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Evaluation of the Influence of Probe Pressure on the B-mode Ultrasound Measurement of Arterial Diameter

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Abstract— **This paper deals with the influence of the echographic probe pressure on the measurement of arterial diameter[1]. It is common sense that arterial diameter measurements are affected by the pressure exerted on the B-mode ultrasound probe but we found no report in the current literature analyzing and quantifying this effect. We demonstrated this influence by the analysis of arterial images obtained with a conventional B-mode ultrasound system while monitoring the pressure exerted on the probe with a force sensor. Our preliminary results demonstrated that continuous measurement was feasible and could be used to assess the relationship between the blood pressure and the arterial diameter. This may prove useful for the quantitative evaluation and the follow-up of patients with cardio-vascular diseases.**

I. Introduction

B-mode ultrasound is a non-invasive measurement technique widely usedfor the evaluation of cardiovascular disorders. Reliable measurements are required, especially for prospective, quantitative studies and for the follow-up of patients with chronic cardiovascular diseases. Peterson's elastic modulus, arterial distensibility, compliance of the arterial wall are determined by measuring diameter and endothelium thickness (Figure 1) on B-mode ultrasound images, correlatively with systolic and diastolic blood pressure.

Measurements on internal arterial diameter and intima media thickness are obtained by the analysis of B-mode images of carotid, femoral, brachial, radial or ulnar arteries. On animals, Graf et al. [9] have computed the arterial diameter and the endothelium thickness using US B mode imaging and IOTEC system (distributed by IODP, Paris, France) with a video frame rate (25Hz), a 640 x 480 image size, 5 to 10 cardiac cycles and a 5 MHz US transducer for a measurement direct on the artery wall. Haluskea et al. [11] have compared manual measurement of arterial diameter (performed independently by two sonographers with a 12MHz US probe) analysed with HDILab software (Philips Advanced Technology Laboratories, Bothell, USA). Beux et al. [4] used a frame grabber at video rate with 768x576 images size and ECG synchronization. They used a mechanical system to maintain the probe in position during the measurement with a 5 MHz US external transducer. Selzer et al. [15] acquired ultrasound images with a 7.5 MHz US external transducer at video rate and used the PROSOUND system for the diameter and thickness measurement.

Globally, diameter measurements reported in the literature range from 0.1 mm up to 4 mm and reach a 0.01 mm precision in prospective clinical studies generally validated using the Bland and Altman method [5].

Image analysis is mostly performed using classical gradient methods which appear to yield the best results [10]: a Region Of Interest (ROI) is analyzed during consecutive cardiac

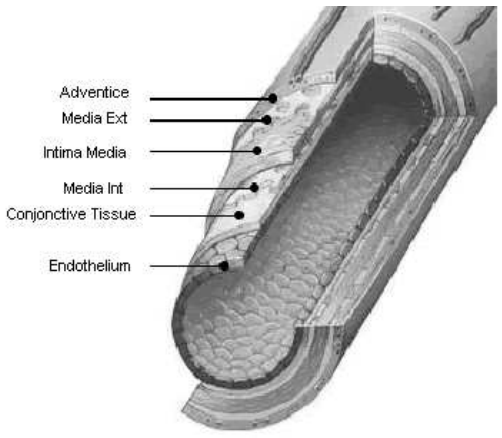

Fig. 1. Different parts of arterial wall

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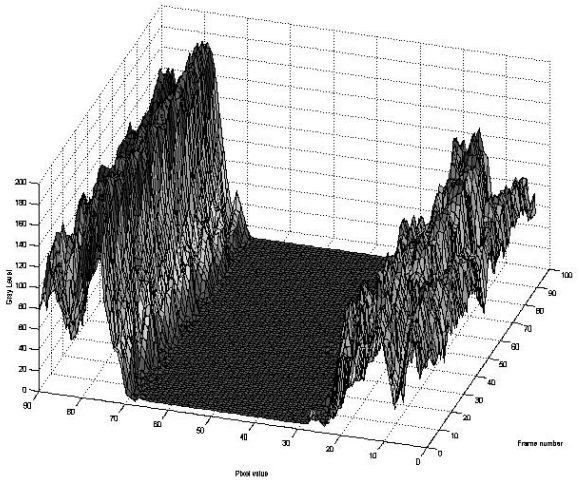


Fig. 2. Line profile evolution due to blood pressure variation.

cycles, and measurement results are mean values of the ROI along the cardiac cycles.

Although none of these studies took into account the pressure exerted by the ultrasound probe during image acquisition, there is a general agreement that this pressure may alter measurements of arterial diameter and diameter changes during the cardiac cycle.

Figure 2 shows the evolution, in a normal subject, of the carotid diameter being only due to a variation in blood pressure (gray levels evolution among a line perpendicular to the arterial walls observed during 4 seconds: 90 frames).

The biomechanical characteristics of the arterial wall are usually estimated from the systolic diameter increase related to the pulse pressure (difference between systolic and diastolic blood pressure). The main parameters commonly used for the diagnosis and evaluation of cardiovascular diseases are described hereunder:

- Peterson's elastic modulus:

$$E_P = \frac{PP}{DD}$$

- Beta stiffness index:

$$\beta = \frac{ln(\frac{P_S}{P_D})}{DD}$$

- Young's modulus:

$$Y_m = E_P \times 0.5 \times \frac{Dmin}{IMT_{D_{min}}}$$

- Arterial distensibility:

$$DIS = 100\% \times \frac{[\frac{(D^2_{max} - D^2_{min})}{D^2_{min}}]}{PP}$$

- Compliance:

$$COMPL = \frac{(D^2_{max} - D^2_{min})}{PP}$$

with:

- D_{min} and D_{max} : minimum and maximum artery diameters,
- $DD = \frac{(D_{max} - D_{min})}{D_{min}}$
- $IMT_{D_{min}}$: intima-media thickness of the arterial far wall at the minimum arterial diameter,
- pulse pressure: $PP = P_S - P_D$ where P_S is the systolic blood pressure and P_D is the diastolic blood pressure.

All these parameters are arterial diameter dependent. Only diastolic and systolic blood pressures are generally used for their calculation, and are usually measured far away from the site of B-mode ultrasound measurement of diameter, (10 up to 20 cm), using a sphygmomanometer device. Therefore, a phase delay should be taken into account between the ultrasound image and the blood pressure measurement.

Force feedback solutions were available in systems like OTELO [7], [8] HIPPOCRATE [14]. In [2], [1] the teleoperation of an ultrasonic probe was done to control pressure on the patient and in [12], to allow guided needle insertion. In [3], visual servoing was used to control ultrasound probe positionning but force feedback was not taken into account. In [13], a similar approach was presented and experimentation done on one animal with force feedback measuring constraints at the insertion point when achieving a minimaly invasive surgery task.

We propose the clinical evaluation of force effect on the arterial diameter measurement using a force sensor (the intima-media thickness of the far wall used for the calculation of Young's modulus will be evaluated later, as dedicated softwares are widely available on the market for this measurement). Our aim is to demonstrate the influence of the pressure force exerted on the ultrasound probe on the measurements of diameter, and measure in a continuous way the force feedback resulting from blood pressure during the cardiac cycle, at the ultrasound probe measurement site.

This paper is presented as follows: in the next section we present measurements and methods then we describe experimental results before concluding.

II. Measurements and methods

A. Image Analysis

We performed preliminary measurements on healthy people and the evolution of the artery diameter correlated to the position, orientation and force of the ultrasonic probe (Siemens Cypress system with a 7L3 probe 5.4 to 6.6 MHz). The B-mode ultrasound image was transmitted to an acquisition board (Matrox Meteor II) during 20 consecutive cardiac cycles. The sequence was stored and the image analysis was made off line. The algorithm is presented in Figure 3. The operator defines a ROI along the artery in the first image. Then, a highpass filter is applied to the image, then finding an adaptative hysteresis threshold and applying Canny gradient approach[6] (see Figure 4) allowing us to identify the arterial near and far walls. A first level of artery wall approximation is a first order polynomial but can be extended to a second order like

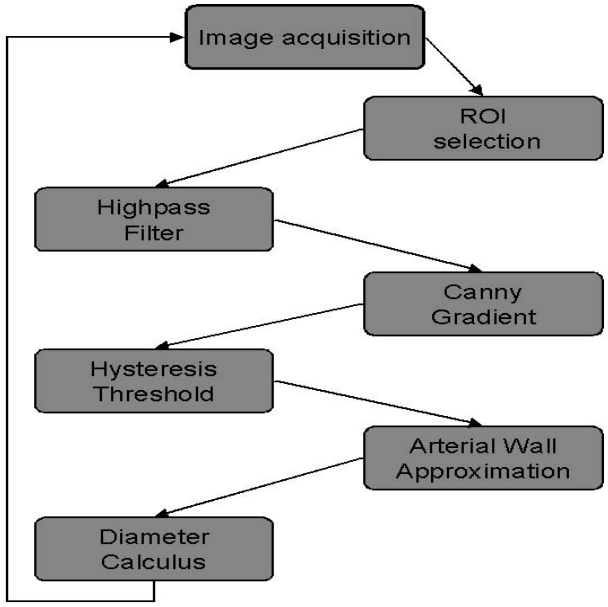
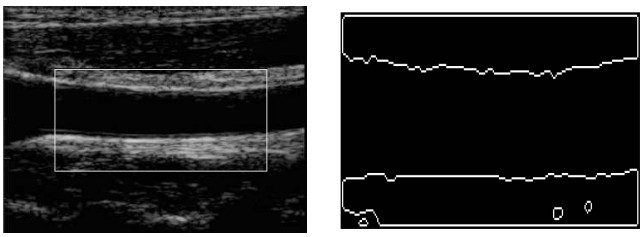


Fig. 3. Gradient approach to extract Intima Media Lumen diameter



(a) (b)

Fig. 4. Image analysis : ROI selection (a), Arterial wall extraction results(b)

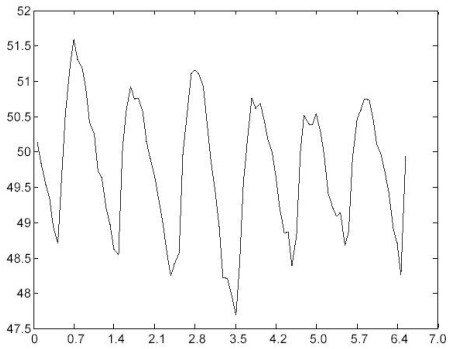


Fig. 5. Arterial diameter in pixels during 7 sec. (6 cycles)

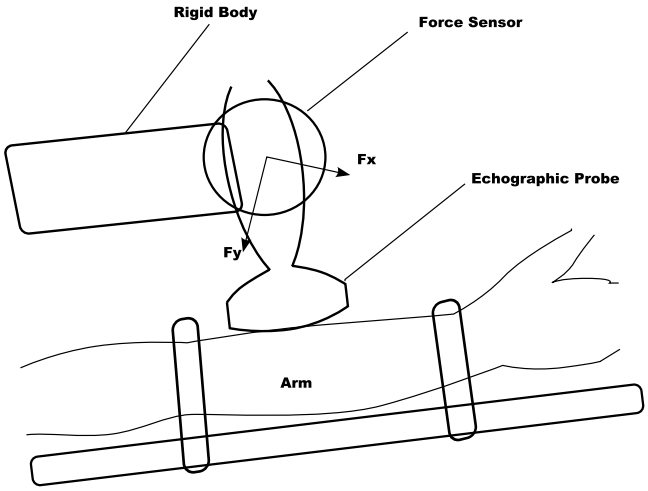


Fig. 6. Experimental setup

[4], then the diameter is evaluated using the mean distance between the near and far walls with a sub pixel precision (0.1 pix). Results were obtained for all the consecutive images of the sequence. Figure 5 shows the diameter changes in pixels during 6 cardiac cycles obtained after image analysis (calibration has yet to be done to obtain measurements in mm). The arterial diameter was of about 50 pixels and its variation of about 4 pixels between systolic and diastolic moments. A small average variation could be observed due to breathing.

B. Probe instrumentation

The echographic probe was fastened to the force sensor (Mini 40 from ATI Industrial Automation) in the (F_x, F_y) measurement plane (F_z is negligible). This unit was attached on a rigid body, then it was possible to maintain a constant force on the patient's arm skin (see Figure 6).

The arm is horizontally fixed (with large elastic bands to avoid artery compression) on a plane unit with a vertical mechanical movement (perpendicular to the plane). Then, force variation is applied, moving the vertical plane with the arm in contact with the echographic probe.
The experiment proceeded according to the following diagram:

1) US image acquisition,
2) 3D force measurements using a National Instruments Data Acquisition Board,
3) US image acquisition,
4) diameter estimation from (1) and (3) images (mean),
5) storing of diameter and 3D force measurements.

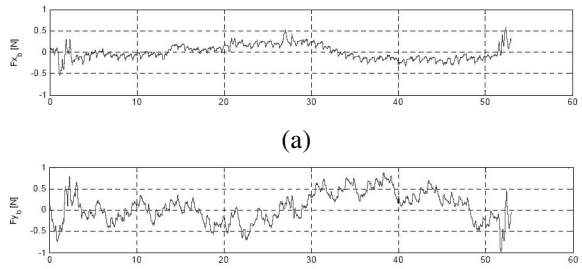
Analysis was then performed off line.

III. RESULTS

So far experiments have been made with only 4 patients but will be extended to a set of 20 patients. We have first verified that the cardiac cycle was correctly detected.

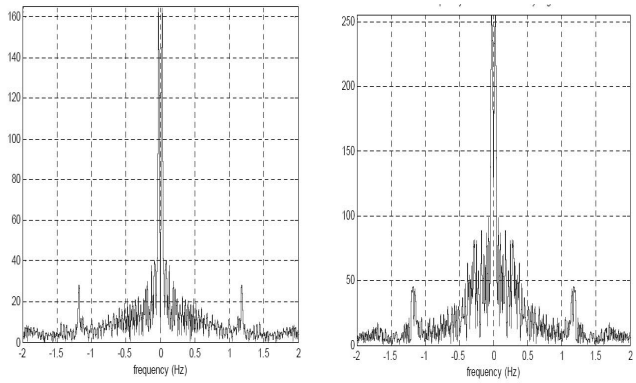
Figure 7 presents the force measurements Fx and F_y during 50 sec with a constant force on the artery (small variation are due to breathing and unperceivable movements of the patient. Figure 8 presents the corresponding frequential information (FFT). A peak appears at 1.2 Hz corresponding to cardiac pulse rate.

We have then analyzed the influence of the probe pressure on the diameter measurement. Figure 9 presents both the force (a) and the diameter evolution (b) with a step of force ranging



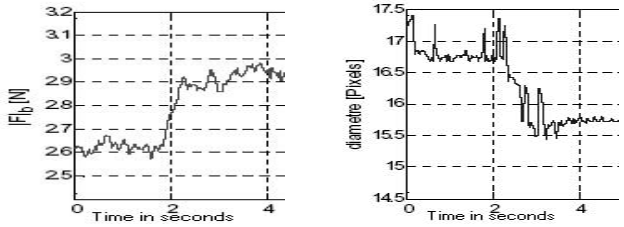
(a) (b)

Fig. 7. F_x (a) and F_y (b) force components during 50 sec.



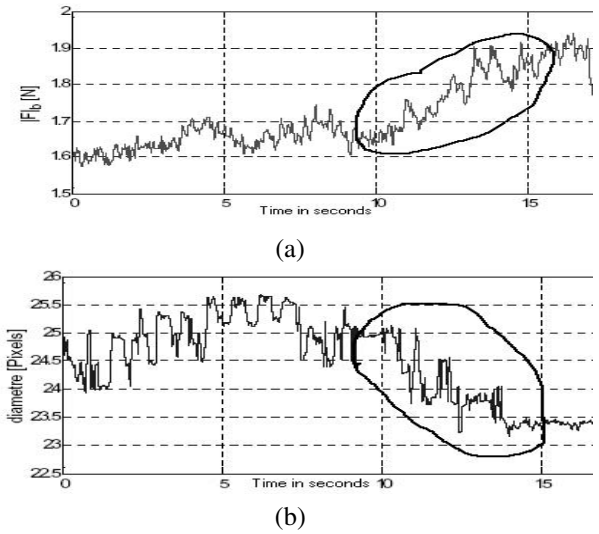
(a) (b)

Fig. 8. Frequency spectrum F_x (a) and F_y (b)



(a) (b)

Fig. 9. Step force variation(a), Corresponding Arterial diameter(b).



(a) (b)

Fig. 10. Linear force variation(a), Corresponding Arterial diameter(b).

from 2.6 N up to 3.1 N. We could observe a diameter variation from 16.8 up to 15.5 pixels.

Figure 10 presents both the force (a) and the diameter evolution (b) with a force varying linearly from 1.6 N up to 1.9 N. We could observe a diameter variation from 25 up to 23 pixels. The arterial response was clear and the influence of pressure is obvious, even when considering the small pressure variations due to the operator.

Finally, we have measured the continuous force variation due to blood pressure variation at the same time and precisely where the echographic probe was measuring.

As the axes of the sensor were F_x and F_y, we presented in Figure 11 the magnitude of the measured force and the corresponding diameter variation. This force had a mean value of about 3 N and varied from 2.5 up to 3.5 N. Arterial diameter had a mean value of about 26.5 pixels and varied from 26 up to 28 pixels.

The important result was that we could observe from 0 to 20 sec a correlation between force and diameter.

Finally, Table I presents the results obtained for the four subjects. The force magnitude was measured at two moments and the corresponding diameter was reported.

Table II shows the systolic - diastolic diameter and force changes calculated from Table I and and demonstrates that theses changes were inversely proportional. The force - diameter correlation seems to be linear and could be estimated from these results but its assessment will need a larger set of experimental data and calibration of ultrasound images.

IV. Conclusion

Our first experimental results demonstrate the feasibility of correlative, simultaneous measurement of the arterial diameter and the force exerted by the probe on the subject's skin. We could thus assess the influence of probe pressure on the arterial diameter. In a second part, we have measured a force proportional to blood pressure. Systematic calibration and validation must now be performed in animal experiments, before this approach can be used for clinical studies.

Our system may help improving the accuracy of the measurement of arterial diameter and diameter changes for the evaluation of arterial wall biomechanical parameters, by taking into account the pressure exerted on the probe by the operator. It may also allow overcoming the problem resulting from the phase shift occurring when blood pressure is measured far away from the site of ultrasound imaging.

It may also provide additional information by allowing to explore remote parts of the pressure - diameter relationship of the artery. This may prove valuable for experimental as well as clinical studies requiring accurate and detailed in vivo measurement of the arterial wall biomechanical parameters.

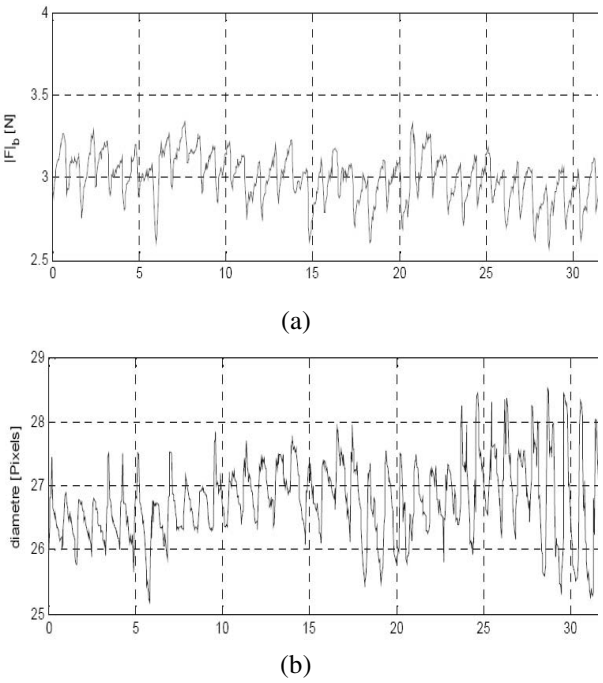


Fig. 11. Continuous arterial pressure measurement during 30 sec.

Patient	Acquisition	Force (N)	Arterial Diam. (pix)
1	1	2.06	18.15
	2	2.94	16.79
2	1	1.82	18.78
	2	2.93	17.21
3	1	0.52	20.12
	2	1.49	18.73
4	1	1.72	24.24
	2	3.00	19.12

TABLE I

FORCE APPLICATION AND CORRESPONDING ARTERIAL DIAMETER

Patient	Δ Force (N)	Δ Arterial Diam. (pix)
1	+0.88	-1.36
2	+1.11	-1.57
3	+0.97	-1.39
4	+1.28	-5.12

TABLE II

FORCE VARIATION AND CORRESPONDING ARTERIAL DIAMETER VARIATION

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