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The analysis of heart valve dysfunction and effectiveness of disc-designed prostheses

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The function of heart (in good condition and in poor condition) and hemodynamic characteristics of mechanical prosthetic cardiac valves influencing hemolysis and blood coagulation have been investigated. The solid-state models of disc-designed prosthetic cardiac valves have been constructed. The finite element analysis of blood velocity and pressure at direct and inverse blood flows has shown better characteristics of bi-leaflet valves. The advantage of the numerical modelling is in the possibility of thorough description of blood flow with determination of turbulence and depression zones for different sizes, curvature and ultimate opening angle of leafs. It allows minimising hydraulic resistance, thrombus formation and hemolysis-preserving high reliability of valve closing. The individual design of valve is discussed and aimed at improving a valve function by creating the blood flow twisting.

Key words: valve function of heart, disc-designed prosthetic cardiac valves, hemodynamics, turbulence, regurgitation, hemolysis, thrombus formation, finite element method

1. Introduction

Surgical treatment of heart diseases still remains one of the most urgent problems in cardiology. Most patients need replacement of affected cardiac valves by prostheses. Especially the parameters of blood flow at the dysfunction of the cardiac valve and solution of similar problem in the analysis of the efficiency of existing constructions of prosthetic cardiac valves are of interest. The quality of this device functioning is ensured by the decrease in hydraulic resistance, reduction of turbulence and equalization of pressure for exception of hemolysis and blood coagulation. The use of anticoagulants in order to prevent thrombus formation is considered to be negative since blood coagulation is a defence-adaptive response of the organism directed to urgent hemostasis and holding blood in vessels. Therefore, in order to decrease thrombogenicity and hemolysis, further improvement of progressive disk valve construction should be achieved taking account of hemodynamic quality criteria. The comparison of estimated characteristics of mono- and bi-leaflet valves is of interest. Moreover, pressure and velocity distributions at a closed valve are not investigated sufficiently. The present study deals with these problems.

2. Biomechanical analysis of aortic valve dysfunction

In systolic phase, blood from atrium flows through open mitral valve and then reaches left ventricle. At the contraction of the left ventricle, when mitral valve is closed, blood is shunted into aorta through open aortic valve. So, the role of the aortic valve (AV) is to ensure a directed blood flow into aorta. The defect of AV resulting in regurgitation, i.e. inverse blood flow in systolic phase, may be quantitatively characterised as the ratio of the area of valve lumen in diastole to the area of its cross section in a fully open state. Statistically, the duration of diastole is longer than that of systole by ca. four times. Due to the dysfunction of the valve there is a shunt of blood into nonpressure capacity of heart. This reduces the amount of both blood and oxygen supplied to the organism at one heart beat. The necessity of covering this obvious deficiency of oxygen causes the increase in pulse rate resulting in growth of heart load and reduction of its resource.

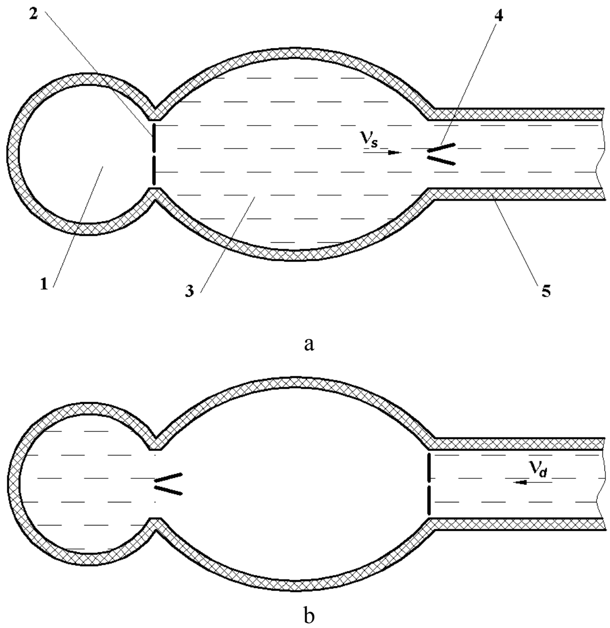


Fig. 1. Scheme of blood flow in systole (a) and diastole (b):
1 – left atrium; 2 – mitral valve; 3 – left ventricle; 4 – aortic valve; 5 – aorta

The volume of the inverse blood flow at diastole in the presence of defect of the valve can be determined by the formula

$$V_d = 60\, q \frac{k_{ds}}{N_0\left(k_{ds}+1\right)}, \tag{1}$$

where: q – intensity of blood flow; k_{ds} – relation of the periods of diastole and systole; N_0 – pulse rate in good condition.

The intensity of blood flow may be estimated as follows

$$q = 1.05 \cdot 10^{-3} k_d \left(\frac{p_{\max} + p_{\min}}{2}\right)^{0.5} \frac{d^2}{\sqrt{\rho}}, \tag{2}$$

where: k_d – degree of dysfunction of the valve, dm^3; $p_{\max}$, $p_{\min}$ – max and min limits of the change of pressure, mm Hg; d – diameter of the valve; ρ – density of blood (1.05 g/cm^3).

Hence, the expression for pulse rate, taking into account the valve dysfunction, can be expressed by

$$N = \frac{N_0 V_s}{V_s - V_d}, \tag{3}$$

where: V_s – systolic volume per minute; V_d – volume of blood being regurgitated per minute.

With the formulas (1)–(3) it is possible to estimate the influence of the dysfunction of the valve on heart rate under three typical conditions of blood-circulation system (see table 1).

Table 1. Characteristics of heart rate,
depending on the condition of blood-circulation system

Parameters	Systolic volume, dm^3	Pressure, mm Hg		Pulse rate
		Systole	Diastole	
Physiological norm	0.07	120	100	60
Aplastic anaemia	0.11	90	45	100
Pre-uraemia	0.03	240	145	70

Figure 2 represents the dependencies of pulse rate on the degree of AV dysfunction for valve section diameter of 20 mm. It is evident that at pathology of circulatory system the valve function is more essential factor compared with physiological norm.

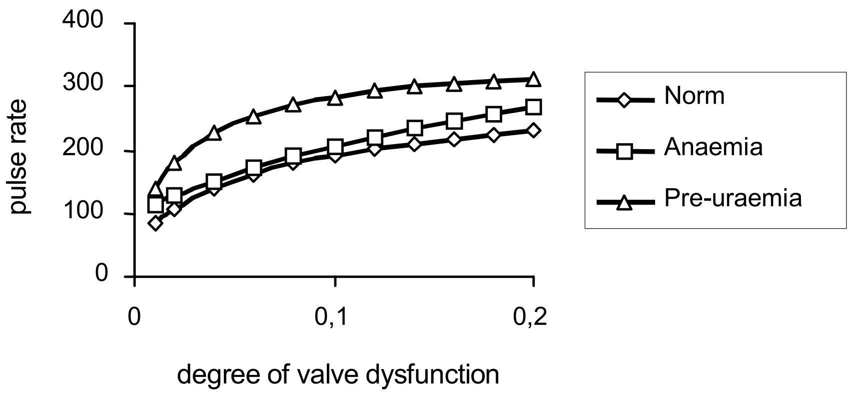


Fig. 2. Dependence of pulse rate on the degree of valve dysfunction

3. Analysis of the effectiveness of disc-designed prosthetic cardiac valves

The problem of adequacy of prosthetic cardiac valves from the viewpoint of biomechanician is considered. As is widely known, evolutionary «optimised» native cardiac valves consist of soft biotissues. A significant durability of the biological valve under conditions of cyclic bend with heart rate is caused by extremely high fatigue strength of biotissue associated with elimination of microdamages of the material in result of directed genesis of connecting tissue.

As the opportunity of reproducing similar adaptive reaction in artificial materials is problematic at present [1], the technical decisions on ball- or disk-locking element made of constructional materials: metals (steel, titanium), ceramics or glassceramics, are offered. Thus, the necessary parameters of reliability and resource of implants are reached due to high strength, hardness and wear resistance of the material.

The research, which had been made earlier, allowed us to compare hemodynamic features of valves of different designs, to optimise the form of locking elements, to evaluate influence of implantation method on function of the valve. In particular, the advantages of mechanical prostheses with one- and especially two-disk removable locking element (figure 3) were revealed. Such prostheses have been widely used in clinical practice due to low (7–11 mm) profile, relatively large angle of opening and high reliability [1], [2]. Naturally, the effective area of opening of monoleaflet valves of disk is by 20% larger than that of bioprosthesis and by 30–35% larger than in ball prosthesis [1].

However, disc-design prostheses have some disadvantages: 1) failure of erythrocytes (hemolysis) due to higher stress level than in native valve and 2) a non-uniform division of the hydraulic canal by the disk inside causes unwanted turbulence of blood flow which is considered to be the reason of emboli formation [2]. Hence, there arises the necessity of determining and controlling the turbulence, depressuring high pressure and localizing the zones of shear (Reynolds's) stresses. Evaluation of pressure distribution is necessary for improvement of AVH effectiveness.

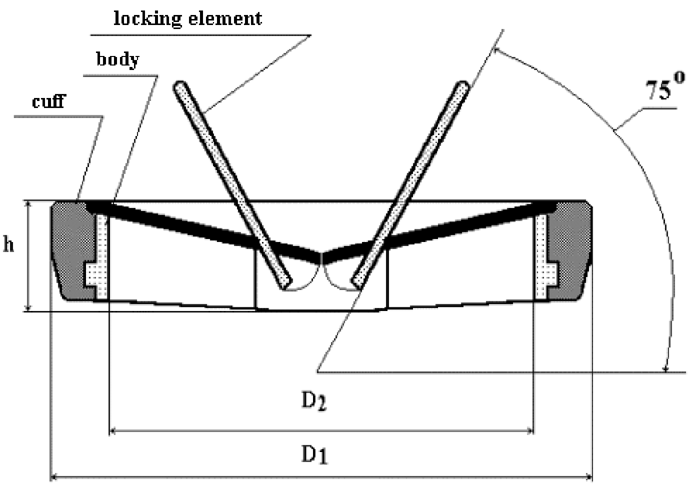


Fig. 3. Bi-leaflet valve of «Planics» type

The locking elements of valve (leafs), which constructively does not open fully for guaranteed return to the closed position, create local resistance, due to which the pressure before the valve is higher than behind it. Losses of pressure at overcoming local resistance of the prosthetic valve are considered to be negative, since they reduce throughput efficiency of blood circulation and cause an increase in pulse rate.

On the other hand, the rare but still dangerous response of the disk-design prosthesis is due to the locking element of the defective valve. The defect may be caused by extremely large critical angle of opening, at which the inverse blood flow appears insufficient for overcoming the moment of resistance to closing.

Thus, it is necessary to find a trade-off, namely, a certain critical angle of opening α, at which optimal size of turbulence zone, minimal pressures and shear stresses, low resistance to blood flow in systole and minimal regurgitation in dyastole, and reliable closing of the valve at inverse blood flow are ensured simultaneously. For definition of this optimum, two conditions should be fulfiled:

$$\begin{gathered} \text{Min } p(\alpha), \tau(\alpha); \\ \text{Max } (V_s(\alpha) - V_d(\alpha)); \\ S_{\text{turb}}(\alpha) = S_{\text{turb}}^{\text{opt}}; \\ M_b(\alpha) = M_f(\alpha), \end{gathered} \tag{4}$$

where: p – local blood pressure; τ – shear stress; S_{turb} – size of turbulence zone; $S_{\mathrm{turb}}^{\mathrm{opt}}$ – optimal size of turbulence zone; α – angle of critical opening of the valve; M_b – moment caused by distribution of blood pressure at reverse blood flow; M_f – moment of resistance to turn of the locking element.

The losses of pressure Δp (mm Hg) caused by valve installation can be estimated based on the formula which takes into account the relation of pressure and flow rate in accordance with approach used in hydrosystem design [3]

$$\Delta p = 20440 \, \frac{\rho}{g^2} \cdot v^2 \left(1 - \frac{S}{S_\alpha}\right)^2 , \tag{5}$$

where: S – sectional area of fully opened valve; $S_\alpha = S(1 - \cos\alpha)$ – sectional area of partially opened valve; v – linear speed of flow through the valve defined by formula

$$v = \left[\frac{V_s \cdot N \cdot (1 + k_{ds})}{S}\right]. \tag{6}$$

The above dependencies allow estimation of AV pressure gradient (figure 4). For example, for 65° angle the losses Δp are equal to 5.8 mm Hg.

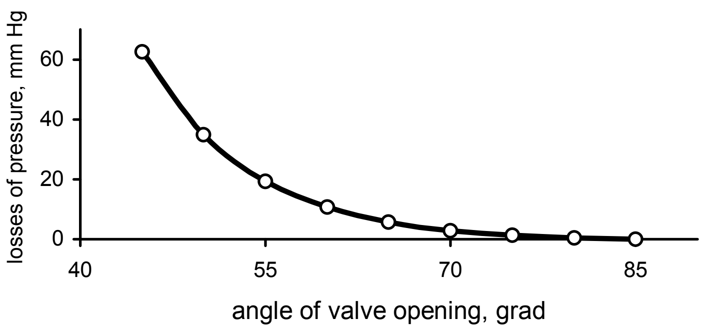


Fig. 4. Dependence of losses of pressure on the angle of valve opening

A more complicated task is to define the pressure and velocity distributions. Fields of velocities and Reynolds's stresses for disc-designed PCV was investigated experimentally by particle method [4]; the method of photochromic visualisation of current line was also applied in the velocity detection [5]. Theoretical analysis is based on computational modelling of the process [2], [6], [7].

In the present study, for the description of the process of blood flow in systolic and diastolic phases, the finite-element solution with implementation of ANSYS computer code is considered to be effective [8]. The advantage of the approach used is the opportunity to describe in detail blood circulation revealing the zones of turbulence and depressure. This allows us to optimise the features of PCV, providing minimal hydraulic resistance and thrombus formation and also high reliability of operation at inverse blood flow.

For optimisation of hemodynamics of disc-designed PCV «Planics» [9], the angle α of leafs' opening was varied (figure 3).

It should be stressed that the 3-D description of blood flow as of non-Newtonian liquid with a possible turbulence causes significant computational complexity. The possibility of more economic estimation of flow in two-dimensional design (slit canal model) is based on the flow symmetry for the construction being described. In this case, the difference in the results for 2-D and 3-D models is not essential (according to our estimations for maximal velocities the error does not exceed 10%). Besides, in *2*-D estimation the interpretation of graphic results is significantly simple.

The solid-state model of blood channel fragment with valve (figure 1) was constructed by means of automated designing Solid Works software [5]. In this connection, we accept the assumption that aorta is not deformed (actually, the increase of aorta root diameter takes place [2]) in systole and promotes the decrease of hydraulic resistance.

4. Numerical results

The series of calculations included definition of fields of pressures and velocities in the field of mono- and bi-leaflet aortal valve.

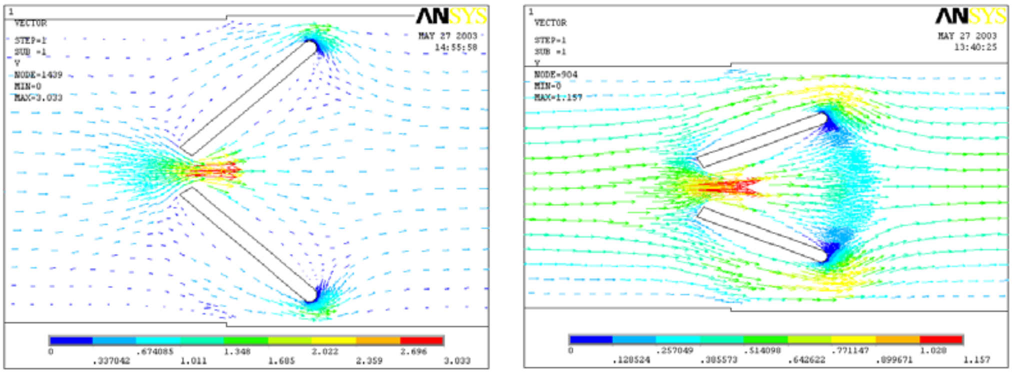
Table 2. Non-dimensioned extreme values of flow parameters, depending on opening angle of valve

Opening angle α, deg	Maximal flow velocity v_{max}	Pressure p	
		min	max
35	8.35 / 8.58	3.00 / 8.37	15.06 / 14.72
40	6.06 / 6.22	1.93 / 5.25	8.88 / 8.68
45	4.39 / 4.51	1.82 / 3.35	5.15 / 5.09
50	3.21 / 3.29	2.16 / 1.82	3.39 / 3.54
55	2.38 / 2.44	1.97 / 1.51	2.34 / 2.48
60	1.85 / 1.90	1.48 / 1.46	1.76 / 1.82
65	1.51 / 1.55	1.23 / 1.36	1.40 / 1.42
70	1.22 / 1.26	1.03 / 1.29	1.16 / 1.16
75	1 / 1.03	1 / 1.07	1 / 1.02
80	0.84 / 0.86	0.95 / 0.91	0.91 / 0.93

It was presumed that in the aperture of the valve with the channel size of 16 mm the laminar stream with the velocity v_s = 0.42 m/s was set which corresponds to Reynolds number Re = 1380. As varying parameter the angle of valve opening α was taken. The results of calculation in the type of non-dimensional pressures and velocities (as referred to the values of the opening angle 75°) are presented in table 2 for the phases of opening (in numerator) and closing (in denominator) of the valve.

In general, the blood flow at closed AV is laminar. In such circumstances, the extreme pressures p^{max}, p^{min} and flow velocities v^{max} increase nonlinearly. It has been established that the increase in the value of critical angle of AV opening to more than 75° does not lead to visible change in hemodynamic parameters, including hemodynamic resistance. It coincides with the above analytical evaluation (figure 4) and correlates with experimental data equal to 75° applied in «Planics» (figure 3).

In particular, figures 5 and 6 represent the pressure and velocity distributions for two values of the opening angle. At the moment prior to closing the AVH, analogous distributions at inverse blood flow and the velocity v_d = 0.42 m/s are shown in figure 7.



a b

Fig. 5. Fields of velocities of blood flow for the angle of valve opening of 50° (a) and 70° (b)

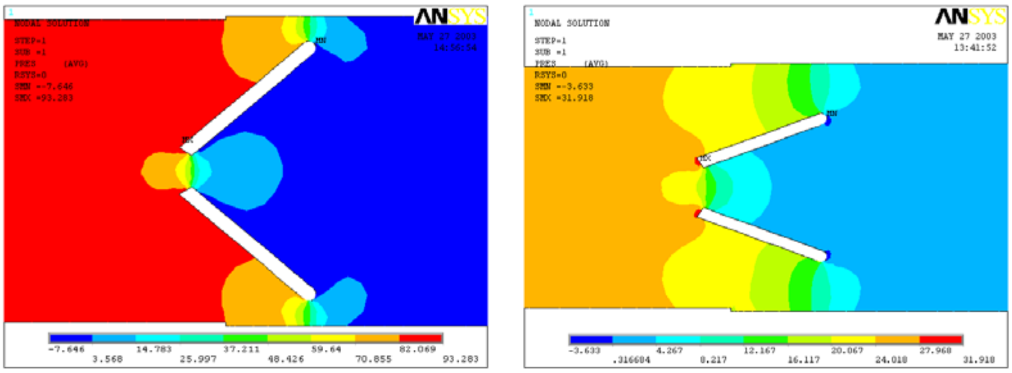


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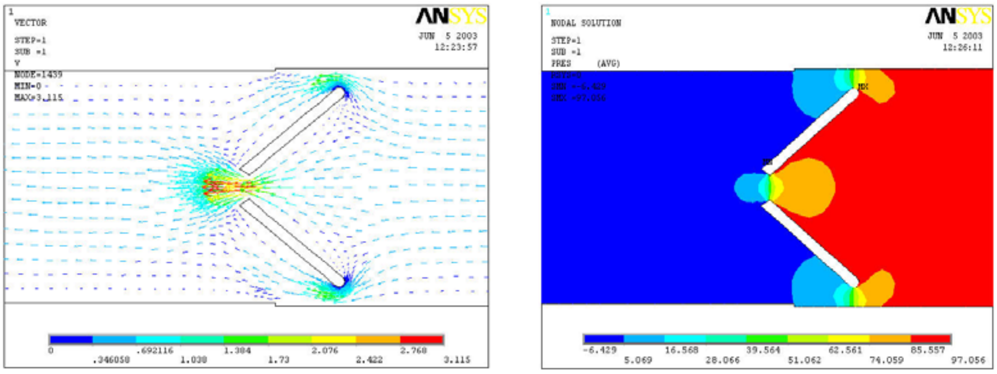
Fig. 6. Fields of pressures of blood flow for the angle of valve opening of 50° (a) and 70° (b)



a b

Fig. 7. Fields of velocities (a) and pressures (b) at inverse blood flow for the angle of valve opening of 50°

Contrary to monoleaflet AVH, there is symmetry of distribution of velocities concerning the axis of flow for two-leaflet ones. At a large opening angle (α = 60°), the local turbulence appears. In accordance to vector diagram of velocities, the inverse current and flow turbulence are detected close to leafs (figure 6).

According to figure 5 the field of pressures is characterised by heterogeneity. The maximal depression is noticed at the beginning of the valve opening (α = 50°) compared to vortex flows occurring at large angles of its opening. This is in agreement with the results reported in [4], where the influence of the leaf edge modification on hemodynamics at closed valve was described and a rapid development of negative pressure on atrial side of the leaf edge was shown. The two zones of high pressure are localised at the edges of leafs from the side of the incident flow, and the zone of negative pressure (depression) adjoins rear edges of leafs. It is possible to notice a significant difference in the velocities in central and mural fields of flow, which is the factor of blood hemolysis due to shear rupture of erythrocytes. The other reason for hemolysis and the wear of valve elements is the cavitation. It can occur at the beginning of the valve opening in the zones of maximal pressure. The results of computational modelling of field of velocities correlate with the experimental data obtained by means of particles method [4] and photochromic visualisation [5]. Moreover, the model designed allows determining the distribution of pressures which is difficult in experiment. The advantage of the present method is the possibility of describing in detail the blood flow with detecting the turbulence and depression zones for different variants of PCV construction, including size, curvature and critical angle of opening of leafs. This allows minimising hydraulic resistance, thrombus formation and hemocoagulation preserving high reliability of the valve closing. Using 3-D modelling it is possible to estimate such a configuration of the prosthesis at which twisting of the blood flow, ensuring reduction of hemodynamic resistance, is provided.

5. Conclusion

The designed computer model of hemodynamics of disc-designed aortic valve prostheses allows minimising the hydraulic resistance, hemolysis and blood coagulation, preserving high reliability of valve closing. The individual design of valve is possible, creating twisting of blood flow to improve the valve function.

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