SPEED CONTROL OF LEFT VENTRICULAR ASSIST DEVICE USING PID CONTROLLER WITH PARAMETER OPTIMIZATION BASED ON PSO

A PROJECT REPORT

submitted by

BENZY J R HYDE
(Reg. No. TKM20EEII08)

to

the APJ Abdul Kalam Technological University
in partial fulfillment of the requirements for the award of the Degree

of

Master of Technology
in
Electrical and Electronics Engineering

with specialisation in

Industrial Instrumentation and Control

Department of Electrical and Electronics Engineering
TKM College of Engineering
Kollam - 691005
KERALA

JULY 2022

DECLARATION

I undersigned hereby declare that the project report entitled "**Speed Control of Left Ventricular Assist Device Using PID Controller with Parameter Optimization Based on PSO**", submitted for partial fulfillment of the requirements for the award of degree of Master of Technology in Electrical and Electronics Engineering with specialisation in Industrial Instrumentation and Control, of the APJ Abdul Kalam Technological University, Kerala is a bonafide work done by me under supervision of *Prof. Fathima M Kasim*, Assistant Professor, Department of Electrical and Electronics Engineering. This submission represents my ideas in my own words and where ideas or words of others have been included. I have adequately and accurately cited and referenced the original sources. I also declare that I have adhered to ethics of academic honesty and integrity and have not misrepresented or fabricated any data or idea or fact or source in my submission. I understand that any violation of the above will be a cause for disciplinary action by the institute and/or the University and can also evoke penal action from the sources which have thus not been properly cited or from whom proper permission has not been obtained. This report has not been previously formed the basis for the award of any degree, diploma or similar title of any other University.

Kollam
July 01, 2022

BENZY J R HYDE

DEPARTMENT OF ELECTRICAL AND ELECTRONICS ENGINEERING

TKM COLLEGE OF ENGINEERING

KOLLAM - 691005

CERTIFICATE

This is to certify that the report entitled " **Speed Control of Left Ventricular Assist Device Using PID Controller with Parameter Optimization Based on PSO** " submitted by **BENZY J R HYDE** , **(Reg. No. TKM20EEI108)** of fourth semester to the APJ Abdul Kalam Technological University in partial fulfillment of the requirements for the award of the Degree of Master of Technology in Electrical and Electronics Engineering with specialisation in Industrial Instrumentation and Control, is a bonafide record of the project work done by her under our guidance and supervision. This report in any form has not been submitted to any other University or Institute for any purpose.

Prof. Fathima M Kasim
Project Supervisor
Assistant Professor
Department of EEE
TKM College of Engineering

Prof. Sumayya Jaleel
Project Coordinator
Assistant Professor
Department of EEE
TKM College of Engineering

Prof. Shanavas T N
PG Coordinator
Associate Professor
Department of EEE
TKM College of Engineering

Dr. Sabeena Beevi K
Head of the Department
Associate Professor
Department of EEE
TKM College of Engineering

Acknowledgement

A lot of effort and hard work has been put into this project in course of its presentation. However, it would not have been possible without the kind support and help of many individuals and other sources. I would like to extend my sincere thanks to all of them. I take this opportunity to express my deep sense of gratitude and sincere thanks to all who helped me to complete this project report successfully.

I express my sincere thanks to *Dr. T A Shahul Hameed*, Principal, TKM College of Engineering for his encouragement in the completion of my project.

I thank *Dr. Sabeena Beevi K*, Head of the Department, Department of Electrical and Electronics Engineering, *Dr. Imthias Ahamed T P*, Professor, TKM College of Engineering, and *Prof. Shanavas T N*, PG Coordinator, Department of Electrical and Electronics Engineering for their support and cooperation.

I am greatly thankful to my Project Supervisor *Prof. Fathima M Kasim*, Assistant Professor, Department of Electrical and Electronics Engineering for her supervision, assistance and helpful suggestions.

I am deeply indebted to *Prof. Sumayya Jaleel*, Assistant Professor, Department of Electrical and Electronics Engineering, and *Prof. Amal A*, Assistant Professor, Department of Electrical and Electronics Engineering, for their excellent guidance, positive criticism and valuable comments.

Finally I thank my parents, friends, near and dear ones who directly and indirectly contributed to the successful completion of my project.

BENZY J R HYDE

Abstract

Our body's most vital organ, the heart, keeps us alive by distributing blood enriched with oxygen and nutrients throughout the body. Heart Failure, on the other hand, is the medical term used when the heart cannot keep up with the demands imposed on it while pumping. There are an estimated 26 million cases of heart failure worldwide, and there are undoubtedly million more cases that go untreated. The best solution for long-term survival in advanced Heart Failure (HF) is a heart transplant, but the lack of donors necessitated the development of another treatment known as VADs (Ventricular Assist Devices). VADs cannot replace a heart, however they can assist the heart. VADs are mainly classified as LVADs (Left ventricular Assist Device), RVADs (Right Ventricular assist Devices), and BiVADs (Bi Ventricular Assist Devices). LVADs are mechanical pumps that are implanted in the left ventricle of patients with left ventricular failure inorder to provide support and meet the function of human heart (pumping blood). Perfusion and suction avoidance are particularly difficult with LVADs though. The LVAD is therefore controlled to maintain physiologic perfusion using a straightforward control method based on PID that simply uses the intrinsic pump measurement. The controller's goal is to keep a differential pump speed setpoint constant. In order to improve the performance of the system, a PID parameter optimization technique based on particle swarm optimization (PSO) is utilised . The effectiveness of the PSO based PID controller was investigated, and the proposed control algorithm proved capable of maintaining the perfusion demand.

Contents

List of Tables

List of Figures

Abbreviations

ABC	Artificial Bee Colony
BiVAD	Bi Ventricular Assist Devices
BLDC	Brushless Direct Current motor
CVS	Cardiovascular System
HF	Heart Failure
IAE	Integral Absolute Error
ISE	Integral Square Error
ITAE	Integral Timed Absolute Error
ITSE	Integral Timed Squared Error
LA	Left Atrium
LV	Left Ventricle
LVAD	Left Ventricular Assist Devices
PID	Proportional Integral Derivative
PSO	Particle Swarm Optimization
RA	Right Atrium
RV	Right Ventricle
RVAD	Right Ventricular Assist Devices
TIC	Taylor Instrumental Company
VAS	Ventricular Assist Devices

Notations

R	Resistances, Ω
P	Pressures, $mmHg$
V	Volumes, V
F	Flow rates, ML/sec
T	Time period, s
C	Compliance, $\frac{m^3}{mmHg}$
L	Inertance, $\frac{m^2}{Kg}$
J	Inertia, $kg.m^2$
K_b	Back emf constant, $\frac{kg.m^2}{A.s^2}$
B	Damping coefficient, $\frac{kg.m^2}{s}$

Chapter 1

INTRODUCTION

1.1 Overview

The heart is a powerful, muscular pump that is roughly the size of a fist in a healthy individual. It continuously circulates blood throughout the body. Through a series of meticulously timed contractions of its four chambers, the heart delivers blood to the lungs and all of the body's tissues. The four chambers of the heart must pulse in a coordinated manner for efficient operation. Heart failure, on the other hand, is the condition in which the heart is unable to function effectively. Heart failure (HF) is one of the major causes of death and is a chronic condition. The heart initially makes an effort to make up for this by:
1)Enlarging: To keep up with the need to pump more blood, the heart extends in order to contract more forcefully. The heart enlarges as a result over time.
2) Increasing muscle mass: The larger cardiac contracting cells are the cause of the increase in muscle mass. This allows the heart to pump more vigorously—at least at first.
3) Swift pumping:The output of the heart is boosted as a result.
The body also makes various attempts to make up for it:
1) Blood arteries constrict in an effort to maintain high blood pressure and compensate for the heart's diminished function.
2) The body diverts blood away from the heart, brain, and less crucial tissues and organs (such as kidneys).
These band-aid solutions only cover up the heart failure issue; they do not address it. Heart failure persists and gets worse until these compensatory mechanisms stop functioning. There-

fore this deadliest problem has to be treated. Heart transplantation is the recommended course of treatment for end-stage heart failure patients, however only 5 to 10 percent of these patients actually obtain a transplant due to a severe lack of donor hearts. Thus emerged the necessity of VADs(Ventricular Assist Devices).

A failing heart can be partially or entirely replaced with a ventricular assist device (VAD), which is a mechanical pump that helps with circulation . VADs are brought in to help either the right ventricle or the left ventricle, or both ventricles. Since left ventricular failure characterises the majority of end-stage HF patients, they may decide to undergo the necessary LVAD therapy. In comparison to the first generation pulsatile-flow left ventricular assist devices (PFVADs), continuous-flow left ventricular assist devices (CFVADs) [1] are currently more widely used because they have shown improved durability and longevity. However, as the length of long-term support increases, it appears that CFVADs may have specific complications and a slower rate of left ventricular recovery associated with diminished pulsality.

A linear proportional integral derivative (PID) controller is employed to overthrow all these difficulties and satisfy the perfusion demand. To meet design constraints and achieve desired performance, PID tuning must be performed in order to determine the values of proportional, integral, and derivative gains. In this paper, we try to build a PID controller using parameter optimization based on PSO. The PSO employs swarm intelligence. The method was inspired by observations of biological behaviour and social interaction in fish and birds. As a result, the perfusion demand is maintained by controlling the speed of the axial flow pump using a PID controller based on the PSO algorithm.

1.2 Objective

VADs(ventricular Assist Devices) are mechanical pumps that are used for persons with left heart failure. The primary difficulty faced by LVADs are physiologic perfusion and suction prevention. So, a new control algorithm that can overcome the difficulties without using embedded sensors have to be developed. The specific objectives are as follows:

(1) Modelling of the Human Circulatory System.

(2) Modelling of the LVAD.

(3) Integration of the LVAD into cardiovascular model.

(4) Develop a proportional– integral- derivative controller for left ventricular assist device.

(5) Tuning of the PID controller to obtain better results.

(6) Validation of the proposed PSO based controller through MATLAB/Simulink..

1.3 Organisation of the report

This report is organised as follows, Chapter 2 presents a deep literature review on various methods to overcome the challenges faced by LVAD. Chapter 3 describes the whole cardiovascular system. Chapter 4 presents the modelling of the Cardiovascular System. Chapter 5 discusses about the proposed system. Modelling of the controllers and the tuning is provided in Chapter 6. Simulation Results are presented in Chapter 7. Chapter 8 includes the conclusion suggestions concerning the work and future scope.

Chapter 2

LITERATURE REVIEW

2.1 Overview

VAD(ventricular assist devices) are in great need due to increase in the number of Heart failure patients. A ventricular assist device (VAD) is an electromechanical device for assisting cardiac circulation, which is used either to partially or to completely replace the function of a failing heart. Since, left ventricular failure characterises the majority of end-stage HF patients, it is necessary to undergo LVAD therapy. A LVAD's primary control goal is to imitate the heart's typical reaction to fluctuations in the demand for cardiac output. Hypoperfusion, damaged and constricted arteries, and other problems arise when the heart demand is not met. Due to the extremely nonlinear, time-varying, and discontinuous character of the human circulatory system (blood flow is not continuous due to the presence of valves), physiological perfusion [2] is particularly difficult with LVADs. In order to overcome these challenges, different control techniques were brought up by many researchers that are discussed further.

2.2 Existing Control Techniques

A numerical integrated model known as Cleveland Clinic Implantable Assist System was developed in [3]. The authors modelled the hemodynamic response of the assisted circulatory system as a function of various, but constant, pump speeds, using a nonlinear static model of the pump. They came to the conclusion that the integrated circulatory system VAD model could be used to create various VAD control algorithms. [4] developed a PI controller for a simple computer model of the circulatory system. However, the assumptions made in this work were unrealistic, including continuous flow throughout the circulatory system, no heart valves, and linear correlation between the pump generated pressure difference,and pump voltage, current, and rpm. A nonlinear, discontinuous model was developed in [5] to maintain the pressure difference between left heart and aorta using a PI controller. The model successfully maintained an adequate perfusion during changing cardiac demand with the help of two implantable pressure sensors. In another literature, a suction index strategy was proposed [6] for maintaining the cardiac demand. The technique successfully maintained perfusion demand during both rest and exercise state and also was capable in avoiding suction. However in both, the use of implantable sensors has proven unsatisfactory due to the risk of sensor failure, clot formation and initiation of sepsis. Estimation of CO(Cardiac Output) using two parallel Kalman Filters were used in [7], However the use of mounted sensors may lead to sensor related issues. [8] proposed a dead beat controller for controlling the pulsating flow in the rotary blood pump. The performance of the controller showed its efficiency in tracking error under model uncertainties.

A number of methods, such as autoregressive exogenous (ARX) modelling, and Extended Kalman Filters (EKF), have been suggested[9]–[10] for model-based parameter estimation utilising the pump speed data to avoid implanting sensors. Model-based approaches to pressure or flow estimation, however, are computationally expensive and subject to variations in blood viscosity, friction , and pump impeller inertia. Clinical studies have shown that flow estimators contain considerable inaccuracies [11] or may need the use of pressure sensors for greater accuracy [12]. Additionally, the model parameters must be rebuilt and validated for various pumps. However,minor modifications to the pump's geometry can greater shifts in the pressure-flow relationship [13]. The design of rotary pumps can be changed to increase their sensitivity to pressure, but this requires time-consuming and expensive FDA approval procedures, and it may not be feasible everywhere.

In order to forecast blood viscosity during CFLVAD support, a machine learning model was also used[14]. However, this method lacks in vivo validation, is susceptible to learning sample mistakes, and adds complexity and processing expense to the controller. [15] used a simple PI controller for maintaining perfusion demand and to prevent ventricular suction during rest, exercise, reduction in preload conditions.

2.3 Conclusion Remarks

Motivated by the above discussion and in order to overcome the issues, a simple control algorithm based on speed measurement was developed.

Chapter 3

THE CARDIOVASCULAR SYSTEM

3.1 Overview

The cardiovascular system, also called the circulatory system is an organ system that enables blood to circulate and carries oxygen, nutrients and hormones and takes them to where they are needed and removes the waste products like carbon dioxide. It helps in providing nourishment, preventing disease, regulating body temperature, and maintaining homeostasis. The heart, blood arteries, and blood make up the cardiovascular system's three parts.This section will provide a detailed understanding on the cardiovascular system.

3.2 The Heart

Heart is the central pumping station of the blood vascular system. It is a hollow muscular organ that beats spontaneously and rhythmically throughout the life. It consists of four chambers- two atria(auricles) and two ventricles. The right atrium receives deoxygenated blood from the body and is passed to the right ventricle through tricuspid valve while the left atrium receives the oxygenated blood from the lungs and passes it to the left ventricle through bicuspid/mitral valve.

The oriface of the pulmonary artery, where it commences from the right ventricle, is guarded by pulmonary valve whereas that between left ventricle and the beginning of aorta is guarded by an aortic valve. These pulmonary and aortic valves are known as semilunar valves.

The right atrium receives blood from superior and inferior venacava as illustrated in Figure 3.1. The blood from the right atrium then goes into the right ventricle. The right ventricle then

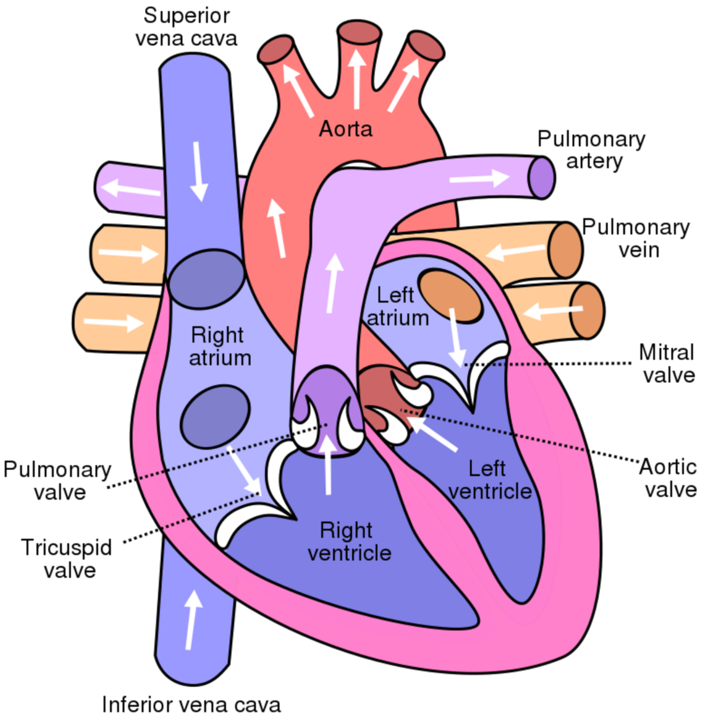

Figure 3.1: The Heart

pumps out its blood, into the lungs via pulmonary artery where it is purified. The oxygenated blood from the lungs is collected into the left atrium via pulmonary veins. Then the blood from the left atrium goes into the left ventricle from where it goes to various body parts via aorta.

3.3 Blood

The body receives nutrition, oxygen, and waste elimination through the blood, which is a fluid that circulates continuously. Blood is "thicker" than pure water because it is largely liquid and contains many suspended cells and proteins. Over a gallon(5 litres) of blood is typically present in an adult. The majority of blood is made up of a liquid called plasma. Plasma comprises proteins that carry substances through the blood, aid in blood clotting, and carry out other tasks. Additionally, glucose and other dissolved nutrients are present in blood plasma.
Blood cells make up around half of the volume of blood:
White blood cells combat pathogens, while red blood cells transport oxygen to the tissues. Platelets,are tiny cells that aid in blood clotting. Blood is transported by blood vessels (arteries and veins).

3.4 Vascular System

Vascular System is made up of arteries, arterioles, capillaries, venules, and veins as shown in Figure 3.2. The interior of all the vessels is the same, making them all function like conducting pipes. The coronary (heart), pulmonary (lung), cerebral (brain), skeletal (bone), renal (kidney), cutaneous (skin), splanchnic (stomach and intestines) and hepatic (liver) circulations are the many types of circulation in the body.

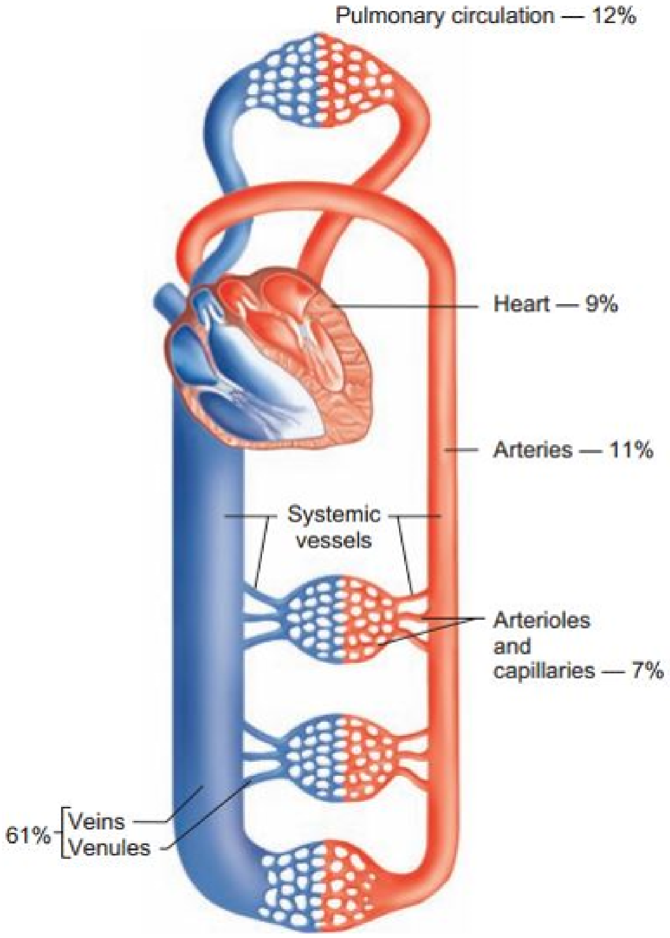

Figure 3.2: Vascular System

1) Artery: Blood is transported away from the heart by arteries. The biggest artery among all is the aorta, which carries blood away from the left ventricle. Depending on their size arteries are classified into three: peripheral(radial and femoral), central(aorta), intermediate(carotid and brachia). The nature of the larger arteries is more elastic, whereas that of the smaller arteries is less flexible.

2) Arteries: The smaller arteries divide into arteries that provide blood to organs, or arterioles.

Arterioles have a substantially higher resistance than arteries. Arterioles' walls are not stretchy. Arterioles' diameters are managed to regulate the amount of blood flow to the corresponding organs.

3) Capillaries: Capillaries carry out the circulatory system's primary function, which is the exchange of nutrients and gases between blood and cells. A capillary is around 0.1 mm away from a cell. Diffusion is the method by which the exchange takes place.

4) Veins: From various sections of the body, veins transport deoxygenated blood to the heart. They are more flexible and larger in diameter. The superior and inferior vena cava together comprise the vena cava, the biggest vein in the body.

3.5 Cardiac Cycle

The sequence of events as in Figure 3.3 takes place during the completion of one heartbeat is known as cardiac cycle. The beat rate is 72 per minute in normal condition. During each beat the Chambers of the heart contract and relax in a specific sequence. The contract and relaxation of cardiac chamber are respectively known as systole and diastole.

The events are divided into four:

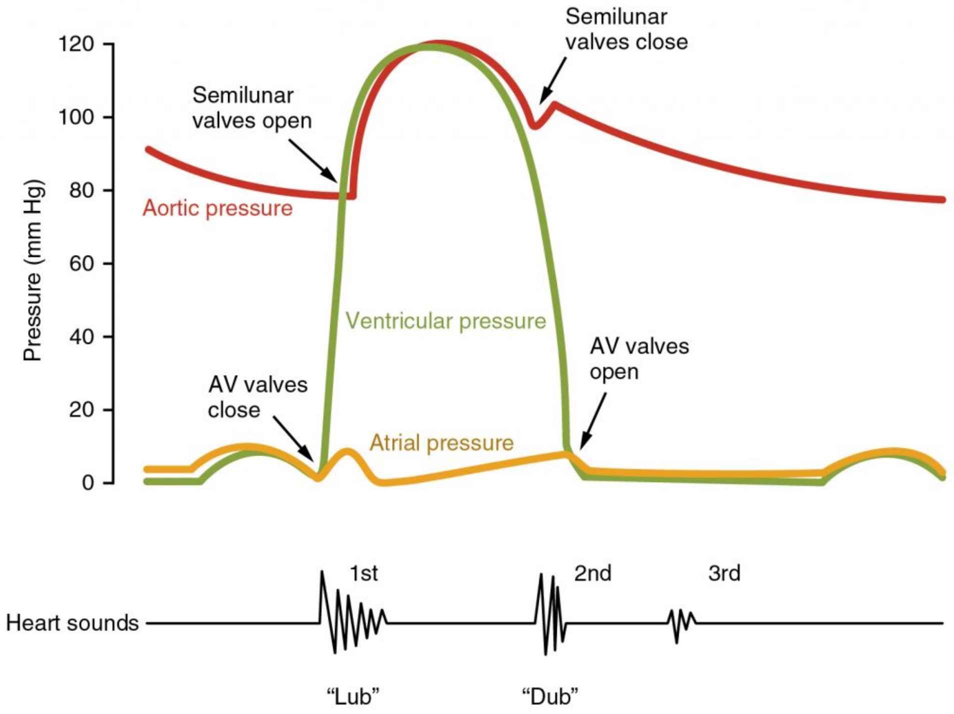

Figure 3.3: Cardiac cycle

1. **Isovolumetric Ventricular Contraction** - Ventricles start to contract when the electrical impulses get to the ventricular muscle cells keeping both semilunar and atrioventricular valves closed. It is called an isovolumetric contraction because it occurs at a constant volume. The pressure in the left ventricle increases significantly during this time.
2. **Ventricular Ejection** - Ventricular pressure increases causing the semilunar valves to open and ejection of blood into the corresponding arteries occurs.
3. **Isovolumetric Ventricular Relaxation** - The ventricles relax, lowering the pressure inside of them, and the semilunar and atrioventricular valves close. It is known as isovolumetric relaxation because the volume is constant.
4. **Ventricular Filling** - Atrio-ventricular valves open as a result of the atria contracting, which also allows blood to flow into the ventricles.

The first two events can be referred to as the systole, and later two can be referred to as the diastole.

3.6 Concluding Remarks

This chapter discussed about the Cardiovascular system. Next chapter presents the modelling of cardiovascular system.

Chapter 4

SYSTEM MODELLING

4.1 Overview

To evaluate the effectiveness and reliability of the suggested control method, a mathematical model of the Cardiovascular system based on [16] was created. The CVS that is fundamentally described in this model is made up of a pump, water, and pipes, which are respectively equivalent to the heart, blood, and blood arteries. The four chambers of the heart (Right Atrium (RA), Right Ventricle (RV), Left Atrium (LA), and Left Ventricle (LV)) and eight segments (Five of these are used for systemic circulation, and other three mimic the pulmonary circulations) of the vascular system are split into 12 blocks that make up the majority of the cardiovascular system. Three lumped parameters—flow resistance (R), vessel wall compliance (C), and interaction of fluid volume—are used to represent each section (L). The following suppositions were used to build the model:

- Blood is presumed to be incompressible and Newtonian.
- Linear lumped parameters are used to represent the vessels.
- Multiple tiers of homoeostatic systems, non-linear walls, and complicated morphology are disregarded.
- Blood vessel elasticity is thought to remain constant.

The following differential equation shows the blood volume in each block of Figure 4.1:

$$\frac{dV_n}{dt} = F_n^{\text{in}} - F_n^{\text{out}} \tag{4.1}$$

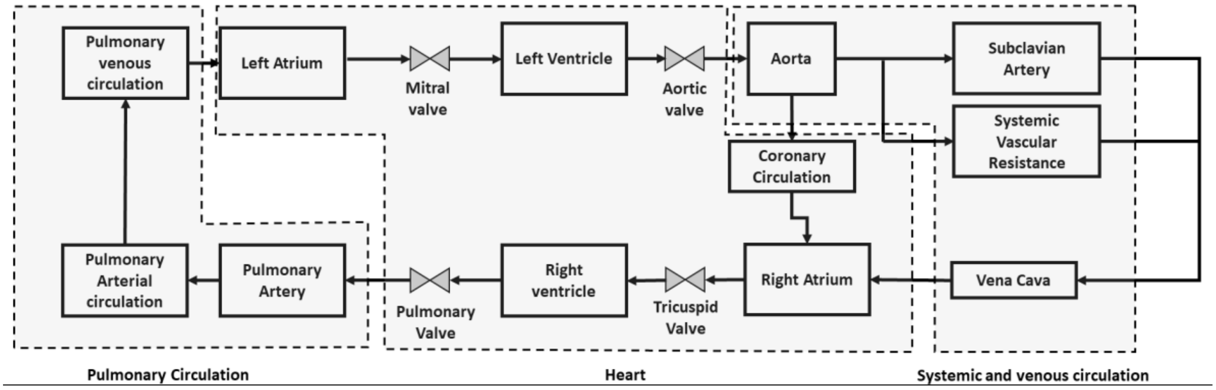

Figure 4.1: Proposed circulatory system

$$C_n = \frac{V_n}{P_n} \tag{4.2}$$

$$F_n{}^{in} = \frac{P_{n-1} - P_n}{R_{n-1}} \tag{4.3}$$

$$F_n{}^{\text{out}} = \frac{P_n - P_{n+1}}{R_n} \tag{4.4}$$

Substitute (4.2),(4.3) and (4.4) in (4.1), we get

$$\frac{dV_n}{dt} = \frac{V_{n-1}}{C_n R_{n-1}} - \frac{V_n}{C_n}\frac{1}{R_{n-1}} - \frac{V_n}{C_n}\frac{1}{R_n} + \frac{V_{n+1}}{C_{n+1} R_n} \tag{4.5}$$

where,

V_n - Volume of block n

$\frac{dV_n}{dt}$ - Rate of change of the volume in block n

F_n^{in} - Flow rate into the block n

F_n^{out} - Flow rate out of block n

P_n - Pressure of block n

C_n - Compliance of block n

4.2 Modelling of Heart

Consider left atrium of left heart:

$$V_{la} = V_{sla} - V_{ula} \tag{4.6}$$

where V_{la} is the volume of Left Atrium, V_{sla} is the stressed volume of Left Atrium and V_{ula} is the unstressed volume of Left Atrium

From (4.1) we get,

$$\frac{dV_{s_{la}}}{dt} = F_{l_a} - F_{Il} \tag{4.7}$$

$$V_{sla} = \int (F_{l_a} - F_{Il}) \tag{4.8}$$

$$V_{la} = [\int (F_{l_a} - F_{Il})] - V_{ula} \tag{4.9}$$

From (4.2) we get,

$$P_{la} = \frac{V_{la}}{C_{la}} \tag{4.10}$$

Valves can be modelled as switch and Flow through mitral valve is given by:

$$F_{Il} = \begin{cases} 0 & \text{if } P_{la} \leq P_{lv} \\ \frac{(P_{la} - P_{lv})}{R_{la}} & : P_{la} \geq P_{lv} \end{cases} \tag{4.11}$$

where,

P_{la} - Pressure of Left Atrium

P_{lv} - Left Ventricular Pressure

C_{la} - Compliance of Left Atrium

R_{la} - Resistance of Left Atrium

Consider left ventricle of left heart:

$$V_{lv} = V_{slv} - V_{ulv} \tag{4.12}$$

where, V_{la} is the volume of Left Ventricle, V_{sla} is the stressed volume of Left Ventricle and V_{ula} is the unstressed volume of Left Ventricle

From (4.1) we get,

$$\frac{dV_{s_{lv}}}{dt} = F_{Il} - F_{Ol} \tag{4.13}$$

$$V_{slv} = \int (F_{Il} - F_{Ol}) \tag{4.14}$$

$$V_{lv} = [\int (F_{Il} - F_{Ol})] - V_{ulv} \tag{4.15}$$

Flow through aortic valve is F_{Ol} , and this is modelled as a switch.

$$F_{Ol} = \begin{cases} 0 & : \text{ if } P_{\text{lvmax}} \leq P_{sa} \\ \frac{P_{\text{lvmax}} - P_{sa}}{Rlv} & : P_{\text{lvmax}} \geq P_{oa} \end{cases} \tag{4.16}$$

where, P_{lvmax} is the pressure in left ventricle(iso volumetric)

Resistance of valve varies with P_{lvmax}

$$R_{lv} = k_{rlv} \cdot P_{\text{lvmax}} \tag{4.17}$$

$$P_{\text{lvmax}} = \varphi(t) \cdot E_{\text{lvmax}} \cdot (V_{lv} - V_{ulv}) + [1 - \varphi(t)] \cdot P \cdot e^{((k \cdot V_{lv})-1)} \tag{4.18}$$

where

,

$$\varphi(t) = \begin{cases} \sin^2 \left[\frac{\pi \cdot T(t)}{T_{sys}(t)} \cdot U\right] & : 0 \leq U \leq \frac{T}{T_{sys}} \\ 0 & : \frac{T}{T_{sys}} \leq U \leq 1 \end{cases} \tag{4.19}$$

where,

$$U(t) = frac \left\lfloor \int_{t_0}^{t} \frac{1}{T(\tau)} d\tau + U\left(\tau_0\right) \right\rfloor \tag{4.20}$$

where,

E_{maxlv} - Elastance of left ventricle during maximum contraction

V_{ulv} - Left ventricular unstressed volume

P and K - constants

$\varphi(t)$ - the activation function

P_{sa} - Pressure of Systemic Artery

V_{lv} - Left Ventricular Volume

T - Time period

T_{sys} - period of systole

Right atrium and ventricle can be modelled using similar equations

4.3 Modelling of Vascular System

Consider Pulmonary artery,

$$V_{pa} = V_{spa} - V_{upa} \tag{4.21}$$

where V_{la} is the volume in Pulmonary artery, V_{spa} is the stressed volume of Pulmonary artery and V_{upa} is the unstressed volume of Pulmonary artery

From (4.1) we get,

$$\frac{dV_{s_{pa}}}{dt} = F_{o_r} - F_{p_a} \tag{4.22}$$

$$V_{spa} = \int (F_{o_r} - F_{p_a}) \tag{4.23}$$

$$V_{pa} = [\int (F_{o_r} - F_{p_a})] - V_{upa} \tag{4.24}$$

From (4.2) we get,

$$P_{pa} = \frac{V_{pa}}{C_{pa}} \tag{4.25}$$

The flow rate from pulmonary artery F_{pa} is given by,

$$\frac{dF_{pa}}{dt} = \frac{P_{pa} - P_{pp} - R_{pa}F_{pa}}{L_{pa}} \tag{4.26}$$

where,

P_{pa} - Pressure of Pulmonary artery

P_{pp} - Pulmonary arterial/ peripheral pressure

F_or - Flow from right ventricle to pulmonary artery

C_{pa} - Compliance of pulmonary artery

L_{pa} - Inertance of pulmonary artery

Using similar equations, the other blood vessels can be modelled.

4.4 Concluding Remarks

The detailed modelling of cardiovascular system is presented in this chapter. Next chapter discusses about Left Ventricular failure. Furthermore, it includes the modelling of LVAD that is used as treatment for Left Ventricular failure.

Chapter 5

HEART FAILURE AND THE LVAD

5.1 Overview

Heart failure (HF) is one of the major causes of death and is a chronic condition. Heart failure, is the condition in which the heart is unable to function effectively. Heart transplantation is the recommended course of treatment for end-stage heart failure patients, however only 5 to 10 percent of these patients actually obtain a transplant due to a severe lack of donor hearts. Due to a lack of available donor hearts, ventricular assist devices (VADs) are now widely used to help failing hearts. A detailed information about VADs, mainly LVAD is presented in this chapter. The modelling of LVAD is demonstrated further in this chapter for the integration of the same into the model of circulatory system.

5.2 Left Ventricular Assist Devices

Ventricular assist devices (VADs) are mechanical pumps that helps with circulation. VADs cannot replace heart but can assist the heart to pump blood efficiently. VADs are of different types like RVAD, LVAD and BiVAD, depending upon where failure has occurred. Since left ventricular failure characterises the majority of end-stage HF patients, they may decide to undergo the necessary left ventricular assist device (LVAD) therapy.

As the name implies, a LVAD is a mechanical pump that assist the heart to function properly. LVAD is inserted surgically right below the heart. The left ventricle, the chamber of the heart that pumps blood from the heart into the body, is where one end of the tube is joined, and the aorta, the body's main artery, is where the other end is linked. The pump receives blood from the heart, which is subsequently pushed into the aorta. A tube called as driveline connects the pump to the external controller and the power source through skin.

The blood flow to a person whose heart has been compromised by heart disease is restored with an LVAD. Some symptoms, such being always exhausted or having trouble breathing, are lessened by this. In some cases, it allows the heart to rest, allowing it to regain its normal ability. It helps with exercise, maintains or improves other organs, and enables cardiac rehabilitation for the individual. However, there are some risks involved after surgery which include infection etc.

5.3 Modelling of LVAD

The axial flow LVAD is modelled as an assistive device in the integrated circulatory model. A brushless DC motor powers the LVAD. The following equations can be used to describe a standard brushless [17] DC motor.

$$J\frac{d\omega}{dt} = T_e - B\omega - T_p \tag{5.1}$$

The phase current has a sinusoidal waveform if the motor's back electromotive force (EMF) is sinusoidal. In this instance, the motor torque is correlated with the phase current's (I) amplitude as,

$$Te = \frac{3}{2}K_B I \tag{5.2}$$

For the relationship between the load torque, generated flow rate, and pump rotational speed, we use the following functional form:

$$T_P = a_0\omega^3 + a_1 F_P \omega^2 \tag{5.3}$$

Combining above equations we get,

$$J\frac{d\omega}{dt} = \frac{3}{2}K_B I - B\omega - a_0\omega^3 - a_1 F_P \omega^2 \tag{5.4}$$

The model for the axial flow pump is given by:

$$\Delta P = b_1\frac{dF_P}{dt} + b_0 F_P + b_2\omega^2 \tag{5.5}$$

where,

J - Inertia of the rotor

ω - Rotor speed

K_B - Back electromotive force constant

I - Phase current

B - Damping coefficient

F_p - Flow rate

$J = 9.16 \times 10^{-7}\ \mathrm{kg} \cdot \mathrm{m}^2$,

$K_B = 0.003\ \mathrm{kg} \cdot \mathrm{m}^2 / \left(\mathrm{A} \cdot \mathrm{s}^2\right)$,

B=6.6 $\times 10^{-7}\ \mathrm{kg} \cdot \mathrm{m}^2/\mathrm{s}$,

$a_0 = 7.38 \times 10^{-13}, a_1 = 1.98 \times 10^{-11}, b_0 = -0.296, b_1 = -0.027, b_2 = 9.33 \times 10^-5$

5.4 Concluding Remarks

The chapter provides a brief idea on Left Ventricular Assist Devices and its modelling using the equations of BLDC motor and axial flow pump.

Chapter 6

METHODOLOGY

6.1 Overview

This chapter explains the method used in this project to maintain the perfusion demand of heart. The presented methodology uses a simple PID controller that utilises speed measurement to meet the objective. To obtain better results , the controller is tuned using particle swarm optimization algorithm.

6.2 Proposed System

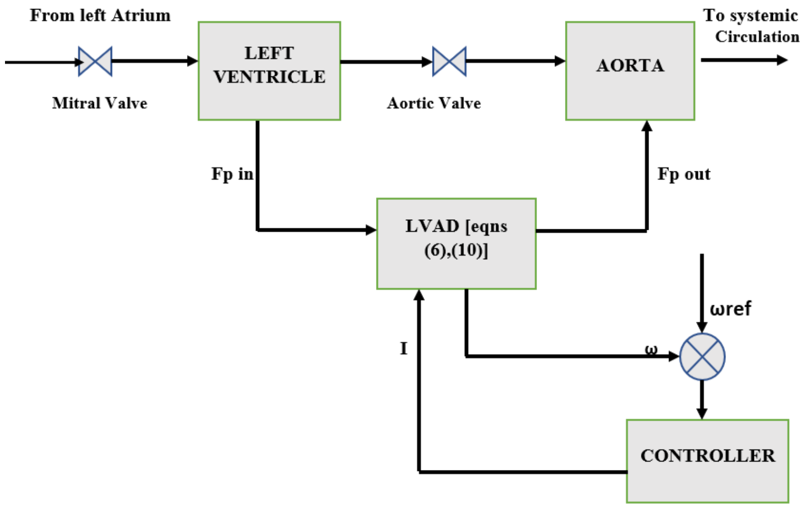

Figure 6.1: Block Diagram

Figure 6.1 depicts the working of the proposed LVAD control. The LVAD is integrated into the cardiovascular model as parallel path between LV and Aorta. The speed of the pump is compared with the reference speed, which is then controlled by a linear proportional integral derivative controller, whose gains are calculated by an intelligence swarm based method called particle swarm optimization(PSO).

6.3 Proportional Integral Derivative Controller

The first version of the PID controller was created in 1911. However, the first pneumatic controller with a fully tunable proportional controller wasn't released until 1933 by the Taylor Instrumental Company (TIC). A few years later, control experts discovered that the steady state error seen in proportional controllers could be eliminated by resetting the point to some hypothetical value, provided the error wasn't zero. The proportional-Integral controller was created as a result of this resetting, which "integrated" the error. The first PID pneumatic controller with a derivative action was then created by TIC in 1940, which minimised overshooting problems. Engineers were finally able to identify and set the proper PID controller parameters in 1942, due to the implementation of Ziegler and Nichols tuning criteria. The main objective of PID controller is meet the setpoint or to maintain a zero error.

The fundamental idea underlying a PID controller which is illustrated in Figure 6.2 is that each of the terms—proportional, integral, and derivative must be independently set. A correction factor is computed and applied to the input based on the variation between these values.

1) P element: Involves correcting the target proportional to the difference. As a result, the target value is never reached because both the applied correction and the difference decrease as they get closer to zero.

2) I element: Integral tuning makes an attempt to fix this by essentially accumulating the incorrect result from the "P" operation to increase the correction factor, but occasionally may result in an overshoot.

3) D element: Derivative tuning is done to minimize the above overshoot by slowing the correction factor that is applied as the goal gets closer.

In the proposed work, since the objective is to control the speed of the LVAD, the speed is taken as the feedback variable. The reference speed(set point) was taken as 860 rpm. The error e(t)

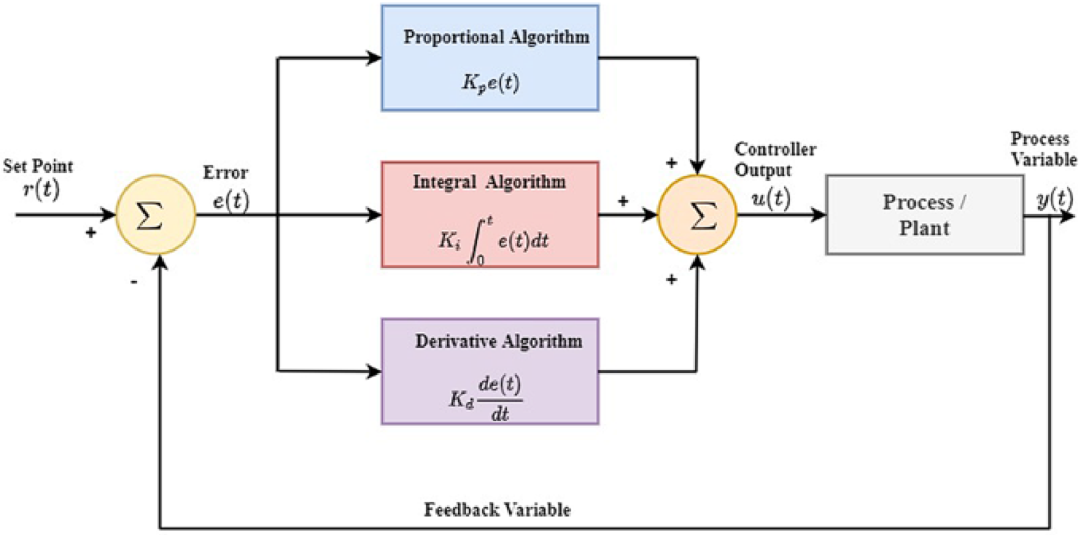

Figure 6.2: Proportional Integral Derivative Controller

which is :

$$e(t) = \omega_{ref} - \omega \tag{6.1}$$

is then given to the PID controller, which will minimize the error and produce a phase current (controller output) as in equation(6.2). This controller output is then fed to the LVAD equation(5.4) to generate the sufficient pump speed, thus maintaining the perfusion demand.

$$I = K_P(\omega_{ref} - \omega) + K_I \int_o^t (\omega_{ref} - \omega)dt + K_D \frac{d}{dt}(\omega_{ref} - \omega) \tag{6.2}$$

where,

I is the phase current,

ω_{ref} is the reference speed

ω is the differential pump speed at any instant

K_P is the proportional gain

K_I is the integral gain

K_D is the derivative gain.

6.4 Tuning of PID

The conventional PI and PID controllers are the most frequently used controllers. Nevertheless, this work focuses on PID . To meet design constraints and achieve desired performance, PID tuning must be performed in order to determine the values of proportional, integral, and derivative gains. Heuristic tuning, rule-based tuning, and model-based tuning are, in general, the three PID tuning techniques that can be used to find the best combination of these parameters. Each approach has benefits and drawbacks. Heuristic and rule-based PID tuning techniques may appear to be simple and inexpensive, but they frequently end up being quite time-consuming and expensive[18–19]. Artificial Bee Colony (ABC), Genetic Algorithm, and other optimization techniques have recently been used to tune PID controllers. In this paper, we try to build a PID controller based on an optimization technique called PSO (particle swarm optimization)[20],[21].

Optimization can be defined as a process through which the maximum or minimum value of a function can be found. The function that we want to minimize or maximize is called the objective function. For a PID- controlled system, there are often four indices to depict the system performance: ISE(Integral Square Error), IAE(Integral Absolute Error), ITAE(Integral Timed Absolute Error) and ITSE(Integral Timed Squared Error). Since we need to supress the error for longer period of time, the objective function used in this work is ITAE(Integral Time Absolute Error) which is given by:

$$\mathrm{ITAE} = \int_0^\infty \mathrm{t}|\mathrm{e}(\mathrm{t})|\mathrm{dt} \tag{6.3}$$

Now to seek a set of PID parameters such that the feedback control system has minimum ITAE, a PSO based optimization technique is used.

The PSO employs swarm intelligence. The method was inspired by observations of biological behaviour and social interaction in fish and birds. It uses a number of agents called particles which forms a swarm moving in a search space looking for the best solution. To move towards the best solution:

- Each particle will update its velocity based on its experience and its neighbours.
- Each particle will keep an account on its best position value that has been achieved so far.This position is called as pbest or personal best.

- Each particle will also keep an account on the best value obtained so far by any particle in its neighbourhood and this value is known as gbest or global best.
- Finally, each particle will update its position based on its current position, current velocity, distance between its current position and pbest, and distance between its current position and gbest.

6.4.1 PSO Algorithm

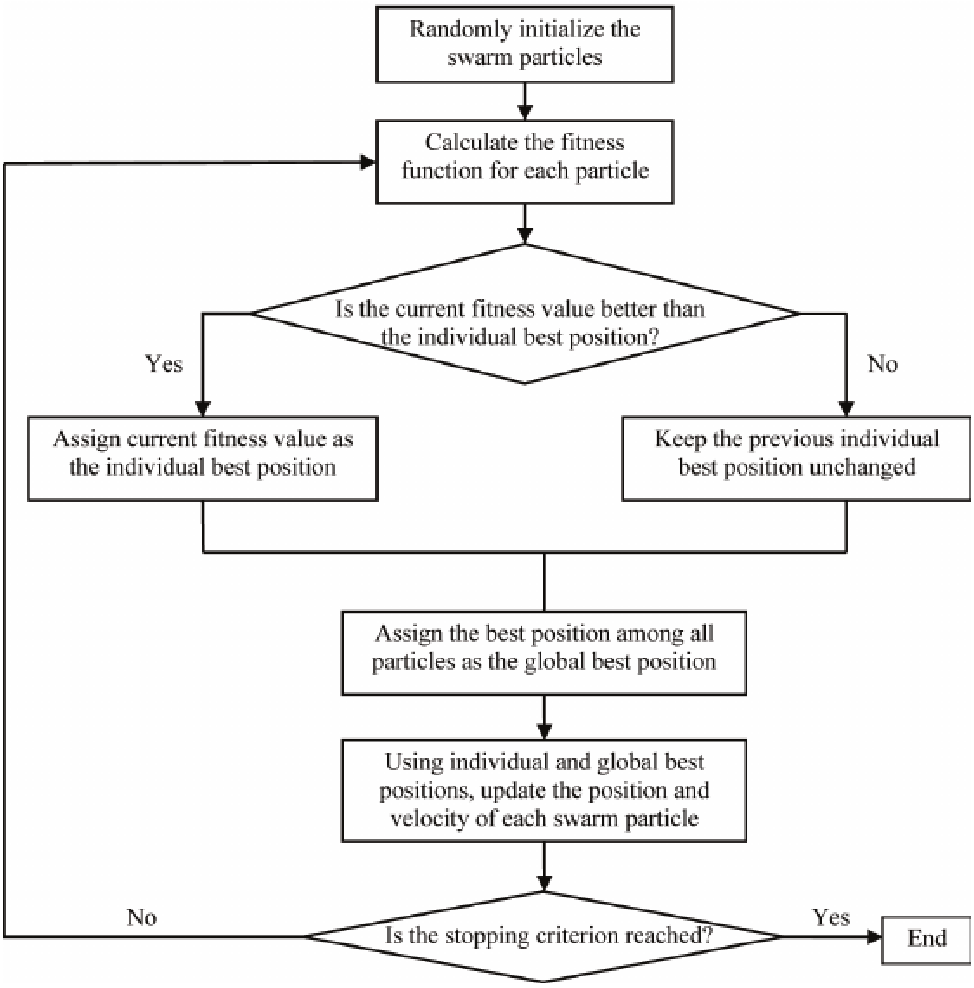

Figure 6.3: Flowchart of PSO

Figure 6.3 shows the working of PSO.

- Initialize the swarm parameters like no of particles, no of iterations.

Table 6.1: Control values in PSO

Sl. No	**Parameters**	**Values**
1	No of particles	50
2	No of iterations	100
3	c1	1.5
4	c2	2

- Evaluate the position of each particle based on objective function:

$$\mathrm{ITAE} = \int_0^\infty \mathrm{t}|\mathrm{e}(\mathrm{t})|\mathrm{dt} \tag{6.4}$$

- Obtain pbest(pb) of all particles. If the particles current position is better than its previous location, update it.
- Determine the gbest(gb).
- Using gbest and pbest update the particle velocity using:

$$v_i^{n+1} = w \cdot v_i^n + c_1 \cdot \mathrm{rand}() \cdot (pb_i^n - x_i^n) + c_2 \cdot \mathrm{rand}() \cdot (gb^n - x_i^n) \tag{6.5}$$

- Move particles to its new position by:

$$x_i^{n+1} = x_i^n + v_i^{n+1} \tag{6.6}$$

- Go to step 2 until the criterion is satisfied.

where,
w - inertia c1 , c2 - positive constant
rand() - random function between 0 and 1
n - iteration.
The control values used in the tuning process is given in Table 6.1. The gains obtained are Kp = 0.0008, Ki = 0.00063, Kd = 0.000119.

6.5 Concluding Remarks

This chapter discussed about the modelling of controller that has been used in the work and its tuning. Next chapter shows the simulation results.

Chapter 7

SIMULATION RESULTS

7.1 Overview

MATLAB/Simulink is used to validate the performance of the controller. The simulations of a healthy heart, failed heart, performance of heart using PID controller are all described in this chapter.

7.2 Graphical results

Simulations on the modelled heart, weakened heart, weakened heart with LVAD using PID controller is studied. Table 7.1 provides various values obtained during simulation. The characteristic plotted are:

1) Aortic Pressure(mmHg) vs time(s)
2) Ventricular Volume(mL) vs time(s)
3) Flow rate(ml/sec) vs time(s)

7.2.1 Cardiac Cycle

Figure 7.1 shows the cardiac cycle obtained from the modelled heart. It gave results similar to a normal heart. The Aortic pressure obtained from the modelling of heart is 117.03/83 which is in normal range, and the Stroke volume(SV)= End Diastolic Volume(EDV) - End systolic volume(ESV) = 53.71, which is also under the normal range(50-100).

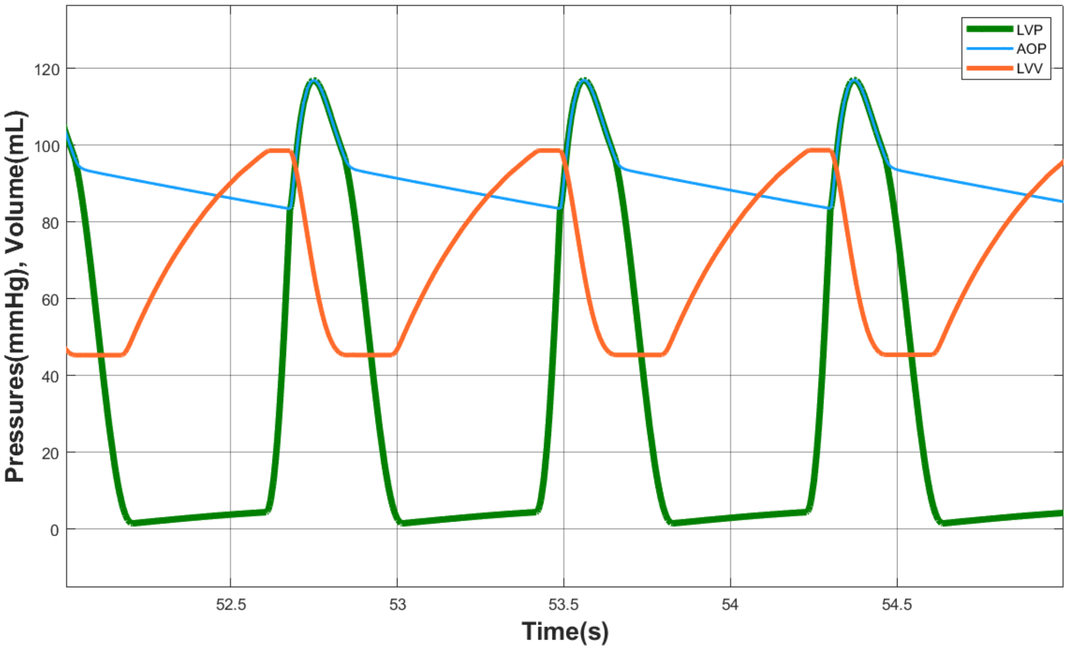

Figure 7.1: Cardiac cycle

7.2.2 Aortic Pressure

Figure 7.2, shows the aortic pressure of healthy heart which is 117.03/83. However due to failure occurred in left ventricle, the aortic pressure falls from the normal value as in Figure 7.3. It shows that aortic pressure is in 136.5/61. Figure 7.4 gives pressure when a LVAD is used. It shows that LVAD with PID controller successfully maintained the aortic pressure in the normal range i.e.,124.6/85.7.

7.2.3 Left Ventricular Volume

Figure 7.5, shows the ventricular volume of healthy heart. It shows a Stroke volume(SV)=53.71 which lies in the normal range(50-100). However due to failure occurred in left ventricle, the stroke volume is not in the normal value indicating heart failure as in Figure 7.6. Figure 7.7 shows the results in volume(LVV) when a LVAD is used. It shows that both LVAD with PID controller successfully maintained the stroke volume in the normal range i.e.,71.66.

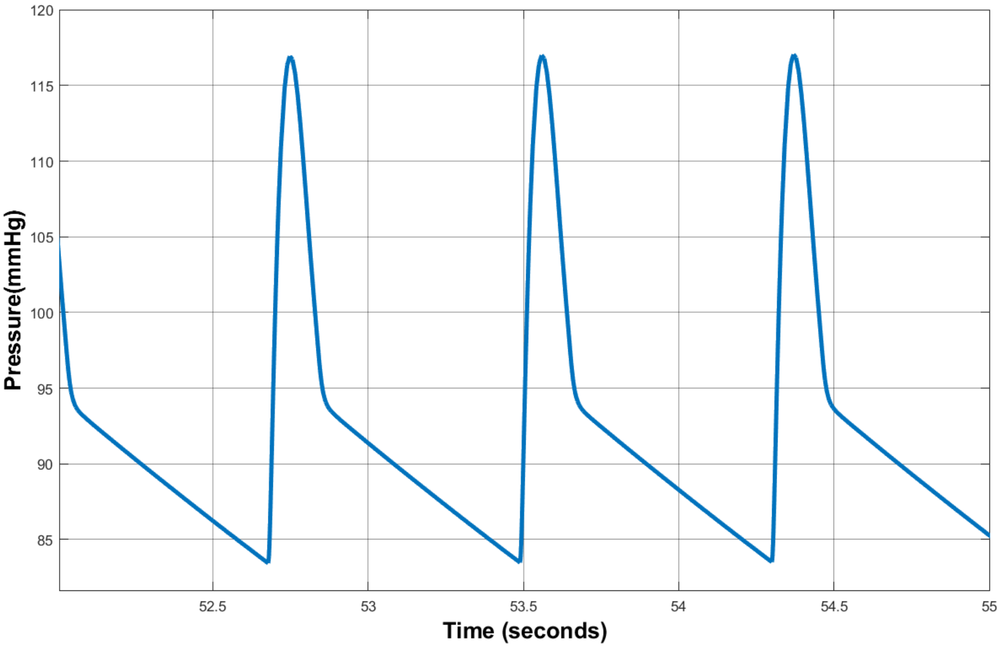

Figure 7.2: Aortic Pressure of Healthy Heart

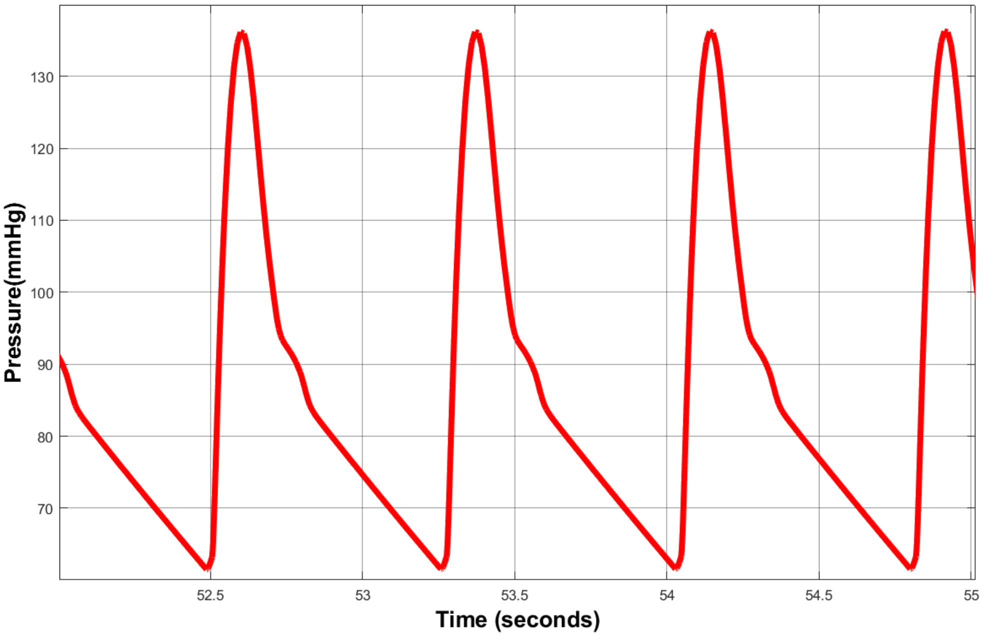

Figure 7.3: Aortic Pressure of Weakened Heart

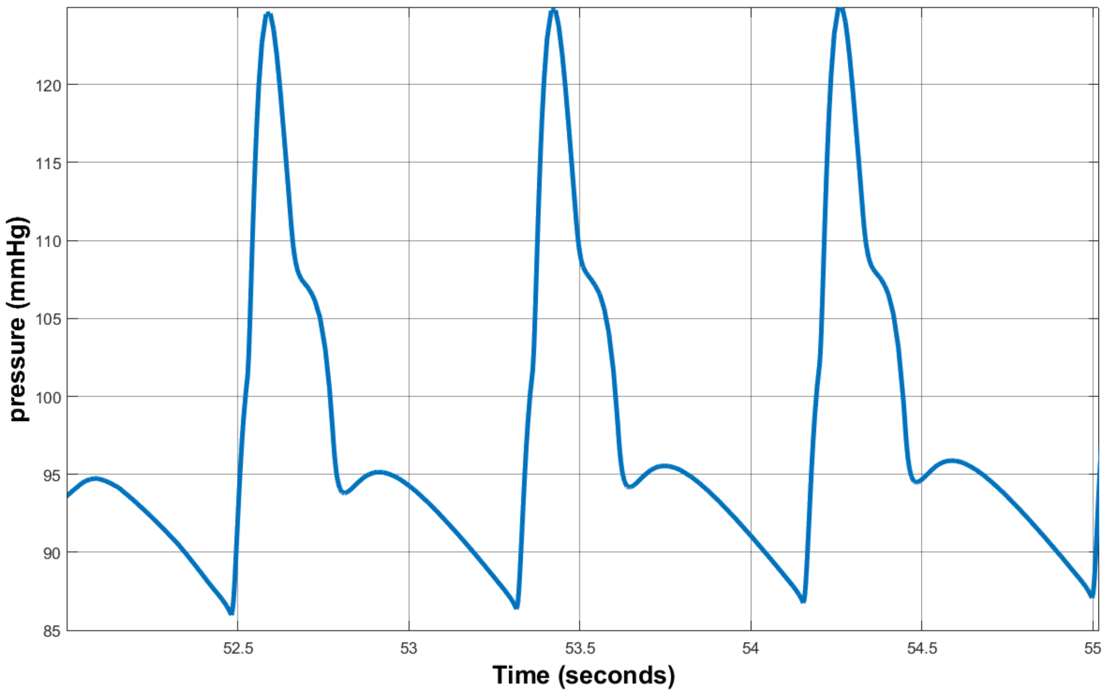

Figure 7.4: Aortic Pressure of Weakened Heart with LVAD using PID

Table 7.1: AOP, SV, and EJF of healthy, weakened and weakened with lvad support

	Aortic Pressure	Stroke Volume	Ejection Fraction
Healthy Heart	117.03/83	53	53.56%
Failed Heart	136.5/61	121.31	72.48%
LVAD support(PID)	124.6/85.7	71.66	62%

7.2.4 Flowrate

Figure 7.8 shows the flow rate when heart failure occurred. Figure 7.9 gives the results in Flow rate when a LVAD is used. It shows that both LVAD with PID controller successfully maintained the flow rate.

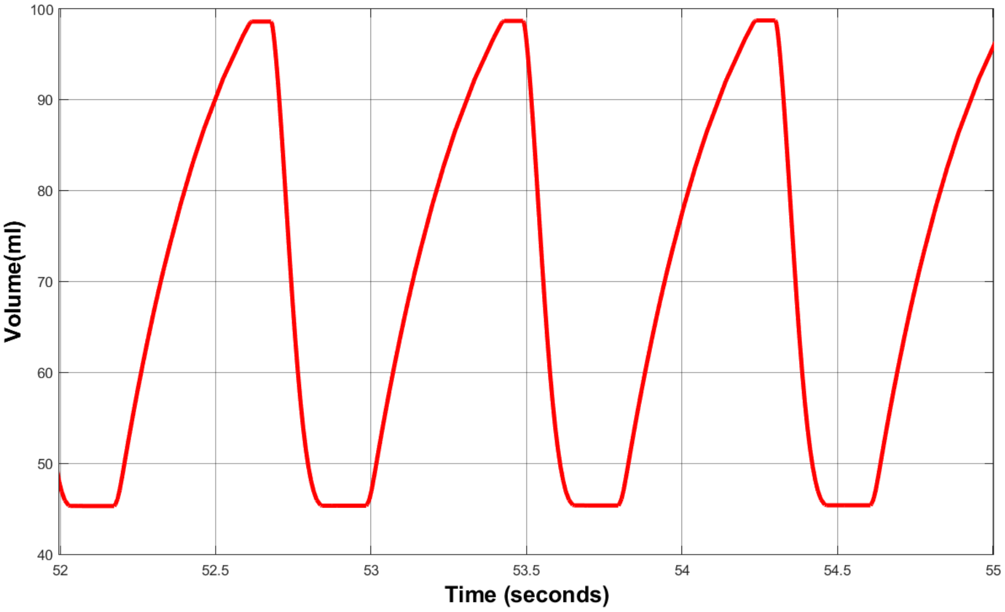

Figure 7.5: Left Ventricular Volume of Healthy Heart

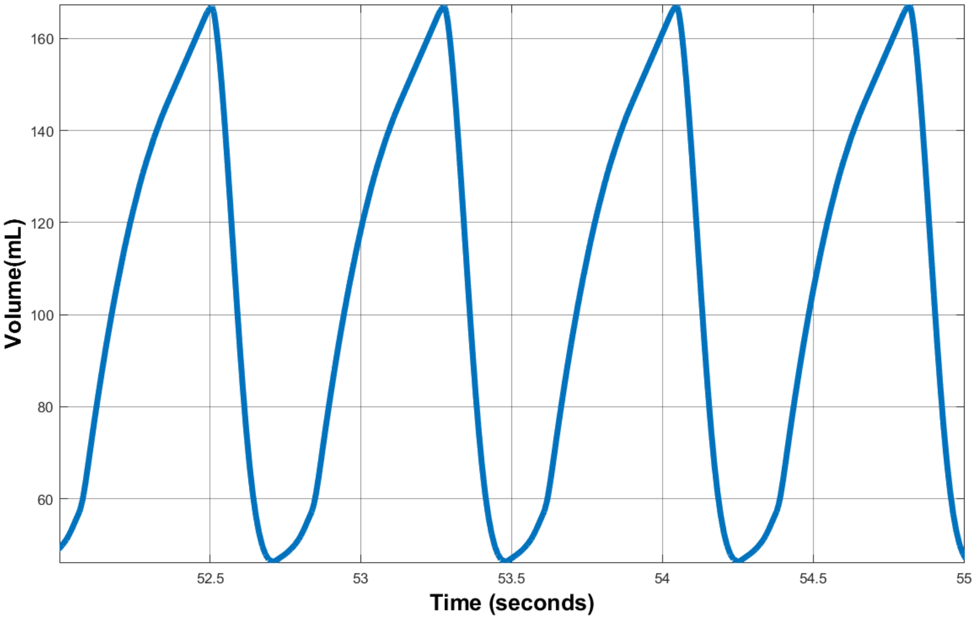

Figure 7.6: Left Ventricular Volume of Weakened Heart

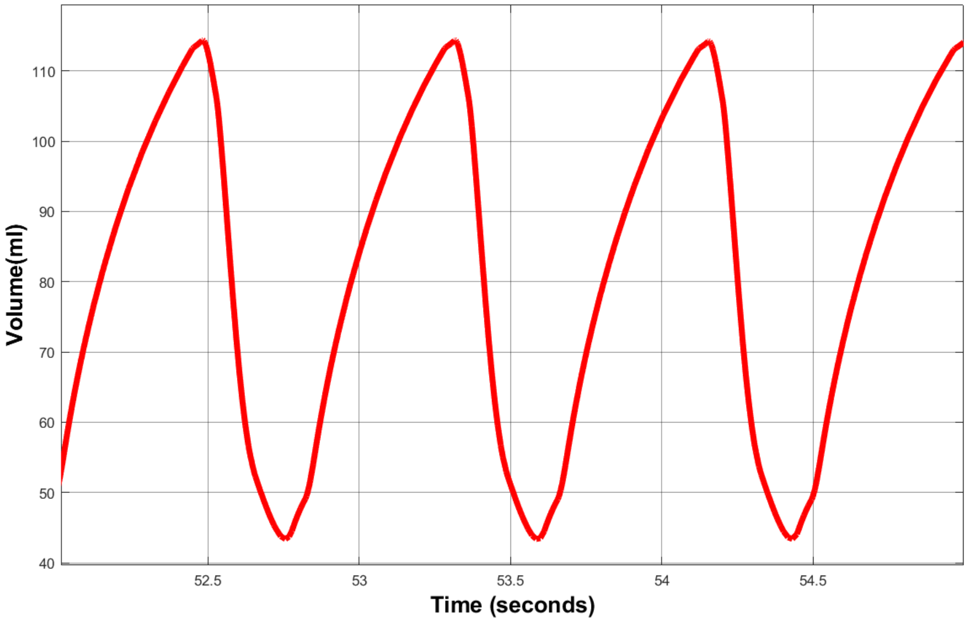

Figure 7.7: Left Ventricular Volume of Weakened Heart with LVAD using PID

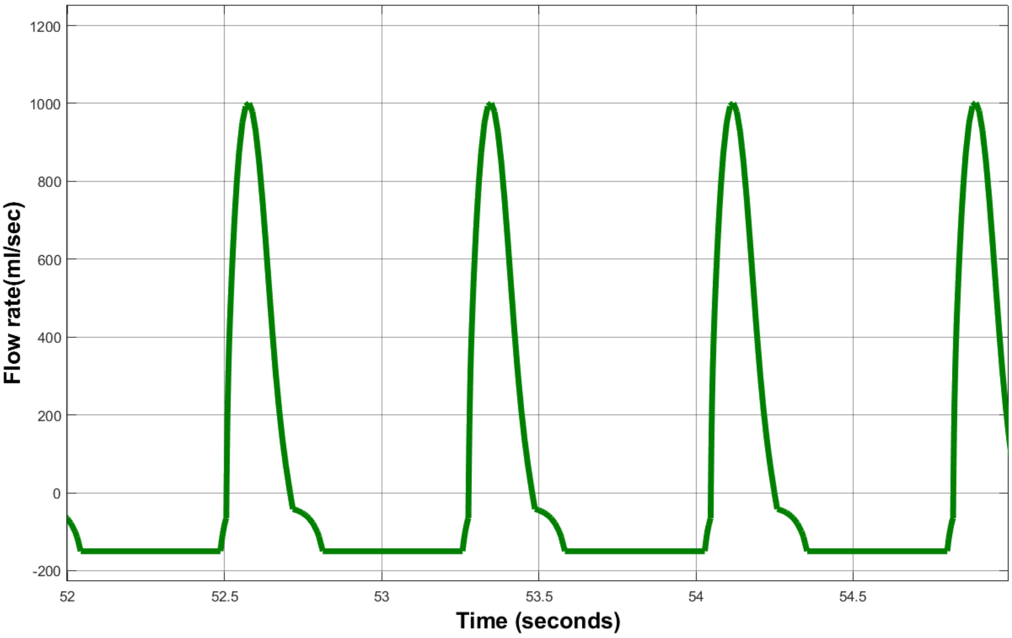

Figure 7.8: Flow rate of Weakened Heart

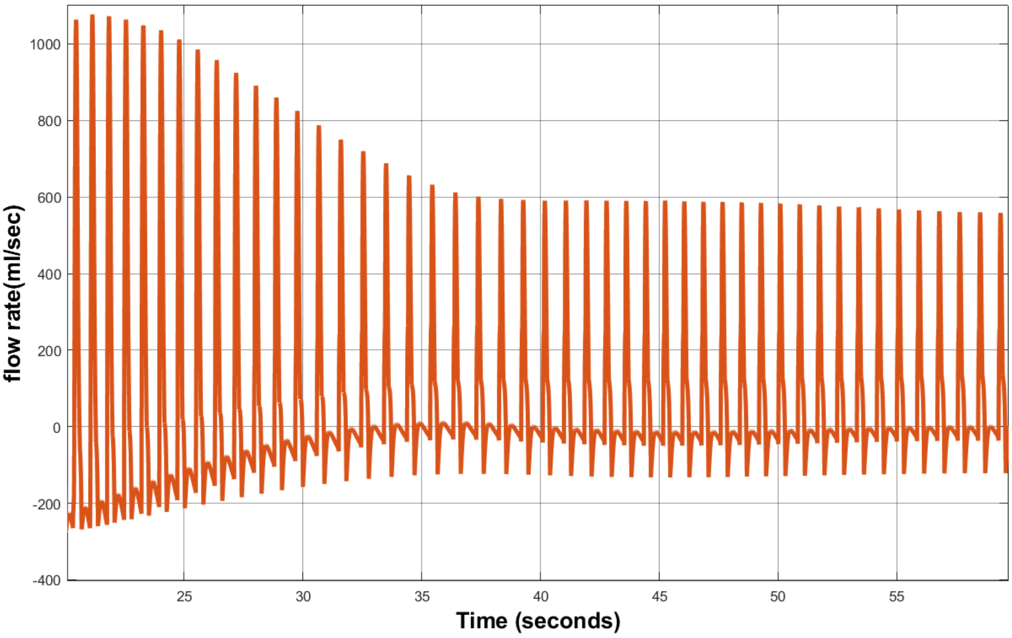

Figure 7.9: Flow rate of Weakened Heart with LVAD using PID

7.3 Concluding Remarks

This chapter verified the effectiveness of proposed control approach in maintaining the perfusion demand. Moreover, the comparison results with PI controller proved the proposed controller better with respect to settling time.

Chapter 8

CONCLUSION AND FUTURE SCOPE

The aim of the work was to develop a simple PID controller using PSO based tuning for the speed control of LVAD so as to prevent the perfusion demand. A model of circulatory system was developed using MATLAB/ Simulink and the simulation results showed the aortic pressure, stroke volume are within the normal range of human heart. Left Ventricular Assist Device was then developed using the modelling equations of a BLDC motor and an axial flow pump. It was then integrated into the modelled heart. Inorder to regulate the speed of the pump a PID controller was utilised, whose parameters were found by a search based algorithm known as Particle Swarm Optimization.

Inorder to study the performance of the proposed PID based LVAD control, a heart failure condition was first initiated in the modelled heart and the performance of the weakened heart was observed. Then by integrating the LVAD into the weakened heart, the performance of LVAD assisted heart was investigated. The results after the integration of LVAD proved its capability in maintaining the perfusion demand within the normal range.

Future work considers the prevention of ventricular suction using PID controller and its effectiveness in maintaining perfusion demand and preventing suction under rest and exercise conditions.

References

[1] A. Cheng, C. A. Williamitis, and M. S. Slaughter, "Comparison of continuous-flow and pulsatile-flow left ventricular assist devices: is there an advantage to pulsatility?," *Annals of Cardiothoracic Surgery*, vol. 3, pp. 573-581, 2014.

[2] J. R. Boston, M. A. Simaan, J. F. Antaki and Yih-Choung Yu, "Control issues in rotary heart assist devices," *Proceedings of the 2000 American Control Conference*, ACC (IEEE Cat. No.00CH36334), 2000, pp. 3473-3477 vol.5, doi: 10.1109/ACC.2000.879214.

[3] Waters T, Allaire P, Adams M, et al., " Motor Feedback Physiological Control for a continuous flow VAD," *Artif Organs*, 23: 480–486, 1999.

[4] Zhou J, Armstrong GP, Medvedev AL, Smith WA, Golding LA, Thomas JD, "Numeric modeling of the cardiovascular system with the left ventricular assist device," *ASAIO Journal*, 45: 83–89, 1999.

[5] Guruprasad A. Giridharan, Mikhail Skliar, Donald B. Olsen, And George M. Pantalos, “Modeling and Control of a Brushless DC Axial Flow Ventricular Assist Device,” *ASAIO Journal* 2002.

[6] Liang, Lixue et al., " A Flow Sensor-Based Suction-Index Control Strategy for Rotary Left Ventricular Assist Devices," *Sensors (Basel, Switzerland)* , vol. 21,20 6890. 18 Oct. 2021, doi:10.3390/s21206890.

[7] Daniel R¨uschen, Miriam Rimke, Jonas Gesenhues, Steffen Leonhardt, Marian Walter, " Continuous Cardiac Output Estimation Under Left Ventricular Assistance," *International Federation of Automatic Control* , Volume 48, Issue 20, 2015, Pages 569-574, ISSN 2405-8963, https://doi.org/10.1016/j.ifacol.2015.10.202.

[8] E. Lim, A. H. Alomari, A. V. Savkin and N. H. Lovell, “Noninvasive deadbeat control of an implantable rotary blood pump: A simulation study,” *International Conference of the IEEE EMBS*, September 2-6, 2009.

[9] P. J. Ayre, N. H. Lovell, and J. C. Woodard, " Non-invasive flow estimation in an implantable rotary blood pump: a study considering non-pulsatile and pulsatile flows," in *Physiol Meas*, vol. 24, pp. 179-89, Feb 2003.

[10] A. H. AlOmari, A. V. Savkin, D. M. Karantonis, E. Lim, and N. H. Lovell, " Non-invasive estimation of pulsatile flow and differential pressure in an implantable rotary blood pump for heart failure patients," in *Physiol Meas*, vol. 30, pp. 371-86, Apr 2009..

[11] M. S. Slaughter, C. R. Bartoli, M. A. Sobieski, G. M. Pantalos, G. A. Giridharan, R. D. Dowling, et al., "Intraoperative evaluation of the HeartMate II flow estimator," in *J Heart Lung Transplant*, vol. 28, pp. 39-43, Jan 2009.

[12] K. A. Pennings, J. R. Martina, B. F. Rodermans, J. R. Lahpor, F. N. van de Vosse, B. A. de Mol, et al., "Pump flow estimation from pressure head and power uptake for the HeartAssist5, HeartMate II, and HeartWare VADs," in *Asaio j*, vol. 59, pp. 420- 6, Jul-Aug 2013.

[13] A. P.J., V. S.S., T. G.D., W. P.A., and L. N.H., "Sensorless Flow and Head Estimation in the VentrAssist Rotary BloodPump," in *Artificial Organs*, vol. 24, pp. 585-588, 2000.

[14] A. Petrou, M. Kanakis, S. Boes, P. Pergantis, M. Meboldt, and M. S. Daners, “Viscosity Prediction in a Physiologically Controlled Ventricular Assist Device,” in *IEEE Transactions on Biomedical Engineering* pp. 1-1, 2018.

[15] Moustafa Meki, Yu Wang , Palaniappan Sethu, Mohammed Ghazal, Ayman El-Baz, and Guruprasad Giridharan, “A Sensorless Rotational Speed-based Control System for Continuous Flow Left Ventricular Assist Devices,” in *IEEE Transactions on Biomedical Engineering* vol. 67, no. 4, pp. 1050-1060, April 2020, doi: 10.1109/TBME.2019.2928826.

[16] Mauro Ursino, “Interaction between Carotid Baroregulation and the Pulsating Heart: a Mathematical Model,” in *Am. J. Physiol.,*vol. 275, no. H1733- H1747,1998.

[17] Choi S, Boston JR, Thomas D, Antaki JF, “Modeling and identification of an axial flow blood pump,” *American Control Conference* Albuquerque, NM: Proceedings, 1997.

[18] Yadav, Nidhi and Chandra Shekhar, “Comparison of pid controller tuning methods for fopdt and sopdt of unstable system,” 2017.

[19] Rajeev Kumar, Sunil K. Singla, Vikram, “A Comparative Analysis of different Methods for the tuning of PID Controller,” in *International Journal of Electronics Communications and Electrical Engineering*, ISSN : 2277-7040 Volume 3 Issue 2 (February 2013).

[20] Mahmud Iwan Solihin, Lee Fook Tack and Moey Leap Kean, “Tuning of PID Controller Using Particle Swarm Optimization (PSO),” *Proceeding of the International Conference on Advanced Science, Engineering and Information Technology* Hotel Equatorial Bangi-Putrajaya, Malaysia, ISBN 978-983-42366-4-9, 14 - 15 January 2011.

[21] K.Lakshmi Sowjanya, L.Ravi Srinivas, “Tuning Of Pid Controllers Using Particle Swarm Optimization,” *International Journal of Industrial Electronics and Electrical Engineering* ISSN: 2347-6982 Volume-3, Issue-2, Feb.-2015.

List of Publications

[1] Benzy J R Hyde, Fathima M Kasim, "Speed Control of Left Ventricular Assist Device Using PID Controller with Parameter Optimization Based on PSO", *International Congress on Biotechnology & Bioinformatics*, 2022 - Accepted.