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A FEM study of aortic hemodynamics in the case of stenosis

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In this report we use a real, two-dimensional geometry of a human abdominal aorta with mild stenosis from images obtained with a MR scanner. Finite element method was used for solving the governing equations for two-dimensional, steady, laminar flow of an incompressible, non-Newtonian fluid in that geometry. The accuracy with which the governing equations were solved using the finite element method was not examined quantitatively in the present study due to a lack of published data. Numerical results were found to be in excellent agreement with Womersley theory and with laser Doppler anemometry velocity data obtained for steady flow in a human model. The distributions of the velocity profile, wall shear stress and pressure along vessel during the cardiac cycle are shown. The results were compared to known values, and peaks were found. The shape of velocity distribution is strongly disturbed by the stenosis, and disturbance is clearly evident whatever instant of the cardiac cycle was considered. The general flow features were accurately predicted based on the finite element flow model, which allows the conclusion that computational fluid dynamics can be used to facilitate improvement of the medical research of cardiovascular physiology.

Key words: finite element method, laminar flow, aortic hemodynamics, stenosis

1. Introduction

The role played by hemodynamics in the development and progression of vascular disease, especially the study of flows through atherosclerotic vessels, is of a great interest due to a substantial health risk [16]. In the aorta, it is observed that

atherosclerotic disease develops first in the abdominal aorta, with the greatest involvement occurring below the celiac artery [14].

In last years, researchers have more frequently used numerical methods trying to understand the mechanics of blood flow in the human vascular system [3]–[5], [7], [8], [10], [15], [18]. They regularly have made use of averaged, idealized, and patient-specific geometric models of blood vessels as a basis for their numerical simulations. However, in order to determine the exact flow conditions in a given individual's vascular system for clinical diagnoses or surgical planning, three-dimensional models, which truthfully represent individual anatomic features and flow conditions, are necessary.

To use numerical simulations for understanding the normal and pathological behaviour of the human vascular system and the altered flow characteristics produced by surgical procedures, it is essential to create the accurate in vivo patient-specific geometric models [1]. It is also important that the geometric model to be smooth enough to produce a finite-element mesh free of local geometric discontinuities, which would create artefacts in the blood flow solution.

The classical numerical hemodynamic models are characterized by a high degree of approximation [21]. Taking into consideration the real case, these models are able to provide only non-confident approaches, pointing from a limited perspective to next main problems:

1. The hypothesis that the blood is Newtonian fluid is not realistic [19].
2. The hypothesis of laminar flow is satisfied in most cases, but some small local turbulences could be observed in different large arteries, the equations of classical models being limitative [9].
3. Blood velocity at the interface with the vessel is one of the most common assumptions.
4. The stationary flow does not represent a real hypothesis because pulsatory phenomena appear almost all over the system [8].
5. The theoretical model of a blood vessel as a uniform cylinder presents limitations, the real shape being irregular, influencing in this way the distributions of velocities and pressures [1].
6. The classical model that does not consider the elasticity of the blood vessels cannot represent the reality [9].

2. Materials and methods

This paper proposes a non-linear realistic model of the blood flow in the arteries. The model is based on a complete set of the Navier–Stokes equations for incompressible Newtonian fluids [9]:

$$\frac{\partial \mathbf{u}}{\partial t} + (\mathbf{u} \cdot \nabla)\mathbf{u} - \nu \Delta \mathbf{u} + \frac{1}{\rho} \nabla p = 0, \tag{1}$$

$$\nabla \cdot \mathbf{u} = 0, \tag{2}$$

where $\mathbf{u} = (u, v, w)^T$ is the velocity vector field, $\nu = \mu / \rho$ represents the kinematics viscosity, □ and □ stand for the constant fluid density and the dynamic viscosity, respectively.

The present FEM model is extremely accurate in terms of geometry, taking into consideration the shape irregularities of the arteries and their tapering as well. The zones where local turbulences are easily induced, influencing in this way the distribution of supplementary stresses in the walls, are also taken into consideration. The 3D patient-specific geometrical model of the abdominal aorta was obtained having as primary materials the series of two-dimensional classical radiographic films of cross-sectional image slices from the University Hospital in Tirgu-Mures. Using scanning devices and image processing software packages the contours of the real system were identified as rigorously as possible [11], [20]. The geometric computational model was obtained as follows:

1. A set of rectangular 2D slice probes through the patient's aorta is precisely positioned in a 3D reference system along a straight line.

2. We use this set of probes to create, by means of Autocad□ 14 (Autodesk), the segmentation of the lumens for each 2D data slice (figure 1).

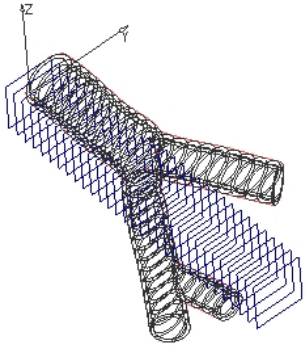

Fig. 1. Human abdominal aorta set of contours defined in a specific reference system

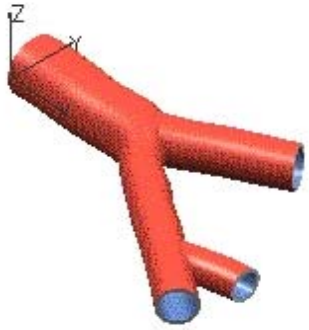

Fig. 2. A 3D geometry of the model considered

3. This set of profiles is fitted with curves of the Bezier type and exported to specialized software for 3D modelling: Catia□ v.4.2.1 (Dassault Systems).

4. A surface is fitted to the set of curves and capped to create a solid model of the vessel (figure 2), which is discretized into a finite element mesh and used in blood flow computations.

Later on, using the IGES standard, the geometry was transferred into the finite element analysis package ANSYS□ v.5.4 (SAS IP, Inc.).

One of the main hypotheses of this study is that for a limited number of variables, after solving the problem, the differences between the 3D model and the 2D model were acceptably small. Due to the complex geometry the used mesh implied a number of over 40,000 elements. The FLUID 141, a FLOTRAN CFD element, was used because it is adequate to model transient or steady-state fluid systems that involve fluid and non-fluid regions [13]. The conservation equations for viscous fluid flow and energy are solved in the fluid region, while only the energy equation is solved in the non-fluid region. The velocities are obtained from the conservation of momentum principle, and the pressure is obtained from the conservation of mass principle [2], [12]. A segregated sequential solver algorithm is used; that is, the matrix system derived from the finite-element discretization of the governing equation for each degree of freedom is solved separately. The flow problem is non-linear and the governing equations are coupled together. The sequential solution of all the governing equations, combined with the update pressure-dependent properties, constitutes a global iteration.

The variation of the blood viscosity as the function of the specific deformation velocity was considered using the Carreau (equation (3)) [6] and the power law models (equation (4)) [22] which enables us to specify variable viscosity dependent on the velocity gradient for such non-Newtonian fluids as blood:

$$\mu = \begin{cases} \mu_0 K D^{n-1}, & D > D_0, \\ \mu_0 K D_0^{n-1}, & D \le D_0. \end{cases} \tag{3}$$

For the power law model, where $\mu_0 = 0.035$ poise is the nominal viscosity of blood, $K = 1\,\text{sec}^{0.39}$ is the consistency index, $D = \sqrt{I_2}$, $D_0 = 226.5\,\text{sec}^{-1}$, the power $n = 0.61$, the second invariant of strain rate tensor $I_2 = \frac{1}{2}\sum_i \sum_j L_{ij} L_{ij}$, $L_{ij} = \frac{1}{2}\left(u_{i,j} + u_{j,i}\right)$ and $u_{i,j}$ is the *i*-th velocity component gradient in the *j*-th direction.

$$\mu = \mu_\infty + \left(\mu_0 - \mu_\infty\right)[1 + (\lambda D)^2]^{\frac{n-1}{2}}. \tag{4}$$

For the Carreau model, □ □ = 0.6 is the viscosity at infinite shear rate, □ 0 = 0.033 is the viscosity at zero shear rate, □ = 3.313 is the time constant and *n* = 0.3568 is the power.

A sequential coupled-field analysis using ANSYS/FLOTRAN was involved. It takes into account the interaction (coupling) between fluid (blood) and structure (vessels wall) [12], [17]. The problem from this work is considered to be a steady-state fluid–structure interaction problem and is solved using the physics environment approach, where the results from fluid analysis become loads for the

structural analysis. Because the analyses are fully coupled, the results of the second analysis will change some input to the first analysis. Coupling is recursive where iterations between the different physics are performed until the desired level of convergence is achieved.

3. Results

Figure 3, at a constant blood velocity of 17 cms^{-1}, presents the variation of blood viscosity computed by the Carreau method, the most classical model considering it as being constant – 0.033 poise.

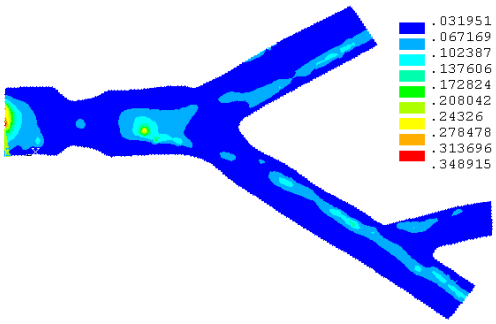

Fig. 3. The variation of blood viscosity (poise) obtained from the Carreau model

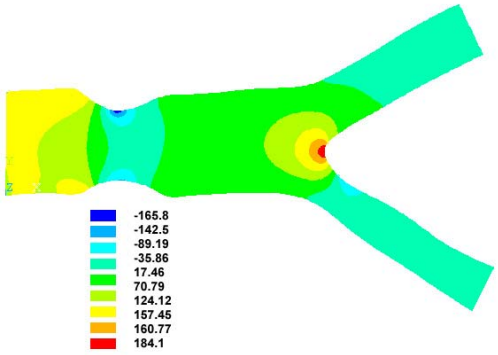

Fig. 4. The distribution of blood pressure (mm Hg)

The distribution of blood pressure (mm Hg) at a variable viscosity computed by the Carreau method is presented in figure 4 for a stenotic model with a channel area reduced by 40 per cent.

Figures 5 and 6 present the distributions of blood velocities (cms^{-1}) and displacements (cm) at a variable viscosity computed by the Carreau method for the stenotic model.

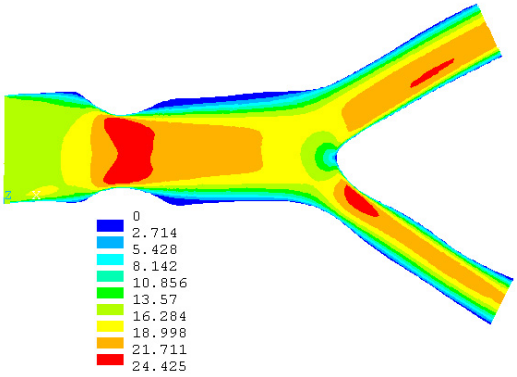

Fig. 5. Blood flow velocity field (cm^{-1})

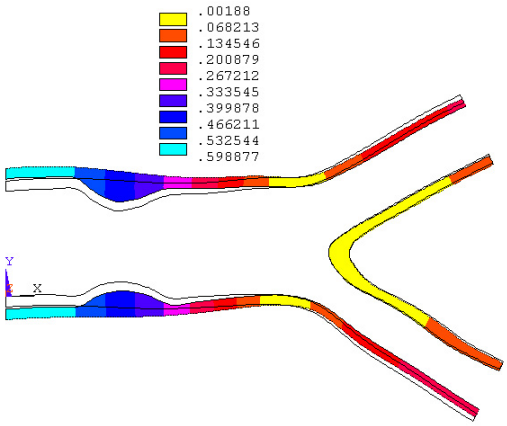

Fig. 6. Displacement distributions (cm)

It is important to mention that using such a model the wall stress (figure 7) can be computed in a rather realistic manner and, in this way, accurate information about the effects of this mechanical solicitation is obtained, especially that from the medical point of view, the zones located in the vicinity of the ramifications are susceptible to possible lesions of the wall.

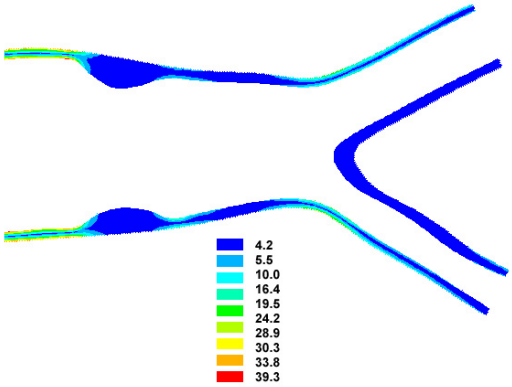

Fig. 7. Wall stress distribution (N/cm^2)

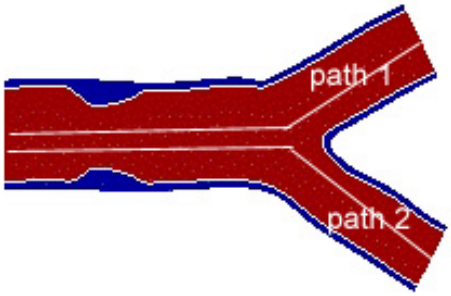

Fig. 8. Analyzed path lines on the model

The axial distribution of the velocities (velocity profile) represents a very important factor in analysing the pressures and the tangential stress over the wall of the blood vessels. The experimental techniques for plotting these profiles for various sections or paths are inaccurate and very time-consuming and they also require a specific equipment. In this respect, the advantages given by a Finite Element Method approach are obvious: figure 8 presents the paths along the median line of the vessel,

and figures 9 and 10 present the velocity (cms^{-1}) and pressure (mm Hg) distributions along it.

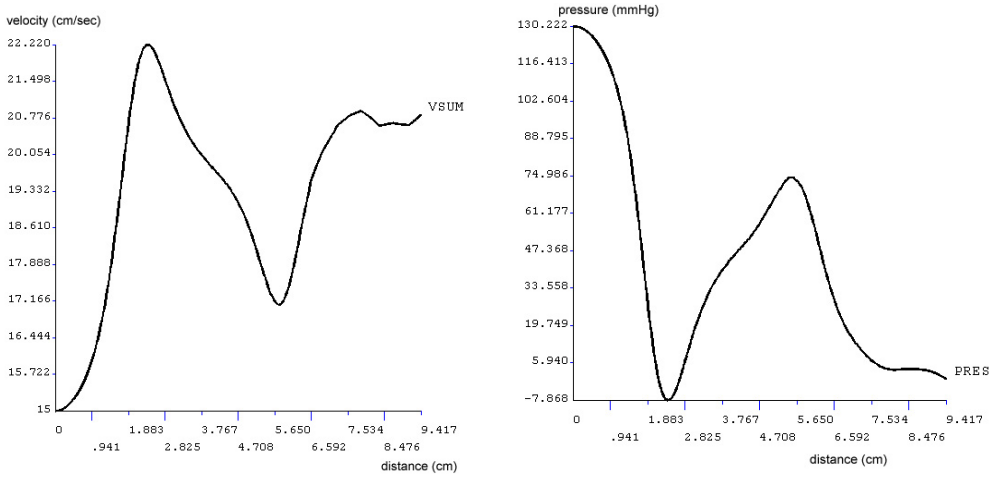

Fig. 9. Velocity (a) and pressure (b) along the path 1

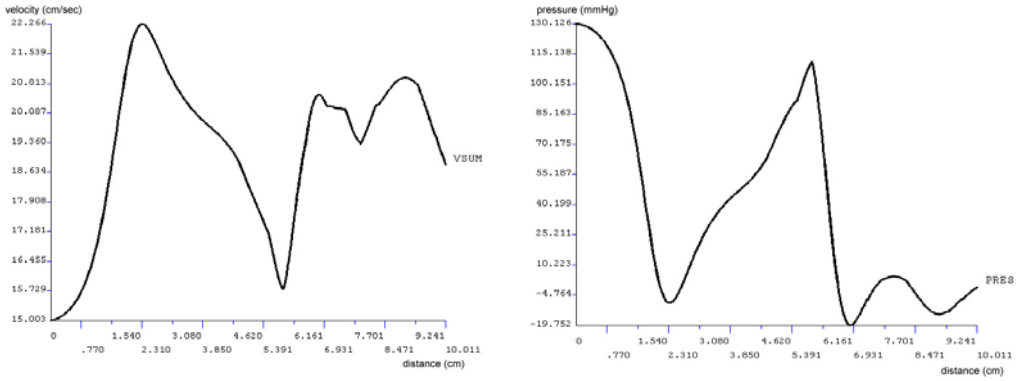

Fig. 10. Velocity (a) and pressure (b) along the path 2

4. Conclusions

The present study emphasizes the fact that the Finite Element Method could represent a very powerful tool in the hands of biomechanicians. In such approaches, complex geometries and delicate phenomena could be modelled and analysed with a high degree of confidence. Despite these advantages, it is also important to mention that the working hypothesis and the type of the elements should be carefully chosen, otherwise the probability of obtaining non-realistic data is very high.

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