Fully automatic three-dimensional visualization of intravascular optical coherence tomography images: methods and feasibility in vivo

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Abstract: Intravascular optical coherence tomography (IV-OCT) is an imaging modality that can be used for the assessment of intracoronary stents. Recent publications pointed to the fact that 3D visualizations have potential advantages compared to conventional 2D representations. However, 3D imaging still requires a time consuming manual procedure not suitable for on-line application during coronary interventions. We propose an algorithm for a rapid and fully automatic 3D visualization of IV-OCT pullbacks. IV-OCT images are first processed for the segmentation of the different structures. This also allows for automatic pullback calibration. Then, according to the segmentation results, different structures are depicted with different colors to visualize the vessel wall, the stent and the guidewire in details. Final 3D rendering results are obtained through the use of a commercial 3D DICOM viewer. Manual analysis was used as ground-truth for the validation of the segmentation algorithms. A correlation value of 0.99 and good limits of agreement (Bland Altman statistics) were found over 250 images randomly extracted from 25 in vivo pullbacks. Moreover, 3D rendering was compared to angiography, pictures of deployed stents made available by the manufacturers and to conventional 2D imaging corroborating visualization results. Computational time for the visualization of an entire data sets resulted to be ~74 sec. The proposed method allows for the on-line use of 3D IV-OCT during percutaneous coronary interventions, potentially allowing treatments optimization.

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Intravascular optical coherence tomography (IV-OCT) is a catheter based imaging modality for the visualization of coronary arteries [1]. Since its inception, IV-OCT is increasingly being used during percutaneous coronary intervention (PCI) to improve the understanding of the anatomy of complex lesions and to elucidate the mechanisms of atherosclerosis. When compared to other vascular imaging modalities, IV-OCT presents an axial resolution (~10-15 µm) improved of approximately one order of magnitude than intravascular ultrasound (IVUS) (>100 µm), coronary angiography (~200 µm) and computed tomography (CT) (>300 µm) [2]. This exceptional resolution makes IV-OCT the most suitable imaging modality for the visualization of small intraluminal structures such as thrombus [3], atherosclerotic plaques, vessel dissections and coronary stent struts [4].

Recent publications pointed to the fact that three dimensional (3D) IV-OCT visualization has potential advantages compared to conventional two dimensional (2D) image representation [5–8]. It was shown that 3D IV-OCT may be of help for a better characterization of stent malapposition and identification of thrombus, guiding stent post-dilation and aspiration thrombectomy [9]. Moreover, it allows for the assessment of jailed side
branches and for the rewiring of a side-branch ostium during percutaneous coronary intervention (PCI) minimizing the risk of floating struts [5,6,8]. Given that this is not something easily achievable with conventional 2D imaging, potential applications of 3D IV-OCT for the management of acute myocardial infarction have been identified.

As indicated by [5,8], the major pitfall of 3D visualization is that image preparation currently has to be done manually. To apply 3D IV-OCT to stent assessment, each stent strut needs to be manually identified and segmented on every image. This results in a labor intensive and time-consuming procedure (i.e., >3 hours per pullback [10]) making the approach not suitable for on-line application during PCI or for large scale use in clinical trials. In fact, the clinical need of a tool for automatic 3D IV-OCT visualization is underlined by several publications [8,11,12]. In particular, the team of Serruys stated that the potential of three-dimensional IV-OCT is real as a complimentary tool to 2D IV-OCT, if available as 'push-of-a-button' at the time of the data acquisition [5]. A similar statement was made by the team of Akasaka, underlining the fact that automatic 3D visualization would bring IV-OCT closer to becoming a practical imaging technique in the daily cardiac catheterization laboratory [8].

As such, the aim of the current study was to achieve a rapid and fully automatic 3D visualization of IV-OCT pullbacks.

2. Algorithm for automatic three-dimensional visualization

The flowchart in Fig. 1 illustrates the entire image processing framework. First, the imaging catheter is segmented in the IV-OCT pullback leading to automatic image calibration. Then guide-wire(s) boundaries are located and vessel lumen and stent struts are automatically segmented through the entire data set. Subsequently, 3-dimensional visualization is generated through dedicated software for volume-rendering.

![Flowchart of the entire method. Following the segmentation of the imaging catheter, the guide-wire(s), stent struts and the vessel lumen, three-dimensional rendering of IV-OCT data is automatically obtained. Automatic pullback calibration is obtained by employing imaging catheter segmentation results.](image)

Fig. 1. Flowchart of the entire method. Following the segmentation of the imaging catheter, the guide-wire(s), stent struts and the vessel lumen, three-dimensional rendering of IV-OCT data is automatically obtained. Automatic pullback calibration is obtained by employing imaging catheter segmentation results.

2.1. Segmentation of the imaging catheter

The IV-OCT imaging catheter appears as multiple bright concentric circular rings (equivalent to vertical lines in the polar image domain). As illustrated by Fig. 2, such object is composed of a plastic sheet (green narrow arrow) covering the internal structures. The diameter of the catheter is typically used for the manual calibration of the IV-OCT images. However, the external plastic sheet often appears to be deformed during vessel flushing, image acquisition and when it is pushed against the vessel wall and can thus not be used for robust calibration. The internal structures of the catheter however (Fig. 2, yellow large arrows), always maintain a fixed appearance. We therefor propose an algorithm for a rapid and robust segmentation of these inner structures allowing for the automatic segmentation of the imaging catheter and for the automatic calibration of the images.

Hereeto polar domain images are processed via a rapid algorithm based on the Hough transform [13]. The standard Hough transform is a global technique for the detection of parameterizable patterns in digital images. By transforming the Cartesian space to a
parametric space (i.e., Hough space) any parametric curve can be defined. In the current situation (Fig. 2c), lines can be parameterized as:

\[ \rho = x \cos(\theta) + y \sin(\theta), \]  

(1)

where \( \rho \) is the distance between the line and the origin and \( \theta \) is the angle of the line according to the positive x-direction as shown in Fig. 2c. By imposing boundaries on the value of \( \theta \) (i.e., \( \theta = 90^\circ \)), detection of vertical lines only is made possible. More specific details about Hough transform are available in literature [13,14].

Through Eq. (1), vertical lines are detected in multiple adjacent images. As the largest concentric ring represents the true position of the catheter, the position of the line with the largest value of \( \rho \) is averaged through the entire pullback. The use of 3D information from multiple frames makes the detection process robust with respect to noise or image artifacts. The detected catheter size is first used to automatically calibrate the entire pullback (catheter size is known a priori); then a mask is automatically applied over the entire pullback to exclude the imaging catheter from further processing.

2.2. Segmentation of the guide-wire(s)

The correct identification and segmentation of the guide-wire (GW) in a single image can be challenging due to the presence of stent struts and artifacts. Moreover, the number of guide-wires present in the image is not known a-priori (e.g., in case of PCI at a vessel bifurcation). Guide-wires are segmented by implying spatial continuity through the entire pullback. Given that the GW is characterized by a bright reflection immediately followed by a shadow, A-lines containing a GW (Fig. 3a) present low accumulated intensities [15]. To exploit this characteristic, an accumulated intensity feature image is generated from the entire pullback. Hereby, every A-line is processed by accumulating pixel intensities as a function of depth. In this way, an entire 2D image could be represented by a single (accumulated intensity) profile. Profiles of multiple consecutive frames generate the final feature image (Fig. 3b).

The left and the right edges of the GW are then located as connected contours in the feature image through morphological operations. The feature image is first processed by global image thresholding using Otsu’s method (Fig. 3c) [16]. Then, morphological closing,
area constrain and morphological majority operations followed by a morphological dilation [13,17,18] are applied (Fig. 3d). This results in a robust and fully automatic procedure able to segment multiple GWs through entire IV-OCT pullbacks (Fig. 3e) and thus on every image (Fig. 3f). Note that in case the guide-wire is wrapped around the A-line with an acquisition angle of 0°, it results to be divided into two pieces: one on the right and the other on the left side of the image. Image padding (by adding a fixed number of A-lines from the right border to the left and vice versa) and recovering of the original image after GW segmentation can easily solve this problem.

### 2.3. Stent struts and lumen segmentation

Stent struts and the vessel lumen are segmented by an algorithm previously developed and validated by our research group [19]. In brief, stent struts are detected through individual A-line analysis in the polar image domain (Fig. 4a). In this domain, a single A-line corresponds to an intensity profile varying as a function of depth I(d). First, A-lines are classified according to their intensity profile properties: (1) maximum intensity level; (2) number of pixels exceeding the full-width partial maximum intensity FWPM; (3) presence of a shadow and (4) shadow length.

![Fig. 4. Stent struts and lumen segmentation. Following A-line classification on the polar image domain, segmentation is obtained by applying spatial continuity over multiple A-lines (a). Thus, the image is scan-converted to Cartesian domain for visualization and quantification of strut apposition (b).](image)
Shadows are located by looking at the average intensity of pixels after the shallowest intensity peak, while shadow length is computed through the use of a sliding window. The use of an intensity level threshold allows identifying the presence of a shadow along an A-line. In a similar way, shadow’s length is quantified. According to these properties, strut A-lines are classified through the entire image.

Stent struts are subsequently segmented by employing spatial continuity. First, luminal stent strut surface is located in every single A-line through a peak detection procedure. Then, 1-dimensional filtering is applied to clusters of adjacent strut A-lines smoothing the detected surface. In a similar way, the vessel wall is segmented by applying spatial continuity through multiple A-lines by the use of a smoothing B-spline (Fig. 4).

Images are then scan-converted to Cartesian domain and strut apposition is subsequently quantified. Discrimination of malapposed from apposed struts is obtained through the use of a priori data on stent thickness [12]. The algorithm was previously validated against the manual analysis of two expert IV-OCT image readers over a large data set.

A more extensive explanation on the development and the validation of this segmentation algorithm is available in literature [19].

2.4. 3-dimensional rendering

Prior to 3D IV-OCT visualization, detected stent struts are automatically painted with a predefined intensity value exceeding the dynamic range of the IV-OCT data. Different values are given to apposed and malapposed struts, respectively. Segmented images are subsequently stored as DICOM and 3D visualization is automatically obtained by the means of dedicated software for volume rendering.

The same look-up table and opacity table were used as an input for the 3D rendering software and applied to all the data sets to be visualized. Opacity was set to be higher for stent struts with respect to the vessel wall and very low intensity pixels were made transparent. The look-up table depicts the vessel wall in red, apposed stent struts in copper and malapposed stent struts in blue. The opacity table and the look-up table are reported on Fig. 5. Real object dimensions are employed for voxels’ rendering.

![Fig. 5. Look-up and opacity tables. Different opacity values (y-axis) are given to stents struts and the vessel wall. The intensity range for the different structures is displayed by the x-axis. Moreover, also the colormap utilized for 3D rendering is reported.](image)

2.5. Implementation details

Algorithms for imaging catheter and guide-wire segmentation were implemented in Matlab 2011a, 64-bit version 7.12.0 (MathWorks, Natick, MA), while the algorithms for image scan-conversion [19,20] were implemented in Visual C++ 2008 (Microsoft, Redmond, WA). The C++ code is executed in Matlab by the means of the MEX libraries (MathWorks).

3D visualization is obtained through the use of the windows based software INTAGE Realia (Cybernet Systems Co., Ltd, Tokyo, Japan). INTAGE Realia is a 3D DICOM viewer free of charge, publicly available for download from Cybernet Systems Co. website [21].
The quantification of computation time was made on a laptop computer with the following characteristics: processor Intel Core i7-3720QM (Intel Corporation, Santa Clara, CA), memory 16GB RAM, 1866MHz DDR3, video card Nvidia Quadro k2000M (Nvidia Corporation, Santa Clara, CA), 64-bit Windows 7 OS (Microsoft) and 128GB solid state hard drive.

Default morphological operations as defined in the Matlab Processing Toolbox 2011 (MathWorks) were used in this study. The algorithm parameter defining the area constrain operation for guide-wire segmentation was set equals to 3500 pixels and kept constant for all the data sets analyzed in this study. The parameters for peak detection in the Hough transform matrix H are set to their default values as defined in the Matlab Image Processing Toolbox (MathWorks) with two exceptions: (1) the parameter defining the maximum number of peaks to be detected was set equal to 4 and (2) the threshold at which the values of H are considered to be peaks was set equals to 0.9.

3. Methods: validation experiments

The algorithms for the segmentation of the imaging catheter and guide-wire were validated against manual analysis performed by an expert image reader blinded to the automatic results. 25 in vivo IV-OCT pullbacks, randomly selected from an already existing database, were both automatically and manually calibrated and the difference between the two procedures (calculated in pixels) quantified as the automatic calibration error. Then, the coordinates of the GW boundaries were automatically located and compared to the manual segmentation over 250 images extracted at random from the previously selected data sets. Given that such pullbacks represent in vivo real-life IV-OCT data, selected images included multiple image artifacts such as incomplete blood clearance, motion artifact (commonly called sew-up artifact), very eccentric catheter position and sunflower artifact. Therefore, validation was obtained using manual analysis as the ground truth by regression analysis and Bland-Altman statistics, defining the range of agreement as mean bias ±1.96 SD [22].

To test final 3D visualization results, IVOCT data sets were selected a priori on the basis of their clinical scenario to illustrate a wide range of different situations. 3D IV-OCT stent patterns were visually compared to photographic pictures of the deployed stents made available by the manufacturers. Similarly, 3D renderings were visually compared to angiographic appearance of the analyzed vessel. Finally, small intraluminal details such as thrombus, stent malapposition, guide-wires and vessel wall dissection were visualized by 3D IV-OCT and compared to conventional 2D images. Both single vessel and bifurcation PCI procedures were taken into account.

Computational times required by the different modules for automatic processing of an entire IV-OCT pullback were also quantified.

3.1. IV-OCT data

IV-OCT pullbacks were acquired through the use of a commercially available FD-OCT system (C7-XR system, St. Jude, St. Paul, Minnesota) and by the means of the Dragonfly catheter (St. Jude) in the context of a clinical trial (University Hospital Leuven – STACCATO, NCT01065519) and other clinical studies. All patients gave informed consent. The system acquires data at a rate of 100 images per second (54,000 A-lines per second) with a pullback length of ~54 mm and a scan range (in contrast medium) of ~5 mm. The system is equipped with a near infrared light source with a central wavelength of 1305 nm and a scan range of ±80 nm. Imaging specifications, according to the manufacturer, are an axial resolution ~20 µm and a lateral resolution of 25-60 µm. Acquisition speeds of both 20 mm/s and 10 mm/s were employed.
4. Results

Figure 6 illustrates the validation results for both the algorithms for automatic segmentation of the imaging catheter and the guide-wire. A correlation coefficient of 0.99 was found in both cases and Bland-Altman statistic did not show any bias together with narrow limits of agreement: mean bias $-0.2 \pm 13.7$ for the GW segmentation algorithm and $-0.5 \pm 4.0$ for the imaging catheter segmentation algorithm (values are expressed in pixels with a pixel size of $\sim 5.2 \, \mu m$). Over a total of 250 images, only in 3 single images ($\sim 1\%$) the presence of a second guide-wire was not correctly detected (false negative).

Figures 7–12 show examples of fully automatic 3D reconstructions. Figures 7 and 8 compare 3D visualization to stent patterns a priori known from the manufacturers. In addition, Fig. 7 illustrates an example of stent edge dissection, while Fig. 8 presents results of an IV-OCT pullback acquired at lower pullback speed (10mm/s vs. 20mm/s). Figure 9 shows 3D renderings of simple lesions compared to angiography.

Figures 10, 11, and 12 show additional examples of 3D IV-OCT images for the visualization of small intraluminal structures. Figure 10a presents an example of thrombus and Fig. 10b an example of a jailed side-branch. Figure 11 deals with the automatic detection and 3D characterization of stent strut malapposition and Fig. 12 illustrates an example of 3D visualization during a side-branch rewiring and reopening procedure (bifurcation PCI).

The total processing time for a 30 mm stent (corresponding to 150 images for 20 mm/s acquisition speed) resulted to be approximately 74 seconds. More specifically, automatic imaging catheter segmentation and pullback calibration took $\sim 1.3$ seconds, guide-wire(s) segmentation $\sim 1.4$ seconds. Stent and lumen segmentation took $\sim 46$ seconds and scan-conversion to Cartesian images of 800·800 pixels $\sim 24$ seconds.

![Fig. 6. Validation results. Above: regression analysis and Bland-Altman statistics for automatic calibration results. Below: guide-wire segmentation results. Values are expressed in pixels, with a pixel size of $\sim 5.2 \, \mu m$. For the validation of the automatic calibration algorithm, reported values correspond to the pixel shift in the original image automatically (y-axis) and manually (x-axis) obtained. For the GW segmentation, reported values correspond to the A-line number where guide-wire boundaries are located both automatically (y-axis) and manually (x-axis).](image-url)
Fig. 7. Fully automatic 3D IV-OCT imaging. The figure compares 3D IV-OCT stent pattern to a priori data from the manufacturer (Promus Element, Boston Scientific, Natick, MA). Small details of the stent pattern (green circles) are visible in both 3D images and manufacturer data. An edge dissection is also clearly visible in the 3D image (yellow arrow). Asterisk (*) indicates guide-wire shadowing artifact.

Fig. 8. 3D IV-OCT stent pattern of another stent device (Xience Prime, Abbott Vascular, Santa Clara, CA). Yellow arrows point to a fine stent detail correctly visualized by 3D IV-OCT in a reproducible way. In addition, the figure shows that a slower acquisition speed (10 mm/s) results in an improved longitudinal resolution.
Fig. 9. Both figures (a) and (b) compare automatic 3D imaging to stent appearance at angiography (arrows point to vessel stenosis). Figure (a) shows a simple lesion before and after stent implantation in the left anterior descending artery (LAD). In fig. (b) a lesion located very close to the ostium of side branches of the LAD is depicted. 3D images are able to better illustrate vessel anatomy compared to conventional 2D images. *Asterisk indicates side-branches – † indicates the guide-wire.

Fig. 10. Panel (a) shows coronary intraluminal thrombus partially covering a stent automatically displayed by 3D IV-OCT (yellow arrows). A stent over a side branch and its different compartments (4 in total) are visible in the 3D rendering in fig. (b). Assessment of the such jailing compartments would be very difficult on conventional 2D IV-OCT images. Asterisk (*) indicates guide-wire shadowing artifact.
Fig. 11. Example of automatic characterization of stent malapposition. 3D spatial distribution of a stent edge malapposition is displayed fully automatically (blue color). If compared to 2D images, 3D IV-OCT better illustrates spatial distribution of malapposed struts.

Fig. 12. Example of automatic IV-OCT 3D visualization of a rewired side-branch in bifurcation PCI. Image on the top illustrates how 3D IV-OCT can guide the positioning of the guide-wire (yellow arrow) in a distal cell for subsequent side-branch reopening. This can potentially optimize side-branch rewiring and reopening procedures. Evaluation of stent cells through 2D images and guide-wire positioning, may be unfeasible. Bottom figure illustrates the final result of side branch reopening, confirming the absence of free floating material. Asterisk (*) indicates the guide-wire in conventional 2D images.
5. Discussion

A fully automatic algorithm for 3D IV-OCT imaging was developed. Multiple examples from real life in vivo clinical cases (selected a priori on the basis of their clinical relevance) were shown to demonstrate the validity of the proposed method. Comparison to angiographic data (Fig. 9) showed visual consistency of the vessel appearance by looking at anatomical features such as side branches and vessel stenosis. Moreover, 3D IV-OCT stent patterns of different devices (Figs. 7 and 8) visually fitted to a priori data from the manufacturers. In fact, from those figures, it is possible to appreciate that also small details of the stent metallic framework were correctly visualized corroborating the results of the proposed method. Furthermore, small intraluminal details such as vessel dissections (Fig. 7), thrombus (Fig. 10a), jailing struts (Fig. 10b), stent strut apposition (Fig. 11), guide-wire and stent cells (Fig. 12) were clearly visualized and discernible from the other structures.

Given that the described method is fast (processing time for an entire data set ~1 min), together with the fact that no manual interaction is required, makes 3D IV-OCT available through the ‘push-of-a-button’ as a complimentary tool to conventional 2D IV-OCT images. As such, it was shown that the proposed method is suitable for the application of 3D IV-OCT visualization in clinical scenarios and it can be applied for the optimization of PCI procedures as indicated by the team of Serruys [5] and the team of Akasaka [8].

5.1. Validation

Algorithms for automatic guide-wire and imaging catheter segmentation were validated against the manual analysis of IV-OCT images. In a similar way, the accuracy of the algorithms for lumen segmentation and stent struts detection was previously reported [19]. Given that manual analysis is considered to be the gold standard for image segmentation [12], the validity of the proposed methods was proven. In addition, IV-OCT 3D rendering was validated by comparing algorithm results to angiographic data, manufacturers’ stent data and to conventional 2D images.

Although the segmentation algorithms were validated through a quantitative procedure, only a qualitative validation of final 3D visualization results was made. The validation of 3D rendering is always a controversial and challenging task. An alternative approach would make use of IV-OCT phantoms [20,23]. Such phantoms were typically used to validate the ability of IV-OCT to quantify vessel morphology measurement such as lumen area and lumen diameters. However, even through the use of phantom models, quantitative validation of 3D visualization algorithms remains a very challenging task. Moreover, to the best knowledge of the authors, this is the first report of an automatic method for 3D visualization of IV-OCT data. As such, comparison to previous studies is not possible. In any case, qualitative visual inspection of algorithm results showed consistency through all the figures from 7 to 12 corroborating 3D visualization results.

Furthermore, 3D rendering was obtained through the same predefined color map and opacity table. In a similar way, the same 3D image illumination and shading are applied to all the data showed in this study. Although this generates uniform results, the evaluation of different solutions for 3D rendering was beyond the scope of this study. Only future standardization efforts from the IV-OCT community would be able to create a consensus about optimal guidelines for 3D IV-OCT visualization and validation.

As an additional consideration, it is important to stress the fact that 3D IV-OCT is intended as a complimentary tool to 2D IV-OCT. In fact, only the combination of both modalities would be potentially able to make IV-OCT a practical tool in percutaneous coronary interventions [5,8].
5.2. Further improvements

Figure 8 shows that acquisitions at lower pullback speed (acquiring a higher number of frames) allow for more detailed 3D IV-OCT. However, slower acquisitions potentially make IV-OCT imaging more vulnerable to motion artifacts (e.g., cardiac motion) [5]. Thus, a future generation of IV-OCT systems with an improved frame-rate will be able to provide additional benefits for 3D applications. Similarly, software based approaches (e.g., longitudinal low pass filtering) could mitigate the sampling effect of IV-OCT in the pullback direction but, as a tradeoff, they would decrease 3D image resolution. Moreover, newer generations of IV-OCT systems designed for the reduction of motion artifacts (e.g., non-uniform rotation distortion and relative motion of the imaging catheter to the vessel wall) [23], would be able to further improve the accuracy of 3D visualizations.

The method proposed in this study is able to segment and depict intracoronary stents in 3D through a fully automatic procedure. Although the method was developed to be robust with respect to common image artifacts and suboptimal image quality [19], automatic strut detection may generate false positive as well as false negative results. However, study results showed the accuracy of 3D stent visualization. Moreover, the automatic correction of false positives or the recovery of missed struts using spatial information would be a very challenging task, given that hundreds of different stent designs are currently present on the market [24].

As an additional consideration, no specific color was given to the guide-wire. Although when working with the on-line 3D rendering tool the guide-wire position is usually totally clear, the use of a dedicated color (e.g., green) would improve the quality of the 3D visualization even more.

For the rendering of an entire stent, a processing time of ~1 min is required. Even if this is suitable for the on-line use of the proposed method, specific customization of the proposed algorithms (e.g., GPU parallel computing implementation) would be able to further reduce the time of the computations, potentially allowing (close to) real-time 3D IV-OCT visualizations.

6. Conclusion

An algorithm for the automatic 3D visualization of IV-OCT data was developed and validated. Given the fact that the method is accurate and that it allows for the rapid assessment of entire pullbacks, it is suitable for the on-line application during coronary interventions. This potentially allows for an expanding use of 3D IV-OCT in clinical routine and PCI optimization, making 3D IV-OCT visualization available through the ‘push-of-a-button’ as a complimentary tool to conventional 2D images.

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