Evaluation of lower limb vein biomechanical properties and the effects of compression stockings, with an instrumented ultrasound probe

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***Abstract*— We present a new approach for the evaluation of the biomechanical properties of lower limb veins based on the simultaneous measurements of the vein cross-sectional area with B-mode ultrasound imaging and of the force exerted on the skin by the ultrasound probe. Ongoing clinical trials allowed us to identify a behavioral model of lower limb veins without and with compression stockings.**

I. Introduction

Chronic venous insufficiency (CVI) concerns mainly lower-limb veins. The venous system is composed of superficial veins located under the skin into the adipose tissue, of deep veins within or around muscles, and of perforating veins linking them. Blood pressure in the veins depends on the body position. When lying down, the pressure in peripheral veins is homogeneous, and is close to 10 mmHg at the ankle. When standing up, there is a large difference in pressure between the heart (~0 mmHg) and the lower-limb veins (~100 mmHg at the ankle). When walking or exercising, contractions of the leg muscles expel the blood contained in deep veins and propel it toward the right atrium, but this 'venous pumping function' depends on the competence of valves distributed along the veins. The biomechanical features of blood vessels are defined by their dimensions (area, wall thickness) and by the elastic properties of their wall. The venous wall is thinner than that of arteries (Fig. 1), with a greater proportion of collagen fibers and a lower proportion of smooth muscle cells. This feature gives the venous wall greater flexibility, allowing veins to change in shape depending on blood pressure and contractions of surrounding muscles.

The venous wall elasticity is defined by Young's modulus E (N/m²):

$$E = \frac{\sigma}{\varepsilon}$$

This parameter is extracted from Hooke's law, which states that the stress σ (N/m²) is proportional to the strain ε (unit-less).

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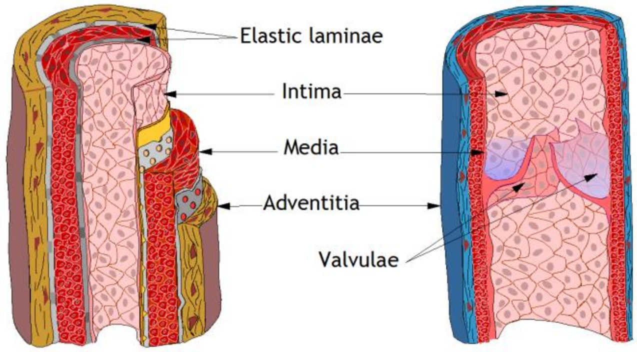

Figure 1. Internal structure of an artery and a vein

The capability of a vein to change its section size and shape as a function of blood pressure is determined by its wall distensibility D (m²/N):

$$D = \frac{\Delta A}{A_0 \times \Delta P} = \frac{1}{E}$$

where the relative variation of area $\Delta A/A_0$ (unit-less) is proportional to the variation of pressure ΔP (N/m²). $\Delta A = A - A_0$ and $\Delta P = P - P_0$, with A and P as current values and A_0 and P_0 initial values. Distensibility is the inverse of the elasticity.

In patients with CVI, lower limb veins are affected by wall deficiency, which can lead to valve dysfunction and incompetence. As a result, vein diameters increase, which in turn raise wall strain, and blood reflux may develop, engaging a vicious circle. To prevent and treat CVI, physicians prescribe wearing medicals compression stockings. These orthotic devices reduce venous transmural pressure and limit orthostatic venous distension. Conventional compression stockings are degressive, applying on the skin a higher pressure at the ankle than the calf, and are classified in a range of increasing pressure (European classification [1]: Class I to IV, from <25 to >45 mmHg at the ankle by steps of 10 mmHg).

Several authors studied arterial biomechanics [2][3][4][5], but venous wall biomechanics remains imperfectly known, particularly regarding the physical and physiological mechanisms influenced by medical compression. Different types of compression stockings for different stages of CVI have been tested [6],[7],[8],[9], but most studies were limited to measuring leg volume before and after wearing the stockings for some time, in relation to the pressure they exerted on the skin. Wang et al [10] modeled the effect of

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compression on lower-limb tissues using magnetic resonance imaging (MRI), but, besides its many limitation, constraints (dedicated room, no ferromagnetic material), and cost, MRI cannot be used with the patient standing up or exercising.

B-mode and Doppler ultrasonography is widely used for vein examination, as it is easy to use, non-invasive and relatively inexpensive. We report here a new method of clinical evaluation of the effects of compression stockings on the wall geometry and viscoelastic characteristics of superficial and deep lower limb veins. We simultaneously measured the force exerted by the ultrasound probe on the patient's skin and the cross-sectional area of the veins. In the first part of this paper, we present our hardware and software developments. In the second part, we report our preliminary clinical results showing the vein area / pressure relationship. To conclude, we discuss the potential applications of this approach.

II. MEASUREMENTS AND METHODS

A. Hardware setup

The acquisition system was composed of a computer equipped with a video frame grabber (Picolo, Euresy, Liege, Belgium) and a data acquisition system (MP150, Biopac Systems, Goleta, USA). The video capture board was connected to the S-video output of a Logiq E ultrasound scanner (GE Healthcare, Milwaukee, USA). A 8-12 MHz linear ultrasound probe (12L-RS, GE Healthcare) was instrumented with a linear force sensor connected to a signal amplifier/conditioner (XFTC300 and ARD154, Measurement Specialties, Hampton, USA) whose output was connected to the data acquisition system (Fig. 2). The digital output of the frame grabber was connected to the data acquisition system for synchronizing the video data stream and the force measurement.

B. Data acquisition

A laboratory-written program using the frame grabber's drivers allowed real-time recording of the B-mode images coming from the ultrasound scanner (25 frame/s), and sending a synchronization signal to the data acquisition and processing software (AcqKnowledge, Biopac Systems, Goleta, USA), with a 100 Hz sampling rate. In this way, 4 force values were averaged for each ultrasound image.

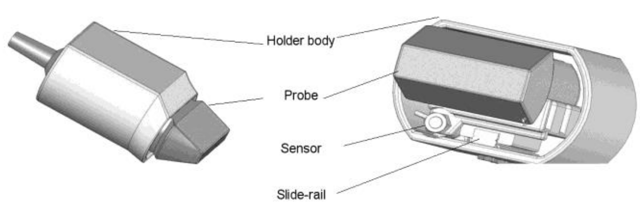

Figure 2. Probe holder with linear force sensor

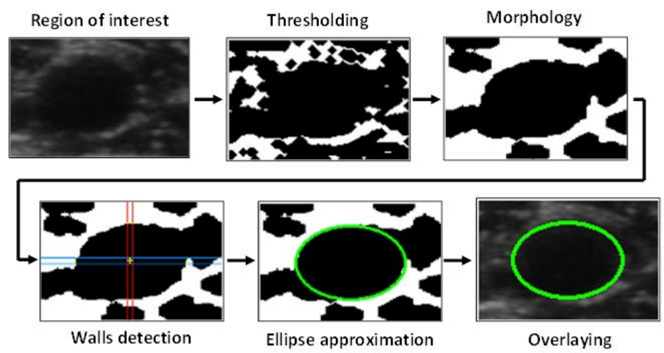

Figure 3. Successive stages of image processing

C. Image processing software

The vein area was extracted from the cross-sectional images of the vein by post-processing software as follows:

- Selection of the saved image sequence
- Image calibration for pixel to mm conversion
- Drawing of the region of interest (ROI) enclosing the observed vein
- Adjustment of the threshold on the gray scale for image binarization
- Morphology adjustment for edge-smoothing
- Detection of the venous wall along the horizontal (X) and vertical (Y) axes
- Computation of the X and Y lengths for ellipse approximation
- Overlay of the calculated ellipse on the initial image (Fig. 3)

The validity of automatic ellipse approximation was visually assessed by clinicians. The ROI center was calculated for each approximated ellipse, allowing to follow the movements of the vein along the sequence. At the end of processing, the program delivered a data file with the values of the vein area for each image.

III. CLINICAL TESTS AND PRELIMINARY RESULTS

A. Measurement protocol

Clinical tests have been conducted on volunteers in collaboration with the angiology department of the Montpellier University Hospital. The pressure / area function of superficial and deep leg veins was assessed in the subjects lying supine or standing up, without and with different types of stockings. For each compression and posture, a physician optimized image settings for the examination of a superficial and a deep vein at mid-calf, and slowly increased the pressure applied on the ultrasound probe until the examined vein collapsed, then slowly decreased the pressure to the minimum keeping the probe at skin contact. The result was a graph on which force and vein area showed opposite phase curves (Fig. 4).

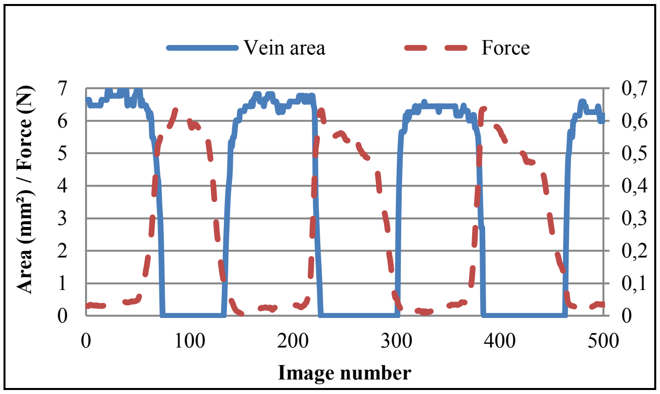

Figure 4. Variations of vein area according to the force applied on the ultrasound probe.

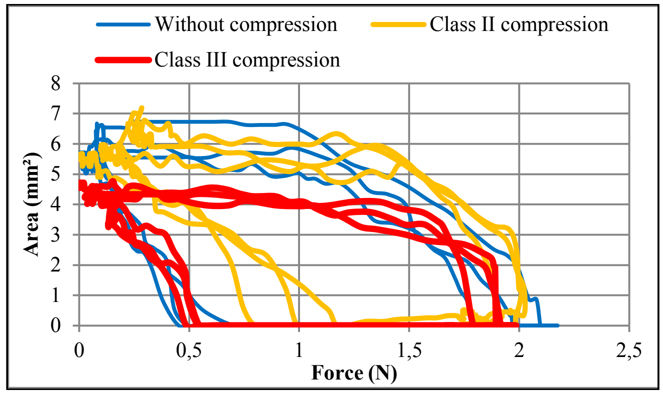

Figure 6. Variations of superficial vein area as a function of force in a normal subject standing up.

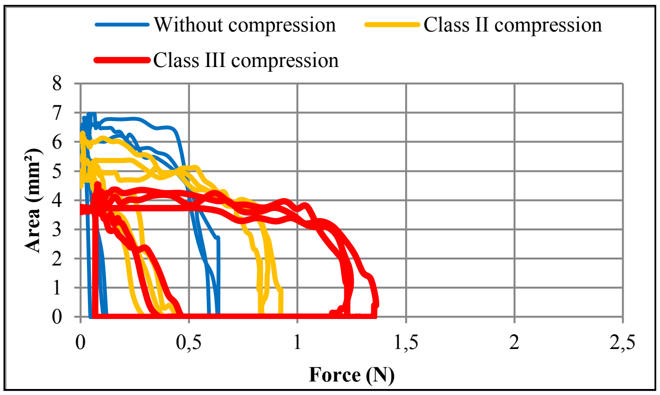

Figure 5. Variations of superficial vein area as a function of force in a normal subject lying supine.

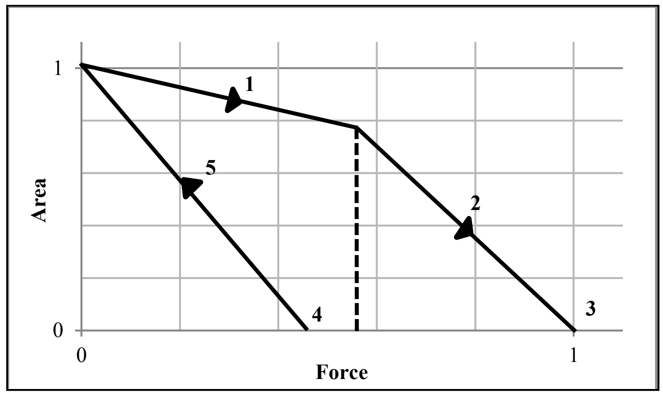

Figure 7. Model of the biomechanical behavior of the lower limb veins.

B. *Relationship between vein area and force*

Drawing the vein area against the force applied on the probe produced a typical hysteresis curve showing the behavior of the vein according to the type of compression. Fig. 5 and Fig. 6 show an example of 3 consecutive cycles of a superficial vein area *vs*. force relationship in a healthy subject without and with compressive stockings, in the supine position (Fig. 5) and when standing up (Fig. 6).

These graphs show that the higher the compression class, the smaller the initial area of the vein and the higher the force needed to collapse it. On this example on the subject lying supine, the initial area of the vein was 6.5 mm² without compression stockings, decreased to 5 mm² with Class II, and to 4 mm² with Class III compression stockings. A 0.6 N force was needed to collapse the vein when the subject wore non-compressive stockings, 0.8 N with Class II, and 1.25 N with Class III compression stockings (Fig. 5).

When the subject was standing up, the force needed to collapse the vein was 1.9 N with non-compressive stocking, 2 N with Class II , and 1.8 N with Class III compression stockings. The vein opened again when force decreased to 0.5 N with non-compressive stockings, 1 N with Class II, and 0.5 N with Class III compression stockings (Fig. 6).

Normalizing these curves enabled us to describe the typical lower limb vein behavior during increasing and decreasing force cycles. The resulting model comprised five remarkable parts (Fig. 7):

1. Compression of the surrounding leg tissues (the ultrasound probe - vein distance decreases while the vein area remains unchanged)
2. Vein closure, with a sharper downslope
3. Vein *collapsus* (area=0)
4. Vein opening
5. Slope of vein, then of tissue, relaxation

The slopes of parts 1 and 2 may be related to the Young's modulus E, and the slope of part 5 may represent distensibility D.

IV. Discussion and Conclusion

The system we developed allowed us measuring the force applied on the skin by the ultrasound probe and the resulting change in area of the observed veins. Our preliminary results illustrate the lower limb vein biomechanics, and our methodology appears particularly well suited for the evaluation of the mechanisms and effects of compression stockings. The assessment of reproducibility would be necessary for longitudinal studies. Although they would need to be confirmed and detailed on a large series of normal subjects and patients with CVI, these results emphasize the complexity of the mechanisms involved by compression therapy, since wearing compression stockings appears to increase and homogenize pressure within leg tissues, so that the directional force needed to collapse the vein would be greater. Therefore, our method would be able to provide new and relevant insights into the vein biomechanics and allow for fully documented, personalized prescription of compression stockings in patients with CVI.

We are now using this approach on normal volunteers and patients with CVI, together with invasive intravenous and muscular pressure measurement, in order to fully validate the biomechanical model of superficial and deep lower limb veins under compression stockings. Additional technological developments will be pursued for a rapid, comprehensive, and reproducible evaluation of venous pathophysiology.

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