A LOW-COST MEASUREMENT SYSTEM FOR THE 3-D EVALUATION OF CAROTID PLAQUES BASED ON ULTRASOUND IMAGES

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Abstract- The paper describes an improved low-cost add-on measurement system for the 3-D analysis of carotid arteries. The system locates in the 3-D space the contours of vessel walls extracted from a series of 2-D acquisitions, thanks to the measurement of the geometrical parameters of the ultrasonic probe positions and orientations. Features of the proposed system include the possibility to realize free-hand measurements, to work with the desired number of images and to process both normal B-mode and Power-angio images. In the paper, the realized prototype will be described and its operation will be detailed and characterized both for the software and the hardware prototype.

I. Introduction

The development and the realization of a measurement system for medical applications is strictly related to the pathology to be analyzed. The atherosclerosis involves the buildup of fatty deposits in the innermost lining of large and medium-sized arteries (Figure 1) and often leads to coronary heart disease, strokes, hardening of the arteries and other disorders because of the occurrence of blood clots in the narrowed arteries. For these reasons many studies are aimed to the detection and location of thickenings and plaques in carotids. Many non-invasive methods, as the ultrasound techniques, such as B-mode, Color Doppler, Color Power Angio (CPA) can clearly detect these phenomena. In fact, the analysis of the obtained images allows a diagnostician to find out a plaque and to evaluate its nature and shape. Most clinical ultrasound tests are based on 2-D image analysis thanks to its simplicity and promptness. However, with this kind of tests some spatial information, in particular about the carotid volume occlusion, could be lost. Consequently, there is a growing clinical interest about 3-D ultrasound imaging, since a 3-D visualization of the carotid artery helps in the accurate definition of the 3-D size and geometry of atherosclerosis plaques [1]-[4]. Different 3-D ultrasound instruments are present on the market, but most of them either are able to make only qualitative measurements or are highly expensive. The 3-D ultrasound devices available on the market can be classified into two classes: i) "add-on" devices, and ii) devices employing sensor arrays. i) "Add-on" devices exploit ordinary 2-D ultrasound probes, but have additional hardware in order to locate each image of the set of acquired images in the 3-D space. By suitable spacing the different image planes, the volume of interest can be covered adequately, but the position and the orientation of the probe has to be measured accurately, in order to avoid errors in the 3-D reconstruction. Besides the free-hand scanning, another solution consists in the mechanical positioning of the probe. In this case, the position and orientation of the probe are evaluated by measuring the orientations of the arms of a mechanical support, allowing the user to program the scanning planes beforehand and to avoid gaps in the volume of interest. ii) Sensor array devices employ special probes, containing arrays of ultrasound emitter/receivers for which the image scanning plane can be changed electronically. Disadvantages of such systems are the dimensions and the cost, even though recently the technology has allowed the realization of smaller probes. Although add-on devices could be less expensive than sensor array devices, the need for accurate probe position measurement leads to employ complex and expensive sensors, such as magnetic sensors, whose prices are between 2500 € and 5000 €.

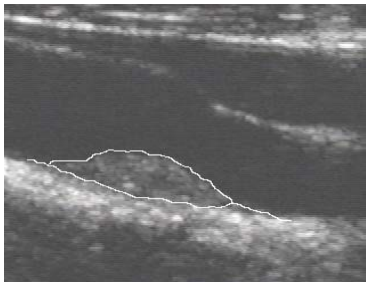

Figure 1. Example of a fatty carotid plaque. The plaque contours have been highlighted

In this framework the authors designed and realized a low cost add-on ultrasound 3-D imaging prototypal system for accurate carotid plaque volume measurements [5-7], which allows to: (i) reconstruct the volumetric image of the carotid artery, (ii) evaluate the plaque volume, and (iii) calculate the percentage of carotid occlusion (stenosis) The realized system is composed of a stage that contains the probe and the necessary hardware to capture the 2-D images and of a measurement sub-system of the angle between the start analysis plane and current scanning plane, called the insonification angle, and of a processing software. In [6] the insonification angle was evaluated applying a package onto the probe provided with a goniometer. Even if perfectly working, the previous realized prototype has shown some limits that have made its use troublesome during a clinical examination made by a doctor: a) the measurement procedure was not flexible, in particular once the probe has been positioned on the patient neck and the operator has reached the best longitudinal view no probe translations were allowed, thus only the insonification angle could be changed. Thus the doctor had to avoid any lateral movement of the probe, since only tilt movements were allowed. b) Due to the typical carotid size and neck width, the insonification angle ranges in the [–5°,+5°] interval. Since the resolution of the manual goniometer was 2°, only a few images were worthy to be analyzed, and a poor reconstruction was thus obtained. c) The 3-D processing software was not optimized to deal with a high number of images, and, mainly, if the edge detection failed on a single image, the whole measurement procedure had to be repeated. This paper proposes an improvement to the previous prototype which overcomes these limitations. Furthermore, the proposed system has been developed with particular attention to the costs, such that in the final version of the prototype this is of the order of 2000 €.

II. The Improved Prototype

In order to overcome the abovementioned limits, many changes were made both to the hardware and to the software parts of the prototype. A schematic of the new realized system is reported in Figure 2, where the ATL HDI 5000 is the ultrasound machine to which the proposed system has been connected. The overall structure of the system can be divided into two parts: the hardware add-on package and the PC processing software. Besides the image acquisition, also geometrical parameters (i.e. both the insonification angle and the lateral translation) related to the acquisition of a single 2-D image are measured in the proposed system and automatically sent to the software procedure. In the following, a detailed description of the changes made is reported together with the advantages obtained.

A. The hardware

The hardware add-on package is composed of four main blocks: i) the angle measurement sub-system, ii) the CCD-based hardware for position tracking, iii) the frame grabber and the (iv) PC.

i) The insonification angle is measured by a specific sub-system based on a AccuSwitch™ Dual Axis Tilt Switch inclinometer with 0.2° resolution and +/- 20° range. The absolute measured X-oriented and Y-oriented angles can be returned using two PWM output signals, which have been interfaced with two counter ports of a low-cost 8-pin PIC12F675 Microchip micro-controller. The acquired data are sent to the PC processing software through a RS232 serial port. Since the micro-controller does not have a compatible RS232 USART port, the implementation of specific code for the serial communication has been necessary. The dual axis inclinometer changes the PWM duty cycle in the [0,1] range according to the angle value in the range [-20°,+20°]; a 0.5 duty cycle value means a 0° tilt orientation.

ii) To allow a correct 3D image reconstruction also the information about the current position of the ultrasound probe is required. In order to reduce the cost of the overall system, instead of measuring an absolute position, a relative measurement is performed. This one is made using an ad hoc device, composed of a small CCD

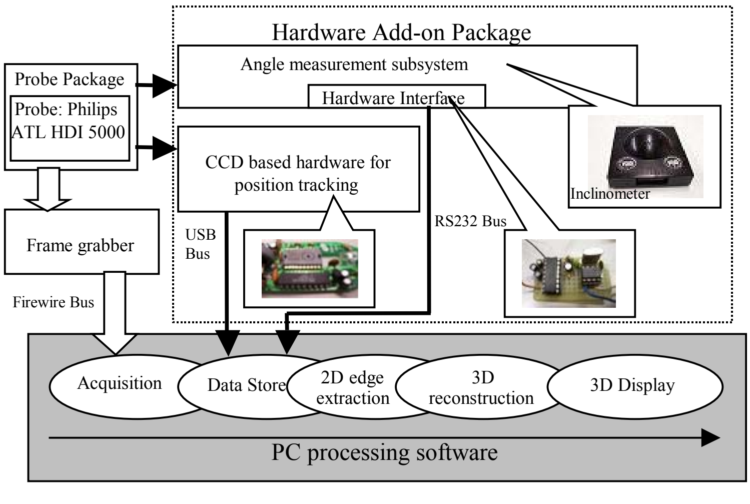

Figure 2. Schematic of the proposed prototype.

sensor and a DSP. This is the same architecture of a common optical mouse. The position sensor device is interfaced to the PC through the USB bus. The CCD continuously grabs images of the neck skin, and the DSP determines the shift vector between the current image and the previous one, thus allowing to know the shift of the probe over the skin. The resolution reached is of the order of 0.1 mm. The only limitation imposed to the measurement procedure is that the doctor must not raise the ultrasonic probe from the patient neck.
iii) To acquire the B-mode and CPA images coming from the ultrasound machine a PAL/NTSC compatible framer grabber device was used. This allows the connection of the add-on system with the most important ultrasound machines available on the market, grabbing the current display of the screen of the ultrasound machine and sending it to the PC using its Firewire bus.
iv) A laptop PC architecture was used. This allows the add-on package to be used also with portable echographic devices. On the PC a processing software runs which is able to acquire continuously all the data sent by the acquisition hardware and required by the measurement software. In the next section, the measurement procedure will be described. Further a metrological characterization will be reported.

B. The operation of the prototype

To start a 3-D measurement section, the doctor has to fix a common ultrasonic probe to the provided add-on package containing the position sensor, the inclinometer and the interfacing hardware (Figure 3) The first steps executed by the processing software are the acquisition and data storing routines. The latter creates an ad-hoc database that links each image grabbed by the frame grabber with the respective geometrical parameters. The acquisition of the image from the frame grabber can be triggered by an external event as the diastolic movement of the heart. In this way the carotid lumen is grabbed always at the same width. In its current version the software is able to store up to 15 bundles, made of a 2-D image and the data about the angles and the position of the probe. At first, the diagnostician has to select a box (Region of Interest, ROI) on any image; this box, that will appear with the same position and dimensions also over all other images, is the area of the image that will be processed in order to extract the 2-D outline of the relevant interfaces between different tissues in the image. Then the 2-D edge extraction. A useful feature of the proposed prototype is that if the edge detection fails on a single image, the diagnostician can choose between two solutions: whether to discard this image or to launch an edge correction routine. After all necessary corrections have been made to the interface contours, the 3-D reconstruction routine can be run, and the 3-D rendering of the carotid and of the possible plaque is shown from a point of view that the user can easily change with the mouse. Beside the 3-D rendering of the carotid, the program yields also other results such as the occlusion percentage, and the volume of the plaque, together with their uncertainties.

Figure 3. The add-on package

C. The processing algorithm

Each 2-D image is processed in order to extract the four interfaces: media-intima, intima-lumen in near wall, lumen-intima and intima-media in far wall. The contour detection algorithm can be applied to both normal B-mode and Power-angio images and employs a dynamic programming procedure [6][7]. This method is based on the recursive application of suitable cost functions. Namely, four cost functions are used, one for each interface. They are constituted by some terms taking into account the characteristics of the investigated contours: for instance, grey level above and below the contour, the contour regularity. In case the user introduces correction points by hand, an additional term considers the closeness of the contour to possible correction points. These terms are weighted by coefficients empirically evaluated during a training phase performed on different carotid images. The 2-D processing algorithm outputs four arrays of the pixel coordinates of the points representing the contours of the four wanted interfaces, for each acquired image.
The objective of the following 3-D reconstruction algorithm is to locate each point of each interface in the 3-D space with respect to a given reference system (O, x_i', y_i', z_i'), based on the knowledge of its pixel coordinates (x_i, y_i) and of the orientation and position values ($\theta_x, \theta_y, d_x, d_y$) of the probe, as measured during the acquisition of the considered image. First, the pixel coordinates are converted into image coordinates (x_i^*, y_i^*) in metric units (mm), obtained by applying conversion factors determined during the acquisition phase, depending on the resolution of the ultrasound machine. The change of

coordinates and the probe shift values achieved from the position measurement device are:

$$\begin{cases} x_i^* = x_i \cdot res_x \\ y_i^* = -y_i \cdot res_y \end{cases} \qquad \begin{cases} d_x^* = d_x \cdot \Delta_x \\ d_z^* = d_z \cdot \Delta_z \end{cases}$$

The center abscissa of the image region of interest can be estimated as: $\bar{x}^* = \left(\max x_i^* - \min x_i^*\right)/2$.

The transformation of coordinates from the image scanning plane and the 3-D space can be represented by the following relationship, in which the transformation matrix depends on the geometrical parameters of the acquisition:

$$\begin{pmatrix} x_i' \\ y_i' \\ z_i' \end{pmatrix} = \begin{pmatrix} \cos\theta_x & -\cos\theta_y * \sin\theta_x & d_x^* + \bar{x}^* * tg\theta_x * \sin\theta_x \\ -\sin\theta_x & \cos\theta_y * \cos\theta_x & \bar{x}^* * \sin\theta_x \\ 0 & \sin\theta_y & d_z^* \end{pmatrix} * \begin{pmatrix} x_i^* \\ y_i^* \\ 1 \end{pmatrix} \tag{1}$$

By processing each point of the four interfaces of each image, the 3-D reconstruction of the carotid can be displayed and saved. The final aim of the realized program is to estimate the values of two parameters of medical interest: the volume of the possible atherosclerotic plaque and the stenosis. The volume of the plaque is evaluated considering only the volume between the lumen-intima and intima-media, and as a summation of the areas of the polygons whose vertices are the points of the contours; each one of these polygons is a section of the plaque taken on a given transversal plane. For each polygon the area can be computed as

$$A_{pj} = \frac{1}{2} * \left| \sum_{i=1}^{N} \left(z_i' * y_{i+1}' - y_i' * z_{i+1}' \right) \right| \tag{2}$$

where N/2 is the number of images. Since the coordinates are already expressed in mm, then the plaque volume can be obtained by multiplying the value of A_{pj} for the calibration coefficient along the abscissa *x*, equal to res_x:

$$V = \sum_j A_{pj} \cdot \Delta_x \tag{3}$$

The stenosis is evaluated as the ratio of the areas of the two sections of the lumen that correspond, respectively, to the minimum and the maximum blood flow:

$$stenosis = A_{pj\,\min} \big/ A_{pj\,\max} \tag{4}$$

III. The prototype characterization

A. The hardware

A first metrological characterization has been done in order to evaluate both the angle subsystem measurement system and position sensor accuracy. The subsystem for the measurement of the angles has been characterized by a comparison with a computer controlled pan-tilt unit (PTU-46 by Directed Perception) having a 0.013° resolution. The experiment has been conducted fixing the angle measurement subsystem onto the pan-tilt unit, and varying the pan-tilt angle $\alpha_{\mu c}$ in the +/- 20° range with a resolution step of 1°. For each step 30 consecutive measurement were executed. For each measure, the PIC 12F675 estimates the T_{on} time of the PWM signal with a counter section with a clock period of 400 ns, and sends the counter reading to the PC. The obtained relationship between the averaged measurements X_c and the pan-tilt unit angles $\alpha_{\mu c}$ can be written as $\alpha_{\mu c} = m \cdot X_c + b = 0.0028 \cdot X_c - 71.097°$, with a *m* standard deviation of 0.0003 and a *b* standard deviation of 0.50. This relationship is assumed with a determination coefficient $R^2 = 0.9998$.

The position sensor has been characterized measuring the translation for the X axis and Y axis and comparing the results with those given by a micrometer measurement system. This one has a position resolution of 0.01mm. The experiment has been realized executing 20 consecutive tests in which the position sensor has been moved between two known positions fixed by the micrometer system. The obtained results shows a position resolution of : d_x=30.00 mm/300 steps = 0.1 mm and a standard deviation of 0.05mm.

B. The software

The uncertainties of the final measurement results (plaque volume and stenosis) have been evaluated by applying the uncertainty propagation law as stated by the ISO EN 13005 [8] to the relationships involved in the determination of these results from the input data. The uncertainty of the volume can be determined by applying the GUM procedure to eq. (3):

$$u_V^2 = \sum_j \left(\frac{\partial V}{\partial A_{pj}}\right)^2 * u_{A_{pj}}^2 + \left(\frac{\partial V}{\partial \Delta x}\right)^2 * u_{\Delta x}^2 + 2*\frac{\partial V}{\partial A_{pj}}*\frac{\partial V}{\partial \Delta x}*u\left(A_{pj},\Delta x\right),$$

where $u_{\Delta x}$ is the uncertainty due to the x resolution (Δ_x) of the frame grabber, and $u\left(A_{pj},\Delta x\right)=0$ and from eq.(2):

$$u_{A_{pj}}^2 = \sum_i \left(\frac{\partial A_{pj}}{\partial y_i'}\right)^2 * u_{y_i'}^2 + \left(\frac{\partial A_{pj}}{\partial z_i'}\right)^2 * u_{z_i'}^2 + 2*\frac{\partial A_{pj}}{\partial y_i'}*\frac{\partial A_{pj}}{\partial z_i'}*u\left(y_i',z_i'\right) = \quad (5)$$

$$= \left(\frac{1}{2}\right)^2 * \sum_i \left(y_{i+1}'-y_{i-1}'\right)^2 * u_{z_i'}^2 + \left(z_{i-1}'-z_{i+1}'\right)^2 * u_{y_i'}^2 + 2*\left(y_{i+1}'-y_{i-1}'\right)*\left(z_{i-1}'-z_{i+1}'\right)*u\left(y_i',z_i'\right)$$

Continuing to solve backwards the relationships, the uncertainties of the 3-D coordinates can be obtained by applying the law to eqs. (1):

$$u_{y_i'}^2 = \left(\frac{\partial y_i'}{\partial \theta_y}\right)^2 * u_{\theta_y}^2 + \left(\frac{\partial y_i'}{\partial \theta_x}\right)^2 * u_{\theta_x}^2 + \left(\frac{\partial y_i'}{\partial y_i^*}\right)^2 * u_{y_i^*}^2 + \left(\frac{\partial y_i'}{\partial x_i^*}\right)^2 * u_{x_i^*}^2 + \left(\frac{\partial y_i'}{\partial \bar{x}}\right)^2 * u_{\bar{x}}^2$$

$$= \left(-\sin\theta_y*\cos\theta_x*y_i^*\right)^2 * u_{\theta_y}^2 + \left[\cos\theta_x*\left(\bar{x}^*-x_i^*\right)-\cos\theta_y*\sin\theta_x*y_i^*\right]^2 * u_{\theta_x}^2 +$$

$$+\left(-\sin\theta_x\right)^2 * u_{x_i^*}^2 + \left(\sin\theta_x\right)^2 * u_{\bar{x}^*}^2 + \left(\cos\theta_x*\cos\theta_y\right)^2 * u_{y_i^*}^2$$

$$u_{z_i'}^2 = \left(\frac{\partial z_i'}{\partial \theta_y}\right)^2 * u_{\theta_y}^2 + \left(\frac{\partial z_i'}{\partial y_i^*}\right)^2 * u_{y_i^*}^2 + \left(\frac{\partial z_i'}{\partial d_z}\right)^2 * u_{d_z}^2$$

$$= \left(\cos\theta_y*y_i^*\right)^2 * u_{\theta_y}^2 + \sin^2\theta_y * u_{y_i^*}^2 + u_{d_z}^2$$

and their covariance term is:

$$u\left(y_i',z_i'\right) = \frac{\partial y_i'}{\partial y_i^*}*\frac{\partial z_i'}{\partial y_i^*}*u_{y_i^*}^2 + \frac{\partial y_i'}{\partial \theta_y}*\frac{\partial z_i'}{\partial \theta_y}*u_{\theta_y}^2 =$$

$$= \cos\theta_y*\cos\theta_x*\sin\theta_y*u_{y_i^*}^2 - \sin\theta_y*\cos\theta_x*\cos\theta_y*y_i^{*2}*u_{\theta_y}^2$$

The terms $u_{x_i^*}^2, u_{y_i^*}^2$ are the uncertainties of the 2-D image coordinates, $u_{\theta_x}^2, u_{\theta_y}^2$ are the uncertainties of the results of the angle measurement system, and $u_{d_x}^2, u_{d_z}^2$ are the uncertainties of the measurements of the shift of the probe. On the other hand, the uncertainty of the stenosis measurement can be determined applying the uncertainty propagation law to eq. (4):

$$u_{stenosi} = \sqrt{\left(\frac{1}{A_{pj\max}}\right)^2 * u_{A_{pj\min}}^2 + \left(\frac{A_{pj\min}}{A_{pj\max}^2}\right)^2 * u_{A_{pj\max}}^2}$$

Tab. I Results of the software metrological characterization

Dist. (mm)	Plaque type	Imposed plaque volume (mm^3)	Estimated plaque volume (mm^3)	error %	$\dot{u}_V$	Imposed stenosys	Estimated stenosys	error %	$\dot{u}_{stenosi}$
12	0	427.3591	430.7079	-0.0078	0.0015	0.5168	0.5170	-0.0004	0.2767
12	A	316.0681	319.7210	-0.0116	0.0171	0.6473	0.6502	-0.0045	0.2282
12	B	224.3322	229.1845	-0.0216	0.0201	0.7746	0.7771	-0.0031	0.1992
12	C	119.3452	121.1094	-0.0148	0.0246	0.8842	0.8833	0.0009	0.2067
15	0	394.0042	396.5680	-0.0065	0.0162	0.5197	0.5199	-0.0005	0.3277
15	A	291.2222	293.3349	-0.0073	0.0180	0.6462	0.6459	0.0005	0.2774
15	B	205.8468	208.2131	-0.0115	0.0212	0.7694	0.7715	-0.0027	0.2417
15	C	124.1292	125.9118	-0.0144	0.0261	0.8830	0.8846	-0.0018	0.1849
18	0	358.4455	362.5473	-0.0114	0.0177	0.5232	0.5208	0.0046	0.4001
18	A	264.8660	268.4824	-0.0137	0.0196	0.6465	0.6459	0.0009	0.3390
18	B	186.7877	189.8595	-0.0164	0.0231	0.7662	0.7665	-0.0003	0.3016
18	C	127.1944	129.8594	-0.0210	0.0295	0.8824	0.8844	-0.0023	0.2671
20	0	333.5509	335.9904	-0.0073	0.0193	0.5260	0.5237	0.0043	0.4526
20	A	246.4068	248.5051	-0.0085	0.0217	0.6475	0.6465	0.0015	0.3904
20	B	173.5689	175.4822	-0.0110	0.0261	0.7651	0.7651	0.0001	0.3478
20	C	117.5873	119.2200	-0.0139	0.0343	0.8790	0.8784	0.0006	0.3146

where the two uncertainties of $A_{pj\ \min}$ and $A_{pj\ \max}$ can be determined as in eqs. (5). Numerical tests have been carried out in order to compare the results given by the proposed algorithm with "reference" parameters of models of carotids designed with a CAD software. The results of tests done on different types of carotid ("0" to "C" types are distinguished by decreasing values of stenosis, from 48% down to 22%, in 2nd column) and with different skin-carotid distances (from 12 to 20 mm, in 1st column) are shown in Tab. I. Systematic errors and relative uncertainties are reported for the different cases.

IV. The experimental results

For sake of brevity only one clinical test is here reported, executed on a 32 years old patient with no evident plaque pathologies, in the "Villa dei Fiori" (Naples, Italy) hospital. The provided add-on package has been mounted on a carotid probe embedded with the ATL HDI 5000 ultrasonic machine. Figure 4.a) shows the realized software during the 2-D edge segmentation procedure, while Figure 4.b) shows a particular of the rendered 3-D image. The measured volume value is 157 mm^3 with a standard

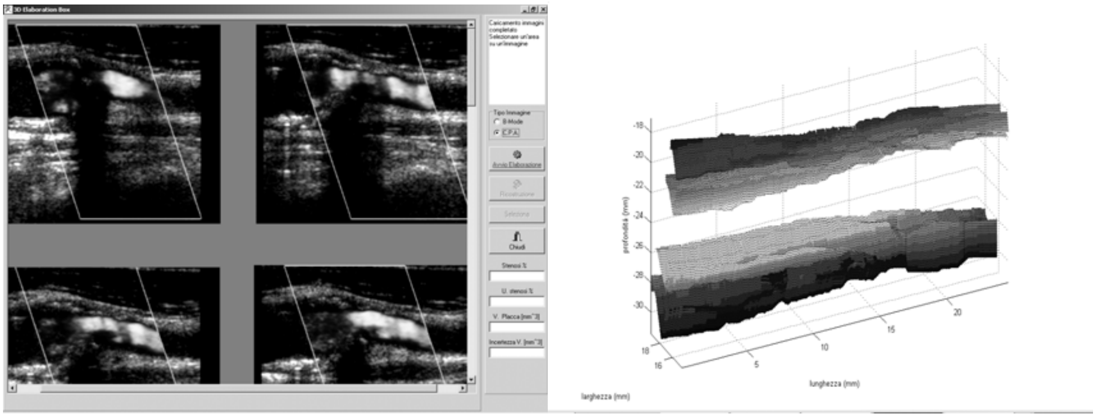

Figure 4: a) The segmentation software; b) a rendered 3-D carotid reconstruction

uncertainty of 16 mm^3, while the stenosis is 0.75 with a relative uncertainty of 0.25.

CONCLUSION

An add-on system for the 3-D reconstruction of carotid arteries to be applied to ordinary ultrasound devices has been realized and characterized. During clinical examinations, the relevant differences with respect to a normal apparatus have been: i) an increase of the weight of the probe; ii) the need of a manual start for the acquisition.

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