259. Research of artificial heart ventricle based on the principle of travelling waves driven by piezoelectric actuation

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Abstract. The present article deals with the piezoelectrically-induced travelling wave that arises in periodically loaded tube and it drives blood through tube intended for application as an artificial heart ventricle. However, frequencies of piezoelectrically actuated running waves lie in ultrasonic frequency band and such big frequencies cannot be used for the blood transfer in an artificial heart. Using finite element method the forms of oscillations in the tube of the artificial heart ventricle, natural frequencies and the characteristics of blood flow through the tube (rate of flow-through, flow and pressure) were calculated. Blood pressure and flow dependencies on different tube characteristics such as tube diameter, wall thickness and frequency and magnitude of systolic driving force were studied. The calculation method allows estimating rational parameters of the artificial heart ventricle.

Keywords: Resonance wave, piezoelectric actuation, artificial heart, valveless pumping, natural frequencies

Introduction

The main function of heart is pumping blood in one direction, since the heart valves ensure this unidirectional flow. Due to different cardiovascular diseases heart cannot perform its function properly and depending on the level of heart dysfunction there are many ways to deal with it: different pharmaceuticals, heart transplantation, heart assist devices or artificial heart, which exist of various constructions, for example pneumatic, peristaltic or rotary blood pumps [1, 2]. Building an efficient and reliable artificial heart is a very hard task because of the variety of different influential factors, complexity of mechanical device and strict requirements for it. It is advantageous to biological systems to sway toward the simpler mechanical systems that are easier to build and less susceptible to failure [3]. One of the fluid pumping methods requiring less complicated mechanical construction is valveless pumping, which basics comes from the ideas of cardiovascular dynamics of early embryonic life where the blood circulates in one direction in spite of the complete lack of valves [4, 5, 6]. Valveless pumping can be achieved by periodically compressing a tube and compression will produce a frequency-dependent net flow [7, 8]. Similar principle of the travelling wave is not new phenomenon and it is widely applied in different fields ranging from industrial to biomedical applications of fluid flow and control [9, 10]. There are many devices operation of which is based on the travelling wave principle, however they function with high frequencies (15 – 30 kHz). Such high frequencies cannot be used for the blood transfer in an artificial heart, since they can produce pathological changes in blood state.

Therefore, the main goals of the present article are estimating the rational parameters of the tube; determine its natural frequencies and explore possibilities reducing the frequencies of travelling waves up to $f \leq 1000$ Hz; using Navier-Stokes equation to investigate the changes in blood velocity, rate and pressure by varying tube geometrical parameters and excitation force amplitude and frequency.

Methods

Following assumptions have been made:

- Artificial heart ventricle is elastic, cylindrical and deformable body with outside diameter d = 20 mm, wall thickness δ = 0.1 mm and length L = 150 mm.
- Blood is Newtonian fluid, laminar and plane flow, constant viscosity μ = 0.0035 kg/ms, density ρ = 1050 kg/m³.
- Blood pressure at the tube inlet p = 10.67 kPa and at the outlet obtained velocity v = 0.318 m/s, pressure p = 16 kPa, rate Q = 5.98 l/min.

- Cylindrical finite elements were used for modeling the fluid flow and quadratic finite elements for tube model, nodes of both meshes where coincident.

Figure 1 shows simplified calculation scheme of meshed finite elements tube model.

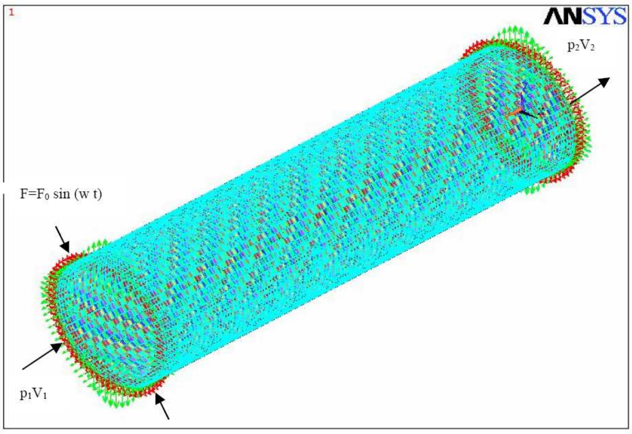

Fig. 1. Simplified calculation scheme of the tube model

Magnitude of the force was chosen in such a way that permanent deformations in tube walls will not be present and frequency of the force would coincide with the natural frequencies of the tube [11].

The blood flow was examined by solving Navier-Stokes equations for incompressible fluids:

$$\frac{du}{dt} + u\nabla u = -\frac{1}{\rho}\nabla p + \nu\nabla^2 u, \qquad (1)$$

and the equation of continuity:

$$\frac{du}{dt} + u\nabla u = 0, \qquad (2)$$

where $\nabla = \frac{\partial}{\partial x} i + \frac{\partial}{\partial y} j + \frac{\partial}{\partial z} k$, i, j, and k are the vectors, $u = (u, v, w)^T$ is velocity field of an element.

Excitation force is in the form of travelling waves which are produced by the deformation of tube walls:

$$F = A_f \, Re^{i(kz - t)}, \qquad (3)$$

where R is real part of complex number; A_f = 1 if the deformation of wall has longitudinal component, and A_f = 0, if the deformation of wall has transverse component.

The interaction of the fluid (blood) and the structure (tube) at a mesh interface causes the pressure to exert a force applied to the structure and the structural motions produce an effective fluid load. The governing finite element matrix equations then become:

$$\begin{cases} M_s\ddot{U} + K_s U - RP = F_s, \\ \rho_0 R^T \ddot{U} + M_f \ddot{P} + K_f P = F_f, \end{cases} \qquad (4)$$

were R is a coupling matrix that represents the effective surface area associated with each node on the fluid-structure interaction (Fig.2).

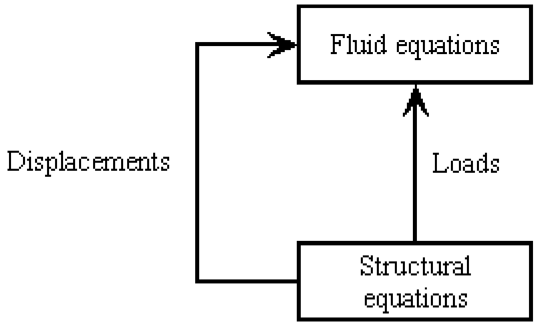

Fig. 2. Fluid-structure interaction

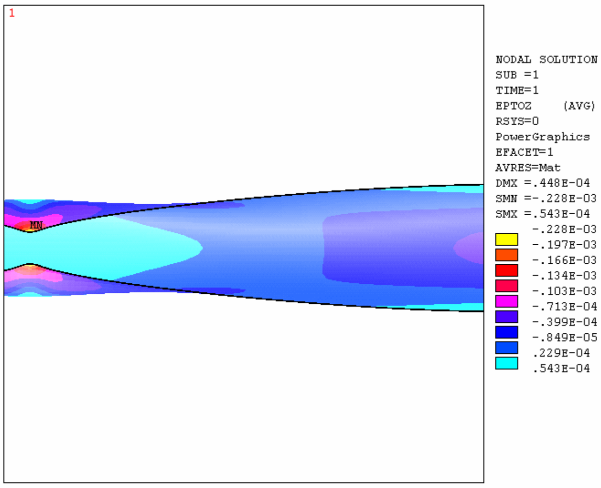

Fig. 3. Deformations of the artificial heart ventricle in Z direction

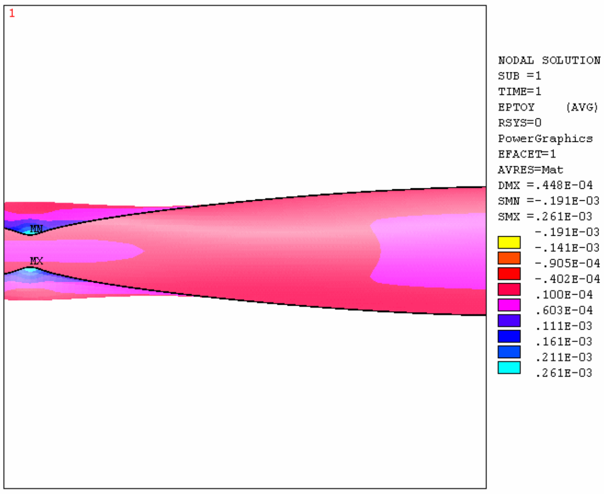

Fig. 4. Deformations of the artificial heart ventricle in Y direction

The changes of blood velocity, rate and pressure were obtained by varying length L = 150 ÷ 250 mm, wall thickness δ = 0.1 ÷ 0.3 mm, amplitude of excitation force A = 0.005 ÷ 0.2 mm.

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Results

From the diagram presented in Figure 5 it can be noted that blood rate is decreasing during the increase of the wall thickness and force staying of the same magnitude. When wall thickness equals to 0.1 mm, the excitation force must be F = 0.05 ÷ 0.1 N in order to produce blood rate Q = 5 ÷ 10 l/min. Consequently, when tube wall thickness equals to 0.2 mm, then F = 0.1 ÷ 0.5 N and when δ = 0.3 mm, F > 0.5 N. Therefore, the magnitude of excitation force should be increased in order to maintain desired blood rate during the change of wall thickness.

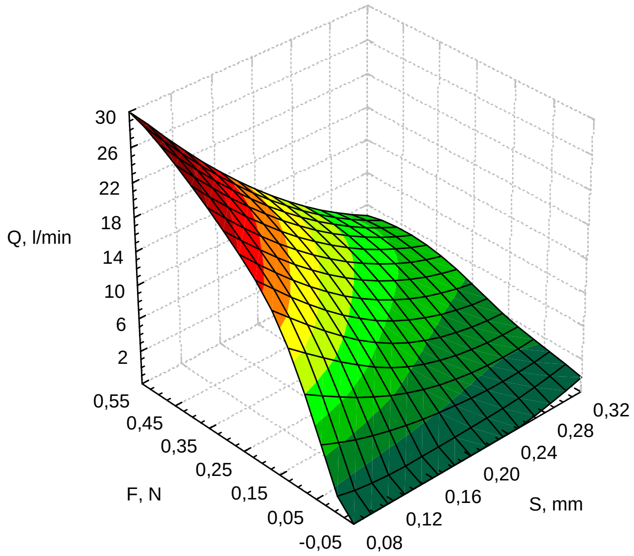

Fig. 5. Blood rate dependence on the tube wall thickness and magnitude of excitation force

Enlarged diameter of the tube increases amount of blood passing through it. When excitation force F = 0.1 N and tube diameter d = 20 mm, blood rate Q = 5,223 l/min, whereas increasing diameter up to d = 30 mm, blood rate increases up to Q = 14, 091 l/min (Fig. 6).

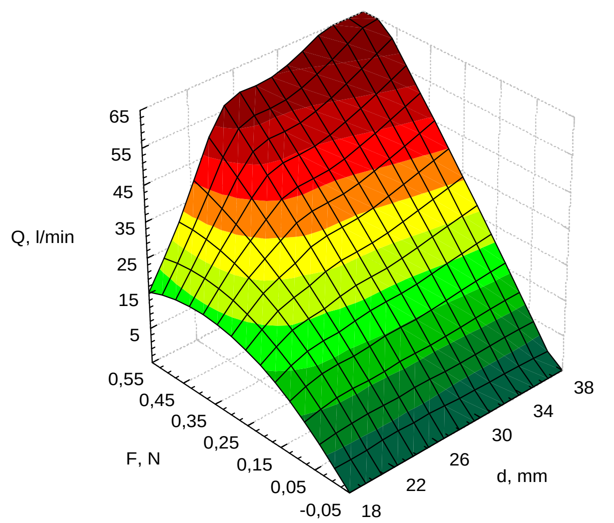

Fig. 6. Blood rate dependence on the tube diameter and magnitude of excitation force

When the tube diameter remains unchanged and increasing the magnitude of excitation force, the blood rate and pressure are increasing (Fig. 7). Largest blood pressure p = 55 kPa will be obtained when the tube will be loaded by 0.5 N force. However, the pressure required at the tube outlet must equal to p = 16 kPa and to reach it the tube must be fluttered by the force with 1002 Hz driving frequency and 0.05 N load.

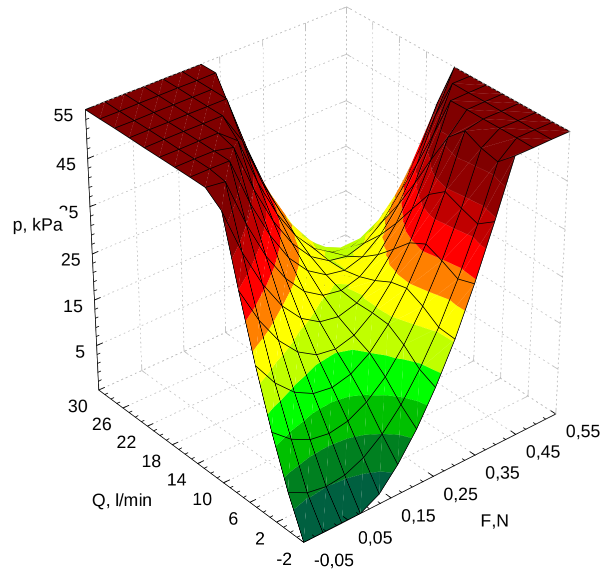

Fig. 7. Blood pressure dependence on the blood rate and force

Increasing amplitude of excitation, the blood pressure increases in the tube (Fig. 8). When amplitude is A = 0.0037 mm and blood rate equals to Q = 0.22 l/min, the pressure obtained at the tube outlet p = 0.67 kPa. Three times increasing the amplitude (A = 0.00872 mm), the pressure at the outlet is p = 5.97 kPa.

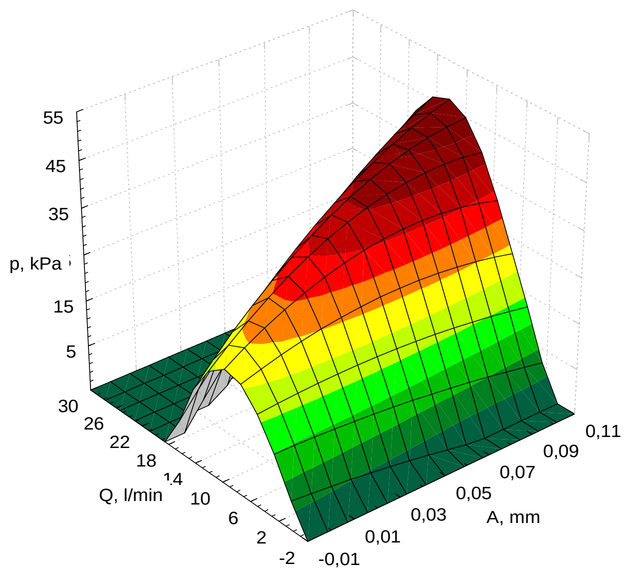

Fig. 8. Blood pressure dependence on the blood rate and amplitude of excitation force

The analysis of calculation results has showed that increasing tube diameter 2 times, efficiency of artificial heart ventricle can be increased up to 4 times (Fig. 7), and increasing amplitude of excitation 10 times, blood pressure increases 2.8 times. Therefore, provided calculation method enables to select rational parameters of artificial heart ventricle.

Conclusions

1. Research has showed that based on the principle of travelling wave it is possible to design artificial heart ventricle, wherein frequency would not exceed 1000 Hz.

2. It was determined that the biggest influence on the efficiency of artificial heart ventricle and blood pressure have the diameter of the tube and oscillation amplitude of its walls. Selecting rational parameters, the efficiency can be increased up to 20 – 30 l/min and the pressure up to 25 kPa.

References

1. **P. Leprince, A. Pavie, N. Bonnet, P. Léger, I. Gandjbakhch:** *Fifteen years single center experience with the total artificial heart Jarvik-7: a bridge to the new era of electrical total artificial heart*. The Journal of Heart and Lung Transplantation, 21(1), 2002, p. 106.
2. **Hogness, J.R., M. Van Antwerp:** *Artificial Heart: Prototypes, Policies and Patients*. Committee to Evaluate the Artificial Heart Program of the National Heart, Lung, and Blood Institute, Division of Health Care Services, Institute of Medicine (National Academy Press, Washington, DC), 1991.
3. **A. I. Hickerson, D. Rinderknecht, M. Gharib:** *Experimental Study of the Behavior of a Valveless Impedance Pump*. Experiments of Fluids, Vol. 38, 2005, p. 534-540.
4. **D. Auerbach, W. Moehring, M. Moser:** *An Analytic Approach to the Liebau Problem of Valveless Pumping*. Cardiovascular Engineering: An Int. Journal, 4(2), 2004, p. 201-207.
5. **L. Loumes:** *Multilayer Impedance Pump: a Bio-inspired Valveless Pump with Medical Applications*. Doctoral Dissertation, California Institute of Technology, 2007, p. 111.
6. **A. I. Hickerson, M. Gharib:** *On the Resonance of a Pliant Tube as a Mechanism for Valveless Pumping*. Journal of Fluid Mechanics, Vol. 555, 2006, p. 141-148.
7. **D. Rinderknecht, A. I. Hickerson, M. Gharib:** *A Valveless Micro Impedance Pump Driven by Electromagnetic Actuation*. Journal of Micromechanics and Microengineering, Vol. 15, 2005, p. 861-866.
8. **E. Jung, C. S. Peskin:** *Two-dimensional Simulations of Valveless Pumping using the Immersed Boundary Method*. SIAM J. Sci. Comput., 23(1), 2006, p. 19-45.
9. **M. L. Cattafesta, J. Mathew, A. Kurdila:** *Modeling and Design of Piezoelectric Actuators for Fluid Flow Control*. SAE transactions, 109(1), 2000, p. 1088-1095.
10. **M. Hamadiche, N. Kizilova:** *Temporal and Spatial Instabilities of the Flow in the Blood Vessels as Multi-Layered Compliant Tubes*. Int. Journal of Dynamics of Fluids, 1(1), 2005, p. 1-24.
11. **M. Mariūnas, G. Četyrkovskis:** *Kairiojo dirbtinio širdies skilvelio savųjų dažnių mažinimo būdas*. Sveikatos mokslai: visuomenės sveikata, medicina, slauga, 1, 2002, p. 46-49.

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