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A physics-based intravascular ultrasound image reconstruction method for lumen segmentation

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ABSTRACT

Intravascular ultrasound (IVUS) refers to the medical imaging technique consisting of a miniaturized ultrasound transducer located at the tip of a catheter that can be introduced in the blood vessels providing high-resolution, cross-sectional images of their interior. Current methods for the generation of an IVUS image reconstruction from radio frequency (RF) data do not account for the physics involved in the interaction between the IVUS ultrasound signal and the tissues of the vessel. In this paper, we present a novel method to generate an IVUS image reconstruction based on the use of a scattering model that considers the tissues of the vessel as a distribution of three-dimensional point scatterers. We evaluated the impact of employing the proposed IVUS image reconstruction method in the segmentation of the lumen/wall interface on 40 MHz IVUS data using an existing automatic lumen segmentation method. We compared the results with those obtained using the B-mode reconstruction on 600 randomly selected frames from twelve pullback sequences acquired from rabbit aortas and different arteries of swine. Our results indicate the feasibility of employing the proposed IVUS image reconstruction for the segmentation of the lumen.

1. Introduction

Intravascular ultrasound (IVUS) refers to the medical imaging technique consisting of a miniaturized ultrasound transducer located at the tip of a catheter that can be introduced in the blood vessels providing high-resolution, cross-sectional images of their interior. The ultrasound transducer transmits sound pulses and receives an acoustic radio frequency (RF) echo signal (i.e., A-line) at a discrete set of angles (commonly 240–360). These signals are processed to reconstruct an image that resembles the interior of the vessels and therefore is easier to interpret by the physicians (i.e., B-mode image). The B-mode reconstruction process consist of the detection of the positive envelopes of each A-line, the application of a time-gain compensation function (TGC), the quantization of the signal, the compression of the dynamic range, the stacking of the signals along the angular direction, a 8-bit gray scale mapping, and finally a geometric transformation from polar to Cartesian coordinates.

The segmentation of the lumen/intima and media/adventitia regions in the IVUS images is an important task necessary to evaluate the degree of stenosis and to identify the characteristics of the atherosclerotic plaque. In practice, manual segmentation of IVUS images may be performed by a trained observer. However, depending on the type of analysis to be performed, the number of frames to be segmented can range from a few hundred to thousands of frames. Therefore, many methods have been proposed for the automatic segmentation of the regions of interest on IVUS data.

Recent approaches for automatic or semi-automatic IVUS image segmentation include computational methods that employ state-of-the-art techniques such as shape-driven approaches [1], deformation of parametric and geometric models [2–5], methods based on wavelet analysis [6,7], multi-class classification methods [8], fast-marching methods [9,10], binary morphological object reconstruction [11], brushlet expansion [12], temporal texture analysis [13], random walks [14], and k-means combined with expectation-maximization approaches [15]. An extensive review of existing segmentation algorithms for IVUS images up to 2012 has been provided by [16]. An evaluation framework that allows a standardized and quantitative comparison of IVUS lumen and media segmentation algorithms was introduced at the MICCAI 2011 Computing and Visualization for (Intra)Vascular Imaging (CVII) workshop [17], which compared the results of eight segmentation methods, including one of our works [5].

Despite all these efforts, the segmentation of IVUS data remains an open problem due to the different challenges present in the

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IVUS data. For example: (i) the variability of the morphology of the lumen; (ii) the various possible compositions of the plaques; (iii) the IVUS artifacts (i.e., speckle noise, ring down artifact, and shadows); (iv) the variability in the IVUS acquisition hardware (i.e., frequency of operation, ultrasound transducer diameter, catheter type, etc.); and (v) the variability on the gray level distributions of the different regions of the vessel depicted in the IVUS images. This variability depends on the B-mode reconstruction settings of the IVUS systems, such as time-gain compensation, dynamic range compression and rejection, persistence, and gamma curves, which are subjectively selected by the interventionist and may change from one intervention to the next, or midway through an intervention [18,19].

A common characteristic shared by many current segmentation methods is that the segmentation is performed using the B-mode reconstruction images. However, the B-mode image reconstruction procedure does not account for the physics involved in the generation of the A-line signals. For example, the size of the transducer and the frequency of operation of the IVUS system determine the divergence of the ultrasound beam. The divergence of the ultrasound beam affects the intensity and distribution of the speckle of the received RF signal because the beam interacts with more scatterers as it moves away from the transducer (Fig. 1). These differences in the gray level distributions of regions corresponding to the same type of tissue may represent a challenge for the automatic segmentation of the different areas of the vessel.

In this paper, we present a novel method to generate an alternative mode of IVUS images that is based on the estimation of the differential backscattering cross-section (DBC) of the scatterers in the vessel from the RF IVUS signal. For this, we employ a physics-based model of the IVUS signal scattered by the structures of the vessel under inspection. We assess the impact of applying the proposed DBC-mode reconstruction for the segmentation of the lumen by comparing the performance of an existing lumen segmentation method when computing texture features from the B-mode and the proposed DBC-mode images.

2. Materials and methods

2.1. Scattering model

Sound waves are mechanical disturbances that move as pressure waves through a medium. The intensity I of a sound wave is defined as the average power carried by a wave per unit area normal to the direction of propagation of the wave over time [20]. The speed of an ultrasound wave c with frequency f through a medium is determined by the density ρ of the medium, while the distance covered by one cycle of the wave is the wavelength $\lambda = \frac{c}{f}$. As a sound wave passes from one medium into another, a portion of the incident wave is reflected at the boundary and another portion spreads in the second medium.

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As an ultrasound beam propagates through a heterogeneous medium, part of its energy is removed from the beam as a function of distance by reflection, scattering, geometric attenuation, and absorption. The attenuation of an ultrasound wave in a medium depends on the frequency of the wave and is described by the attenuation coefficient μ which is the sum of the individual coefficients for scattering and absorption in units of decibels per centimeter. A simple phenomenological model used in practice to write the intensity $I(r)$ at a distance r from the transducer is:

$$I(r) = I(0)e^{-\mu r}, \tag{1}$$

where $I(0)$ corresponds to the initial intensity of the ultrasound beam.

For a standard disc-shaped transducer (Fig. 2), initially the beam is of comparable diameter to the transducer D as the series of ultrasound waves that make up the beam travel parallel to each other. This is known as the near-field or Fresnel zone F_z and can be computed as:

$$F_z = \frac{D^2}{4\lambda}. \tag{2}$$

Beyond the Fresnel zone, some of the energy escapes along the periphery of the beam to produce a gradual divergence of the ultrasound beam. In this region, called the far-field zone, the axial pressure decreases approximately according to $1/r$ [20]. The angle of divergence $\Delta\Theta$ (in degrees) of the beam can be computed as:

$$\sin(\Delta\Theta) = 1.22\frac{\lambda}{D}. \tag{3}$$

The structures in the vessel imaged by IVUS, such as collagen fibers or red blood cells (RBC), are smaller than the wavelength of the ultrasound wave. Such small structures produce scattered waves that return to the transducer through multiple pathways. The sound that returns to the transducer from such non-specular reflectors is no longer a coherent beam. It is instead the sum of

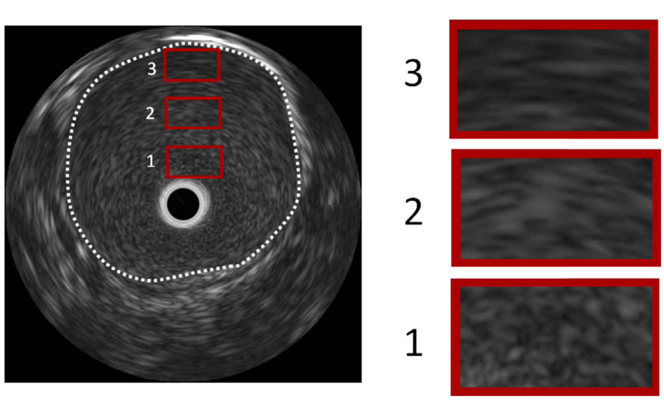

Fig. 1. Example of different gray level distributions for regions corresponding to lumen. Note how density of speckle reduces as the region moves farther away from the transducer. The dotted line indicates the region corresponding to lumen.

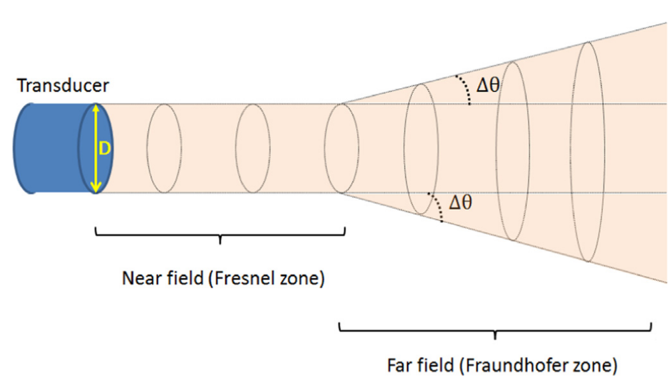

Fig. 2. Characteristics of the ultrasound beam for a standard disc-shaped transducer.