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Error propagation in the characterization of atheromatic plaque types based on imaging

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ABSTRACT

Imaging systems transmit and acquire signals and are subject to errors including: error sources, signal variations or possible calibration errors. These errors are included in all imaging systems for atherosclerosis and are propagated to methodologies implemented for the segmentation and characterization of atherosclerotic plaque. In this paper, we present a study for the propagation of imaging errors and image segmentation errors in plaque characterization methods applied to 2D vascular images. More specifically, the maximum error that can be propagated to the plaque characterization results is estimated, assuming worst-case scenarios. The proposed error propagation methodology is validated using methods applied to real datasets, obtained from intravascular imaging (IVUS) and optical coherence tomography (OCT) for coronary arteries, and magnetic resonance imaging (MRI) for carotid arteries. The plaque characterization methods have recently been presented in the literature and are able to detect the vessel borders, and characterize the atherosclerotic plaque types. Although, these methods have been extensively validated using as gold standard expert annotations, by applying the proposed error propagation methodology a more realistic validation is performed taking into account the effect of the border detection

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algorithms error and the image formation error into the final results. The Pearson's coefficient of the detected plaques has changed significantly when the method was applied to IVUS and OCT, while there was not any variation when the method was applied to MRI data.

1. Introduction

Diagnostic imaging plays an important role in medicine today. Delineation of vessels solves practical applications such as diagnosis of coronary artery disease and assessment of the degree of stenosis in carotid artery pathology [1]. Advantages in imaging technology led to a wide variety of imaging modalities for depicting the silhouette and the interior of blood vessels. These modalities can be either invasive [2,3], i.e. itravascular imaging (IVUS), optical coherence tomography (OCT), near-infrared spectroscopic imaging (NIRS), etc., or non-invasive [4,5], i.e. X-ray angiography (XRA), magnetic resonance imaging (MRI), computed tomography (CT), etc., each one having its advantages and disadvantages [6].

Imaging systems perform measurements and all measurements have uncertainties due to the equipment imperfection, human imprecision, instrumental calibration, etc. These uncertainties [7] describe how accurately a mathematical model can represent the true physics, and are determined after system tests and by taking into account the measurement procedure to estimate its reliability. To estimate the error of a system (group of instruments), the accuracy of a specified output approximated by a given numerical method is measured. Imaging systems [8] transmit and acquire signals using a detector in order to construct the image. However, all imaging systems are subject to error sources and the detected signals that construct the image include the variations of the detector or possible calibration errors [9]. Those artifacts can affect image quality [10,11], even though some of them are corrected by medical equipment software, resulting sometimes in false image representation. These artifacts, along with the image quality can affect automated or semi-automated computer-aided image processing methods. Spatial resolution [12] is a major determinant of image quality and describes the level of detail that can be distinguished on an image. Spatial resolution relates to how small an object can be in order to be detected by a particular imaging system. Imaging resolution of a digital image can be described in many different ways, including spatial resolution. Therefore, spatial resolution of an imaging system affects its resolution and the ability to resolve vessel microstructures; consequently, producing an error in the final image. Image formation errors include the limitations produced both by the pixel resolution (spatial resolution) of the image acquisition system and by any other possible systematic errors. Spatial resolution, usually is reported in line pairs per centimeter (lp/cm) and provides an adequate metric of the uncertainty implied by imaging system's formation errors.

The latest advances in computer technology and the need for automation of the diagnosis of vascular or carotid artery disease have made possible the development of automated methods for processing vessel images derived from various imaging modalities. These methods can be mainly grouped to methods segmenting the vessels [13–15] and methods characterizing the atherosclerotic plaque [16–20] of the segmented vessels, regardless of the imaging modalities used. Segmentation methods vary depending on the imaging modality or method being automated or semi-automated. Vessel segmentation methods are the key components of the automated plaque characterization methods; the results of a segmentation method are used as input in order for the plaque characterization method to be applied. However, all the plaque characterization methods have a serious drawback: plaque characterization and segmentation steps are validated independently. This leads to overestimated results as the segmentation error computed in the validation step and the acquired system image formation errors are not propagated into the final result.

In this work, a methodology to study the error propagation for both segmentation step and image formation is presented. The proposed error propagation methodology is validated using three different plaque characterization methods [16,17,20]. The first method [17] processes IVUS images and segments the artery using a semi-automated border detection method [13], based on the use of deformable models [21]. The pixels of the segmented area are then classified to dense calcium (DC), necrotic core (NC), fibrous tissue (FT), fibro-fatty tissue (FFT) and normal tissue-media (M), using a classification algorithm. The second method [20] uses active contours to detect the luminal borders in MRI images. It implements a knowledge-based algorithm to characterize the lipid core (LC) within the segmented area. Finally, the third method [16] segments and characterizes the artery automatically by classifying the plaque in four different plaque types: calcified plaque (Ca), lipid plaque (LP), fibrous tissue (FT) and mixed plaque (MP).

Although, the methods [16,17,20] have been validated thoroughly, the segmentation step error and the image formation error to the final plaque characterization results have not been taken into account. Currently, the proposed methodology is applied to present a more thorough and realistic validation. Specifically, for the IVUS method the influence of segmentation error and the image formation error to the detection of DC, NC, FT, FFT, M and the entire plaque area are validated. For the MRI approach, the influence of both segmentation error and the image formation error to the detection of LC are validated. Similarly to IVUS, for OCT the influence of both segmentation and image formation error to the detection of Ca, LP, FT, and MP are validated.

Although the plaque characterization results are affected by the propagated errors, still the results of the three methods [16,17,20] are accurate enough (Pearson's coefficient > 0.86 in all plaque types) to be used for research or clinical applications.