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Robust segmentation methods with an application to aortic pulse wave velocity calculation

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ABSTRACT

Aortic stiffness has proven to be an important diagnostic and prognostic factor of many cardiovascular diseases, as well as an estimate of overall cardiovascular health. Pulse wave velocity (PWV) represents a good measure of the aortic stiffness, while the aortic distensibility is used as an aortic elasticity index. Obtaining the PWV and the aortic distensibility from magnetic resonance imaging (MRI) data requires diverse segmentation tasks, namely the extraction of the aortic center line and the segmentation of aortic regions, combined with signal processing methods for the analysis of the pulse wave. In our study non-contrasted MRI images of abdomen were used in healthy volunteers (22 data sets) for the sake of non-invasive analysis and contrasted magnetic resonance (MR) images were used for the aortic examination of Marfan syndrome patients (8 data sets). In this research we present a novel robust segmentation technique for the PWV and aortic distensibility calculation as a complete image processing toolbox. We introduce a novel graph-based method for the centerline extraction of a thoraco-abdominal aorta for the length calculation from 3-D MRI data, robust to artifacts and noise. Moreover, we design a new projection-based segmentation method for transverse aortic region delineation in cardiac magnetic resonance (CMR) images which is robust to high presence of artifacts. Finally, we propose a novel method for analysis of velocity curves in order to obtain pulse wave propagation times. In order to validate the proposed method we compare the obtained results with manually determined aortic centerlines and a region segmentation by an expert, while the results of the PWV measurement were compared to a validated software (LUMC, Leiden, the Netherlands). The obtained results show high correctness and effectiveness of our method for the aortic PWV and distensibility calculation.

1. Introduction

Aortic stiffness is an important factor in estimating the cardiovascular risk in several disease conditions. The pulse wave velocity (PWV) is a good indicator of the aortic stiffness in patients with hypertension [1], Marfan syndrome [2,3], metabolic syndrome [4], Diabetes [5], etc. The PWV is a cardiovascular parameter which is intensively studied [6] both in humans and animals [7]. Aortic PWV is a strong predictor of cardiovascular events and all-cause mortality [8]. The PWV is measured using various techniques [9,10]. The main idea behind the PWV calculation is to track the propagation of the pulse wave from the ascending level of the aorta to its abdominal level in order to obtain the transition speed.

We classify approaches for the PWV measurement into three categories: the blood pressure, ultrasound Doppler and magnetic resonance (MR) techniques. The blood pressure PWV analysis is done using a cuff sphygmomanometer on a peripheral limb, where the measured values reflect the pressure throughout the arterial tree in large conduit arteries. The commercially available hardware (TensioClinic Arteriograph) [11] uses the time interval between the pulse wave (the early systolic peak) and its reflection from the abdominal aortic bifurcation (the late systolic peak). However, the PWV values are obtained using an estimate of the aortic

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length between the jugulum and the symphysis, which introduces inaccuracies in measurements. The pulsed wave (PW) ultrasound Doppler hardware (e.g. Siemens, Philips) uses the Doppler effect to determine the transition time of the pulse wave in the aorta. However, the length of the aorta still needs to be estimated, resulting in a lower accuracy of the PWV measures. The same approach is taken if a heart sound sensor is used.

Various approaches exist to PWV calculation using MR images. The transit time (TT) approach requires modulus and phase-contrast images at various aortic levels to calculate the velocity curves and a whole MR abdomen and thorax image to calculate aortic lengths between the given levels. The PWV is calculated as the speed of wave propagation. Often only two cross-sectional slices are required for calculating the velocity wave profiles, where the aortic length is measured from an oblique sagittal slice. The time interval between pulse wave is often determined by the "foot" of the wave curve [12]. Algorithms based on finding the maximum (or the minimum) velocity of the pulse wave depend on the assumption that the sampling rate at which the images were taken is sufficiently high to accurately capture the peak of the wave. Similar approach is to define the wave arrival moment as the time instant in which the wave reaches its mid-range value (the average of the minimum and the maximum value). The flow area (QA) approach [13] uses one data set at one site across the aorta (cross-sectional through-plane velocity encoding), where the aortic area is acquired from magnitude images and the flow from phase-contrast images. The PWV is estimated as the ratio of difference in the total (the maximum and the minimum) flow and difference in the total area at early systole. The extension to TT approach is a *multisite method* that uses a single para-sagittal slice of aorta, which allows for obtaining flow waveforms at multiple locations along the aorta. The multisite method [14] uses the maximum velocity change (at a single cross section) to define the arrival moment of the pulse (at the given cross section). The cross-correlation (XC) approach [15] is a multisite method that uses cross-correlation to determine the time interval between flow waves at different locations along the aorta. A flow-sensitive 4D MRI approach for PWV measurements has also been proposed [16]. An axial velocity profile method for estimating the PWV of the descending aorta was developed in [17], which allows visualization of the pulse wave propagation. Apart from these, methods based on deformable surfaces are applied to the segmentation of modulus images for an aortic distensibility calculation [18]. The comparison of MR PWV calculation approaches was done in [9] and concluded that TT and XC methods result in a closer and a more reproducible aortic PWV measurement than in the QA method.

The current reviews on vessel extraction techniques [19,20] show a wide variety of segmentation methods developed for angiographic vessel images using level-sets [21], machine learning [22], edge detection [23], etc. "Black-blood" vessel extraction methods show good results on segmenting the vessel wall, usually with some user interaction needed: [24] uses discrete dynamic contours, [25] uses deformable models with the Markov Random Fields and a multiscale method is used in [26], robust enough to deal with slice discontinuities and blood flow artifacts. However, the method is not computationally efficient (it needs 8 min on average to segment a single MRI data set). The method of [27] (MiaLite application) is optimized for the segmentation of the abdominal aorta and uses a modified sparse field level set method, with a periodic monotonic speed function, resulting in coherent propagation of the contour boundary. The ITK-SNAP application [28] implements two well-known 3-D active contour segmentation methods: Geodesic Active Contours [29] and Region Competition [30]. The method of [31] (implemented in 3D Slicer) uses seed points to extract the local robust statistics to describe the object features. It evolves several active contours simultaneously with their interactions being motivated by the principles of action and reaction converging to equilibrium.

The main idea in our work is to design a complete PWV analysis software tool for subjects scanned without injection of contrast agent (for aorta check-up purposes), as well as for patients (studied with contrast images) [32]. First, we propose a novel method for extracting the centerline of the abdominal aorta in contrasted and black-blood MR images by combining graphs and our previous work on multiscale profiling [33]. Next, we introduce a method for the segmentation of the aortic region in modulus images by selection of candidate regions obtained using projections [34]. Finally, we propose a novel method for the analysis of the pulse wave by taking into account its steepest slopes and deformation over time. In order to validate our algorithm, we compare our results to the results of already validated method for pulse wave analysis using the FLOW package for analysis of phase-contrast images and an in-house developed PWV tool [35–37] (LUMC, Leiden, the Netherlands).

2. Materials and methods

2.1. Problem definition and data sets

Data sets for the PWV calculation consist of a 3-D MRI image (as a series of 2-D slices) of an abdomen and thorax and series of modulus and phase-contrast images at different aortic levels represented as time series of 2-D slices. The number of aortic levels at which the Cine series are recorded is arbitrary (unlimited). In this paper, we use four different aortic positions: ascending, descending, diaphragmal and abdominal levels (see Fig. 1). The main idea of our research is to develop an image processing toolkit for the PWV calculation for contrast enhanced or non-contrast enhanced MRI images. Hence, the experiments are conducted on images of healthy volunteers (22 data sets) and Marfan syndrome patients (8 data sets) obtained from the Ghent University Hospital (UZ Gent). We will now describe the used Cine imaging and, in succession, structural 3-D imaging used for volunteers and patients.

Fast imaging with steady-state free precession (TrueFISP) modulus images are acquired as a retrospective electrocardiogram (ECG) gated scan with reconstruction of 40 images per RR-interval, slice thickness 6 mm, repetition time (TR) 26 ms, echo time (TE) 1 ms, flip angle (FA) 80°, field of view (FOV) 240 mm × 320 mm and matrix size 192 × 256 pixels adjusted to body habitus to avoid ghosting artifacts (pixel size 1.25 mm × 1.25 mm).

Fast Low Angle Shot (FLASH) 2-D phase-contrast images were acquired as a retrospective ECG gated scan with reconstruction of 40 images per RR interval, slice thickness 6 mm, TR = 61 ms, TE = 3 ms, FA = 30°, velocity encoding (VENC) of 150 cm/s, where the matrix size and slice position were adjusted to the TrueFISP image allowing their mapping.

The 3-D MR structural images of healthy volunteers were acquired using the Half Fourier Acquisition Single Shot Turbo Spin Echo (HASTE, properties: slice thickness 6 mm, TR = 700 ms, TE = 26 ms, FA = 160°, FOV 233 mm × 340 mm and a matrix of 176 × 256 pixels adjusted to a body habitus, pixel size 1.32 mm × 1.32 mm) technique on a 1.5 T scanner using a high inter-slice spacing of 6.6 mm without injected contrast agent, yielding low quality "black-blood" images (see Fig. 1). The black-blood aorta images pose a hard segmentation problem due to unclear boundaries between the aorta and surrounding organs (veins, heart and lungs). However, the advantage of such scanning process is the increased acquisition speed with a non-invasive approach.

The 3-D MR structural images of patients were obtained using Fast Low Angle Shot (FLASH, properties: slice thickness 0.9 mm,