Automatic Measurement of Carotid Diameter and Wall Thickness in Ultrasound Images

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Abstract

Carotid vessel ultrasound imaging is a reliable non-invasive technique to measure the arterial morphology. Lumen Diameter (LD), intima-media thickness (IMT) of the far wall, and plaque presence can be reliably determined using B-mode ultrasound.

In order to measure the carotid IMT as well as any other more complex quantitative indices of vessel morphology, it is necessary to identify lumen-intima and media-adventitia borders in the ultrasound images.

In this paper we describe an automatic approach to measure LD and IMT based on an active contour technique improved by a multiresolution analysis.

The measurements of LD and IMT were compared to manual tracing of the vessels border in terms of coefficients of variability (CV) and correlation (R). The results have shown that the method is a reliable and reproducible way of assessing the LD and far wall IMT in the carotid artery.

1. Introduction

Ultrasound imaging is widely used to depict carotid, brachial, femoral and other peripheral arteries. There are several major advantages of using ultrasound in comparison to other techniques. Most importantly, B-mode ultrasound imaging is non-invasive and allows real-time visualization of the arterial morphology that is not currently possible with any other imaging modality[1,2].

Moreover, the non-invasive approach provided by B-mode ultrasound imaging and its low cost has allowed the use of this technique for large-volume clinical studies that demonstrates the direct relationship between the carotid intima-media thickness and cardiovascular disease[3].

Measurements of lumen diameter (LD) and intima-media thickness (IMT) of carotid and femoral arteries from B-mode ultrasound are defined as the average distance of interfaces between vessel tissue layers. In order to determine the interface location, a manual tracing is commonly used. However, this approach is a time consuming procedure and based on subjective operator assessment that inevitably results in inter and intra-observer variability.

Efforts have been made to make the measurement less operator dependent by introducing automated image analysis procedures[4,5].

In this paper we describe an automatic approach to measure IMT and LD based on an active contour technique improved by a multiresolution analysis. A series of carotid images using B-mode ultrasound is acquired during systole and diastole. The operator selects a region-of-interest (ROI) in this set of images that is convolved with the corresponding partial derivatives of the Gaussian filter. This process is repeated using different filter spatial resolution – or scales – in both x and y directions. The first order derivative images are used to compute a 2D gradient magnitude image in order to extract the vessel's boundaries. Using an active contour technique, the vessel's border is determined automatically.

The user can also interactively adjust the detection in regions where the border has not been properly identified due to lower image quality. The near wall media-adventitia (NWMA), far wall media-adventitia (FWMA) and far wall lumen-intima (FWLI) borders are obtained by a least-square fitting of the active contours result. The distance between NWMA and FWLI (vessel diameter) and between FWLI and FWMA (far wall intima-media thickness) are obtained for all images and the mean value is computed during systole and diastole.

2. Methods

Non-invasive ultrasonic B-mode imaging is an increasingly important method for studying progress and regress of atherosclerotic lesions in the carotid artery.

Figure 1 shows a representative B-mode ultrasound image of the carotid artery and a schematic illustration of the relevant leading edges of echo responses. Previous studies[4,5,6] have shown that the leading edges can be mapped to the following interfaces: near-wall media-adventitia, far-wall lumen-intima and far-wall media-

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adventitia. The LD is defined as the distance between the media-adventitia interface of the near-wall and the lumen-intima interface of the far-wall. The far-wall IMT is defined as the distance between the far-wall lumen-intima and the far-wall media-adventitia interfaces.

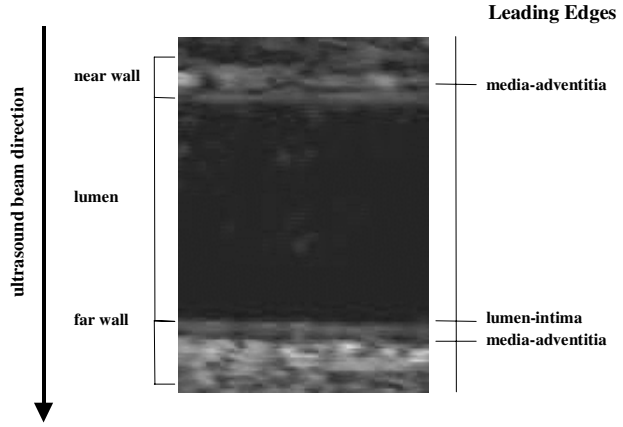

Figure 1 – Interfaces between carotid tissue layers obtained from B-mode ultrasound

The determination of ultrasonic measurement of the artery becomes equivalent to accurately detecting the echo boundaries presented in Figure 1. However, the existence of ultrasonic imaging artifacts such as speckle, reverberations and dropouts make the accurate definition of a boundary very difficult.

2.1. Artery boundary enhancement

To enhance border detection accuracy, a multiscale border identification was implemented using filters in the form of scaled convolution operators[7,8]. The scale space of an image is constructed through convolution of the image with a two-dimensional (2D) Gaussian density kernel with zero mean and standard deviation:

$$G(\vec{x},\sigma)=\frac{1}{\sqrt{2\pi\sigma^2}^{D}}e^{-\frac{\|\vec{x}\|^2}{2\sigma^2}} \qquad (1)$$

where D denotes the dimension of the input domain. A blurred replica of the original image is obtained by convolution with $G(\vec{x};\sigma)$ for a specific σ. The stack of images as a function of increasing scale parameter σ is coined a linear scale space. Hence, as σ increases, the detailed object structures vanish, while gross structures persist.

Based on these features a scaled artery image is used to identify the approximated position of the near and far walls. Two complementary images, based on the gradient value in y-direction are obtained: one that enhances pixel values transitions from high to low echoes, such as edges encountered in near wall tissue interfaces, and other that enhances pixel values transitions from low to high echoes (such as edges encountered in far wall tissue interfaces). Figure 2 shows the boundary enhancement of the near and far wall.

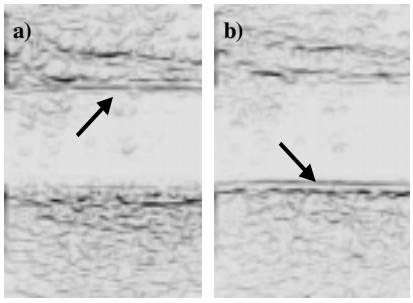

Figure 2 – Boundary enhancement of the near wall a) and far wall b)

2.2 Contour modeling

The contour of each wall is modeled following the Geometrically Deformed Model proposed by Lobregt and Viergever[9]. In this model, a set of vertices which are connected by straight line segments or edges forms the basic contour structure (Figure 3).

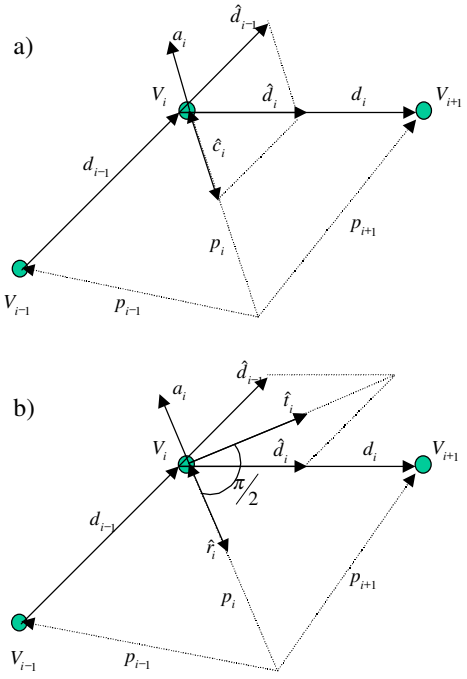

Figure 3 – Contour model consisting of a set of vertices V_i which are connected by segments or edges.

The position of a vertex V_i is represented by a vector p_i, and the edge between V_i and V_{i+1} by a vector d_i. The contour deformation is caused by a combination of forces which act on the vertices. The resulting

acceleration in vertex V_i is denoted by a vector a_i.

The contour local curvature at a vertex V_i is defined as the difference between the directions of the two edge segments that join at that location:

$$\hat{c}_i = \hat{d}_i - \hat{d}_{i-1} \tag{2}$$

The local tangential unit vector is defined as the normalized sum of the unit vectors of two joining edge segments:

$$\hat{t}_i = \frac{\hat{d}_i + \hat{d}_{i-1}}{\left\| \hat{d}_i + \hat{d}_{i-1} \right\|} \tag{3}$$

The local radial direction at a vertex V_i is obtained from $\hat{t}_i$ by a rotation over $\pi/2$ radians:

$$\hat{r}_i = \begin{bmatrix} 0 & 1 \\ -1 & 0 \end{bmatrix} \hat{t}_i \tag{4}$$

2.3. Dynamic force formulation

In the model definition, the dynamic in each vertex V_i must satisfy the Newton´s second law,

$$\begin{aligned} F_i &= F_{\text{int},i} + F_{ext,i} + F_{damp,i} \\ F_i &= \mu_i a_i \end{aligned} \tag{5}$$

where μ_i is a coefficient that has a mass unit, $F_{damp,i}$, $F_{\text{int},i}$ and $F_{ext,i}$ and are the damping (or viscous), the internal and the external forces, respectively.

The internal force can be estimated from the local contour curvature along the local r-axis :

$$F_{\text{int},i} = \left(c_i \cdot \hat{r}_i\right) \tag{6}$$

The external force acting in each vertex can be approximated by some image feature. In this paper we used the information obtained from the local image gradient as the external force.

The damping force is proportional to the velocity of the vertex and points in opposite direction:

$$F_{damp,i} \approx -k.v_i \tag{7}$$

The total force F_i acting on a vertex is a weighted combination of damping, internal and external forces :

$$F_i = w_{\text{int}} F_{\text{int},i} + w_{ext} F_{ext,i} + w_{damp} F_{damp,i} \tag{8}$$

where w_{int}, w_{ext} and w_{damp} are the weighting factors.

The deformation process over the contour is implemented as a numerical time integration process in which the complete state of the contour is calculated at a sequence of discrete positions in time [9].

3. Evaluation procedure

In this study a total of 180 images from 30 patients (3 images in diastole and 3 images in systole for each patient) were analyzed, all of which included manually defined interfaces for reference. The minimum and maximum artery diameter were measured for each patient using the manual and the automatic procedure. However, for clinical purposes, some interactive tools for manual tracing were incorporated to the model in order to correct remaining detection errors in regions with poor image quality.

In order to study the variability between the automatic and manual definition of artery boundaries, the pooled mean, $\overline{\mu}$, and the standard deviation, σ, for the difference between automated and manual measurements of lumen diameter were computed. The coefficient of variation, CV, was calculated according

$$CV = \left(\frac{\sigma}{\overline{\mu}\sqrt{2}} \times 100 \right)\% \tag{10}$$

The strength of the relationship between automated and manual methods is indicated by the correlation, $R_{a,m}$, between the two measurements:

$$R_{a,m} = \frac{Cov_{a,m}}{\sigma_a . \sigma_m} \tag{11}$$

where $Cov_{a,m}$ is the covariance between the automated and manual and σ_a and σ_m are the standard deviation of automated and manual measurements, respectively.

4. Results

Means ($\mu_{a,m}$) and standard deviations ($\sigma_{a,m}$) for the differences between the automatic and manual methods were calculated for the population (n=30). The coefficients of variability ($CV_{a,m}$) and the correlation ($Corr_{a,m}$) between both methods were also obtained.

The results obtained for the parameters $\mu_{a,m}$, $\sigma_{a,m}$, $CV_{a,m}$ and $Corr_{a,m}$ are summarized in Table 1.

Table 1.Lumen diameter (LD) and Intima-Media thickness measured using automatic and manual methods (n=30). The difference, Δ, the coefficient of variability, CV and the correlation, $Corr_{a,m}$, between both measurements are also presented

	Automatic $\mu_a \pm \sigma_a$ (mm)	**Manual** $\mu_m \pm \sigma_m$ (mm)	**Difference** Δ $\mu_{a,m} \pm \sigma_{a,m}$ (mm)	**Variability** CV (%)	**Correlation** $Corr_{a,m}$
Lumen Diameter (diastole)	$7{,}85 \pm 1{,}01$	$7{,}78 \pm 1{,}01$	$0{,}13 \pm 0{,}09$	0,83	0,99
Lumen Diameter (systole)	$6{,}81 \pm 1{,}06$	$6{,}77 \pm 1{,}05$	$0{,}12 \pm 0{,}10$	1,00	0,99
Intima Media Thicknes	$0{,}72 \pm 0{,}14$	$0{,}63 \pm 0{,}12$	$0.09 \pm 0{,}06$	6,16	0,90

5. Discussion

Measurements of lumen diameter (LD) and intima-media thickness (IMT) of carotid and femoral arteries from B-mode ultrasound are defined as the average distance of interfaces between vessel tissue layers. In order to determine the interface location a manual tracing is commonly used. However, this approach is a time consuming procedure and based on subjective operator assessment. This procedure, inevitably results in inter and intra-observer variability. This is due to the complex nature of the echogenic zones, specially at the lumen-intima interface, which frequently present weak echoes, echo dropouts and irregularities caused by scattering.

We have proposed a method that uses the active contour technique where the external forces are proportional to the local image gradient obtained from a multiscale analysis. The automated measurements, when compared to those obtained by manual tracing, are equally accurate and the coefficients of variability between both methods are below 1,0% for Lumen Diameter and 6,5% for Intima-Media thickness measurements. However, for clinical purposes, some interactive tools for manual tracing may still be need in order to correct remaining detection errors due to poor image quality.

Acknowledgements

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