

# A fully-automatic method to segment the carotid artery layers in ultrasound imaging – Application to quantify the compression-decompression pattern of the intima-media complex during the cardiac cycle

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## Abstract

The aim of this study was to introduce and evaluate a contour segmentation method to extract the interfaces of the intima-media complex in carotid B-mode ultrasound

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images. The method was applied to assess the temporal variation of the intima-media thickness during the cardiac cycle.

The main methodological contribution of the proposed approach is the introduction of an augmented dimension to process two-dimensional images in a three-dimensional space. The third dimension, which is added to the two spatial dimensions of the image, corresponds to the tentative local thickness of the intima-media complex. The method is based on a dynamic programming scheme that runs in a three-dimensional space generated with a shape-adapted filter bank. The optimal solution corresponds to a single medial axis representation that fully describes the two anatomical interfaces of the arterial wall. The method is fully-automatic and does not require any input from the user.

The method was trained on 60 subjects and validated on 184 other subjects from six different cohorts and four different medical centers. The arterial wall was successfully segmented in all analyzed images (average pixel size of  $57 \pm 20 \mu\text{m}$ ), with average segmentation errors of  $47 \pm 70 \mu\text{m}$  for the lumen-intima interface,  $55 \pm 68 \mu\text{m}$  for the media-adventitia interface, and  $66 \pm 90 \mu\text{m}$  for the intima-media thickness. The amplitude of the temporal variations in IMT during the cardiac cycle was significantly higher in the diseased population compared to healthy volunteers ( $106 \pm 48$  vs.  $86 \pm 34 \mu\text{m}$ ,  $p = 0.001$ ).

The introduced framework is a promising approach to investigate an emerging functional parameter of the arterial wall, by assessing the cyclic compression-decompression pattern of the tissues.

*Keywords:* Contour segmentation, Carotid artery, Ultrasound, Dynamic programming, Front propagation, Intima-media thickness

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## 1 Introduction

2 Cardiovascular diseases are the leading cause of mortality and morbidity world-  
3 wide (WHO, 2015). An early sign of the atherosclerotic process is arterial wall  
4 thickening. Carotid intima-media thickness (IMT, Fig. 1) has been demonstrated  
5 to be strongly correlated with cardiovascular diseases (O’Leary et al., 1999; Baldas-  
6 sarre et al., 2000; Polak et al., 2011), and is considered an established biomarker  
7 of atherosclerosis (Lorenz et al., 2007): IMT values  $< 1.1$  mm indicate a healthy  
8 artery (Jarauta et al., 2010), values  $> 1.5$  mm are a surrogate marker for an atheroscle-  
9 rosis plaque (Touboul et al., 2012), and transitional values define an intermediate  
10 risk. Therefore, assessing IMT is crucial to evaluate the cardiovascular risk. The fo-  
11 cus of the present study is the assessment of IMT at an early stage of atherosclerosis,  
12 in images without atheromatous plaque.

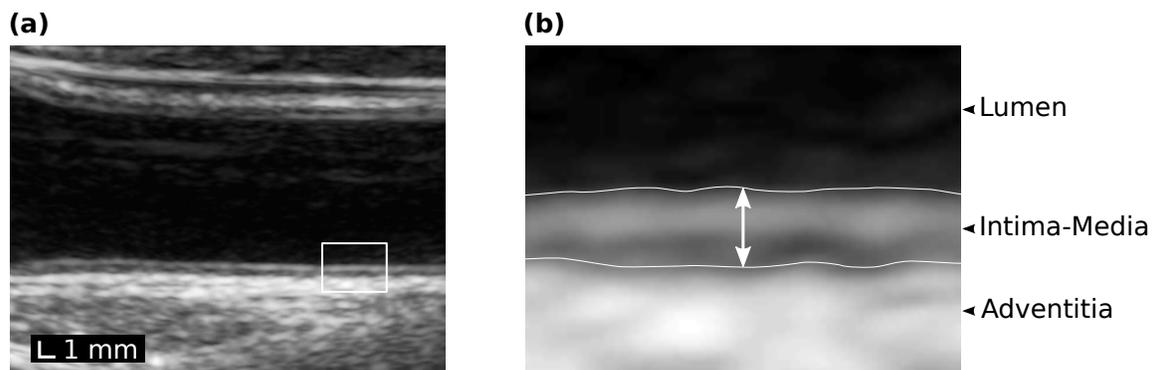


Figure 1: (a) Longitudinal view of the common carotid artery in B-mode ultrasound imaging. (b) Enlarged region detailing the intima-media complex. The lumen-intima and media-adventitia interfaces are represented by the contour lines, and the intima-media thickness is indicated by the arrow.

13 B-mode ultrasound (US) imaging has been extensively used to assess carotid IMT  
 14 *in vivo*. The problem of quantifying the IMT is generally addressed by extracting the  
 15 two contours of the intima-media complex – *i.e.*, the lumen-intima (LI) and media-  
 16 adventitia (MA) interfaces, Figure 1 –, and calculating the distance between these  
 17 two contours. According to the Mannheim consensus (Touboul et al., 2012), the far  
 18 wall of the artery is generally analyzed, mainly due to better depiction of the intima-  
 19 media double-line pattern, and to possible overestimation of the IMT in the near  
 20 wall (Skilton et al., 2011). A large number of computerized segmentation methods  
 21 have been proposed to (semi-) automatically extract the LI and MA interfaces in two-  
 22 dimensional (2D) US images. Recent comprehensive review articles (Molinari et al.,  
 23 2010; Ilea et al., 2013; Loizou, 2014) propose categorizations of these methods ac-  
 24 cording to image-processing techniques used, and provide their synthetic description  
 25 together with main results, etc. Hereafter, we cite only examples of major studies.

26 Let us note that the interfaces of interest may be extracted either sequentially or  
 27 simultaneously. Most of the existing methods fall in the broad category of sequential  
 28 contour extraction, *i.e.*, the two contours are extracted more or less independently  
 29 (sometimes one extracted contour being used as initialization to extract the second  
 30 one), and then possibly refined to improve their spatial consistency. Among these  
 31 approaches, methods based on feature extraction and region classification have been  
 32 recommended to extract the pixels that belong to the intima-media complex (Moli-  
 33 nari et al., 2011; Menchón-Lara et al., 2014). An edge operator based on statistical

34 filtering has been introduced to extract the anatomical boundaries by analyzing the  
35 local statistics of the image (Faita et al., 2008). A method based on adaptive  
36 thresholding followed by a morphological operation was proposed to detect the con-  
37 tours of the layers (Ilea et al., 2013). Various implementations of the active-contour  
38 framework have been put forward by different teams (Mao et al., 2000; Cheng et al.,  
39 2002; Loizou et al., 2007; Xu et al., 2012; Bastida-Jumilla et al., 2015). Dynamic  
40 programming methodology has also been adopted in combination with various tech-  
41 niques, such as fuzzy expression forms representing image features and geometrical  
42 characteristics of vessel-wall interfaces (Liang et al., 2000), smooth intensity thresh-  
43 olding surfaces and geometric snakes (Rocha et al., 2010), and directional Haar-like  
44 filter (Lee et al., 2010).

45 Cheng and Jiang (2008) proposed a dual-dynamic-programming method control-  
46 ling the simultaneous evolution of two curves and devised to jointly extract both  
47 contours of the intima-media complex. They argued that their approach was able to  
48 robustly cope with noise and irregular contrasts. In particular, the article demon-  
49 strated that dual dynamic programming was more efficient in handling the spatial  
50 consistency of the contours than traditional dynamic programming, where the con-  
51 tours, extracted sequentially, may intersect or collapse onto each other. Following  
52 a similar strategy, we have developed a method (Zahnd et al., 2012, 2013b) based  
53 on a representation known as *medial axis*, *i.e.*, a single object fully describing the  
54 two contours of the intima-media complex by means of *i)* the radial *position* of a  
55 curve located mid-way between LI and MA interfaces, and *ii)* the local *distance* be-  
56 tween each contour and this curve. That information was separately estimated in  
57 each column of the image by means of a shape-adapted filter bank. The algorithm  
58 processed a whole image sequence (hereafter referred to as *cine-loop*) after an inter-  
59 active initialization performed in the first frame. The method efficiently controlled  
60 the location and smooth shape of the central curve, but could not guarantee the  
61 smoothness of the contours.

62 Overall, accuracy of the methods reported by the above-cited studies, in IMT  
63 quantification, as well as in both LI and MA interfaces extraction, ranged between 10  
64 and 90  $\mu\text{m}$ . Nevertheless, these results were obtained using different images, distinct  
65 reference annotations, and possibly not the same evaluation protocols. According to  
66 recent surveys (Molinari et al., 2010; Ilea et al., 2013; Loizou, 2014), for the majority  
67 of the studies available in the literature, evaluation was carried out using a relatively  
68 limited number (typically, below 50) of images that were generally acquired using  
69 only one scanner. Additionally, in most of these studies, training data were not  
70 specified; if the same images were used to tune a method and determine its optimal  
71 parameter settings, and then to evaluate it, the robustness of this methods with

72 respect to new data and scanners may be questionable. Moreover, although some  
73 approaches are fully automatic (Molinari et al., 2011; Ilea et al., 2013; Menchón-  
74 Lara et al., 2014; Bastida-Jumilla et al., 2015), the majority of the existing methods  
75 require manual input from the user. Such user interaction, which can range from  
76 several mouse-clicks to interactive parameter tuning, may be necessary to overcome  
77 various difficulties, *e.g.*, related to acquisition-specific variability, but it is known  
78 to introduce inter- and intra-observer variability, when dealing with the same case,  
79 while automatic methods are fully reproducible, when applied to the same data.  
80 Additionally, to limit the user’s workload, methods devised for cine-loops containing  
81 hundred frames or more must be maximally automated.

82 The aim of the present study was to develop a fully automatic method capa-  
83 ble of simultaneously extracting both contours of the intima-media complex, in US  
84 cine loops, with sufficient accuracy to quantify minute variations of IMT during the  
85 cardiac cycle. To this purpose, we introduced a dynamic-programming algorithm  
86 that conducts a minimum-cost-path search in a three-dimensional (3D) space built  
87 of the two spatial dimensions (width and height) of the image, and of all potential  
88 IMT values. The minimum-cost-path defines a smooth 3D medial axis that describes  
89 the spatial location and shape of the intima-media complex, as well as its thickness  
90 in each column of the image. A fully-automatic framework is presented to run the  
91 method without any user interaction. Nevertheless, the user may choose to manually  
92 perform part or all of the otherwise automatic initialization phase.

93 The introduced contour segmentation method was first trained on a subset of  
94 *in vivo* data corresponding to 60 subjects from four different cohorts provided by  
95 three medical centers, then it was validated on a set of *in vivo* data with variable  
96 image characteristics and quality, corresponding to 184 subjects from six distinct  
97 cohorts, imaged with different scanners in four medical centers. Furthermore, as  
98 the amplitude  $\Delta_{\text{IMT}}$  of the temporal variations in IMT during the cardiac cycle  
99 has been demonstrated to be associated with cardiovascular risk factors (Boutouyrie  
100 et al., 2001; Meinders et al., 2003; Polak et al., 2012b,a; Zahnd et al., 2013b), the  
101 method was finally applied to quantify this emerging and potentially relevant clinical  
102 parameter.

## 103 Methods

104 Similarly to our previous work (Zahnd et al., 2012, 2013b), the contour-extraction  
105 method is based on the well-known concept of medial axis. In 2D images, this concept  
106 corresponds to a representation of (mainly elongated) objects by a line located in  
107 the object half-way between its contours; each point of the line carries information  
108 about the distance to the contours. In our implementation, the gap between the  
109 contours, designated by the variable  $\Delta$ , is calculated column-wise; in the sequel, this  
110 gap will be referred to as distance or IMT, although it is not rigorously equal to  
111 the actual distance calculated in the direction perpendicular to the contour, when  
112 the wall is curved or tilted in the image. Figure 2 represents the work-flow of the  
113 method applied within a region of interest (ROI)  $\mathcal{I}_W$  of a single frame, where  $W$   
114 stands for the width of the ROI. The steps of the work-flow will be detailed in the  
115 subsequent sections; here we give its outline. The first step (a-b) uses the shape  
116 and location of the medial axis from the previous frame (or from the initialization,  
117 if the very first frame of the cine loop is considered), in order to “flatten” the region  
118 of interest centered on the intima-media complex, so that the contours of the latter  
119 become nearly horizontal. In the second step (c), a bank of filters sensitive to the  
120 typical two-line pattern of the intima-media complex is applied in each point of  
121 the image. The response of a filter is strong if the point, in which it is applied,  
122 is located half-way between two edges. Since the IMT can vary spatially and/or  
123 temporally (Selzer et al., 2001) each filter in the bank is tuned to a different spacing  
124  $\Delta$  between edges, and the range of these tentative spacings is set as a function of the  
125 IMT value estimated in the previous frame (or provided by the initialization). The  
126 responses of the filters form a 3D map, where the first two dimensions correspond to  
127 the  $x$  and  $y$  spatial dimensions in the image, while the third dimension corresponds  
128 to the spacing  $\Delta$ . In the third step (d), filter responses are transformed into costs:  
129 stronger responses give rise to lower costs. In the fourth step (e), front propagation is  
130 carried out starting from  $x = 0$ : the front propagates faster where the cost is lower,  
131 while its non-horizontal displacements and changes in  $\Delta$  are penalized. The result  
132 of this step is a map of arrival times. In the subsequent step (f), the minimum-cost  
133 path is found by a gradient descent in this map. After the last step (g), which applies  
134 the inverse transformation with respect to the “flattening”, this path defines a new  
135 medial axis, and the next frame can be processed. Let us note that the first step  
136 (a-b) was identical in our previous method (Zahnd et al., 2012, 2013b), as well as  
137 the filter bank. These will be summarized in the subsection “Background”, whereas  
138 the actual contributions of the current work will be described in the subsections  
139 “Contour extraction” and “Automatic initialization”.

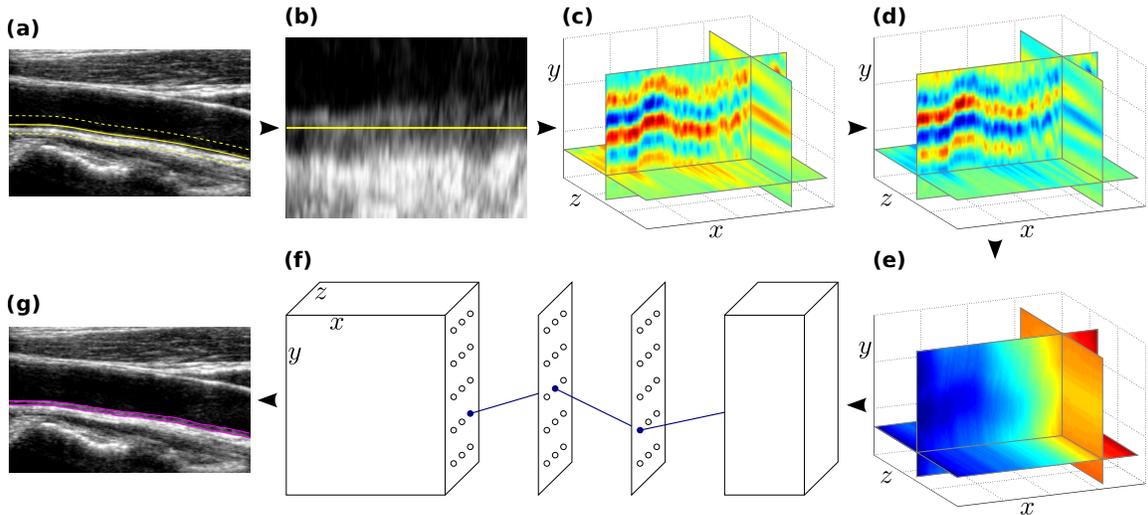


Figure 2: Illustration of the main steps of the segmentation method, applied here to the full width of an image (*i.e.*,  $\mathcal{I}_W = \mathcal{I}$  in this example). (a) Initial image  $\mathcal{I}$ , with the medial axis from the previous image (solid yellow line). (b) Sub-image  $\mathcal{I}_T$ , resulting from the  $T$  transformation of the region located between the dashed yellow lines in (a), such that the anatomical interfaces around the previous medial axis (solid yellow line) are nearly horizontal. (c) 3D volume, corresponding to the set of  $K$  velocity maps  $F_{\Delta^k}$  generated with the filter bank  $\mathcal{H}_{\Delta}$ . (d) 3D cost function  $\mathcal{C}$ . (e) 3D cumulative cost function  $\mathbb{C}$  corresponding to a left-to-right direction of propagation. (f) Schematic representation of the optimal path (blue line), extracted by backtracking the decreasing values in  $\mathbb{C}$ , corresponding to the medial axis: for each column ( $x$ ), the radial position ( $y$ ) of the medial axis as well as the local IMT ( $z$ ) are fully described by the coordinates of the corresponding node. (g) Resulting contour extraction (violet lines) after the inverse  $T^{-1}$  transformation.

140 *Background*

141 *Spatial transformation  $T$ .* The aim of the transformation  $T$  is to generate a sub-  
 142 image  $\mathcal{I}_T$ , centered around the far wall, in which the anatomical interfaces of possibly  
 143 tilted or curved wall become nearly horizontal (Fig. 2a,b). Briefly, the transforma-  
 144 tion  $T$  consists in mapping the medial axis from the previous frame onto the current  
 145 frame, and then selecting, for each column in  $\mathcal{I}_W$ , a region of height  $H$  centered  
 146 around the medial axis, which is assumed to be smooth. This column-by-column  
 147 re-sampling is associated with longitudinal smoothing by a Gaussian filter  $G_{\sigma_x}$  of  
 148 standard deviation  $\sigma_x$ . The values of  $H$ ,  $\sigma_x$ , and of all other parameters will be

149 specified in the experimental part.

150 *Filter bank.* The shape-adapted filter  $\mathcal{H}_\Delta$  introduced in our previous study (Zahnd  
151 et al., 2013b) is defined as

$$\mathcal{H}_\Delta = \pm G'_{\sigma_y} * \mathcal{M}_\Delta, \quad (1)$$

152 where  $(*)$  is the convolution operator,  $G'_{\sigma_y}$  is the first derivative of a Gaussian with  
153 standard deviation  $\sigma_y$ , and  $\mathcal{M}_\Delta$  is a pair of Diracs separated by a distance  $\Delta$ . The  
154 term  $G'_{\sigma_y}$  alone, convolved with an image, is usually used to calculate the intensity  
155 gradient, and thus highlight the edges. Combined with the term  $\mathcal{M}_\Delta$ , it highlights  
156 the expected position of the medial axis, located mid-way between two edges (in our  
157 case, LI and MA interfaces) separated by the gap  $\Delta$  and having the same sign of the  
158 gradient. The  $\pm$  sign corresponds to the expected gradient orientation: it is negative  
159 for the far wall used in this study, and would be positive if the near wall was assessed.  
160 Column-wise convolution of the  $\mathcal{H}_\Delta$  filter with  $\mathcal{I}_T$  generates a response  $F_\Delta$ :

$$F_\Delta = \mathcal{H}_\Delta * \mathcal{I}_T. \quad (2)$$

161 Let us remind that the scope of the present study is early stage atherosclerosis;  
162 therefore no atheromatous plaques are present in the image, and the IMT is expected  
163 to be almost constant across the image. Nevertheless, the thickness of the layers  
164 remains subject to a certain amount of variability that can occur both spatially  
165 (*i.e.*, within a single frame) and temporally (*i.e.*, between consecutive frames) (Selzer  
166 et al., 2001). To cope with such minute variation of the IMT, a range of different  
167  $\Delta$  values is assessed. Since the spatial resolution is relatively coarse with respect to  
168 the expected IMT variation, the image  $\mathcal{I}_T$  is first interpolated by a factor 10 in the  
169 radial direction to increase the precision. Then, a set of  $K$  filters  $\mathcal{H}_{\Delta^k}$  is successively  
170 applied to  $\mathcal{I}_T$  to build  $K$  functions  $F_{\Delta^k}$ . The  $K$  tentative values of the local IMT  
171 are defined as:

$$\Delta^k = \Delta(n-1) + k\delta, k \in \mathcal{K} \equiv \left\{ -\frac{K-1}{2}, \dots, 0, \dots, \frac{K-1}{2} \right\}, \quad (3)$$

172 where  $\delta$  is an increment corresponding to the minimal IMT variation investigated,  
173 and  $\Delta(n-1)$  is the IMT value estimated in the previous image  $\mathcal{I}(n-1)$ . When  
174  $n = 1$ , the approximate IMT value found in the initialization step is used. In the  
175 current implementation, the range of these tentative values is restricted by threshold  
176 values  $\Delta_{\text{MIN}}$  and  $\Delta_{\text{MAX}}$  to discard non-realistic IMT solutions, and thus increase  
177 the robustness of the method and avoid a potential divergence during successive  
178 time-steps.

179 In the context of front propagation,  $F_\Delta$  can be interpreted as a velocity function:  
180 the front tends to propagate towards the points with larger values of  $F$ . Additionally,  
181 the propagation is constrained to ensure the smoothness of the resulting minimum-  
182 cost path. Before running the front propagation and the subsequent minimum-cost-  
183 path search, the velocity function  $F$  is transformed into a positive cost function  $\mathcal{C}$ ,  
184 where small values correspond to strong filter responses:

$$\mathcal{C} = \exp(-F). \quad (4)$$

185 In our previous work, the function  $F$  was two-dimensional: it was obtained by se-  
186 lecting, for each point of spatial coordinates  $(x, y)$ , the largest response among  $K$   
187 filters applied in the same point:  $F(x, y) = \max_{k \in \mathcal{K}}(F_{\Delta^k}(x, y))$ . This blind selection  
188 of the maximal output, without fully exploiting the smoothness of the anatomical  
189 structures of interest, was the limitation of that method. Indeed, artefactual patterns  
190 similar to the actual intima-media complex may produce the strongest filter response  
191 for inconsistent spatial locations and  $\Delta$  values; whereas constraining the front propa-  
192 gation by means of a penalty proportional to vertical displacements guaranteed that  
193 the radial positions of the medial axis were smooth, there was no smoothness con-  
194 straint on the selected  $\Delta$  values. It was therefore possible to assign highly different  $\Delta$   
195 values to adjacent columns in the image, thus generating irregular contours despite  
196 a smooth medial axis (Fig 3 left).

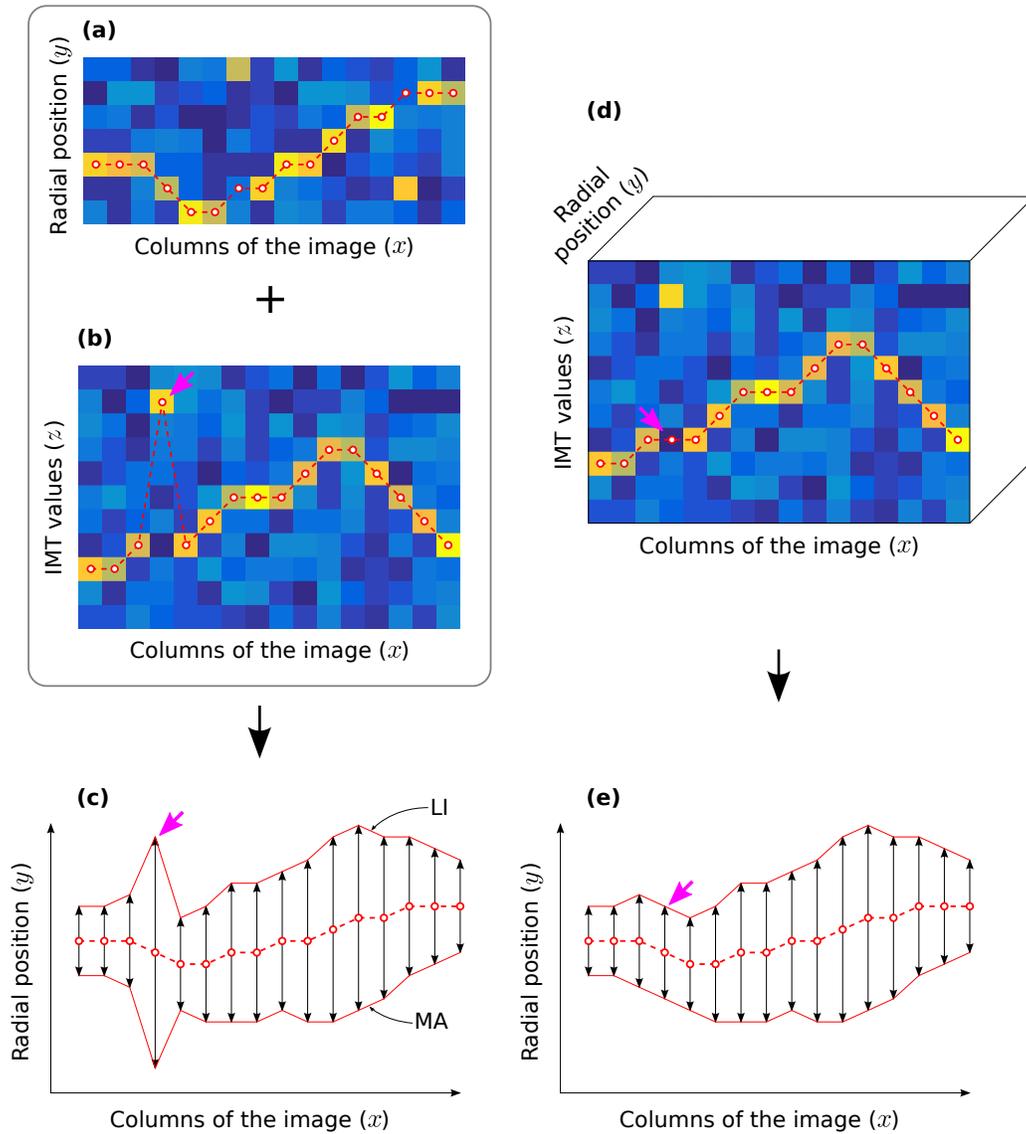


Figure 3: Schematic representation of the previously introduced segmentation method (left), compared with the approach proposed in the present study (right). The medial axis is represented by the red dashes, the two lumen-intima (LI) and media-adventitia (MA) contours are indicated by the red lines, and the  $\Delta$  value (IMT) of each column in the image is displayed with the black arrows. In the previous method, a pair of 2D maps was used: a minimal path was extracted in a 2D space to describe the optimal radial position of the medial axis (a), and the  $\Delta$  information was simply determined by selecting the maximum of a filter output (b), which may lead to erroneous results (c, pink arrow). In the present method, the minimal path is extracted in a 3D space (d) to describe the optimal medial axis (e). Please note that the spatial dimension  $y$  of the path is not fully represented in (d).

198 *Contour extraction*

199 In the new proposed method, all the  $K$  functions  $F_{\Delta^k}$  are concatenated to build a  
 200 single 3D velocity function  $F$  (Fig. 2c). More precisely, each map  $F_{\Delta^k}$  is first down-  
 201 sampled by a factor 10 in the radial direction, back to the original image resolution,  
 202 in order to reduce the computational time as compared to down-sampling at a later  
 203 stage of the work-flow. Subsequently, a 3D cost function  $\mathcal{C}$ , where the lowest values  
 204 (blue color in Figure 2d) correspond to the points most likely to belong to the medial  
 205 axis, is generated according to Equation 4.

206 *Dynamic programming.* Using the 3D cost function  $\mathcal{C}$ , a front propagation scheme  
 207 is run to build a 3D cumulative cost function  $\mathbb{C}$  (Fig. 2e). Within the 3D space,  
 208 the following  $x, y, z$  notation was adopted:  $x$  and  $y$  correspond to the horizontal and  
 209 vertical coordinates in the 2D US image, respectively, and  $z$  represents the additional  
 210 dimension corresponding to  $\Delta$ , *i.e.*, tentative IMT values.

211 The propagation is unidirectional along  $x$  (*i.e.*, from the left to the right border of  
 212 the sub-image  $\mathcal{I}_T$ ), and favors the valleys in  $\mathcal{C}$  while penalizing abrupt changes along  $y$   
 213 and  $z$ , corresponding to non-horizontal contours and IMT variations, respectively.  
 214 The values of  $\mathbb{C}$  where  $x = 1$  (*i.e.*, the leftmost column of the ROI) are initialized  
 215 with the corresponding values of  $\mathcal{C}$ . Then, the function  $\mathbb{C}$  is iteratively generated for  
 216 increasing values of  $x$ , with  $x$  ranging from 2 (*i.e.*, second leftmost column of the  
 217 ROI) to  $W$  (*i.e.*, rightmost column of the ROI), according to Equation 5.

$$\begin{aligned} \mathbb{C}(x, y, z) = \min_{d_y, d_z} \left\{ \mathbb{C}(x-1, y+d_y, z+d_z) \right. \\ \left. + \left( \mathcal{C}(x, y, z) + \mathcal{C}(x-1, y+d_y, z+d_z) \right) \cdot \left( (1 + \alpha_y \cdot d_y^2) + (1 + \alpha_z \cdot d_z^2) \right) \right\} \end{aligned} \quad (5)$$

218 Here,  $d_y \in \{-N_y, \dots, 0, \dots, N_y\}$  and  $d_z \in \{-N_z, \dots, 0, \dots, N_z\}$  respectively represent  
 219 displacements along the axes  $y$  and  $z$ , so the respective numbers of reachable neigh-  
 220 bors along these directions are  $2N_y + 1$  and  $2N_z + 1$ . The parameters  $\alpha_y$  and  $\alpha_z$   
 221 control the smoothness of the medial axis along  $y$  and  $z$  directions, respectively. Let  
 222 us also note that the propagation function (Eq. 5) is symmetrical along the  $x$  di-  
 223 rection, namely it yields identical results for left-to-right or right-to-left propagation  
 224 schemes.

225 Finally, the point on the right border of  $\mathcal{I}_T$  with the minimal cumulative cost is  
 226 found, and the minimum-cost path,  $P$ , corresponding to the global optimal solution,  
 227 is extracted by back-tracking the decreasing values in  $\mathbb{C}$  from this identified point to  
 228 the left border (Fig. 2f). By construction, for each value of  $x$ , there is a single point  
 229 in  $P$ , so it can be written as  $P \equiv \{(x, y, z) : x \in [1, W], y = P_y(x), z = P_z(x)\}$ .

230 *Inverse transformation  $T^{-1}$ .* For each column  $x$  of the image,  $y$  and  $z$  coordinates  
 231 of the minimum-cost path respectively indicate the radial position,  $P_y(x)$ , of the  
 232 corresponding medial-axis node in the transformed image  $\mathcal{I}_T$  and the local IMT,  
 233  $P_z(x)$ . The final medial axis position is determined by mapping back the positions  
 234  $P_y(x)$  onto the original image  $\mathcal{I}$  by the  $T^{-1}$  operation. Let  $y_p(x)$  stand for the  
 235 position  $P_y(x)$  transformed by the  $T^{-1}$  operation. For each column  $x$  of the image,  
 236 the coordinates of the two contours  $y^-(x)$  (LI interface) and  $y^+(x)$  (MA interface) of  
 237 the intima-media complex are then deduced from the final medial axis (Fig. 2g and  
 238 Fig. 3 right), according to the relation

$$y^\pm(x) = y_p(x) \pm 0.5 * P_z(x). \quad (6)$$

239 Finally, the global IMT of the image is calculated as the median radial distance  
 240 between the LI and MA interfaces over the width  $W$ .

#### 241 *Automatic initialization*

242 Prior to processing a given frame  $\mathcal{I}(n)$ , three parameters are required to run the  
 243 method: *i)* the left and right boundaries of the ROI,  $\mathcal{I}_W$ , to be segmented, *ii)* the  
 244 approximate location and shape of the far wall, and *iii)* the approximate IMT. When  
 245 processing the  $n^{\text{th}}$  frame of a cine-loop, this input is obtained from the segmentation  
 246 results of the previous  $n - 1^{\text{st}}$  frame. For the first frame of the cine-loop, or a static  
 247 image, these parameters must be provided by the initialization phase. In this work,  
 248 the initialization is fully automatic, *i.e.*, no user interaction is required, whereas it  
 249 was manual in our previous work. This automatic process is composed of three main  
 250 steps described in the following paragraphs.

251 *Border clipping.* First, the outer top, bottom, left and right borders of the image  
 252 that contain various text information such as the date and the scanner settings  
 253 are removed to keep only the central region corresponding to the image of the vessel  
 254 (Fig. 4a). For every pixel of the image, a score is computed by calculating the average  
 255 absolute difference of gray level between every pair of consecutive images in the  
 256 cine-loop. Outer borders correspond to low-score regions since pixels remain quasi-  
 257 identical through the entire cine-loop, whereas the central vessel image corresponds  
 258 to high-score regions since pixels are constantly changing between different frames.  
 259 The image is thus clipped by applying a threshold  $\tau$  on the horizontal and vertical  
 260 profiles of the calculated scores.

261 *Localization of the far wall.* Second, an approximate location of the far wall is deter-  
 262 mined by finding the regions with the strongest positive vertical gradient. To avoid

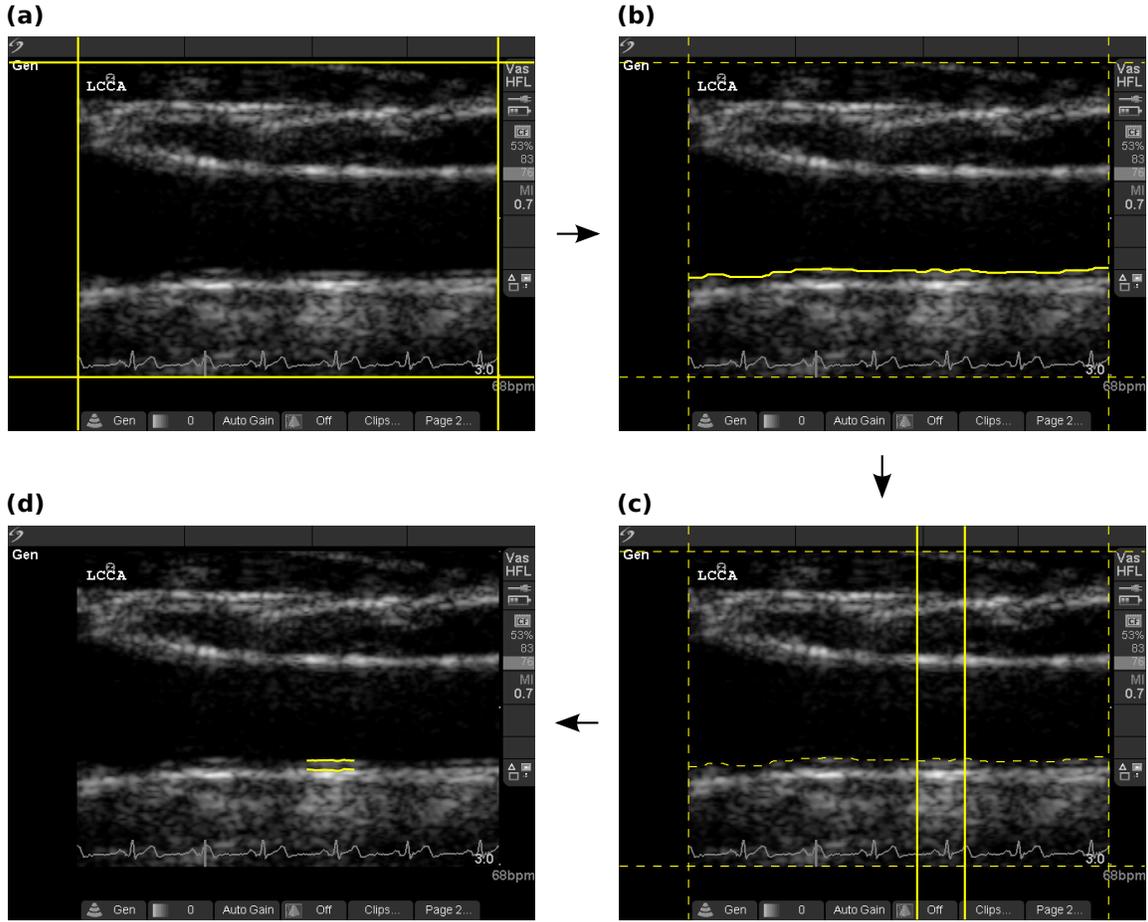


Figure 4: Illustration of the 4 stages of the fully-automatic process. (a) Border clipping. (b) Localization of the far wall. (c) Determination of the optimal region. (d) Contour segmentation.

263 potential errors that could be caused by the presence of the jugular vein, which ex-  
 264 hibits a similar structure and is located (in the image) above the carotid artery, the  
 265 search is only conducted in the bottom of the image, up to  $h\%$  of the total image  
 266 height. A heavily blurred vertical intensity gradient is calculated by column-wise  
 267 convolution with a filter  $G'_B$ , defined as the first derivative of a Gaussian with a large  
 268 standard deviation  $B$ . It is then used to build a cost function where lower costs  
 269 correspond to larger positive gradients. A single contour is subsequently extracted  
 270 (Fig. 4b) by a classical 2D minimum-cost-path search with this cost function, imple-

271 mented in a dynamic-programming scheme. The resulting rough contour of the far  
272 wall constrains the search of the initial medial axis in the last step of the initialization.

273 *Determination of the optimal region and of the initial IMT.* Since some regions of  
274 the far wall are frequently too noisy and not sufficiently contrasted to be segmented  
275 accurately, a ROI,  $\mathcal{I}_W$ , with the best image quality, is localized. From our experi-  
276 ence, its width  $W = 3$  mm provides the optimal results to extract a reproducible  
277 compression-decompression pattern (Zahnd et al., 2013b); a region too narrow of-  
278 ten yields noisy curves of IMT temporal variation, and a region too broad generally  
279 yields blurred curves. To find this optimal region, the contour segmentation method  
280 introduced in this study is tentatively applied to the entire width of the image. Let  
281 us recall however, that – at this stage – only an approximate location and shape of  
282 the far wall are available, but IMT is not. Estimating the initial IMT, localizing  
283 the optimal ROI, as well as refining the initial location of the line supposed to fall  
284 mid-way between the two contours, are jointly performed in this last step of the  
285 initialization process. The previously extracted location of the far wall is used to  
286 apply the “flattening” transformation  $T$ . A set of  $K$  tentative IMT values, linearly  
287 distributed in a large range between 0.4 mm and 1.5 mm, are assessed. Then, a score  
288 is calculated for every column of the image, by extracting the output value of the  
289 velocity function  $F_\Delta$  (Eq. 2) at the location of the medial axis resulting from the  
290 segmentation. Here, regions with good image quality are expected to have a high  
291 score since they present well perceptible lumen-intima and media-adventitia inter-  
292 faces. The optimal region  $\mathcal{I}_W$  is thus selected by finding the 3 mm segment with  
293 the highest total score (Fig. 4c). Finally, the initial IMT is selected like in the main  
294 method, by selecting the median radial distance between the contours resulting from  
295 this tentative segmentation, within  $\mathcal{I}_W$ . The results of this step, *i.e.*, the borders of  
296  $\mathcal{I}_W$ , the approximate IMT, and the spatial location of the medial axis resulting from  
297 the tentative segmentation, are passed as input parameters to the actual contour  
298 segmentation applied within  $\mathcal{I}_W$  (Fig. 4d).

## 299 Experiments

### 300 *Study population*

301 All subjects were retrospectively selected from already acquired data used in  
302 previous studies. Subjects presenting an atheromatous plaque in the imaged region,  
303 as visually assessed by a medical doctor, were excluded from our study. A total of 244  
304 subjects were included in the present study. To assess the accuracy and robustness  
305 of the proposed method on data with various image characteristics and quality, the  
306 present study involved a total of six cohorts from four different research centers, as  
307 presented in Table 1. Representative examples of images from the six different cohorts  
308 are displayed in Figure 5. The image data collection was conducted in compliance  
309 with the requirements of the Institutional Review Boards in the different research  
310 centers. Informed written consent was obtained from all subjects. The characteristics  
311 of the cohorts involved are described hereafter.

312 *Erasmus MC, The Netherlands.* The first 51 subjects included in the PARISK study  
313 (Truijman et al., 2014) at the Erasmus Medical Center, the Netherlands, were consid-  
314 ered. The subjects were individuals with high cardiovascular risk; the main inclusion  
315 criteria were a transient ischemic attack or a minor stroke in the past six weeks be-  
316 fore enrollment, and a stenosis  $< 70\%$  in the carotid bifurcation. Eight subjects were  
317 rejected from the study, because they had an atherosclerotic plaque in the imaged  
318 region. The remaining 43 subjects (denoted NLd) were included in the study.

319 *University of Lyon, France.* Fifty-seven healthy controls (FRc) and 25 patients  
320 (FRd) at high cardiovascular risk were involved. Data acquisition was performed at  
321 Louis Pradel Hospital (Lyon, France), as previously described (Zahnd et al., 2013a).  
322 The inclusion criterion for the at-risk patients was the presence of one of the following  
323 diseases diagnosed at least 1 year before: metabolic syndrome, or type 1 or 2 dia-  
324 betes. No other criterion, including clinical characteristics, was used to select these  
325 subjects. Healthy controls were cardiovascular-risk-factor free (tobacco use, hyper-  
326 cholesterolemia, diabetes, hypertension, or particular family history), as assessed by  
327 an oral questionnaire.

328 *Sydney Medical School, Australia.* Thirty healthy controls (AUc) and 30 patients  
329 (AUd) were selected from the database of the PerioCardio study (Kapellas et al.,  
330 2014). The images were collected at the Menzies School of Health Research in Dar-  
331 win, Northern Territory, Australia. The at-risk patients were Aboriginal Australian  
332 with moderate periodontitis, an emerging cardiovascular risk factor (Bouchard et al.,  
333 2010). The healthy controls were age-matched Caucasian subjects without cardio-  
334 vascular risk factors.

335 *University of Cyprus, Cyprus.* Data were gathered by downloading a publicly-available  
 336 dataset containing 100 images recorded at the Cyprus Institute of Neurology and Ge-  
 337 netics (Nicosia, Cyprus) (Loizou et al., 2007). All subjects were at risk of atheroscle-  
 338 rosis and had already developed clinical symptoms, such as stroke or transient is-  
 339 chemic attack. Forty-one patients were imaged twice, and the corresponding 41  
 340 images were rejected to keep a single image per patient. The remaining 59 images  
 341 (CYd) were included in the study.

Table 1: Characteristics of the six cohorts involved in the present study. (\* The 100 initial images of the dataset from the University of Cyprus correspond to 58 males and 42 females with a mean age of 54 y.o. Nevertheless, 41 of these subjects were imaged twice. Therefore, the actual sex and age information for the 59 subjects of the CYd cohort is unknown.)

Medical center	Cohort name	Diseased/Control	n	Training set	Testing set	Gender (males)	Mean Age (years)
Erasmus MC, The Netherlands	NLd	Diseased	43	13	30	36	$69.1 \pm 7.5$
University of Lyon, France	FRc	Control	57	19	38	24	$37.9 \pm 14.1$
University of Lyon, France	FRd	Diseased	25	8	17	16	$56.2 \pm 10.5$
Sydney Medical School, Australia	AUc	Control	30	0	30	17	$41.0 \pm 16.4$
Sydney Medical School, Australia	AUd	Diseased	30	0	30	22	$40.3 \pm 8.9$
University of Cyprus, Cyprus	CYd	Diseased	59	20	39	*	*

#### 342 *Acquisition of carotid artery ultrasound images*

343 Longitudinal B-mode US images of the CCA were acquired for all subjects. The  
 344 subjects were examined in the supine position. Image acquisition was performed  
 345 with a clinical scanner equipped with a linear array transducer. The characteristics  
 346 of the specific scanner used for each cohort are described in Table 2. For all but one  
 347 (CYd) cohort, images were recorded through at least two consecutive full cardiac  
 348 cycles, whereas for the CYd cohort only a single static image was available for each  
 349 patient. The average ( $\pm$  std) number of frames for each cine loop was  $206 \pm 145$  (range  
 350 [66 – 491]). All image data were digitally stored and transferred to a commercial  
 351 computer for off-line image analysis.

#### 352 *Generation of training and testing sets*

353 Each cohort was divided into a training set and a testing set. The training  
 354 datasets were used during the development phase of the method, to empirically de-  
 355 termine the optimal parameter settings. The testing datasets were used only during  
 356 the evaluation phase of the method, to determine the performance of the developed  
 357 framework on independent test data. The distribution of the training and testing  
 358 sets for each cohort is detailed in Table 1.

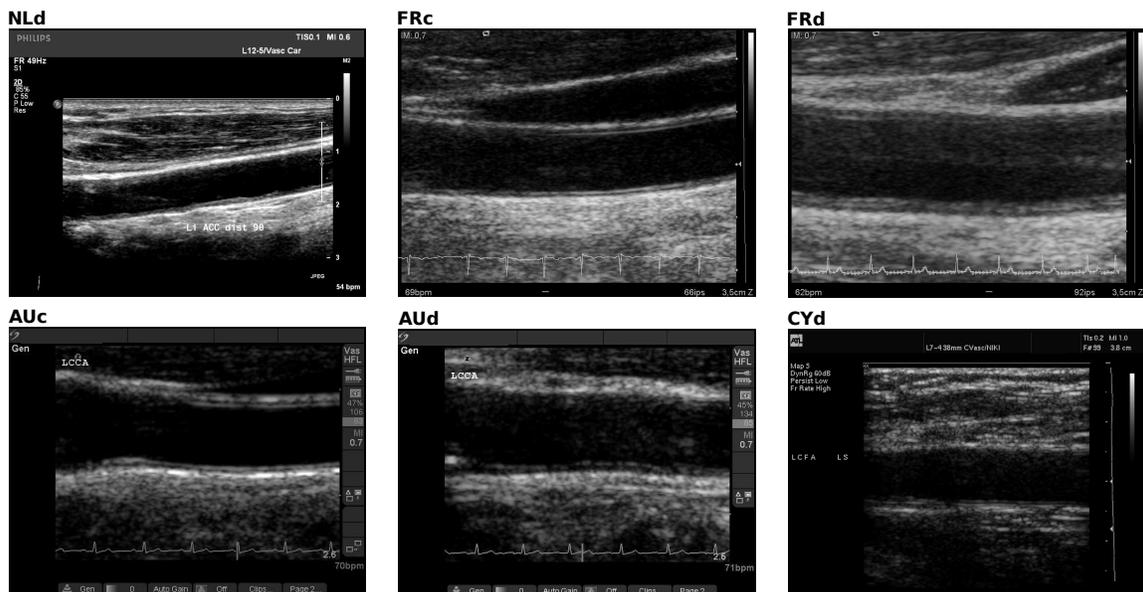


Figure 5: Representative images from the six different cohorts.

Table 2: Characteristics of the different ultrasound scanners used for the six cohorts.

Cohort name	Scanner	Transducer (MHz)	Pixel size ( $\mu\text{m}$ )	Frame-rate (fps)
Nld	iU22, Philips Medical Systems Bothell, USA	5–17 or 5–12	$94 \pm 9$	40
FRc, FRd	Antares, Siemens Erlangen, Germany	7.5–10	$34 \pm 5$	26
AUc, AUd	Sonosite MicroMaxx WA, USA	5–10	$52 \pm 4$	24
CYd	HDI-3000, Advanced Technology Laboratories, Seattle, USA	4–7	$67 \pm 0$	–

359 For all but two (AUc and AUd) cohorts, the training set was generated by ran-  
 360 domly selecting a sub-sample of subjects (*i.e.*, roughly one third of the total number  
 361 of subjects), and the testing set was generated with all the remaining subjects. For  
 362 both AUc and AUd cohorts, the training datasets are empty. The reason for this  
 363 decision was to evaluate the method on images with new unknown characteristics, by  
 364 directly applying the proposed framework on the full AUc and AUd datasets without  
 365 any prior training on images from these two specific cohorts.

### 366 *Independent standard*

367 To quantify the accuracy of the method, despite the lack of ground truth inherent  
 368 to *in vivo* imaging, reference contours were traced by an experienced analyst  $\mathcal{A}_1$ , on

369 the entire exploitable width of one image per subject. For the NLd, FRc, FRd,  
 370 AUc and AUd cohorts, the first frame of the cine-loop was selected. First, as the  
 371 left and/or right borders of the image sometimes show blur, low echoes, or high  
 372 noise, the analyst  $\mathcal{A}_1$  was asked to clip out the poor-quality borders, if necessary,  
 373 thus defining the ROI to be considered. Subsequently, the analyst  $\mathcal{A}_1$  performed  
 374 the manual segmentation of both LI and MA interfaces, using a graphical interface  
 375 that was developed in house specifically for this purpose. Continuous contours were  
 376 generated by fitting a curve through a collection of carefully placed control points.  
 377 The curve interpolation was done by means of the MATLAB *pchip* function. Two  
 378 other analysts,  $\mathcal{A}_2$  and  $\mathcal{A}_3$ , who were blinded to the other results, manually traced the  
 379 anatomical interfaces LI and MA, within the same ROIs of the same set of images.  
 380 These tracings were used to assess inter-operator variability. Similarly, to assess  
 381 intra-operator variability,  $\mathcal{A}_1$  performed one week later a second manual delineation  
 382 of the contours<sup>1</sup>.

### 383 *Parameter settings and implementation*

384 The optimal parameter settings were empirically determined using the training  
 385 set ( $n = 60$ ), then the method was applied once to all the images of the testing set  
 386 ( $n = 184$ ) using the following configuration: height of the region centered around  
 387 the medial axis used in the transformation  $T$ ,  $H = 3$  mm; minimal and maximal  
 388 IMT threshold values,  $\Delta_{\text{MIN}} = 0.4$  mm and  $\Delta_{\text{MAX}} = 1.5$  mm, respectively; standard  
 389 deviation of the Gaussian functions  $G_{\sigma_x}$  and  $G_{\sigma_y}$ ,  $\sigma_x = 150$   $\mu\text{m}$  and  $\sigma_y = 100$   $\mu\text{m}$ ,  
 390 respectively; smoothness parameters,  $\alpha_y = 2$  and  $\alpha_z = 0.1$ ; number of reachable  
 391 neighbors along the  $y$  and  $z$  dimensions,  $2N_y + 1 = 5$  and  $2N_z + 1 = 5$ . The number  $K$   
 392 of possible values  $\Delta^k$  of the IMT depends of the spatial resolution of the image: the  
 393 range between  $\Delta^{\text{min}}$  and  $\Delta^{\text{max}}$  was equal to 1 mm, with an increment  $\delta$  between  
 394 the  $K$  slices equal to 0.1 pixel (for example,  $K = 200$  for an image with a spatial  
 395 resolution of 50  $\mu\text{m}$ ). Setting  $\delta$  in pixels<sup>2</sup> was motivated by the consistency between  
 396 the three dimensions of the search space: with a finer image resolution one can expect  
 397 a finer estimate of the IMT. The standard deviation  $B$  of the Gaussian function  $G_B$   
 398 was equal to 1 mm. In the automatic initialization process, the parameter  $\tau$  used  
 399 in border clipping, was equal to 3% of the highest score calculated in the given cine

---

<sup>1</sup>For the CYp cohort, reference tracings were collected together with the images of the publicly-available dataset (Loizou et al., 2007). For this specific cohort, the reference annotations were already performed by two different analysts. Tracings from a third analyst and intra-analyst variability were not available.

<sup>2</sup>Setting  $\delta$  in millimeters would yield a constant value of  $K$  and a constant ‘‘slice thickness’’ of the 3<sup>rd</sup> dimension in the search space.

400 loop, whereas the height,  $h$ , limiting the search of the far wall in the bottom of the  
401 image was set to 60% of the clipped-image height.

402 The algorithm was implemented in a graphical user interface (GUI), both devel-  
403 oped in MATLAB (MATLAB 7.14, The MathWorks Inc., Natick, MA, USA, 2011).  
404 A compiled executable version, along with a user guide, were made publicly avail-  
405 able: <https://www.creatis.insa-lyon.fr/carolab/>. The GUI also includes an  
406 optional interactive initialization enabling the user to manually define the inputs of  
407 the algorithm, like in our previous work: *(i)* select a sub-image  $\mathcal{I}_W$  by clipping out,  
408 if necessary, the left and/or right borders of the image beyond which the wall cannot  
409 be clearly perceived due to poor image quality, *(ii)* indicate the approximate location  
410 and shape of the far wall by positioning three points within the intima-media com-  
411 plex, and *(iii)* provide a rough initial estimate of the IMT by positioning two radially  
412 aligned points on the LI and MA interfaces. Although the focus of this work is the  
413 fully automatic method, the manual initialization was also used in the evaluation, to  
414 assess different steps of the whole process, as described in the next subsections.

#### 415 *Evaluation of the segmentation accuracy*

416 To assess the accuracy of the proposed method, two distinct experiments were  
417 carried out: with the fully-automatic procedure, and with the manual initialization  
418 procedure described in the previous subsection. Both were performed for all images  
419 annotated by  $\mathcal{A}_1$ ,  $\mathcal{A}_2$ , and  $\mathcal{A}_3$  in the testing set. The resulting contours and IMT  
420 were compared to those corresponding to the manual annotations by  $\mathcal{A}_1$ . For both LI  
421 and MA interfaces, the unitary segmentation error  $\epsilon(x)$  was defined as the absolute  
422 distance between the reference contour manually traced by  $\mathcal{A}_1$  and the estimated  
423 contour, in the column  $x$  of the image. For each image, the IMT estimation error  
424 was defined as the difference between the IMT estimated by the method and the  
425 reference IMT value derived from  $\mathcal{A}_1$ 's annotations. The latter was calculated as the  
426 median column-wise distance between the LI and MA reference contours traced by  
427  $\mathcal{A}_1$ , over the width of the ROI used in the given experiment, *i.e.*,  $W = 3$  mm in the  
428 first experiment, and whole exploitable width defined by  $\mathcal{A}_1$ , in the second one.

429

430 *Fully automatic process.* The aim of the first experiment was to assess the robustness  
431 of each step in the fully-automatic initialization process (Fig. 4), as well as the accu-  
432 racy of the segmentation method initialized by this process. The analyst  $\mathcal{A}_1$  labeled  
433 the result of each step either as a success or as a failure. During the evaluation,  
434 failures were reported and corrected in a cascade, *i.e.*, failures at the  $k^{\text{th}}$  step were  
435 manually corrected prior to conducting the  $k + 1^{\text{st}}$  step, as described below.

- 436 • For the first step, success was defined as clipping out the entire top, bottom, left  
437 and right borders of the frame, while conserving the full image of the artery  
438 (Fig. 4a). Failures were corrected by the user manually indicating the real  
439 borders (1 mouse click per erroneous border).
- 440 • For the second step, success was defined as getting a contour that remained  
441 close to the far wall over the full width of the image (Fig. 4b). Failures were  
442 corrected by the user manually tracing the approximate location and curvature  
443 of the far wall (3 mouse clicks).
- 444 • For the third step, success was defined as a ROI of width  $W = 3$  mm located in  
445 a region with good image quality, where both LI and MA contours were clearly  
446 visible (Fig. 4c). Failures were corrected by the user manually indicating the  
447 center of the new ROI on the region with optimal image quality (1 mouse click).
- 448 • For the fourth step, success was defined as a correct segmentation of both LI  
449 and MA interfaces (Fig. 4d). Failures were corrected by the user manually in-  
450 dicated the IMT (2 mouse clicks) before re-running the segmentation method.

451 Let us note that the images of the cohort CYP are static frames, therefore the  
452 first step could not be applied, because a cine-loop is required. Hence, this step was  
453 not evaluated for that cohort, and the borders were manually indicated by  $\mathcal{A}_1$ .

454 *Segmentation with interactive initialization.* The aim of the second experiment was  
455 to assess the ability of the method to accurately extract the LI and MA interfaces in  
456 the whole exploitable width,  $W$ , determined by the user, and to compare the results  
457 with our previous work (Zahnd et al., 2013b). The initialization was performed by  
458 the analyst  $\mathcal{A}_1$  who defined the ROI  $\mathcal{I}_W$ , indicated the approximate location and  
459 shape of the wall, and roughly determined the IMT. The same initialization was  
460 used to run the proposed segmentation method and the previous one. To investigate  
461 the impact of the manual initialization phase, the reproducibility of the segmentation  
462 algorithm was also assessed. The method was thus applied twice on each image, with  
463 the analyst  $\mathcal{A}_1$  performing the initialization phase a second time – one week later –  
464 within the same clipped  $\mathcal{I}_W$  region.

#### 465 *Evaluation of the temporal variation of the IMT during the cardiac cycle*

466 The intima-media complex has been demonstrated to undergo a reproducible  
467 compression-decompression pattern during the cardiac cycle (Selzer et al., 2001),  
468 namely the IMT is maximal at the end of diastole and minimal during systole. The  
469 amplitude of the IMT compression, denoted  $\Delta_{\text{IMT}}$  and corresponding to the difference

470 between the diastolic and systolic values of the IMT, has been shown to be positively  
471 correlated with cardiovascular risk factors (Boutouyrie et al., 2001; Meinders et al.,  
472 2003; Polak et al., 2012b,a; Zahnd et al., 2013b).

473 To investigate this phenomenon, the proposed segmentation method was applied  
474 to all subjects from both the training and testing datasets, except those from the  
475 CYP cohort, as only one static image was available instead of a temporal cine-loop.  
476 The fully-automatic initialization scheme was used to determine a region of width  
477  $W = 3$  mm located in the far wall, and the IMT was then assessed in all frames of the  
478 cine-loop with the contour segmentation method. For all the 185 analyzed subjects,  
479 the amplitude  $\Delta_{\text{IMT}}$  of the temporal variation of the IMT was finally measured by  
480 the analyst  $\mathcal{A}_1$  on the resulting time-series by indicating the maximal (*i.e.*, during  
481 the diastole) and minimal (*i.e.*, during the systole) IMT values.

#### 482 *Statistical analysis*

483 The Mann-Whitney U test was used to compare the values of the end-diastole  
484 IMT, as well as the total amplitude of the IMT variation  $\Delta_{\text{IMT}}$  during the cardiac  
485 cycle, between at-risk patients and healthy controls. The paired t-test was used to  
486 compare the errors between the present method and the one previously developed  
487 by our team (Zahnd et al., 2013b). The value  $p < 0.05$  was considered to indicate a  
488 statistically significant difference. The Pearson’s correlation coefficient  $R$  was used  
489 to calculate the correlation between the reference tracings and the results from the  
490 method, as well as between the end-diastolic IMT and the amplitude  $\Delta_{\text{IMT}}$ . All  
491 statistical analyzes were performed using MATLAB.

492 **Results**

493 *Fully-automatic process*

494 Segmentation with fully-automatic initialization was evaluated in the 184 images  
 495 of the testing set. Qualitatively, the robustness of each step was assessed as follows: 2  
 496 failures for the first step (border clipping), no failure for the second step (localization  
 497 of the far wall), 7 failures for the third step (determination of the region with the  
 498 best image quality), and 3 failures for the fourth step (contour extraction). Results  
 499 from the quantitative evaluation of the contour accuracy are presented in Table 3.

Table 3: Segmentation absolute errors (mean  $\pm$  SD) in  $\mu\text{m}$ , between the fully-automatic segmentation method and the manual tracings from  $\mathcal{A}_1$ , in the testing set. Manual correction of the fully-automatic process was performed for 12 images.

	All ( $n = 184$ )	NLd ( $n = 30$ )	FRc ( $n = 38$ )	FRd ( $n = 17$ )	AUc ( $n = 30$ )	AUd ( $n = 30$ )	CYd ( $n = 39$ )
Lumen-Intima	$47 \pm 70$	$49 \pm 37$	$38 \pm 35$	$29 \pm 20$	$34 \pm 30$	$36 \pm 35$	$94 \pm 144$
Media-Adventitia	$55 \pm 68$	$74 \pm 65$	$29 \pm 26$	$34 \pm 29$	$51 \pm 42$	$52 \pm 60$	$110 \pm 116$
IMT	$66 \pm 90$	$88 \pm 68$	$35 \pm 28$	$19 \pm 14$	$39 \pm 31$	$50 \pm 50$	$130 \pm 156$

500 *Segmentation in the full exploitable width, with interactive initialization*

501 The proposed method was successfully applied to all the images, in both training  
 502 and testing sets. Both LI and MA interfaces of the wall were delineated, and the  
 503 corresponding IMT was extracted. Representative examples of resulting segmented  
 504 contours for subjects from the six cohorts involved are displayed in Figure 6.

505 For all the 244 subjects involved, the average width  $W$  of the analyzed sub-image  
 506  $\mathcal{I}_W$  was  $20 \pm 8$  mm (range 3 – 51 mm), which corresponded to an average ratio of  
 507  $60 \pm 22$  % (range 10 – 100 %) of the full image width.

508 For both training and testing sets, the mean absolute errors ( $\pm$  standard de-  
 509 viation) of the segmentation process are displayed in Table 4. These results were  
 510 calculated by aggregating all the corresponding unitary segmentation errors in a  
 511 single vector, prior to generating the mean absolute error and standard deviation.

512 To further analyze the results, it is insightful to compare the error magnitude  
 513 with the size of the analyzed tissues, namely, the IMT. The mean value ( $\pm$  standard  
 514 deviation) of the reference IMT, in the first frame of all the subjects assessed, was  
 515  $688 \pm 164$   $\mu\text{m}$ . The relative errors were calculated as the ratio between each absolute  
 516 segmentation error  $|\epsilon(x)|$  and the corresponding individual reference IMT. The mean  
 517 value ( $\pm$  standard deviation) of the relative error was equal to  $7.5 \pm 10.3$  % for the LI  
 518 interface,  $8.9 \pm 9.2$  % for the MA interface, and  $7.6 \pm 8.2$  % for the estimated IMT.

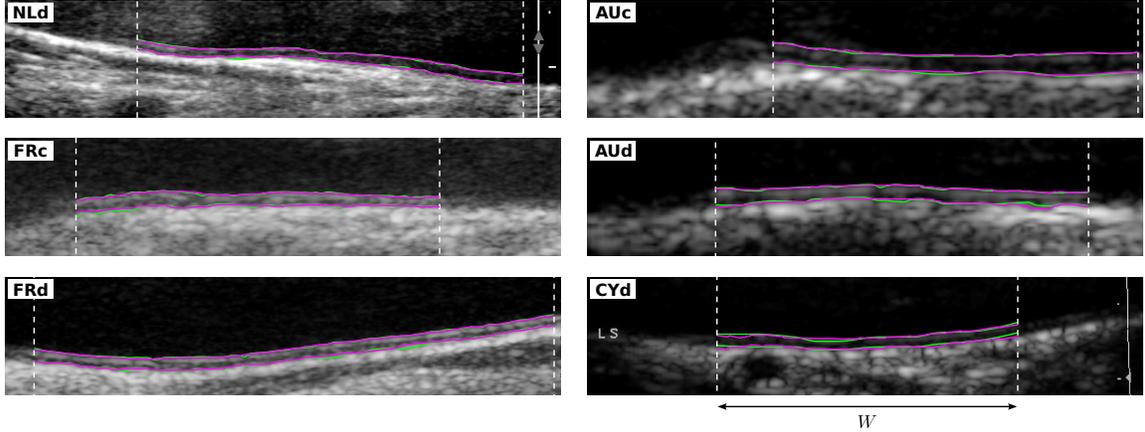


Figure 6: Results of the segmentation method (violet lines) compared to the reference contours traced by  $\mathcal{A}_1$  (green lines), in subjects with varying image quality from the six cohorts involved. The left and/or right borders presenting poor image quality have been clipped out (vertical dashed lines) prior to segmenting the contours in the sub-image  $\mathcal{I}_W$  of width  $W$ .

519 The IMT estimated with the proposed method was in good correlation with  
 520 reference tracings performed by  $\mathcal{A}_1$  ( $R = 0.87$ ), as depicted in Figure 7. The good  
 521 agreement between the estimated IMT and the reference is also demonstrated in the  
 522 Bland-Altman plot (Fig. 8), with a 95% confidence interval equal to  $158 \mu\text{m}$  and a  
 523 signed bias of  $-11 \mu\text{m}$ . The error introduced by the proposed method was very close  
 524 to the inter- and intra-analyst variability, as displayed in Figure 9.

525 The computation time is proportional to the width of the investigated ROI and  
 526 inversely proportional to the pixel size. On average, for a pixel size of  $57 \mu\text{m}$  and a  
 527 segment of width  $W = 20 \text{ mm}$  (350 pixels), the processing time was 0.8 s per image.

528 The errors obtained with the previous method developed by our team (Zahnd  
 529 et al., 2013b), are reported in Table 5. These results are to be compared with light-  
 530 gray rows in Table 4. Comparing mean errors and standard deviations in the first  
 531 column of these tables, it can be seen that the present method overall more accurately  
 532 extracted LI and MA interfaces. More specifically, the mean absolute error for both  
 533 LI and MA interfaces was smaller, with a near-statistically significant  $p$ -value of 0.08  
 534 and 0.07, respectively. Although the mean absolute error resulting from the IMT  
 535 estimation was slightly larger, the difference was not statistically significant ( $p =$   
 536 0.59). Moreover, contours resulting from the previous method had a much rougher  
 537 aspect, since their smoothness was not explicitly controlled, as depicted in Figure 10.

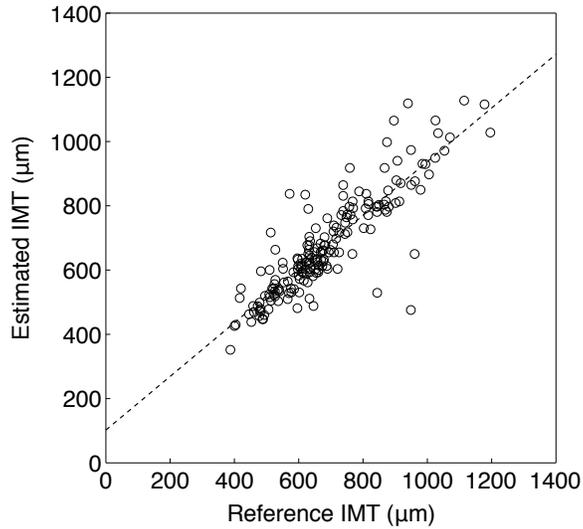


Figure 7: Linear regression line between the IMT estimated by the proposed method and the manual reference performed by  $\mathcal{A}_1$ , for all 184 images of the testing set.

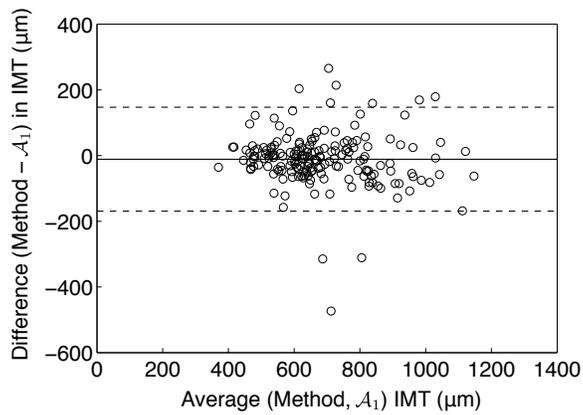


Figure 8: Bland-Altman plot, comparing the estimates of the IMT between the proposed method and the manual tracings performed by  $\mathcal{A}_1$ , for all 184 images of the testing set. The solid and dashed lines represent the bias and the 95% limits of agreement, respectively.

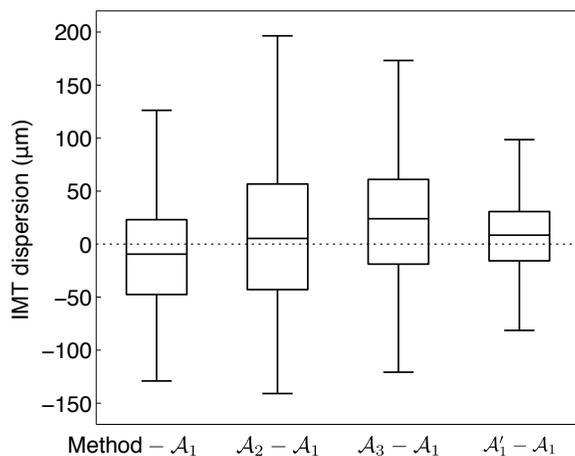


Figure 9: Box plot representing the dispersion between the IMT estimated with the proposed method and the manual tracings performed by  $\mathcal{A}_1$ , compared to the inter- and intra-analyst variability, for all 184 images of the testing set. Percentiles are indicated by boxes (25<sup>th</sup> and 75<sup>th</sup>), inner lines (50<sup>th</sup>), and error bars (5<sup>th</sup> and 95<sup>th</sup>).

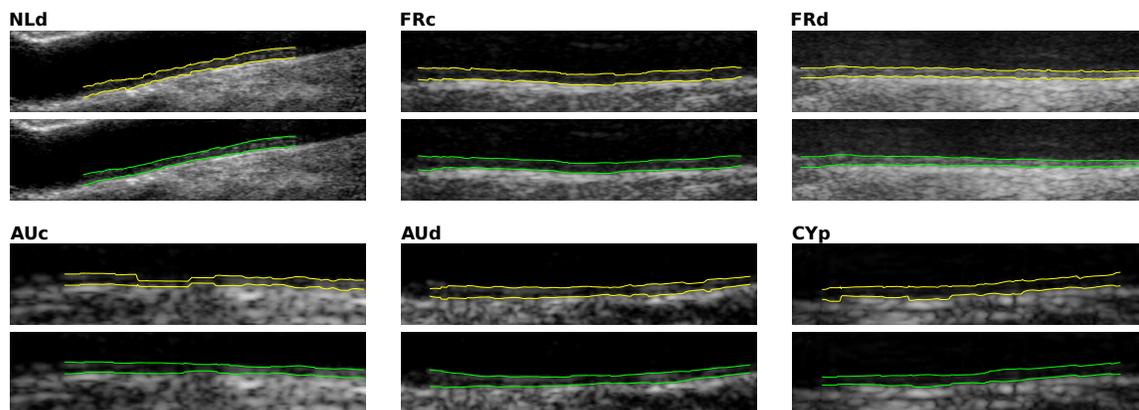


Figure 10: Example of segmented contours, for each cohort, resulting from the method previously proposed (Zahnd et al., 2013b) (yellow) and the present approach (green). The results show that the proposed method controls the smoothness of the contours.

Table 4: Segmentation absolute errors (mean  $\pm$  SD) in  $\mu\text{m}$ , for the proposed method with manual initialization and the manual tracings performed by the analysts  $\mathcal{A}_{1,2,3}$ .

Cohort	All ( $n = 244$ )	NLd ( $n = 43$ )	FRc ( $n = 57$ )	FRd ( $n = 25$ )	AUc ( $n = 30$ )	AUd ( $n = 30$ )	CYd ( $n = 59$ )
<b>Training set</b>	$n = 60$	$n = 13$	$n = 19$	$n = 8$	$n = 0$	$n = 0$	$n = 20$
<b>Lumen-Intima</b>							
Method vs $\mathcal{A}_1$	$52 \pm 73$	$64 \pm 96$	$33 \pm 27$	$32 \pm 27$	–	–	$97 \pm 110$
Method reproducibility	$27 \pm 85$	$33 \pm 28$	$11 \pm 10$	$12 \pm 10$	–	–	$67 \pm 170$
Inter-analysts	$49 \pm 53$	$76 \pm 75$	$44 \pm 42$	$31 \pm 30$	–	–	$102 \pm 70$
Intra-analyst	$31 \pm 33$	$54 \pm 50$	$28 \pm 26$	$19 \pm 16$	–	–	–
<b>Media-Adventitia</b>							
Method vs $\mathcal{A}_1$	$57 \pm 61$	$66 \pm 66$	$44 \pm 51$	$37 \pm 33$	–	–	$93 \pm 76$
Method reproducibility	$30 \pm 125$	$33 \pm 36$	$11 \pm 10$	$13 \pm 11$	–	–	$77 \pm 255$
Inter-analysts	$54 \pm 60$	$102 \pm 90$	$41 \pm 42$	$43 \pm 42$	–	–	$81 \pm 65$
Intra-analyst	$35 \pm 43$	$58 \pm 49$	$30 \pm 41$	$28 \pm 33$	–	–	–
<b>IMT</b>							
Method vs $\mathcal{A}_1$	$67 \pm 90$	$31 \pm 24$	$28 \pm 20$	$22 \pm 24$	–	–	$144 \pm 122$
Method reproducibility	$16 \pm 73$	$5 \pm 8$	$2 \pm 3$	$4 \pm 4$	–	–	$42 \pm 125$
Inter-analysts	$57 \pm 46$	$61 \pm 46$	$51 \pm 40$	$42 \pm 30$	–	–	$149 \pm 54$
Intra-analyst	$25 \pm 24$	$31 \pm 34$	$25 \pm 18$	$17 \pm 16$	–	–	–
<b>Testing set</b>	$n = 184$	$n = 30$	$n = 38$	$n = 17$	$n = 30$	$n = 30$	$n = 39$
<b>Lumen-Intima</b>							
Method vs $\mathcal{A}_1$	$49 \pm 62$	$52 \pm 59$	$39 \pm 33$	$35 \pm 38$	$36 \pm 30$	$42 \pm 41$	$92 \pm 113$
Method reproducibility	$18 \pm 28$	$28 \pm 24$	$11 \pm 15$	$11 \pm 10$	$17 \pm 15$	$17 \pm 16$	$30 \pm 55$
Inter-analysts	$56 \pm 51$	$68 \pm 60$	$47 \pm 40$	$31 \pm 29$	$70 \pm 57$	$80 \pm 52$	$85 \pm 68$
Intra-analyst	$36 \pm 36$	$50 \pm 43$	$34 \pm 36$	$26 \pm 34$	$34 \pm 28$	$37 \pm 35$	–
<b>Media-Adventitia</b>							
Method vs $\mathcal{A}_1$	$59 \pm 67$	$72 \pm 60$	$43 \pm 42$	$40 \pm 42$	$52 \pm 49$	$54 \pm 61$	$104 \pm 109$
Method reproducibility	$17 \pm 26$	$31 \pm 31$	$11 \pm 10$	$12 \pm 10$	$17 \pm 18$	$14 \pm 13$	$23 \pm 49$
Inter-analysts	$69 \pm 72$	$113 \pm 95$	$40 \pm 38$	$39 \pm 40$	$100 \pm 78$	$89 \pm 79$	$92 \pm 72$
Intra-analyst	$45 \pm 50$	$65 \pm 55$	$33 \pm 40$	$34 \pm 46$	$53 \pm 48$	$59 \pm 63$	–
<b>IMT</b>							
Method vs $\mathcal{A}_1$	$67 \pm 67$	$52 \pm 44$	$29 \pm 21$	$36 \pm 21$	$53 \pm 31$	$62 \pm 58$	$145 \pm 87$
Method reproducibility	$5 \pm 8$	$6 \pm 10$	$2 \pm 3$	$3 \pm 5$	$6 \pm 6$	$7 \pm 10$	$7 \pm 10$
Inter-analysts	$59 \pm 52$	$87 \pm 67$	$48 \pm 36$	$35 \pm 27$	$52 \pm 51$	$58 \pm 53$	$106 \pm 28$
Intra-analyst	$34 \pm 37$	$36 \pm 26$	$31 \pm 29$	$27 \pm 28$	$31 \pm 23$	$44 \pm 64$	–

Table 5: Segmentation absolute errors (mean  $\pm$  SD) in  $\mu\text{m}$ , between the previous method that was developed by our team (Zahnd et al., 2013b) and the manual tracings from  $\mathcal{A}_1$ , in the full exploitable width of each image of the testing set.

	All ( $n = 184$ )	NLd ( $n = 30$ )	FRc ( $n = 38$ )	FRd ( $n = 17$ )	AUc ( $n = 30$ )	AUd ( $n = 30$ )	CYd ( $n = 39$ )
Lumen-Intima	$55 \pm 81$	$46 \pm 53$	$37 \pm 35$	$33 \pm 34$	$56 \pm 92$	$45 \pm 51$	$119 \pm 138$
Media-Adventitia	$68 \pm 81$	$67 \pm 69$	$42 \pm 44$	$42 \pm 37$	$71 \pm 61$	$65 \pm 73$	$138 \pm 130$
IMT	$64 \pm 75$	$45 \pm 37$	$34 \pm 30$	$28 \pm 22$	$54 \pm 47$	$59 \pm 63$	$131 \pm 109$

538 *Temporal variation of the IMT during the cardiac cycle*

539 The temporal variation of the IMT was assessed in all available cine-loops, and  
 540 a cyclic compression-decompression pattern could be observed during consecutive  
 541 cardiac cycles, as displayed in Figure 11. For all subjects, the diastolic IMT was  
 542 systematically greater than the systolic IMT, namely  $766 \pm 174 \mu\text{m}$  *vs.*  $670 \pm 163 \mu\text{m}$ ,  
 543 respectively. The average amplitude  $\Delta_{\text{IMT}}$  of the temporal variation of the IMT  
 544 during the cardiac cycle was  $97 \pm 43 \mu\text{m}$ , and corresponded to  $13 \pm 5\%$  of the end-  
 545 diastole IMT.

546 Measurements from the diseased subjects (*i.e.*, NLd, FRd and AUd,  $n=98$ ) were  
 547 then compared to the healthy controls (*i.e.*, FRc and AUc,  $n=87$ ), from both training  
 548 and testing sets. The end-diastole IMT value, estimated with the fully-automatic  
 549 segmentation method, was significantly higher in at-risk patients compared to healthy  
 550 controls ( $855 \pm 161$  *vs.*  $666 \pm 129 \mu\text{m}$ ,  $p < 0.0001$ ). Furthermore, the amplitude  
 551  $\Delta_{\text{IMT}}$  was also significantly higher in the diseased population compared to healthy  
 552 volunteers ( $106 \pm 48$  *vs.*  $86 \pm 34 \mu\text{m}$ ,  $p = 0.001$ ), as shown in Figure 12. The  
 553 end-diastole IMT value was not correlated with the amplitude  $\Delta_{\text{IMT}}$  of the temporal  
 554 variation of the IMT during the cardiac cycle ( $R = 0.3837$ ).

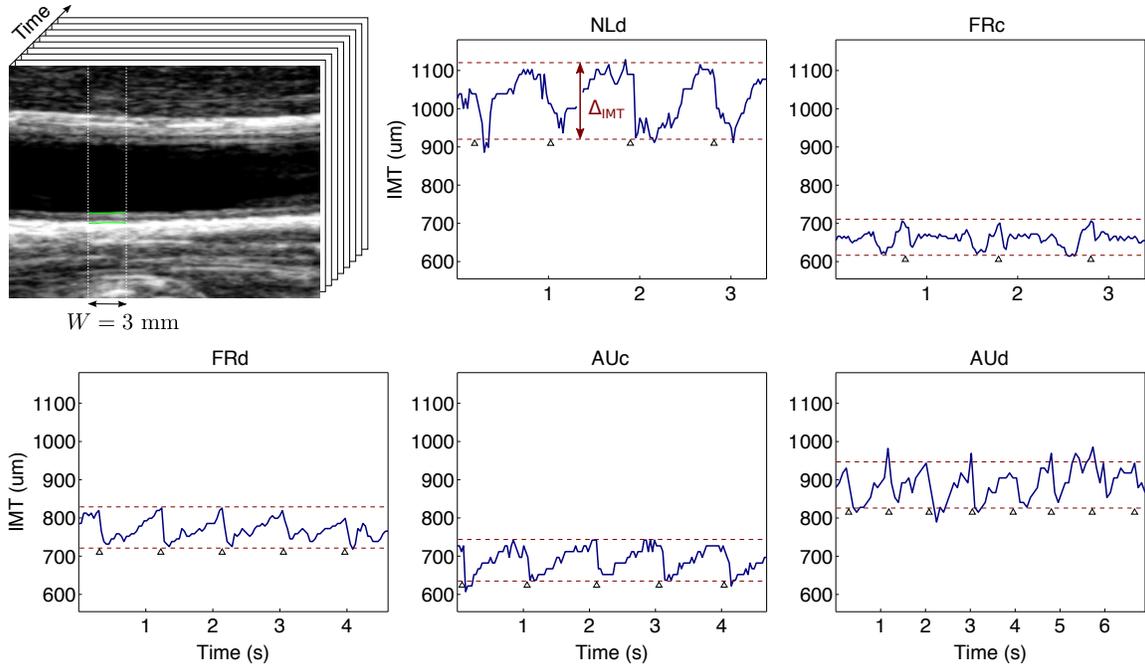


Figure 11: Temporal variations in IMT during several consecutive cardiac cycles. The IMT was quantified on a local region of reduced width ( $W = 3$  mm) for all frames of the cine-loop (top left panel). Examples of the cyclic compression-decompression pattern of the IMT, resulting from the proposed fully-automatic method, are provided for subjects from different cohorts. The  $\Delta_{\text{IMT}}$  parameter corresponds to the total amplitude of the IMT variations (dashed lines). The end of the diastole is indicated by the triangle markers.

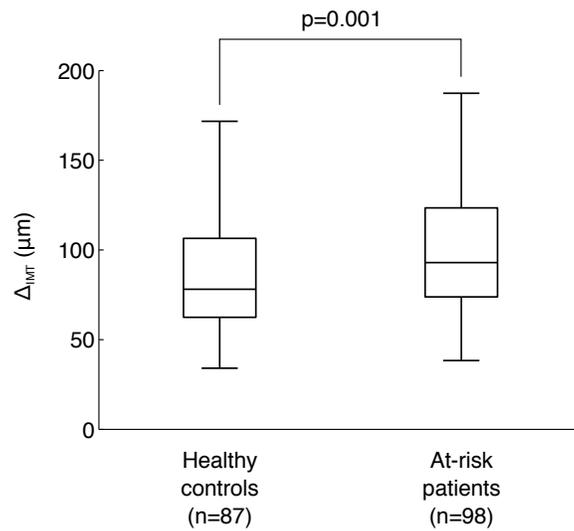


Figure 12: Box plot representing the amplitude  $\Delta_{\text{IMT}}$  of the temporal variations in IMT during the cardiac cycle, for healthy controls and at-risk patients. Percentiles are indicated by boxes (25th and 75th), inner lines (50th) and error bars (5th and 95th). The result of the Mann-Whitney U test is indicated by the  $p$  value.

555 **Discussion**

556 The main contribution of this study is an original segmentation method devised  
557 to simultaneously extract, from noisy images, both interfaces of thin structures with  
558 nearly parallel contours. Although the method is fully-automatic, its implementation  
559 allows quick interactive corrections, if necessary, at each step of the algorithm. It  
560 was tuned to segment out the carotid artery intima-media complex in 2D US images.  
561 These anatomical interfaces were successfully extracted for 244 subjects from six  
562 different cohorts. The contours obtained and the calculated IMT values were close  
563 to manual tracings from expert analysts, with spatial errors of the order of the pixel  
564 size. The temporal variation of the IMT during the cardiac cycle was finally assessed  
565 with use of the proposed method, and the amplitude of this variation was significantly  
566 higher in at-risk patients compared to healthy controls, which is in accordance with  
567 recent findings.

568 *Contour segmentation framework*

569 The method extracts a single medial axis, from which the contours of both LI  
570 and MA interfaces can be fully deduced. Among all possible solutions, the globally  
571 optimal medial axis is determined by means of a specific dynamic programming  
572 scheme, in which a 3D cost function represents, for each column  $x$  of the image, all  
573 potential radial positions,  $y$ , of the medial axis, as well as all potential IMT values,  $z$ .  
574 Similar approaches have previously been introduced, to extract a globally optimal  
575  $(n+1)$ -dimensional path that describes the  $n$ -dimensional contour of tubular objects,  
576 with a front propagation technique based on a multi-scale filter (Wink et al., 2004),  
577 as well as with fast-marching methods exploiting spheres of varying radius (Li and  
578 Yezzi, 2007; Benmansour and Cohen, 2011).

579 The front propagation scheme (Eq. 5) favors solutions that are located mid-way  
580 between two strong positive gradients (namely, LI and MA interfaces), while penaliz-  
581 ing contours with irregular shapes. More precisely, the smoothness of the medial axis  
582 is ruled by two penalty terms: the first one penalizing abrupt variations of the radial  
583 position of the medial axis between adjacent columns, and the second one penalizing  
584 sudden IMT variation, which would result in spiky contours even with a perfectly  
585 smooth medial axis. In the case of a cine-loop, the medial axis from the previous  
586 image is exploited as *a priori* information to characterize a spatial transformation  $T$ ,  
587 used to encourage solutions presenting a similar shape and curvature. In the case of  
588 a single image, a similar process is used by exploiting the information derived from  
589 the initialization phase. The resulting output corresponds to a pair of continuous  
590 and smooth contours that still capture the subtle spatial variation of the IMT within  
591 a given image (Fig. 6).

592 The process is fully-automatic and does not require any user interaction. Nev-  
593 ertheless, manual specification of part (or all) of the initialization settings can still  
594 be performed by the operator in a quick and simple way *via* a convenient graphical  
595 interface. Such manual corrections are sometimes necessary to increase the overall ro-  
596 bustness of the method by avoiding errors that can be caused by the fully-automatic  
597 process in some challenging cases. Manual initialization also enables the user to per-  
598 form a case-by-case analysis by specifying the center of the desired ROI as well as  
599 its width for each subject.

### 600 *Method evaluation*

601 The contour extraction method was applied in a set of 244 *in vivo* images gath-  
602 ered from six different cohorts and four different research centers. Quite a large  
603 discrepancy could be observed between the different cohorts in terms of image char-  
604 acteristics and quality: FRc and FRd had a high image quality with generally well  
605 visible anatomical interfaces, AUc and AUd had a medium image quality with more  
606 blurry interfaces, NLd had an overall high image quality, but the interfaces were  
607 not clearly detailed as the images were zoomed out, and CYd had a challenging  
608 image quality with poorly visible interfaces. Image acquisition was performed with  
609 a different scanner in each of the four medical centers, and a discrepancy was also  
610 present in the spatial resolution of the images, with a pixel size equal to  $94 \pm 9 \mu\text{m}$   
611 for NLd,  $34 \pm 5 \mu\text{m}$  for FRc and FRd,  $52 \pm 4 \mu\text{m}$  for AUc and AUd,  $67 \pm 0 \mu\text{m}$  for  
612 CYd. Moreover, the discrepancy in image quality among the analyzed data was also  
613 reflected by significantly varying width of the exploitable region,  $\mathcal{I}_W$ , as assessed by  
614 observer  $\mathcal{A}_1$  (range 10 – 100 % of the full image width).

615 The accuracy of the method, for both the fully-automatic and manual initial-  
616 ization, was of the same order of magnitude as inter- and intra-analysts variability  
617 (Tables 3, 4). Overall, the mean absolute error corresponded approximately to one  
618 pixel. Variability of the method with the manual initialization process was sub-  
619 stantially smaller than inter- and intra-analysts variability, which reflects a better  
620 reproducibility. Variability of the method with the fully-automatic initialization  
621 process was null (*i.e.* fully reproducible results), except when the automatic process  
622 failed (*i.e.* 12 cases out of 184) and had to be manually corrected. Variability due to  
623 such a manual correction was not assessed, but it is likely to be very close to vari-  
624 ability of the manual initialization process for these specific cases (5% of the whole  
625 population), but – averaged over the whole population – it is likely to be negligible.

626 The robustness of the method could also be verified, since images were separated  
627 between a training set, used to determine the optimal parameters settings, and a  
628 testing set, used to evaluate the method in independent test images. Let us first

629 note that the errors for the testing set were of the same magnitude as the errors  
630 for the training set, which reflects a certain consistency in the performances of the  
631 method. Moreover, when applying the method on the AUc and AUd datasets without  
632 any prior training on these two cohorts, good results were also obtained. In view  
633 of the above, the robustness of the method was demonstrated, since 244 images  
634 were successfully segmented using the same parameter settings, despite differences  
635 in arterial geometry, scanner configuration, and image quality.

636 A variety of methods have been introduced to segment the carotid artery in longi-  
637 tudinal B-mode US images. When comparing the results of the present framework to  
638 previous work reported in recent review articles (Molinari et al., 2010; Ilea et al., 2013;  
639 Loizou, 2014), the present method would rank among the most accurate state-of-the-  
640 art techniques. Nevertheless, such comparisons are to be taken carefully, since the  
641 design of the evaluation protocol can greatly differ between distinct studies. Among  
642 the most encountered differences are the nature of the cohorts involved (*e.g.*, number  
643 of subjects and images, healthy volunteers and/or at-risk patients), the characteris-  
644 tics of the images (*e.g.*, use of several different scanners, spatiotemporal resolution),  
645 the distance metric (*e.g.*, mean absolute radial distance, or polyline distance), the  
646 dimensions of the ROI (*e.g.*, full image width, limited region), the properties of the  
647 initialization process (*e.g.*, fully-automatic or with manual input from the user), and  
648 the characteristics of the manual reference annotations (*e.g.*, discrete control points  
649 or continuous tracings, inter- and intra-analysts variability). To thoroughly estab-  
650 lish a genuine comparison between multiple state-of-the-arts approaches, a challenge  
651 would be necessary, where evaluation would be carried out homogeneously and in  
652 the same datasets for all the methods.

### 653 *Temporal variation of the IMT*

654 The compression-decompression pattern of the arterial layers during the cardiac  
655 cycle was analyzed by extracting the IMT in every frame of the cine-loop (Fig. 11).  
656 This operation was performed in a ROI of reduced width  $W = 3$  mm, located in a  
657 well-contrasted region with clearly visible interfaces. From our experience, such a  
658 narrow region with good image quality is better suited to extract the minute IMT  
659 variations compared to the full width of the image (Zahnd et al., 2013b).

660 The IMT was significantly higher in at-risk patients compared to healthy controls  
661 ( $855 \pm 161$  vs.  $666 \pm 129$   $\mu\text{m}$ ,  $p < 0.0001$ ), which reflects arterial wall thickening (Bal-  
662 dassarre et al., 2000). Moreover, the diastolic IMT was systematically greater than  
663 the systolic IMT for all subjects ( $766 \pm 174$   $\mu\text{m}$  vs.  $670 \pm 163$   $\mu\text{m}$ ), which is in  
664 accordance with previous studies (Selzer et al., 2001; Meinders et al., 2003). The  
665 amplitude  $\Delta_{\text{IMT}}$  of the IMT variation during the cardiac cycle was also quantified,

666 and the average value for all subjects was  $97 \pm 43 \mu\text{m}$ . This is higher than the  
667 results that were reported in previous studies that indicated a  $\Delta_{\text{IMT}}$  value around  
668  $50 \mu\text{m}$  (Boutouyrie et al., 2001; Meinders et al., 2003; Polak et al., 2012b,a). These  
669 differences may be explained by the differences in the characteristics of the subjects  
670 involved, contour segmentation methods, and measurement protocols.

671 Interestingly,  $\Delta_{\text{IMT}}$  was significantly increased in at-risk patients compared to  
672 healthy controls ( $106 \pm 48$  vs.  $86 \pm 34 \mu\text{m}$ ,  $p = 0.001$ , Fig. 12); this result is  
673 in agreement with recent findings showing that the amplitude of the compression-  
674 decompression pattern of the arterial wall is associated with cardiovascular risk fac-  
675 tors (Boutouyrie et al., 2001; Meinders et al., 2003; Polak et al., 2012a; Zahnd et al.,  
676 2013b). More specifically,  $\Delta_{\text{IMT}}$  has been demonstrated to be associated with pulse  
677 pressure, LDL-cholesterol, and age (Polak et al., 2012b). Although it is beyond the  
678 scope of the present study, these three factors might contribute to explain the overall  
679 increase of the parameter  $\Delta_{\text{IMT}}$  in the diseased population.

680 Each subject can thus be characterized by two different parameter values, namely  
681 the diastolic IMT and the parameter  $\Delta_{\text{IMT}}$ , as displayed in Figure 11. Although the  
682 exact association of  $\Delta_{\text{IMT}}$  with cardiovascular risk is still exploratory, this finding  
683 can contribute to shed further light on potential early-stage symptoms of atheroscle-  
684 rosis. IMT is a *static* parameter that marks the *presence* of atherosclerotic disease.  
685 Nevertheless, it has recently been demonstrated to have limited additional predic-  
686 tive power over conventional risk factors (Kavousi et al., 2012). On the other hand,  
687 the amplitude  $\Delta_{\text{IMT}}$  is a *dynamic* parameter that is likely to be directly related to  
688 the *effect* of the disease on mechanical properties of the arterial wall, as well as the  
689 functional role of arterial-wall cells. Furthermore, compression of the arterial wall is  
690 a crucial physiological mechanism, as it enables fluid exchange between blood and  
691 surrounding tissues (Chuong and Fung, 1984). Let us also recall that the end-diastole  
692 IMT value was not correlated with  $\Delta_{\text{IMT}}$  ( $R = 0.3837$ ). For all these reasons, the  
693 parameter  $\Delta_{\text{IMT}}$  is likely not to replicate information already provided by static IMT  
694 measurements, but may instead yield complementary information in cardiovascular  
695 risk assessment. Such analysis was out of the scope of the present study.

696 When attempting to assess the temporal variation of the IMT in US cine-loops,  
697 several recommendations should be considered. First, the spatial resolution of the  
698 scanner plays a major role. The total amplitude of the compression phenomenon  
699 is around  $100 \mu\text{m}$ , therefore a pixel size  $> 100 \mu\text{m}$  is likely to yield inaccurate re-  
700 sults, despite the tenfold interpolation used in the method. The accuracy of the  
701 method is expected to increase when decreasing the pixel size. Although a finer spa-  
702 tial resolution might result in an increment of the computational time, this may be  
703 compensated by as smaller interpolation factor. Second, as the studied phenomenon

704 is dynamical, the temporal resolution (frame-rate) should be sufficient to capture  
705 several images per cardiac cycle. From our experience, frame-rates  $> 15$  fps are  
706 generally sufficient to assess the compression amplitude, but higher values may be  
707 necessary for subjects with particularly fast heartbeat and/or to study the shape of  
708 the compression curves. It should however be noted that increasing the temporal  
709 resolution generally results in decreasing the spatial resolution. Third, image acqui-  
710 sition should be performed during at least one entire cardiac cycle to capture the full  
711 compression-decompression pattern. Acquiring several cardiac cycles may be useful  
712 to exploit the reproducibility of this pattern, but all the acquired frames must rep-  
713 resent the same section of the arterial wall. Therefore, special care shall be taken to  
714 maintain the transducer steady, while the patient refrains himself/herself from mov-  
715 ing, swallowing, and – preferably – breathing. In practice, we recommend acquiring  
716 three cardiac cycles, as a reasonable trade-off between reproducibility, and the risk  
717 of undesired effects due to patient’s movements or unsteady transducer position.

### 718 *Limitations and perspectives*

719 The method is based on a 3D front propagation approach (namely, along the  
720 radial, longitudinal, and IMT dimensions) to extract a pair of spatially smooth con-  
721 tours from a 2D image. However the temporal dimension *per se* is not implemented  
722 in the front propagation scheme (Eq. 5). When processing several consecutive images  
723 in a cine-loop, the output from a given frame is used as a robust *a priori* information  
724 to initialize the search in the following frame. Although the contours are accurately  
725 extracted in each individual frame, the up-and-down motion of these contours be-  
726 tween successive images is not regularized by a smoothing parameter. Accordingly,  
727 taking into consideration the spatio-temporal consistency of the contours by extend-  
728 ing a previously proposed surface-fitting approach (Metz et al., 2011) to a 3D+t  
729 volume could potentially contribute to the improvement of the method.

730 According to the Mannheim expert consensus (Touboul et al., 2012), optimal  
731 IMT measurements should be performed at least 5 mm from the carotid bifurcation.  
732 In the proposed fully-automatic scheme, the selection of the assessed region is based  
733 on image quality, and may therefore, in some cases, be closer than 5 mm from the  
734 bulb. The expert consensus also recommends to carry out IMT measurements over  
735 a section of 10 mm. In the present study, the IMT variation was evaluated on a  
736 narrow region of 3 mm because, from our experience, this width provides the opti-  
737 mal results to extract a reproducible compression-decompression pattern (Fig. 11).  
738 Curves obtained from regions  $\ll 3$  mm are often corrupted by noise because too few  
739 local measurements are averaged to generate the IMT value and, conversely, curves  
740 from regions  $\gg 3$  mm are generally flat smooth lines without much variation due to

741 excessive averaging; in both cases the cycles cannot be clearly perceived, and the am-  
742 plitude  $\Delta_{\text{IMT}}$  cannot be reliably quantified. It is also noteworthy that the temporal  
743 variation of the IMT is an emerging dynamical parameter and is not mentioned in  
744 the consensus. Moreover, the full exploitable width of all the analyzed images ranged  
745 from 3 to 51 mm, and therefore some cases could not have met the 10 mm required  
746 by the consensus. Nevertheless, the user can easily and quickly select the center of a  
747 more suited region (1 mouse click) and/or modify the width of the analyzed section  
748 (2 mouse clicks), if necessary.

749 The purpose of the method is to extract both LI and MA interfaces of the ar-  
750 terial wall in images of asymptomatic subjects to extract quantitative markers of  
751 early-stage atherosclerosis, such as arterial wall thickening. The method might be  
752 sub-optimal when applied to images presenting an atherosclerotic plaque, since the  
753 rationale is to detect two nearly-parallel contours. A distinct set of parameters (*viz.*,  
754 with smaller smoothness penalties and a larger tolerated thickness) could enable  
755 processing of such images with a plaque – thus yielding a measurement of plaque  
756 thickness and length, but potentially also (semi-) automatic plaque detection – yet  
757 it is likely to hinder the performances of the method in asymptomatic regions.

## 758 *Conclusion*

759 A fully-automatic method based on front propagation was introduced to segment  
760 the the intima-media complex in carotid B-mode ultrasound. A globally optimal 3D  
761 path was extracted to describe a pair of 2D contours. The method was devised to  
762 accurately extract the IMT in every column of the image. The performances of the  
763 framework were validated in 184 independent test images. The proposed approach  
764 was successfully applied to quantify the amplitude of the compression-decompression  
765 pattern of the intima-media complex during the cardiac cycle, which may potentially  
766 be a risk marker of early-stage atherosclerosis. The algorithm was implemented in a  
767 graphical user interface and is available on-line: [https://www.creatis.insa-lyon.  
768 fr/carolab/](https://www.creatis.insa-lyon.fr/carolab/).

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