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Calibration-Free Coronary Artery Measurements for Interventional Device Sizing using Inverse Geometry X-ray Fluoroscopy: *In Vivo* Validation

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Abstract

Proper sizing of interventional devices to match coronary vessel dimensions improves procedural efficiency and therapeutic outcomes. We have developed a novel method using inverse geometry x-ray fluoroscopy to automatically determine vessel dimensions without the need for magnification calibration or optimal views. To validate this method *in vivo*, we compared results to intravascular ultrasound (IVUS) and coronary computed tomography angiography (CCTA) in a healthy porcine model. Coronary angiography was performed using Scanning-Beam Digital X-ray (SBDX), an inverse geometry fluoroscopy system that performs multiplane digital x-ray tomosynthesis in real time. From a single frame, 3D reconstruction of the arteries was performed by localizing the depth of vessel lumen edges. The 3D model was used to directly calculate length and to determine the best imaging plane to use for diameter measurements, where out-of-plane blur was minimized and the known pixel spacing was used to obtain absolute vessel diameter. End-diastolic length and diameter measurements were compared to measurements from CCTA and IVUS, respectively. For vessel segment lengths measuring 6 mm to 73 mm by CCTA, the SBDX length error was -0.49 ± 1.76 mm (SBDX – CCTA, mean $\pm$ 1 SD). For vessel diameters measuring 2.1 mm to 3.6 mm by IVUS, the SBDX diameter error was 0.07 ± 0.27 mm (SBDX – minimum IVUS diameter, mean $\pm$ 1 SD). The *in vivo* agreement between SBDX-based vessel sizing and gold standard techniques supports the feasibility of calibration-free coronary vessel sizing using inverse geometry x-ray fluoroscopy.

Keywords

x-ray fluoroscopy; inverse geometry; cardiac interventions; quantitative coronary angiography

1. INTRODUCTION

1.1 Motivation

Obtaining accurate vessel diameters and segment lengths during coronary interventions is essential for sizing interventional devices, such as angioplasty balloons and stents. To maximize the therapeutic benefit and reduce the risk of complications, the length of the

interventional device should match the length of the lesion,^{1–5} and the diameter of the device should match the diameter of the healthy vessel on either side of the lesion (“reference diameter”).^{6–9}

The gold standard method for measuring vessel dimensions during an intervention is intravascular ultrasound (IVUS).^{10, 11} IVUS imaging is invasive and time consuming because it requires positioning the IVUS probe within the lesion, which limits its routine utilization. Alternatively, vessel dimensions can be obtained from cineangiograms using quantitative coronary angiography (QCA) techniques.¹² While less invasive than IVUS, QCA measurements can be time consuming and error prone because the vessel magnification needs to be determined to convert image measurements (in pixels) to absolute dimensions at the plane of the vessel (in millimeters). Vessel length is also underestimated if the vessel is not parallel to the image plane (i.e. foreshortened).

Scanning-Beam Digital X-ray (SBDX) is an inverse geometry x-ray fluoroscopy system which produces tomosynthetic images of the patient volume.^{13, 14} We have developed a technique that uses the depth resolution inherent to SBDX tomosynthesis imaging to localize points along a coronary vessel and determine the vessel’s 3D trajectory and magnification from a single angiographic frame.¹⁵ In this paper, we apply this method *in vivo* using a healthy porcine model and compare SBDX-based measurements of coronary artery dimensions against gold standard measurements from IVUS and coronary computed tomography angiography (CCTA).

1.2 Scanning-Beam Digital X-ray

Scanning-Beam Digital X-ray technology uses an inverse geometry beam scanning methodology designed to achieve high dose efficiency in cardiac angiographic and fluoroscopic applications.^{13, 14} The SBDX x-ray tube has a 100×100 array of collimated focal spot positions that are scanned in a raster pattern to create a series of narrow overlapping images of the patient volume (Figure 1). In a typical 7” 15 Hz imaging mode, 71×71 holes are used, and each hole is scanned 8 times within a 1/15 sec frame period. The “beamlet” images captured during scanning are streamed to a real time reconstruction engine that produces full field-of-view images at 15 frames per second. The inverse geometry reduces dose through scatter rejection (narrow beam and large airgap) and a large entrance field below the patient, which can reduce skin dose by a factor of 3 to 7 compared to a conventional geometry.¹⁴

The SBDX prototype system in this study uses a 2 mm thick CdTe photon counting detector with $5.3 \text{ cm} \times 10.6 \text{ cm}$ area and $0.66 \text{ mm} \times 0.66 \text{ mm}$ detector elements (80×160 array). The detector is positioned 150 cm from the source plane. Shift-and-add tomosynthesis is performed at 32 planes throughout the patient volume for each 1/15 sec image frame period, in real time. Plane positions are independently programmable in software; typically a stack of 32 planes with 5 mm between planes is reconstructed for each image frame.

Tomosynthesis images have the property that objects near the image plane are in focus and objects that are out of plane appear blurred. To show all features in focus in the live display, a multiplane composite image is created from the tomosynthesis images in real time.¹³

2. ALGORITHM

The feasibility of coronary vessel sizing with SBDX has been demonstrated previously in phantoms.¹⁵ The algorithm works on the image data provided by a single SBDX image frame. Both the stack of tomosynthesis images and the multiplane composite image are inputs into the vessel sizing algorithm. The only user interaction required by the algorithm is to identify two vessel segment endpoints. The rest of the algorithm steps are completed automatically in software. Recently, the algorithm was redesigned to be more robust in the presence of image noise and complex background structures. The basic steps of the algorithm are described in the following sections.

2.1 2D Vessel Centerline and Edge Segmentation

Using the user-specified segment endpoints, an initial vessel centerline is defined on the multiplane composite image using a wavefront propagation approach.^{16, 17} Briefly, the composite image is converted to a “propagation speed” image, where the image pixel intensity is proportional to the likelihood that the pixel belongs to a vessel. Using dynamic programming, the wavefront propagation algorithm determines the optimal path as the path with the minimal propagation time between the two endpoints. The result is a sequence of 8-neighborhood connected pixels as shown in Figure 2A. A smooth cubic spline is fit to these pixels and parameterized by the arclength to enable regular sampling along the length of the curve.

To extract the two vessel edges from the multiplane composite image, a minimum cost path approach is used.¹⁸ Intensity profiles are extracted perpendicular to the initial centerline at regular intervals along its length. The intensity profiles are gradient filtered to identify edge pixels, and dynamic programming is used to find the optimal paths connecting edge pixels between adjacent profiles.¹⁹ Once the edge paths are calculated, they are converted to smooth cubic splines. A new, more accurate centerline is defined by sampling the two edge contours at regular intervals, calculating the midpoint between the two edges, and defining a spline through the midpoints. Using the new centerline in place of the initial centerline, the entire process is repeated two times to produce a high quality centerline and edge contours (see Figure 2B).

2.2 3D Vessel Edge Depth Localization

The vessel edge depth localization step is outlined in Figure 3. Using the vessel edge contours calculated from the multiplane composite image, the depth of the vessel edges along the source-detector (Z) axis is calculated from the edge sharpness across the tomosynthesis image stack. For each edge contour, perpendicular profiles centered on the edge are defined and extracted from the tomosynthesis images. The profiles are stored as rows where the row position indicates position along the centerline, thereby creating a “straightened” vessel image (see Figure 3B). The image is filtered to produce pixel values proportional to the local contrast. The filtered edge images for both sides are summed together to produce a single image in which the central column corresponds to the vessel edge and the pixel intensity corresponds to the edge strength. This process is repeated for each image plane position.

Next, at each point on the vessel, a distribution of edge strength versus plane position (Z) is calculated. Figure 3E shows an example distribution for a fixed point on the vessel. To calculate the distribution value at a particular point and image plane, the edge strength value is summed within a circular ROI centered on the vessel point. Repeating this calculation for all points along the vessel yields the image shown in Figure 3F, where image row corresponds to image plane position, image column corresponds to distance along the vessel, and image intensity is the edge strength at a particular vessel point and image plane.

The final depth calculation is performed by fitting an active contour model through the edge strength image.^{20, 21} The active contour model is defined along the entire length of the vessel edge and deformed along the image plane position direction (Z). The internal and external forces direct the deformation of the contour. Internal forces enforce a smooth curvature of the path, while external forces guide the path toward image planes where the edge strength is greatest. The cost function for the curve is the weighted sum of the internal and external energies. After iteratively deforming the active contour, the position of the contour along the Z direction gives the final depth value for a point along the vessel (dashed line in Figure 3F).

2.3 Absolute Length and Diameter Measurements

To calculate the absolute length of the vessel segment, a 3D vessel centerline is constructed using the 2D centerline from the multiplane composite (Sec 2.1) and the depth values calculated using the active contour (Sec 2.2). Then, the 3D arclength (in millimeters) is calculated as a function distance along the 2D centerline (in pixels), as shown in Figure 4A. From this graph the 3D length (in millimeters) can be calculated between any two vessel points identified in the 2D image display.

The absolute diameter of the vessel is calculated at multiple points along the length of the vessel using a parameterized model fitting approach.²² For each point along the 2D vessel centerline, the tomosynthesis plane closest to the vessel is identified, and an intensity profile perpendicular to the centerline is extracted. Pixel distances in this profile are converted to physical distances (in millimeters) using the known pixel dimensions in the tomosynthesis plane. The vessel model used for fitting is an ideal cylindrical projection parameterized by the radius, position of the cylinder center along the profile, background intensity, and a gain factor. These parameters are optimized to minimize the squared error between the model and the observed profile, and the final diameter is calculated from the final radius parameter. This process is repeated at regular intervals to calculate the vessel diameter along the entire length of the segment, as shown in Figure 4B.

To boost the SNR of the extracted profiles and create a smooth diameter function, adjacent intensity profiles are averaged prior to fitting the parameterized model. For each diameter measurement, the 31 closest profiles are averaged together before calculating the diameter. Perpendicular profiles are typically extracted in 1 pixel intervals along the 2D centerline, and the nominal pixel spacing of SBDX images is 0.161 mm. Therefore, each diameter measurement corresponds to the average diameter over a 5 mm length of vessel.

3. IN VIVO IMAGING

3.1 Animal Prep

Three healthy juvenile swine (~40 kg) were imaged under a protocol approved by the institutional animal care and use committee at the University of Wisconsin – Madison. After initial anaesthetization, the animals were intubated and mechanically ventilated with 2% isoflurane. Vascular access to the femoral artery was obtained for angi ography and IVUS imaging. Venous access through the ear was used for administration of drugs and iodinated contrast during CCTA imaging.

3.2 IVUS

Intravascular ultrasound was used as the gold standard technique for coronary vessel diameter. A conventional fluoroscopy system (H5000, Philips Healthcare, The Netherlands) was used for image guidance to position the IVUS catheter. A 40 MHz rotating IVUS probe (Revolution, Volcano Corporation, San Diego, CA) was inserted over a guide wire into the distal LAD. Serial cross sectional IVUS images were recorded along the length of the LAD up to the left main artery using continuous imaging during a fixed-rate mechanical pullback (30 frames per second, 0.5 mm/s or 1.0 mm/s pullback speed). During the IVUS pullback, the IVUS frame indices showing major branch points were annotated using the IVUS software. This procedure was repeated for the LCx. The IVUS recordings were exported in DICOM format for offline analysis.

3.3 CCTA

Coronary CT angiography was used as the gold standard technique for measuring vessel segment length. The CCTA was performed using a 64 slice CT scanner (GE Discovery CT750, GE Healthcare, Waukesha, WI). After acquiring scout images, but before the CCTA, zatebradine HCl (Torcis Bioscience, Bristol, UK) was administered to lower the heart rate to below 80 BPM. The CCTA was performed under breath hold conditions by suspending mechanical ventilation at endexpiration during scanning. If needed, vecuronium was administered prior to scanning to prevent residual respiratory motion. The CCTA was performed with a 66 mL injection of iodixanol contrast agent (320 mgI/mL, Visipaque, GE Healthcare, Waukesha, WI), injected at 6 mL/sec, followed by a 60 mL saline chase at 6 mL/sec. Cardiac helical scanning parameters were: 120kV, 600 mA, pitch 0.24, 0.35 sec/ rotation. Images were reconstructed at 75% R-R (enddiastole), with $0.39 \times 0.39 \times 0.625$ mm voxel size.

3.4 SBDX

Angiography with SBDX was performed similar to conventional angiography, using femoral access to position a diagnostic catheter into the left main artery. Cineangiography of the left coronary vessels (LAD and LCx) was performed at multiple views and x-ray tube settings (typically 20 RAO and 30 LAO, 90–120 kVp, 10–15 frames/s) to create different imaging scenes and SNR levels. Contrast injections during angiography were performed by hand using iohexol (Omnipaque 350), and each angiographic run lasted approximately 10

seconds. Raw scan data were downloaded after the experiment for offline reconstruction and analysis.

To identify end-diastolic image frames, a scintillator and photomultiplier tube (PMT) with amplifier was positioned to detect the start of each SBDX image frame from scattered x-rays. The raw (1000×) ECG signal from a physiological monitor connected to the animal (Vital-Guard 450C, Ivy Biomedical Systems, Inc., Branford, CT) was simultaneously recorded. These ECG and PMT signals were then used to calculate the percent R-R interval for each SBDX image frame.

4. ANALYSIS

4.1 Diameter Measurements from IVUS

A semi-automated segmentation method was applied to the IVUS imaging sequences to obtain luminal contours on the transverse (cross sectional) images. An image-based cardiac gating algorithm was applied to identify the end-diastolic image frame from each heartbeat.²³ The end result of the algorithm was a sequence of indices corresponding to the enddiastolic IVUS image frames from each heartbeat.

A semi-automated segmentation method was implemented to identify the lumen borders for the purpose of measuring lumen diameter. The analysis method closely follows previously described methods.^{24–26} Two methods were used to calculate the lumen diameter from the transverse lumen contours (Figure 5). The vessel diameter was derived from the lumen cross sectional area by assuming the lumen had a circular cross section (i.e. $diameter = 2 \sqrt{area/\pi}$). The vessel diameter was also calculated as the minimum diameter from all possible diameters of the lumen contour. The area method was more robust against image noise but was prone to overestimation if the IVUS probe axis was not perfectly parallel to the true vessel centerline. The minimum diameter method was insensitive to probe orientation but tended to underestimate the vessel diameter due to noise in lumen contour. For the purposes of comparison to SBDX vessel sizing diameters, both methods of calculating IVUS lumen diameter were considered.

To register the IVUS images to the other modalities, vessel segments were identified based on the distal carina of branching vessels. For each vessel branch, the end-diastolic image frame on the distal side of the branch was identified in which the main vessel lumen and the lumen of the branch were continuous. Vessel segment endpoints were confirmed with annotations made during the IVUS recording, and segment lengths were cross checked against CCTA length measurements for consistency.

4.2 Length Measurements from CCTA

Analysis of the CCTA images was performed on a GE Advantage W Workstation (GE Healthcare, Waukesha, WI). Vessel lengths were measured at end-diastole by analyzing images reconstructed at 75% R-R interval. Semi-automated 3D centerline tracing of the LAD and LCx was performed by identifying the distal end of the vessel and the branch point from the left main. Curved reformatted images of the LAD and LCx were generated based on the 3D centerlines, which presented the vessel as 2D cut plane through a “straightened”

3D vessel volume, as shown in Figure 6. From these images, the position of the distal edge of each branching vessel was manually identified and recorded. The distances between each of these reference points served as the segment length ground truth values.

4.3 Segment Measurements from SBDX Angiography

A software-based simulation of the hardware reconstructor was used to reconstruct SBDX tomosynthetic images and multiplane composites. For each image frame, 32 tomosynthesis planes were reconstructed spaced 5 mm apart along the Z direction and centered about mechanical isocenter. For each image frame, a multiplane composite was created using the 16 tomosynthesis image planes closest to the main branch of the artery being analyzed. The vessel sizing algorithm described in Section 2 was used to calculate the vessel dimensions from SBDX angiographic frames. An image frame was analyzed if 1) the frame was recorded at end-diastole (~75% R-R interval) and 2) the main branch and side branches were filled with contrast.

The location of the distal carina each vessel branch was identified in the multiplane composite images. In addition, the most proximal and distal points along the LAD and LCx were recorded. Using the full vessel endpoints, the vessel sizing algorithm was applied, resulting in 3D length and diameter measurements of the entire vessel as a function of position along the 2D vessel centerline (see Figure 4). Measurements for individual segments were performed by finding the points on the full 2D vessel centerline closest to the user defined segment endpoints, calculating the 3D segment length from the corresponding range on the full 3D vessel centerline, and extracting diameter measurements over the range defined by the segment endpoints.

4.4 Vessel Segment Comparison and Statistical Analysis

Vessel segments of the LAD and LCx were defined in each SBDX image frame based on the vessel branches that were positively identified in the image. The set of branches used varied depending on the view angle of the angiogram and the degree of filling with contrast within the vessel. In order to test segments with a range of lengths, segments were defined by each possible pair-wise combination of branches as shown in Figure 7. Thus, the longer segments were made up of one or more smaller segments.

For each segment, the length from SBDX was compared to the segment length from CCTA, and the length error was calculated as (SBDX length – CCTA length). The average length error and standard deviation were calculated for each segment within each experiment and across all experiments as a whole. The least squares linear regression between CCTA and SBDX lengths was calculated to determine the linear relationship between the two methods. Pearson's correlation coefficient was calculated to measure the correlation between the two techniques. SBDX-derived results were plotted against the CCTA results as a calibration curve to show the correlation between the two methods.

Vessel diameters calculated with SBDX were compared to the IVUS using the lumen area and minimum diameter methods. Each vessel segment was divided into 5 mm long, non-overlapping sub-segments. The SBDX diameter measurement from the point in the middle

of a sub-segment corresponded to the average diameter over that segment due to the profile averaging applied prior to measuring the diameter. The SBDX diameter for a sub-segment was compared to the average IVUS diameter over the same 5 mm section of the vessel. Diameter error was calculated for each subsegment as (SBDX diameter – IVUS diameter). Mean diameter error and standard deviation were calculated across each experiment and for all experiments together. Least squares linear regression, Pearson's correlation coefficient, and calibration curve plots were also calculated for diameter measurements.

5. RESULTS

Table 1 summarizes the length measurements based on CCTA and the SBDX algorithm. The number of measurements in each experiment varied with the number of SBDX image frames and the number of segments in each frame. Table 2 compares length measurements between SBDX and CCTA. The CCTA lengths of the segments analyzed ranged from 6.10 mm to 72.50 mm, with an average segment length of 25.32 mm. Overall, length errors were similar across all three experiments, with average errors of -0.65 ± 1.99 mm, -0.78 ± 1.90 mm, and -0.22 ± 1.42 mm. Overall error was -0.49 ± 1.76 mm. The negative mean difference between the measurements indicated that the vessel sizing algorithm tended to underestimate the true segment length. Length measurements correlated strongly between the two methods with correlation coefficients of 0.99 in all cases. Least squares line fits between the SBDX and CCTA length data had near unity slope. Figure 8 shows this strong relationship graphically.

Diameter measurements using SBDX and IVUS are summarized in Table 3. The results were similar across all three experiments. The measurement differences between SBDX and IVUS are shown in Table 4. When using the “area” method for IVUS diameter, mean differences with SBDX were -0.18 ± 0.29 mm, -0.09 ± 0.28 mm, and -0.15 ± 0.24 mm, respectively, with the overall difference of -0.14 ± 0.26 mm. Using the minimum diameter from the IVUS lumen segmentation, the mean differences compared to SBDX were -0.02 ± 0.28 mm, 0.11 ± 0.27 mm, and 0.09 ± 0.26 mm, with an overall difference of 0.07 ± 0.27 mm. Based on mean error, the SBDX diameter measurements agreed slightly better with the IVUS measurements calculated using the minimum diameter method.

Figure 9 shows the diameter results graphically (minimum diameter method). Data points were clustered around the unity slope line. Due to the limited range of diameters measured, it was difficult to obtain a reliable linear fit. To restrict the behavior of the fit results, a line passing through the origin was used, resulting in slopes ranging from 1.03 to 1.06 for the lumen area method and from 0.95 to 1.00 for the minimum diameter method. For each sub-segment, one IVUS measurement was compared to multiple SBDX measurements. When plotted as shown in Figure 9, this one-to-many comparison appeared as horizontal clusters, which skewed the linear fit model to have a slope less than one. Ideally, one IVUS measurement would be compared to one SBDX measurement, but the time required for multiple IVUS acquisitions and the increased risk of complications made multiple IVUS measurements impractical.

The Pearson correlation between IVUS and SBDX diameter measurements ranged from 0.24 to 0.83. Within each experiment, most of the vessel sub-segments measured had similar diameters, ranging between 1.93 and 3.86 mm. Given that the variability (standard deviation) in the diameter error was about 0.25 mm, the data form a cloud around the unity slope line. Compare this to the length data shown in Figure 8, where the variability is on the order of 2 mm, but the measurements span a range of about 70 mm. To better characterize the correlation between the two measurements, a wider range of vessel diameters would need to be imaged with the two modalities.

6. DISCUSSION

The SBDX vessel sizing algorithm measured both vessel length and diameter in a healthy swine model with submillimeter accuracy. Fifty vessel segment lengths were measured across multiple image frames resulting in 429 total measurements. The mean segment length from CCTA was 24.8 mm, and the SBDX measurements were within 0.49 mm on average. The diameters of 33 different sub-segments were compared between IVUS and multiple SBDX measurements for a total of 347 comparisons. For vessel diameters ranging between 2 and 4 mm, the mean diameter with SBDX was within 0.14 mm or 0.07 mm, depending on how the IVUS diameter was calculated. Given that interventional devices come in length increments of 4–5 mm and diameter increments of 0.25 mm, SBDX vessel sizing should be accurate enough to guide an interventionalist to select a device that matches vessel dimensions. The variability of the measurements (quantified by the standard deviation of the error) was 1.97 mm for the length and 0.26–0.27 mm for the diameter. These values represent the uncertainty of a measurement from a single image frame. With the algorithm fully automated (after the user identifies the segment of interest), the results of multiple image frames could be averaged together to improve the precision.

The measured variability of vessel measurements with SBDX depended on a number of factors, some of which are not related to the algorithm itself. In these experiments, each vessel was measured only once using the gold standard method (CCTA for length, IVUS for diameter). To put the variability of SBDX measurements into perspective, the multiple gold standard measurements of the same vessel would need to be performed to calculate the statistical variability of those modalities and compare to the SBDX method.

Some of the diameter errors may be attributed to the fact that IVUS and SBDX imaging were not performed at the same point in time. Physiological changes over the course of the experiment may have caused vessel dimensions to vary between modalities. This potential problem could be mitigated by performing the IVUS and SBDX imaging closer together, thereby minimizing the time discrepancy between the two measurements.

Other groups have compared diameter measurements *in vivo* between conventional QCA and IVUS. Davies et al.²⁷ showed that conventional QCA underestimated vessel diameter by 0.12 ± 0.37 mm and 0.02 ± 0.34 mm (mean $\pm$ 1 standard deviation), when one or two angiographic views are used, respectively. Sinha et al.²⁸ measured differences between QCA and IVUS of 0.02 ± 0.12 mm, 0.04 ± 0.15 mm, 0.17 ± 0.23 mm, and 0.19 ± 0.17 mm, depending on the distance from the ostium of the artery. In both studies, the minimum IVUS

diameter was compared to the QCA result. For the same type of measurement, the SBDX vessel sizing algorithm had a mean difference of 0.07 ± 0.28 mm, which put its accuracy and precision on par with existing methods.

7. CONCLUSION

This work reports the first *in vivo* validation of the accuracy and precision of a calibration-free SBDX vessel sizing algorithm. Measurements using this method had sub-millimeter accuracy on average when compared to gold standard methods. All measurements with SBDX were performed on single image frames without the use of a magnification calibration object (such as a contrast filled catheter) and without optimal, non-foreshortened views. This method may be less cumbersome and error prone than conventional QCA, and less invasive and time consuming than IVUS. By providing an accurate and rapid method for determining vessel sizes, this method could be used clinically to assist in the selection and sizing of interventional devices and potentially improve procedure efficiency.

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Figure 1.

(A) SBDX uses a scanned x-ray beam and multihole collimator to produce a series of narrow overlapping x-ray projections. (B) A stack of full field-of-view tomosynthesis images are reconstructed from the narrow beam images for each frame period. (C) A multiplane composite image is created from the tomosynthesis image stack in which all features are in focus, similar to a conventional fluoroscopic projection.

Figure 2.

The vessel sizing algorithm starts with 2D vessel segmentation on the multiplane composite. (A) A rough, initial centerline is automatically defined between two user defined points. (B) The initial centerline is used to create edge contours and precise centerline.

Figure 3.

The degree of tomographic blur versus plane position is used to determine the depth (Z position) of each point along the vessel edge. (A) Perpendicular profiles are extracted along each edge from each tomosynthesis image. (B) The profiles create a straightened vessel edge image for each image plane. (C) The straightened vessel images are filtered to delineate edge pixels. (D) The two edge images are summed together for each plane. (E) For each position along the vessel, the edge strength values within an ROI are summed to produce a distribution of edge strength versus plane position. (F) The ROI sums are stored in an image where the row indicates the plane position and the column indicates position along the 2D vessel centerline. An active contour (dashed line) is deformed through the brightest regions while maintaining a smooth curve. Note that in (A–E), only three planes are shown for clarity; 32 planes were typically used.

Figure 4.

The results of the algorithm are used to look up the length (A) and diameter (B) for any part of the measured segment based on the position along the 2D vessel centerline. (C) The results can also be visualized as a surface rendering to show the trajectory and size of the artery.

Figure 5. After segmenting the vessel lumen, the diameter was calculated using two methods: (A) the cross sectional area and (B) the minimum lumen diameter.

Figure 6. Coronary vessels were segmented from the CT volume. (A) Three-dimensional centerlines of the LAD and LCx were defined using semi-automatic methods. (B) Curved reformatted images of the coronary vessels were used to make length measurements between branch points.

Figure 7.

Vessel measurements were compared to gold standard techniques based on segments between branching vessels. (A) Vessel segment lengths were compared CCTA, while average sub-segment diameters were compared to IVUS. Segment lengths varied depending on which branches were used for endpoints, while the sub-segments used for diameter comparisons were all approximately 5 mm in length. (B) Vessel branches were combined in different ways to produce segments of various lengths. The four branches shown can be combined pairwise to create six different segment lengths.

Experiment 1

Experiment 2

Experiment 3

Combined

Figure 8.
SBDX lengths from the vessel sizing algorithm are plotted against ground truth (CCTA) measurements. Line represents unity slope.

Figure 9.

Vessel diameters calculated using the vessel sizing algorithm are plotted against the average IVUS diameter, which was derived from the minimum diameter across the segmented lumen. The line represents unity slope.

Table 1

Summary of Length Results. Length measurements presented as mean [min, max].

Experiment	Number of Segments	Number of Measurements	SBDX Length [mm]	CCTA Length [mm]
1	24	145	24.79 [3.74, 62.52]	25.44 [6.10, 59.40]
2	16	99	31.58 [4.41, 73.80]	32.36 [6.60, 72.50]
3	10	185	21.23 [7.36, 46.57]	21.45 [8.50, 46.50]
All	50	429	24.82 [3.74, 73.80]	25.32 [6.10, 72.50]

Table 2

Comparison of Length Measurements. Error was calculated as (SBDX length – CCTA length) and presented as mean $\pm$ 1 standard deviation. Linear fit assumes y = CCTA length and x = SBDX length.

Experiment	Number of Measurements	Length Error [mm]	Linear Fit	Pearson Correlation (r)
1	145	-0.65 ± 1.99	$y = 1.00x + 0.60$	0.99
2	99	-0.78 ± 1.90	$y = 1.00x + 0.76$	0.99
3	185	-0.22 ± 1.42	$y = 1.01x + 0.11$	0.99
All	429	-0.49 ± 1.76	$y = 1.01x + 0.34$	0.99

Summary of Diameter Measurements. Diameter measurements presented as mean [min, max].

Table 3

Experiment	Number of Sub-Segments	Number of Measurements	SBDX Diameter [mm]	IVUS Diameter (Area) [mm]	IVUS Diameter (Min) [mm]
1	10	71	2.51 [2.08, 3.13]	2.69 [2.23, 3.04]	2.53 [2.06, 2.89]
2	12	104	2.69 [1.93, 3.62]	2.77 [2.42, 3.90]	2.58 [2.24, 3.60]
3	11	172	2.92 [2.27, 3.86]	3.07 [2.71, 3.64]	2.83 [2.32, 3.42]
All	33	347	2.76 [1.93, 3.86]	2.90 [2.23, 3.90]	2.69 [2.06, 3.60]

Table 4

Comparison of Diameter Measurements. Error was calculated as (SBDX diameter – IVUS diameter) and presented as mean $\pm$ 1 standard deviation. Linear fit assumes $y = \text{IVUS diameter}$ and $x = \text{SBDX diameter}$. Due to the narrow range of diameters measured, the linear fit was constrained to go through the origin.

Experiment	IVUS (Area)			IVUS (Min)		
	Diameter Error [mm]	Linear Fit	Pearson Correlation (r)	Diameter Error [mm]	Linear Fit	Pearson Correlation (r)
1	-0.18 ± 0.29	$y = 1.06x$	0.24	-0.02 ± 0.28	$y = 1.00x$	0.26
2	-0.09 ± 0.28	$y = 1.03x$	0.82	0.11 ± 0.27	$y = 0.95x$	0.83
3	-0.15 ± 0.24	$y = 1.05x$	0.51	0.09 ± 0.26	$y = 0.97x$	0.47
All	-0.14 ± 0.26	$y = 1.04x$	0.72	0.07 ± 0.27	$y = 0.97x$	0.70