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Detection of Inflating Balloon in Optical Coherence Tomography Images of a Porcine Artery in a Beating Heart Experiment

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ABSTRACT

Suboptimal results of angioplasty procedures have been correlated to arterial damage during balloon inflation. We propose to monitor balloon inflation during the angioplasty procedure by detecting the balloon contours with intravascular optical coherence tomography (IVOCT). This will shed more light on the interaction between the balloon and the artery and to assess the artery's mechanical response. An automatic edge detection algorithm is applied for detection of the outer surface of an inflating balloon in a porcine artery in a beating heart experiment. A compliant balloon is inflated to deform the artery. IVOCT monitoring of balloon inflation is performed at a rate of 30 frames per second. During inflation, the balloon engages the arterial wall. Therefore, the characterization of the diameter of the inflated balloon leads to a characterization of the luminal diameter of the vessel. This provides precise information about the artery response to a simulated angioplasty procedure, information currently not provided by any other existing technique. In the current experiment, balloon inflation characterization is based on 356 IVOCT frames during which the estimated balloon diameter increases approximately from 1.8 mm to 2.9 mm.

Keywords: Intravascular optical coherence tomography, beating heart, edge detection.

1. DESCRIPTION OF PURPOSE

In the literature, excessive mechanical deformation of the arterial wall during angioplasty has been correlated to undesired outcomes, e.g. restenosis.¹ Development of a setup to study, in real-time, the deformation of the artery wall during balloon inflation can lead to improvement in balloon inflation strategies and in development of percutaneous coronary intervention (PCI) devices. With this aim, we propose to detect the contours of an inflating balloon using intravascular optical coherence tomography (IVOCT). In this paper, we demonstrate the method with images acquired in a porcine artery during a beating heart experiment. The beating heart model provides realistic conditions allowing for simulation of an angioplasty procedure.

Intravascular imaging techniques, such as intravascular ultrasound (IVUS) and IVOCT, provide resolutions higher than x-ray fluoroscopy, which is a silhouette imaging technique. Furthermore, IVOCT provides a resolution almost 10 times better than that of IVUS. Therefore, balloon detection in IVOCT images provides more precise measurements of different parameters, e.g. diameter, rendering it an efficient technique to achieve the described objectives.

The paper is organized as follows. In section 2, we first describe the structure of IVOCT images. We then demonstrate how an edge detection algorithm is applied to detect the balloon which is in contact with either blood or the artery wall. The algorithm is sufficiently fast to be applied in real-time. Characterization of the balloon diameter is presented in section 3. The concluding remarks are included in section 4.

2. METHODS

2.1 The beating heart setup

The beating heart model was recently presented as a valuable tool for the development of intravascular optical coherence tomography (IVOCT) technology.² In this paper, the same model is further used to explore the advantages of IVOCT monitoring during percutaneous coronary interventions. The beating heart model is helpful in generating conditions

similar to *in-vivo* conditions without the associated complications of a full animal preparation. The setup remains functional for several hours.

2.2 The swept-source optical coherence tomography (SS-OCT) system

A custom-built SS-OCT system is used for imaging.² We use a wavelength-swept laser source (Santec, HSL2000), operating with a sweep rate of 30 kHz and a sweep range of over 108 nm around 1.33 μm wavelength to provide a measured axial resolution of about 15 μm in air. The SS-OCT system is configured as a Mach-Zehnder interferometer with balanced detection and is packaged as a mobile unit.

The intravascular probe is illustrated in Fig. 1. The IVOCT probe is composed of a single-mode fiber enclosed within a spiral metallic tube in the proximal region and in a flexible polymer tube in the distal region. Near the tip of the catheter, the light exiting from the fiber is focused by a gradient-index (GRIN) lens and is redirected by a right-angle prism. The ensemble is enclosed in a stainless steel ferrule with an outer diameter of 0.7 mm. The probe assembly rotates in a liquid within a transparent polymer sheath. The polymer sheath has an internal diameter of 1 mm and an external diameter of 1.4 mm. SS-OCT imaging is typically performed at 20 to 30 fps. A compliant balloon is used to deform the artery. OCT imaging of the vessel walls is performed through the balloon. Inflation pressure applied to the balloon is usually less than 2 atm.

Figure 1. Intravascular probe (B: Balloon, F: ferrule, PS: polymer sheath, GRIN: gradient-index lens, LB: light beam, OF: optical fiber, PT: polymer tube, RAP: right-angle prism)

2.3 IVOCT images

Figure 2 presents a sample OCT image in polar coordinates (Fig. 2(a)) and Cartesian coordinates (Fig. 2(b)), illustrating the balloon inside the artery. In this figure, the balloon is not yet inflated. This measurement was performed in a beating heart setup with a heart perfused with a mixture of blood and Krebs-Henseleit solution. Therefore, the arterial wall is not visible at many angles, where the blood is occluding the view by scattering the light. The polymer sheath and the balloon are represented each by two contours, corresponding to their inner and outer surfaces.

In order to process and analyze the OCT image, we use the polar coordinates (Fig. 2(a)), which is the format in which the image is acquired. Figure 2(a) presents the grayscale image corresponding to the acquired intensity matrix. The horizontal and the vertical axes correspond to the probe rotation angle and the radial depth, respectively. The primary axes represent pixel values. Each column in the image matrix represents depth scanning at a particular angle and is called an A-scan. Two sample A-scans, A-scan "A" and A-scan "B" are depicted by red lines in Fig. 2. The nodes on these A-scans correspond to the detected edges, which will be described later. A-scan A provides a sample depth profile for the segment of the image where visualization of the vessel wall is partially obstructed by blood. A-scan B provides a sample depth profile for the segment of the image where the balloon is in the vicinity of the vessel wall and where there is almost no blood trapped in between. In this segment, better visualization of the vessel wall structure is provided; the vessel wall layers, namely, intima, media and adventitia can be distinguished, as demonstrated in Fig 2(a).

Assuming that the probe rotates uniformly, the number of acquired A-scans in a full rotation determines the angle between each two adjacent A-scans, or the angular resolution. In our images, there are about 1000 A-scans per rotation. The transverse pixel size at 1 mm depth in true geometrical distance is approximately 6 μm . The radial pixel size in the presented images is 8 μm in optical length, i.e. the product of geometrical length and the refractive index. The values on the secondary horizontal and vertical axes represent the angle in degrees and the depth in millimeters, respectively.

Figure 2. IVOCT image inside the artery when the balloon is not inflated

2.4 Balloon detection

To detect the balloon, we focus on the outer surface of the balloon. We localize the edge that is formed due to contrast in brightness between the region inside the balloon and the surrounding structure, which is either the blood or the vessel wall, depending on the segment of the image being processed. The edge is estimated by a contour, composed of n nodes. These nodes can be detected on all A-scans. Therefore, the value of n can be as large as the number of A-scans. For real-time application, detection of fewer nodes provides sufficient precision and increased processing speed. In the results of this paper 24 points are detected for each contour.

Let us assume that j_k , where $k=0, 1, 2, \dots, n$, is the index to the A-scans to be processed to extract the contour nodes. On each A-scan, a depth index, i_k , should be estimated, in order to fully determine the coordinates of the contour node.

The steps to provide i_k are described as follows:

1. We select n A-scans, distributed at equal angles over a full rotation. Figure 3 demonstrates two sample A-scans, which are used to illustrate the detection steps. These are A-scan A and A-scan B previously illustrated in Fig. 2.
2. Each element of a selected A-scan is averaged with the elements at the same depth of the n_{av} neighboring A-scans on each side of the selected A-scan. This may slightly decrease the precision of localization of the edge. However, it increases the robustness of detection and is an essential part of the algorithm. This step is particularly important when the selected A-scan is located where there is discontinuity in the image due to deflection of light. In Fig. 3, it can be seen that application of this step to the selected A-scan more clearly defines different regions, e.g. blood, vessel wall. The prism surface is represented by R, since it is used as the reference for automatically adjusting the depth in each image. The depth corresponding to the prism surface is determined by the known thickness of the prism. The prism surface is automatically detected in each image by a peak detection technique.
3. In the next step a median filter with a coarse scale is used to keep the edge of interest and eliminate the unwanted ones. For example, this step removes the strong edge between media and adventitia in Fig. 3(b), which is not sought in this paper. It also eliminates single peaks, such as those corresponding to the prism surface and the inner surface of the balloon. This step leaves a pulse corresponding to the blood and the vessel wall, from which the edge of interest can be extracted. The use of a median filter is preferred over other smoothing filters since it can eliminate single peaks without deteriorating edge characteristics, significantly.
4. In this step, we detect the low-to-high edge of the pulse, generated by the median filter. As can be seen in Fig. 3, this edge can be detected as the maximum of the gradient of the resulting signal, obtained from the previous step. The detected edges for the two sample A-scans are represented by red dots in Fig. 2.

Figure 5 demonstrates the characterization of balloon diameter values, based on area of the detected contour, for all 356 images. For diameters larger than 2.4 mm, the contact between the balloon and the artery is revealed by oscillations in the measured diameter, a clear signature of the artery contractions.

Figure 4. Detection of the balloon in OCT images during the balloon inflation process

Figure 5. Estimated balloon diameter versus time and image index

4. CONCLUSIONS

In this paper, we proposed balloon detection as a helpful feature to improve the balloon inflation process. The results can be advantageous both in cardiovascular research and in the development of PCI devices. Balloon detection can help assess the mechanical response of the artery. Although no stent was used in our experiments, balloon detection can be applied to monitor stent deployment. The high resolution, provided by OCT, together with different segmentation algorithms result in a detailed and accurate assessment of tissue deformation. This motivates further research addressing segmentation of IVOCT images, which may lead to a greater role for the OCT technology in cardiovascular research.

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