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EVALUATION OF THE EFFICIENCY OF A NEW PULSATILE FLOW-GENERATING CIRCULATORY-ASSIST SYSTEM IN ROTARY BLOOD PUMPS. RESEARCH ON A MATHEMATICAL MODEL

G.P. Itkin[1, 2], A.I. Syrbu[2], A.P. Kyleshov[1], A.S. Buchnev[1], A.A. Drobyshev[1]

[1] Shumakov National Medical Research Center of Transplantology and Artificial Organs, Moscow, Russian Federation

[2] Moscow Institute of Physics and Technology, Moscow, Russian Federation

Objective: to study the effect of a pulsatile flow-generation (PFG) device on the basic hemodynamic parameters of the circulatory system using a mathematical model. **Results.** Modelling and simulation showed that the use of PFG significantly (76%) increases aortic pulse pressure. The proposed mathematical model adequately describes the dynamics of the circulatory system and metabolism (oxygen debt) on physical activity in normal conditions and heart failure, and the use of non-pulsatile and pulsatile circulatory-assist systems. The mathematical model also shows that the use of PFG device blocks the development of rarefaction in the left ventricular cavity associated with a mismatch of blood inflow and outflow in diastolic phase when there is need to increase systemic blood flow by increasing the rotary pump speed.

Keywords: mathematical model, circulatory system, pulsatile rotary blood pump, pulsatile flow generation.

INTRODUCTION

In recent decades, rotary nonpulsatile flow pumps (NFPs) have almost replaced pulsatile flow pumps (PFPs) in clinical practice as a bridge to heart transplantation and targeted therapy, due to the advantage of these pumps in weight-size, energy and performance characteristics. This promoted higher survival among patients with end-stage heart failure (HF) [1–3]. However, prolonged use of NFPs leads to a number of complications associated with low aortic pressure pulsation. These include gastrointestinal bleeding, arteriovenous malformation, aortic valve insufficiency, etc. [4–7]. Besides, these pumps showed low cardiac unloading in comparison with PFPs, which is one of the main factors of native myocardial contractility restoration [8, 9]. A number of works have shown the important role that pulsatile flow plays not only in implantable systems, but also in extracorporeal short-term mechanical circulatory assist systems, including cardiopulmonary bypass systems [10, 11]. Therefore, in the last decade, the attention of researchers has been directed towards the development of various pulsatile flow generation (PFG) methods [12–14].

One of the main directions for effectiveness assessment of these methods is the use of mathematical models (MM) allowing to analyze the operation of assisted circulatory systems (MCAD) more fully in the dynamic range of circulatory system parameter changes (under physical load, myocardial contractility changes, etc.) compared to the studies using hydrodynamic circulatory system simulators. In particular, the use of MM methods will allow for comparative assessment of the efficiency of MCAD techniques using NFPs in standard mode and in PFG mode. In this work, we carried out a comparative analysis of the circulatory system when including left ventricular assist devices (LVADs) using NFPs in standard non-pulsatile mode and in PFG mode [15].

MATERIALS AND METHODS

The basic structure of the mathematical model of the cardiovascular system (CVS) is presented in A.I. Syrbu et al. [16]. This model additionally includes NFP modules, pulse flow generator (PFG) [15], as well as an element simulating the effect of left ventricular (LV) discharge development, occurring at high NFP speeds [17]. This MM is developed in Matlab Simulink environment using electrohydraulic analog method and describes a large circulatory circuit, consisting of the following modules (Fig. 1): left ventricle (LV), left atrium (LA), aortic valve (AV), mitral valve (MV), aortic section (AS), peripheral section (PS), venous section (VS), coronary vessels (CV), and regulation circuits: baroreceptor (B), oxygen debt (O_2) and heart rate (HR). The dashed lines in Fig. 1 highlight the NFP of PFG and the element simulating LV rarefaction (R).

This MM allows for a comparative study of CVS operation with inclusion of NFPs in the standard mode

Corresponding author: George Itkin. Address: 1, Shchukinskaya str., Moscow, 123182, Russian Federation.
Phone: (916) 129-78-33. E-mail: georgeitkin@mail.ru

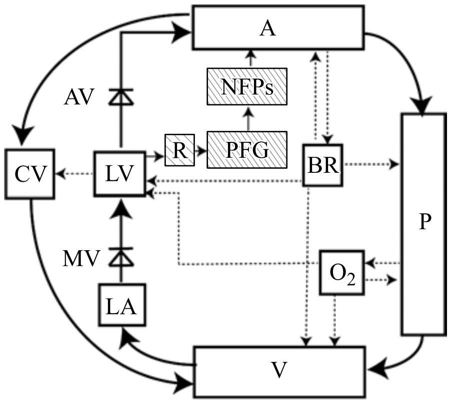

Fig. 1. Block diagram of the mathematical model

and with operation of PFPs with inclusion of PFG. The additional modules included in the CVS mathematical model are described by the following basic relationships.

Blood flow through NFPs is defined by the expression [17]:

$$dQ_{vad}/dt = (P(Q_{vad}) - (P_{ao} - P_{lv}) - R_c Q_{vad}) / L_c,$$

where $P(Q_{vad})$ is the dependence of pressure created by NFPs = on the flow through it, which is determined from the flow-pressure characteristic of NFPs, P_{ao} is aortic pressure, P_{lv} is LV pressure, R_c and L_c are cannula resistance and inertia.

The pressure generated by the NFP is described by the following differential equation [17]:

$$P(Q_{vad}) = aQ_{vad} + bdQ_{vad}/dt + cw^2,$$

where Q_{vad} is blood flow through NFPs; w is pump rotor speed; a, b, c are coefficients; dQ/dt is the time differentiation operator. The coefficients were selected from the flow-pressure characteristic of the HeartWare HVAD centrifugal pump [18].

PFG contains a hydraulic resistance made in the form of connector, in which there is a tube of elastic biocompatible material, hermetically sealed by the ends along the butt ends of the cylinder from its inner side (Fig. 2).

This tube opens fully at a pressure greater than a certain P_2 in the systolic phase and partially closes when the pressure drops below P_1 in the LV diastolic phase, thereby increasing the hydraulic resistance to blood flow.

The following equation was used to describe the PFG device:

$$R_{PFG} = \frac{(P_2 - P_{lv})R_{max}}{P_1 - P_2},\ R_{PFG} \geq 0,$$

where R_{max} is the highest hydraulic resistance in diastole, P_1 is the pressure at which the elastic tube partially collapses, P_2 is the pressure at which the elastic tube opens and resistance is minimal ($R_{min} = 0$).

One of the problems related to NFP operation in elevated rotor speed regime, necessary to normalize systemic blood flow and improve LV unloading, is the danger of developing LV rarefaction related to mismatch of blood inflow and outflow through the pump during diastolic phase. It can lead to tissue damage in the inlet cannula, interventricular septum displacement, deterioration in right ventricular function, arrhythmia, cardiac ischemia, and hemolysis [19].

To describe this effect, we used the following piecewise function [20, 21]:

$$R = \begin{cases} 0; P_{lv} \geq P_{th} \\ -3.5P_{lv} + 3.5P_{th}; P_{lv} < P_{th} \end{cases},$$

where R is additional resistance at the NFP inlet, P_{th} is threshold value.

RESULTS

Fig. 3 shows time dependences of aortic pressure (AP), systemic blood flow, and HR in the simulations: HF (time interval I: 0 to 30 sec), connection of NFPs without PFG (time interval II: 30 to 60 sec), and with PFG connection without change in NFPs (time interval III: 60 to 90 sec) and with increase in NFP speed (time interval IV: 90 to 120 sec).

Analysis of the dependencies shown in Fig. 3, shows the following: NFP connection leads to changes in the main parameters of hemodynamics: increase in mean AP to 114/100 mmHg and blood flow to 4.2 L/min and decrease in HR to 75 beats/min. At the same time, the qualitative nature of the dependencies and the obtained values of hemodynamic parameters correspond to the data given in the literature [22]. In turn, PFG connection at constant NFP rotor speed leads to decreased blood

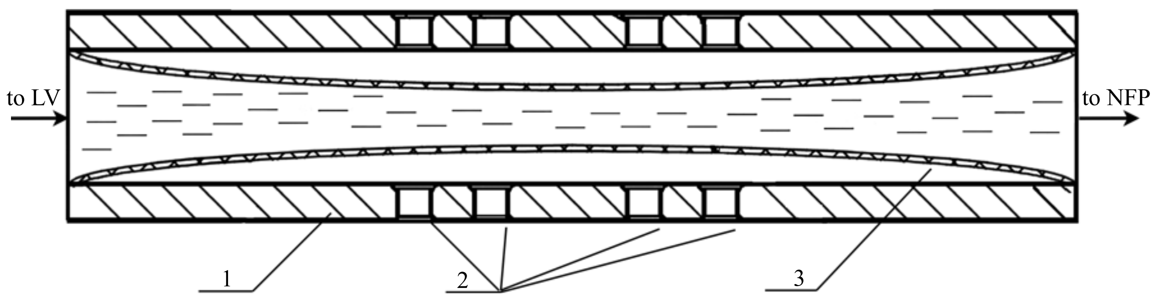

Fig. 2. PFG scheme (1 – connector, 2 – holes with an outlet to the atmosphere or compensation chamber, 3 – elastic tube)

flow and aortic pressure with normalization of the pulsatile nature of blood flow in the aorta. Therefore, to normalize mean AP and systemic blood flow, the NFP rotational speed was increased by 15%. In this case, a concomitant effect is a further increase in pulsatile AP by 76% compared to NFP without PFG.

This model also allows for evaluation of NFPs under physical stress. It was obtained that when PFG was connected, pulse pressure increased from 10 to 24 mmHg, or by 140%.

Fig. 4 shows the time dependences of aortic pressure (AP), systemic blood flow (SBF), HR and oxygen debt (DO) in the simulations: physiological norm (a), HF (b), physical activity in HF conditions (c), physical activity in HF conditions with connected NFP (d) and with PFG with increased NFP turnover (e).

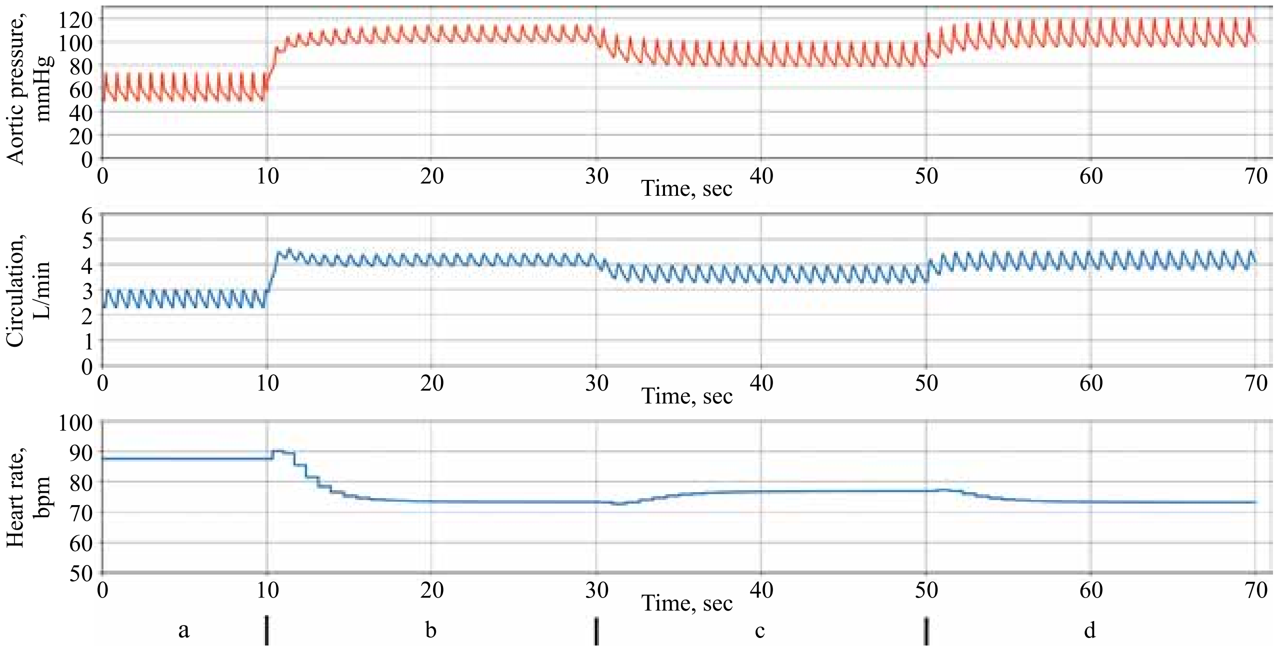

Fig. 3. Dependence of blood pressure, systemic circulation and heart rate on time for simulated states: a, HF; b, NPBP connection; c, NPBP with PFG without increased rotor speed; d, NPBP with PFG with increased rotor speed

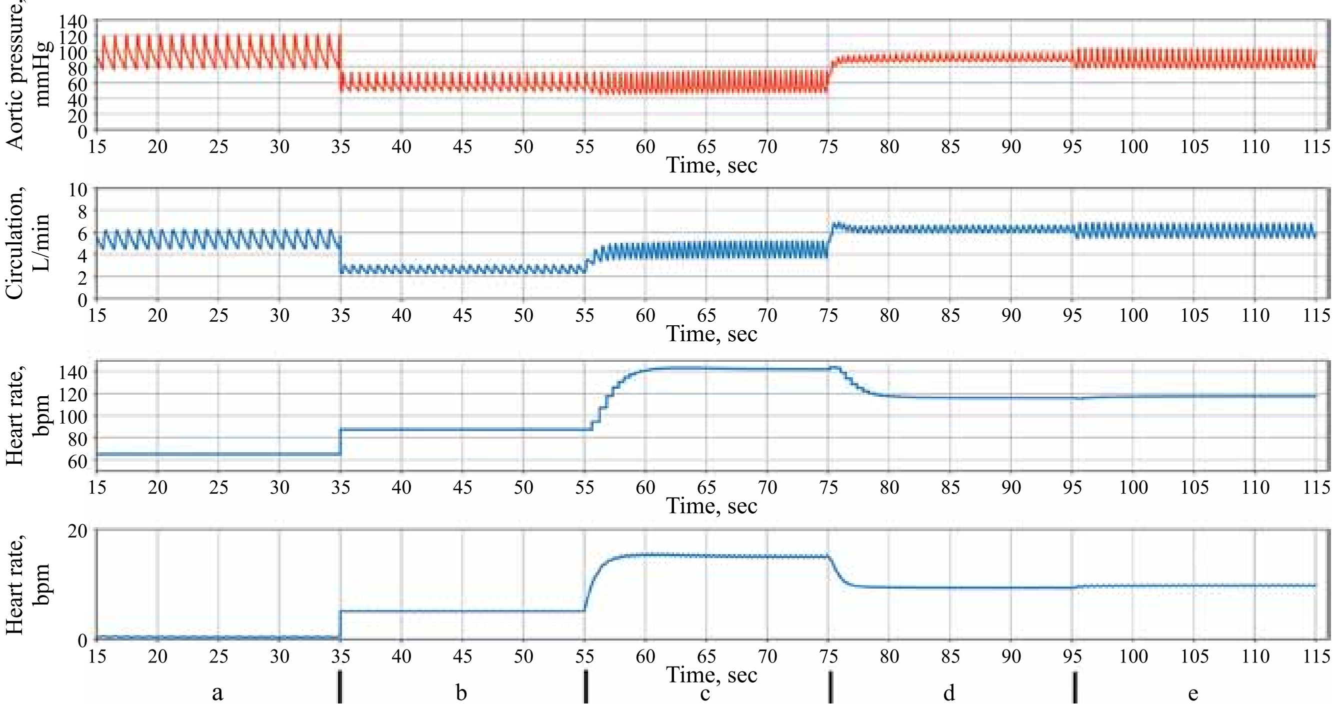

Fig. 4. Dependence of blood pressure, systemic circulation, heart rate and oxygen debt on time in the simulation of physiological norm (a), HF (b), physical activity under HF conditions (c), with connected NPBP (d) and PFG connection (e)

The results obtained are consistent with the experimental data obtained in [22] in the study of physical activity in HF patients with NFPs.

As indicated above, one of the problems of clinical application of PFPs is the effect of development of rarefaction in the diastolic phase of LV when systemic blood flow should be increased by increasing PFP speed [23, 24].

On MM, we evaluated this effect with increased speed of the PFPs (15%) in standard mode and when the PFG is connected (Fig. 5).

The simulation results show that in standard NFP mode, rarefaction reached –58 mmHg, and when PFG was connected, the dangerous effect of LV inlet rarefaction completely disappeared.

Fig. 6 shows the dependence of arterial pressure on LV elasticity in HF modeling with NFP and PFG. The increase in elasticity led to increased mean AP from 90 to 110 mmHg and pulse pressure from 20 to 30 mmHg.

CONCLUSION

The computer MM of human systemic circulation, built on the basis of the model developed earlier by us [16], allows for investigation of the state of the circulatory system in heart failure when MCAD is connected in LVAD mode in standard NFP mode and when PFG is connected. Simulation results show a significant (76% and 140% without and with physical load) increase in pulse pressure in the aorta as well as prevention of rarefaction area in the left ventricle when using variable hydraulic resistance. It has been shown that it is possible to model a new device – PFG, which opens the possibility of optimizing both the parameters of the PFG device itself and combination of PFG and NFP parameters (NFP

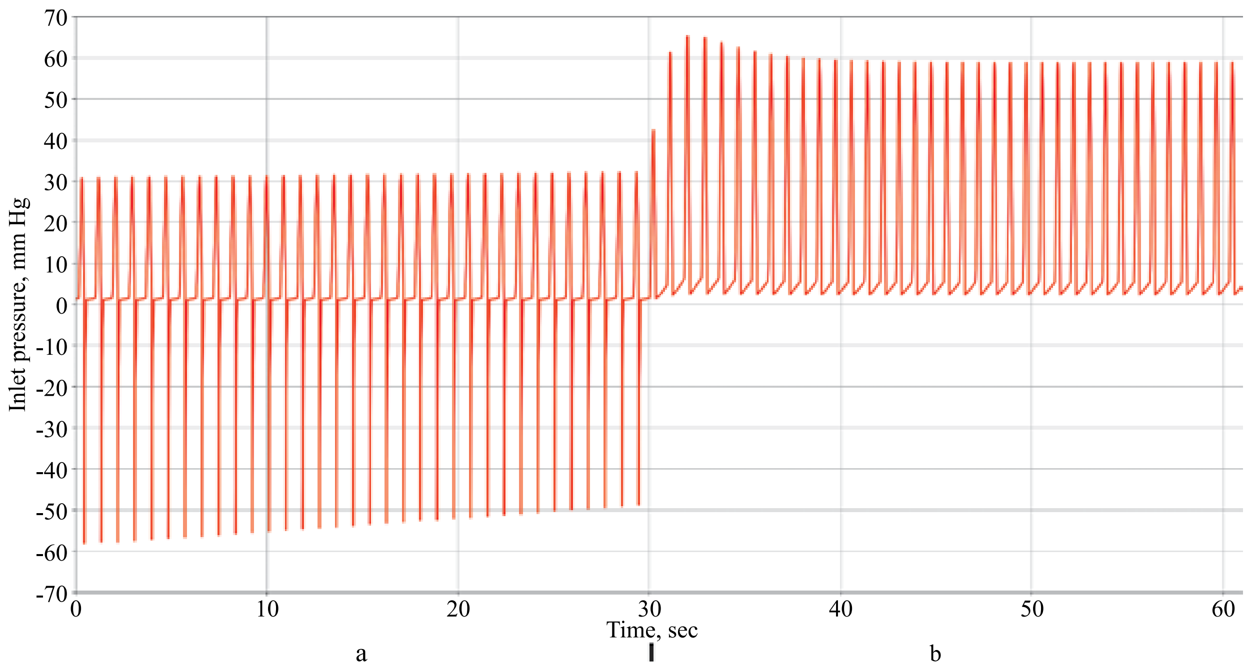

Fig. 5. Dependence of NPBP inlet pressure on high rotor speed without (a) and with PFG (b)

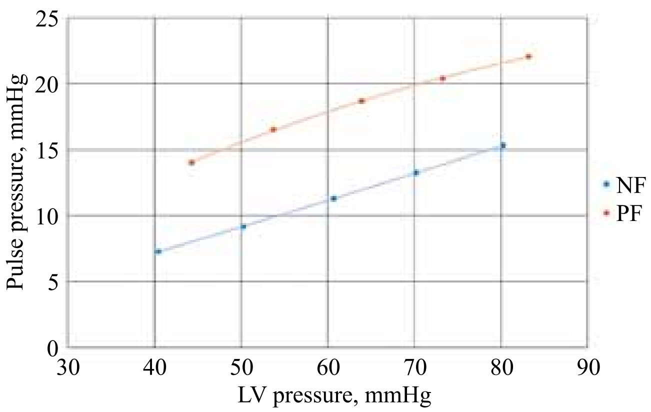

Fig. 6. Changes in pulse pressure with increasing LV pressure in heart failure with NPBP and PFG

operation modes, NFP outlet pressure at different PFG parameters and others). The further direction of research is precisely on addressing the problems of optimizing these devices.

The authors declare no conflict of interest.

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