# Numerical Simulation of a Biventricular Assist Device with Pulsatile Control Operation for Bridge to Recovery

**Ronnachit Deepankaew**[1], **Arnon Plianchaisuk**[1], and **Phonphop Naiyanetr**[1],

## ABSTRACT

In this paper, the cardiovascular model was developed base on cardiovascular system to estimate the aortic pressure, pulmonary pressure, and blood flow during both left ventricular assist device (LVAD) and right ventricular assist device (RVAD) so called bi-ventricular assist device support (BiVADs). The physiological control system of BiVADs is developed for simulated a hemodynamics during BiVAD support. The cardiaovascular model, BiVAD model, and control system are performed using MATLAB /SIMULINK. The VAD-operating speed range can be varied between 7500 and 12,000 rpm in pulsatile control mode. At constant speed control mode, the pump operating speed during LVAD and RVAD support was 12,000 and 9400 rpm respectively. Control system is proposed to control blood flow to meet the cardiovascular system demand by using dynamics function of VAD. Parameters included the heart failure condition of both left and right ventricles that set to be 71% and 53% of normal heart contractility, respectively. Both normal and heart failure conditions have been simulated during BiVADs support via constant speed and pulsatile conditions. Results, in the control system constant speed vs pulsatile speed controls, show ejection fraction, pressure volume area, and ventricular efficiency of ventricle have been increased (improved) during pulsatile speed control mode comparison with constant speed mode. In conclusion, the pulsatile speed control mode of BiVADs system can increase the ventricular load during support. Therefore, the pulsatile speed control can potentially use for bridge to recovery treatment.

**Keywords**: Biventricular assist devices, control system, mechanical circulatory support.

## 1. INTRODUCTION

Left ventricular assist device (LVAD) is a mechanical circulatory support (MCS) device which is used to maintain normal hemodynamics in left ventricular heart failure (LVHF) patients. LVAD transplantation alleviates LVHF by driving excess blood remained in left ventricle into systemic circuit to reduces pressure and volume inside left ventricle during diastole[1] and improving reverse remodeling process and contractility of cardiac tissue[2]. However, single LVAD implantation is not commonly used as a bridge to recovery (BTR). There is an evidence reported that up to 30% of patients with heart failure experiences biventricular failure after get LVAD implantation[3]. Despite the benefits of LVAD, systemic blood flow accretion increases venous return and then induces right ventricular heart failure (RVHF) consequently. Therefore, Biventricular assist devices (BiVADs) become essential for supporting both ventricles. Additional right ventricular support decreases ventricular volume by unloading in right ventricle then improves reverse remodeling process of right ventricle[4].

For BTR concept, pulsatility significantly associates with maintenance of normal cardiovascular tissue functions through reverse remodeling process[5]. Comparing to constant speed LVAD, which does not generate pulse pressure (PP) in cardiovascular system (CVS), several computational models with various types of LVAD elicited better capabilities of pulsatile LVAD than constant-speed VAD to maintain hemodynamic function[6, 7, 8]. Pulsatile LVAD also showed better left ventricular systolic and diastolic functions by reduction in extracellular matrix (ECM) turnover and fibrosis in heart failure patients[9].

Therefore, pulsatile flow control of BiVADs may be beneficial to recover the function of cardiovascular tissue in heart failure patients. However, the studies of BiVADs pulsatility control for BTR are not sufficient. This study aimed to investigate and compare capabilities between constant-speed and pulsatile control operation in BiVADs as a BTR for heart failure patients by simulation of hemodynamic parameters from computational CVS models.

## 2. MATERIALS AND METHOD

### 2.1 Model Description

The numerical model was implemented in SIMULINK®(Math Works, Inc., Natick, MA, United States) using the ODE45s solving algorithm. Pre and post-processing of the values is performed in MATLAB®. A schematic diagram of the CVS and BiVADs model that is used to develop this simulation is shown in Figure 1. The model of the CVS consists of lumped parameter blocks; each characteristic by resistances (R), compliances (C), inertances (L), elastances (E), diodes (D), flow (Q) and pressure (P). In

Manuscript received on August 9, 2015 ; revised on December 12, 2015.
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its simple model, the CVS has four parts including the pulmonary and systemic circulations, the right heart and left heart. The CVS model parameters have been adjusted to procreate pressure, flow and volume distributions in a healthy human [10]. As reported in [10], the model has been carefully validated using published data and the parameters were then adjusted from the healthy condition to heart failure condition as a precondition for simulations. The most relevant parameters included blood volume, systemic peripheral resistance, and contractility of the LV and RV end-systolic elastances so called Maximum elastane (Emax), are given in Table 1. In addition to the structure of the circulatory model, the parameter values used will be discussed and compared to published results[8]. As the numerical simulation will be used to simulate BiVADs support, discussion will be made regarding the changes to the models parameters for healthy and heart disease patients. To integrate the both LVAD and RVAD models into the circulatory system model, a VAD model for both the pressure and flow characteristics of the device is developed using axial flow VAD based on the Micromed DeBakey VAD (MicroMed Cardiovascular Inc., Houston, TX, USA). The differential equation is the measured pressure difference, and the control input is the pump rotational speed ($\omega$) which can be calculated from the pump characteristic equation (1)[11].

$$H(t)B = B - \omega^2 - R_p - Q(t) - L_n - (dQ(t)/dt) \quad (1)$$

The meaning of each parameters compose of H(t) represented head pressure, Q(t) represented the pump flow, $\omega$ represented the pump rotational speed while B, $R_p$ and $L_p$ represented a constant speed related coefficient, the constant slope of the linear pressure-flow characteristics, and the constant of fluid inertance respectively. (B = 8.56e-5 mm Hg$\times s^2/rad^2$, $R_p$ = 0.52 mm Hg×s/mL, and $L_p$ =0.063 mm Hg$\times s^2$/mL). The BiVADs components operate between 7,500 and 12,000 rpm.

## 2.2 Control Systems Description

In this control system, this study base on high speed pump, it is assumed that while the aortic valve is closed. The pump flow can be presumed as a total cardiac output. The input reference signal to the system has been selected, arbitrarily, to be at around a physiological value, which is about 5 L/min. The system is derived with feedback control mechanism incorporating a reference model, where the proportional-integral-derivative (PID) is implemented as feedback part over a cardiac cycle. The input of the Control system is the estimated pump flow, while the outputs of the controller are speed of BiVADs, both on constant mode and pulsatile mode.

### 2.2.1 Varying speed control

In this method, to regulate the BiVADs operating speed, the reference signal flow rate was selected as

***Table 1:***: Model parameters to simulate the HF condition

| **Variable** | **Symbol** | **Unit** | **Healthy** | **Heart failure** |
|---|---|---|---|---|
| Left ventricular contractility | Elv | mmHg/mL | 3.54 | 0.71 |
| Right ventricular contractility | Erv | mmHg/mL | 1.75 | 0.53 |
| Systemic peripheral resistance | Rsa | mmH/.mL | 0.74 | 1.11 |
| Total blood volume | Vtotal | (mL) | 5300 | 5800 |

the control variable with PID control as a control method. For PID control application, the output of a PID controller, equal to the control input to the plant, in the time-domain is as follows (2).

$$U(t) = K_p e(t) + K_i \int e(t)dt + K_d \frac{de}{dt} \quad (2)$$

This error signal (e) will be computed both the derivative and the integral at the PID controller. The control signal (u) to the plant is there by equal to the summation of the product of proportional gain ($K_p$) and the magnitude of the error, the product of integral gain ($K_i$) times and the integral of the error, and the product of the derivative gain ($K_d$) and the derivative of the error. Therefore, the transfer function is defined in equation (3).

$$G(S) = K_p + \frac{K_1}{S} + K_D S \quad (3)$$

$K_p$, $K_I$ and $K_D$ represent the proportional, integral and derivative gains of the controller, respectively. The values of these constant gains, with minimal overshoot, were adjusted to achieve a 5% settling time of 10 seconds.

### 2.2.2 Pulsatility control

The reference flow rate ($Q_{ref}$), modeled by a sinusoidal function in pulsatile-speed BiVADs assistance (Equation (4)): as shown in Figure 2.

$$Q_{ref} = \begin{cases} Asin2\pi * 2t * 2T, & t \leq T/4 \\ 0, & t > T/4 \end{cases} \quad (4)$$

,where $A$ is the amplitude of the reference signal, $t$ is the relative time and $T$ is the cardiac cycle duration. During the simulation, the reference flow signal, resembling aortic systolic flow, consist of the positive portion of a sinusoidal signal for the first quarter and is remained zero for the remainder period of the cardiac cycle. To achieve a synchronized between the

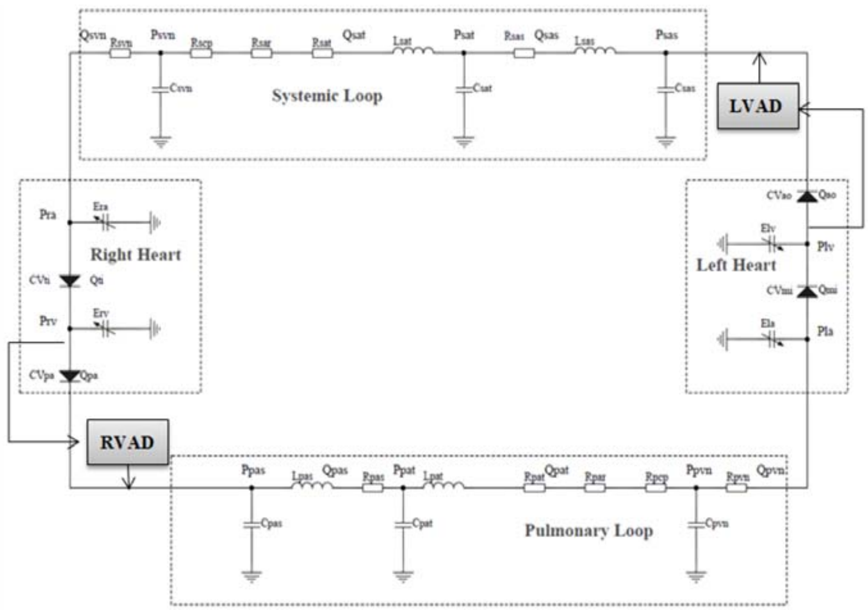



***Fig.1:***: Illustration of electrical equivalent circuit analogue of CVS - BiVADs interaction.

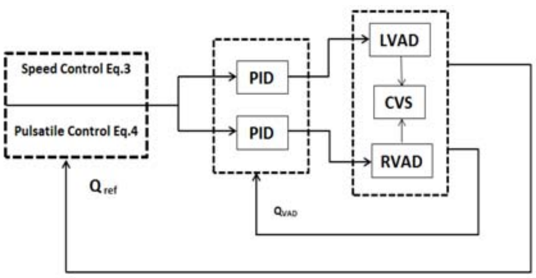



***Fig.2:***: Block diagram of pulsatile control.

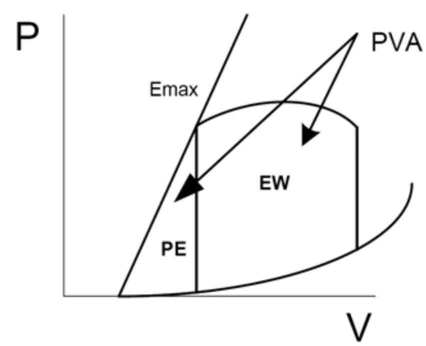



***Fig.3:***: Illustration of the ventricular pressurevolume (P-V) loops PVA, pressurevolume area; PE, potential energy; EW, external mechanical work.

reference flow signals and the beating heart, a triggering algorithm was developed [12]. For pulsatile-speed support, the operating speed varied between 7500 and 12,000 rpm over a cardiac cycle.

### 2.2.3 Simulation Study Protocols

Numerical simulations are simulated with the heart rate (HR) preset at 75 bpm, and the ventricles are characterized by 100% of contractility as normal ventricles and reduced the percentage of contractility to simplify the heart failure condition. Once the simulation is started, after the steady non-assisted pathologic state is reached, the BiVADs model is switched on (after 30 seconds). The BiVADs model is made to work in two different modes of control, constant flow and pulsatile flow (synchronized with the working heart). The steady-state performance of pulsatile control is compared to the constant speed mode. For all of the simulations, hemodynamic variables are waited to be settled and steady before transitioning from the baseline into the testing conditions. The simulations are continued until the steady state post-transition are achieve.

### 2.2.4 Monitoring and calculation of derived parameters

The derived parameters such as stroke volume (SV), pulse pressure (PP), ejection fraction (EF) can be determined directly using pressure-volume loop (PV loop) of the ventricle. Furthermore, the area is relate to in the PV loop also represented the function of the heart regarding energy in two parts. First, the area inside the PV loop is defined as the ventricle stroke work (SW), or also refers as external work (EW), which is the energy subjected to the blood by the contraction of the ventricle. Hence the stroke work may be used to evaluate the function of the heart. The second part is the elastic potential energy store in the left ventricle which can be represented by

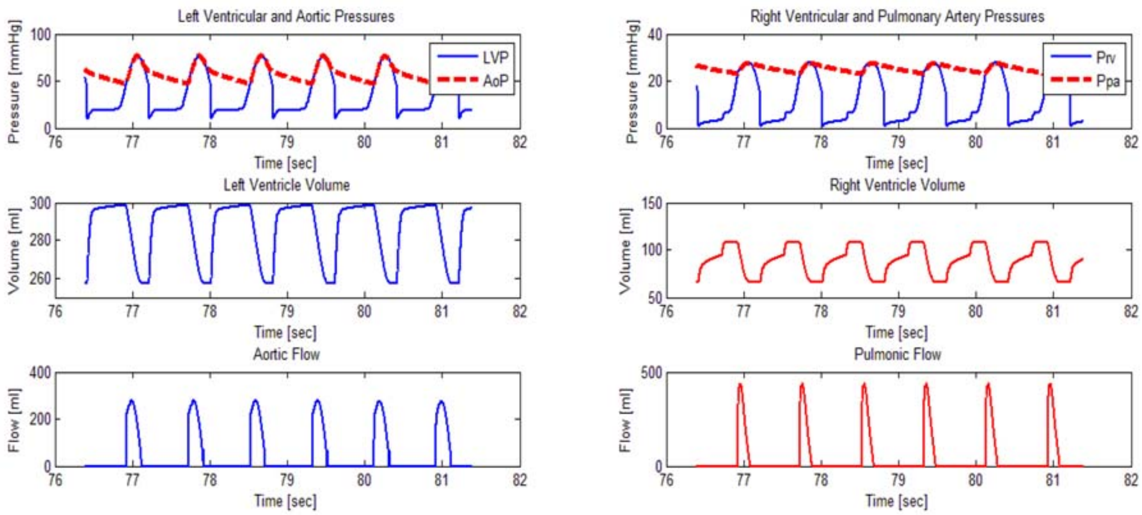
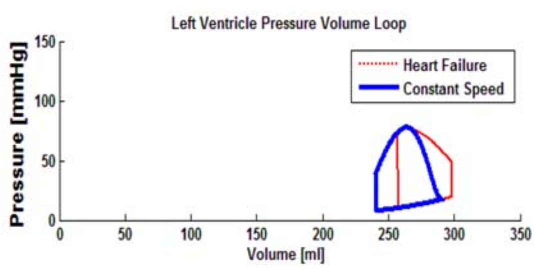
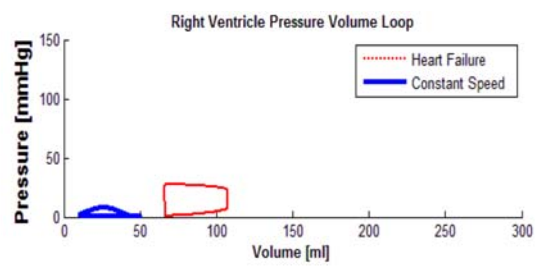



***Fig.4:***: Pressure curves are shown of the heart failure at 100% contractility left and right ventricular (AoP: arterial pressure, LVP: left ventricular pressure, Ppa: pulmonary arterial pressure, Prv: right ventricular pressure).




(a)

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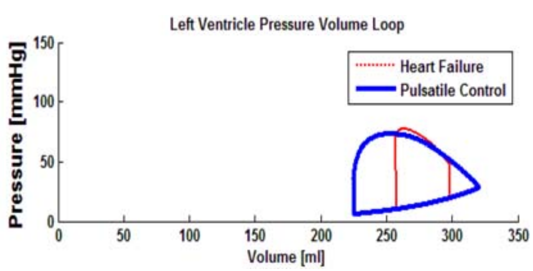


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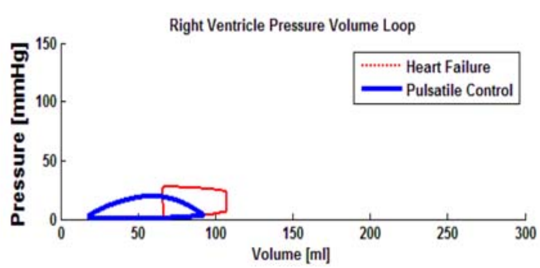


(d)

***Fig.5:***: Comparison of pressure-volume loops in heart failure with or without BiVADs assistance: (a) left ventricle assisted by constant speed, (b) right ventricle assisted by constant speed, (c) left ventricle assisted by pulsatile control, and (d) right ventricle assisted by pulsatile control.

***Table 2:***: Simulated hemodynamic data at healthy and heart failure with different control system.

| **Computational Model** | | | | |
|---|---|---|---|---|
| | **Healthy** | **Heart Failure** | **Constant Speed** | **Pulsatile Control** |
| **HR (bpm)** | 75 | 75 | 75 | 75 |
| **SBP (mmHg)** | 114.70 | 76.85 | 120.60 | 119.20 |
| **DBP (mmHg)** | 80.02 | 46.97 | 110.70 | 59.12 |
| **PP (mmHg)** | 34.68 | 29.88 | 9.90 | 60.08 |
| **MAP (mmHg)** | 94.3 | 58.48 | 114.0 | 79.15 |
| **CO (L/m)** | 5.43 | 3.71 | 4.40 | 5.40 |
| ***Left Ventricular*** | | | | |
| **EF (%)** | 65.4 | 13.8 | 17.6 | 29.7 |
| **Qao (L/m)** | 5.4 | 3.71 | 0 | 0 |
| **QLVAD (L/m)** | 0 | 0 | 4.4 | 5.4 |
| **EW (ml • mmHg)** | 3451 | 2089 | 2137 | 5902 |
| **PE (ml • mmHg)** | 1905 | 8446 | 8110 | 7346 |
| **PVA (ml • mmHg)** | 5356 | 10535 | 10247 | 13248 |
| **VE** | 0.64 | 0.20 | 0.21 | 0.44 |
| ***Right Ventricular*** | | | | |
| **EF(%)** | 62.3 | 38.4 | 79.1 | 80.2 |
| **Qpo (L/m)** | 5.4 | 3.71 | 0 | 0 |
| **QRVAD (L/m)** | 0 | 0 | 4.4 | 5.4 |
| **EW (ml • mmHg)** | 158 | 69 | 56 | 143 |
| **PE (ml • mmHg)** | 587 | 816 | 51 | 80 |
| **PVA (ml • mmHg)** | 745 | 885 | 107 | 223 |
| **VE** | 0.21 | 0.08 | 0.52 | 0.64 |

the triangular are under Emax line. Furthermore, the ratio of EW to pressure-volume area (PVA) can be considered as an indicator for the ventricular working efficiency (VE; Equation (5)) [13]; as shown in Figure 3.

$$VE = \frac{EW}{EW + PE} = \frac{EW}{PVA} \quad (5)$$

## 3. RESULTS

Figure 4 shows typical results of pressure, volume and flow patterns in the time domain (condition presented at arterial pressure (AoP) = 77 mmHg, severe left and right heart failure, LVAD and RVAD assisting at a speed of 12,000 rpm and 9,400 rpm respectively). MAP, PP, SBP, DBP, CO increase compare with heart failure.

Figure 5 shows PV loops under constant speed conditions on the ventricle have the ejection fraction and ventricular efficient (EW/PVA) lower than pulsatile control. Table 2 summarizes the simulated hemodynamic data the pump speed and pulsatile control.

HR, heart rate; MAP,mean arterial pressure; SBP, systolic blood pressure; DBP, diastolic blood pressure; PP, pulse pressure; CO, cardiac output; EF, ejection fraction; Qao, aortic mean flow; Qpo, pulmonary mean flow; QLVAD, LVAD mean flow; QRVAD, RVAD mean flow; EW, external work; PE, potential energy; PVA, pressure-volume area;VE, ventricular working efficiency.

## 4. DISCUSSION AND CONCLUSION

At the present, medical technology is well developed to support heart failure by VADs implantation for bridge to transplantation (BTT) purpose. However, the number of patients supported by VADs who waited for heart transplantation still continually rises. From OPTN/SRTR 2013 Annual Data Report, the proportion of American heart transplantation candidates with VADs substantially increases from 7.5% in 2003 to 27.4% in 2013 [14]. The median waiting for candidates with VADs increased sharply from 6.2 months for candidates listed in 2010-2011 to 9.3 months for candidates listed in 2012-2013 [14]. With the trend of waiting time extension, VADs insertion will be more useful if they can act as BTR for maintenance of CVS regular function during waiting for new heart transplantation. BiVADs have more potential to be a BTT than LVAD by assisting both ventricular functions with lower risk of RVHF occurrence in long term transplantation. [4] Moreover, BiVADs also operates reverse remodeling of cardiovascular tissue to keep up CVS regular functions.

Our results exhibited the competency of both constant and pulsatile BiVADs to sustain normal homeostatic level of adult hemodynamic parameters including systolic blood pressure (SBP; 90-140 mmHg), diastolic blood pressure (DBP; 60-90), pulse pressure (PP; 0-80 mmHg), mean arterial pressure (MAP; 70-105 mmHg), and cardiac output (CO; 4.0-8.0 L/min). [15] There is no difference between these parameters due to types of BiVADs except EF and VE.

EF measurement displays percentage of blood leaving from filled ventricle in each heart contraction. Heart failure group showed diminution in left ventricular EF from 65.4% to 13.8% and right ventricular EF from 62.3% to 38.4%, normal to heart failure respectivelty. After BiVADs were switched on, pulsatile control BiVADs prominently improved EF of left ventricle (29.7%) when constant control BiVADs gave a slight increase in left ventricle EF (17.8%). For right ventricle, pulsatile mode and constant speed mode can increase EF to 80.2% and 79.1%, repectively. The EF of right ventricles had no effects from different types of BiVADs control. To sum up the model, pulsatile control BiVADs increased left ventricular ejection fraction in heart failure patients.

Although EF increase when BiVADs were supported, EF value still cannot refers to function of heart in blood pumping process. So, VE was measured to determine ventricular pump performance. From our model, VE of left and right ventricles in heart failure patient dropped from 0.64 and 0.21 as healthy conditions to 0.20 and 0.08 respectively. The results displayed VE increase only by pulsatile control BiVADs support (VE = 0.42). And this indicates capability of pulsatile control BiVADs to improve ventricular function in heart failure patient.

The numerical model could be developed which can be used for modeling the interaction between the cardiovascular system and BiVADs. The model can be used for repeatable observations and comparisons of effects caused by different parameter variations, but cannot obviously reproduce some quantitative data particularly right ventricle that not clearly limits the applicability of physiological model is used.

Pulse is physiologically important for human body because pulse regulates vasodilation and vascular remodeling through various cellular signaling pathways in endothelial cell, smooth muscle cell and fibroblast cell inside circulatory system. [5] To conclude, our simulation study manifested the capabilities of pulsatility control in BiVADs to sustain cardiovascular function in heart failure patients by control of hemodynamic parameters and enhancement in EF and VE values. To consider flow through semilunar valves and BiVADs after BiVADs was fully supported, blood flow completely transmitted through BiVADs instead of semilunar valves ($Q_{ao} = 0$ and $Q_{po} = 0$). Therefore, our future study will perform BiVADs partial support to adjust amount of flow through semilunar valves in order to recover cardiac pumping functions by time.

## 5. ACKNOWLEDGEMENTS

The authors would like to thank all members of Cardiovascular Engineering & Artificial Organ Laboratory (CardioArt Lab) and all lecturers from the Department of Biomedical Engineering, Faculty of Engineering, Mahidol University.

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