Modeling of Circulatory System With Coronary Circulation and the POLVAD Ventricular Assist Device

Alicja Siewnicka, Bartlomiej Fajdek, Krzysztof Janiszowski

Abstract:

This paper presents an application of a numerical package for modeling and simulation of human circulatory system. The model includes a coronary circulation and the parallel heart assistance. The cases of the simulation of the proper and the pathology circulation conditions, such as left or right heart failure are shown. A description of the coronary circulation system is presented and obtained coronary sinus occlusion simulation results are included. An implementation of the whole package as a part of PExSim application is contained. The identification experiment for the ventricular assist device has been described and different methods of the artificial ventricle modeling are presented. An example of use of a fuzzy logic to presentation the dynamics of the POLVAD device is also included. Advantages of developed simulation platform are discussed.

Keywords*: PExSim, modeling of the circulatory system, modeling of the coronary system, modeling of the ventricular assist device, POLVAD.*

1. Introduction

The continuous development of technology enabled for the more common use of its achievements in medical applications. Therefore, in recent years many scientific projects were run to allow the use of technology to save human life and health. One of the biggest bio-engineering projects in Poland is the Polish Artificial Heart Program, whose aim is to develop of the construction and control algorithms for the heart assist device. Application of the developed solutions requires accurate testing. For this purpose the modeling methods are widely used. In recent years some different models of the human circulatory system were developed, both numerical and physical ones, for example electrical or hydraulic [1, 2]. All of them are widely used to reproduce hemodynamic conditions of circulation system. Besides this, they can be applied for the testing of medical devices such as the blood pumps or assist devices. For this purposes also models of the same new devices are created. This way the possibility of simulating of the influence of the heart support can be obtained without carrying out experiments on living organisms. This paper contains description of the developed circulatory model with the possibility of connection of the simply models of the extracorporeal ventricle assist device. In our case, the main aim of development the mathematical description of circulatory system was to create a research platform, for general purpose, which could be easily adopted to solve different problems. For

example it could be used for determination and testing of a Polish Ventricular Assist Device (POLVAD) control and diagnostic algorithms.

2. Model of the circulatory system

The main part of the developed research platform is the mathematical model of the circulatory system. It is based on the description proposed by Ferrari [3] and implemented as a part of the PExSim application [4]. This software consists of predefined function blocks that represent basic mathematical and logic relationships, dynamic and static elements or support an input and output operations. The possibility of easy extension with user-written objects makes it a flexible tool that can be used for emulation of complex dynamic system. For the simple and clear presentation of circulation system and to ensure the ability of easy parameters changes, each of the blocks is responsible for reproducing behavior of different part of human circulatory system. They are grouped in *Human Circulatory System* library, which consists such elements and systems as:
–   left and right ventricle (LH, RH),
–   systemic arterial circulation (SAC),
–   systemic venous circulation (SVC),
–   pulmonary arterial circulation (PAC),
–   pulmonary venous circulation (PVC).

The model is based on a Starling`s law [3], which defines the conditions for a balance between the filling and ejection characteristics of the ventricle. The basic relation is a function that makes it possible to calculate the ventricular pressure [3, 5]:

$$P_v(t) = (V_v(t) - V_{v0}(t)) \cdot E_{\max} \cdot f(V_v(t), \dot{V}_v(t), V_{v\max}(t))$$
$$+ A \cdot e^{k \cdot Vv(t)} + B \cdot e^{-j \cdot Vv(t)} + C \qquad (1)$$

where: $P_v(t)$ – the ventricle pressure, $V_v(t)$ – the ventricle volume, V_{v0} – the ventricle rest volume, $E_v(t)$ – the normalized elastanse function, $E_{\max}$ – the maximum value of the elastance (end-systolic), $f(V_v(t), \dot{V}_v(t), \dot{V}_{v\max}(t))$ – the correction function dependent on the ventricle volume and ejection rate, and A, B, C, j, k – constant parameters.

The values of the flows are calculated as the ratio of the proper pressure difference and the vascular resistance. The rest of the pressures values are calculated as the solution of the differential equation which defines the pressure derivative as the quotient of the sum of flows and the capacity of the system. In the PExSim modeling platform, the simple model of ventricular assist device (VAD), based on the ventricle activity description, was added. The parameters values were adopted to ensure

work consistent with the theoretical data. The model of the ventricular assist device is included into the system between the atrium and the arterial system, creating a bypass of the model of the natural ventricle. The mathematical equations in the circulation blocks were respectively adopted to join the parallel ventricle support. The proper connection of all elements creates a complete model of circulatory system with the external assist device (Fig. 1).
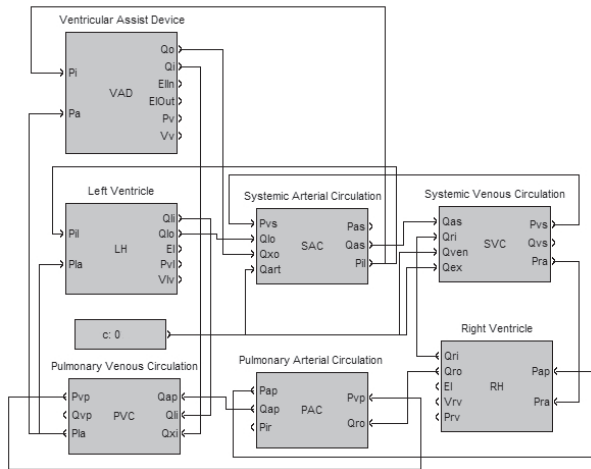

Fig. 1. The circulatory model in PExSim application with the parallel connection of ventricular assist device

A full description of structure and function of the blocks one can find in [6]. As a result of this work we received a useful tool, that yields an investigation of the proper and pathological circulation conditions and the influence of the ventricle assistance for the hemodynamic conditions. For example, Fig. 2 presents the influence of the left heart support on the atrium (P_{la}) and arterial (P_{as}) pressures.
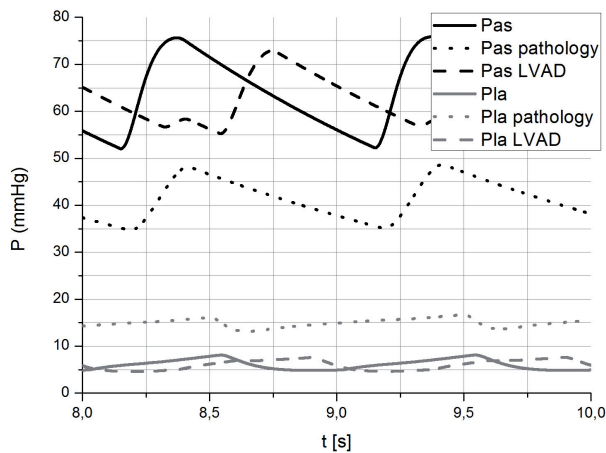

Fig. 2. The modeled waveforms of the atrium (P_{la}) and arterial (P_{as}) pressures for normal, pathological and pathological with left ventricle assistance (LVAD) conditions

In the ventricular failure state the blood accumulates in atrium causing the rise of the pressure volume. At the same time the pressure in arteries is low due to insufficient stroke volume of the heart. In simulation we can observe the reduction of the atrium pressure value and increase of the arterial pressure as a result of left ventricle support which confirms medical observations.

3. The coronary circulation model

The system described above did not contain the coronary circulation model. As the extension of functionality of the developed platform, the coronary circulation model was developed and included. This way a possibility of simulating the caval occlusion was achieved. The applied
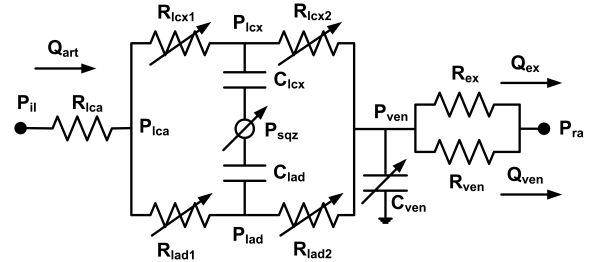

Fig. 3. The electric analogue of the coronary circulation system

mathematical description is the combination of the models proposed in [7] and [8]. The schematic representation of this, as an analogy to electric circuit diagram, is shown in Fig. 3. It was added to the mentioned PExSim application as a new CC (Coronary Circulation) element.

The driving pressure for the coronary circulation (P_{sqz}) is taken as a proportional to left ventricle pressure. According to this, the input values for the model are: pressure values in aorta, left ventricle and right atrium (P_{il}, P_{lv}, P_{ra}). The others pressures are obtained by the equations:

$$P_{lca}(t) = P_{il}(t) - R_{lca} \cdot Q_{art}(t) \tag{2}$$

$$P_{lcx}(t) = \frac{V_{lcx}(t)}{C_{lcx}} + P_{sqz}(t) \tag{3}$$

$$P_{lad}(t) = \frac{V_{lad}(t)}{C_{lad}} + P_{sqz}(t) \tag{4}$$

$$P_{ven}(t) = \frac{V_{ven}(t)}{\chi_{ven}} \cdot \exp[\sigma(V_{ven}(t) - V_{ven0})] \tag{5}$$

where:
P_{lca} – the pressure in bifurcation of coronary arteries, P_{il} – the aortic pressure, Q_{art} – the coronary arterial flow, P_{lcx} and P_{lad} – the coronary capillaries pressures (lad – left anterior descending artery, l_{cx} – left circumflex artery), R_{lca} – the arterial resistance, C_{lcx} and C_{lad} – the coronary capillaries compliances, V_{lcx} and V_{lad} – the coronary capillaries volumes, V_{ven} – the coronary veins volume, χ_{ven}, σ, V_{ven0} – the parameters for venous compliance.

The coronary arterial flow value (Q_{art}) is obtained as a sum of the capillaries input flows (Q_{lcx1}, Q_{lad1}). Input and output flows are calculated based on pressures difference and blood vessels volumes as follows:

flows in direction "to the coronary capilares system" ($P_{lca} \geq P_{lcx}$, $P_{lca} \geq P_{lad}$, $P_{ven} \geq P_{lcx}$, $P_{ven} \geq P_{lad}$, $P_{ra} \geq P_{ven}$):

$$Q_{lcx1}(t) = \frac{P_{lca}(t) - P_{lcx}(t)}{R_{lcx1}} \tag{6}$$

$$Q_{lad1}(t) = \frac{P_{lca}(t) - P_{lad}(t)}{R_{lad1}} \qquad (7)$$

$$Q_{lcx2}(t) = \frac{P_{lcx}(t) - P_{ven}(t)}{R_{lcx2}} \qquad (8)$$

$$Q_{lad2}(t) = \frac{P_{lad}(t) - P_{ven}(t)}{R_{lad2}} \qquad (9)$$

$$Q_{ven}(t) = \frac{P_{ven}(t) - P_{ra}(t)}{R_{ven}} \qquad (10)$$

where:

Q_{lcx1} and Q_{lad1} – the capillaries input flows, R_{lcx1} and R_{lad1} – the capillary resistances, Q_{lcx2} and Q_{lad2} – the capillary output flows, R_{lcx2} and R_{lad2} – the capillary resistances, Q_{ven} – the flow supplying the right atrium, P_{ra} – the right atrium pressure, R_{ven} – the coronary venous resistance.

- flows in direction "from the coronary capilares system" ($P_{lca} < P_{lcx}$, $P_{lca} < P_{lad}$, $P_{ven} < P_{lcx}$, $P_{ven} < P_{lad}$):

$$Q_{lcx1}(t) = \frac{P_{lca}(t) - P_{lcx}(t)}{R_{lcx1} + \beta \cdot (V_{lcx})^{-2}} \qquad (11)$$

$$Q_{lad1}(t) = \frac{P_{lca}(t) - P_{lad}(t)}{R_{lad1} + \beta \cdot (V_{lad})^{-2}} \qquad (12)$$

$$Q_{lcx2}(t) = \frac{P_{lcx}(t) - P_{ven}(t)}{R_{lcx2} + \beta \cdot (V_{lcx})^{-2}} \qquad (13)$$

$$Q_{lad2}(t) = \frac{P_{lad}(t) - P_{ven}(t)}{R_{lad2} + \beta \cdot (V_{lad})^{-2}} \qquad (14)$$

$$Q_{ven}(t) = \frac{P_{ven}(t) - P_{ra}(t)}{R_{ven} + \beta \cdot (V_{ven})^{-2}} \qquad (15)$$

$$Q_{ex}(t) = \frac{P_{ven}(t) - P_{ra}(t) + P_{venoff}}{R_{ex} + \beta \cdot (V_{ven})^{-2}} \qquad (16)$$

where:

β – the parameter for volume dependent resistances, Q_{ex} – the extraordinary venous outflow, R_{ex} – the extraordinary venous resistance, P_{venoff} – the venous pressure offset.

The volumes of vessels are determined from the equations:

$$\dot{V}_{lcx}(t) = Q_{lcx1}(t) - Q_{lcx2}(t) \qquad (17)$$

$$\dot{V}_{lad}(t) = Q_{lad1}(t) - Q_{lad2}(t) \qquad (18)$$

$$\dot{V}_{ven}(t) = Q_{lcx2}(t) + Q_{lad2}(t) - Q_{ven}(t) - Q_{ex}(t) \qquad (19)$$

This complex model was added to the PExSim application as a new Coronary Circulation (CC) function block. In order to join the coronary circulation function block to the main model of circulation system, the modifications in equations for systemic arterial and venous circulation models had to be implemented. In SAC element the input for coronary arterial flow (Q_{art}) was added and the description was modified. In SVC element two extra inputs were added, for the coronary venous and extraordinary flows (Fig. 4).
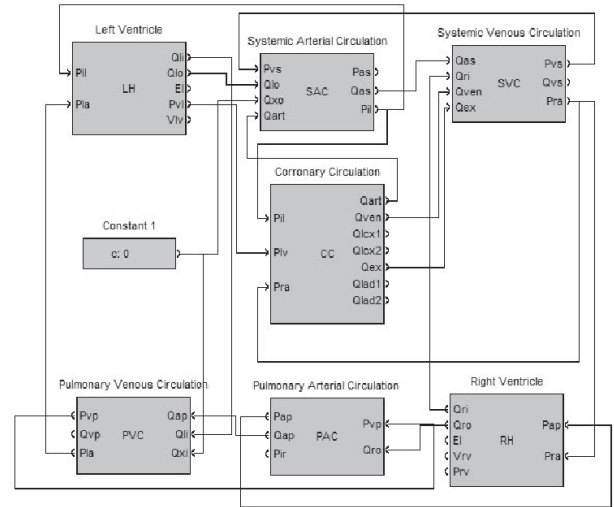

Fig. 4. The circulatory model in PExSim application with the coronary circulation system model

As a result we gained the opportunity to simulate various dysfunctions of the coronary circulation system, e. g. coronary sinus occlusion. In Fig. 5 we can see the value of flow from the capillaries to veins in normal state and for the simulation of sinus occlusion.
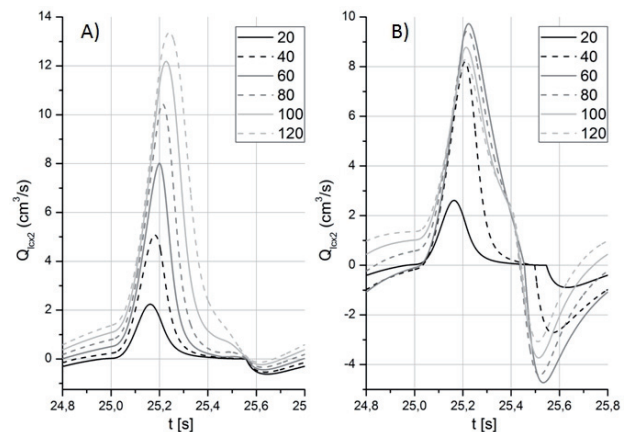

Fig. 5. The modeled coronary flow from the capillaries to veins under variation of perfusion pressure for the normal (A) and coronary sinus occlusion (B) conditions

For the coronary vein occlusion conditions we can observe the increase of the invert flow value as a result of vascular obstruction.

4. The POLVAD modeling

The next step for development of complete simulation platform was to create a model of the real assist device based on measurement results. For this purpose, an identification experiment of an extracorporeal ventricular assist device (POLVAD) was carried out at the Foundation of Cardiac Surgery Development in Zabrze (Poland). The Polish Cardiac Assist System POLCAST undergoing testing consists of the pneumatic driving unit POLPDU-402 and the pneumatic drive pulse pump POLVAD-MEV with disc valves and two chambers separated by a flexible membrane (Fig. 6). The blood flow is a result of membrane movement, which is caused by air pressure changes in the pneumatic chamber. For the

in vivo experiments, instead of blood, a fluid with similar properties was used. The assist device was connected to a hybrid circulatory system model, which represents the circulatory system.
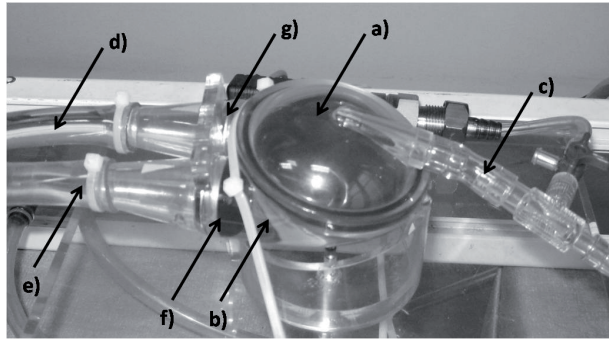

Fig. 6. The extracorporeal heart support ventricle POL-VAD: a) pneumatic ventricle, b) blood ventricle, c) pneumatic pipe, d) input cannula, e) output cannula, g), f) disc valves

The following variables were measured: a supply pressure (P_{pn}), and pressures and flows in input and output cannula (P_{in}, P_{out}, Q_{in}, Q_{out}). As a result of the experiment, data series for different supply and load conditions were obtained. They were used as a basis for modeling the dynamics of the ventricular assist device which is not a trivial task. The change in parameters strongly affects the nature of the process. What is more, a model of mechanical valves has to be developed. The next problem is a strong dependence of the process state: full filling and emptying of blood chamber. For cardiological reasons, the most important value is the volume of the cardiac output. This could be estimated from the output flow value. For this purpose, proper modeling of the output flow is the basic factor in a control system design. The value of the input flow is also very important. That is why we have been trying to develop a model of the structure presented in Fig. 7, in which Pmax, Pmin, %SYS, BPM are the maximal and minimal values of the control pressure, the percentage ratio of the systolic phase time to the cycle time, and the beats per minute, respectively.
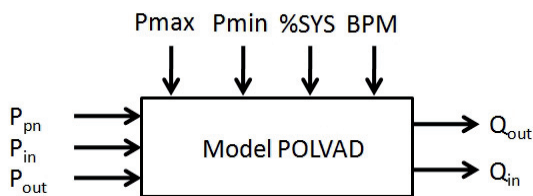

Fig. 7. A schematic diagram of the dynamics of a ventricular assist device

The original idea for more accurate mathematical description of the chamber was to modify the shape of the elastance function. The valves, usually modeled as ideal diodes [3], were replaced by inertia elements. The method of determining the pressure in the blood chamber was also modified [9]. The unknown parameters of the description were determined using dedicated program called PExSim Optimizer [10]. The waveforms obtained by modeling approximately reflected the nature of the

measured flows values. However this model was independent from the control pressure value. It means that description would never be universal – we would have to adjust parameters to every particular situation.

Another approach to the proper description of the dynamics of the assist device was an attempt to modeling using the fundamental principles of hydraulics and pneumatics. First, the pressure in the blood chamber was determined as a function of the control pressure. An algorithm was obtained to detect changes in the control pressure phases, which facilitated the modeling the opening and closing of the valves. The output flow was calculated on the basis of the pressure balance and changes in the valve resistance. The parameters of the process were selected manually. The exemplary result of the simulation is presented in Fig. 8.
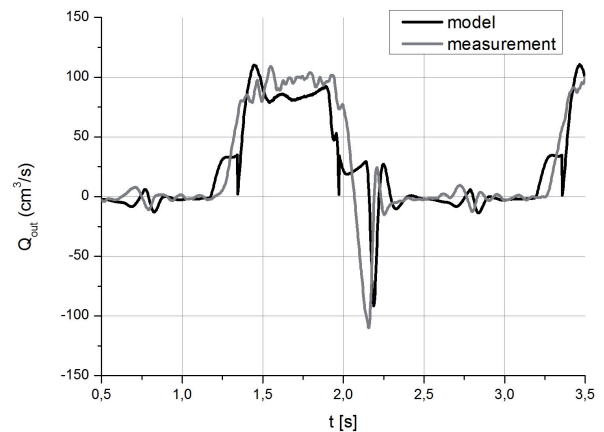

Fig. 8 The modeled and measured assist device output flow (Q_{out})

The obtained waveforms are not fully identical with real values, but they reflect the nature of the appropriate flows. The advantage of this description is that for the same parameter values it gives a similar quality of rendering for different supply parameters. In this model, the control pressure is one of the model inputs and it provides information about the extreme values and the filling ratio of the control signal and a BPM value.

Different approach to the problem was the fuzzy modeling. It is based on an approximate description of the dynamics using a set of transfer functions, being the sum of the impact of individual inputs to the model output. For example, in our case, the output flow value, for zero initial values, can be represented according to the relation:

$$Q_{out}(s) = G_{QoutPin}(s)P_{in}(s) + G_{QoutQin}(s)Q_{in}(s) + \\ + G_{QoutPout}(s)P_{out}(s) + G_{QoutPpn}(s)P_{pn}(s) \tag{20}$$

Partial transfer functions have the form (21) of general transfer function with unknown coefficients α and β and the delay values T_j:

$$G_{QoutVj}(s) = \frac{[\beta_{0j} + \beta_{1j}s + \dots + \beta_{nj}s^n]e^{-sT_j}}{1 + \alpha_1 s + \dots + \alpha_n s^n} \tag{21}$$

Determination of the coefficients is performed mostly in an indirect way of parametric identification method. On the base of sampled data the discrete transfer functions $G_{QoutVj}(z^{-1})$ are determined in such a way that the modeled signal, $\tilde{Q}_{out}(k\Delta)$, where Δ is the sampling step, should be as closely as possible to the measured one. We get a description consisting of a vector of inputs to the model v(kΔ), containing the signals measured in the respective moments of time, and the vector θ of unknown coefficients of the investigated transfer functions.

$$\tilde{Q}_{out}(k\Delta) = v(k\Delta) \cdot \theta \qquad (22)$$

Vector of transfer function coefficients can be estimated using various methods, for example, by the smallest sum of squared errors (LS). The dynamic model determined in this way usually accurately reproduces the dynamics of the process but is very local. This means that the change of supply parameters makes it necessary to re-selection transfer function coefficients. That why, the fuzzy modeling is very useful method. For the same structure of the system (22) vector of coefficients θ is dependent on some fuzzy variables, defined by membership functions. For example, separate models can be determined for the value of low, medium and high value of the fuzzy variable. The model is the weighted sum of the partial models and the corresponding membership function values. General fuzzy model can be determined as a linear combination of several local models set for the different intervals of membership function. Estimation algorithms of partial models coefficient vectors are more complex and require the simultaneous calculation of the vectors for all the partial models. The basic difficulty is also the designation of the proper shape and number of membership functions. In our case, in order to obtain a general description of the dynamic properties of the device, as a fuzzyfying variables were used two signals: output flow value Q_{out} and a set pressure different ΔP. The first signal determines different valve states (Fig. 9a): positive flow (ejection, valve open), backward flow (closing of the valve) and the closed valve phase (no flow). For the second variable the membership function was divided for three areas (Fig. 9b): low, medium and high values of pressure.
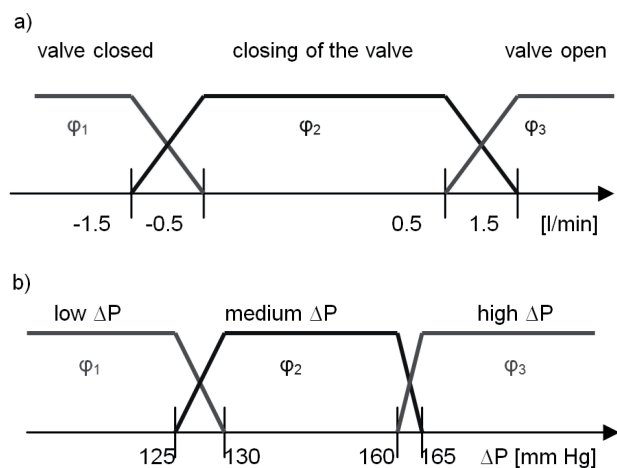
a)

b)

Fig. 9 The membership functions for the fuzzy variables: a) output flow value, b) pressure difference

As a result of the presented fuzzy modeling method the nine partial models had to be determined. The measurements were carried out for five values of pressure difference. For the estimation of the models the measurement data for maximum, medium and low pressure differential were used. To verify the obtained model, the remaining series were used. The sample waveforms of the modeled and measured output flow are shown in Fig. 10.
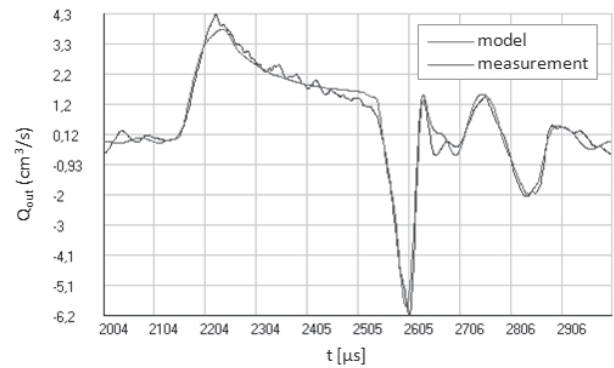

Fig. 10 Sample measured and modeled flow waveform for the verification data set

The estimated parametric fuzzy model reproduces output flow value relatively well both, for the data on which he was appointed and the verification series. It yields a good representation of the dynamics of the POLVAD artificial ventricle. The extended description of the fuzzy modeling applied to determine the assist device model one can find in [11].

5. Summary

The paper presents a new numerical library for modeling and simulation of human circulatory system with the extension of coronary circulation model and possibility of the parallel assist device connection. The main result of the coronary model addition was to allow simulation and verification of the influence of the left ventricle assistance on the coronary flow conditions. The whole library was implemented as a part of PExSim application. Modular construction of the plugin ensures flexibility because it can be easily modified to solve different problems within modeling and support of the human circulatory system. The modeling methods of the ventricular assist device were presented. We received an approximate representation of the output flow value dependent on the supply pressure. Method based on fuzzy modeling made it possible to achieve better results. However, in the case of even small modification of the device construction, the measurements and the whole modeling procedure will have to be carried again. Also, we do not receive the direct dependence of the output signal from the power supply parameters. Work on the determination of a better model of the mechanical construction of the ventricular assist device is still carried out. However, as a result of presented work we have the useful tool, which gives the opportunity to study functions of individual components of the human circulatory system as well as the system as whole. It enables a simulation of the proper and the pathology circulation conditions, such as left or right heart failure. Implemented model of the ventricle assist device

gives the opportunity for modeling the influence of the heart assistance on the hemodynamic co

Acknowledgements

AUTHORS

Alicja Siewnicka*, Bartłomiej Fajdek, Krzysztof Janiszowski – Warsaw University of Technology, Faculty of Mechatronics, *Institute of Automatic Control and Robotics,* ul. Św. Andrzeja Boboli 8, 02-525, Poland,

E-mail: alicja.siewnicka@gmail.com, b.fajdek@mchtr.pw.edu.pl, kjanisz@mchtr.pw.edu.pl

*Corresponding author

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