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Application of a PExSim for modeling a POLVAD artificial heart and the human circulatory system with left ventricle assistance

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This paper presents a model of the human circulatory system with the possible addition of a parallel assist device, which was developed for the purpose of artificial heart monitoring. Information about an identification experiment of an extracorporeal ventricle assist device POLVAD is included. The modelling methods applied and the corresponding functional blocks in a PExSim package are presented. The results of the simulation for physiological conditions, left ventricle failure and pathological conditions with parallel assistance are included.

Key words: PExSim, modeling of the ventricular assist device, POLVAD, modeling of the circulatory system.

Introduction

The use of artificial organs is of significant importance in today's medicine. A large number of human organs have an auto regeneration function. One of them is the heart muscle. That is why ventricular assist devices (VADs) are widely used. They can ensure a proper functioning of the circulation system during ventricular failure and/or heart transplants. This kind of therapy is also used for the supporting of life functions in cardiac surgery. In recent years, many types of the assist devices have been developed. They vary depending on working conditions and the place of introduction into the circulation system. The pumps can be divided into two main categories: continuous flow pumps or pulsatile pumps. They can be connected in parallel or in series to the heart ventricle [2]. Before being used in clinical practice, every medical device must be carefully tested and proper operation parameters have to be selected. For in vivo testing, modelling of the heart and the circulation system can be applied.

The main aim of our work was to develop a numerical model of the ventricular assist device and to create a research platform for control algorithms testing. To achieve this, a numerical model of the human circulatory system was developed as a part of the PExSim (Process Explorer and Simulation) application [9]. It is a flexible tool that can be used for numerical calculations and simulations of the complex dynamic systems. It contains predefined function blocks which represent, among other things, fundamental mathematical and logical functions, basic dynamic and static elements and input/output components. The application, written in C++, can be easily expanded by additional plugins. As an extension, we developed a HCS (Human Circulatory System) library, which include a group of blocks representing functions of the circulation system elements. By creating a proper connection between individual blocks and changing some selected parameters, we can reproduce and simulate various physiological conditions (e.g. weakness of the left heart). What is more, the system developed can be applied to the analysis of data, and it can also predict a trend of some selected hemodynamic variables that cannot be measured. In order to develop a complete simulation tool, a model of a ventricle assist device has to be created. The present paper contains a description of an identification experiment of a POLVAD assist device. Various approaches to the problem of modelling the dynamics of an artificial heart chamber are presented, including the use of a modified left ventricle description and a model based on the fundamental principles of hydraulics. Comparative diagrams of the modelled volume waveforms and the corresponding measurement data are also demonstrated. The results of simulations of a cardiovascular system with a left ventricular assistance model using the above descriptions are presented.

Implementation of the circulatory system model

The first stage in developing a platform for automatic selection of control parameters for an artificial heart ventricle was to implement a model of the human circulatory system. One of the best models, widely reported in literature, is one with concentrated parameters, presented in the form of an analogy to an electrical system [5, 6]. The model

of the left and right ventricle [1] is based on Starling's law, 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 [5, 3]:
Ejection:

$$P_v(t) = \left(V_v(t) - V_{v0}(t)\right) \cdot E_v(t) \cdot E_{\max} \cdot f\left(V_v(t), \dot{V}_v(t), \dot{V}_{v\max}(t)\right) \tag{1}$$

Filling:

$$P_v(t) = A \cdot e^{k \cdot V_v(t)} + B \cdot e^{-j \cdot V_v(t)} + C \tag{2}$$

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

Flow values are strongly dependent on the shape and the maximum value of the elastance function, which is the most important part of the ventricle work description. In our approach, the time-varying elastance function was adopted according to G. Ferrari [5], and the change in ventricular contractility is achieved by modifying the value of the end-systolic elastance. The remaining blocks represent functions of other components of the circulatory system and are based on fundamental physical properties [4]. As a result of model implementation, a library of six functional blocks was created.

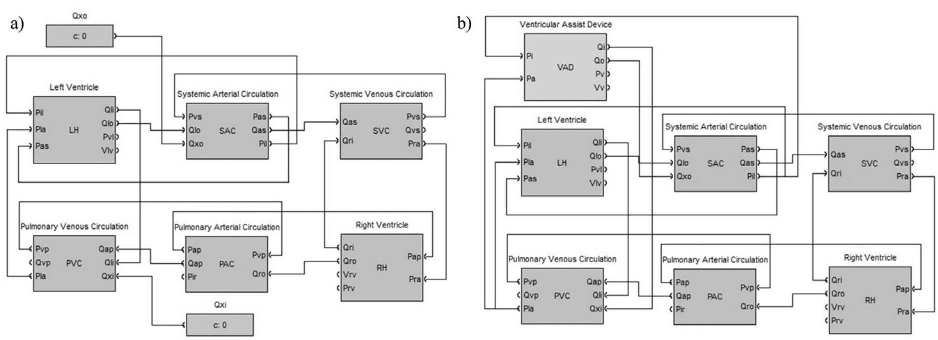

Figure 1. A model of human circulatory system in a PExSim package

They correspond to particular elements of the circulatory system such as the left and right ventricle of the heart (LH, RH), the systemic arterial circulation (SAC), the systemic venous circulation (SVC), the pulmonary arterial circulation (PAC), and the pulmonary venous circulation (PVC). The necessary mathematical relationships for the parallel assist device were established, and a simple model of a pulsatile pump, based on the description of the heart ventricle, was developed (VAD). The parameters of the blocks were matched so that the counter-pulse support could be simulated. The proper connection of the blocks creates a complete model of the human circulatory system with (Figure 1b) or without (Figure 1a) heart assistance.

As a result of the simulation, the waveforms of the main hemodynamic variables such as: flows, pressures and ventricular volumes can be reproduced. By analyzing the pressure-volume loop, an energetic balance can be described [2] and the stroke volume (SV) value can be calculated. Indirectly, we can also determine the cardiac output as a product of SV and the heart rate. It is a volume of blood pumped by the heart per minute and it constitutes the main parameter for evaluating heart efficiency. A sample of the pressure-volume loop obtained by the circulation system modelling is shown in Figure 2.

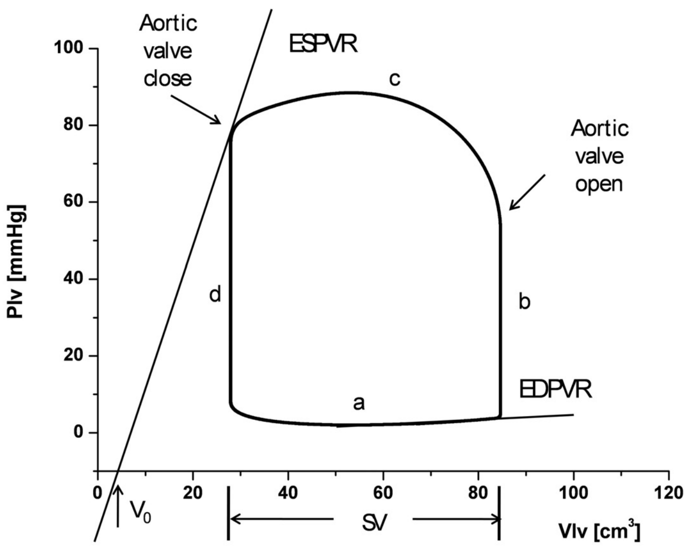

Figure 2. The left ventricle pressure-volume plane: filling (a), isovolumetric contraction (b), ejection (c), isovolumetric relaxation (d), the end-systolic pressure-volume relationship (ESPVR), and the end-diastolic pressure-volume relationship (EDPVR) [8]

The left ventricle failure can be modeled by change of LH block parameters: increase of the rest volume (V_0) and reduction of end systolic elastance value. As an effect we can observe the decrease of systemic arterial pressure, which is caused by insufficient efficiency of left ventricle. Furthermore, the blood is accumulating in pulmonary venous system, so the left atrium pressure is too high. The simulation results for pathological conditions are shown in Figures 3 and 5.

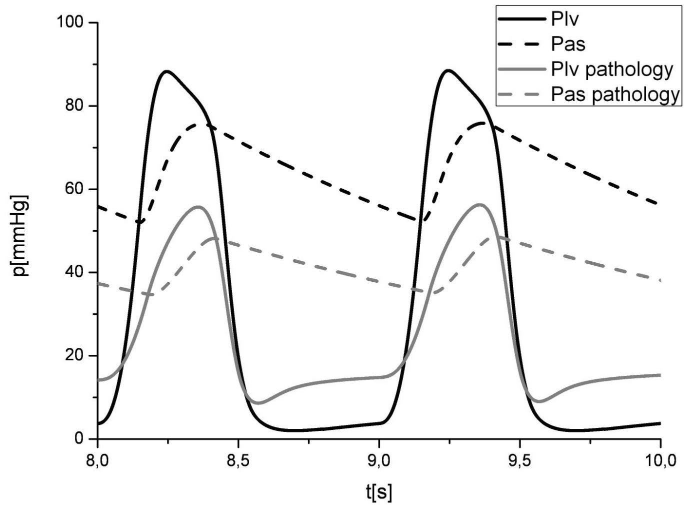

Figure 3. The left ventricular pressure (Plv) and the systemic arterial pressure (Pas) for a healthy heart and left ventricle pathology

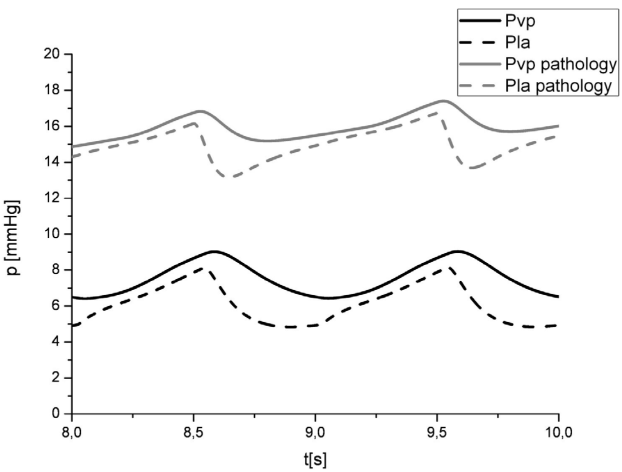

Figure 4. The venous pulmonary pressure (Pvp) and the left atrium pressure (Pla) for a healthy heart and left ventricle pathology

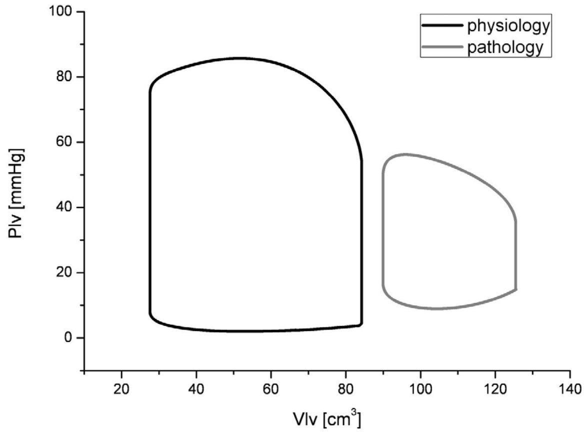

Figure 5. The left ventricle pressure-volume loop for physiological and pathological conditions

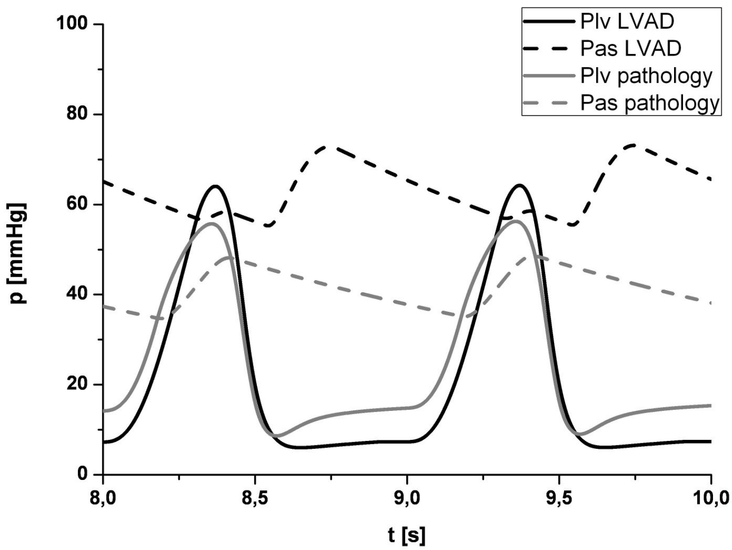

Figure 6. The left ventricular pressure (Plv) and the systemic arterial pressure (Pas) for a pathological condition, and a pathological condition with a parallel assistance (LVAD)

The addition of a simple left ventricle assist device model provides an opportunity for modeling circulation conditions with a parallel support. An artificial ventricle fills up during the heart systolic phase, and the ejection occurs when the natural ventricle is in a diastolic phase. This causes an increase in the systemic arterial pressure and a decrease in the atrium pressure. The heart is less loaded, which can be seen in the pressure-volume loop as a reduction of the stroke volume value. The expected changes are caused by the connection of a VAD block. The pulmonary venous pressure decreases (Figure 4), the systemic arterial pressure increases (Figure 3), and the pressure-volume loop shrinks (Figure 5).

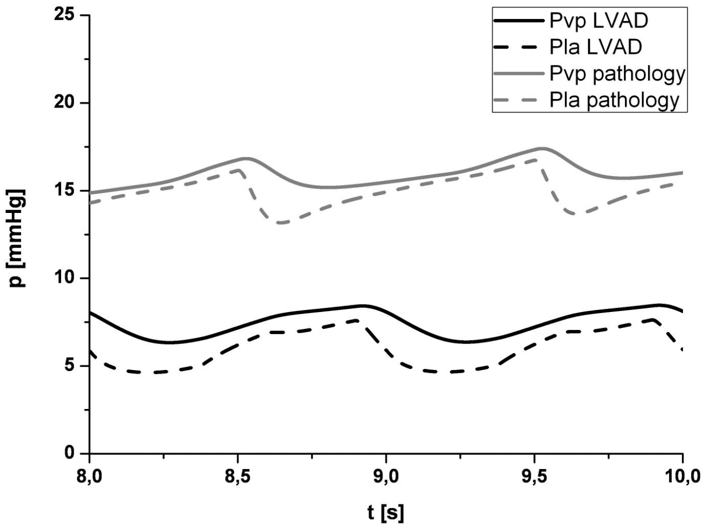

Figure 7. The venous pulmonary pressure (Pvp) and the left atrium pressure (Pla) for a pathological condition, and a pathological condition with a parallel assistance (LVAD)

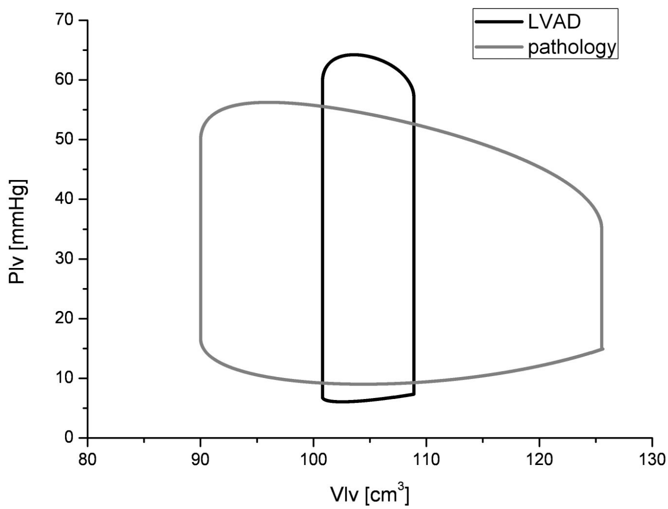

Figure 8. The left ventricle pressure-volume loop for a pathological condition, and a pathological condition with a parallel assistance (LVAD)

Identification experiment

The next step, after implementing the circulatory system model with a parallel support, is to create a proper model of the dynamics of a real assist device based on measurement results. For this purpose, a big 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 is a product of the Polish Artificial Heart Programme and has been clinically used since 1999 [7]. It consists of a pneumatic drive pulse pump with disc valves and two, blood/pneumatic, chambers separated by a flexible membrane (Figure 9), and a pneumatic driving unit POLPDU-401. The blood flow is a result of membrane movement, which is caused by air pressure changes in the pneumatic chamber. In a high-pressure phase, the output valve opens (input valve remains closed throughout) and the blood is ejected from the device. In a low-pressure phase, the working conditions are reversed, the input valve opens (the output valve is closed) and the blood fills up the chamber.

Figure 9. The extracorporeal ventricular assist device POLVAD

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 is based on a numerical description of the circulation system and includes impedance converters developed at the Nalecz Institute of Biocybernetics and Biomedical Engineering of the Polish Academy of Sciences [10]. These converters transfer the mathematically determined values from the numerical model to physical equivalents – real flows and pressures. Thus a possibility is provided of connecting and testing different types of assist devices. The simple method of changing parameters in the mathematical model made it possible to obtain different working conditions such as steady input and output loads. The following variables were measured: a supply pressure (Ppn), and pressures and flows in input and output cannula (Pin, Pout, Qin, Qout). As a result of the experiment, 643 data series for different supply and load conditions were obtained. They were used as a basis for modelling the dynamics of the ventricular assist device.

Modelling methods

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 modelling 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 Figure 10, 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.

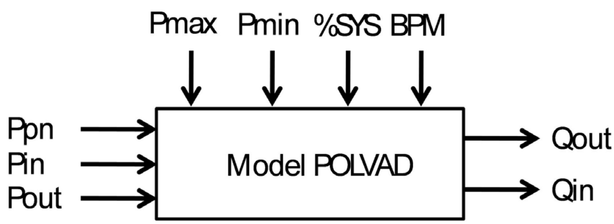

Figure 10. A schematic diagram of the dynamics of a ventricular assist device

The dynamic modelling of a ventricular assist device is not a trivial task. The change in parameters strongly affects the nature of the process, therefore making the operating range very wide. What is more, a model of the operation of mechanical valves must be developed. The next problem is a strong dependence of the process conditions from appearing of full filling and emptying of blood chamber. The original idea for obtaining a proper mathematical description of the chamber was to modify the shape of the elastance function. The valves, modelled as ideal diodes, were replaced by inertia elements. The method of determining the pressure in the blood chamber was also modified according to the relation:

$$P_v(t) = \left(V_v(t) - V_{v0}(t)\right) \cdot \left(E_{in}(t) \cdot E_{in\max} + E_{out}(t) \cdot E_{out\max}\right) \cdot f_1\left(V_v(t), \dot{V}_v(t), \dot{V}_{v\max}(t)\right) \quad (3)$$

where: P_v is the pressure in the blood chamber, V_v is the volume of the blood chamber, V_v0 is the rest volume of the chamber, E_{in}, E_{inmax} are the elastance function values and the maximum elastance values for the chamber filling, E_{out}, E_{outmax} are the elastance function value and the maximum elastance value for the chamber emptying.

The above description contains some unknown parameters such as the time constant of the valves, their input and output resistances, and the coefficient of shape and displacement of the elastance function. In order to determine all these parameters, a dedicated program called PExSim Optimizer [11], was employed. Among various calculation methods and quality indexes, the best result was obtained using the PSO (Particle Swarm Optimization) algorithm and the difference between the modelled and the measured value of flows adopted as a quality index. The comparison of the waveforms obtained is presented in Figure 11 and Figure 12.

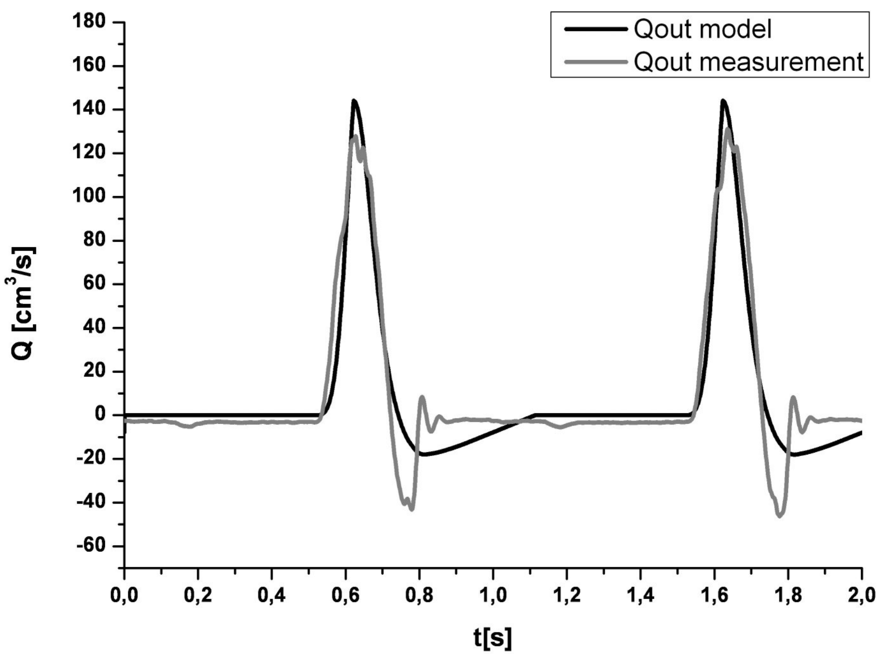

Figure 11. The modeled and measured output flow (Qout)

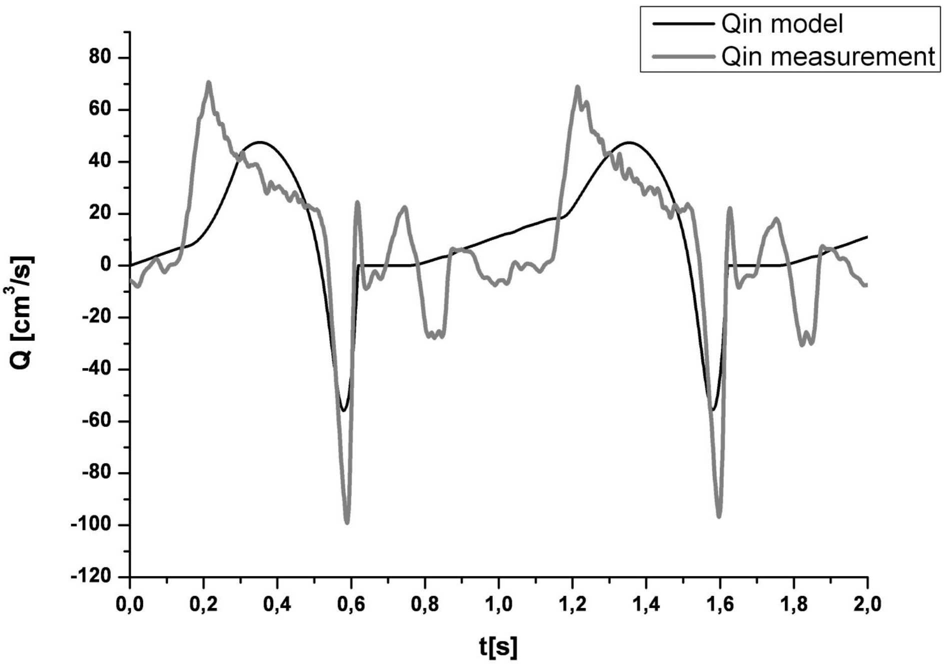

Figure 12. The modeled and measured input flow (Qin)

The waveforms obtained by modelling approximate those measured. However, the model presented here is independent from the control pressure value. It means that we can have parameters adjusted to a particular situation, but the description will never be universal. For each working condition, the parameters must be set independently. This procedure could be very difficult and time consuming, and the effect would probably not good enough because of a large number of possible operation settings.

Another approach to the description of the dynamics of an assist device was an attempt to determine the model 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 modelling the opening and closing of the valves. The 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 results of the simulation are presented in Figure 13 and Figure 14.

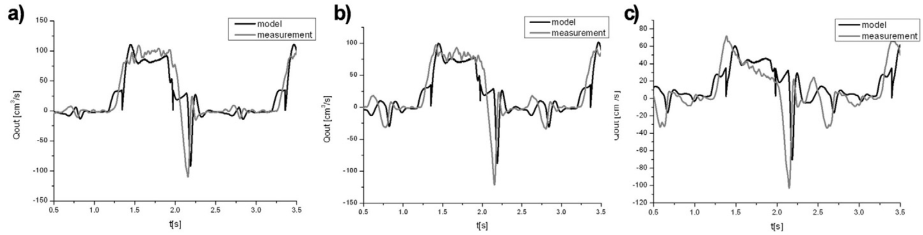

Figure 13. The modelled and measured assist device output flow (Qout) for various maximal control pressure values: (a) P_{max} = 175 mmHg, (b) P_{max} = 150 mmHg, (c) P_{max} = 100 mmHg

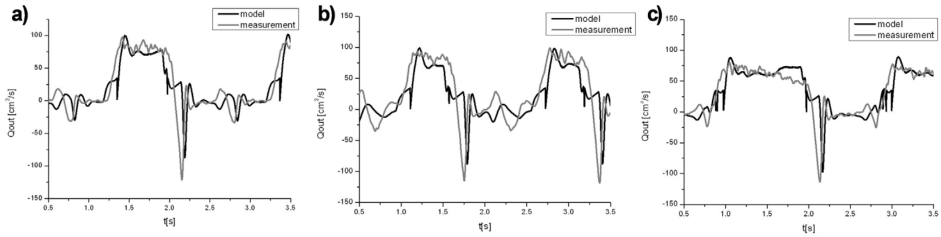

Figure 14. The modelled and measured assist device output flow (Qout) for the same maximal control pressure values and different time parameters: (a) BPM = 60 beats/min, %SYS = 40%, (b) BPM = 75 beats/min, %SYS = 40%, (c) BPM = 60 beats/min, %SYS = 60%

The waveforms obtained 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 pressures and time working conditions. 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 as a frequency of action.

Results of left ventricle assistance modelling

The models presented above were implemented in the PExSim program as two functional blocks. By using a proper connection of blocks, which are a part of the plugin, we can reproduce a complex model of the circulatory system with or without mechanical assistance. This method was used to simulate the left ventricle assistance on the model

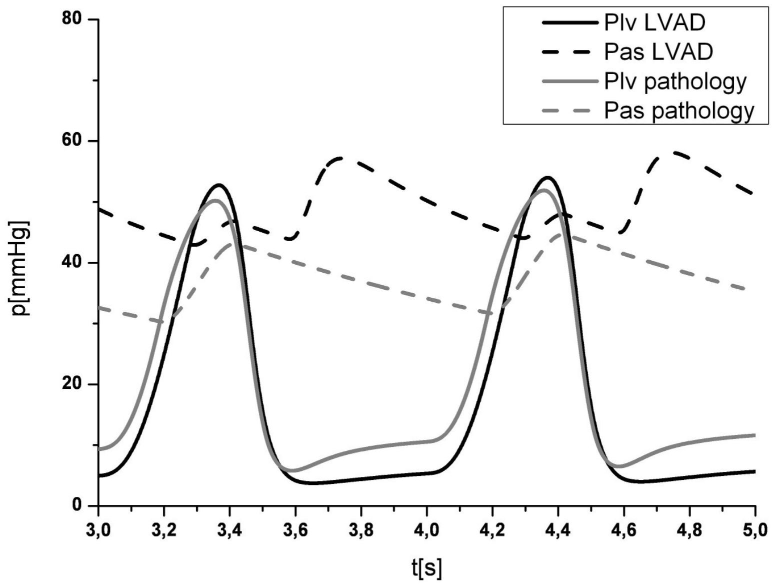

Figure 15. The results of modelling of the ventricular pressure (Plv) and systemic arterial pressure (Pas) for pathological conditions and pathological with assistance (LVAD) conditions

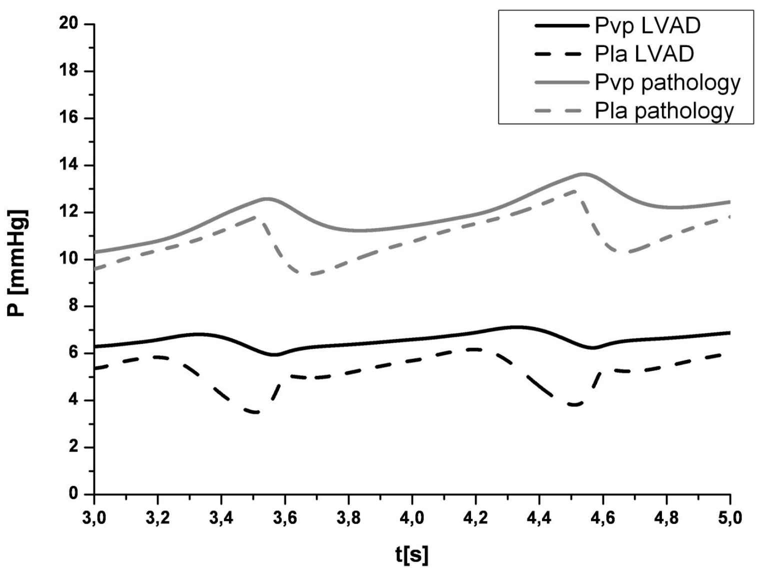

Figure 16. The results of modeling of the pulmonary venous pressure (Pvp) and atrium pressure (Pla) for pathological conditions and pathological conditions with parallel assistance (LVAD)

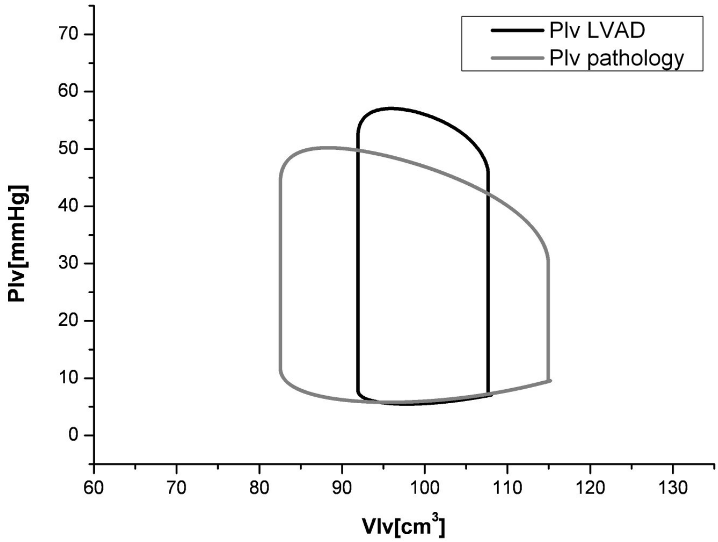

Figure 17. The results of modelling of the left ventricle pressure-volume loop for pathological conditions and pathological conditions with parallel assistance (LVAD)

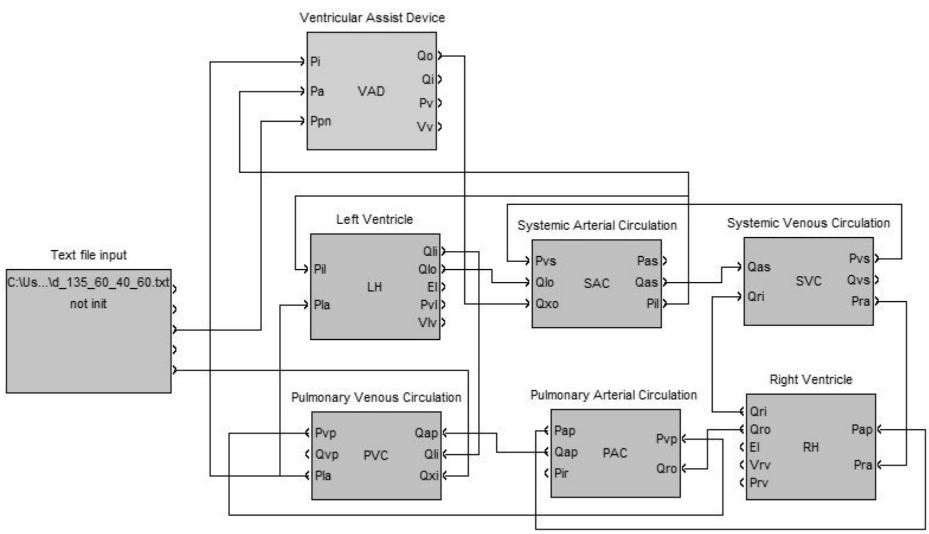

Figure 18. A model of the human circulatory system with the POLVAD model in a PExSim package

based on a modified elastance description (eq. 5) of the POLVAD device. The results of simulation with this parallel assistance are shown in Figures 15-17.

In this case, we can note the expected influence of the assistance on the main haemodynamic values. The heart is relieved (Figure 17), the atrium pressure decreases (Figure 16) and the systemic arterial pressure increases (Figure 15). For a modelled pulmonary venous pressure (P_{vp}), a small increase in the main value in time can be observed. This is the effect of fitting the model to measurements, when there is a quantity discrepancy between the total input and output flow for one work cycle. The elimination of this effect requires implementation of an algorithm which will ensure a balance between the stroke and filling volumes.

Next, a model based on basic physical relations was connected to a circulatory model. In this case, an LVAD block has an additional input for the control pressure value. This signal and the input flow value are taken from the measurement file (Figure 18). For this case, we also obtained an expected increase in the systemic arterial pressure (Figure 19) as a heart relief, which caused the stroke volume to decrease (Figure 20).

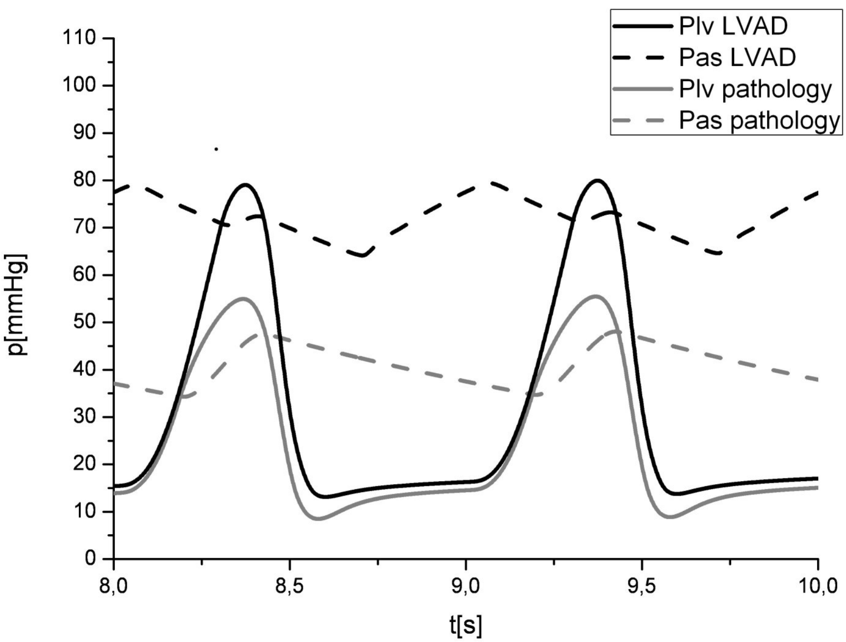

Figure 19. The results of modelling of the ventricular pressure (Plv) and the systemic arterial pressure (Pas) for pathological conditions and pathological conditions with assistance (LVAD)

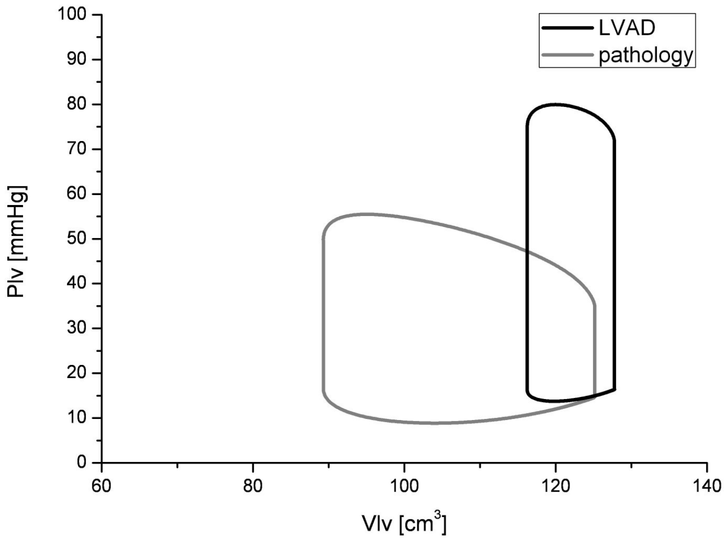

Figure 20. The results of modelling of the left ventricle pressure-volume loop for pathological conditions and pathological conditions with parallel assistance (LVAD)

Conclusions

As a result of our research, a complete PExSim plugin for the numerical modelling of the human circulatory system with parallel assistance has been developed. The implementation of this device as a function block, with the possibility of a simple change of parameters, makes possible to represent the system and to make a simulation of specific physiological conditions, both normal and pathological. The modification of the mathematical description of the circulatory model permits the connection of a ventricular assist device. A simple model for artificial heart ventricle was attached, which provided the possibility of studying the impact of the prosthetic operation on hemodynamic conditions. As a result of an extensive identification experiment for the artificial chamber POLVAD, unique results were obtained. They were used for dynamic modelling of the device. The descriptions based on a modified elastance function and basic physical relations were proposed. For each of them, the parameters optimization was made. The results of modelling the cardiovascular system are presented for normal physiological conditions, left ventricular failure and pathology with attached support models created on the basis of measurement data.

The future study should lead to a better description of the dynamics of the ventricle assist device and more accurate presentation of the way POLVAD works. Currently, investigations of automatic detection of the full discharge and full filling of the blood chamber are being carried out. This task is of fundamental importance because of the high risk of thrombosis in the case of incorrect chamber working conditions. Thrombosis could lead to blood vessel occlusion and the death of the patient. Detection of extreme positions is not a trivial task because it can only be determined on the basis of a control pressure analysis, which is the only measured signal, which is always available. By developing this type of an algorithm, it will also be possible to obtain a more accurate description of the physical phenomena in the artificial heart chamber, so that the model would be able to reproduce a wider range of device working conditions.

The platform developed in our study can be used to carry out various types of scientific research, including the development of control algorithms for artificial heart implants. It can be also employed as a useful tool in the learning process, improving the attractiveness of teaching and allowing a better understanding of the work of the human circulatory system. The proposed system can be also used in medical tasks, for example for modelling of gradual offloading of the patient heart by modifying the mechanical support level. Numerical simulation makes it also possible to create a model of values which are otherwise difficult or even impossible to measure in clinical conditions.

Acknowledgements

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