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FLUID MECHANICS

NUMERICAL MODELING OF THE VENOUS VALVE FUNCTIONING

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ABSTRACT: The present work investigates the response of an air valve in terms of dynamic pressure due to kinematic support excitation. For the purpose of the analysis a full-scale model was installed in a laboratory setting, employing a shaking table to induce the dynamic input excitation in the form of a sine wave. The considered valve is an air release type, as a representative of those widely used in Japanese irrigation systems. Experiments were performed regarding the initial static pressure and the shaking parameters. The dynamic pressure and the movement of the valve's float was measured to establish the relation between the peak pressure and the mechanical response of the valve. After the analysis of the measured data some results and discussions are presented.A 3-D numerical model of the function of saphenous venous valve with a constant elastic modulus $E_{\mathrm{kl}} = 500$ kPa is developed depending on vein elastic modulus in series of $E_{\mathrm{v}} = 100$ kPa and 30 kPa, diameter 4 mm, thickness 0.66 mm and length 600 mm, and constant blood flow pressure at inlet 4.5 kPa and pulsatile one at outlet from 0 to 6 kPa according to the law of cosines. The mechanical behavior of the system "blood flow-vein-valve" is modeled using the Fluid-Structure Interaction application of computer program Ansys, in a case of initially damaged vein valve, unclosed with two-sided aperture of 0.1 mm.

The established distributions of the blood flow pressure and velocity in the deformed vein volume in a set with the established displacements and Mises stress in the valve and vein wall obtained during the cyclic valve opening/closing lead to stable working regimes in a vein with elastic modulus 100 kPa and to unstable ones in a vein with elastic modulus 30 kPa. Closing/proximal and opening/distal differences in the blood pressure are definite for the normal valve function.

It is established with $E_{\mathrm{v}} = 100$ kPa that under cosine-pulsatile inlet pressure from 3.6 to 4.8 kPa, an out-side pressure on the vein wall in the range of 4.25–4.5 kPa influences essentially the vein extensions and leaflet displacements and

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leads to decreasing aperture and/or to impact in the leaflets.

KEY WORDS: vein, venous valve, blood flow, fluid-structure interaction, elasticity.

1 INTRODUCTION

The venous system (VS) is an essential part of the human blood system. Its basic physiological function is the return of the non-oxide blood from the periphery back to the heart. Moreover, the VS represents an important blood reservoir which could be activated under different conditions of exercise. Due to this reason, the parameters of the blood flow are different along all the VS, and they perform values of optimal operation ranges at constant temperature of 37°C.

FUNCTIONAL MORPHOLOGY OF THE VEINS

The venous wall and the arterial wall are characterized by significant diversity in reference to their histological structures. The venous wall has thinner layers and contains epithelial cells, muscular fibers, elastine and collagen filaments. Thereby, the venous lumen is more flexible. It has compressible and relaxing capacities in larger ranges in comparison to the arterial ones. The small thickness of the venous wall allows its easy deformation under external pressure thus performing vein stretchability during filling up with blood stages. This veins' specificity plays a key role for the blood transport. The surrounding muscles shortening compress the veins while conducting their activity. As a result, the blood is pushed up to the heart. By default, the healthy vein valves prevent the blood flow downwards.

MECHANISMS OF THE BLOOD MOVEMENT IN THE VEINS

In the veins, the blood moves according to pressure gradient nearly 10–15 mmHg, which is rather lower than the values that are typical for the arteries. However, this value is completely sufficient to ensure conducting the return of the blood to the heart. In body stand position, the transport conditions are changed significantly due to the gravity influence. Therefore, additional mechanisms are needed to assist the necessary blood flow volume realization, so in fact, the muscles act as a pump during their contraction. Approximately, 0.77 mmHg per centimeter are added to the venous pressure in most body segments below the heart. This value is subtracted in all body parts above the heart. In stand position the muscle pump activity results in venous pressure growth. The effect appears in an equilibrium disturbance between the filtration and reabsorption.

The synchronized activity of the venous valves function in a set with the muscle contractions and the respiratory movements of the diaphragm operate as a thoracic pump which directs the blood flow from the lower limbs towards the heart.

The venous flow is regulated by the sympathic part of the vegetative nervous system. Alfa-adnergic receptors appear in the smooth-muscle cells of the venous wall. Due to this reason, venous constriction (contraction of the venous lumen) is pronouncedly observed during the in phase stimulation, thus decreasing the blood volume in the veins.

The valves, structured with flexible elastic fibers, are situated in the vein, Fig. 1. They have two arc-form leaflets. Their edges are freely coming into contact along the vein diameter thus, forming cusps. Along their periphery, the leaflets are fixed to the vein wall where the elastic fibers are more concentrated in them than in the vein wall. The two leaflets switch on/off the venous blood flow by contacting with and moving apart their free edges (cusps) oppositely. Like that, unidirectional movement of the blood is ensured. It is established that in the time this process is performed in four phases [1,2], namely: opening phase with duration of 0.27 ± 0.05 s; equilibrium phase of 0.65 ± 0.08 s with oscillations of the opened valve leaflets in amplitude of 0.01 cm to 0.16 cm; closing phase of 0.41 ± 0.07 s; closed phase of 0.45 ± 0.05 s.

The process described above is pulsatile, characterized by a frequency which depends on the muscles' action. During the closed phase, the valves keep portions of blood in the venous spaces bordered by them. In this way the whole blood vessel volume is separated in parts and the veins are prevented from expanding due to an eventual blood leaks to the lower body limbs, driven by gravity. The venous valve opening, and tight closing are complex mechanical-biological processes [3]. As it was already highlighted, they are conducted by the successful interaction of the vegetative nerve system, the muscle pump, and the respiration body movements. In a mechanical discourse, the blood flow is driven by the difference between the venous pressures distally and proximally of any valve. It also strongly depends on the mechanical properties of the two valve leaflets, as well as the venous wall characteristics. The press force experienced by the venous wall from outside, caused by the surrounding tissues and the muscles' contractions of the lower limbs, is another key factor. In general, a combination of three parameter' optimal values is necessary for a clinically healthy venous blood system best functioning:

— the pressure value of the blood flow before the valve P_{inlet};

— the pressure value of the blood flow after the valve P_{outlet};

— the outside pressure value exercised upon the venous wall P_{wall}.

The elastic characteristic of the valve and the vein are factors which ensure the optimal conditions for unidirectional blood flow. The damages in their structures lead to loosening of the wall, causing valve disfunction, which result in a reflux emergence (a back-turn of the blood flow to the periphery).

The valve insufficiency appears based on the chronic venous disease. The presence of reflux increases the blood volume and determines the disease advancement. Many explanations in a pathophysiological and biochemical discourses highlight this complex process. However, the development of the contemporary digital processing capacities and methods [4–6] in a set with the development of the MRI-measurements [7] and the newest observation and analysis instrumentation made possible the mathematical/numerical modeling of the blood flow within the vascular system. This allows a new approach to describing the mechanical influence of the blood fluids upon the venous structures. In this elaboration, all nerve-pathological impacts on venous wall, as well as the inflammatory processes possible to influence the valve leaflets are not considered. The subject of this elaboration is focused only on the velocity and the pressure exercised by the movement of liquids in the vessel's lumen. Regarding the phenomenon of the venous blood flow (as described above), the aims of this work are reduced to

- Development of a numerical model of the work of a venous valve considering the elastic properties of a three-component system involving "blood flow–vein wall–valve leaflets".

- Determination of optimal function values and dependencies between the parameters of the blood flow, the characteristics of the venous wall and the venous valve mechanical responses during their interaction under different operational regimes.

- Defining some critical regimes of this three-component system operation.

For this purpose, the pressures, velocities, and flow of fluids in a set with displacement, deformation and stresses in the vein wall and vein valve are observed. A mechanical model of the phenomenon "Functioning of the venous valve" will be defined from a mechanical point of view applying both: the Methods of Finite Elements and Finite Volumes in an approach of fluid-structure interaction. Work parameters compatible with really observed human ones will be assigned to this model. The model performance will be detected, and the values development will be read as results. Analyses and comparisons of these results will be presented in a trend to search for causal-effective lines and dependencies between the fluid and the structural parameters of the process modeled. Some conclusions will be finally presented.

2 Fluid-Structure Interaction Model of the Venous Valve Function

2.1 Assumptions

In the developed model the phenomenon of blood transport across a venous valve is recognized as a mechanical stop and start of a laminar flow by elastic elements. Elastic ruling elements possessing arc-form, imitating the two valve leaflets, are placed perpendicularly of the axis, and bounded in periphery to the inside of a tube imitating the venous vessel. Because of the fluid pressure gradient and the materials' elasticity, a fluid flow in the tube and an elastic deformation of the tube wall are initiated. Both valves' elements start acting in a pulsative mode. Thus, a fluid-structure interaction appears. In this work, the influence of the material mechanical characteristic on the flow will be only modeled following an FSI-approach applied to a carotid artery [8]. The data interpretations will be also made from the corresponding mechanical treatments.

The blood is modeled as continuous medium by applying of Newtonian fluid. The elastic vein wall and the elastic valve of two leaflets are modeled as solid media according to Hookean law. The corresponding volumes occupied by the fluid medium and the three structured media are accordingly divided in finite number of volumes and finite number of elements containing corresponded nodes. Jointed nodes translate the interaction between the fluid and the structure. The characteristics of the blood flow and the venous-valve response will be numerically obtained in all the nodes, and will be graphically presented and discussed in some chosen representative points/nodes belonging to the blood and to the vein. In fact, the blood-transport process is quite complicated, so to some extent this is an idealization because of the discretization of the two continuous inhomogeneous media in two isotropic and homogeneous interacting models. Boundary conditions are acting on the models on the inlet/outlet fluid surfaces, as well as on the outside surfaces of the vein.

The following basic assumptions are available during the development of the model. The fluid medium continuously fills in the space closed by the inside surfaces of the vein wall, the valve and the two inlet and outlet surfaces. In the case of closed valve by the leaflet walls this space is separated in two volumes - distal and proximal. Solid media continuously feel the volume of the vein and these of the valve leaflets.

The mechanical properties of these media coincide with averaged characteristics of the real media and could be taken from references. The vein and valve materials were considered as homogeneous, linearly elastic materials with different Young's module. The blood was assumed as Newtonian fluid. Interactions between the structures inside of the vein, valve and blood are not discussed in this research.

The modeling time of the FSI-process as described, is essentially larger than the time of the free displacements of the real particles of both media.

With the above assumptions, and under availability of conservation laws of mass, of energy, of momentum and moment of momentum, as well as both the Newton's and Hook's constitutive laws characterizing the corresponding media, the mechanical behavior in geometric points of the vein/valve and the blood flow, represented by nodes belonging to the corresponding finite element model and finite volume model, describe the averaged mechanical behavior of the two interacting real media according to given boundary conditions.

2.2 Geometrical parameters, finite volume model of the blood and finite element model of the vein/valve

A 3-D numerical simulation of the blood flow – the vein and the valve interaction is developed. The 3-D model of the vein is assumed as a tube possessing inside radius of 0.002 m, thickness of 0.00066 m [9] and length of 0.6 m, Fig. 1a. The 3-D valve is constructed by two leaflets possessing arc-leaf forms with equal radii of 3.5 mm and thickness of 0.2 mm, and bonded to the inside tube surface, Fig. 1b. The leaflets are

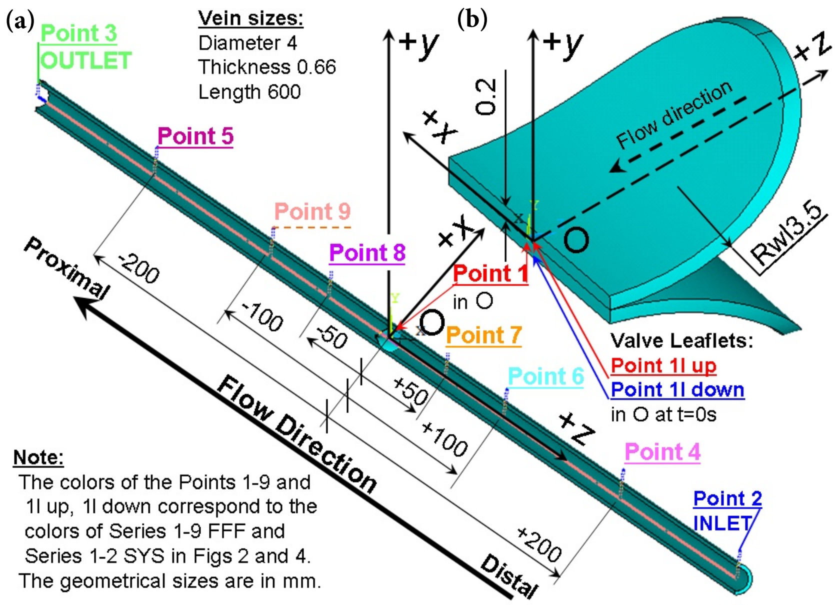

Fig. 1: Models of the discussed vein and valve. Situations of the points where: (a) blood flow parameters Points 19 and (b) valve displacements Points 1lu-1ld, are read.

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closing along the tube diameter in the mean of the tube, where the center of $Oxyz$ Cartesian coordinate system is embedded, Fig. 1a. The plus direction of the tube z-axis is orientated along the venous axis directed to the lower limbs. The x- and y-axes are orientated along the horizontal and vertical venous diameter. Consequently, the modeled blood flow runs to the opposite z-direction, i.e., from lower limbs, positive z-direction, below the valve to the heart, negative z-direction above the valve. The center of the valve is placed in the center O of the coordinate system. The leaflets are directed along x-axis, i.e., horizontally with a valve hole. The displacements of the leaflets during opening/closing phases are read along the y-axis, from the center O. Initially, a symmetric hole of 0.1 mm along Ox, imitating physiological damage orientated symmetrically, is assigned between the leaflets' cusps. In this model, this initial unclosing of the valve could be interpreted as a consequence of some clinical deviations observed elsewhere [3].

The vein is held immovably at the frontal inlet $+z$ and outlet $-z$ surfaces. The vein volume including this of the valve is meshed by 677,788 finite elements Solid 186 with 97921 nodes. The volume of the blood flow is meshed in 188,288 finite volumes with 40,516 nodes. The distributions of the pressure and the blood velocity are numerically established along the z-tube axis in chosen representative points with coordinates $z = 0$ m (Point 1) placed in the $Oxyz$-center, at inlet $z = +0.3$ m and outlet $z = -0.3$ m (Point 2 and 3, respectively), before the valve in distal direction with $z = +0.2$; $+0.1$; $+0.05$ m corresponding to Points 4, 6 and 7; and after the valve in $z = -0.05$; -0.1; -0.2 m corresponding to Points 8, 9, 5, Figs. 1a, b. In the following diagrams the curves representing the data obtained in these points are given with Series for FFF from 1 to 9, correspondingly. The numerically established mesh displacement of the leaflets is represented in Points 1lu (Leaflet Up) and 1ld (Leaflet Down), Fig. 1b, with initial coordinates $x = z = 0$ and initial valve unclosing of 0.1 mm reflected by $y = +0.05$ and $y = -0.05$ mm. The displacements are given in the figures by Series for SYS for 1 and 2. The displacements and the Mises stresses in the valve/vein wall are analogically presented in Oyz-plane by colored diagrams.

2.3 Material characteristics of the vein, valve and blood

The process of interaction between the vein, the valve and the laminar blood flow is modeled under followed material properties assumed according to experimentally measured values:

— Two cases of elastic mechanical behavior of venous wall are treated:

1. Vein wall possessing Young's modulus $E_{\mathrm{v}} = 30$ kPa. The value is chosen at the lowest limit between a normal Elastic module and an elastic module reflecting high risk of vein incompetence [10].

2. Vein wall possessing Young's modulus $E_{\text{v}} = 100$ kPa. The value is chosen as a representative of a normal elasticity, but it is lower than an averaged one, which appears near to 200–250 kPa [10].

— Equal elasticity of the two valve leaflets of $E_{\text{kl}} = 500$ kPa [4]. This value is assumed as a constant in this elaboration, and it is chosen to be equal to the Elastic modulus of the elastine, with the presumption that the leaflets are lesser elastic than the vein because of their structure where the elastine is prevailing at expense of muscle's tissue, as it was pointed out in the Introduction.

— Equal Poisson ratios of 0.48 and equal densities of 1050 kg/m^3 are assumed for the vein and the valve leaflets.

— Blood density is of 1060 kg/m^3; and blood viscosity is of 0.00345 Pa s.

The environment temperature is 37°C. Gravity is assumed along the positive z-direction.

2.4 Boundary conditions

With these geometrical and material parameters the following boundary conditions imitating different blood loads on the valve and vein with different elasticity are assigned

— I$^{\text{st}}$ boundary condition: at the tube inlet (i.e., below the valve – distally from the heart), firstly constant pressure and after pulsatile according the law of cosines, given by blue line in the diagrams.

— II$^{\text{nd}}$ boundary condition: at the tube outlet (i.e., over the valve – proximally to the heart) only pulsatile pressure according to law of cosines but in a larger range, given by green line in the diagrams.

The two pressures are assigned with pulse frequency of 72 min^{-1} in phase, similarly to the observed in [6]. It is assumed in the model theoretically that the valve opens when the pressure over the valve P_{proximal} induced by P_{outlet} (P5, 8, 9) becomes lower than the pressure below the valve P_{distal} induced by P_{inlet} (P4, 6, 7), and conversely the valve is closing when the pressure over the valve P_{proximal} (P5, 8, 9) induced by P_{outlet} becomes higher then the pressure below the valve P_{distal} (P4, 6, 7), i.e.,

at opening $P_{\text{outlet}} < P_{\text{inlet}}$ inducing $P_{\text{proximal}} < P_{\text{distal}}$;

at closing $P_{\text{outlet}} > P_{\text{inlet}}$ inducing $P_{\text{proximal}} > P_{\text{distal}}$.

The precise values of these pressures $P_{proximal}/P_{distal}$ depend on the ratio of elasticity of the venous wall (E_{v}) to the elasticity of the valve (E_{kl}). By their variable values, the boundary conditions assigned like that provoke a cosinus-pulsating blood flow similar to this experimentally observed in [9]. This flow opens and closes the valve while it also deforms the valves leaflets according to E_{kl}.

The tube length given of 0.6 m, including inlet- outlet-sections, agrees with a venous length which begins from the leg above the knee, where the theoretical venous pressure is in order of 40 mmHg/5320 Pa, and ends around the beginner part of Inferior Vena Cava where the venous pressure is about 22 mmHg/2926 Pa. In the present model, the approximate values are pointed out as selected values for the two-boundary inlet-outlet conditions.

In the tube inlet pulsatile fluid pressure is given according to

$$P_{\text{inlet}} = C_{\text{in}} + hA_{\text{in}} \cos(nt + 0.36), \tag{1}$$

where t is the time, s, $n = 2\pi f$ and $f = 1.2$ is the natural frequency of oscillation, here chosen to be correspond to 72 min^{-1}, C_{in} is the average-constant value of inlet pressure, A_{in} is the amplitude of the pulsating part of the inlet pressure, $h < 1$ is a divisor of an initially assumed amplitude, Table 1. If we assume that there exists an optimal A_{in} for clinically healthy persons obtained as results of statistical data. Then, h could be interpreted as a parameter of declinations. Its values will play a managing role in the tube's deformation behavior under constant or pulsating inlet pressure. In the tube outlet as a boundary condition a pulsatile fluid pressure is given according to:

$$P_{\text{outlet}} = C_{\text{out}} + A_{\text{out}} k \cos(nt + 0.36), \tag{2}$$

analogically, here t is the same time, s, $n = 2\pi f$ and $f = 1.2$ is the same natural

Table 1: Mechanical characteristics and boundary conditions of the numerical model

			P_{inlet}, eq (1)			P_{outlet}, eq. (2)			
No	E_{v}, kPa	E_{kl}, kPa	C_{in}, kPa (mmHg)	h	A_{in}, kPa (mmHg)	C_{out}, kPa (mmHg)	k	A_{out}, kPa (mmHg)	P_{wall}, kPa
1	100	500	4.5 (33.83)	0	3 (22.55)	3 (22.55)	1	3 (22.55)	0
2	30	500	4.5 (33.83)	0	3 (22.55)	3 (22.55)	1	3 (22.55)	0
3	100	500	4.2 (31.57)	0.1	3 (22.55)	3 (22.55)	1	3 (22.55)	0
3a	100	500	4.2 (31.57)	0.1	3 (22.55)	3 (22.55)	1	3 (22.55)	4.25
3b	100	500	4.2 (31.57)	0.1	3 (22.55)	3 (22.55)	1	3 (22.55)	4
3c	100	500	4.2 (31.57)	0.1	3 (22.55)	3 (22.55)	1	3 (22.55)	4.25

frequency of oscillation of 72 pulses/min, C_{out} is the average-constant value of the outlet pressure, A_{out} is the amplitude of the pulsating part of the outlet pressure, k is similar to h, Table 1. Both the pressures P_{inlet}, P_{out} are sinphased. The FSI-process between the laminar blood flow and the vein/valve is considered for time period of 6 seconds. This process time is modeled in 72 steps thus applying a time step of 0.0833333 s.

In this research a numerical model is presented in phase of development. So, different parameters are initially prescribed to this model in order to outline its possible working space, and to observe qualitatively the initial mechanical behavior of the built model system "fluid-structure", while the real cases with precise practically measured parameters will be situated as a sub-ensemble somewhere in between the ensemble of values defined by these cases, which here are treated as a function $(P; V_z) = f(P_{\text{inlet}}; P_{\text{outlet}}; E_{\text{v}}; E_{\text{kl}}; P_{\text{wall}})$.

3 Results

There are numerically established:

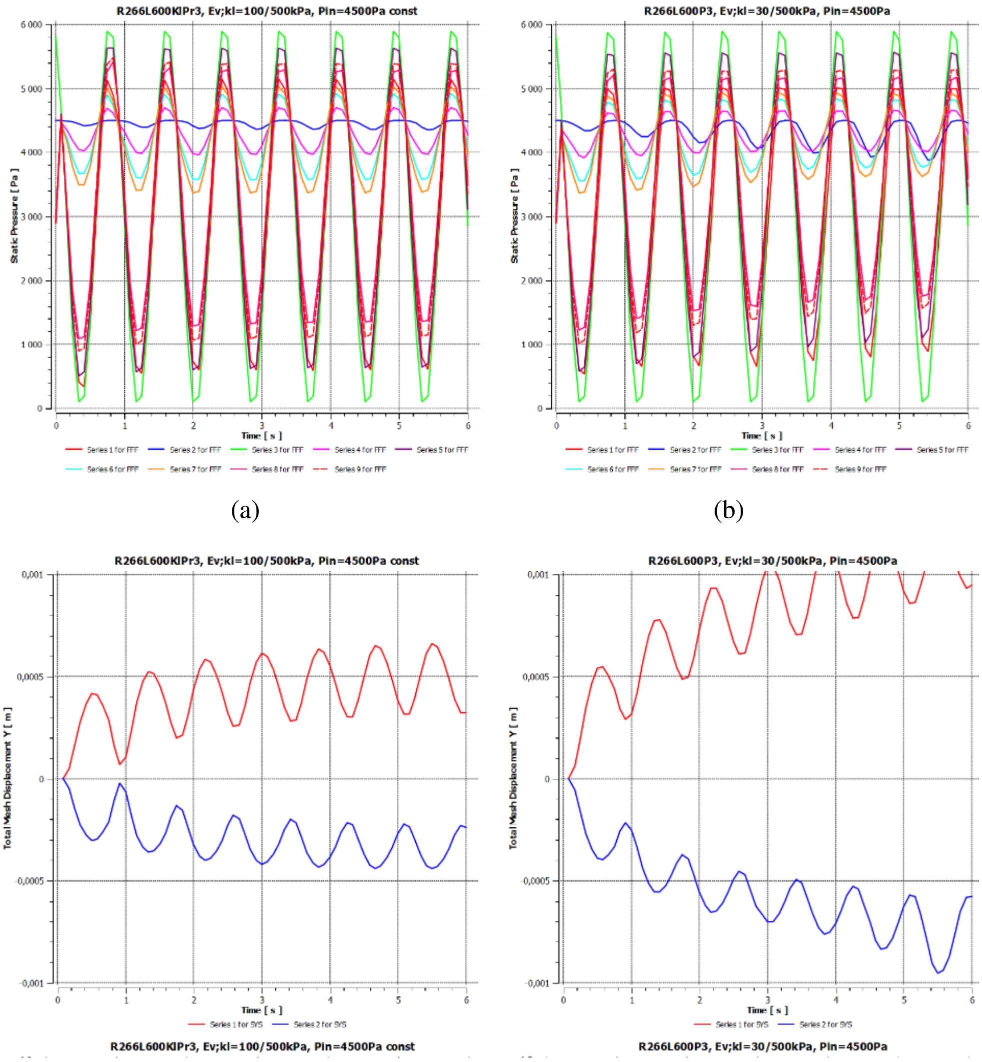
— pressures $P_{\text{distal; proximal}}$ (in Points 1–9) which provoke opening of the valve and its closing subsequently, Figs. 2a, b.

— displacements of the points of the valve (Points 1lu, 1ld) which determine its opening/closing Figs. 2c, d and displacements in the vein wall Fig. 3.

— distributions of the blood flow velocity along the vein axis, Figs. 2d, e and Fig. 3.

The permanently increased pressure exercised on each arbitrary given point of VS can damage it. The mechanical damage will produce poor valve closing and gradual leaflet loosening. Under such circumstances, the interactions between the fluid and the elastic tube, e.g., between venous blood flow and vein/wall, is discussed. The blood pressures bring to the venous wall loosening and failures are computed. It was assumed, that the vein is only elastically deformed, during its optimal operation at the various regimes of interaction with the blood flow passing through the valve, Table 1. In the numerical examples No 1 and No 2, the given $E_{\text{v}} = 100$ and $E_{\text{v}} = 30$ kPa are aiming to establish the influence of the various venous elasticity on the interaction "blood flow–vein/valve". The other model parameters are identically assigned including $E_{\text{kl}} = 500$ kPa, constant $P_{\text{inlet}} = 4500$ Pa and cosine-pulsatile P_{outlet} about 3000 Pa symmetrically in the range of 0–6000 Pa, Figs. 2a, b, Table 1.

With these assumed elastic properties of the vein different pulsating pressures are established in the distal and proximal sections of the vein, below and over the valve,



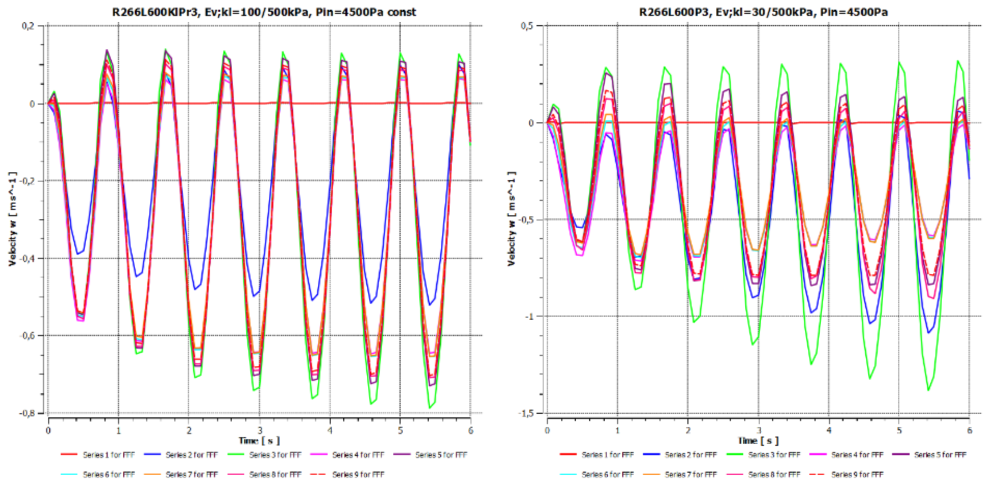
(a) (b)

(c) (d)

Fig. 2: Continued on the next page.

respectively. It results in deferent mechanical behavior in the system "vein/valve–blood flow", as with $E_{\mathrm{v}} = 30$ kPa the vein bends and loses its linear stability.

In case of $E_{\mathrm{v}} = 100$ kPa the system is stably deformed. The maximal values of the PROXIMAL-pressure (after the valve) in Points 5; 8 $\max P^{\mathrm{Pt5;8}}_{\mathrm{PROX}} \approx 5600$ Pa (42.10 mmHg); and 5300 (39.84 mmHg), Fig. 2a, are higher than the maximal ones



(e) (f)

Fig. 2: Parameters of FSI at (), (c), (e) $E_{\text{v;kl}} = 100$ kPa/500 k and at (b), (d), (f), $E_{\text{v;kl}} = 30$ kPa/500 k under pressure $P^{\text{const}}_{\text{inlet}} = 4.5$ k and $P^{\text{const}}_{\text{outlet}} = 3$ kPa $+P^{\text{ampl}}_{\text{outlet}} = \pm 3$ k: (), (b) venous blood pressure; (c), (d) displacement of leaflets; (e), (f) z-axial blood velocity.

of the DISTAL-pressure (before the valve) in Points 4; 7 $\max P^{\text{Pt4;7}}_{\text{DIST}} \approx 4700$ Pa; 5000 (35.33 mmHg; 37.59 mmHg). Thereby, the higher maximal PROXIMAL-values close the valve exercising the higher pressure upon the concave arch-wise sides of its leaflets, and upon the inside venous wall emerged over the valve. A difference of 900 /6.76 mmHg is read between the pressures in Points 5; 4, both are located at a distance of $z \mp 0.2$ m from the valve, respectively. Between the Points 7; 8 which are fixed at a distance from the valve ∓ 0.05 m, this pressure difference is 300 Pa or 2.25 mmHg. In this way, a *closing difference* $\Delta P_{\text{close}} = \max P^{\text{Pt5;8}}_{\text{PROX}} - \max P^{\text{Pt4;7}}_{\text{DIST}} \Delta P^{5/4;8/7}_{\text{close}} = 900$ /6.76 mmHg; 300 /2.25 mmHg is defined. It is evident, in the same couples of Points 4/7 and 5/8, when the minimal values of the PROXIMAL-pressure $\min P^{\text{Pt5;8}}_{\text{PROX}} \approx 600$ /4.51 mmHg; 1300 /9.77 mmHg are lower than the minimal values of the DISTAL-pressure $\min P^{\text{Pt4;7}}_{\text{DIST}} \approx 4000$; 3400 (30.07 mmHg; 25.56 mmHg), the valve opens in consequence of the higher minimal DISTAL-pressure exercised upon the convex arch-wise sides of its leaflets, and also upon the inside venous wall emerged below the valve. Difference of 3400 Pa is formed in the minimal pressures between Points 4/5. Between Points 7/8 this difference is 2100 Pa. Thereby, the *opening difference* $\Delta P^{4/5;7/8}_{\text{open}} = 3400$; 2100 (25.56 mmHg; 15.78 mmHg) is defined. The difference induced by P_{inlet} and P_{outlet} does not allow the valve to get closed under constant pressure $P_{\text{inlet}} = 4500$ /33.83 mmHg.

The closing and opening differences are in dependence of both the boundary conditions in a combination with the elasticities defined by the modules E_{v}; E_{kl}. The operating regimes commented here are determined only in this way at a ratio $E_{\mathrm{v}}/E_{\mathrm{kl}} = 100/500$ kPa.

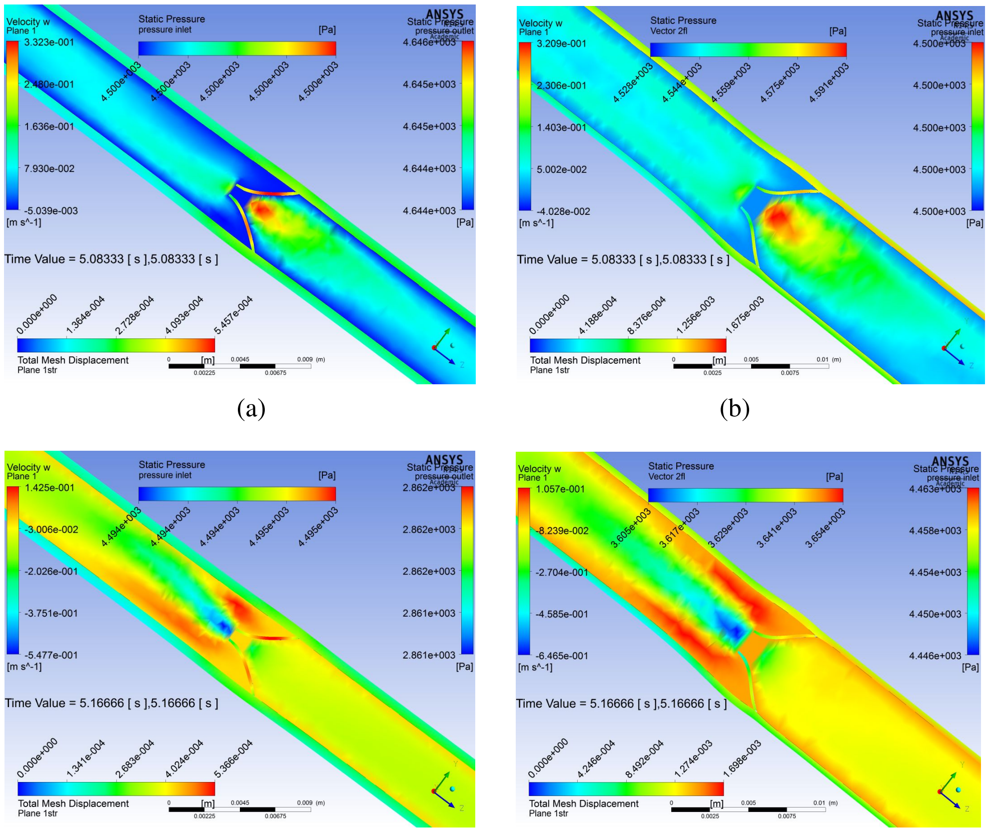
The increased elasticity of the vein with venous elastic module $E_{\mathrm{v}} = 30$ kPa brings to maximal pressures in Points 5; 8 (behind/over the valve) of $\max P_{\mathrm{PROX}}^{\mathrm{Pt5;8}} \approx 5500$ Pa; 5200 (41.35 mmHg; 39.09 mmHg) and in Point 4; 7 (before the valve) of $\max P_{\mathrm{DIST}}^{\mathrm{Pt4;7}} \approx 4600$ Pa; 4900 (34.58 mmHg; 36.84 mmHg), Fig. 2b. Similar $\Delta P_{\mathrm{close}}^{5/4;8/7} = 900$; 300 (6.76 mmHg; 2.25 mmHg) is defined, but it emerges at lower maximal pressures. Accordingly, the minimal ones are $\min P_{\mathrm{PROX}}^{\mathrm{Pt5;8}} \approx 1000$ Pa; 1700 (7.51 mmHg; 12.78 mmHg) and $\min P_{\mathrm{DIST}}^{\mathrm{Pt4;7}} \approx 4000$ Pa; 3600 (30.07 mmHg; 27.06 mmHg). Difference $\Delta P_{\mathrm{open}}^{4/5;7/8} = 3000$ Pa; 1900 (22.55 mmHg; 14.28 mmHg) is defined. It is obtained approximately as $E_{\mathrm{v}} = 100$ kPa, but in this case the wall is more elastic. A venous dilatation around the valve is clearly expressed – the venous diameter increases in this location, and as a result, the valve leaflets cannot tightly contact, i.e., a diametrical hole emerges freely. In this case, it should be mentioned that for the period of 6sec, the minDISTAL-pressures in Points 6; 7 increase, and maxDISTAL-ones remain constant. The opening difference increases, and the closing remains constant, Fig. 2b. This brings to pulsatile enlargement of the venous DISTAL-volume, before the valve, because the DISTAL-pressures are aspired to the maximum. They do not sufficiently fall in order to close additionally the valve, because the venous wall elasticity acts against them. As a result, at the 8th second the system loses the stability of the pulsating elastic deformation; the deformation becomes irreversible, and finally, the vein becomes bended. Consequently, in case of $E_{\mathrm{v}} = 30$ kPa, the $P_{\mathrm{inlet}} = 4500$ is limited by the pressure bearing capability of the venous tissue. To keep the system stability, this pressure should be decreased after the period of 6 s. Similar working phases, including similar variability of the PROXIMAL/DISTAL-pressures, are experimentally observed [2].

The periods of the working phases in this model could be read following the displacements of the cusps, Figs. 2c, d. The dependency of these displacements on the diverse venous elasticity is obvious under equal other parameters. It is established, that the closing phase of the valve continues 0.3333 s. It is commensurable with the maximal time limit for such phase in experimental investigations [1]. In this elaboration such a phase begins, for example from $t = 4.75$ s, when the leaflets are open, the DISTAL-pressures become equal, and the closing process continues until they are increasing, but the PROXIMAL-pressures are also more intensively increasing in the same period. The closing phase finishes when all DISTAL- and PROXIMAL-pressures along z-axis achieve equal values, for example, in $t = 5.0833$ s. As it is visible from Figs. 2a, b, Figs. 3a, b, and moreover the PROX-pressure begins to fall more intensively than the DISTAL-pressure. Then the elasticity of the system “vein-valve” is connected to the PROX-pressure and begins also to act against the DISTAL-pressure until an equilibrium is achieved. In case of a weaker elastic system, as well as in case of the elastic force superiority, caused by “vein-valve” to the pressure force induced by the opening pressure difference $\Delta P_{\text{open}}^{4/5;7/8}$, an impact in the leaflets could be followed in this modeled phase. During the period when the vein and the valve elasticity perform a balance of both pressures, the valve remains closed. After that, the valve opens driven by the increasing DISTAL pressure. The opening phase in this model, Figs. 2c, d, also continues 0.3333 s, for example from $t = 5.166$ s, when all decreasing pressure become equal, Figs. 2a, b and the intensity of the decreasing PROXIMAL- becomes higher. Next, the intensity of its increase is lower than this of DISTAL pressure. The time of the observed model opening phase continues until 5.5 s, and it is also commensurable to the minimal time of such a phase observed experimentally [1].

In the case of $E_{\text{v}} = 100$ kPa the radial displacement in the valve (Point 1lu) achieves values from 0.3 mm to 0.65 mm, Fig. 2c, while in the case of $E_{\text{v}} = 30$ kPa the same displacements achieve almost twice more from 0.85 to 1.5 mm, Fig. 2d. The valve opens/closes with lower displacements with $E_{\text{v}} = 100$ kPa, because of the higher elastic venous module, which successfully acts in reverse and as a result, DISTAL-pressure falls. Following the leaflets’ displacements, Figs. 2c, d, in Figs. 2e, f, the blood flow velocity is read during working phases depending on the pressure pulses, Figs. 2a, b. Applying $E_{\text{v}} = 100$ kPa periodical refluxes are observed along all the tube length with maximum of 0.1 m/s and minimum of 0.05 m/s, at a distance of 0.1 m from the inlet, Points 2; 4.

During the valve closing time, for example from 5.5 s to 5.8333 s, decreases of both pressures are evident. However, the DISTAL one is higher than the PROXIMAL. The balance occurs in 5.6 s. After this moment, both the pressures increase but the PROXIMAL one does it more intensively. This circumstance becomes a condition for reflux. Proximally, after the valve, the blood flow velocity has also minimum of 0.12 m/s, Points 3–5. In the case of $E_{\text{v}} = 30$ kPa, reflux is only established at $z = +0.05$ m, Point 7, Fig. 2f, the average velocity Vz on both sides of the valve has value of -0.4 m/s, while in the case of Ev $= 100$ the average velocity V_z is of -0.3 m/s. The observed maximum speed along the vein varies from $V_{z_{\max}} = 0.65$ m/s up to $V_{z_{\max}} = 0.72$ m/s in open valve position.

In Fig. 3, along y-axis near to the valve radial venous enlargements of 1.25 and 0.025 mm are correspondingly established for $E_{\text{v}} = 30$ and 100 kPa at $P_{\text{inlet}} = 4500$. The plus-values of the velocity, Figs. 2e, f, indicate returns of the velocity direction



(a) (b)

(c) (d)

Fig. 3: Displacement in the vein and the valve during interaction (in equal scales) with blood flow and distributions of the blood flow velocity V_z in the Oyz-plane at the end of closing phase $t = 5.083$ s and at beginning of opening phase $t = 5.166$ s under const $P_{\text{inlet}} = 4500$ Pa: (a), (b) with venous elastic module $E_{\text{v}} = 100$ kPa; (c), (d) with venous module $E_{\text{v}} = 30$ kPa.

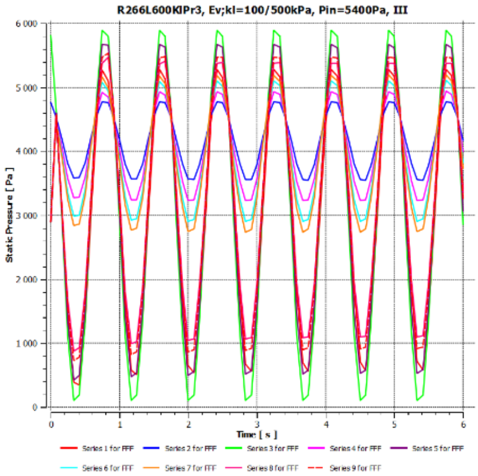
(reflux) with decreasing values in Points 3; 5; 8, 9 i.e., behind the valve only at a distance of 5 cm, Point 8. In this case, it should be highlighted that these V_z-distributions are related to the chosen representative 9 points along z-axis, Fig. 1a. Actually, the blood velocity gets various values in the different cross-sections along the z-axis, as well as at the different geometrical points (nodes) of the fluid volume, Fig. 3. Moreover, in vivo the blood is a multiphase fluid. In the suggested treatments, an increasing or decreasing of the elastic module could be interpreted as a consequence of different biological processes which could be developable in real circumstances [3].

The assignment of pulsating pressure in the inlet of the vein with $E_{\text{v}} = 100$ kPa, Fig. 4a, Table 1, induces maximal pressure in Points 5; 8 (after the valve) $\max P_{\text{PROX}}^{\text{Pt5;8}} \approx 5650$ Pa (42.48 mmHg); 5400 (40.60 mmHg), and in Points 4; 7 (before the valve) $\max P_{\text{DIST}}^{\text{Pt4;7}} \approx 4900$ Pa (36.84 mmHg); 5200 (39.09 mmHg). The closing difference is $\Delta P_{\text{close}}^{5/4;8/7} = 750$ (5.64 mmHg); 200 (1.50 mmHg). The minimal pressures are $\min P_{\text{PROX}}^{\text{Pt5;8}} \approx 500$ Pa (3.76 mmHg); 1100 (8.27 mmHg), and $\min P_{\text{DIST}}^{\text{Pt4;7}} \approx 3200$ Pa (24.06 mmHg); 2700 (20.30 mmHg). The opening difference is $\Delta P_{\text{open}}^{4/5;7/8} = 2700$ Pa (20.30 mmHg); 1600 (12.03 mmHg). An additional decrease in the unclosing of the valve is observed. The leaflets are radially opened in distance of 0.55 mm, and unclosed with radial hole of 0.25 mm, Point 1lu, Fig. 4c at the initially given radial unclosing distance of 0.05 mm, i.e., the venous enlargement around the valve area is 0.2 mm because of the decreased pressure $P_{\text{inlet}}^{\text{min}}$, which falls until 3600 Pa, Table 1, Fig. 4a. The observed reflux blood flow velocity is lower than 0.1 m/s with synchronic velocities V_z. The maximal reflux V_z in a constant pressure condition of 4500 kPa was above 0.1 m/s.

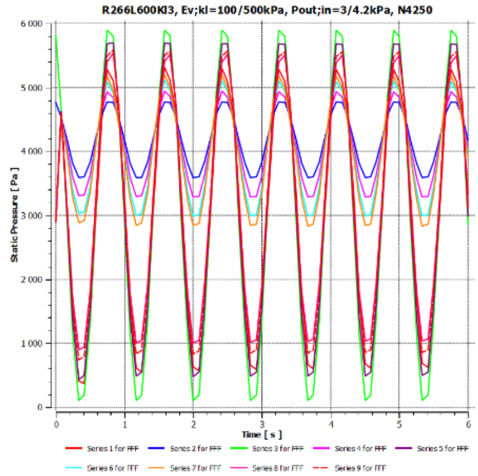
Up to here, all performed investigations were focused on the closing/opening conditions of the valve depending on the $P_{\text{inlet}}/P_{\text{outlet}}$ ratio only. The influence of the opening/closing pressure differences on the displacements of leaflet's cusps was established as an important ruling parameter.

The venous enlargement, Figs. 3b, d, could be compensable by pressure upon the outside tube wall, respectively upon outside vein wall, P_{wall}. This outside pressure could bring to decrease of the unclosing of 0.1 mm preliminary given between the leaflets, and this pressure could even advantage a tight closing of the valve, but it could also cause impacts to the leaflets.

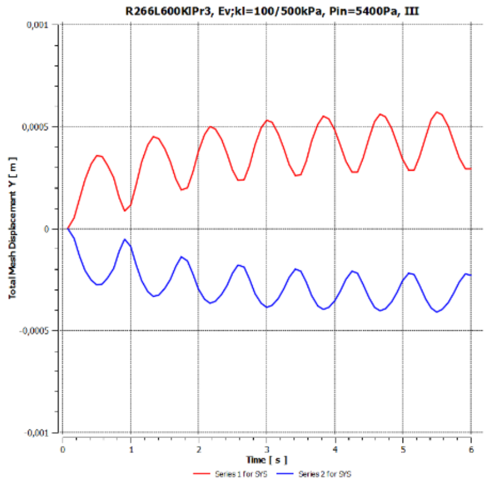
In the work frame of the previous theoretical setting, a pressure of 4.25 kPa is applied on the outside venous wall, Fig. 4b. It is commensurable with the average value of the 4.2 kPa of the assigned P_{inlet} but higher with 50 Pa (0.376 mmHg). Due to this small difference, this pressure could be also recognized as analogous to a pressure exercised by the muscles, or some other external reactions aiming to prevent venous damages in a trend of their permanent enlargement. Importantly, in the case commented above, the influence of the blood pressure along the z-axis is not essential, Figs. 4a, b. However, it is evident that an additional pressure of just 0.05 kPa coming from outside the wall causes the valve closure, Fig. 4d and may bring to its over-closure. Moreover, the emerging decrease of the blood flow velocity could be explained by this over-closure effect. The sensitivity of the system "blood flow – vein/valve" could be examined by variations of this external pressure P_{wall}. Decreasing P_{wall} from 4.25 kPa to 4.00 kPa results in an increasing distance between the valve leaflets, at the end of the closing phase. An increase with 300 Pa (2.25 mmHg) causes



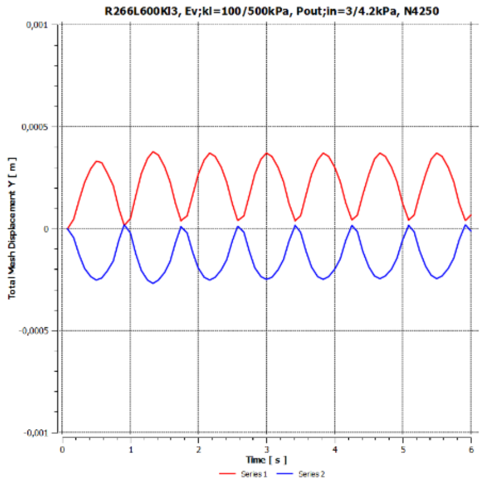
(a)



(b)



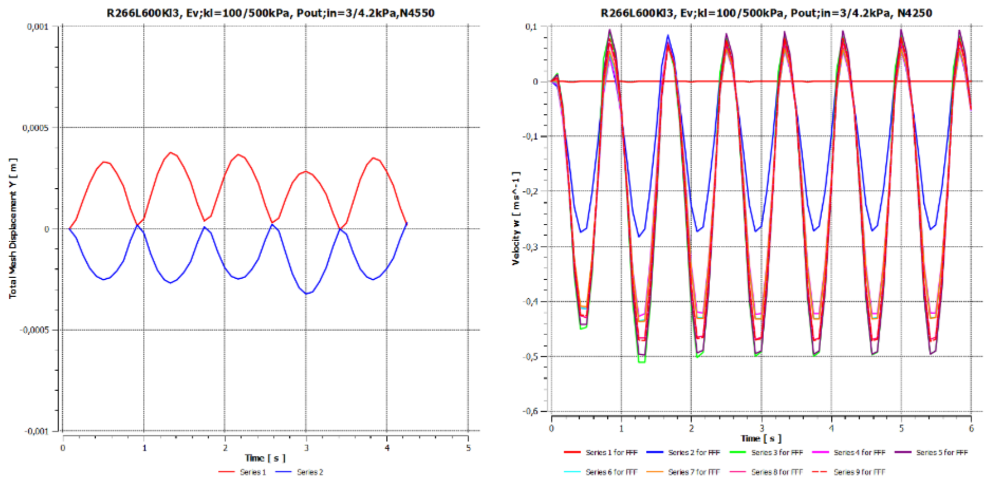
(c)



(d)

Fig. 4: Continued on the next page.

a considerable impact on the leaflets' stability, Fig. 4e. In Fig. 4f, the stably pulsating blood flow velocity Vz is about average of -0.2 m/s which results in decreasing reflux lower than 0.1 m/s. Stabilized maximal V_z-values from -0.58 m/s to -0.5 m/s and from -0.52 to -0.42 m/s are observed before and after the valve, respectively. Consequently, low uncompensated fluctuations in the external press values exercised on the vein could cause changes of the blood flow velocity, as well as changes of the venous

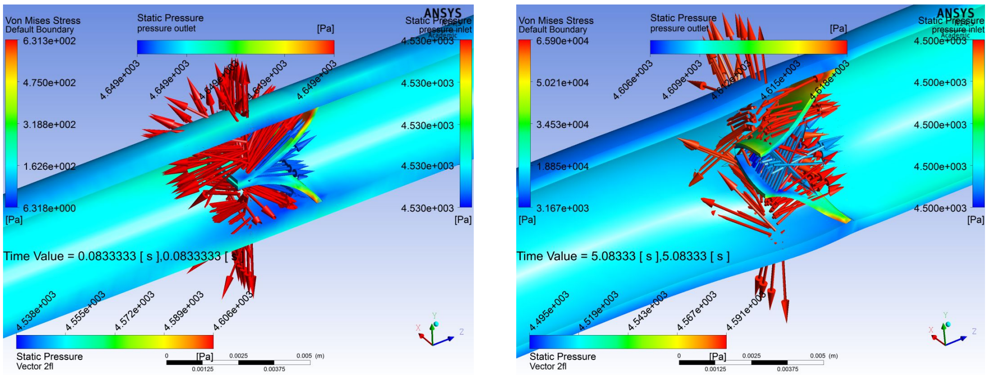


(e) (f)

Fig. 4: Parameters of FSI with $E_{\text{v;kl}} = 100/500$, pulsatile pressures $P^{\text{const}}_{\text{inlet}} = 4.2$ k + $P^{\text{ampl}}_{\text{inlet}} = \pm 0.6$ k and $P^{\text{const}}_{\text{outlet}} = 3$ kPa + $P^{\text{ampl}}_{\text{outlet}} = \pm 3$ k: blood pressure under: (a) $P_{\text{wall}} = 0$ Pa and (b) $P_{\text{wall}} = 4250$ Pa; displacements of the valve leaflets under: (c) $P_{\text{wall}} = 0$; (d) $P_{\text{wall}} = 4250$ Pa and (e) $P_{\text{wall}} = 4550$ Pa; blood velocity V_z; (f) under $P_{\text{wall}} = 4250$ Pa.

valve functioning during its closure. This effect is evident at Fig. 5a, where in case of $E_{\text{v}} = 100$ kPa and outside $P_{\text{wall}} = 4250$ Pa, the venous enlargement is not observed as it is seen in Fig. 5b. Also, in Fig. 5 (a and b), the compared Mises stresses, and the directions of pressure vectors upon the leaflets, numerically obtained with $E_{\text{v}} = 100$ kPa and 30 kPa, demonstrate the effects of the different venous elasticity under equivalent geometrical scale of the initially undeformed venous/valve forms. It could be concluded that, for the different combinations of venous/valve elasticity and blood flow parameters, there may exist an "opening/closing difference" between the three pressures (discussed above), to which the system could be considered as the optimal one resulting due to the change of the elasticities (remodeling), and/or external press on the vein, caused by the surrounding tissue.

The underscored changes of the elastic properties of the vein and the valve could be recognized as consequent from the blood flow values, and also consequent from the wall shear stress (WSS) initiated on the venous wall by the blood velocity, in addition to the responses of the corresponding venous and valve structures. These dependencies have complex biochemical and physiological origin. They were not commented in this investigation, but they could be subjects of next works based on the data here reported.



(a) (b)

Fig. 5: Deformed shapes (in equal scales), Mises stress and vector of blood pressure on the valve leaflets at the end of closing phase at $t = 5.0833$ s with: (a) $E_{\text{v}} = 100$ kPa, pulsating P_{inlet} and P_{outlet} and $P_{\text{wall}} = 4250$ Pa const $P_{\text{inlet}} = 4500$ Pa and with (b) $E_{\text{v}} = 30$ kPa, const $P_{\text{inlet}} = 4500$ Pa.

4 Conclusions

A numerical model of the sapheno-femural venous valve function is developed, Fig. 1. Using this model, distributions of the parameters of venous blood flow in reference to interactions with elastic vein and valve are established, Figs. 2a, b, e, f and Fig. 3, under boundary conditions with constant and pulsating blood pressures at the inlet/outlet of the vein, Table 1. Pressures' variability before and after the valve, that open and close it depending on $E_{\text{v}} = 100$ kPa and 30 kPa at $E_{\text{kl}} = 500$ kPa, are established. The displacements of the valve leaflets, Figs. 2c, d and Fig. 3 are also established. Opening and closing pressure differences necessary for the valve function are defined. Stable working regimes of elastic venous/valve deformation and corresponding changes of the venous/valve forms are established for the vein with $E_{\text{v}} = 100$ kPa, Figs. 2a, 3a, 4a, b and 5a, and unstable working-deformation regimes for the vein with $E_{\text{v}} = 30$ kPa, Figs. 2b, 3b, 5b. The importance of additional external pressure, applied to the venous wall, for the effective valve movement is established in a set with opening and closing working regimes of the valve, Figs. 4b, 4d, as well as regime impacts on the valve leaflets, Fig. 4e, are commented. Comparisons of Mises stresses and vectors of the pressures on the valve are given at stable and unstable venous deformation, Fig. 5.

Based on the results obtained, a hypothesis is made about the availability of optimal combinations between the treated parameters of the blood flow and the venous/valve materials' characteristics. These combinations are concluded as an elastically deformable system "blood flow– vein–valve" in a trend of the vein remodeling to keep optimal conditions for the blood transportation. The disturbance of the ranges of optimal combinations can cause nonreversible deformations of the vein and the valve, as well as their damages, corresponding to clinically observed vein disease pictures. Finally, we believe that the wall shear stresses, not commented here but achieved by the developed model application, play an essential role in keeping the optimal regimes for a venous valve proper functioning.

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