Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

Dynamics of mechanical heart valve prostheses

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Abstract

A numerical code for analyzing the dynamic behaviour of heart valve prostheses is presented. The study is conducted with the valve in a simplified cardiovascular system, represented schematically by a circuit with a pump, the valve under test, a rugged elastic tube and a capacity. The pump represents the left ventricle, while the tube and the capacity represent the arterial system. The tube, unlike the other components, presents propagative effects. Input includes the physiological data and valve type, geometry and masses, while time-dependent output include fluid pressure and velocity in the circuit and downstream of the valve, valve opening time and the angular speed and acceleration of moving parts.

1 Introduction

Mechanical aortic and mitral valve prostheses are designed and constructed to reproduce natural heart valves as faithfully as possible. Such prostheses consist essentially of a rigid ring carrying one or two mobile members which alternately permit and prevent the passage of blood. From the mechanical standpoint, these valves are passive members and can be regarded as check valves which are subject in operation to the action of the surrounding fluid. In particular, it has been observed that valve closing is determined by blood backflow, and takes place in a way which does not adequately reproduce the phenomenon of early partial closing typical of natural valves. Kang [1] and Van Steenhoven [2] have shown that most mechanical aortic prostheses now on the market close only 5% of the passage section during systole, a value significantly below that of the natural aortic valve, which is estimated at around 74%. This means increased blood backflow, larger swings in pressure after the valve closes, and a less gradual closing process, leading potentially to cavitation (Kepletko [3]).

The many published studies of heart valves have used both experimental and theoretical-numerical approaches. For the static case, for example, contributions include those of Idelsohn, Costa & Ponso [4], Milton Swanson & Clark[5], Belforte, Raparelli & Romiti [6], Handle, Harrison, Yoganathan, Allen & Corcoran [7], Hansekam, Westphal, Nygaard, Reul, Giersiepen & Stodkilde-Jorgensen [8]. For the dynamic case, which better reflects actual operating conditions, the literature is more meager (Peskin [9], Graf, Fisher, Reul & Ran [10]).

This paper presents a numerical calculation program for analyzing the dynamic behaviour of heart valve prostheses. The study is conducted with the valve in a simplified cardiovascular system, represented schematically by a circuit with a pump, the valve under test, an elastic tube and a capacity. The pump represents the left ventricle, while the tube and the capacity represent the arterial system.

2 System model

The circuit layout used for the theoretical study of the valve is illustrated in Figure 1. It includes a pump V simulating the left ventricle of the heart, a check valve V_a simulating the aortic valve, an elastic tube of length l, and a capacity C_a. The latter two items simulate the arterial system.

Pump. This is an ideal component which generates a pressure similar to that of the left ventricle. In Figure 2, the solid line represents effective pressure, while the dash line represents the pressure obtained by approximating the actual curve with a 6th degree polynomial and straight sections.

Heart valve. The recent "mobile bileaflet" design shown in Figure 3a was selected from amongst the various types of artificial valve now on the market. Valve passage area a depends on leaflet position θ (Figure 3b) as follows:

$$a = A(1 - \cos\theta) \tag{1}$$

where A is the maximum passage cross-section, viz. that of the aortic tube.
The equation for flow rate through the valve is:

$$Q = C_d a \sqrt{(2\Delta p/\rho)} \tag{2}$$

where C_d is the outflow coefficient, Δp is the pressure drop across the valve, and ρ is blood density. In the case of reverse flow (*backflow*), the outflow coefficient was reduced by 1/4 given the different geometry.

With J as the leaflet's moment of inertia relative to the rotational axis and y_c as the center of thrust (which coincides with the center of gravity if pressure is uniform over the area), the leaflet's equation of motion is:

$$(p_0 - p_1)\frac{A}{2} y_c \cos(\theta) = J\ddot{\theta} \tag{3}$$

where p_0 and p_1 represent the pressures at the valve/ventricle and valve/aorta interfaces respectively.

Aortic tube. Assuming non-compressible, homogeneous and single-dimensional flow, the state variables pressure p and velocity u, which are functions of time t and the axial coordinate x along the tube, are linked by the motion equation (4) and the continuity equation (5):

$$\frac{\partial u}{\partial t}+u\cdot\frac{\partial u}{\partial x}+\frac{1}{\rho}\cdot\frac{\partial p}{\partial x}+f\frac{u|u|}{4r}=0 \tag{4}$$

$$\frac{\partial (Au)}{\partial x}+\frac{\partial A}{\partial t}=0 \tag{5}$$

where $A=\pi\cdot r^2$ is tube cross-section, r is tube radius and f is the friction coefficient. Cross-section A depends on the transmural pressure (the difference in pressure between vessel interior and exterior) in accordance with the following algebraic relation provided by Fung [11]:

$$r=r_0+\frac{\alpha}{2}(p-p_0) \tag{6}$$

where r_0 is the radius of the vessel when $p=p_0$ and α is the vessel's compliance (m/Pa), which is assumed to be constant.
Given (6), (5) becomes:

$$\frac{\partial u}{\partial x}+\frac{\alpha}{r}\frac{\partial p}{\partial t}+\frac{\alpha}{r}u\frac{\partial p}{\partial x}=0 \tag{7}$$

Partial differential equations (4) and (7) can be formulated as ordinary differential equations. Multiplying (7) by an appropriate coefficient and summing it with (4) gives two ordinary differential equations:

$$\frac{du}{dt}\pm\frac{1}{\rho C}\frac{dp}{dt}\pm\frac{fu|u|}{4r}=0 \tag{8}$$

These equations are valid along the lines defined by

$$\frac{dx}{dt}=u\pm C \tag{9}$$

where $C=\sqrt{r/\rho\alpha}$. This value represents the propagation speed of the small perturbations in the tube. Expressions (8) are called *characteristic equations* and are indicated with C^+ or C^{--}, depending on whether a + or - sign is selected.
Capacity. This is an ideal element capable of maintaining constant internal pressure.

3 Numerical simulation

Circuit simulation entails discretizing all the equations describing the components. The space-time integration plane is also discretized (Figure 4). The aortic tube is divided into n elements of equal length Δx, according to the desired spatial definition. Two characteristic lines whose slope is given by equations (9) depart from each point of the tube. Two of the characteristic lines converge at instant $j+1$ in a *nodal point* where both equations (8) are valid. It is thus possible to calculate the state variables at instant $j+1$ once those at instant j are known. The values at instant j are calculated by linear interpolation between the values for adjacent nodal points. The time increment is Δt. In the sections delimiting the tube (valve interface section *1* and capacity interface section *n*), there is only one characteristic, i.e. C^- in section *1* and C^+ in section *n*. To calculate the state variables in these sections, it is thus necessary to define the *boundary conditions*. In section *n*, given that the reservoir guarantees a constant pressure, the boundary condition is defined by the value of that pressure. In section *1*, the boundary condition is defined by equation (10), which is obtained by equalizing the flow through the valve [equation (2)] with the flow entering section *1*, and from characteristic equation C^- [equation (8) as indicated above]. Equation (10) is:

$$u_1 = c_d(1 - \cos\theta)\sqrt{\frac{2(p_o - p_1)}{\rho}} \qquad (10)$$

Motion equation (3) gives angular acceleration $\ddot{\theta}$ and, with successive integrations, angular velocity $\dot{\theta}$ and angular position θ.

4 Calculation program

The flow-chart for the program is shown in Figure 5. The program is written in GWBASIC and runs on personal computer. The time-dependent output which can be calculated with this program include:

1) Pressure immediately downstream of valve.
2) Flow velocity through valve.
3) Extent of valve opening.
4) Angular velocity of mobile leaflet.
5) Angular acceleration of mobile leaflet.

Input data include:

1) Maximum cycle duration.
2) Valve outflow coefficient.
3) Leaflet radius.
4) Leaflet angle, velocity and angular acceleration at instant t=0.
5) Fluid density.
6) Friction coefficient.
7) Initial pressure and velocity.
8) Tube length.
9) Vessel compliance.
10) Tube discretization.
11) Integration interval Δt.

The large amount of input data which must be entered from keyboard was necessary in order to achieve a certain flexibility for the program.

5 Results

The study was organized into two stages.

In the first stage, the physiological data were established and tests were carried out to determine the magnitudes of the simplified cardiovascular circuit which make it equivalent to the actual circuit. Figures 6, 7, 8, 9 and 10 show pressure downstream of the valve, flow velocity downstream of the valve, opening angle, leaflet angular velocity and leaflet angular acceleration. The data used to obtain these graphs are: maximum time = 0.8 s, outflow coefficient = 0.4, leaflet radius = 0.0125 m, leaflet density (assumed to be uniform) = 1500 kg/m^3, initial angle, angular velocity and angular acceleration = 0, fluid density = 1050 Kg/m^3, friction coefficient = 0.05, initial pressure = 11304 Pa, initial flow velocity = 0, tube length = 0.11 m, and vessel compliance = 4.762 E-7 m/Pa.

These data resulted from the study conducted in the first stage, and reflect a physiologically normal situation.

In the second stage, a number of the problem's input parameters were varied, to determine their influence on the valve's dynamic behaviour.

By way of example, Figures 11, 12, 13, 14 and 15 respectively show curves for valve downstream pressure, position, velocity and angular acceleration for ventricular pressures of 1.5 times (solid line curves) and 0.8 times (dash-line curves) the reference pressure. These two pressure levels were used to simulate a heartbeat as produced under physical exertion, and one in a condition of rest.

For the case of exertion, more severe valve operating conditions are obtained. Curves are similar to those for normal conditions, though the peaks increase. Valve opening and closing take place earlier.

In the second case the peaks are lower, as are the dynamic stresses on the valve.

6 Conclusions

A numerical code for analyzing the dynamic behaviour of mechanical heart valve prostheses was developed and validated. The program runs on personal computer.

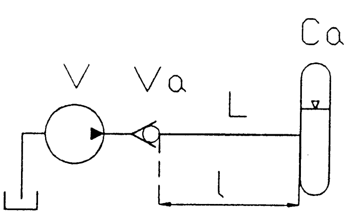

Figure 1:Circuit diagram.

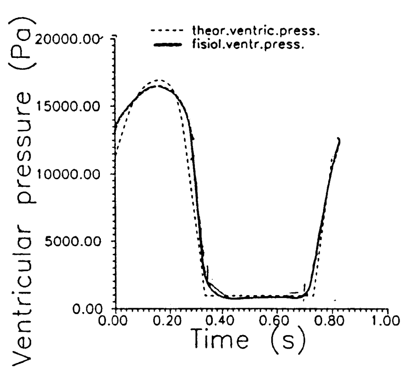

Figure2:Ventricular pressure.

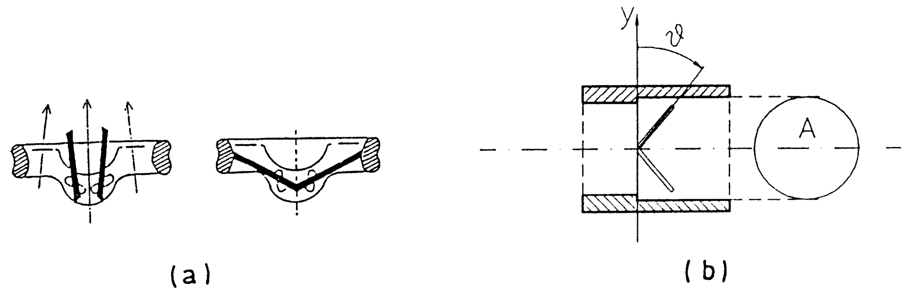

Figure 3:Valve schematics.

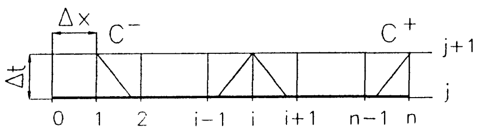

Figure 4:Integration plane.

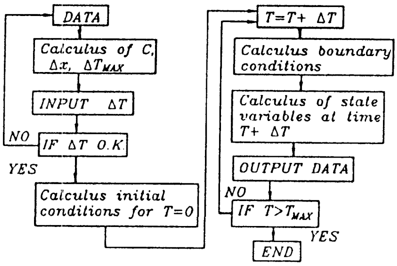

Figure 5:Flow Chart.

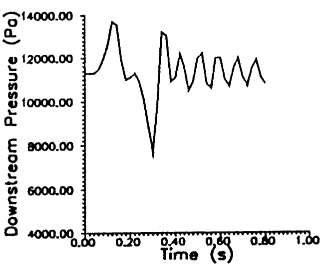

Figure 6:Downstream pressure.

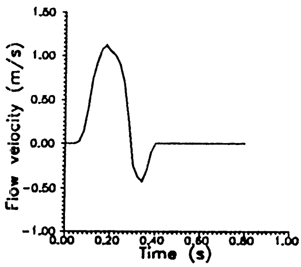

Figure 7:Flow velocity.

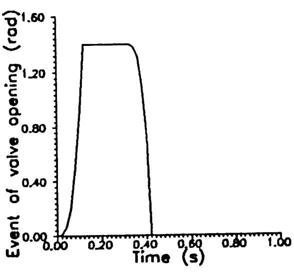

Figure 8:Extent of valve opening.

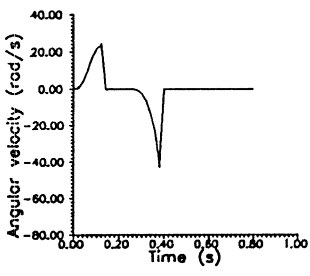

Figure 9:Angular velocity of mobile leaflet.

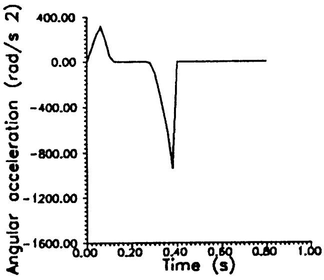

Figure 10:Angular acceleration of mobile leaflet.

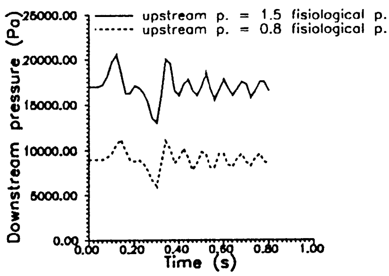

Figure 11:Downstream pressure.

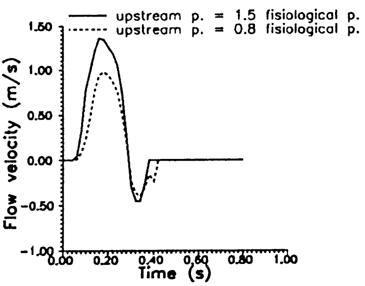

Figure 12:Flow velocity.

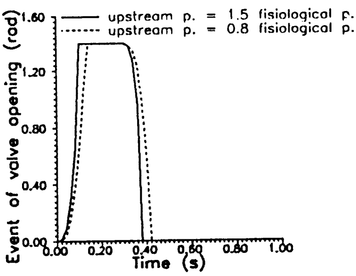

Figure 13:Extent of valve opening.

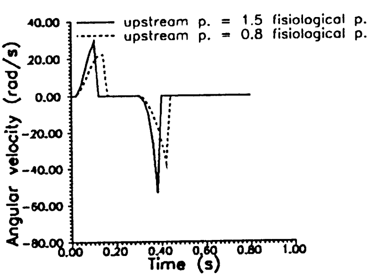

Figure 14:Angular velocity of mobile leaflet.

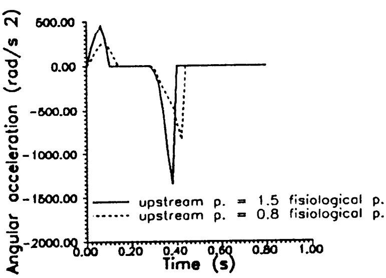

Figure 15:Angular acceleration of mobile leaflet.

The code provides effective support during valve design in as much as it makes it possible to perform parametric studies, as well as during valve implantation, as the performance of different valve types can thereby be compared.

The cardiovascular system developed as part of the investigation can be readily extended to permit the study of a more complex downstream circuit.

Acknowledgements
The authors would like to thank Maurizio Cerrone for his assistance.
The research described in this paper was made possible by MURST funding.

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