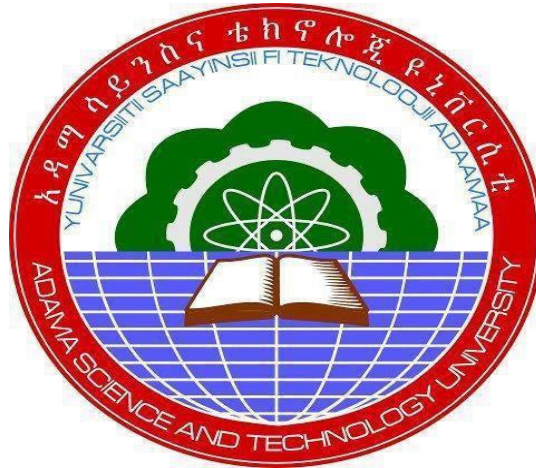


Heart Rate Regulation Using Cardiac Pacemaker System with Adaptive Neuro-Fuzzy Inference System Based Fractional-Order PID Controller



Nigatu Abeme Seyoum

A Thesis Submitted to

The Department of Electrical Power and Control Engineering

School of Electrical Engineering and Computing

Presented in Partial Fulfillment of the Requirement for the Degree of Master's in Electrical Power and Control Engineering (Control Engineering)

Office of Graduate Studies

Adama Science and Technology University

November 2022

Adama, Ethiopia

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Advisor:

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DECLARATION

I hereby declare that this Master Thesis entitled “**Heart Rate Regulation Using Cardiac Pacemaker System with Adaptive Neuro-Fuzzy Inference System Based Fractional-Order PID Controller**” is my original work. That is, it has not been submitted for the award of any academic degree, diploma, or certificate in any other university. All sources of materials that are used for this thesis have been duly acknowledged through citation

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I, the advisor of this thesis, hereby certify that I have read the revised version of the thesis entitled “**Heart Rate Regulation Using Cardiac Pacemaker System with Adaptive Neuro- Fuzzy Inference System Based Fractional-Order PID Controller**” prepared under my guidance by Nigatu Abeme submitted in partial fulfillment of the requirements for the degree of Masters of Science in Control Engineering. Therefore, I recommend the submission of a revised version of the thesis to the department following the applicable procedures.

Dr. Shubhashish Bhakta

Major Advisor

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Date

APPROVAL PAGE

I, the advisors of the thesis entitled “**Heart Rate Regulation Using Cardiac Pacemaker System with Adaptive Neuro- Fuzzy Inference System Based Fractional-Order PID Controller**” and developed by Nigatu Abeme hereby certify that the recommendation and suggestions made by the board of examiners are appropriately incorporated into the final version of the thesis.

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LIST OF ACRONYMS

ANFIS	Adaptive Neuro- Fuzzy Inference System
ANN	Artificial Neural Network
ANS	Automatic nervous system
AV	Atrioventricular
BPM	Beats Per Minutes
ECG	Electrocardiograph
FOFPID	Fuzzy Fractional Order Proportional Integral Derivative
FOMCON	Fractional–Order Modeling and Control
FOPID	Fractional Order Proportional Integral Derivative
FPID	Fuzzy Proportional Integral Derivative
GA	Genetic Algorithm
HR	Heart Rate
IAE	Integral Absolute Error
ISE	Integral Square Error
ITAE	Integral of Time Absolute Error
ITSE	Integral of Time Square Error
PID	Proportional Integral Derivative
SA	Sino Atrial
VFiB	Ventricular fibrillation
VDP	Van der pol
WHO	World Health Organization

ABSTRACT

Nowadays, heart disease is becoming the main cause of death worldwide. Therefore, the need arises to find advanced approaches to keep the patients safe. An implantable cardiac pacemaker is a standardized medical electronic device for the management and treatment of cardiac arrhythmias aimed at improving healthcare. This paper describes the nonlinear control of the heart rate model. ECG signals were generated using Zeeman's heart rate model. The existing heartbeat model was analyzed, and revised by consolidating the control inputs and adding a control mechanism as a pacemaker. In this study, the FOPID controller and ANFIS-based FOPID controller were applied to the heartbeat model to track and generate real ECG signals. The proposed method ANFIS-based FOPID controller has shown better performance in terms of tracking ECG signal of the actual data, obtained from the MIT- Boston's Beth Israel Hospital (BIH) and the physioNet database by eliminating chattering compared with the use of FOPID controller that increases the life span of a pacemaker. In addition, since the heartbeat pacemaker is disturbed by the brain signal, the robustness of the system to disturbance was analyzed. On the other hand, in this thesis, an ANFIS-based FOPID controller was developed to improve the performance of cardiac pacemakers that regulate heart rate. The design of the ANFIS-based FOPID controller ensures the stability and performance of the rate-adaptive pacing system, achieves better heart rate tracking and regulation at desired values. The developed control system was successfully validated using three cases of preset heart rates. The proposed model is designed, tested, and simulated along with tuning the controller gains using Matlab/Simulink software. The proposed controller performance for tracking and regulating different heart rates has been compared with FOPID and GA-tuned FOPID controllers. The comparison results show that the proposed system improved the parameters of cardiovascular system response in terms of stability, steady-state error and transients response.

Keywords: ANFIS, ECG signals, FOPID, GA, heartbeat model, heart rate, pacemaker.

CHAPTER ONE

1. INTRODUCTION

1.1. Background of Study

The human heart is a significant and intricate organ that circulates blood around the body through the capillaries of the circulatory system, delivering nutrients and oxygen to the tissues while expelling waste products like carbon dioxide. The main function of the human heart is to move blood throughout the body, control the rhythm and speed of our heart rate, and maintain blood pressure. The heart is made up of elements that are similar to those found in a house, including walls, chambers (rooms), valves (doors), blood vessels (plumbing), and an electrical conduction system (electricity). There are two states in a cycle of the heart of a human heartbeat namely diastole and systole (Farooq and Ahmad, 2017). The heart beats around 100,000 times a day. The heart has its electrical systems that adjust the rhythm of the heartbeat (Momani et al., 2019). The electrical signals cause the heart to contract and pump the blood to travel throughout the body with every heartbeat. Heartbeat is generated by firing the pulse of the SA node, which is the natural pacemaker of a healthy heart. Then, this electrical pulse propagates across the atria to ventricles via the atrioventricular node and the His-Purkinje system, resulting in the contraction and relaxation sequence of the cardiac muscle, i.e., normal heart rhythm (Karar, 2018). The average heartbeat for a normal adult is 75 beats/min, for a young active person is around 85 beats/min, and for old age, a person is around 65 beats/min (Momani et al., 2019).

The electrocardiogram (ECG) signal is among the most significant signals connected with how the human heart functions. The electrical potential created by the electrical activity of heart tissue is represented as a time-varying signal. The heart's contraction and relaxation during one ECG cycle are what cause the heart to beat. By observing the electrical potential between two electrodes placed at certain locations on the skin's surface, an ECG can be detected. The condition of cardiac health as well as a possible heart issue can be indicated by characteristic data collected from the ECG signal (Thanom et al., 2012).

Recent surveys have shown that cardiovascular diseases are one of the leading causes of death in the world, taking an estimated 17.9 million lives each year (WHO, 2021). The problem related to the heartbeat's rate or rhythm is considered one of such heart diseases. Bradycardia and Tachycardia usually happen when a person's heart rate is too low and too high respectively, which causes the heart to not pump or pump more than enough oxygen

to itself, causing, for example, a heart attack (Arunachalam et al., 2016). In addition to that the risk of sudden cardiac arrest, which is the cessation of typical circulation of the blood due to disruptions of the heart to contract successfully, which strikes an individual to death without any forebode, whether or not they have analyzed heart condition (W. V. Shi and Zhou, 2012). Thus, to avoid accidental patients death, early and exact determination and medical treatment of heart disease are very vital (W. V. Shi and Zhou, 2011a). To deal with this problem, the heart will be electrically stimulated by the so-called cardiac pacemaker, which can restore normal heart activity (Yadav and Rani, 2013).

A pacemaker is one of the implantable cardiac devices for the medical treatment of heart diseases and has been widely used nowadays. Cardiac pacemakers provide the electrical stimulus to restore or maintain the normal electromechanical activity of the heart. Early diagnosis and medical treatment of heart diseases can prevent the sudden death of a patient. The initial 10 seconds are crucial for comparing controller performance and demonstrating how quickly a response may be given (W. V. Shi, 2015). The accelerometer in a pacemaker detects patient movement and physical activity that increases heart rate and generates a signal electronically proportional to that activity (Johnson et al., 2003). Although it reacts slowly to the start of exercise, the QT interval rate response is physiologically realistic. Two sensors are utilized for compensation in order to decrease these factors, resulting in the ideal controller converging only when the system is stable (W. V. Shi, 2013) (W. V. Shi and Zhou, 2011b).

Cardiac pacemakers have become a therapeutic apparatus used around the world with more than 250,000 pacemaker inserts each year (W. Shi, 2012). A pacemaker is a device that produces electrical stimuli and conveys them to the muscles of the heart, in a way to causes the muscle to contract and the heart to beat (Haddad et al., 2006). One of the treatments to recover a victim's rhythm is to deliver the electrical pulse, which is well demonstrated as the most effective treatment for cardiac arrest (Gois et al., 2009). It comprises a pulse generator with a rechargeable battery and lead connectors. The pacemaker's battery is usually made of lithium iodine, which gives a life span of up to 10 to 12 years (Karar, 2018). The heart rate and/or beat are checked by the electrocardiogram detector within the integrated circuit of the pacemaker to apply electrical simulations of the pulse generator to the myocardium through pacing leads if irregular heart beating is identified.

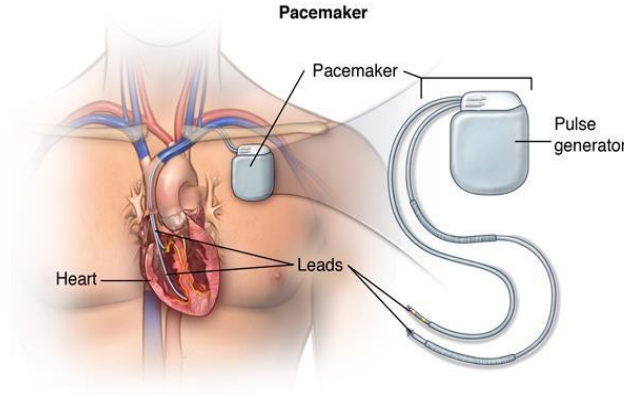


Figure 1.1: A typical demonstration of a cardiac pacemaker (Momani et al., 2019)

To improve the efficiency of such cardiac pacemakers, robust dynamic control system designs have been carried out for this device by many researchers. Implementing efficient pacing system control is required due to uncertainties, including imperfect information from the combination of physiological demands. This paper describes advanced cardiac pacemaker control using an adaptive neuro-fuzzy inference system based fractional-order PID controllers.

1.2. Statement of the Problem

The ability of the SA node to accurately pace a heart's rhythm can be disturbed by aging, medicines, changes in autonomic nerve activity (sympathetic and vagal), circulating hormones, serum ion concentrations, cellular hypoxia, or heart disorders, leading to slower, faster, or halted beats. Such types of heart diseases are now being treated with implantable artificial cardiac pacemakers that use electric simulation to synchronize the heart as closely as possible to the normal heart rhythm.

Diagnosis of cardiac abnormalities in pacemakers is serious, but detection and analysis of arrhythmias are challenging. It also exhibits nonlinear properties and the heart rate is also affected by the instructions fired from the brain signal, the control mechanism should be nonlinear and robust against such undesired physiological complications and disturbance of the heart. The implantable cardiac pacemaker also exhibits an inability to generate an impulse resulting in a heart rate lower than the programmed rate limit and the inability to pace an impulse to produce an evoked potential due to an inefficient control mechanism inside it. This results in an inappropriate inhibition of the pacing stimulus leading to potentially life-threatening consequences.

Due to magnetic interference, recent wireless pacemakers are severely attacked in their sensor outputs. Therefore, when using conventional linear controllers, such as PID control, the shocking signal supplied to the heart muscles may be delayed or missing to restore normal beating, resulting in a patient's sudden demise (Sargolzaei1 et al., 2014). For example, it often wrongly adds shocking voltages of 600 to 700 to the heart even if no blockage, which is too risky (Nazari et al., 2014).

Pacemaker performance depends not only on sensors and the pacemaker circuitry but also on the performance of the controller. Recently, there has been observed an increasing interest in combining artificial intelligent control tools with classical control techniques. The ANFIS based controller is designed to obtain improvements in system performance, cost-effectiveness, efficiency, dynamism, reliability and handling nonlinearity during different operating conditions. Therefore, this thesis presents an ANFIS-based controller for the cardiac pacemaker system.

1.3. Objective

1.3.1. General Objective

The main objective of this thesis is to design an ANFIS-based FOPID controller for heart rate regulation with a cardiac pacemaker system.

1.3.2. Specific Objective

- To study the heart's electric conductivity dynamics through mathematical modeling.
- To design a controller to track and generate synthetic ECG signals using reference information from the physioNet database.
- To design intelligent control and select suitable optimization algorithms to track and regulate heart rates.
- To simulate the entire system in the MATLAB Simulink ®.
- To compare and evaluate the performance of optimized FOPID with ANFIS-based controller trained by optimized FOPID controller data.

1.4. Scope

The scope of the work includes design and simulation of an ANFIS-based FOPID controller for a dual sensor cardiac pacemaker used to track ECG signals and control heart rate. In this study, it was assumed that the patient was at rest. Therefore, the dual sensor's active state is zero. The electrical activity of the heart is disturbed by brain signals. The parasympathetic

nervous system's impulses that lower heart rate under resting conditions were not taken into account in this thesis. However, the sympathetic signal simulation that causes the heart rate to rise as a result of anxiety or disease was regarded as a disturbance while tracking an ECG signal. Additionally, physiological issues that slow electrical propagation were not taken into account. Design and implementation of the system was done in MATLAB Simulink.

1.5. Limitations of the Study

The thesis is limited to the simulation of the system using MATLAB/Simulink by showing how the different components of the system interact with each other. This is because the implementation of a real system is difficult to procure the components. Along with it, it takes the additional time and cost incurred in making the actual prototype implementation. In addition, to the optimization of pulses, the pulse generator of a pacemaker may not always generate very accurate pulse amplitude. The sensitivity of the system performance needs to be investigated. The design of desired implantable medical devices requires a tradeoff between the quality of operation of the device and its power consumption, i.e., its longevity in the human body after implantation. An ANFIS-based controller is complex and may need significant power to be operated. Its energy consumption issues should be addressed.

1.6. Motivation of the Study

Last year I lost my younger brother suddenly by heart attack and I started exploring more about heart disease. Heart disease has inspired me to learn more about the circulatory system of the heart, the physiology of cardiac muscle, heart signals, the heart's excitation and conduction system, pacemaker history, applications, and sensing systems. These studies have helped me better understand how the heart works and how I might be able to illustrate a heart model and use different control mechanisms to enhance pacemaker functionality and benefit those who have heart issues. I have been inspired to make connections between my study and my field.

1.7. Significance of the Study

- The application of ANFIS based controller to Zeeman nonlinear heartbeat dynamic was introduced. The control algorithm used has better performance in the presence of disturbance of heartbeat system.
- In thesis an implantable pacemaker with an ANFIS based FOPID controller that eliminates the overshoot in the heart-rates. The ANFIS based controller is more

accurate compared to FOPID and GA tuned FOPID controller for pacing and correcting heart-rate of a cardiac patient.

- It provides better control schemes for generation of pacing pulse parameters.
- It offers good performance for tracking the heart rate and other ECG parameters to the physiological needs of patient.

1.6. The Organization of the Thesis

This thesis is organized in six chapters. The first chapter introduces about human heart and pacemaker, statement of problem, objectives, significance and scope of the thesis. In chapter two, literature review about human heart and sensing system and different related works are presented. Chapter three describes about methods used and overall system block diagram. Chapter four describes the design controller for cardiac pacemaker system. Chapter five describes about the results and discussions of overall system. Last chapter presents the conclusion and the recommendation for future work.

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CHAPTER TWO

2. LITERATURE REVIEW

2.1. The Human Heart

The human heart is the organ responsible for pumping blood through the circulatory system's blood vessels, feeding tissues with oxygen and nutrients, and removing waste products like carbon dioxide.

According to the National Institutes of Health, there are four major chambers in the human heart. The upper chambers are known as the atria, while the lower ones are known as the ventricles. The right atrium and the right ventricle are the "right heart," while the left atrium and the left ventricle are the "left heart." The heart's outer wall is made up of three layers. The inner wall of the heart is made up of the epicardium, while the middle layer is composed of the myocardium. According to the British Heart Foundation, the inner layer, also known as the endocardium, is responsible for keeping the heart's chambers aligned.

The heart is the primary organ of the human body, and it's responsible for delivering oxygenated blood to various body parts. It also carries out other tasks such as carrying out deoxygenation and carrying out blood pressure regulation. Besides regular exercise and physical activity, various factors can also affect the heart's contractility. Some of these include drugs, dehydration, and emotional disorders. For adults, the resting heart rate ranges from 60 to 100 beats per minute.

2.2. The Human Heart Circulatory System

The cardiovascular system is a type of system that connects the heart and the blood vessels. It can be described as a simple loop with the endpoints of the heart and the beginning points of the blood vessels. The blood moves toward the heart and is then returned to the body through the veins. Oxygen, nutrients, and hormones are delivered to cells by the circulatory system, which also eliminates waste materials like carbon dioxide. The heart, blood, and blood arteries make up the three components of the circulatory system. Through a system of arteries and veins, the heart pumps blood to the body (blood vessels) (Arthur C., 2006).

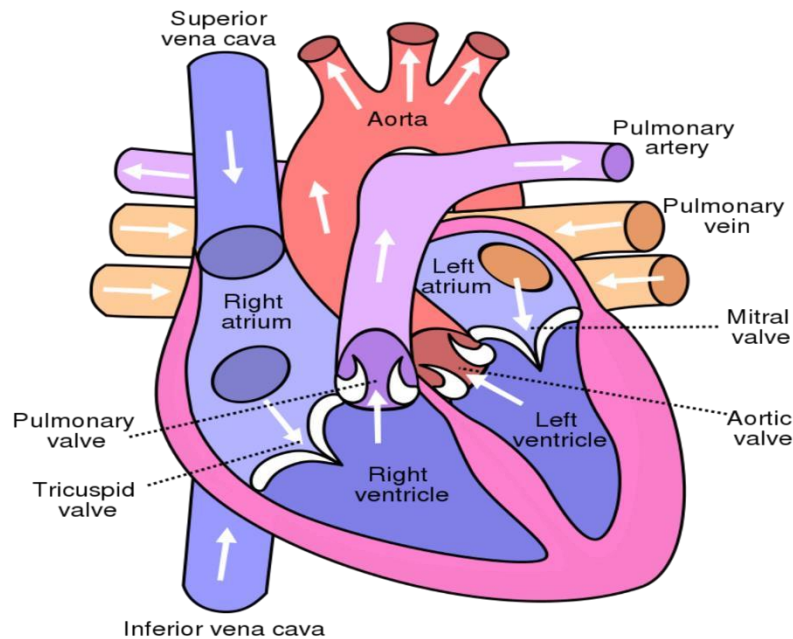


Figure 2.1: Circulatory system of the human heart (Arthur C., 2006)

Blood circulation throughout the body is the role of the circulatory system. In order to sustain life, blood circulation maintains our organs, muscles, and tissues healthy and functioning. The circulatory system also aids in the removal of waste from our bodies, including waste from food and drink as well as various chemical byproducts from our organs and carbon dioxide from breathing. Its arteries, veins, and capillaries all play a role in how it works. These veins let the heart and lungs pump blood constantly throughout our body.

2.3. The Cardiac Cycle and the Conduction System

2.3.1. The Cardiac Cycle

In the cardiac cycle, the relaxation phase of diastole, during which the heart fills with blood, is followed by the contraction phase of systole. The beginning of the cardiac cycle is signaled by SA node depolarization and atrial contraction. Each cycle takes a specific amount of time to complete. Throughout each cycle, changes in pressure, volume, electricity, and sound take place. Instead of contraction, the main cause of heart sounds is the turbulence in blood flow brought on by valves shutting.

Atrial and ventricular contraction is regulated by two different networks of cardiac fibers. Although the heart is innervated by the autonomic nervous system, cardiac contraction is neither caused nor influenced by it. Depolarization waves are produced and carried across the myocardium by the conduction system, which is made up of specialized muscle tissue.

When the ventricles contract, ventricular muscle fibers are arranged in whirls that squeeze blood out of the ventricles. Systole denotes the phase of contraction, while diastole denotes the moment of filling. The atria contract while the ventricles relax, and the ventricles relax while the atria contract.

The Figure 2.2 below shows the many cardiac cycle events for the left side of the heart. The top three curves represent the corresponding pressure variations of the left atrium, left ventricle, and aorta. The left ventricular volume changes are shown on the fourth curve, the ECG is shown on the fifth, and the phonocardiogram, which is a recording of the noises made by the heart, mostly by the heart valves as it pumps, is shown on the sixth curve (Arthur C., 2006).

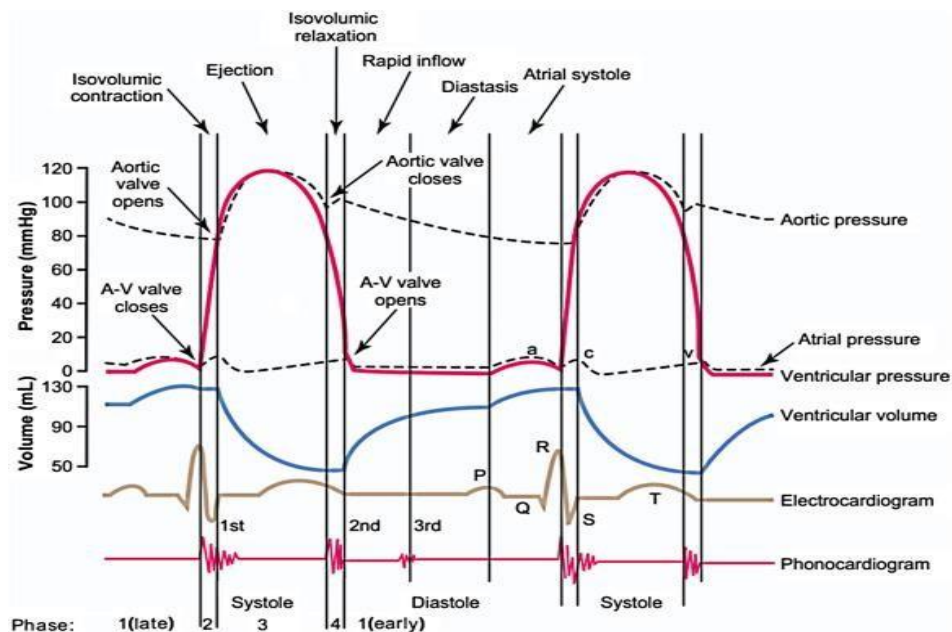


Figure 2.2: Cardiac cycle events for left ventricular function (Arthur C., 2006)

2.3.2. The Cardiac Conduction System

The heart's intrinsic conduction system consists of several specialized cells that either electrical activity spontaneously generates (pacemaker cells) or preferentially conduct this stimulus in a coordinated manner through the four chambers of the heart. In order for the heart to pump efficiently, the individual myocardial fibers must contract and relax in a coordinated, rhythmic fashion. This property of a healthy heartbeat is called synchrony. Synchronization is maintained by the heart's intrinsic electrical system, which generates and transmits electrical impulses through specialized pathways. Just as the pulse is evidence of the heart's mechanical activity, the electrocardiogram (ECG) is evidence of the heart's electrical

activity. When synchrony is lost, myocardial fibers contract in a random and uncontrolled manner called ventricular fibrillation, and the heart can no longer pump effectively. If the oxygenated blood does not reach distant tissues, the patient dies. The cardiac conduction system is composed of highly specialized cells that histologically resemble neural tissue. Five distinct anatomical structures that comprises the cardiac conduction system:

- The sinoatrial (SA) or sinus node is a small mass of tissue, about the size of a match-head, located high in the wall of the right atrium near the entrance of the superior vena cava. This is the heart's normal pacemaker, which automatically triggers impulses faster than other than any other part of the conduction system.
- The atrioventricular (AV) node is also located on the heart at the right side, just beneath the surface of the interventricular septum. The AV node and the conduction tissue surrounding it are known as the AV junctional tissue.
- The atrioventricular bundle, or bundle of His, is a band of nerve fibers that originates at an AV node, and then passes along the interventricular septum to the ventricles. Wilhelm His was the Swiss physician who first described this tissue.
- The right and left bundle branches are continuations of the bundle of His. They proceed along the right and left sides of the interventricular septum to the tips of the two ventricles.
- The terminal branches of the right and left bundles-of-thousands of fibrils that stretch between cardiac fibers for roughly one-third to one-half the thickness of the ventricular wall-are known as Purkinje fibers after their discoverer, Johannes von Purkinje.

2.4. The Heart as an Electrical Circuit

The SA node normally produces 60-100 electrical signals per minute; this is heart rate or pulse. With each pulse, signals from the SA node follows a natural electrical pathway through the heart walls. The movement of the electrical signals causes the heart's chambers to contract and relax. Natural pacemakers of the heart tell the heart when to beat. The master pacemaker is located in the atrium (upper chamber). It works like a spark plug, igniting in a regular rhythmic pattern to regulate the rhythm of the heart. This "spark plug" is called the sinusoidal node (SA) or sinus node. First, the atria contract as soon as the signal is sent. Electrical signals from the sinus node go through the atrium like a pebble in a puddle. The signal next makes its way to the area between the atria and ventricles. It's crucial to have this electrical connection. Without them, the signal never reaches the ventricles;

instead, it is sent to the rest of the heart by the big pumping chambers, which causes the muscles to contract.

The electrical signal reaches the atrioventricular node, another natural pacemaker. As the signal travels to the ventricle, it passes through a bundle of tissues called the AV bundle, also called the His bundle. The fascicles divide into thin wire-like structures called fascicles that extend into the right and left ventricles of the heart. The electrical signal travels down the bundle branch into a thin fiber. Ultimately, these fibers send signals to the ventricular muscles, causing them to contract and pump blood into the arteries. This coordinated sequence of electrical signals occurs in a typical heart once every second or so to keep the heart's regular rhythmic patterns.

The electrical system of the heart is in charge of producing the impulses that cause the heart to beat and the contraction of the heart muscle. All over the body, blood is pumped with each contraction. The atria, the upper chambers of the heart that pump blood to the lower chambers, are where this process starts (ventricles). The ventricle then pumps blood to the body and lungs. The heart is "wired" to send electrical signals that instruct the heart chambers when to contract, resulting in this coordinated activity. Thus, the electrical wiring in a home and the heart are analogous.

2.5. Electrocardiogram (ECG)

Heart electrical impulses may be recorded intracellularly using a microelectrode inserted into a cardiac muscle cell. Additionally, the surface of the thorax may be used to record the electrical potential caused by cardiac electrical activity. The extracellular electrical activity of the heart muscle is measured by the ECG. The transmission wavefront of the cardiac electrical signal across the body is an extremely complex operation. Leads are used to monitor potential fluctuations on the body surface between sites, as suggested by a number of cardiac signal measurement equipment. A dipole projection along the path indicated by the leads is the predicted signal. According to the position of the leads, different measurements are thus acquired, although normally the signal consists of the following waves (Gois et al., 2009).

The **P-wave**, which appears as the first wave on the ECG after the sinus simulation, shows the activation of the atrium. It typically lasts between 60 and 90 milliseconds in adults, has a rounded form, and has maximum amplitude between 0.25 and 0.3 mV.

The time between the beginning of the P wave and the beginning of the QRS complex is used to compute the **PR-Interval**, which lasts 90 ms.

Regardless of whether the first wave is a Q or R wave, the **QRS-Complex** refers to the ventricular contraction and is measured from the first wave to the last wave (R or S wave). Due to the high signal frequencies, the complex has a clear contour and lasts an average of 80 ms. The form of the lead device varies greatly depending on the application.

The **ST-Interval**, which corresponds to a portion of the ventricular repolarization, lasts from the end of the QRS-complex until the start of the T wave.

T-Wave: This has a circular configuration, amplitude of approximately 0.6 mV, and it represents ventricular stimulation. The three waves that make up the ECG signal shown schematically in figure 2.4 are the P-wave (depolarization of the atrium), R-waves (depolarization of the ventricles), and T-wave. Finding the proper R wave levels allows for accurate heart rate measurement and subsequent recording (W. V. Shi and Zhou, 2011a).

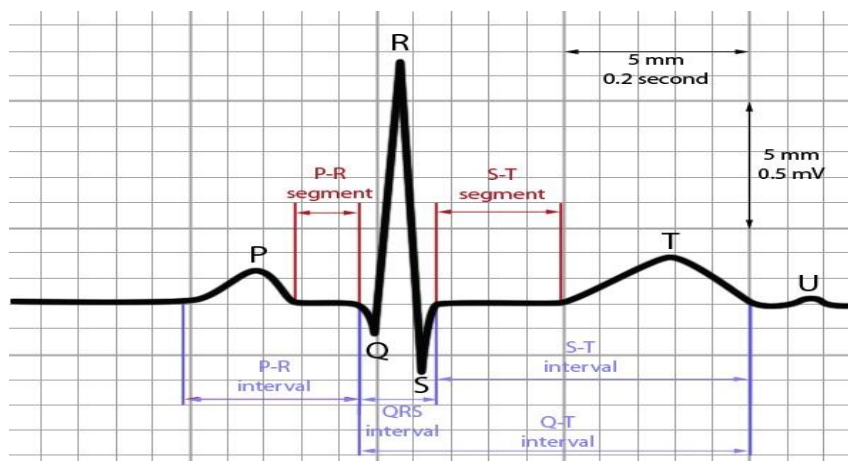


Figure 2.3: ECG waves and interval (Leti and Bricout, 2013)

2.6. Cardiac Diseases

The causes of cardiac diseases (arrhythmias) are usually one or a combination of the following abnormalities in the rhythmicity-conduction system of the heart (Arthur C., 2006).

1. An irregular rhythm from the pacemaker,
2. The pacemaker moving from the sinus node to a different location in the heart,
3. Blocks at various points throughout the impulse's path through the heart,
4. Abnormal pathways of impulse transmission through the heart,

5. Spontaneous generation of spurious impulses in almost any part of the heart.

Coordinating the electrical function of the heart can be affected by the conduction defects and refractory properties in the cardiac tissue. This may be classified into bradycardia, tachycardia, and cardiac arrest.

A. Bradycardia

It may be due to Impulse propagation dysfunction, where the conduction from the atria to the ventricle is disrupted or even blocked, or impulse generation failure with anomalies in the SA node. This indicates that a low heart rate, which is commonly defined as less than 60 bpm as indicated in Figure 2.4, might result in inadequate blood flow.

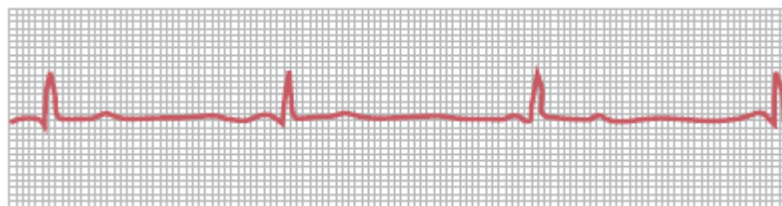


Figure 2.4: Sinus bradycardia (Arthur C., 2006)

B. Tachycardia

In an adult person, it is commonly defined as a heartbeat that is more than 100 beats per minute (bpm). The ECG of a patient with tachycardia is shown in the Figure 2.5. The ECG is normal, but the heart rate is higher than the typical 72 bpm, as indicated by the time interval between the QRS complexes. The most common signs of tachycardia are an increase in body temperature, sympathetic nerve activation of the heart, or cardiac toxicity (Arthur C., 2006).



Figure 2.5: Sinus tachycardia (Arthur C., 2006)

C. Atrial Fibrillation

The irregular heart rhythm shown in the next Figure often has a very high heart rate. Electrical activity becomes unpredictable during atrial fibrillation, causing an abnormal and rapid pulse that prevents blood from flowing through the body effectively (Dutta and Ghorbel, 2005).



Figure 2.6: Atrial Fibrillation (Arthur C., 2006)

D. Ventricular fibrillation (VFib)

It is a type of abnormal heart rhythm in which the lower cardiac ventricles repeatedly contract excessively, causing the heart rate to be irregular and frequently rapid. When the electrical signals within the ventricular muscle tissue become disordered, no reliable rhythm results, VFib occurs (Charles Patric and Jerry R. balentine). The heart trembles and barely pumps any blood to the body during VFib. It can cause unexpected heart arrest if not treated right away or properly controlled.

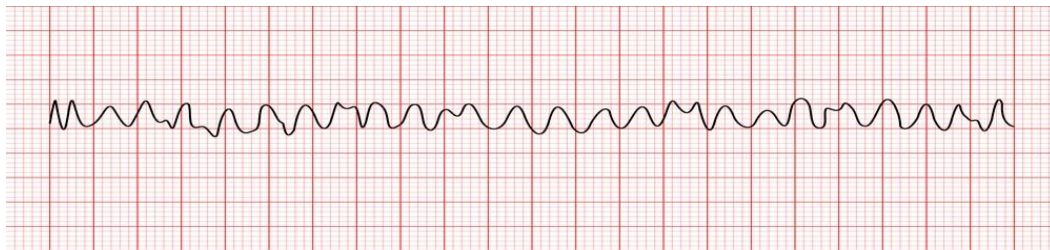


Figure 2.7: Ventricular fibrillation (Arthur C., 2006)

2.7. Heart Rate Variability

Heart rate is the RR interval's reciprocal, and heart rate has a significant impact on blood pressure. The ANS, which controls the autonomic nervous system, controls blood pressure. The medulla of the hypothalamus and the brain stem make the decision for automatic function. These centers send instructions to the heart via the nerves connecting the brain and heart. The parasympathetic and sympathetic kinds of ANS, both of which innervate the heart, are represented by two primary nerve sets. Parasympathetic (vagal) activity can lower heart rate whereas sympathetic activity can raise it (Jafarnia-Dabanloo et al., 2007).

The SA node has an intrinsic heart rate, but the autonomic nervous system, which is part of the central nervous system, typically modifies it (ANS). The sympathetic nervous system

is activated during stressful times. Additionally, it innervates the myocardium, speeding up depolarization wave propagation primarily through the AV node and boosting mechanical contraction force. This speeds up the pace at which SA nodes fire, which in turn speeds up heart rate (Reisner et al., 1949).

2.8. Sensing System and Heart Pacemaker

The heart's muscles (myocardium) receive electrical signals from a pacemaker that cause them to contract and relax. It is used to treat irregular heart rhythms, such as those that are too slow, too rapid, or both. The illustration below shows how a pacemaker is implanted in a human body. An abnormal heart beat can now lead a more active lifestyle thanks to a pacemaker. The pacemaker needs to be tuned to provide an obsessive, rapid pulse that transforms dangerous cardiac abnormalities back into normal rhythm (W. V. Shi and Zhou, 2011a).

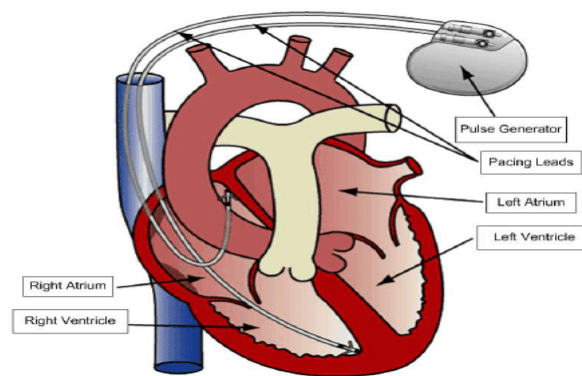


Figure 2.8: Schematic view of the heart with implantable pacemaker (Karar, 2018)

2.9. Cardiac Pacemaker System

2.9.1. Introduction

The heart receives electrical impulses from a pacemaker, a small battery-operated medical device, to ensure regular contractions. Worldwide, more than one million cardiac pacemakers are inserted each year. These numbers are anticipated to rise as a result of an aging population and an increase in pacing inductions. Pacemakers are used to treat heart rhythm irregularities, deliver an electric pulse that simulates the heart's natural pacing system, and maintain a healthy heart rate by delivering controlled, rhythmic electrical stimulation to the heart's chambers, and shield people from the dangers of low heart rate and sudden cardiac arrest. The pacemaker takes over rate control if the SA node is ill or the conduction system becomes clogged (Mulpuru et al., 2017).

2.9.2. Artificial Pacemaker

The first artificial pacemaker was created by Albert Hyman in 1932. He referred to it simply as the artificial pacemaker. An artificial pacemaker is a device that gives the heart muscle a controlled, rhythmic electrical simulation in order to maintain an effective heart rhythm for an extended period of time and maintain effective hemodynamic functionality (Haddad et al., 2006). A pacemaker is made up of three components, as shown in the Figure 2.8: an electrode (lead) system, a power source (battery), and an electrical pulse generator.

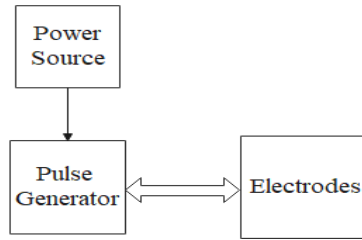


Figure 2.9: Basic pacemaker functional block diagram (Haddad et al., 2006)

2.9.3. Implantable Pacemaker

Wilson Greatbatch invented the implanted pacemaker in 1958. He placed a resistor with the wrong resistance while constructing an oscillator to record heartbeats. It started to emit a constant electrical pulse. In order to power the pacemaker, Greatbatch later created the corrosion-free lithium battery. The method was primarily utilized to treat Stokes Adams disease patients who had a complete block of the AV and required single chamber ventricular pacing. The pacemaker has a 6 cm diameter, a 1.5 cm thickness, and weighs about 180 g in total. The electrode received pulses from the pacemaker circuit that were 1 ms duration, 10 mA in amplitude, and repeated at a rate of 60bpm. When powered by ten mercury-zinc cells, the circuit's average current drain under these circumstances was about 12 A, giving it a roughly five-year continuous operational life (Haddad et al., 2006).

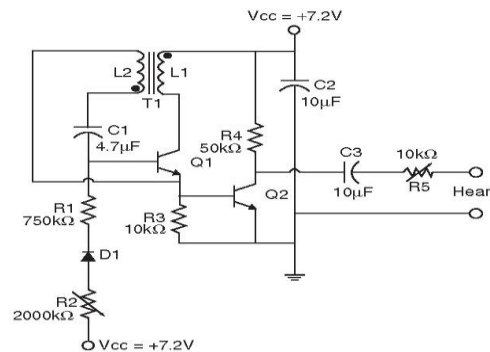


Figure 2.10: A schematic of the first implanted pacemaker (Haddad et al., 2006)

2.9.3. Demand Pacemaker

Demand pacemakers are pacemakers that only start up when they detect symptoms that the heart isn't beating naturally enough. A gadget of this type monitors the heart rhythm and only produces electrical pulses when the heart beats too slowly or skips a beat. A pacemaker on demand has two benefits over fixed-rate pacemakers: it prevents competing beats from happening and has a longer battery life. Regardless of any sporadic cardiac events, the early pacing system essentially communicated a constant pulse rate at a predefined rhythm to the ventricle. These pacemakers, also known as intermittent or fixed rate pacemakers, disrupt the heart's regular rhythm and frequently also cause arrhythmias or ventricular fibrillation (VF). A demand pacemaker provides electrical impulse for imitating the heart even in the absence of a regular beating by connecting a sensing amplifier to the asynchronous pacemaker to sense endogenous cardiac activity and so escape this interference. The battery life of the machine is increased since the demand pacemaker only activates when pacing impulses are needed, as opposed to the fixed rate device. The basis for all current pacemakers was introduced by Berkov in June 1964, and the schematic is given in Figure (Haddad et al., 2006).

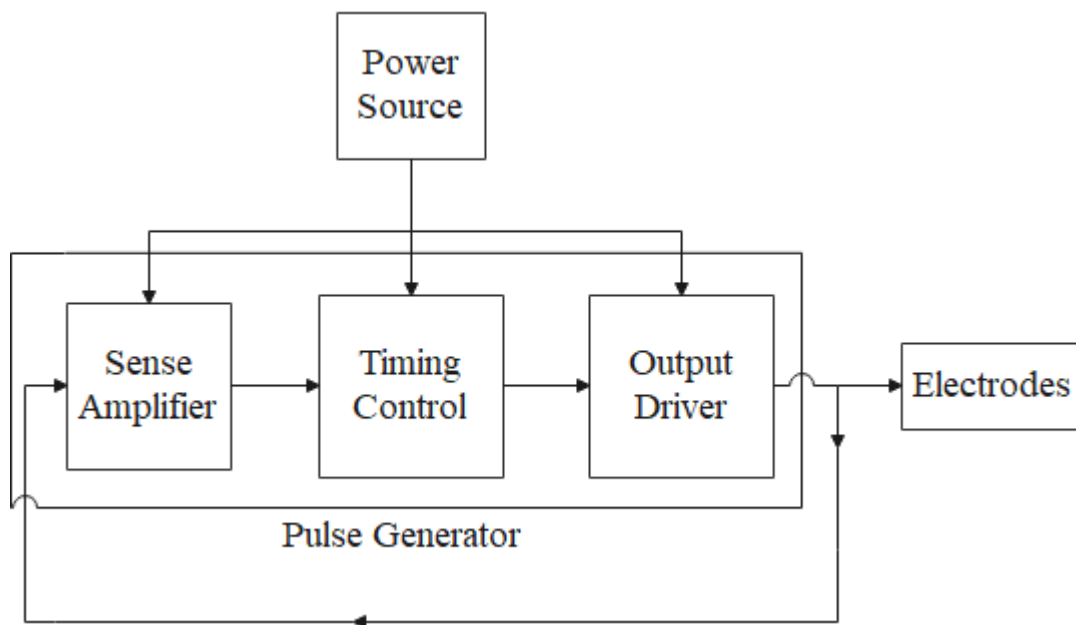


Figure 2.11: Basic demand pacemaker functional block diagram (Haddad et al., 2006)

2.9.4. Dual Chamber Pacemaker System

Both the upper and lower chambers of the heart contain electrodes while using this pacemaker. As the pacing moves sequentially from the atria to the ventricles, this type of pacing most nearly resembles the heart's natural conduction pattern and enhances the heart's

pumping capacity. The pulse generator can continuously modify the electrical activity of the heart in both ventricles through electrodes in both the atria and ventricles. These are the most commonly used pacemakers today, having been on the market since 1970. In 1971, Berkovits first reported a bifocal (AV) pacemaker that sensed only in the ventricles and controlled all chambers, offering one of his dual chambered his pacemakers. Bifocal pacemakers pace the atria and, given sufficient time, can pace the ventricles in atrial fibrillation or sinus node syndrome and atrioventricular block (Haddad et al., 2006). The schematic configuration of the system is shown in the Figure 2.12.

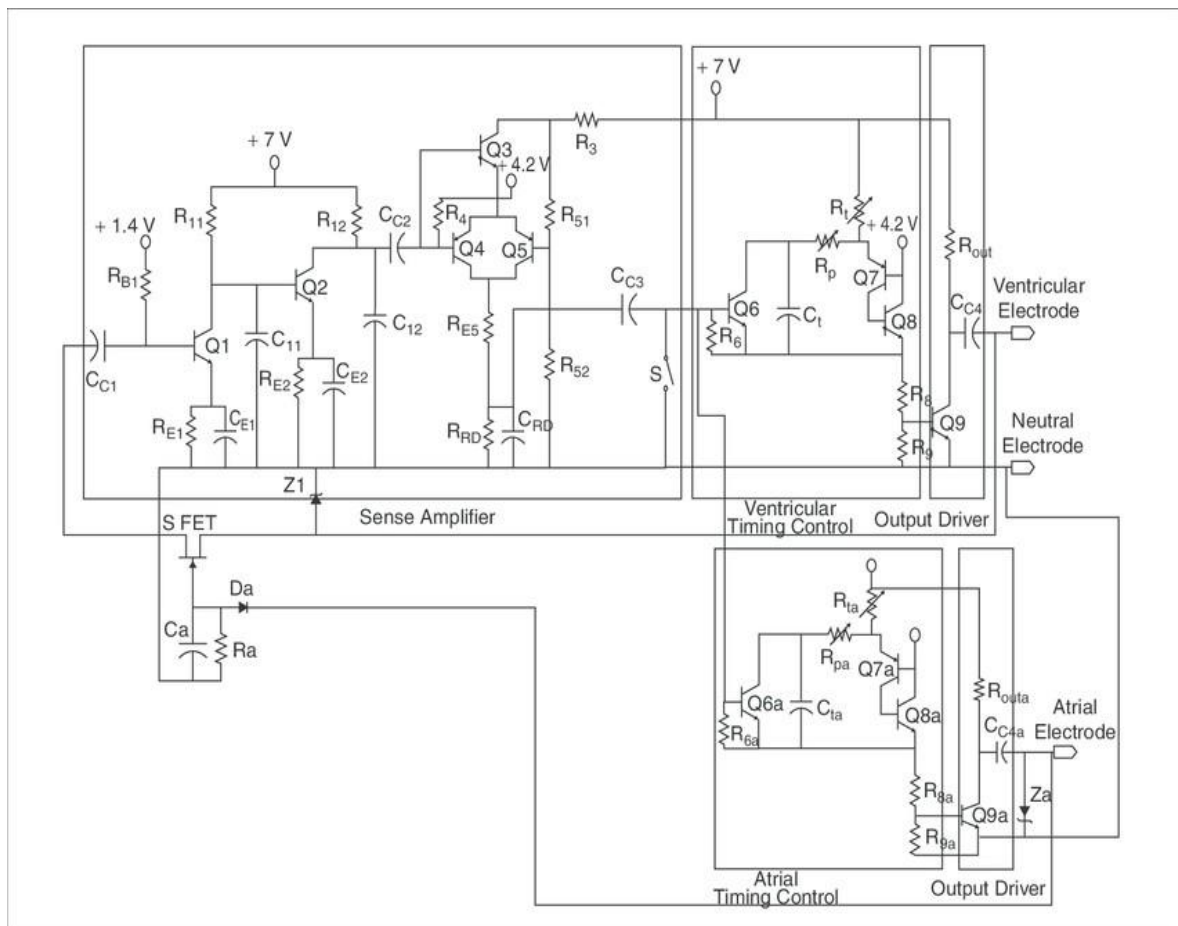


Figure 2.12: Schematic of the dual-chamber demand pacemaker (Haddad et al., 2006)

2.9.5. Rate Responsive Pacemakers

Recent developments in the early 1980s included the creation of rate-responsive pacemakers that could control the pacing rate based on the performance of the sensor device embedded in the pacemaker. The sensor system consists of a mechanism that calculates certain body parameters (body position, respiratory rate, PH, blood pressure, etc.) and a pacing algorithm that can change the pacemaker's output according to the calculated amount. Such

devices speed up or slow down the heart rate depending on the person's activity. Optimal heart rate is determined by an additional sensor for human activity based on body movement or a respiration sensor that detects breathing rate.

Current rate-responsive pacemakers can respond to various sensor details about the patient's health and physical activity. The machine relies on the pacemaker to provide the demand pulse generator and adapts to calculated parameters (Haddad et al., 2006). Figure 2.13 shows the block diagram of a pacemaker that responds to heart rate.

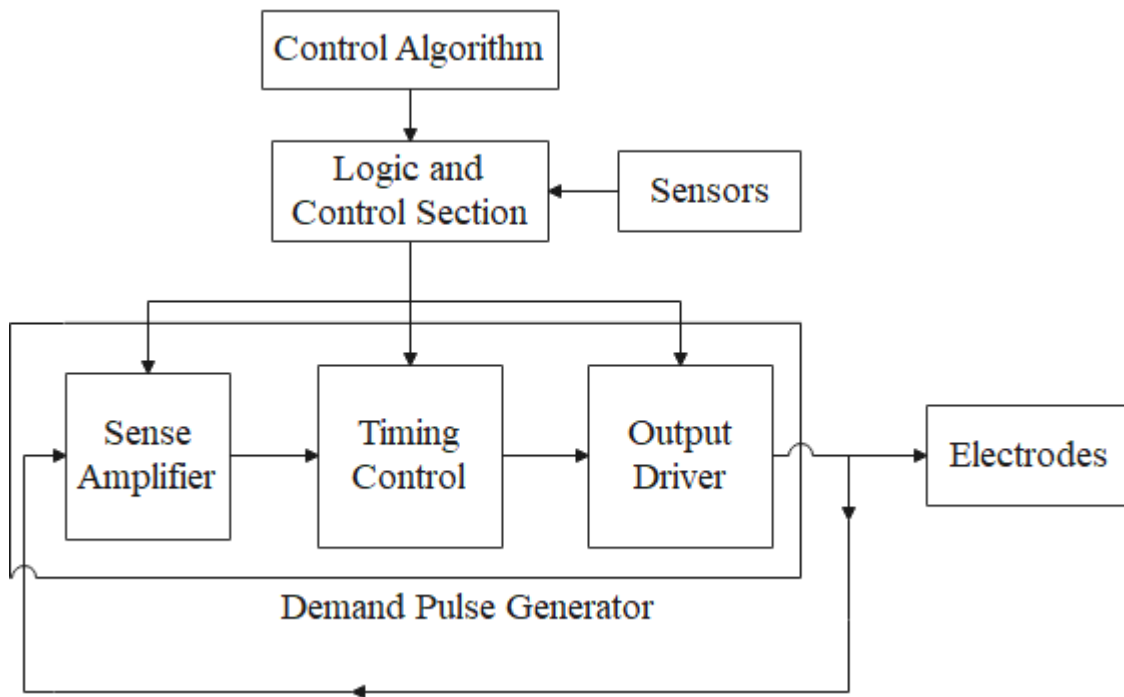


Figure 2.13: A block diagram of the rate-responsive pacemaker (Haddad et al., 2006)

2.10. Biosensors in Pacemaker System

Biosensors are devices for the detection of analytes that combine biological and physico-chemical detector components. Due to the proximity of the sensor units to each other, a biosensor network can have a base station that handles all processing, communications, and power. Biosensors are built into most pacemakers as a programmable option. As sensor technology has advanced, pacemakers are now able to detect many different types of physiological variables and cardiac signals used for diagnostics such as: body physical activity, QT interval, minute ventilation, intraventricular ECG, respiratory rate, intraventricular pressure, oxygen saturation, and impedance. Biosensors are classified into those that detect physical activity and respond to movement (accelerometers) and those that measure the

consequences of physiological changes during exercise or otherwise (QT interval) (W. V. Shi and Zhou, 2012).

In addition, the sensor's role has been expanded to include functions other than frequency boost, such as: the detection of atrial and ventricular capture and monitoring of heart failure, sleep apnea and hemodynamics. Using sensors to monitor cardiac hemodynamics, right ventricular pressure has been shown to be a good estimate of pulmonary artery diastolic pressure and capillary wedge pressure. Therefore, fully implanted devices have been used to reduce hospitalizations for heart failure. Table below illustrates the categories of sensors for pacemaker systems (W. V. Shi and Zhou, 2013) (W. V. Shi and Zhou, 2012).

Table 2.1: Features of main biosensors

Activity sensor (Accelerometer)	Minute Ventilation	QT interval	Dual sensors
Body movement or vibrations	Volume of air inhaled/ exhaled in one minute	Intervals in pacing pulse spike and T-wave	Dependent on two self-sensing devices
Rapid response to exercise	Physiological correlated with metabolic demand	Important ECG diagnostic parameter	Control each other
Less Physiologically accurate	Slow respond to exercise	Unstable chronically	High power consumption

2.10.1. Dual Sensors

Dual sensors are used to prevent excessive rate fluctuations for more accurate diagnostic data analysis, allowing the two sensors to be taken into account with each other. Both sensors monitor each other during crosschecks, and the pacing rate can only be adjusted when both or the dominant sensor agrees (W. V. Shi and Zhou, 2013) (W. V. Shi and Zhou, 2012).

2.10.2. Activity Sensor

Activity sensors are old and common, and react quickly to exercise by evaluating your body's vibrations and movements. Accelerometers record changes in posture and physical activity associated with body movement. Tracking body movements using accelerometers is a simple yet reliable method of action capture (Mulpuru et al., 2017). An activity

sensor is a simple device that can be used for motion tracking applications. A triaxial accelerometer measures acceleration and provides raw data along each axis (Hayes et al., 2008). Because it is non-invasive (the sensing system is attached inside the pacemaker without physical interaction with the body), it is the strategy of choice used in most rate-responsive pacemakers (Petrutiu et al., 2007). Figure 2.14 shows a schematic of how to represent an activity sensor with a pulse generator placed on an accelerometer.

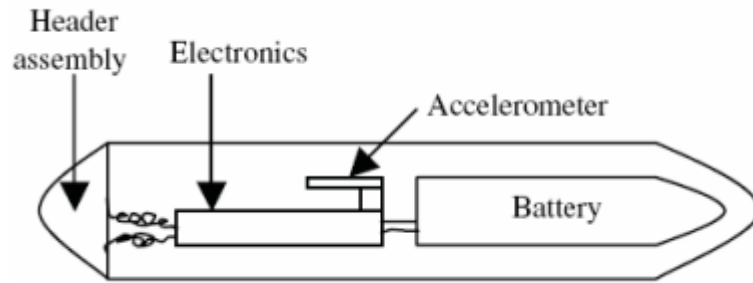


Figure 2.14: Activity sensing system (Hayes et al., 2008)

2.10.3. QT Interval Sensor

It is a form of metabolic sensor that measures the actual heart rate values. QT length in the ECG represents contraction to relaxation of ventricle (W. V. Shi and Zhou, 2012). Long QT intervals on the ECG are associated with an increased risk of arrhythmia and sudden death.

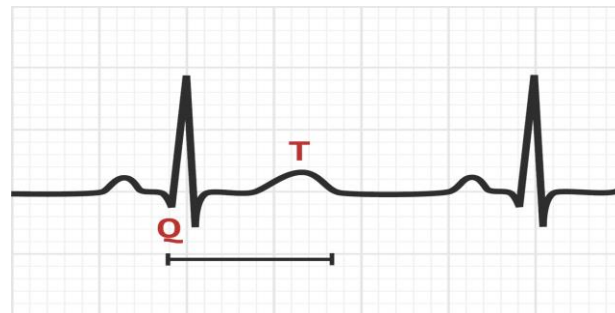


Figure 2.15: QT interval (Prenner et al., 2016)

2.11. Related Works

Srivastava and Kumar (Srivastava and Kumar, 2022) presented the design and hardware implementation of an Adaptive Neuro-Fuzzy Inference System (ANFIS) based controller for a cardiac pacemaker. Improve the parameters of cardiovascular system response in terms of stability, steady-state error, transients, etc. However, the system response takes long settling time around 1.3813sec. The settling time is the major constraint in the design of a

pacemaker controller; ideally, it is required to have no delay between the need of heartbeat and the stimulus delivery by the pacemaker.

Momani et al. (Momani et al., 2019) Designed FOPID heart Rate Controllers for Cardiac Pacemaker. A significant improvement of the step response has been achieved using El-Khazali's approximation over that of others. In particular, using the El-Khazali approximation has the lowest overshoot and shows higher effectiveness than the others. This is important for pacemaker control, but the rise time is slightly longer than resting regulators. The FOPID controller achieved a very fast rise time of 0.2263 seconds and a settling time of 0.3573 seconds with an overshoot of 0.8597%. Furthermore, FOPID controllers are not optimal due to limitations in the tuning process.

Khan et al. (Khan et al., 2020) proposed use of a pacemaker with a FOPID controller in a secure medical cyber-physical system to monitor and stabilize heart rate. In this work, an FOF-PID controller is optimized for using in conjunction with a pacemaker in the heart of a human being. This controller helps in mitigating the overshoots in heart-rate, but it has root mean square error of 0.55 and 1.55% maximum error for the female heart and root mean square error of 0.49 and 0.72% maximum error for male heart. It also needs optimal tuning procedure to overcome the errors.

Fedal and Malathi (Fedal and Malathi, 2018) presented advanced control strategies for cardiovascular system. In this system, an adaptive PID controller based on the delta rule and adaptive correction factors was proposed and compared with a conventional fuzzy controller under different settings, i.e., 65, 75, 85 bpm. The results show excellent performance of the adaptive PID controller achieving an output time response of 0.00003 s rise time and 0.02 s settling time. However, it still has an overshoot of 1.53%.

Aghdam et al. (Aghdam et al., 2017) designed a new algorithm for the next generation of cardiac pacemakers using ANFIS. In this work, different control techniques are analyzed to design Heart Rate controller. Initially a PID controller tuned with the help of MATLAB tune GUI tool to satisfy the different performance parameters. To improve the performance of the system, an intelligent ANFIS controller is designed with ANFIS toolbox, it is observed from the response of fuzzy controller that the parameters (maximum overshoot and RMSE and Maximum error) of ANFIS controller is better as compared to the conventional PID controller and other methods. The ANFIS controller achieved output response of 1.55 sec settling time, and 1.7 sec rise time; however, it still has 1.08% overshooting.

Arunachalam et al. (Arunachalam et al., 2016) presented a robust FOPID controller based on the Ziegler-Nichols tuning method compared to conventional PID controllers and conventional fuzzy controllers in terms of output system response parameters such as rise time, settling time, and overshoot. The FOPID controller achieved a very fast rise time of 0.1 s and a settling time of 0.2 s with a 2.05% overshoot. Furthermore, FOPID controllers are not optimal due to limitations in the tuning process.

Govind and Sekhar (Govind and Sekhar, 2014) introduced a method to regulate the heart rate of the pacemaker. At three different set points, 65, 75, and 85 bpm, an adaptive PID controller based on the Delta rule and adaptive correction factor has been developed and compared with the traditional fuzzy controller. The results show better performance of the adaptive PID controller, which achieved output time response of 0.00003 sec rise time, and 0.02 sec settling time; however, it still has 1.53% overshooting.

Yadav and Rani (Yadav and Rani, 2013) presented a closed-loop, unity negative gain artificial pacemaker system that was operated by two different types of controllers: i) a PID controller tuned with Tyreus-Luyben, Zeigler-Nichols, and relay techniques and ii) the fuzzy controller with two inputs/ one output, nine membership functions for each input/ output, and 81 fuzzy rules for fuzzification and defuzzification process worked at (65, 75, 85) bpm different set points. The result shows the superiority of the fuzzy controller compared to the PID controller; however, it still has overshooting percentage level ranging from 2-3.5%.

Shi and Zhou (W. V. Shi and Zhou, 2011a) proposed fuzzy PID controller for dual sensor pacing system for bradycardia patient at rest. New cardiac layout, sensor device, and comparative tests revealed the dominance of the FPID controller relative to fuzzy logic controller in obtaining a better correlation between natural heart rate and desired profile. This model, though, has several disadvantages, such as 2.32.6 percent error levels and obtained a low overshooting result with gradual time increase and time settlings.

Shi and Zhou (W. V. Shi and Zhou, 2012) Presented an overview of the body sensors used in pacemakers, new features and advancements in the latest pacemakers introduced, and advances in the various types of body sensors incorporated in pacemakers, presenting their rationale, function, and application. Using one sensor is not ideal to for heart rate adaptation. So, combining different kinds of sensors (dual sensors) is better than one sensor for optimal rate adaptation.

Zhou et al. (Zhou et al., 2010) Presented a comparative evaluation of six algorithms to measure QT interval prolongation using simulated ECG. However, relying on just one sensor is not ideal. However, using two sensors will improve the performance of heart rate detection and adjustment.

López et al. (López et al., 2010) Simulated heart rhythm dynamic and control. It was implemented through proportional control law based on ECG signal reference tracking, and the control algorithm was tested for various cardiac pathologies. For the simulation of cardiac rhythm dynamics, a mathematical model based on six differential equations with delay time was used. The result was noticed that the system, with a very low error magnitude, returned the chaotic unpredictable rhythms to the standard rhythm. In this case, the authors obtained the controller gain manually.

Thanom et al. (Thanom et al., 2012) presented a nonlinear system control hypothesis focused on linearization of input output and an observer were introduced to estimate unknown variables. For this analysis, Zeeman models were selected because they not only explain the heartbeat but also have a clear biophysical connection to the complex variables with a minimum phase. The simulation findings revealed with asymptotic stable tracking error and the outputs of the system tracks the existing ECG data from the Physio Net database.

Finally, it is concluded from preceding studies the pacemakers' great difficulty is to develop an effective controller that tracks and generates ECG data with a near matched rate of pacing by eliminating abnormal rhythm correlated with cardiac pathologies. Observer dependence is, however, one of the key challenges of the methods mentioned in feasible applications. In addition to tracking and generating ECG signals problem there is a great challenge also in design of a pacemaker controller for tracking and regulating heart rates that satisfies all desired response parameters, including the rise time, and settling time without overshooting and with a small permissible error. Consequently, the present work proposed a design of ANFIS based controller gains help to achieve close pacing rate compared to the desired heart rate with fast time response.

CHAPTER THREE

3. METHODOLOGY AND ANALYSIS OF HEARTBEAT MODEL

3.1. Introduction

This chapter mainly focuses on the methodology that has been followed in this thesis work. The mathematical model of the human heartbeat, cardiovascular system and analysis of heartbeat model is discussed.

3.2. Materials

In this research work, software such as MATLAB/2021a, Microsoft Office 2021, and Math-Type 7.0 equation editor are used. MATLAB is computing software which consists of technical toolboxes and Simulink. Microsoft office is used for editing the thesis documentation. Math Type 7.0 is used for writing mathematical formulas and equations in Microsoft office word and PowerPoint presentation tools. Mendeley is software. It is installed on the computer, which automatically inserts citations of the reference paper and bibliography. Edraw-Max is also block diagram and flowchart maker. It is used to draw the diagram that is not easy to draw on Microsoft office word.

3.3. Methods

The methodology of this thesis mainly focuses on software algorithms. To attain the general and specific objectives of the study stated in the previous chapter, the following methods and techniques will be followed. Starts from defining the problem, to collect information related to the cardiac pacemaker system. After analyzing all of the resources, the next step is to complete the mathematical modeling for the cardiovascular system. Depending on the mathematical model and the characteristics of the cardiac pacemaker system, the effective controller for this system is selected and the parameters of the controller are tuned using genetic algorithm. To evaluate the system's performance, MATLAB/Simulink will be used as simulation tools.

In general, the method followed throughout the research study can be summarized in the following flow chart of Figure 3.1 below.

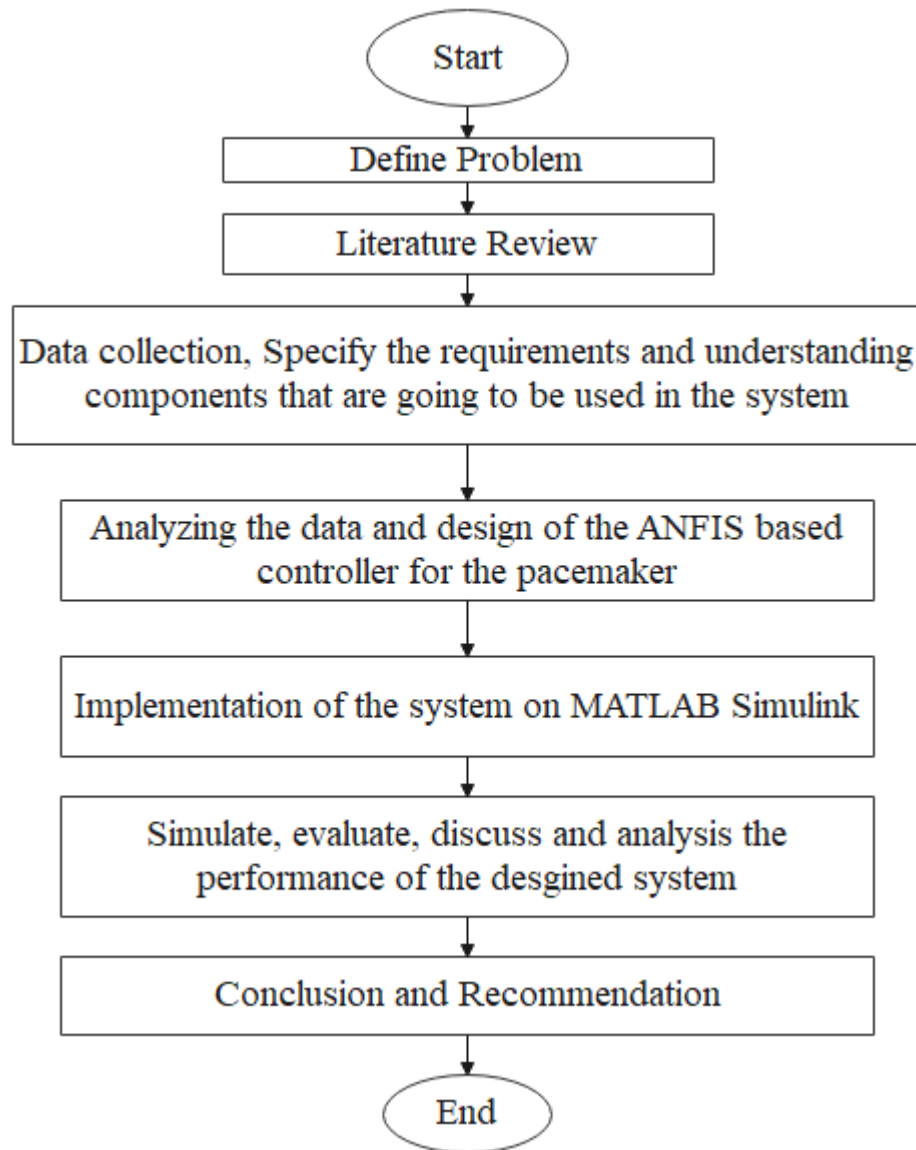


Figure 3.1: Work flow

3.4. Block Diagram of the Proposed System

The suggested system block diagram of the ANFIS based FOPID controller for cardiac pacemaker system is a single input single output system, as depicted in Figure 3.2 below. ANFIS is a network structure that makes it easier to find and refine membership function parameters using either the back propagation gradient descent (GD) approach alone or in combination with Least Squares Estimates (LSE). To acquire ideal timing settings for the cardiac pacemaker system, a FOPID controller is tuned properly. These data are used as input to train with ANFIS controller's neural network.

The pacemaker rate is adaptive because the controller enables modification in stimulus delivery to the heart based on the human body. The controller detects the heart's intrinsic

pacing and sends the appropriate control signal to the pacemaker's pulse generator. The time duration, amplitude, and sensitivity of the pulse to be sent to the heart are all contained in this control signal.

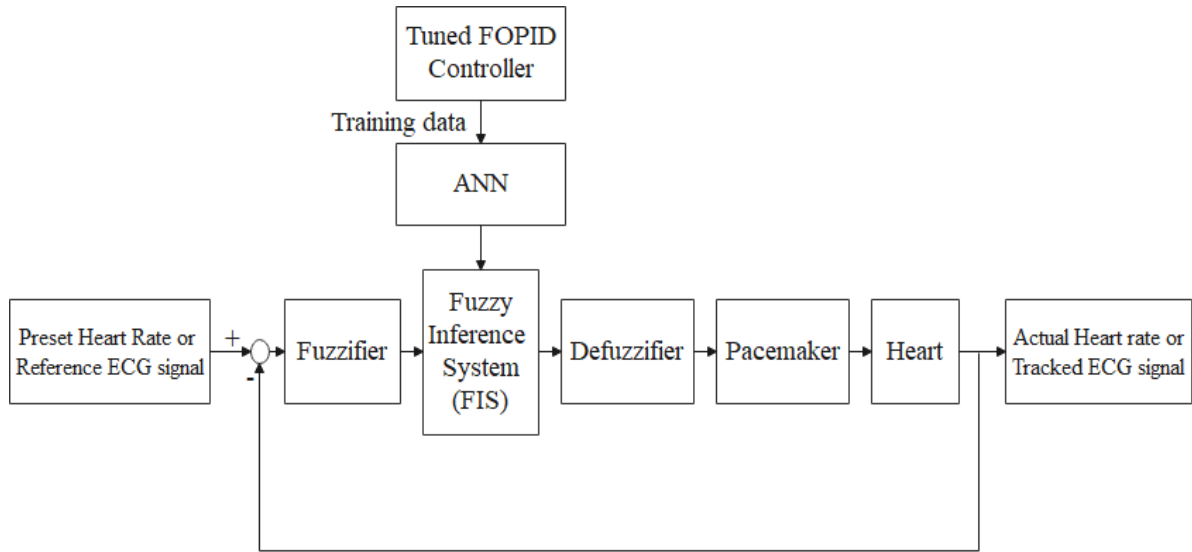


Figure 3.2: Block Diagram of ANFIS based FOPID controller for cardiac pacemaker system

3.4. Mathematical Model of Human heart

The mathematical modeling of biological signals is extremely difficult, and this is relatively new study. This allows for a better understanding of the physical processes that result in numerous physiological signals in the human body, such as the electrocardiogram (ECG), electroencephalogram (EEG), electromyography (EMG), and so on (Leif and Laguna, 2005) (Noble, 1962). These physiological signals are frequently used by physicians throughout the year to diagnose abnormal human organ function.

Various modeling approaches with different levels of complexity have been developed for replicating electrical processes of the human heart, complete or partial, linear or nonlinear models. Some models use a heart's dynamical system to generate genuine ECG data (Abdelhady et al., 2019)(Selgrade, 1994)(Jones et al., 2009)(Zeeman, 1973). Others use various modifications of nonlinear relaxation oscillators, notably coupled van der pol oscillators, to create a heart model based on natural cardiac pacemakers (Dos Santos et al., 2004)(Grudziński et al., 2004)(Gois et al., 2009).

3.4.1. Heartbeat Modeling Using Dynamic System

In 1972, Zeeman published a paper (Zeeman, 1973) that looked at two types of excitable biological systems: nerves and the heart. He created an electric neural stimulation system using nonlinear-coupled differential equations and van der pol linear equations. Both models focus on the pace making triggered by the Sino Atrial node, which is a superior pacemaker than the atrioventricular (AV) junction's slower one. The sympathetic and parasympathetic modulations that are responsible for HRV production were not incorporated in these models (Jones et al., 2009). The Zeeman heartbeat model is described as follows:

$$\begin{aligned}\varepsilon \dot{y}_1 &= -(y_1^3 - Ty_1 + y_2) \\ \dot{y}_2 &= y_1 - y_d\end{aligned}\tag{3.1}$$

Where $y_1(t)$ defines the length of muscle fiber in the heart, $y_2(t)$ is the variable related to electrochemical activity, and could be measured as membrane potential across muscle fiber. ε is a small positive constant associated with the fast eigenvalues of the system, and y_d is the average cardiac muscle fiber length during diastole (the duration of time when the blood is refilled by the heart contraction). $T > 0$ is the muscle tension related to blood pressure.

Zeeman's second-order heartbeat differential equations (ODE) have been revised by adding a switch (on/off) control variable showing the contraction and relaxation function of the pacemaker to the heart in (Jafarnia-Dabanloo et al., 2007). It will be described as follows:

$$\begin{aligned}\varepsilon \dot{y}_1 &= -(y_1^3 - Ty_1 + y_2) \\ \dot{y}_2 &= (y_1 - y_d) + (y_d - y_s)u(t), T > 0\end{aligned}\tag{3.2}$$

Additional parameters, y_s represent the typical length of muscle fiber during the systolic state, $u(t)$ the heart pacemaker control signal that guides the heart into diastole and systolic states. (Jafarnia-Dabanloo et al., 2007) revised the third-order Zeeman nonlinear model by introducing control parameter which influences the oscillation frequency to regulate the HRV utilizing a neural network to produce ECG signal (Van Der Pol et al., 2000). The third order heartbeat is described as:

$$\begin{aligned}\varepsilon \dot{y}_1 &= -(y_1^3 + y_1 y_2 + y_3) \\ \dot{y}_2 &= -2y_1 - 2y_2 \\ \dot{y}_3 &= -y_2 - 1 + u(t)\end{aligned}\tag{3.3}$$

Where $y_1(t)$ represents the length of muscle fiber; $y_2(t)$ represents a tension in muscle fiber; $y_3(t)$ related to electrochemical activities; $u(t)$ describes the cardiac pacemaker control signal that guides the heart into diastole and systole states. The model dynamics of the third-order heartbeat system have no significant difference with the second-order system except tension in muscle fiber caused by blood pressure is considered as a state variable.

3.4.2. Heartbeat Modeling Using Natural Pacemakers

The oscillator can be seen as the starting point of modeling because the qualitative properties of the heart's excitation potential are extremely close to the classical Van der pol dynamical behavior. The AV node is increasingly involved in pacemaking thanks to the combined Van der pol (VdP) oscillator models. The model aids in comprehending the effort required for genuine electrophysiological dynamics to be coupled between SA and AV pacemakers.

In one of the earliest research, Van der Pol and Van der Mark discovered that HECS (Heart Electrical Conduction System) was the coupled VdP oscillator in 1928 (Van Der Pol et al., 2000). The ECG, on the other hand, was not created.

$$\ddot{y} + \alpha(y^2 - \mu)\dot{y} + w^2 y = F(t) \quad (3.4)$$

Where, $\alpha > 0$ and $\mu=1$ nonlinear dumping forces; w normal frequency of oscillator; $F(t)$ external driving pulse; $y(t)$ corresponds to the electrical heart pulse amplitude (action potential). Then, many studies have been developed to model cardiac rhythms. A modification of the original Van der Pol oscillator was suggested in (Grudziński et al., 2004) to capture some significant aspects of the potential for heart activity represented by the natural pacemaker.

$$\ddot{y} + \alpha(y - v_1)(y - v_2)\dot{y} + \frac{y(y+d)(y+e)}{ed} = F(t) \quad (3.5)$$

Where, modifies the pulse shape, which changes time that the heart receives the stimuli, v_1 , and v_2 replaces the damping terms, e controls the atrial or ventricular contraction period, d is a parameter that arises when the harmonic forcing of the classic equation is replaced by cubic terms and is an external forcing. Where, $v_1 v_2 < 0$ and $d, e, a > 0$.

The model, further, was modified (Dos Santos et al., 2004), who presented a model of the cardiac dynamics composed of two coupled modified Van der Pol oscillators, representing the behavior of two pacemakers.

$$\begin{aligned}
\dot{y}_1 &= y_2 \\
\dot{y}_2 &= k(y_1 - w_1)(y_1 - w_2)y_2 - b_1y_1 + c_1(y_3 - y_1) + a_1 \sin(w_1t) \\
\dot{y}_3 &= y_4 \\
\dot{y}_4 &= k(y_3 - w_1)(y_3 - w_2)y_4 - b_2y_3 + c_2y_3 + c_2(y_2 - y_3)
\end{aligned} \tag{3.6}$$

Where, the pairs (y_1, y_2) and (y_3, y_4) , show SA and AV Oscillators, respectively, and $k, w_1, w_2, b_1, b_2, c_1, c_2$ represent model parameters (Gois et al., 2009) simulated ECG signal via a mathematical model consisting of three updated Van der Pol oscillators, representing the Sino Atrial(SA), Atrioventricular(AV), and the His-Purkinje mechanism, linked by time delay couplings. Numerical calculations were conducted to generate the ECG signal, which could be changed by adjusting the parameters of the coupling.

3.4.3. Mathematical Modeling of Human Heart Using Transfer Function

The dynamics of the cardiac muscle any represented by the following model in Figure below (Khan et al., 2020).

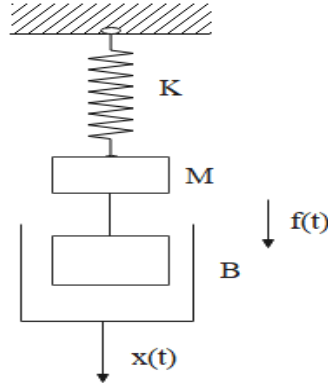


Figure 3.3: Basic mechanical model of cardiac muscle (Khan et al., 2020)

With the use of control modeling, it is possible to obtain the dynamic equation for the movement of the heart muscle from the Figure 3.3 above, as shown below.

$$M \frac{d^2x}{dt^2} + B \frac{dx(t)}{dt} + Kx(t) = f(t) \tag{3.7}$$

Where M denotes the mass of cardiac muscle, B is constant of viscous drag of one myocardial muscle, K represents the elasticity of cardiac muscle, $x(t)$ is the movement of cardiac wall generated due to the exerted force, and $f(t)$ is the force by electrical and electrochemical activity effects on the vascular system. Take the Laplace transform of equation (3.7):

$$Ms^2 X(s) + BsX(s) + KX(s) = F(s) \tag{3.8}$$

$$\frac{X(s)}{F(s)} = \frac{1}{Ms^2 + Bs + K} \quad (3.9)$$

$$\frac{X(s)}{F(s)} = \frac{K_1}{s^2 + k_2s + k_3} = T(s) \quad (3.10)$$

Where, $K_1=1/M$, $K_2=B/M$ and $K_3=K/M$

Now compare this equation with the second-order transfer function.

$$\frac{X(s)}{F(s)} = \frac{w_n^2}{s^2 + 2\xi w_n s + w_n^2} = \frac{K_1}{s^2 + K_2s + K_3} = \frac{\frac{1}{M}}{s^2 + \frac{B}{M}s + \frac{K}{M}} \quad (3.11)$$

From the equation (3.11)

$$w_n = \sqrt{\frac{1}{M}} = \sqrt{\frac{K}{M}}, \xi = \frac{B}{2\sqrt{M}} \quad (3.12)$$

The damped frequency of oscillation is given by:

$$w_d = w_n \sqrt{1 - \xi^2} \quad (3.13)$$

Considering the heart rate as 72 beats per minute; that is, the damped frequency (f_d) is given by $f_d = 72/60 = 1.2\text{Hz}$. So that the damped frequency of oscillation of heart as $w_d = 2\pi f_d = 7.54$ radian/second. Due to various condition of mankind like blood pressure, blood sugar, mental stress, tumor etc. the heartbeat may change causing the variation in the values of damping coefficient. As heart is under damped palpitating organ, it was assumed that the (damping coefficient) $\xi=0.815$ for this study.

By using equation (3.13), the value of w_n will be obtained the follows:

$$w_n = \frac{w_d}{\sqrt{1 - \xi^2}} = \frac{7.54}{\sqrt{1 - 0.815^2}} = 13.01 \text{ rad / sec}$$

Now putting the values of ξ and w_n in the forward path transfer function of a unity feedback second order system, I get:

$$\frac{w_n^2}{s(s + 2\xi w_n s)} = \frac{13.01^2}{s(s + 2*0.815*13.01s)} = \frac{169.2601}{s^2 + 21.2063s}$$

Which is close to the actual transfer function of heart $G_h(s) = G_h(s) = \frac{169}{s^2 + 20.8s}$ (Khan et al., 2021). Therefore, practically the transfer function varies depending on the damping coefficient and damped frequency of oscillation. In this thesis work the actual transfer function of heart that used with the proposed controller to track and regulate different heart rates is:

$$G_h(s) = \frac{169}{s^2 + 20.8s} \quad (3.14)$$

The pacemaker block above block diagram in Figure 3.2 is a first-order linear time invariant system which can be represented as:

$$G_p(s) = \frac{K}{1 + s\tau} \quad (3.15)$$

Where k and τ represent the steady-state gain and time constant respectively. With $\tau = 1/2\pi f_d = 1/(2\pi (1.2)) \approx 1/8$ second and $k = 1$, the transfer function of the pacemaker becomes:

$$G_p(s) = \frac{8}{s + 8} \quad (3.16)$$

Therefore, the pacemaker behaves as a frequency selective device which blocks high frequency signals from affecting the heart of a cardiac patient. The proposed ANFIS FO-PID controller actuates the pacemaker to overcome the problems of bradycardia (slow heart rate) and tachycardia (high heart rate) in cardiac patients and stabilizes the heart rate.

Currently, there is no well-recognized mathematical models related to the activities of heart-beat. However, in this study mathematical modeling of human heart using transfer function was used to track and regulate different heart rates and also Zeeman second order nonlinear heartbeat model was used to track ECG signals.

3.5. Zeeman Heartbeat Dynamics and Model Validation Analysis

There are two extreme states of balance during the heartbeat cycle: diastole, which is the relaxation phase, and systole, which is the contraction phase. The presence of a pacemaker at the apex of the atrium helps the heart beat more efficiently. Systole is now possible for the heart. In other words, it generates an electrical pulse that progressively travels through the atria, contracting muscle fibers and forcing blood into the ventricles, contracting the ventricle and sending a huge blood pump down to the arteries (Zeeman, 1973). A good

mathematical model which describes the heartbeat activity should contain three basic characteristics(Thanom and Loh, 2010):

- i. A stable equilibrium phase representing diastole;
- ii. A threshold to activate the electrochemical wave that triggers the heart to reach the systole and
- iii. The rapid returning to diastolic state.

The original Zeeman model in equation 3.1 is used to perform the above criteria and the following parameter values have been taken for analysis.

Table 3.1: Parameters of the model (Sargolzaei et al., 2014) (Thanom et al., 2012)

Parameter	y_d	ε	T	y_s
Value	1.024	0.2	1	-1.3804

For $y_d=0$, the equilibrium point is at $(0, 0)$.

$$A = \begin{bmatrix} \frac{-1}{\varepsilon}(3y_1^3 - T) & \frac{-1}{\varepsilon} \\ 1 & 0 \end{bmatrix},$$

Where, A is the Jacobean matrix for equation (3.1).

At an equilibrium point $(0, 0)$, A can be defined as:

$$A = \begin{bmatrix} \frac{T}{\varepsilon} & \frac{-1}{\varepsilon} \\ 1 & 0 \end{bmatrix}$$

Since both eigenvalues A are positive i.e., 3.618 and 1.382, the origin is unstable.

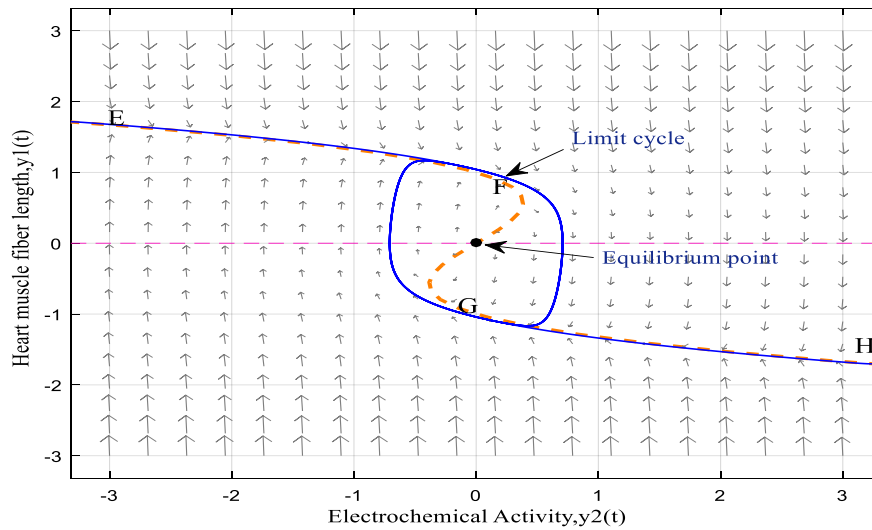


Figure 3.4: Phase portrait of heartbeat system for $\varepsilon=0.2$, $T=1$ and $y_d=0$

Figure 3.4 shows the phase portrait of equation (3.1) with the initial conditions. The steady state was represented by the dashed curve. In the limit cycle, both paths end at the same place. Because the vector field in the limit cycle points away from the equilibrium point, the equilibrium point is clearly unstable. The vector field over portions EF and GH commonly leads to the dotted line in figure 3.4, indicating stable regions, whereas points along the segment FG are unstable. Points F and G are crucial since they define the threshold, ensuring that the heartbeat model selection criteria are met (ii).

The eigenvalue is negative for $3y_1^2 - T > 0$. Therefore, the system model is stable if $y_1 \geq \sqrt{T/3}$ which belongs to the segment EF, and $y_1 \leq -\sqrt{T/3}$ which belongs to segment GH. Besides, the thresholds for triggering from diastolic to systolic states on point F i.e., $y_1 = \sqrt{T/3}$, and $y_1 = -\sqrt{T/3}$ on point G.

The diastole state can be calculated by varying value of y_d in such a way as to fulfill the above stabilization condition. The requirements will be met if the y_d is 1.024. Therefore, the stable equilibrium is at (1.024, -0.049) because of the eigenvalues are negative i.e., -0.48826 and 10.2404.

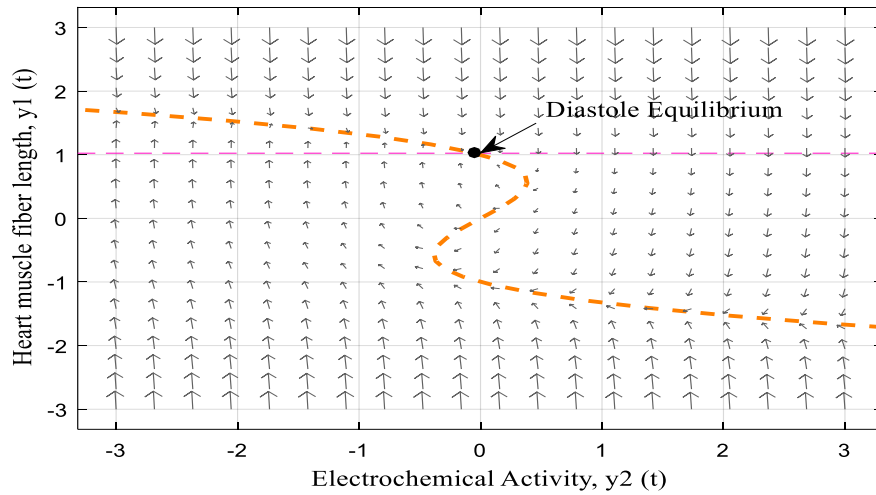


Figure 3.5: Phase portrait of heartbeat system during diastole for $T = 1$, $\varepsilon = 0.2$ and $y_d = 1.024$

The phase portrait of the system for $y_d = 1.024$ was shown in Figure 3.5. The equilibrium point at (1.024, -0.0497) is stable and defines the diastole equilibrium condition, i.e. it satisfies the first characteristic i: a stable equilibrium. In this situation, regardless of the primary conditions, all trajectories reach the diastole equilibrium point. Except for an external excitation that causes the system to transition to a new point of equilibrium, the system remains

at the same point of equilibrium indefinitely. The heartbeat model is merged with a control input in equation (3.17) based on this new stable system (equation 3.2) as written below:

$$\begin{aligned}\varepsilon \dot{y}_1 &= -(y_1^3 - T y_1 + y_2) \\ \dot{y}_2 &= (y_1 - y_d) + (y_d - y_s) u(t), T > 0\end{aligned}\tag{3.17}$$

Another parameter has been added: y_s , which describes average fiber length when the heart is in systole, and $u(t)$, which describes the control mechanism of the cardiac pacemaker that guides the heart into diastole and systole. By inserting a pacemaker signal in either on or off form, the heartbeat system's equilibrium condition between diastole and systole would be modified.

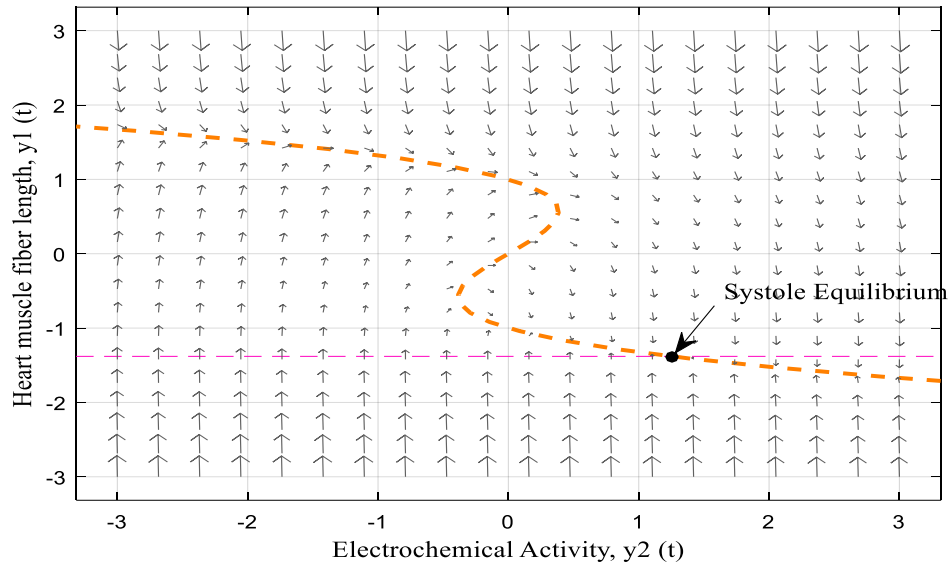


Figure 3.6: Phase portrait of heartbeat system during systole

Figure 3.6 shows the phase portrait plot when $u(t)$ was turned on with $y_s = -1.3804$. A stable equilibrium point is found at $(-1.3804, 1.25)$. As a result, the on off control technique is effective in mimicking state transitions from diastole to systole dependent on the control signal (III satisfied). The physical behavior of the heart is supposed to meet the three criteria listed above. As a result, the heartbeat model is validated.

3.6. Considered Disturbances of Cardiovascular System

Feelings, thoughts, action, and all other processes are controlled by the brain. Scientists estimate that a typical adult human brain contains one hundred billion neurons, with each neuron linked to two hundred fifty thousand others. In contrast to skeleton muscles, which require nervous system simulation, the heart can generate its own electrical stimulation.

Interestingly, the heart can beat without brain orders, but it must be changed in situations when the brain controls the heart rhythm, such as (exercise, hormones such as epinephrine (adrenaline)) are released (typically connected with sensations of apprehension, anxiousness, and excitement) (Amy Kuklev). The neurological system can make the heart rate move faster or slower, but it cannot make it go faster or slower. It is quite common for the if overdosed, the cardiovascular system's function will be disrupted. The frequency bands of brain waves are commonly split into four categories: Beta, Alpha, Delta, and Theta. Delta has a low frequency band and high amplitude, whereas beta has a high frequency band and small amplitude(Aabid et al., 2016).

Table 3.2: Brain wave frequency bands (Aabid et al., 2016)

Categories of brain wave taken as a disturbance		
BETA	(14-100Hz)	Arousal, Anxiety, disease, fight
ALPHA	(8-13.8Hz)	Pre-sleep, pre-walking, drowsiness
THETA	(4-7.9Hz)	Deep mediation, Hypnagogic
DELTA	(1-3.9Hz)	Loss of body awareness, Dreamless

3.7. Heart Rate Calculations and Reference ECG signal

To determine the heart rhythm, the ECG signal is measured using wavelet transforms built in function. It can be determined using the following formula (Kabari and Onwuka, 2019).

$$\text{Heart Rate} = \frac{60}{\text{R-R interval in seconds}} \quad (3.18)$$

A healthy human's heart rate is 72 beats per minute. The average person's heart rate is 60–100 beats per minute (Kabari and Onwuka, 2019). Figure 3.7 shows the real signal, which was collected from the MIT-BIH physio Net database under the name 16773m.mat and matlab code to generate the real ECG signal is shown in appendix [D] . In the lower channel, which has a sample frequency of 128Hz and 1280 columns (samples/signal), there are irregular high-frequency and severe noise surges. Pre filtering was used to remove noise from the ECG data using high pass filtering. The ECG signal is then squared to remove the negative and make the R-R peaks highly apparent or prominent.

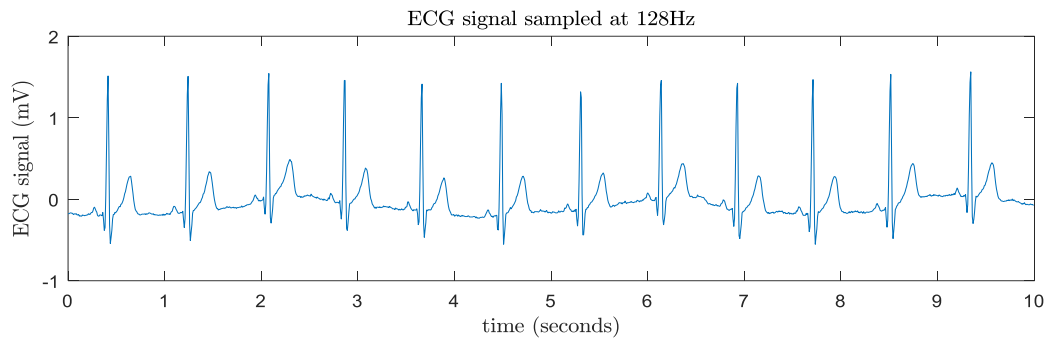


Figure 3.7: ECG data

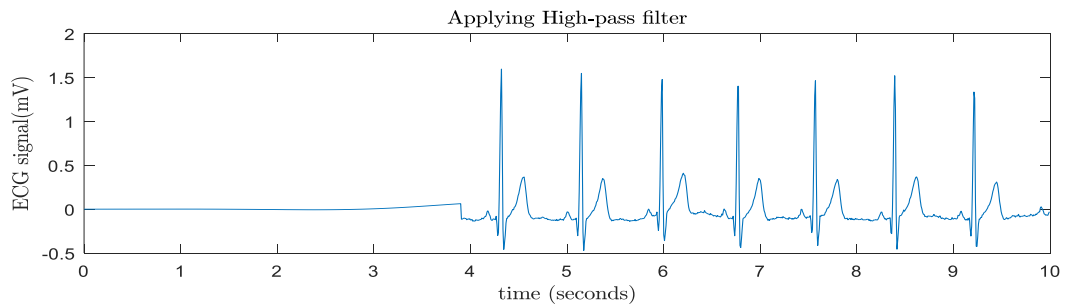


Figure 3.8: Filtered ECG signal

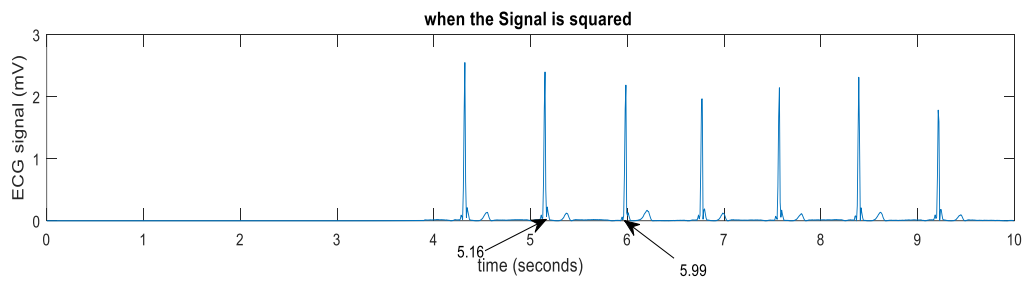


Figure 3.9: Squared ECG signal

From Figure 3.9, the R-R peak interval is calculated as:

$$\begin{aligned} \text{Heart rate} &= \frac{60}{R - R \text{ intervals in seconds}} \\ &= \frac{60}{5.99 - 5.16} = \frac{60}{0.83} \cong 72 \text{ bpm} \end{aligned}$$

CHAPTER FOUR

4. CONTROLLER DESIGN

4.1. Introduction

There are various types of control methods that can be employed to control a system. In the investigation of nonlinearities, especially in biological systems, regular and aberrant dynamic behavior has been described, and it differs from person to person. For decades, the behavior of biological systems has been observed linearly, removing a wealth of information on the natural processes at hand. According to new research, certain cardiac arrhythmias include nonlinear features that are frequently associated with a disordered response. The nonlinear system theory and the chaos associated with biological systems have attracted scholarly interest (Lopes and Morgado de Gois, 2018).

Nonlinear control methods are increasingly widely employed and can be used to both linear and nonlinear systems. These controllers can be applied to both linear and nonlinear systems, addressing issues such as system uncertainty and external disturbance resistance.

Researchers have claimed that fractional order systems are better at modeling various materials than traditional methodologies in the previous two decades. Fractional-order systems have a significant impact on controller system behavior, such as increasing response speed and decreasing steady-state error and relative stability.

In this chapter, FOPID and ANFIS based FOPID controllers are designed for tracking and regulating human heart rate and tracking of ECG signal. The parameters of the controllers are tuned using a genetic algorithm (GA) based on an objective function.

4.2. Review of FOPID Controller

Control systems and/or controllers that are characterized by fractional order differential equations are referred to as fractional order control (FOC). Recently, there has been increasing interest in improving the performance of ordinary PID controllers using the concept of fractional calculus with non-integer orders of derivatives and integrals (Sáez et al., 2010). A.Oustaloup first introduced the fractional-order controller in 1991 with the Commande Robuste d'Ordre Non-Entier (CRONE) controller. Later, in 1999, Igor Podlubny introduced the most popular form of fractional-order PIDs in the form of $PI^\lambda D^\mu$. This includes an integrator of order λ and a differentiator of order μ , where the values of λ and

μ are 0 to 1. The values of λ and μ allow many choices for the type of controller. He also showed that the response of this type of controller is superior compared to his traditional PID controller. Better control of dynamic systems described by fractional-order mathematical models is one of the most important advantages of FOPID controllers. Another advantage is that FOPID controllers are less sensitive to changes in the properties of the controlled system. This is due to the two additional degrees of freedom that allow a fractional-order control system to better modify its dynamical features. However, up till now, there is no systematic way to set the value for λ and μ (Belaso Beyene, 2021). The fractional integrodifferential equation of the FOPID controller is given by (Verma et al., 2017):

$$u(t) = k_p e(t) + k_d D^\mu e(t) + k_i D^{-\lambda} e(t) \quad (4.1)$$

The transfer function of the FOPID controller is obtained through Laplace transform as follows:

$$G_c(s) = \frac{U(s)}{E(s)} = k_p + \frac{k_i}{s^\lambda} + k_d s^\mu \quad (4.2)$$

Here $E(s)$ is an error and $U(s)$ is the controller's output. It is clear that a fractional-order PID controller not only needs to be designed with three parameters K_p , K_i , and K_d but also two fractional-orders λ and μ for the integral and derivative controllers. Figure 4.1 shows the block diagram configuration of the FOPID controller.

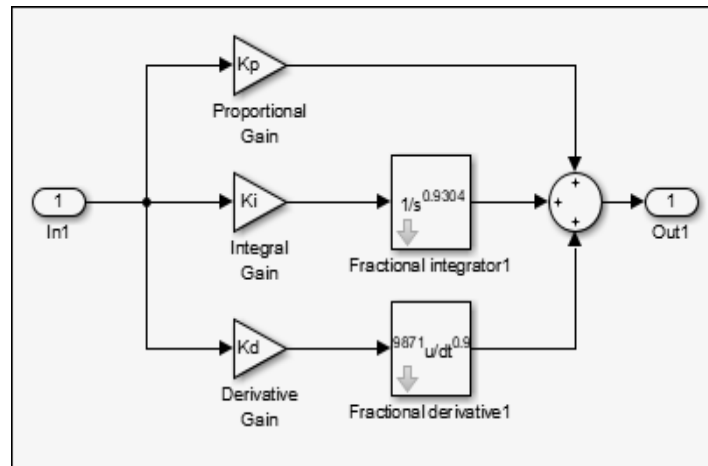


Figure 4.1: Simulink block diagram of the FOPID controller

In MATLAB the fractional-order PID controller can be implemented using the FOMCOM Toolbox. As shown in Figure 4.2, the fractional-order PID controller generalizes the ordinary integer-order PID controller and expands it from point to plane. Point (0, 0) corresponds

to P controller, point (0, 1) corresponds to ordinary PD controller, point (1, 0) corresponds to an ordinary PI controller, and point (1, 1) corresponds to ordinary PID controller, whereas the shaded portion between four corners represents the FOPID controllers. All these classical types of PID controllers are special cases of the FOPID controller when the values of λ and μ are integer values of 0 or 1 (Belaso Beyene, 2021).

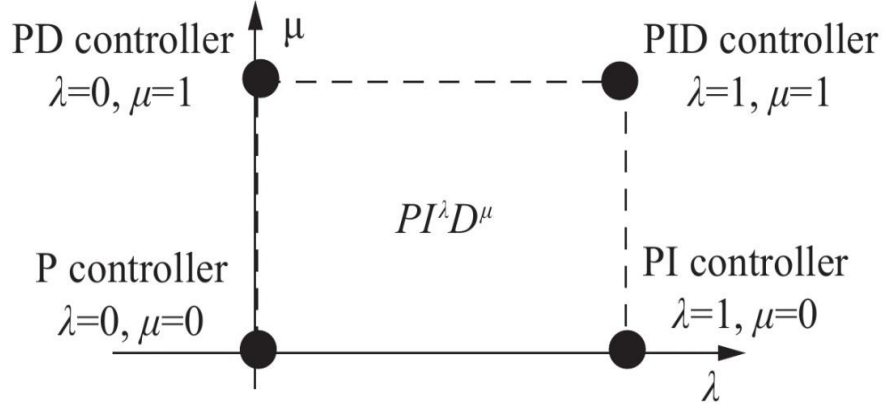


Figure 4.2: Pictorial representation of the fractional-order PID controller (Belaso Beyene, 2021)

The FOPID control loop is represented by the block diagram as shown in Figure 4.3, where $G_p(s)$ and $G_c(s)$ are the plant and the FOPID controller models, respectively. The FOPID receives the error signal $E(s)$ and produces the control signal $U(s)$ to control the output signal $C(s)$ and regulate the disturbance signal $D(s)$, referring to the reference input $R(s)$.

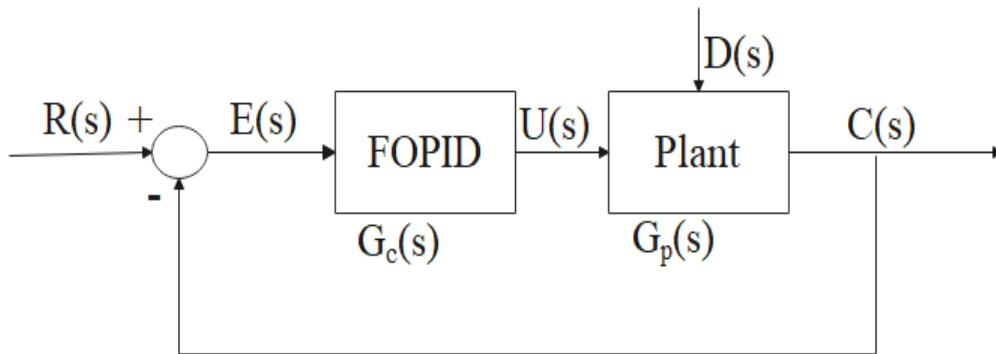


Figure 4.3: FOPID control loop

4.2.2. FOPID Controller Design for the Tracking of the ECG Signal

The second-order heartbeat model in Equation (3.17) is considered to design a control signal, by letting $y(t) = y_2(t)$ as output measurement (measured as potential across muscle fiber) and $y_1(t)$ as desired signal. The tracking error rate is defined as:

$$e(t) = y_r(t) - y(t) \quad (4.3)$$

The objective is to enforce $e(t)$ to 0 and $y(t)$ to $y_r(t)$ as time t approaches to infinite. The heartbeat model correlated with disturbance is defined as:

$$\begin{aligned}\varepsilon \dot{y}_1 &= -(y_1^3 - Ty_1 + y_2) \\ \dot{y}_2 &= (y_1 - y_d) + (y_d - y_s)u(t) + d(t)\end{aligned}\quad (4.4)$$

Equation (4.4) is a nonlinear function and has uncertainty. Additionally, the $d(t)$ assumed to be a system disturbance. The following assumptions are required in order to design the control input.

- System states are observable.
- $d(t)$ is unknown, however, constrained, i.e., $d(t) < D$. Where, the positive constant of $|D|$ is known.
- The control rule is based on the ECG signal.

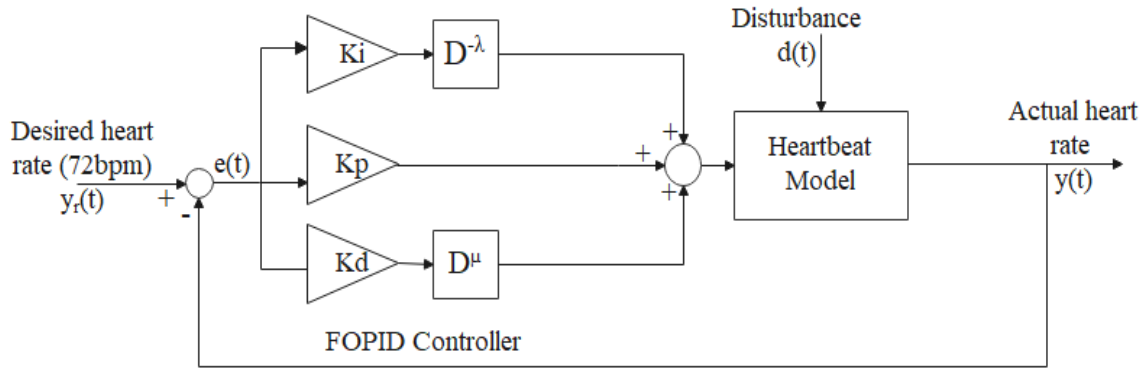


Figure 4.4: Block diagram of FOPID controller for tracking of ECG signals

4.2.3. FOPID Controller Design for Heart Rate Regulation

The closed-loop system for the cardiac pacemaker with controller and unity feedback, is shown in Figure 4.5. Such system illustrates the heart rate regulating model which contains a controller, a pacemaker, and the heart. Actual transfer function of heart in equation (3.14) and the transfer function of the pacemaker in equation (3.16) are considered to design a control signal.

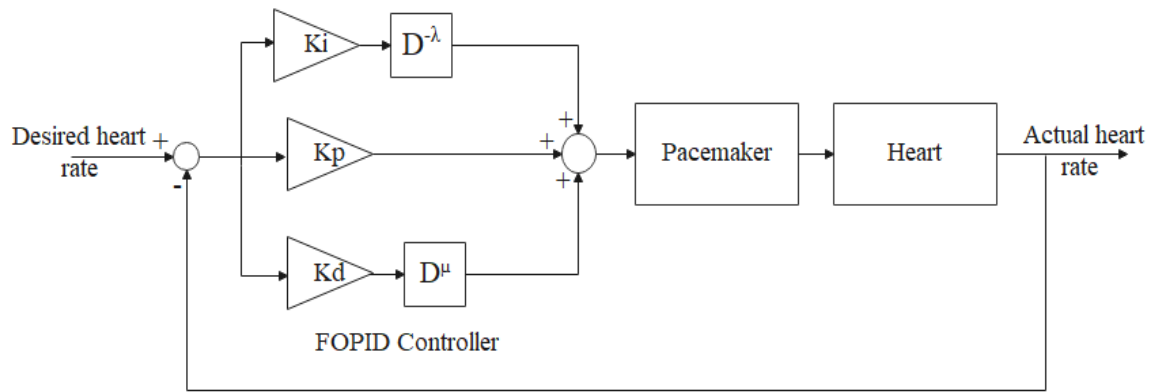


Figure 4.5: Block diagram of FOPID controller for heart rate regulation

The parameters of the FOPID controller are tuned using FOMCON Toolbox. The integer order system is identified to fractional order system by FOMCON Toolbox. FOMCON stands for ‘‘Fractional–Order Modeling and Control’’. It is a MATLAB toolbox and is built upon an existing mini toolbox called FOTF (Fractional–Order Transfer Functions). FOMCON provides time domain and frequency domain fractional order system analysis and confirming system stability (Zheng and Teng, 2016). The FOMCON toolbox consists of the following modules:

- Main module (fractional system analysis). The system analysis module is the main module of the FOMCON toolbox. It serves as a foundation for fractional-order control systems engineering and hence for all other modules.
- Identification module (system identification in the time and frequency domains).
- Control module (fractional PID controller design, tuning and optimization tools as well as some additional features).

The controller design is based on MATLAB Graphical user interface. To design a FOPID controller using MATLAB the author strictly followed the following steps:

1. First add FOTF in MATLAB,
2. Checking the stability of plant,
3. From FOTF Viewer click Tools menu and select Fractional PID design tools,
4. Click tuning tools of Fractional PID design and choose Optimize button,
5. From FPID optimization tool approximation filter, maximum and minimum tuning values, performance metric and maximum number of iterations are determined,
6. Finally click the optimize button and check optimal values of FOPID controller that stabilizes the cardiac pacemaker control System.

The lower and upper bounds for tuning selected here is tuning all parameters with the range of $k_p \in [1, 1.5]$, $k_i \in [0.01, 0.2]$, $k_d \in [0.1, 0.5]$, $\lambda \in [0.2, 0.4]$ and $\mu \in [0.9, 0.95]$ but available options are tune all parameters, fix exponents, and fix gains. Fractional PID design tool graphical user interface, FPID optimization tool, FOTF viewer graphical user interface and new fractional order transfer function details available in Appendix [F].

Table 4.1: Using FOMCON Toolbox Simulated in MATLAB the Optimal Values of FOPID Controller Parameters

Controller	Optimized parameters and values					
FOPID	Parameter	Kp	Ki	Kd	λ	μ
	Value	1.3644	0.19908	0.28406	0.23148	0.94378

4.3. GA Optimization Technique

The most popular evolutionary algorithm is the Genetic Algorithm (GA), invented by John Holland at the University of Michigan in 1975. This type of algorithm uses mutation, inheritance, crossover, and selection, methods inspired by evolutionary biology in order to find solutions to problems. GA is a type of searching algorithm. It looks for an optimal solution to a problem in a solution space. The method of searching is a key feature of the GA (Meena and Devanshu, 2017). Genetic optimization algorithm is used in MATLAB either in code or GA optimization toolbox app.

The following are the fundamental steps in a genetic algorithm:

1. Start: Generate a random population of n chromosomes which have a proper form for the problem.
2. Run fitness function: Evaluate the fitness function using each chromosome in the population.
3. Result test: Judge the n evaluated results. If conditions are satisfied, then GA stops and outputs the best chromosome of current population.
4. Selection: Select two or more parent chromosomes from a population according to their fitness (the better the fitness of a chromosome, the greater is its chance of being selected).
5. Crossover: To create a new offspring, cross over the parents with a certain probability (children). If there was no crossing, the child is a perfect replica of its parents.

- Figure 4.6 shows, GA optimization of FOPID controller for cardiac pacemaker system. The GA optimizes the FOPID controller based on the upper and lower boundaries of tuning parameter. The lower boundaries of the tuning parameter are the starting values of the tuning parameter. In contrast, the upper boundaries of the tuning parameter are the ending value of the tuning parameter. For more understanding, the parameters of GAs such as maximum generation, population size, number of parameters, lower boundaries, and upper boundaries of the tuning parameters are given as shown in Table 4.2. The GA optimization technique is run according to upper and lower boundaries of the parameter that set on GA Matlab code. The GA optimization technique is written on Matlab editor. It has also population size and maximum generation value. GA optimizes the parameter that set on the Simulink model. The name of the Simulink model is assigned on GA Matlab code. The GA tuner tunes the parameter based on the Simulink model name. For more understanding, the Matlab code of GA optimization technique is mentioned in appendix [B].



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error (ISE), or integral of time-weighted-squared-error (ITSE) and integral of time absolute error (ITAE) (Luo et al., 2013).

$$IAE = \int_0^{\infty} |e(t)| dt \quad (4.5)$$

$$ISE = \int_0^{\infty} e^2(t) dt \quad (4.6)$$

$$ITSE = \int_0^{\infty} te^2(t) dt \quad (4.7)$$

$$ITAE = \int_0^{\infty} t|e(t)| dt \quad (4.8)$$

Where $e(t)$ is the error between the desired profile $y_r(t)$ and the actual one $y(t)$. Among the listed objective function above, in this thesis, ITAE is used since it results in small overshoot and oscillation in which minimized error between actual and reference pace rate is obtained.

The GA parameters that have been set to obtain the optimal values of the FOPID gains are presented in table 4.2.

Table 4.2: GA optimization parameter setting for FOPID controller

Parameter	Values
Population size	30
Maximum generation	50
Encoding	Binary
Selection	Uniform
Cross Over	Single Point
Mutation	Uniform
Number of parameters	5
Lower bound	[1 0.1 0.2 0.2 0.9]
Upper bound	[2 0.5 0.4 0.3 0.99]

Based on the above parameter setting, the step fitness curve of ITAE objective function for the cardiac pacemaker system is shown in Figure 4.7 below.

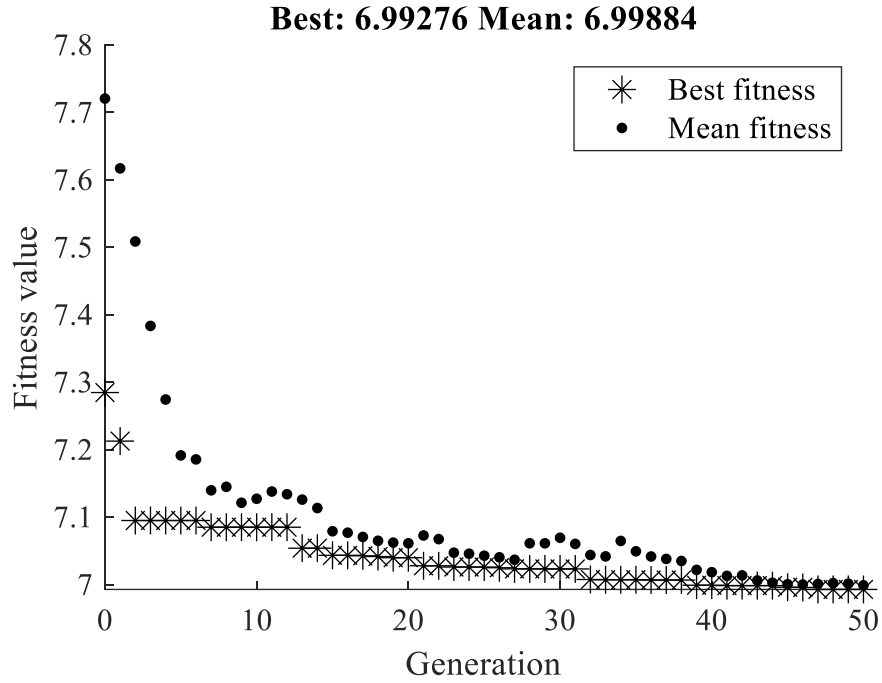


Figure 4.7: fitness vs generation of ITAE

The genetic algorithm is applied to the FOPID controllers. Firstly, GA is exported to MATLAB by coding. The proposed algorithms are tuned for 50 iterations to get the minimum value of the cost function (J). Table 4.3 shows the optimal values of FOPID parameters after tuning, corresponding to the minimum cost function value provided by the optimal GA.

Table 4.3: Parameter setting of GA tuned FOPID controller for regulation of heart rate

Controller	Optimized parameters and values					
GA tuned FOPID	Parameter	Kp	Ki	Kd	λ	μ
	Value	1.3088	0.1417	0.2500	0.2755	0.9900

4.4. Adaptive Neuro Fuzzy Inference System

A fuzzy-based controller produces a control signal that is produced by triggering a rule base written based on previous experience, which is then triggered to obtain the final output. As a result, the controller results are also random and sub-optimal. To avoid this, a suitable rule-based context-sensitive selection can be achieved using the ANFIS controller. This provides an integrated approach to control objectives with excellent results. The ANFIS scheme, designed for pacing rate control in pacemaker systems, uses neural network techniques to select an appropriate rule base. This is achieved using a hybrid algorithm. This

integrated approach improves system performance, cost-effectiveness, efficiency, dynamics, and reliability of the designed controller.

A model that maps input characteristics to input membership functions, input membership functions to rules, rules to a set of output characteristics, output characteristics to output membership functions, and the output membership functions to a single-valued output or a decision associated with the output is the fundamental building block of the Mamdani fuzzy inference system. Such a system makes use of arbitrarily selected fixed membership functions and rule structures that are fundamentally predetermined by the user's interpretation of the properties of the model's variables (Biniam Abera, 2020).

Neuro-fuzzy control combines the mapping and learning capabilities of artificial neural networks with the linguistic and fuzzy inference advantages of fuzzy logic. Fuzzy logic has the ability to self-modify the membership functions to achieve the desired performance. Therefore, neuro-fuzzy controllers may outperform traditional ANNs or fuzzy-logic controllers. The Adaptive Network-Based Fuzzy Inference System (ANFIS) uses the Takagi Sugeno Kang (TSK) type fuzzy inference system (Girma Kassa, 2019).

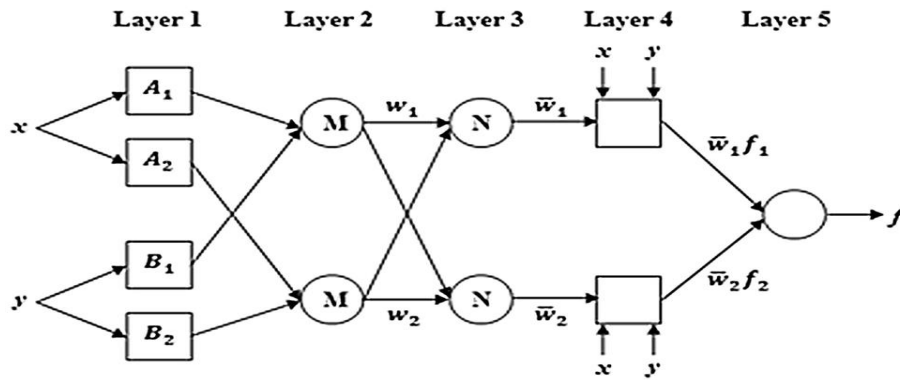


Figure 4.8: Equivalent ANFIS structure (Housny et al., 2020)

For simplicity, assumed that the fuzzy inference system under consideration has two inputs and one output. The rule base contains Takagi and Sugeno type fuzzy if-then rules such as: If x is A and y is B , then z is (x, y) , where A and B are the antecedent fuzzy sets and $z = (x, y)$ is the consequent crisp function. Usually (x, y) is a polynomial in the input variables x and y . However, other functions are available that can approximately describe the output of the system within the fuzzy domain specified by the assumptions. If (x, y) are constant, a 0th order Sugeno fuzzy model is formed. It can be viewed as a special case of the Mamdani fuzzy inference system (Mrinal, 2008) where each of the following rules is

specified by a fuzzy singleton: Assuming (x, y) to be a first-order polynomial, a first-order Sugeno fuzzy model is formed. For a Sugeno fuzzy inference system with two first-order rules, the two rules can be specified as:

Rule 1: if x is A_1 and y is B_1 , then $f_1 = p_1x + q_1y + r_1$

Rule 2: If x is A_2 and y is B_2 , then $f_2 = p_2x + q_2y + r_2$

Here, a type 3 fuzzy inference system proposed by Takagi and Sugeno is used. In this inference system, the output of each rule is a linear combination of the input variables added with an additional constant term. The final output is a weighted average of each rule's output. The individual layers of this ANFIS structure are described below.

Input Node (Layer 1): Nodes in this layer contains membership functions. Every node i in this layer is square and adaptive with a node function:

$$O_1^i = \mu_{A_i}(x), \text{ for } i=1,2 \text{ or} \quad (4.9)$$

$$O_1^i = \mu_{B_{i-2}}(y), \text{ for } i=3,4 \quad (4.10)$$

where, x is the input to node i , A_i is the linguistic variable associated with this node function and μ_{A_i} is the membership function of A_i . Similarly, y is the input to node i , B_{i-2} is the linguistic variable associated with this node function and $\mu_{B_{i-2}}$ is the membership function of B_{i-2} . Here the membership function can be any appropriate parameterized membership function, such as the generalized triangular function:

$$\mu_{A_i}(x) = \max \left(\min \left(\frac{x-a_i}{b_i-a_i}, \frac{c_i-x}{c_i-b_i} \right), 0 \right), \quad i=1,2. \quad (4.11)$$

Where x is the input and $\{a_i, b_i, c_i\}$ are parameter sets. Parameters in this layer are called premise parameters.

Rule Nodes (Layer 2): Every node in this layer is fixed and circle node labeled M, whose output is the product of all incoming signals:

$$O_i^2 = w_i = \mu_{A_i}(x) * \mu_{B_i}(y), \quad i=1,2. \quad (4.12)$$

Each node's output represents the firing strength of a rule. This layer chooses the minimum value of two input weights. Any other T-norm operators that perform fuzzy AND (e.g., min) can be used as the node function in this layer.

Layer 3: Every node in this layer is a fixed node. Each i th node calculates the ratio of the i th rule's firing strength to the sum of firing strengths of all the rules.

The output from the i th node is the normalized firing strength given by,

$$o_i^3 = \bar{w}_i = \frac{w_i}{w_1 + w_2}, \quad i = 1, 2 \quad (4.13)$$

For convenience, outputs of this layer are called normalized firing strengths.

Layer 4: Every node i in this layer is an adaptive node with a node function

$$o_i^4 = \bar{w}_i f_i = \bar{w}_i (p_i x + q_i y + r_i) \quad (4.14)$$

Where w_i is a normalized firing strength from layer 3 and $\{p_i, q_i, r_i\}$ are the parameter sets of this node. Parameters in this layer are referred to as consequent parameters.

Layer 5: The single node in this layer is fixed node labeled Σ , which computes the overall output as the summation of all incoming signals:

$$\text{overall output} = o_i^5 = \sum_i \bar{w}_i f_i = \sum_i \frac{w_i f_i}{w_i} \quad (4.15)$$

4.4.1. Design of an ANFIS for Cardiac Pacemaker System

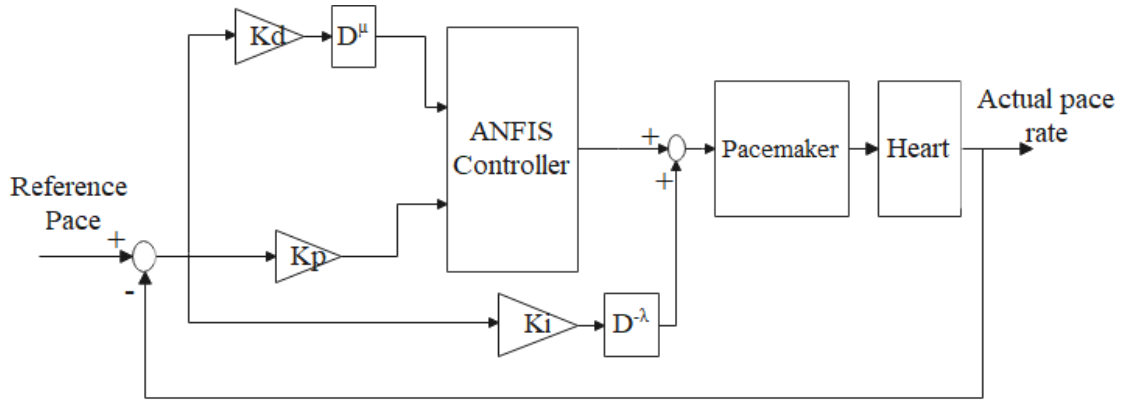


Figure 4.9: The proposed ANFIS based FOPID controller for heart rate regulation

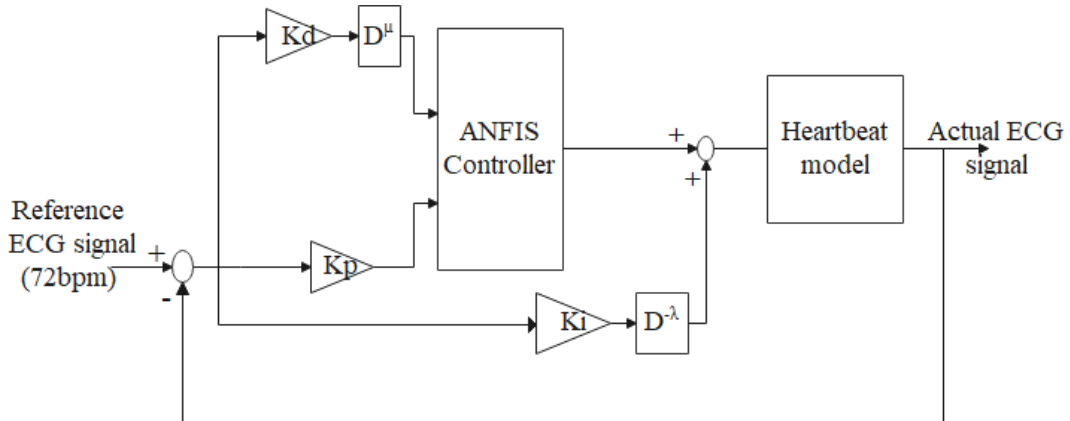


Figure 4.10: The proposed ANFIS based FOPID controller for ECG signal tracking

4.4.2. Training of ANFIS Controller Using Data from the FOPID Controller

ANFIS-based modeling combines the transparent linguistic representation of fuzzy systems with the learning ability of neural networks so that they can be trained to perform input/output mapping. When designing, we cannot tell what your membership functions should be just by looking at the data. ANFIS allows automatic tuning of parameters. Membership functions capture the dynamics of data. The set of rules provided by ANFIS provides information about the underlying system and is therefore invaluable for further insight into the process model. Starting with FIS as a controller, there is a special requirement that the membership function parameters should be tuned to ensure the best performance of the system.

The system pace error and fractional order derivative rate of pace error are inputs and the command signal is the output of the neuro-fuzzy controller. Sugeno fuzzy model with five-layer ANN structure is used in the proposed controller. The following figure below shows the input/output of the ANFIS controller.

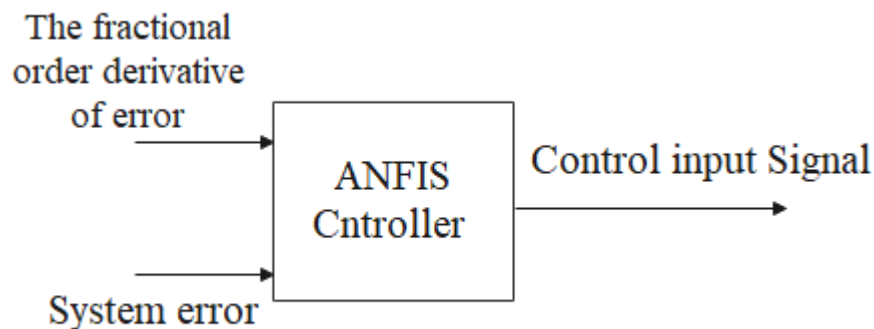


Figure 4.11: ANFIS controller input/output

In order to generate an adaptive neuro-fuzzy system, GA tuned FOPID controller is designed and simulated using MATLAB®/2021a to collect training and checking data for 10 seconds. After collecting the input/output data, the data is divided into training and checking data sets randomly. 200,050 data points for heart rate regulation and 8539 data pairs for ECG signal tracking are collected and out of these data, 80% of total data have been considered as training data sets and the remaining 20% of total data have been considered as checking data. Since ANFIS involves the combined properties of neural network and fuzzy logic, it provides a method for the fuzzy modeling procedure to learn information about a data set, and to compute the membership function parameters that allow the associated fuzzy inference system (FIS) to track the given input/output data. A network type structure resembling a

neural network that maps inputs through input membership functions and related parameters and output membership functions and associated parameters to outputs might be taken into consideration in order to create the FIS structure. During the learning process, the parameters related to the membership functions will alter. The computation of these parameters is done by a hybrid learning algorithm. In this work, the number of membership functions considered during the development of ANFIS is seven, and the type of input membership functions used are generalized triangular functions. The linear membership function type is selected for output, a total of 49 rules are generated, and the number of generations considered is 50.

The performances of ANFIS models of both training and checking data were evaluated and the best training/checking data set was selected according to RMSE. To find the ideal ANFIS system, the system trained with a variety of settings for items such as data set sample, epoch number, membership function type, and number, and number of inputs to achieve the best performance. The MATLAB toolbox function ANFIS uses specified input/output data sets to build a fuzzy inference system (FIS) whose membership function parameters are tuned using a hybrid learning algorithm method. The data were split into two separate sets, a training set, and a checking set. ANFIS was trained using the training data set, and the accuracy and effectiveness of the trained ANFIS model for adapting the learning content were validated using the checking data set. Two pair data sets were made with different combinations of 80% and 20% of the samples to improve the generalization properties of the adopted ANFIS. The training and checking data pairs were gathered from the simulation result of the block diagram shown in Figure 3.2 using a fractional order PID controller. After collecting the input/output data, the data is divided into training and checking data sets randomly. Once the input/output data pairs have been collected, the first step was loading training and checking data pairs to the ANFIS GUI of MATLAB toolbox, then initializing the FIS was the second step (the initialization step includes the number and type of input and output membership functions and also several epoch numbers). After initializing FIS and specifying the number of epochs, the next step was to train the training data pairs until the minimum training and checking errors are recorded. If the errors are acceptable, the final step was testing the training data using checking data to validate the model. After loading the training data and checking data sets in the neuro-fuzzy designer, the following results have been observed. The training and checking data sets for tracking ECG signals and heart rate tracking and regulating after loading in the ANFIS editor toolbox are shown in the appendix [C].

Figure 4.12 shows the step input training and checking data sets after loading in the ANFIS editor toolbox for tracking and regulation of heart rate.

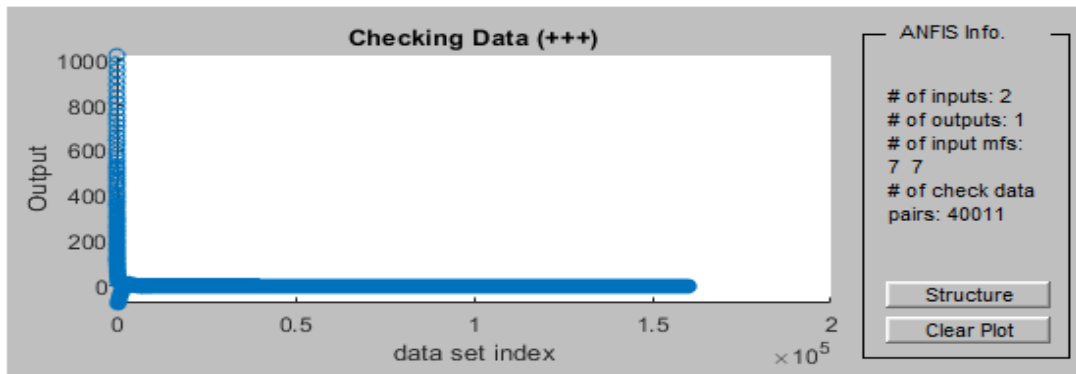


Figure 4.12: Training and checking data sets of step input for tracking and regulation of heart rate

Checking data is used for testing the generalization capability of the fuzzy inference system at each epoch. The checking data has the same format as that of the training data. This data set is used to validate the fuzzy inference model. This validation is done by applying the checking data to the model and then seeing how well the model responds to this data. When the checking the data option is used using the ANFIS Editor GUI, the checking data is applied to the model at each training epoch. The FIS membership function parameters are computed using the ANFIS editor GUI when both training and checking data are loaded. The checking data error diminishes as training progresses because the checking data and training data are sufficiently close. The checking error is the difference between the output of the fuzzy inference system that corresponds to the same checking data input value as the checking data output value and the output that is associated with that checking data output value. The RMSE for the checking data at each epoch is recorded in the checking error. As the system is trained, the ANFIS Editor GUI plots the checking error against epochs curve. Tracking ECG signals, tracking and regulation of different heart rates minimal training RMSE, and minimal checking RMSE are shown in the appendix [C]. The tracking and regulation of heart rate minimal training RMSE is 0.105702 and minimal checking RMSE is 0.0733346. When checking data is tested against the trained FIS, it looks satisfactory. The tracking and regulation of heart rate training and checking errors are shown in Figure 4.13.

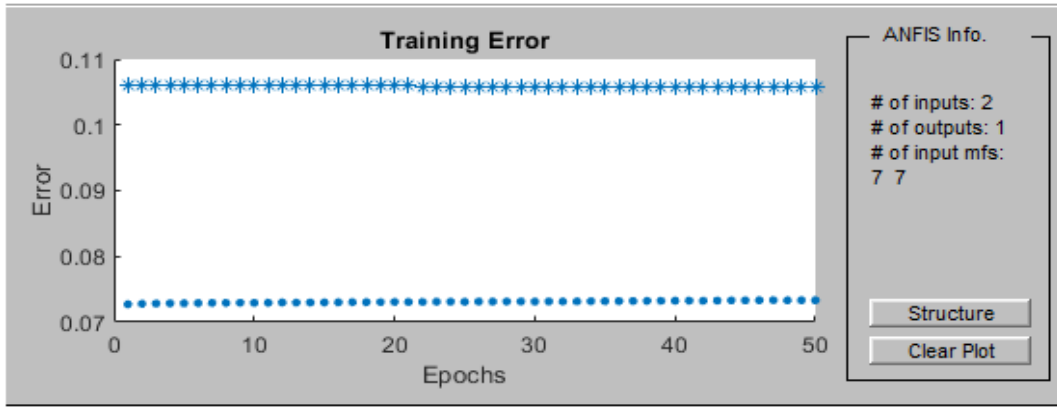


Figure 4.13: Training and checking errors

As shown in Figure 4.13 the training and checking errors as the training starts and became about number of epochs reaches 50 is almost constant. The RMS errors are negligibly small, and hence the ANFIS captured the essential dynamics of the system. Figure 4.13 also shows how much the training and checking data are related and they are superimposed on each other. The following figure shows the structure of the proposed ANFIS controller.

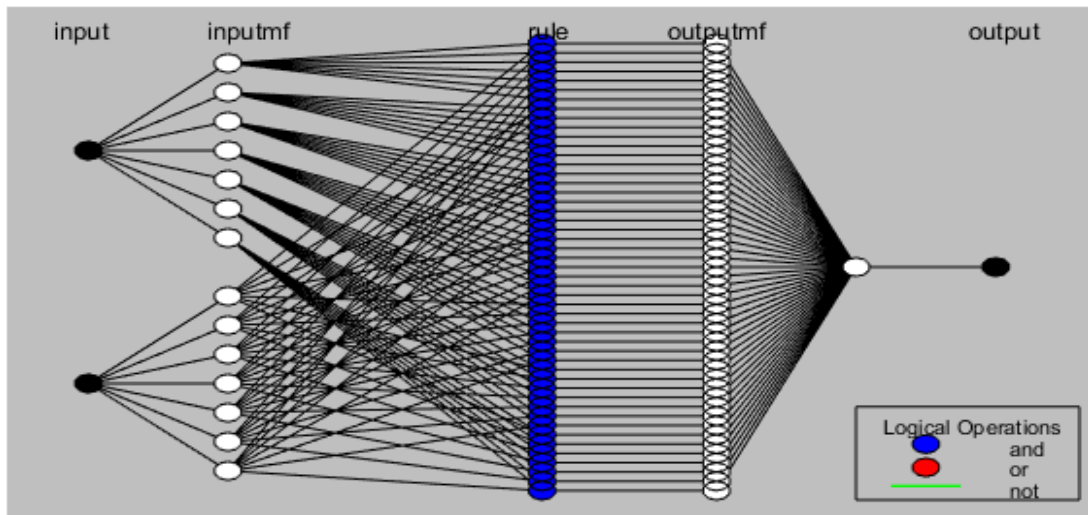


Figure 4.14: Schematic of the proposed ANFIS controller

From the ANFIS structure shown in Figure 4.14, it has been observed that when the values of the premise parameters are fixed, the overall output can be expressed as a linear combination of the consequent parameters.

Table 4.4: Specifications of the developed adaptive neuro-fuzzy inference system

Parameters	Description/Value	Parameters	Description/Value
Optimization method	Hybrid	Number of outputs	1
Structure of FIS	Sugeno first order	Output MF type	Linear
Or method	Max	No. of rules	49
And method	Min	No. of epochs /iterations	50
Implication method	Prod	No. of training data pairs	80% of total generated data
Aggregation method	Max	No. of checking data pairs	20% of total generated data
No. of inputs	2	No. linear parameters	147
No. of input MF	7	No. of nonlinear parameters	42
Type of input MF	Triangular	Total No. of parameters	189
No. of output MF	49	No. of nodes	131

CHAPTER FIVE

5. SIMULATION RESULTS AND DISCUSSIONS

5.1. Simulink Modeling

Simulink Modeling MATLAB/Simulink is a software package that is used to model, simulate, and analyze dynamic systems. Simulink has the advantage of being capable of complex dynamic system simulations, a graphical environment with visual real-time programming, and a broad selection of toolboxes.

5.1.1. Simulink Model for Tracking of ECG Signal

The simulation is carried out using MATLAB/Simulink. The validity of the heartbeat model was analyzed in Chapter 3. The overall control system is shown in Figure 5.1. The second order of the Zeeman heartbeat model has been used to perform a simulation of the proposed controller to track ECG data. The tracking comparison performance of the FOPID and ANFIS-based FOPID in terms of eliminating chattering was tested. In addition, the robustness of the proposed controller was tested on disturbance rejection.

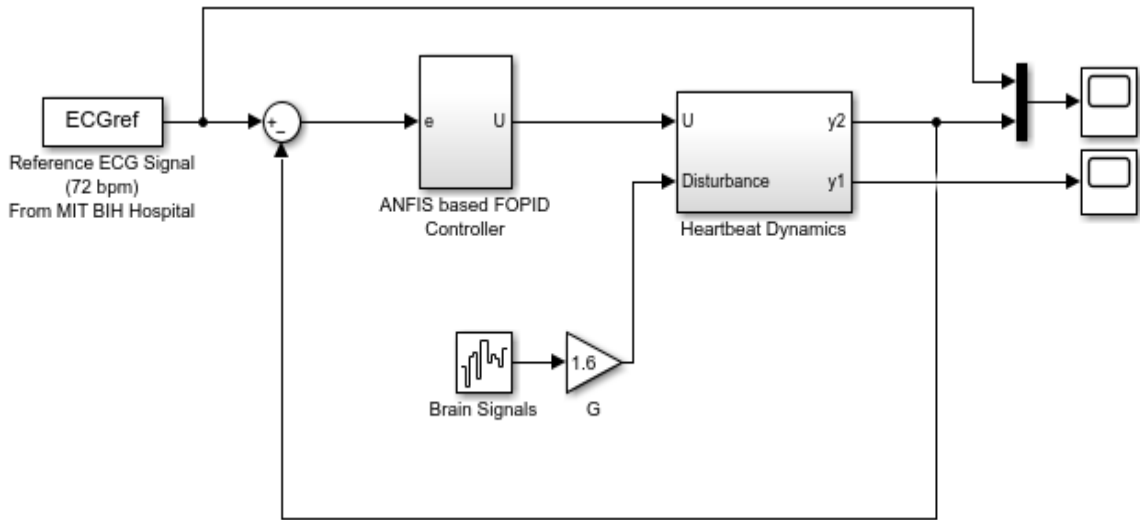


Figure 5.1: Simulink realization of ANFIS-based FOPID controller for Zeeman heartbeat model

In this thesis, the parameters of the FOPID controller for tracking ECG signals were tuned manually by trial and error method. The average healthy heartbeat for adult humans (72bpm) obtained from the PhysioNet database was used as a reference signal and the simulation was explained in chapter three.

The disturbance of electrical stimulation of the heart with noise (like anxiety, or disease) coming from the human brain was considered system disturbance.

Table 5.1: Values of the brain frequency (Aabid et al., 2016)

Value	
Wave Form	Random
Amplitude	20
Brain Frequency (Anxiety, disease)	100Hz

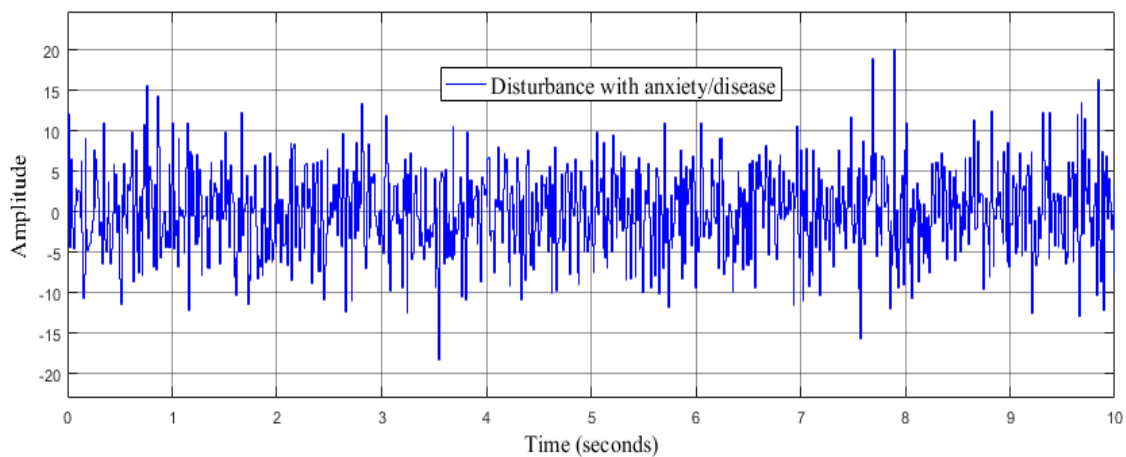


Figure 5.2: Disturbance signal

5.1.2. Simulink Model for Human Heart Rate Tracking and Regulation

The mathematical modeling of the human heart using the transfer function presented in Chapter 3 is used to model ANFIS-based FOPID controller with cardiac pacemaker system in MATLAB/Simulink R2016a environment. Figure 5.3 shows the complete Simulink model for heart rate regulation and tracking using a cardiac pacemaker system. Once the block diagram has been developed in MATLAB/Simulink it can be simulated using any number of different solvers. These solvers can compute the internal state variables of the blocks by solving their respective ordinary differential equations in MATLAB modeling configuration parameters. The solver is significant to decrease the computation time and improve the accuracy of the simulation.

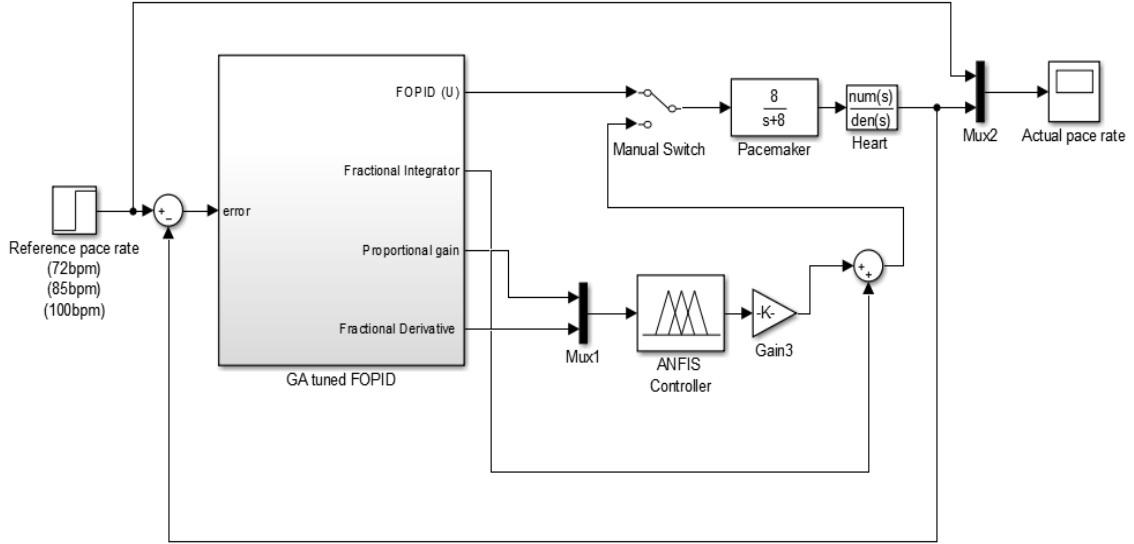


Figure 5.3: Simulink realization of ANFIS-based FOPID controller for heart rate regulation

5.2. The Results

5.2.1. Performance Comparison of the System to Track ECG Signals with FOPID and ANFIS-based FOPID Controller

5.2.1.2. ECG Signal Tracking Using FOPID Controller without Disturbance

Table 5.2: Parameter setting of FOPID controller for ECG signal tracking

Parameter	KP	Ki	Kd	λ	μ
Value	3.357	8.457	350.965	0.965	0.95

The simulation result with the FOPID controller in the Figure 5.4 indicated that the output measurement $y_2(t)$ tracks the ECG signal from the physio net database without any external disturbance applied to the system. The Figure 5.5 showed that there is too much chattering in pacemaker signal $u(t)$. Based on the above parameter setting, the response of the system is obtained as:

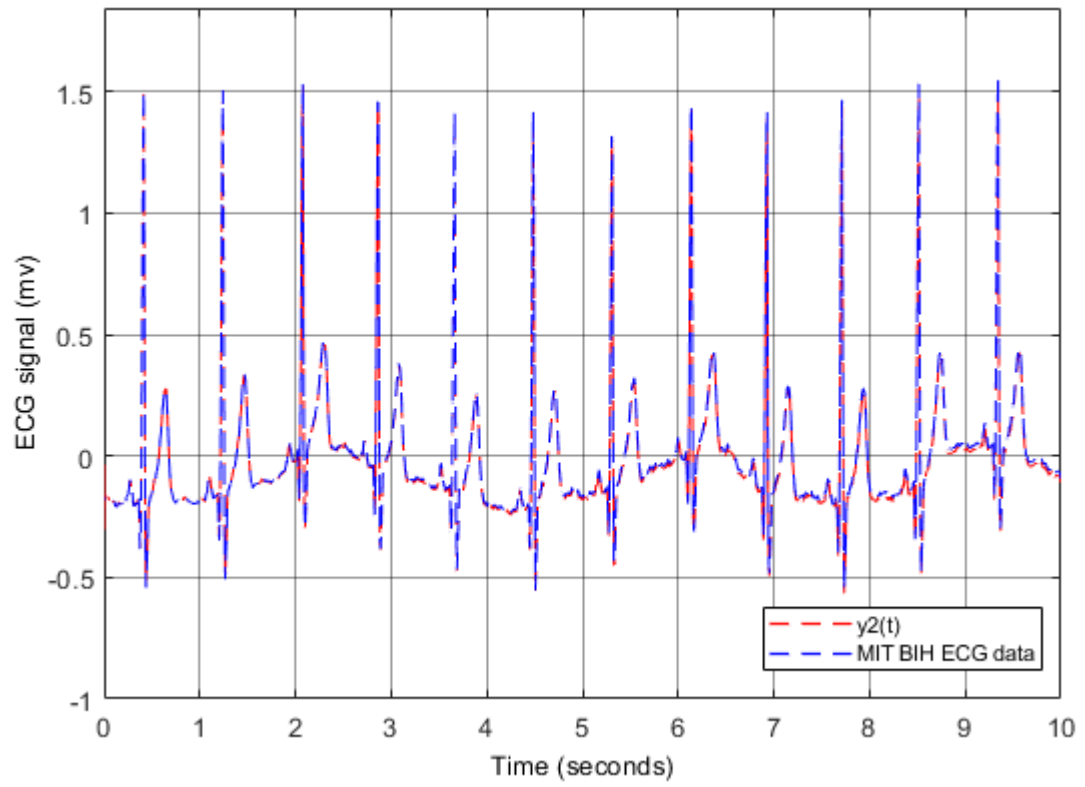


Figure 5.4: ECG signal tracking of FOPID controller for heartbeat system without disturbance

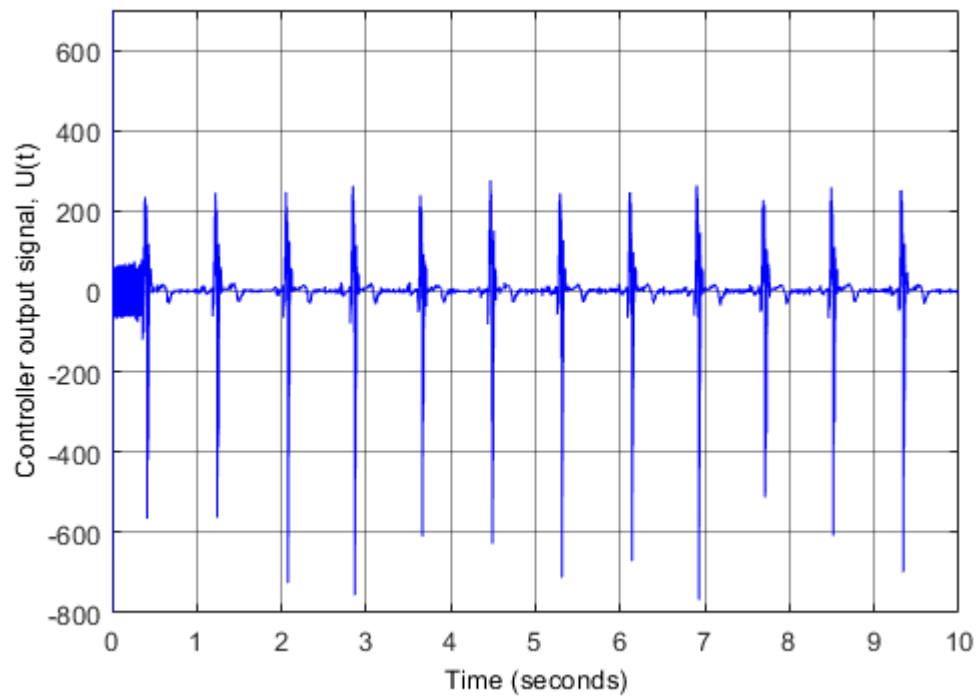


Figure 5.5: Control signal for heartbeat system with FOPID controller

5.2.1.2. ECG Signal Tracking Using ANFIS Based FOPID Controller without Disturbance

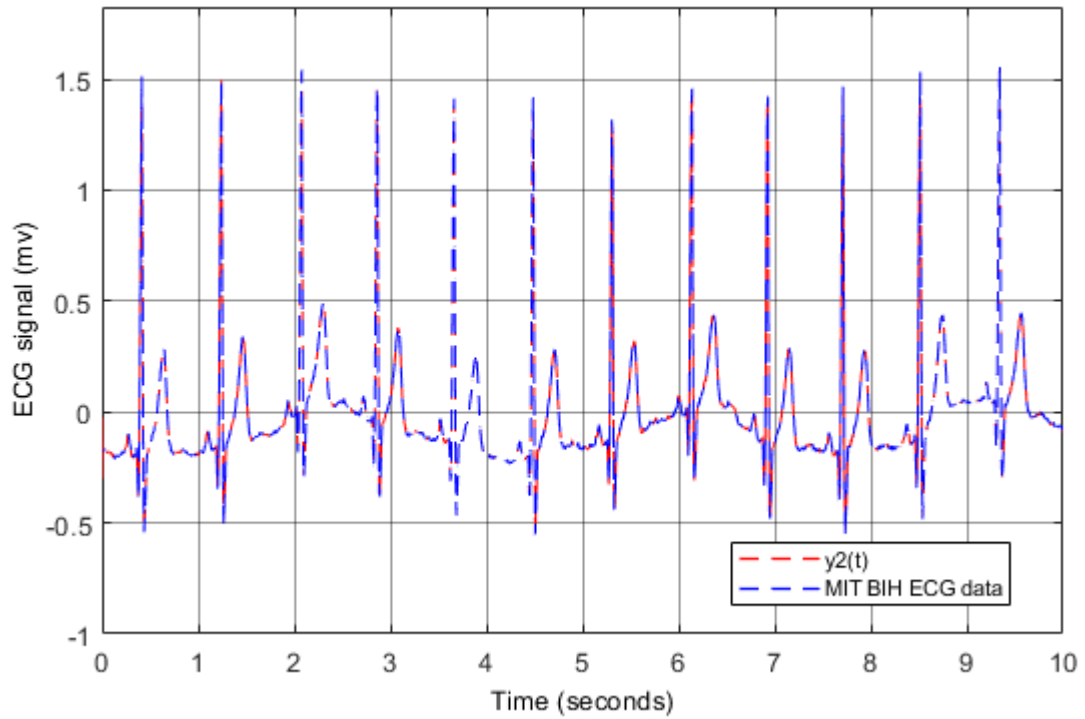


Figure 5.6: ECG signal tracking of ANFIS FOPID controller for heartbeat system without disturbance

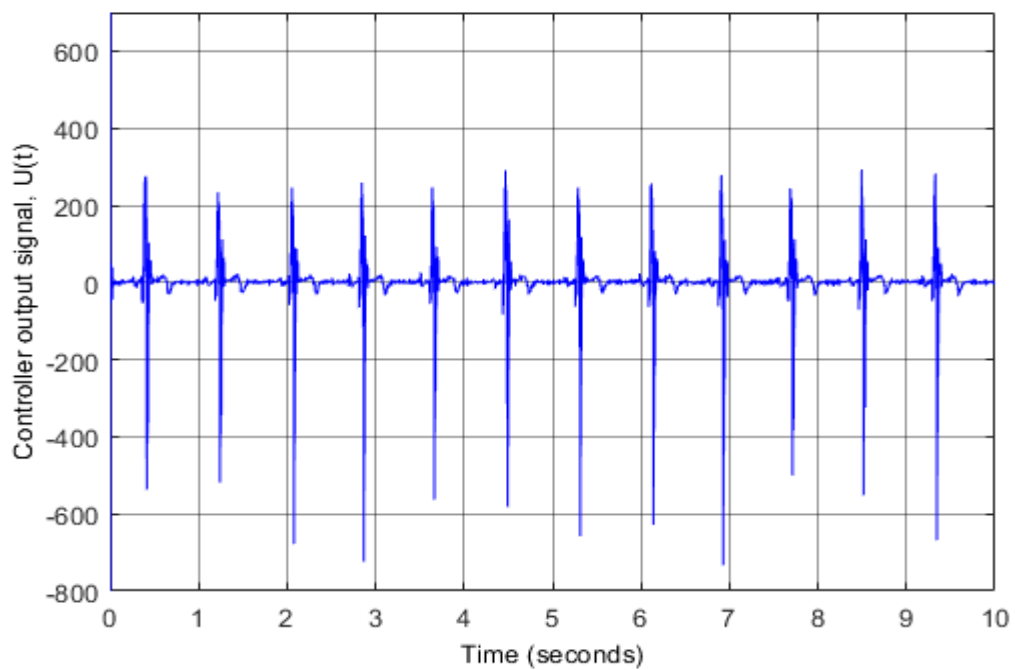


Figure 5.7: Control signal for heartbeat system with ANFIS-based FOPID controller

The result in the figure shows the tracking performance of the ANFIS-based FOPID controller. The simulation result shows that the ANFIS based FOPID controller is superior to the FOPID controller in eliminating the chattering phenomenon which appeared in the pacemaker signal as it is shown in the Figure 5.7.

Table 5.3: Comparison of controller performance without disturbance

Performance comparison criteria	Controller algorithms	
	FOPID controller	ANFIS-based FOPID controller
Mean square error	0.000271	0.0000946
Root mean square error	0.01646	0.00973
ITAE	0.7029	0.0527

5.2.1.3. ECG Signal Tracking Using FOPID Controller with Disturbance

The simulation result with FOPID shown in the Figure 5.8 indicates that the output measurement $y_2(t)$ tracks the ECG signal from the Physio Net database and rejects external disturbance.

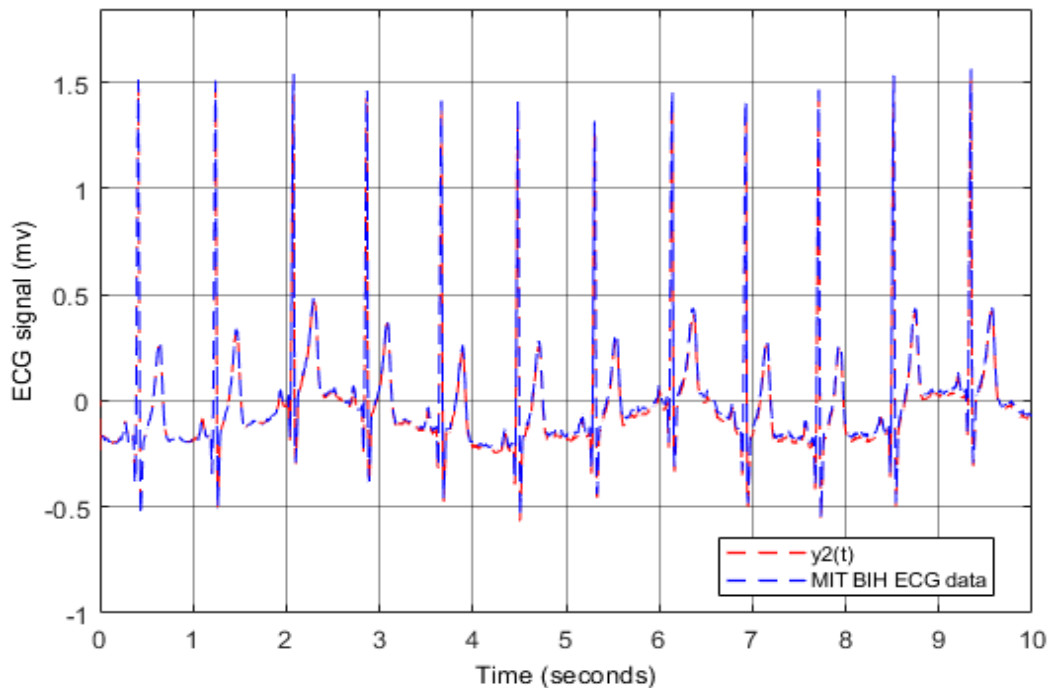


Figure 5.8: ECG signal tracking of FOPID for heartbeat system with disturbance

5.2.1.4. ECG Signal Tracking Using ANFIS Based FOPID Controller with Disturbance

The result in Figure 5.9 shows the tracking performance of the ANFIS-based fractional order PID controller. It rejects external disturbance and also shows that the ANFIS-based FOPID controller is superior to FOPID in eliminating the chattering phenomenon which appeared in the pacemaker signal.

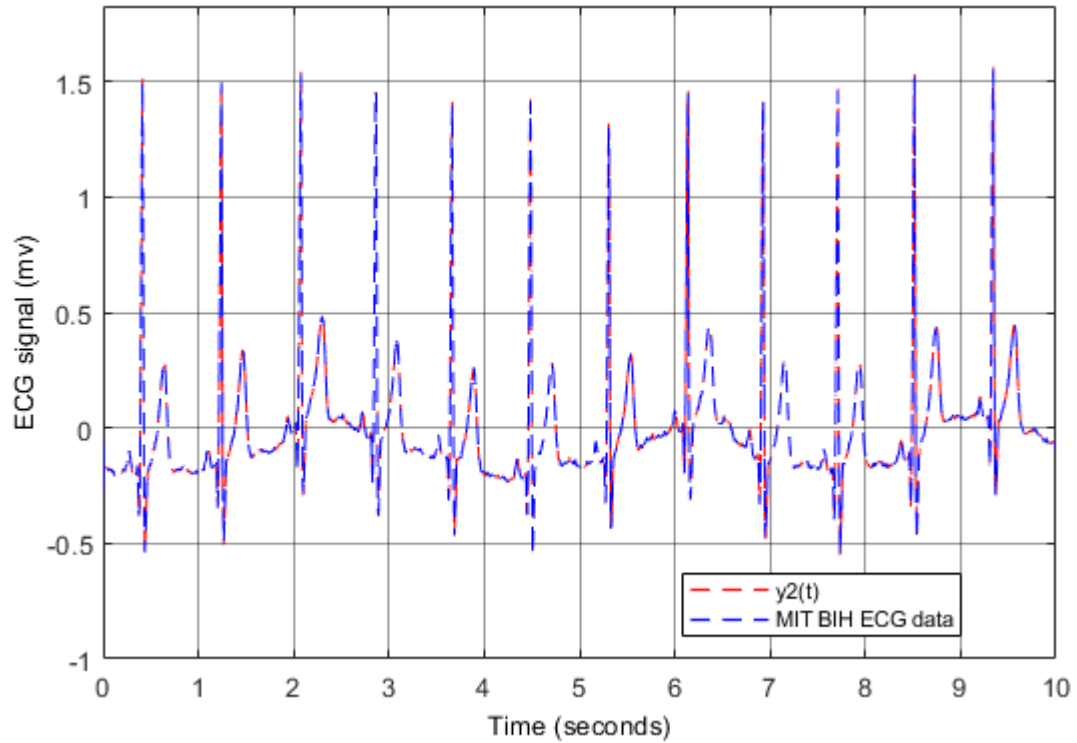


Figure 5.9: ECG signal tracking of ANFIS-based FOPID for heartbeat system with disturbance

Table 5.4: Comparison of controller performance with disturbance

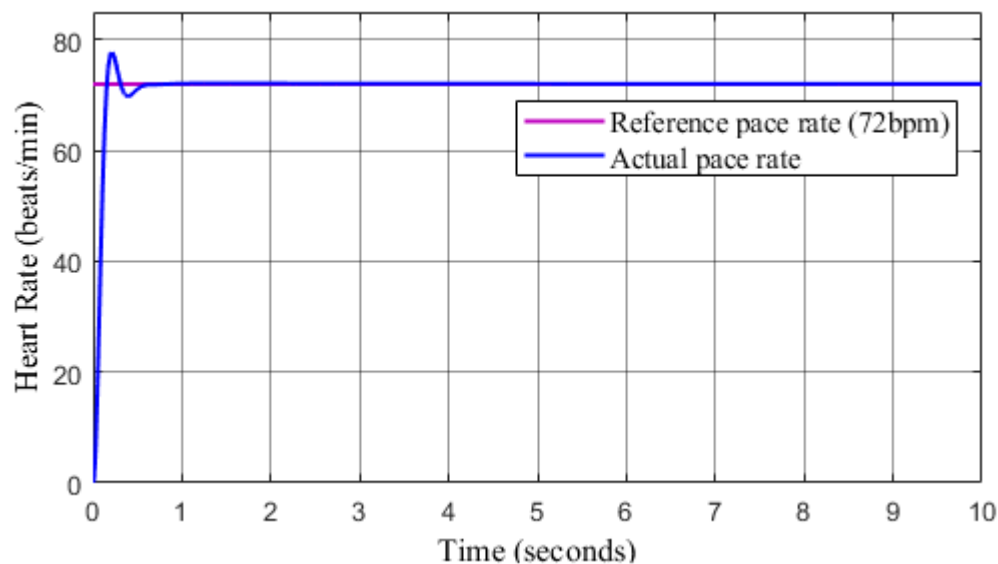
Performance comparison criteria	Controller algorithms	
	FOPID controller	ANFIS-based FOPID controller
Mean square error	0.000421	0.00011
Root mean square error	0.0204	0.010488
ITAE	1.014	0.06918

5.2.2. FOPID Controller for Heart Rate Tracking and Regulation

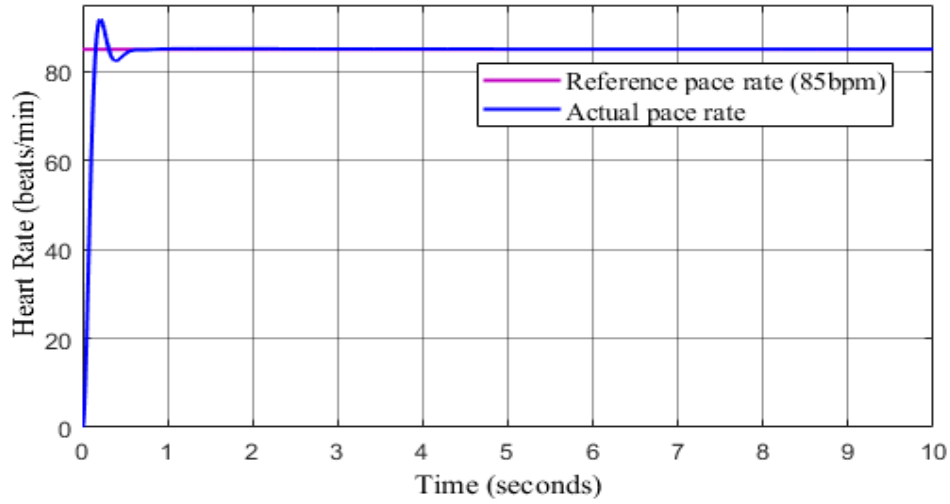
Initially, the FOPID controller is tuned with the help of the FOMCON toolbox. The response of FOPID controller tuned with FOMCON toolbox for different cases are shown in Figure 5.10 (a-c) respectively. Three different cases of HR are taken into consideration separately as reference inputs. The performance parameter values for HR=72, HR=85, and HR=100 using FOPID controller are depicted in Table 5.5. It is observed from Figure 5.10 (a-c) that overshoot and settling times are very high by using the FOPID controller tuned with the FOMCON toolbox. The oscillations are also observed in the response of the process. Therefore, to check the feasibility of getting improved performance genetic algorithm is used for tuning of the FOPID controller. Based on the parameter setting in Table 4.1 the response of the system is obtained as:

Table 5.5: Response parameter of FOPID controller

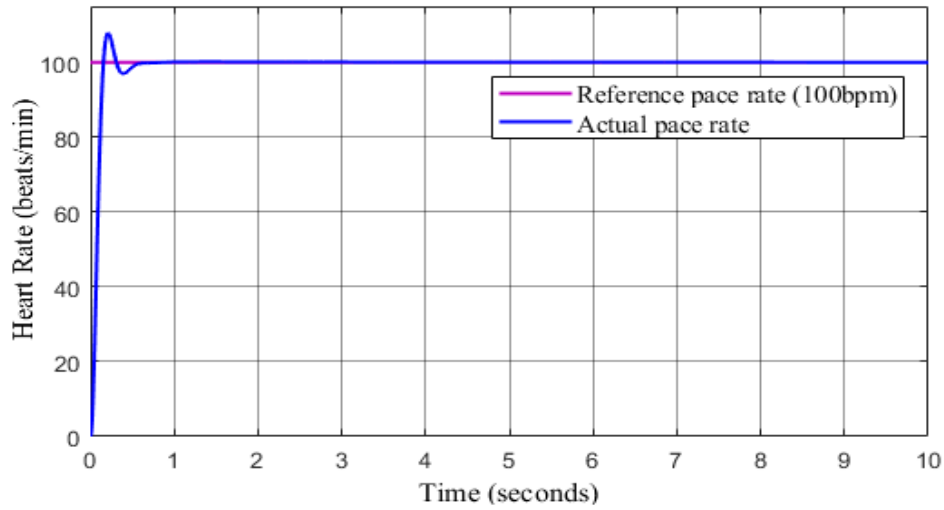
Heart Rate (HR)	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
72bpm	0.1020	0.4654	7.8883
85bpm	0.1020	0.4655	7.8864
100bpm	0.1117	0.5195	7.9317



(a)



(b)

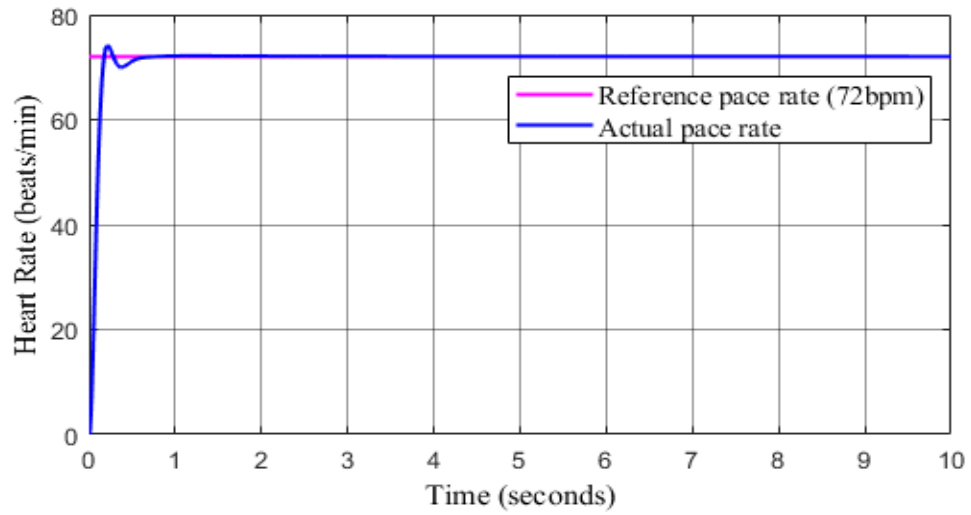


(c)

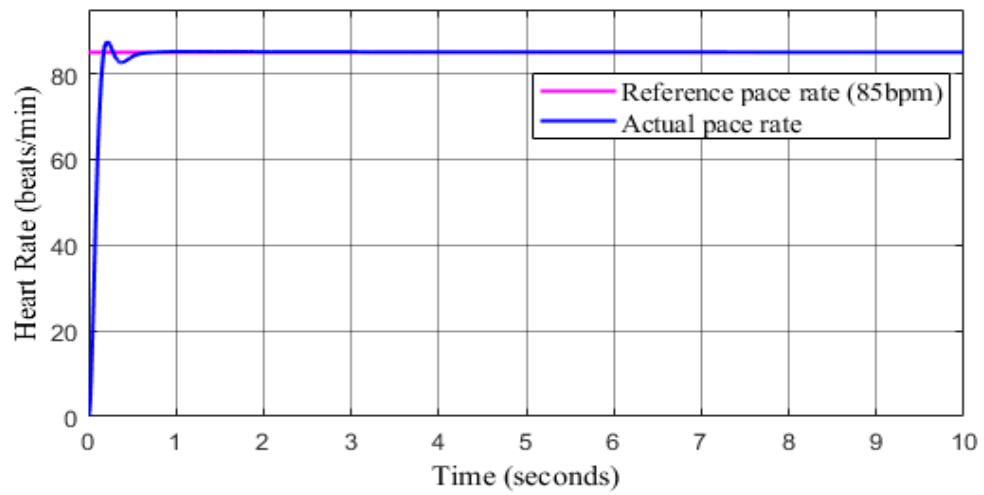
Figure 5.10: Response of the FOPID controller for (a) HR=72, (b)HR=85 and (c) HR=100

5.2.3. GA Tuned FOPID Controller for Heart Rate Tracking and Regulation

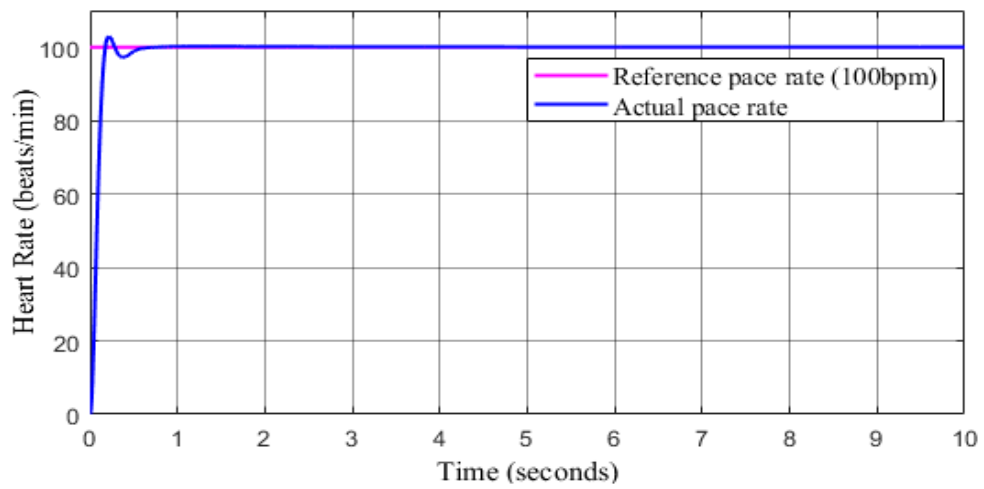
The response of FOPID controller tuned with GA for different cases are shown in Figure 5.11 (a-c) respectively. The performance parameters of GA tuned FOPID controller are depicted in Table 5.6. It is observed from Figure 5.11 (a-c) that settling time and overshoot is significantly less as compared to the response of the FOPID controller tuned with the FOMCON toolbox. Still, the response needs to be improved and to check the feasibility of getting desired results. Therefore, to improve the overall performance of the FOPID controller a robust system is needed. Based on the parameter setting in Table 4.3 the response of the system is obtained as:



(a)



(b)



(c)

Figure 5.11: Response of the GA tuned FOPID controller for (a) HR=72, (b)HR=85 and (c) HR=100

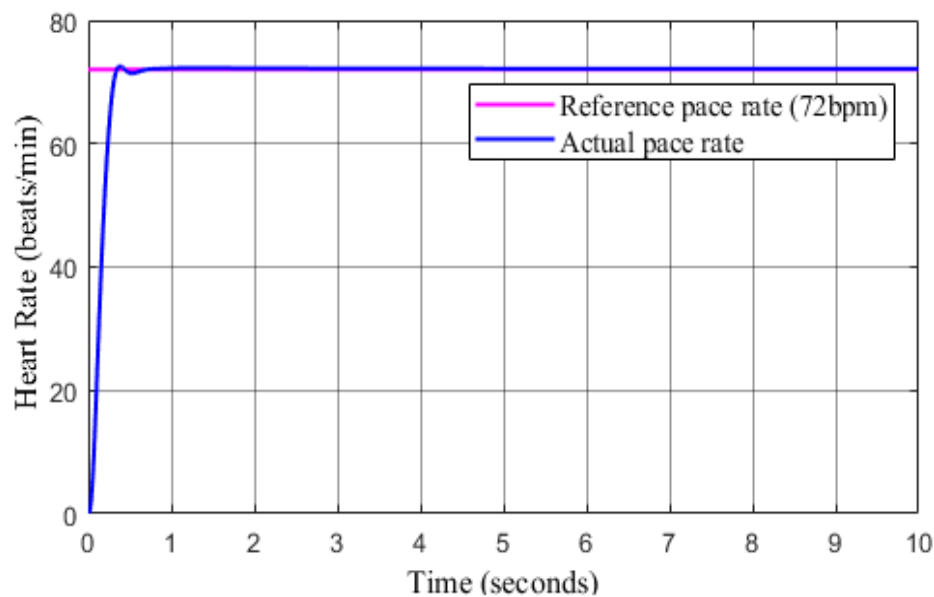
Generally, ANFIS based systems give better system response. Some of the studies using ANFIS in engineering applications have proved that, nonlinear ANFIS controller gives better response than conventional controller and local stability can also be achieved using ANFIS based controller. Therefore, to improve the performance of the PID controller fuzzy based control system is designed.

Table 5.6: Response parameter of GA tuned FOPID controller

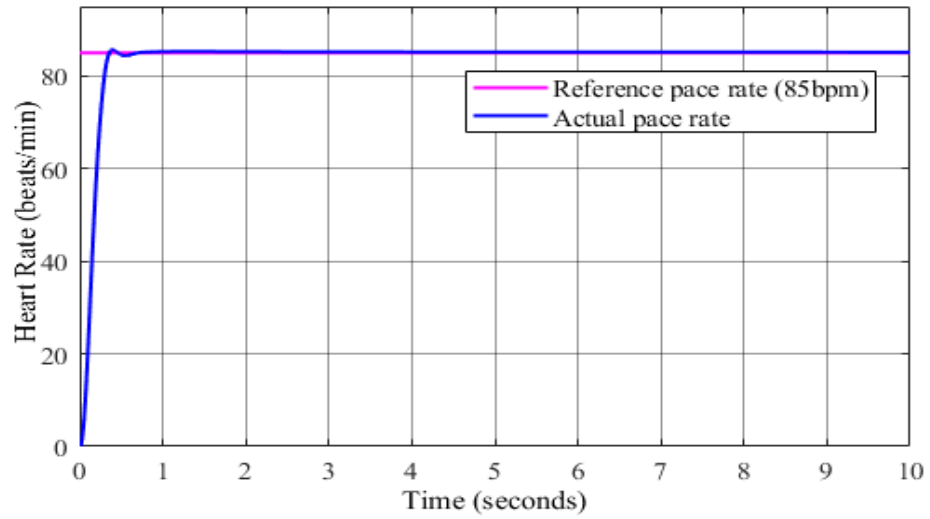
Heart Rate (HR)	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
72bpm	0.1103	0.4503	2.9261
85bpm	0.1103	0.4504	2.9197
100bpm	0.1103	0.4503	2.9257

5.2.4. ANFIS-based FOPID Controller for Heart Rate Tracking and Regulation

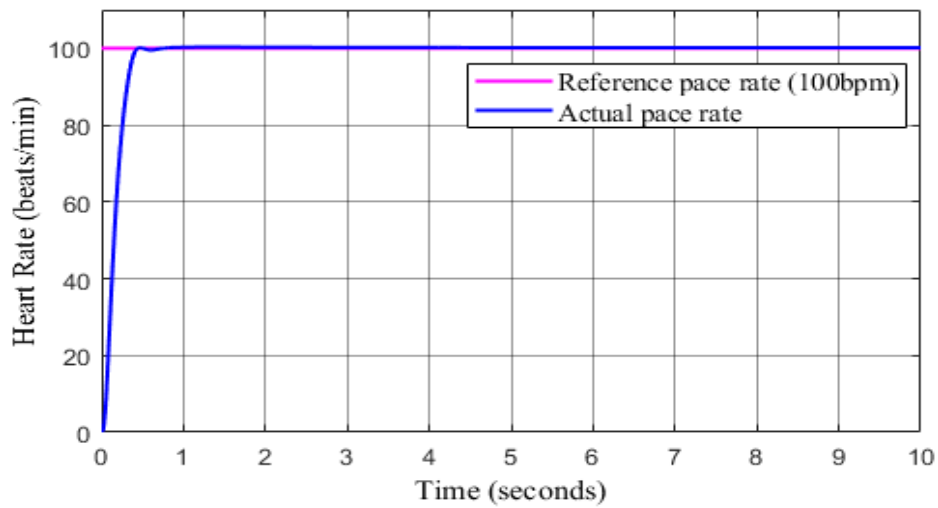
The responses of ANFIS based FOPID controller for different cases of HR are shown in Figure 5.12 (a-c) respectively. The performance parameters of ANFIS based controller are depicted in Table 5.7. It is also observed from Figure 5.12(a-c) and Table 5.7 that the settling time and overshoots are comparatively less as compared to FOPID tuned with FOMCON toolbox and GA methods and overall performance of fuzzy controller is satisfactory.



(a)



(b)



(c)

Figure 5.12: Response of the ANFIS based FOPID controller for (a) HR=72, (b)HR=85 and (c) HR=100

Table 5.7: Response parameter of GA tuned FOPID controller tuned with FOMCON toolbox

Heart Rate (HR)	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
72bpm	0.2047	0.3168	0.5830
85bpm	0.2151	0.3297	0.6522
100bpm	0.2551	0.3893	0.1760

5.2.5. Comparison of FOPID, GA tuned FOPID and ANFIS Based FOPID Controller Results

The result obtained by FOPID controllers tuned with FOMCON toolbox and genetic algorithm methods and ANFIS based FOPID controller is depicted in Table 5.5, 5.6 and 5.7 for different cases of HR. It is observed from Figure 5.10-5.12 and Tables 5.5-5.7 that rise time of FOPID controller tuned FOMCON toolbox is less as compared to response of FOPID controller tuned with GA and ANFIS based FOPID controller method. It is observed in Figure 5.10-5.12 and also in Tables 5.5-5.7 that the settling time and overshoot of ANFIS-based FOPID controller significantly lesser than the response of FOPID controller tuned with the FOMCON toolbox and GA method. The overall performance of the ANFIS-based FOPID controller is more promising than that of the FOPID controller tuned with the FOMCON toolbox and GA method performance.

Table 5.8: Performance measures of the FOPID, GA tuned FOPID and ANFIS-based FOPID controller for 72bpm heart rate

Controller	72 beats per minute heart rate		
	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
FOPID	0.1020	0.4654	7.8883
GA tuned FOPID	0.1103	0.4503	2.9261
ANFIS based FOPID	0.2047	0.3168	0.5830

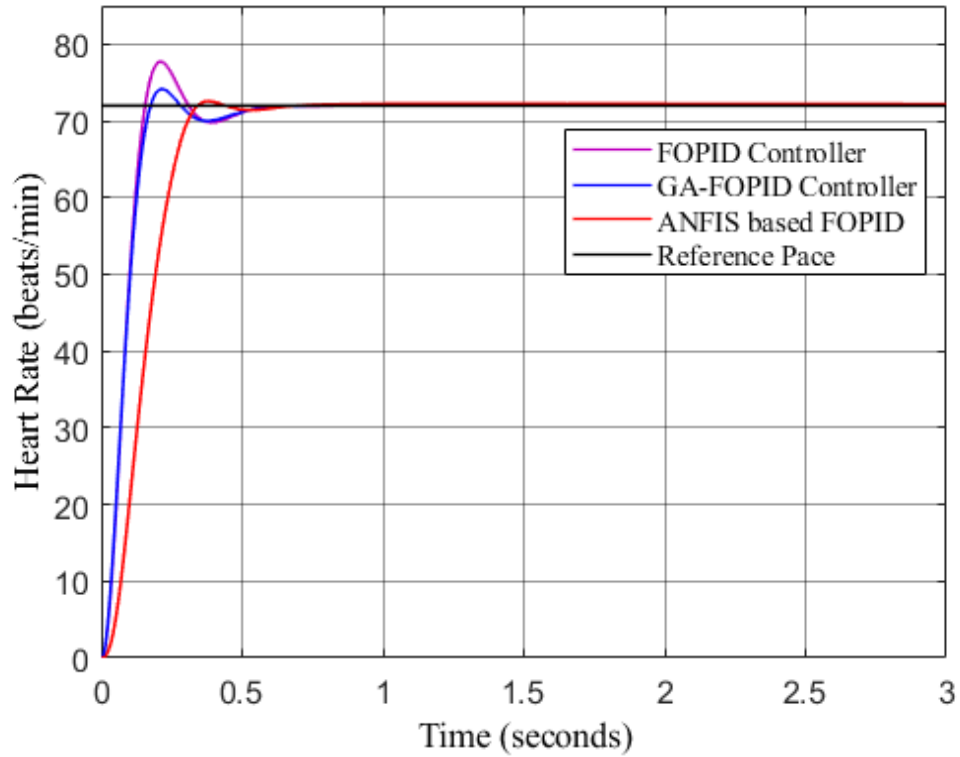


Figure 5.13: Comparison of result obtained by FOPID, GA tuned FOPID and ANFIS based FOPID controller HR=72 bpm

Table 5.9: Performance measures of the FOPID, GA tuned FOPID, and ANFIS-based FOPID controller for 85bpm heart rate

Controller	85 beats per minute heart rate		
	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
FOPID	0.1020	0.4655	7.8864
GA tuned FOPID	0.1103	0.4504	2.9197
ANFIS based FOPID	0.2151	0.3297	0.6522

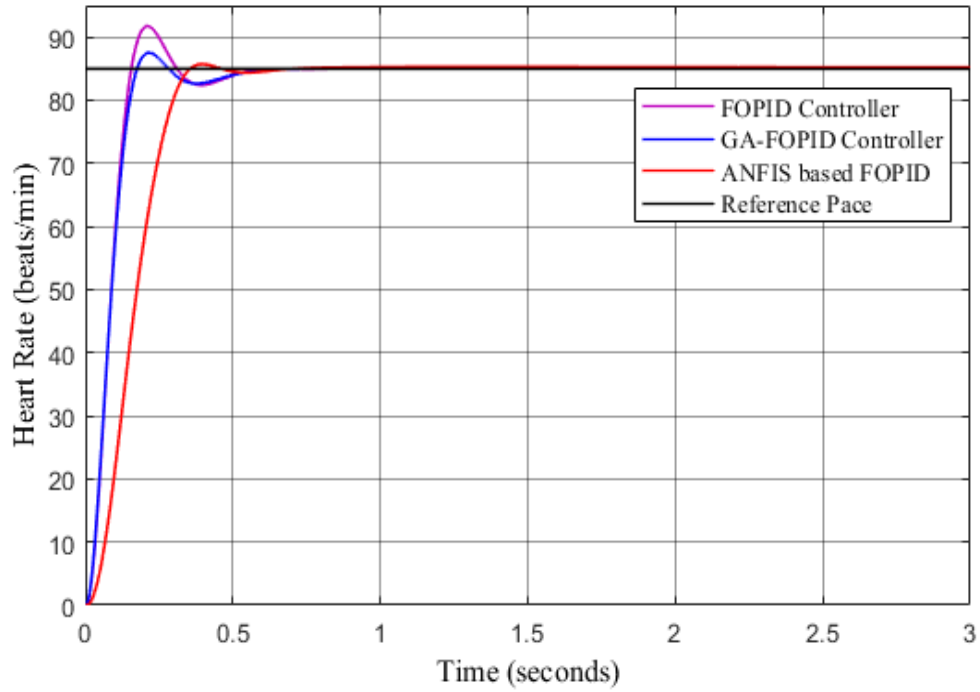


Figure 5.14: Comparison of result obtained by FOPID, GA tuned FOPID and ANFIS based FOPID controller HR=85 bpm

Table 5.10: Performance measures of the FOPID, GA tuned FOPID, and ANFIS-based FOPID controller for 100bpm heart rate

Controller	100 beat per minute heart rate		
	Rise Time (sec)	Settling Time (sec)	Maximum Overshoot (%)
FOPID	0.1117	0.5195	7.9317
GA tuned FOPID	0.1103	0.4503	2.9257
ANFIS based FOPID	0.2551	0.3893	0.1760

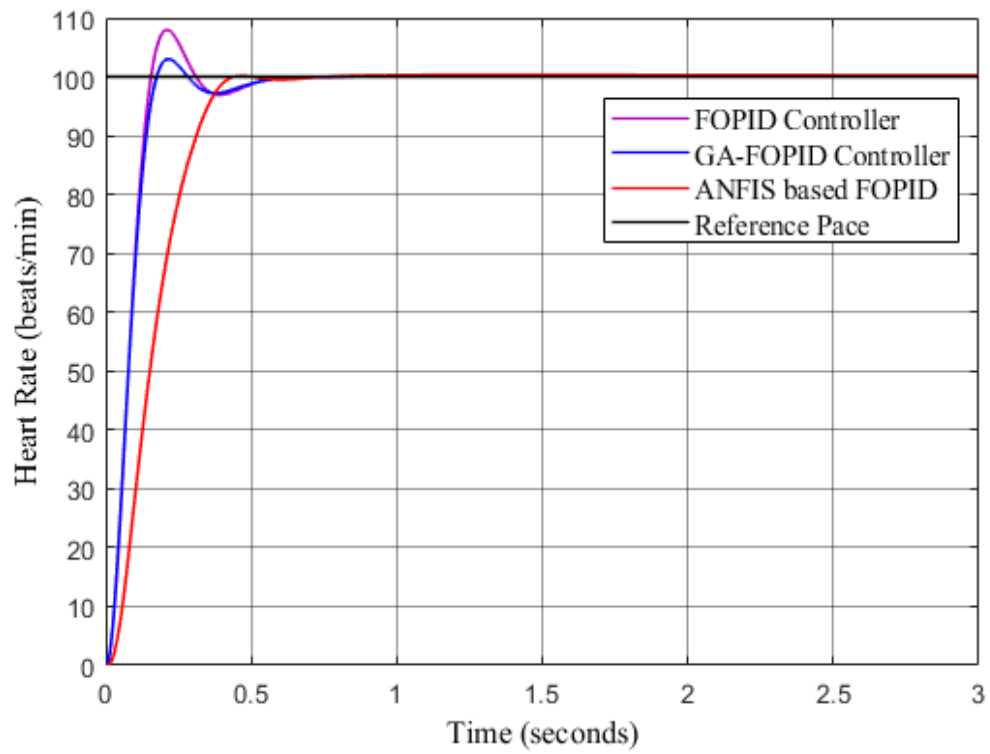


Figure 5.15: Comparison of result obtained by FOPID, GA tuned FOPID and ANFIS based FOPID controller HR=100 bpm

CHAPTER SIX

6. CONCLUSION AND RECOMMENDATION

6.1. Conclusion

In this thesis, mainly four tasks have been performed. These are design and simulation of conventional FOPID Controller and ANFIS based FOPID controller for tracking of real healthy ECG signal, design of FOPID and GA tuned FOPID controller for tracking and regulation of heart rates, design of ANFIS based FOPID controller for tracking and regulation of heart rates and finally, comparative analysis based on MATLAB/Simulink results of the above three controllers have been performed.

Several cardiac-related mathematical models associated with the heartbeat were studied and the Zeeman model was selected. The model validation was tested by using the Breitenstein principle. It is seen that both FOPID and ANFIS-based FOPID controllers applied to the Zeeman heart model were able to track ECG data obtained from the physioNet database accurately. The average heart rate (72 bpm) used for the reference signal was calculated from the ECG signal pattern in terms of a beat per minute by finding the interval between two consecutive R waves. In addition to this, the brain signal can indirectly affect the heart's electrical conduction activity. The BETA-type brain signals like anxiety, disease was considered as a disturbance. To compare the performance of the controller developed by ANFIS with FOPID, different evaluation conditions have been selected. The chattering observed in the FOPID result was eliminated and root mean square error was also reduced while using the proposed ANFIS-based FOPID controller for heart model.

Heart rate signals are used as a reliable indicator of heart diseases. These HR signals form the basis of the functioning of a pacemaker. Pacemaker performance depends not only on sensors and the pacemaker circuitry but also on the performance of the controller. In this thesis work, different control techniques are analyzed to track and regulate heart rates. Initially, a FOPID controller is tuned with the help of the FOMCON toolbox and genetic algorithm optimization method respectively to satisfy the different performance parameters. The FOPID controllers thus tuned with FOMCON toolbox and GA method do not provide an overall satisfactory response. Therefore, to improve the performance of the system, an ANFIS-based FOPID controller is designed with the help of different rule bases and knowledge bases. It is observed from the response of the ANFIS-based FOPID controller that the parameters settling time and maximum overshoot is within the limits and a smooth and

satisfactory response is obtained. Therefore, it is concluded that the overall performance of the proposed ANFIS-based FOPID controller for tracking and regulation of heart rate is better compared to the FOPID and GA-tuned FOPID controller.

6.2. Recommendation

Even if successful result was obtained based on the control objective, there are other several ways where this research can also be improved in the future. Some of those are mentioned below:

- In this thesis, the FOPID controller parameters for tracking of ECG signal were tuned manually. Appropriate optimization algorithm and other nonlinear control techniques may be applied to get better results in the future.
- I encourage future researchers to consider external disturbances, parameter variations, and time-delay attacks on the sensed loop while tracking ECG signals using nonlinear control techniques.
- The pacemaker controller can be done for various body conditions of the patient such as rest, walking, running and other cases need to be studied for temperature or environmental changes in the future.
- Heartbeat model used in this thesis is based on the Sino-atrial (SA) node pace making, dominant pacemaker compared with the lower one produced by Atrial-ventricular (AV). In the future, researchers can use coupled VDP oscillator models at AV node and His purkinjie system to make pace making better.
- The thesis can be also extended for hard ware implementation using Application specific integrated chips (ASICs), Digital signal processors (DSPs) and Field programmable gate array (FPGA).
- Finally, I recommend future researchers to consider different sensors input which incorporate with cardiac pacemaker system such as accelerometers, HRV, GSR, minute ventilation, blended sensor...etc. while designing control mechanism.

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APPENDIXES

Appendix A: Detailed Simulink representation of the system

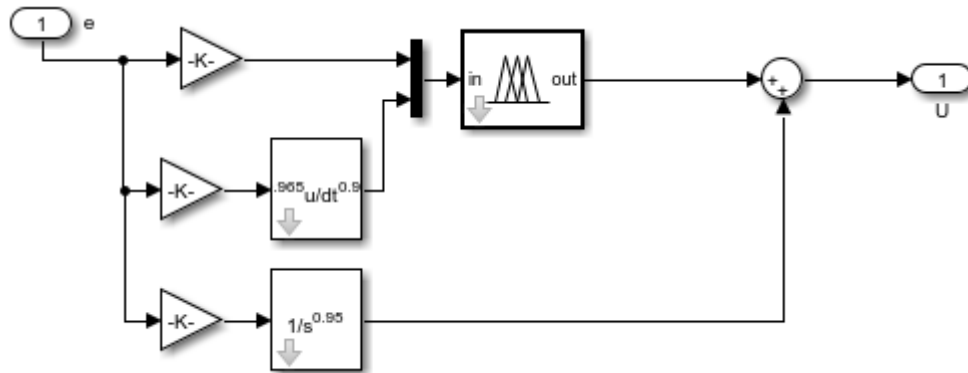


Figure A.1: Realization of ANFIS based FOPID controller for heartbeat system

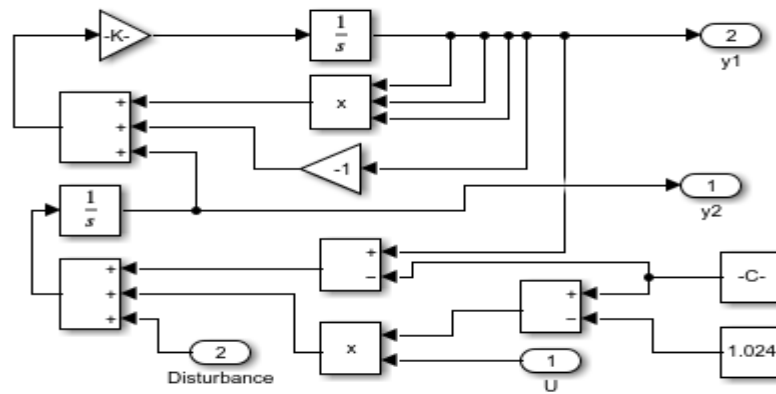


Figure A.2: Realization of heartbeat system

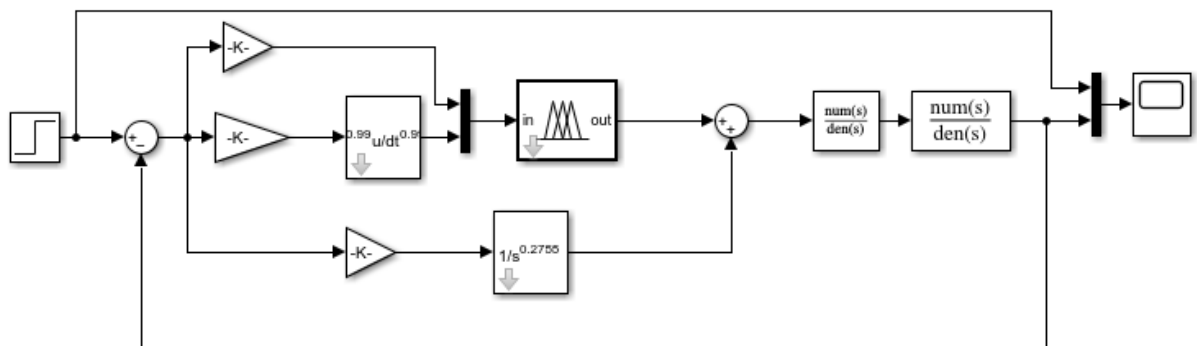


Figure A.3: Realization of ANFIS based FOPID controller for heart rate regulation

Appendix B: Appendixes Related To M-Files for GA Optimization

```
function Tune_fopidmodel1_SP
clear all
options= optimoptions('ga','PlotFcn', @gaplotbestf);
options.PopulationSize= 30;
options.MaxGenerations= 50;
nvar= 5; % Number of parameters
LB= [1 0.1 0.2 0.2 0.9]% Lower bound of tuning parameters
UB= [2 0.5 0.4 0.3 0.99];% Upper bound of tuning parameters
[x,fval]= ga(@EvalObj,nvar,[],[],[],[],LB,UB,[],options)
simulink_model= 'Tuned';
load_system(simulink_model);
gains= get_param(simulink_model,'modelworkspace');
gains.assignin('E', x(1));
gains.assignin('F', x(2));
gains.assignin('G', x(3));
gains.assignin('H', x(4));
gains.assignin('I', x(5));
function obj= EvalObj(x)
simulink_model= 'Tuned';
load_system(simulink_model);
gains= get_param(simulink_model,'modelworkspace');
gains.assignin('E', x(1));
gains.assignin('F', x(2));
gains.assignin('G', x(3));
gains.assignin('H', x(4));
gains.assignin('I', x(5));
simOut= sim(simulink_model,'SaveOutput','on');
t= simOut.find('t');
e= simOut.find('e');
n= length(e);
ee= abs(e);
obj= 0;
for i= 2: n
stepsize= t(i)-t(i-1);
obj= obj+0.5*(ee(i-1) +ee(i)) *stepsize; % ITAE
end
```

Appendix C: Tracking of ECG signal and heart rate regulation using Adaptive Neuro Fuzzy Inference System using MATLAB procedure

Seven Linguistic Variables have been defined for both inputs and output as Negative Big (NB), Negative Medium (NM), Negative Small (NS), Zero (Z), Positive Small (PS), Positive Medium (PM) and Positive Big (PB)). Their respective plots are shown below.

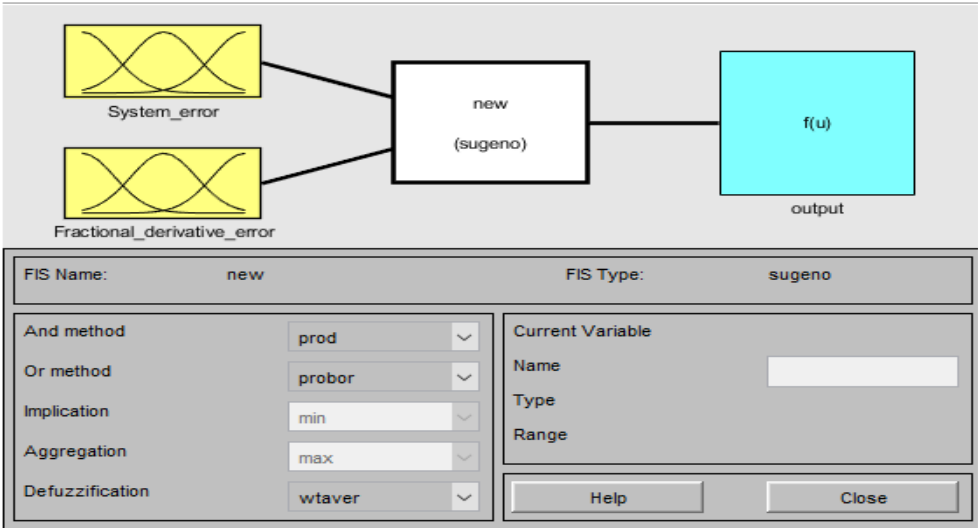
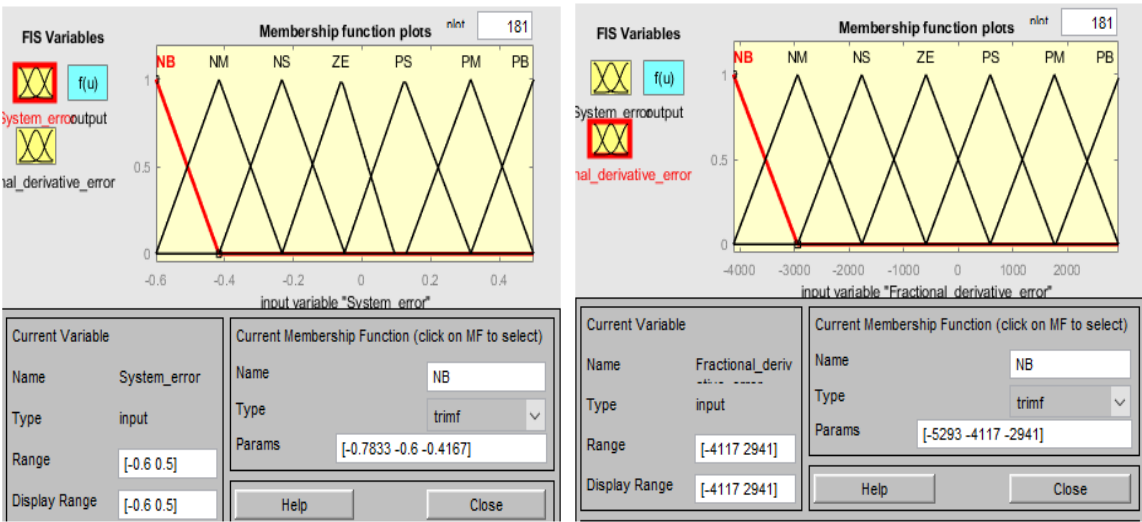


Figure C.1: FIS Property Editor for Sugeno type Fuzzy Model with two inputs and one output

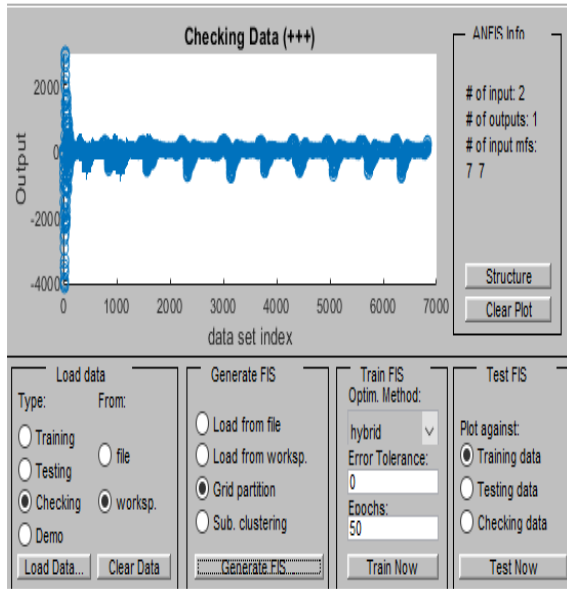
✓ **Tracking of ECG signal using Adaptive Neuro Fuzzy Inference System using MATLAB procedure.**



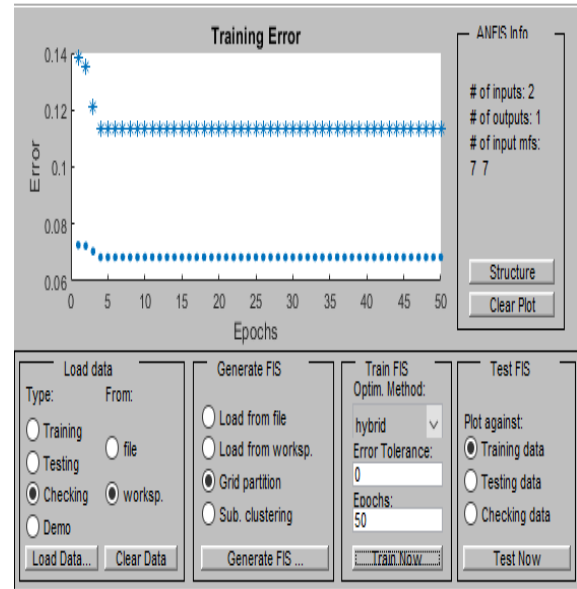
(a) System error MFs

(b) Fractional derivative of error MFs

Figure C.2: System Error and Fractional Derivative of Error MFs for ECG Signal Tracking

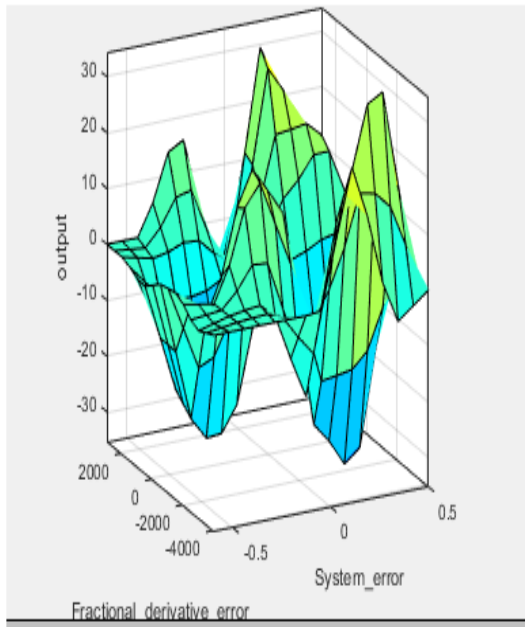


(c) Training and checking data sets

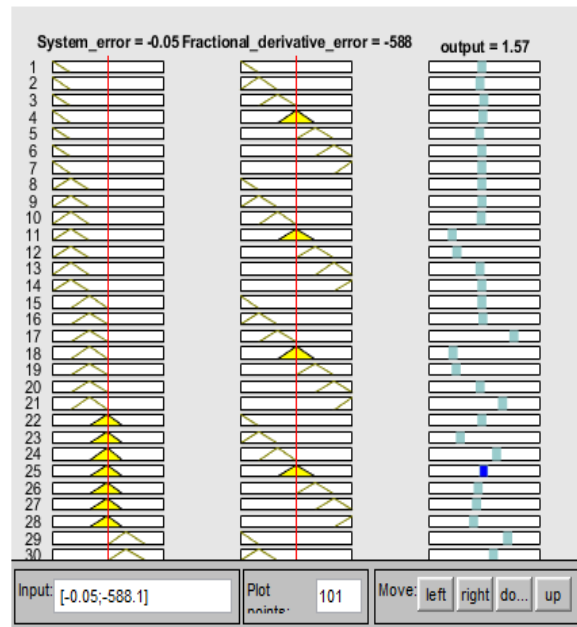


(d) Training and checking errors

Figure C.3: Training and Checking Data Sets and Errors of ECG SignalTracking



(e) Surface viewer



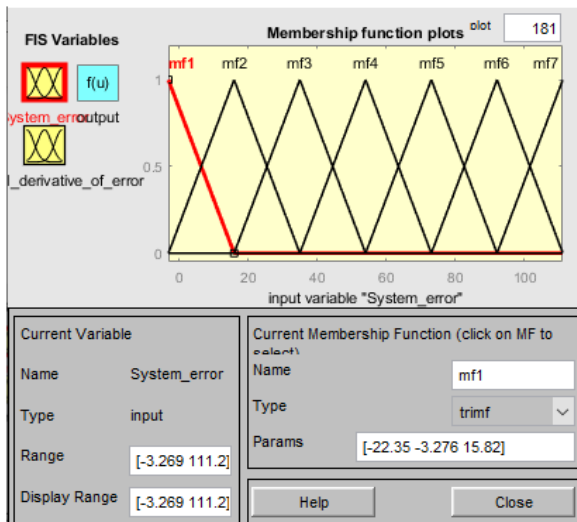
(f) Rule viewer

Figure C.4: Surface and Rule Viewer of ECG Signal Tracking

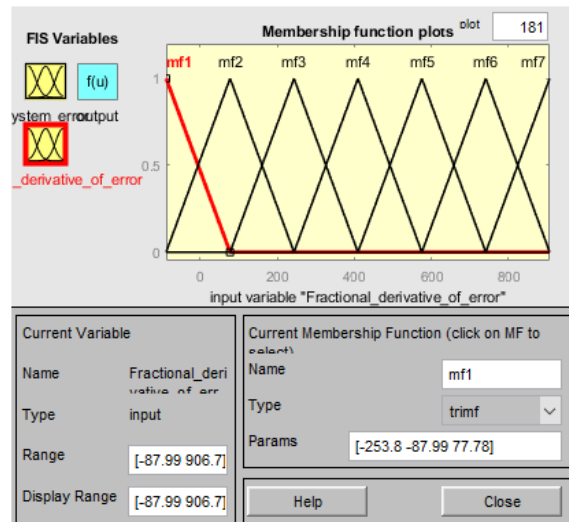
Minimal training RMSE is 0.113162 and minimal checking RMSE is 0.0664283 is achieved for ANFIS, as a controller. 50 epochs training conducted with hybrid training and zero error tolerance. Satisfactory performance is obtained with the above RMSE.

✓ **Heart rate tracking and regulating using Adaptive Neuro Fuzzy Inference System using MATLAB procedure.**

For tracking and regulating of heart rate seven Linguistic Variables have been defined for both inputs as mf1, mf2, mf3, mf4, mf5, mf6 and mf7. Their respective plots are shown below.

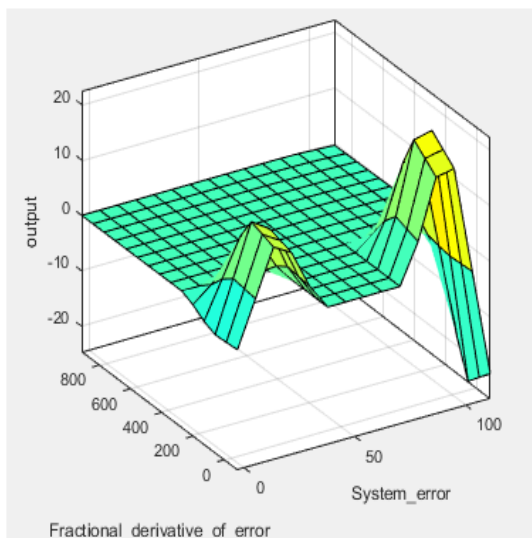


(g) System error MFs

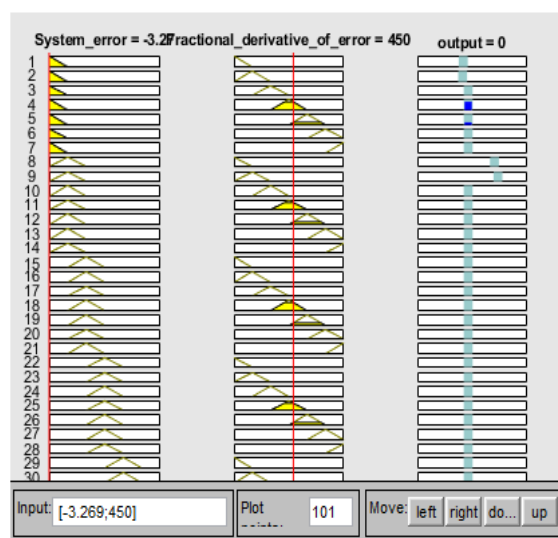


(h) Fractional derivative of error MFs

Figure C.5: System Error and Fractional Derivative of Error MFs for Heart Rate Regulation



(i) Surface viewer



(j) Rule viewer

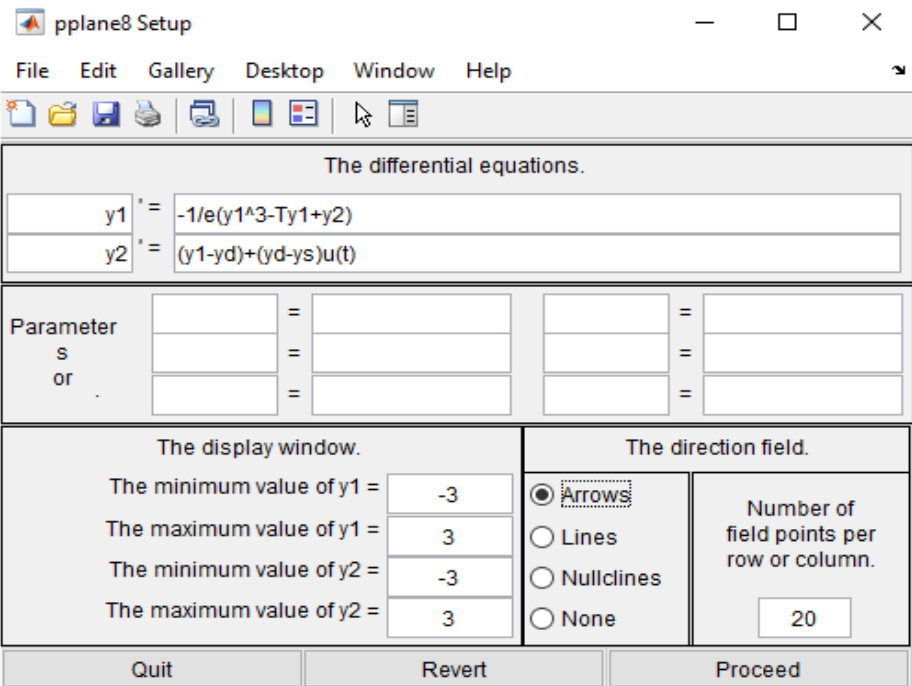
Figure C.6: Surface and Rule Viewer of Heart Rate Regulation

Appendix D: MATLAB code for Generating ECG signal From PhysioNet database

and constants

```
load ('16774m.mat'); % load val signal and 16773m is heartbeat id for a person
valu=transpose(val);
Esignal= (valu-0)/400; % 0 is for base and 400 is for gain
Fs=128; % sampling freq in Hz
t = (0: length (Esignal)-1)/Fs; % time is 1/Fs
time=transpose(t);
ECGref=[time Esignal];
subplot (3,1,1);
%%
plot(t,Esignal);
xlabel('time (seconds)');
ylabel('ECG signal (mV)');
title ('ECG signal sampled at 128Hz');
%%
% When high pass filter is applied
hf=fir1(1000,1/360*2,'high');
Esignal_filter = filter (hf,1, Esignal);
subplot (3,1,2);
%%
plot(t,Esignal_filter);
xlabel('time (seconds)');
ylabel('ECG signal(mV)');
title('Applying High-pass filter');
%%
% When the signal is squared
desqr=(Esignal_filter. ^2);
subplot (3,1,3);
%%
plot(t,desqr);
xlabel('time (seconds)');
ylabel('ECG signal (mV)');
title ('when the Signal is squared ');
%%
yd=1.024; % muscle fiber length during diastole
T=1; % muscle tension at low blood pressure
T1=10; % muscle tension at high blood pressure
e=0.2; % e and e1 fast eigenvalues of the system
e1=10;
ys=-1.3804; % muscle fiber length during systole state
```

Appendix E: pplane 8.m was used to draw the phase portraits plot of heartbeat system



Appendix F: FOMCON Toolbox details for the optimizing FOPID parameters

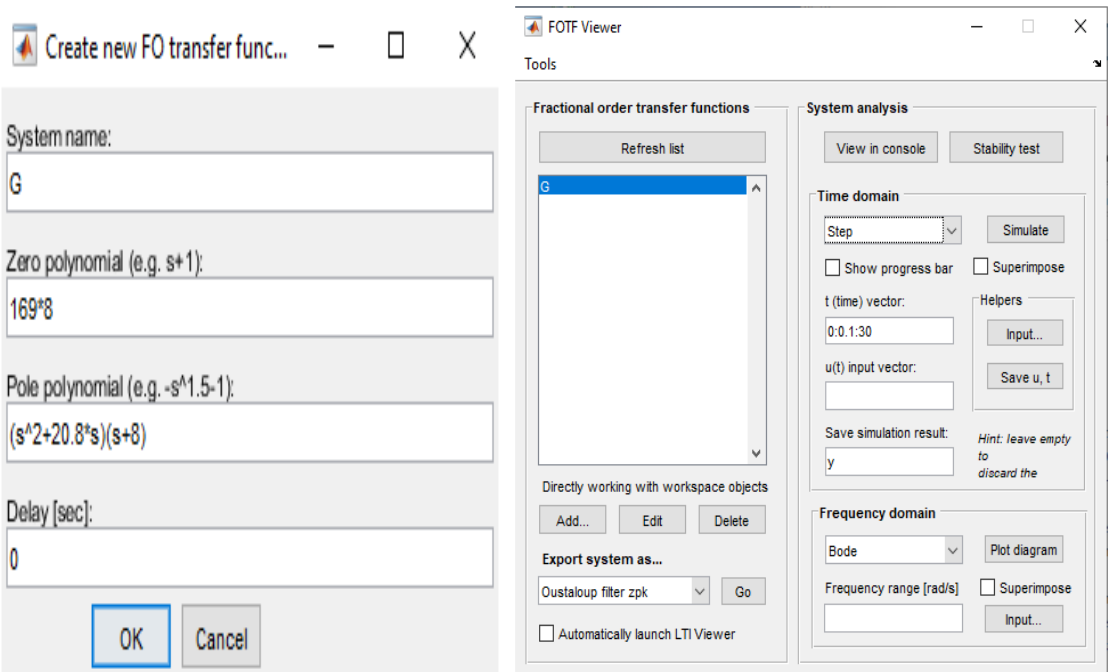


Figure F.1: New FO transfer function and FOTF Viewer graphical user interface respectively

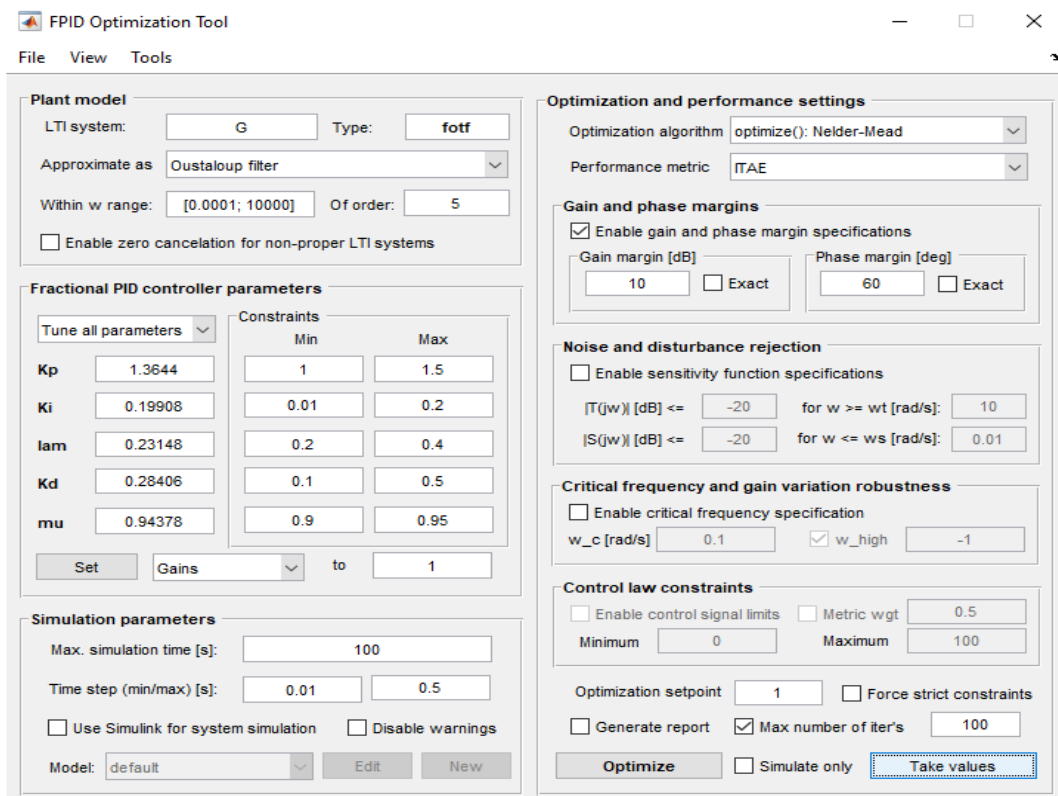


Figure F.2: FPID Optimization Tool

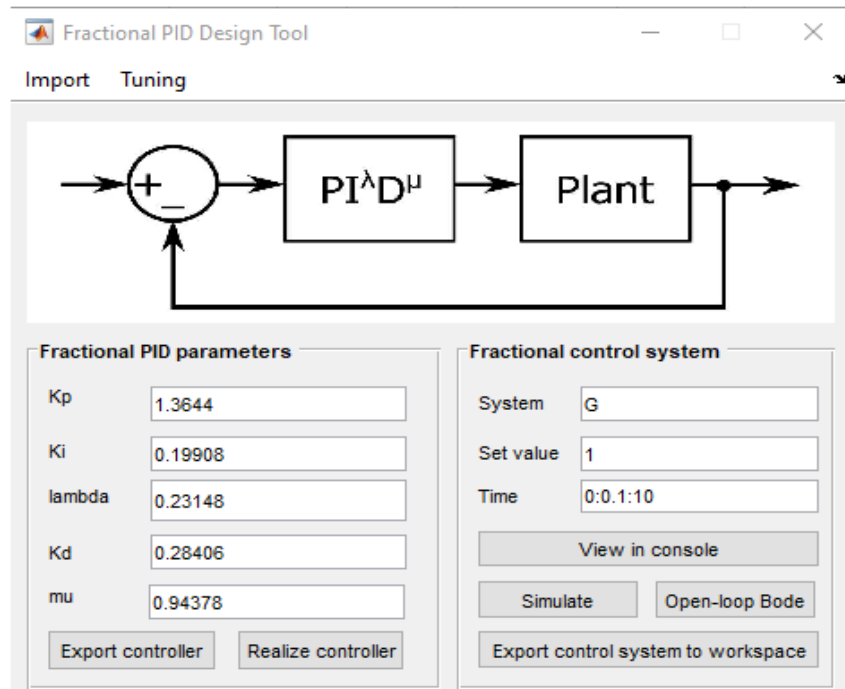


Figure F.3: Fractional PID Design Tool graphical user interface