Calculation of Human Femoral Vein Wall Parameters *In Vivo* From Clinical Data for Specific Patient

Blas Herrera[1]
e-mail: blas.herrera@urv.net

Gerard Fortuny

Department of Computer Engineering and Mathematics,
Universitat Rovira i Virgili,
Avinguda Països Catalans 26,
43007, Tarragona, Spain

Francesc Marimón
Department of Internal Medicine,
Sant Joan University Hospital,
Universitat Rovira i Virgili,
Avinguda Països Catalans 26,
43007, Tarragona, Spain

A common problem in the elaboration of biomechanical models is determining the properties and characteristics (measures) of the physical behavior of in vivo tissues in the human body. Correct estimates must be made of the tissue's physical properties and its surroundings. We suggest a method to compute the constitutive modeling of venous tissue, for every specific patient, from clinically registered ultrasounds images. The vein is modeled as a hyperelastic, incompressible, thin-walled cylinder and the membrane stresses are computed using strain energy. The approach is based on a strain-energy function suggested by Holzapfel capturing the characteristic nonlinear anisotropic responses of femoral veins with its collagen fibers. [DOI: 10.1115/1.4007948]

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1 Introduction

4.5% of the population risk suffering from venous thromboembolism disease with an approximate mortality rate of 11% [1,2].

It is well known that deep vein thrombosis (DVT) originates in the deep venous region, predominantly in the lower extremities and mainly in the femoral vein near the inguinal region. DVT takes place when blood clots form inside the femoral vein and impede circulation. Sometimes these clots break and travel through the venous system until they are deposited in some other place in the cardiovascular system [3]. This detachment can generate serious lesions in the human body, in particular lung thromboembolism. If we want to study the conditions that cause these clots to break, and if we want to generate a biomechanical model that will allow us to study and to reproduce this process, then we need specific knowledge of the physical properties of the tissues involved.

Cardiovascular tissue has been modeled by various authors [4,5], and there are even precise biomechanical models of the arterial system [6]. However, as far as we know, before our study [7] about the Young's modulus and the Poisson's coefficient, there has been no specific research into live venous tissue and its surroundings in humans. These physical parameters have already been calculated by other authors [8,9], but only for non live venous tissue.

The physiology of the cardiovascular system in general, and in particular of the femoral vein, has already been sufficiently documented [Williams (1995)]. Our main aim is to make a detailed biomechanical model of the femoral vein (where most DVT occurs) that can be adapted to specific patients.

Specifically, in this work, we suggest a method to compute the constitutive modeling of venous tissue for every specific patient from clinically registered ultrasounds images. Works like Holzapfel et al. [10] and Stålhand [11] determine some constitutive modeling of arterial tissue walls. We determine a constitutive modeling of the tissue in the venous walls in their real situation: inside the human body.

2 Method

2.1 Original Data. To carry out the present study we have used images captured by projecting ultrasound in real time, which is a typical method for clinically registering the pressure and radius. The ultrasound probe is linear, it scans at 4.5 MHz, and it uses the MyLab Xview 70 high resolution image projection system (Esaote, Genoa, Italy). This new-generation ultrasound tool eliminates particles while preserving the information needed for diagnosis. We have used a time sequence of 10 s to register in a file of a healthy 40 year-old person.

2.2 Mechanical Model. The mechanical model is described in this section. Two sets of stresses are determined for the mechanical model: equilibrium stresses and constitutively determined stresses. These sets are then used in the parameter determination where the underlying idea is that the vein wall stress computed in two different ways should be equal; the parameters are obtained by a least-square fitting of the vein wall stressed and the stressed computed by enforcing global equilibrium.

The readers of this journal are familiarized with the vein histology: its three distinct layers, the intima, the media, and the adventitia, and its mechanical contributions. Then, here we only want to remark that the media is the middle layer of the vein that consists of a complex three-dimensional network of smooth muscle cells, and elastic and collagen fibrils; and the orientation of and close interconnection between them constitute a continuous fibrous circle, circumferentially oriented [12].

Let the femoral vein be given by a thin-walled, incompressible cylinder of length $l(t)$, and inner and outer radius $r_0(t)$ and $r_1(t)$, respectively at the instant t (we will write only r_0, r_1, and l). Let $\overline{L}$, $\overline{R}_0$, and $\overline{R}_1$ be the length, the inner, and outer radius, respectively, of the referential (diastole) vessel, and are constant. The cylinder is subjected to a hydrostatic pressure $p_r(t)$ at the inner boundary wall while the outer boundary wall is traction free.

Throughout this paper, an overline on a variable means that variable is a constant.

We consider cylindrical coordinates $x = (\theta, z, r)$ for the global full cylinder (θ, z, and r denotes the circumferential, axial, and radial directions, respectively), and we notate $B_x = \left\{\overrightarrow{e}_\theta, \overrightarrow{e}_z, \overrightarrow{e}_r\right\}_x$ as the cylindrical orthonormal base at point x.

The stretches $\lambda_\theta(t)$, $\lambda_z(t)$ at point x of the midwall are (only λ_θ, λ_z)

$$\lambda_z = \frac{l}{\overline{L}}, \quad \lambda_\theta = \frac{r_0 + r_1}{\overline{R}_0 + \overline{R}_1} \tag{1}$$

Now, we denote the vector tangent space of $\mathbb{R}^3$ (oriented Euclidean three-dimensional space) at any point y as $T_y(\mathbb{R}^3)$. Also we denote $\left\{E_\theta = \overrightarrow{e}^*_\theta, E_z = \overrightarrow{e}^*_z, E_r = \overrightarrow{e}^*_r\right\}_y$ as the dual base of B_y for the one-covariant tensor space $T^1_y(\mathbb{R}^3) = \left(T_y(\mathbb{R}^3)\right)^* = \mathcal{L}(T_y(\mathbb{R}^3); \mathbb{R})$ linear applications. The duality $\overrightarrow{v}^*$ is given by the standard scalar Euclidean product $\cdot$ at y; i.e., $\overrightarrow{v}^*(\overrightarrow{w}) = \overrightarrow{v} \cdot \overrightarrow{w}$ $\forall\left\{\overrightarrow{v}, \overrightarrow{w}\right\} \subset T_y(\mathbb{R}^3)$, then

[1]Corresponding author.

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