[TYPE THESIS TITLE HERE]ARRHYTHMIA DETECTION AND CLASSIFICATION USING RHYTHM ORIGIN AND HEART RATE VARIATION

byBy

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A thesis submitted to the faculty of Computer Science Department, Military College of Signals, National University of Sciences and Technology, Rawalpindi in partial fulfillment of the requirements for the degree of ~~M~~BS in Computer S~~ciences~~oftware

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ABSTRACT

ARRHYTHMIA DETECTION AND
CLASSIFICATION USING RHYTHM ORIGIN
AND HEART RATE VARIATION

~~[THESIS TITLE]~~

~~by [Your Name]~~

~~Chairperson of the Supervisory Committee: Professor [Name]~~
~~Department of [Name]~~

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~~The ECG is the electrical manifestation of the contractile activity of the heart~~

We have developed an arrhythmia diagnosis system with the name CARDIOGENIC which discriminates supra-ventricular rhythms from ventricular ones. ~~This paper explains the modular details of the system that has been tested thoroughly.~~ The system uses a power spectrum of QRS complex to separate the origin of a rhythm and then it classifies about 12 types of cardiac arrhythmias based on the heart rate and its variations. The system has been developed using human ECG data recordings from the MIT-BIH arrhythmia database and AHA database. Two kinds of origin, supra-ventricular and ventricular, were distinguished with a sensitivity and specificity greater than 80%. The discrimination has been achieved using linear discrimination method with weighted summation of two power spectrum inputs (0.6104 and 1.2207 Hz). The phase of arrhythmia classification is accomplished by using the RR-intervals for calculating the Heart Rate (HR) and the variation thresholds. The beat-by-beat accuracy is found to be well above 75%.

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DECLARATION

No portion of the work presented in this dissertation has been submitted in support of any other award or qualification either in this institution or elsewhere.

DEDICATION

In the name of Allah, the All Compassionate, the Most Merciful

To our parents whose prayers and support has led us to produce a work of this magnitude

ACKNOWLEDGMENTS

We wish to thank Allah Almighty who gave us the strength and determination to pursue this project.

We gratefully acknowledge the continuous guidance and encouragement provided to us by Lt. Col Naveed Sarfraz Khattak. We would like to thank our project supervisors Assisstant Professor Tauseef Ahmed Rana (MCS), Brigadier Syed Mohammad Imran Majeed (AFIC/NIHD) and Dr. Shoaib (CARE) without whose personal supervision, advice and help, timely completion of this project would have been impossible. We extend very special thanks to the Head of Department Col. Raja Iqbal for his moral and technical support.

We would also like to thank Mr. Robert Tratnig who helped us with the understanding of the development environments, Ms. Andy Coldrey, the customer service manager of Del Mar Reynols Medical Ltd., Britain for providing us with necessary information to put the system in real time use and Mr. Pat Anderson from EP Ltd who helped us with the analysis queries during design of the system architecture.

We deeply treasure the unparallel moral support that we received from our friends. We are also deeply indebted to our families, especially our parents whose prayers and heartiest appreciation have brought this work into existence.

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LIST OF ABBREVIATIONS

AED	Automatic External Defibrillator
AF	Atrial Fibrillation
AFIC	Armed Forces Institute of Cardiology
AHA	American Heart Association
ANN	Artificial Neural Network
ATHD	Atrial Threshold
AV	Atrial Ventricular
BBB	Bundle Branch Block
BPA	Back Propagation Algorithm
bpm	Beats per Minute
DWT	Discrete Wavelet Transform
ECG	Electrocardiogram
EP	Electro-Physiology
FFT	Fast Fourier Transform
FR	Functional Requirement
HF	High Frequency
HOS	Higher Order Statistics
HR	Heart Rate
HRV	Heart Rate Variation
ICD	Implantable Cardio-verter Defibrillator
LDA	Linear Discriminator Analysis
LF	Low Frequency
NIHD	National Institute of Heart Diseases
PAC	Premature Atrial Contraction
PIMS	Pakistan Institute of Medical Sciences
PSD	Power Spectral Density
PST	Power Spectrum Transformation
PVC	Premature Ventricular Contraction

QR	Quality Requirement
RSA	Respiratory Sinus Arrhythmia
SD1	Standard Deviation 1/minor axis
SD2	Standard Deviation 2/major axis
SDNN	Standard Deviation of all Normal RR intervals
SNR	Signal-to-Noise Ratio
SVM	Support Vector Machine
SVR	Supra-Ventricular Rhythm
SVT	Supra-Ventricular Tachycardia
TFD	Time Frequency Distribution
ULF	Ultra Low Frequency
VF	Ventricular Fibrillation
VLF	Very Low Frequency
VR	Ventricular Rhythm
VT	Ventricular Tachycardia
VTHD	Ventricular Threshold

Chapter 1

~~Glossary~~
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Introduction

1.1 Introduction

The understanding of the research area under consideration is staunchly supported by the background knowledge on it. In the following, a glance at the project's domain, the focus of the work, the problems to be solved and the motivation on them are reviewed.

1.2 Project Domain Overview

The field of *Biomedical Engineering* uses traditional engineering expertise to analyze and solve problems in biology and medicine, providing an overall enhancement of health care. The concept guiding the pursuit of biomedical engineering is the application of advanced technology to the complex problems of medical care. The wide interaction of engineering with medicine has called upon a huge range of capacities like designing instruments, devices, and software, to bring together knowledge from many technical sources to develop new procedures, or conducting research needed to solve clinical problems.

Systems Physiology is the term used to describe that aspect of biomedical engineering in which engineering strategies, techniques and tools are used to gain a comprehensive and integrated understanding of the function of living organisms ranging from bacteria to humans. Computer modeling is used in the analysis of experimental data and in formulating mathematical descriptions of physiological events. Living systems have highly regulated feedback control systems that can be examined with state-of-the-art techniques. Examples are the applications of expert systems and artificial intelligence to clinical decision making, that is, computer-based systems for diagnosing diseases.

Electrophysiology is one area where an extensive and growing application of engineering techniques is found. Especially, electrocardiography has benefited immensely from the computer-based automation systems. The significance of automation in clinical diagnosis of heart disorders arises from the fact that it requires extremely prompt responses from the

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monitoring hardware devices such as pacemakers, ICD and AED etc and the degree of accuracy required of the diagnosis is also very high. Therefore, from engineering prospective, there lies a challenge between the timeliness of the solution and the precision of the solution. In recent years, much research has been done to develop new algorithms or methods based on the micro-processor or minicomputer to allow physicians to derive more information for diagnosis.

1.3 Brief Anatomy of the Human Heart

The human heart is basically a strong muscle. The cardiac muscle has a vital function in the human body and the implications of its proper functioning are imperative to the study of the research done.

1.3.1 Structure

The human heart is primarily a shell. There are four cavities, or open spaces, inside the heart that fill with blood. Two of these cavities are called atria. The other two are called ventricles. The two atria form the curved top of the heart. The ventricles meet at the bottom of the heart to form a pointed base which points toward the left side of the chest. The left ventricle contracts most forcefully, so that the heart can be felt pumping at the left side of the chest.

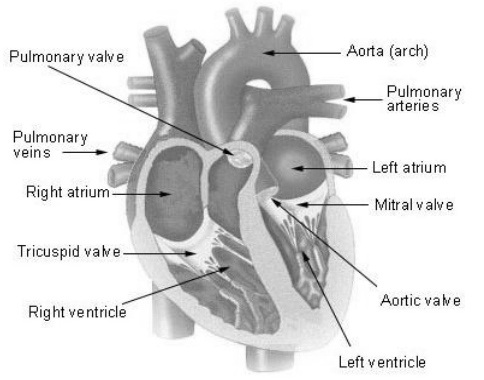

Figure 1.1 Structure of the Heart

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The left side of the heart houses one atrium and one ventricle. The right side of the heart houses the others. A wall, called the septum, separates the right and left sides of the heart. A valve connects each atrium to the ventricle below it. The mitral valve connects the left atrium with the left ventricle. The tricuspid valve connects the right atrium with the right ventricle.

The top of the heart connects to a few large blood vessels. The largest of these is the aorta, or main artery, which carries nutrient-rich blood away from the heart. Another important vessel is the pulmonary artery which connects the heart with the lungs as part of the pulmonary circulation system.

The heart's structure makes it an efficient, never-ceasing pump. From the moment of development through the moment of death, the heart pumps. The average heart's muscle, called cardiac muscle, contracts and relaxes about 70 to 80 times per minute. As the cardiac muscle contracts it pushes blood through the chambers and into the vessels. The processes of contraction and relax are termed as depolarization and repolarization respectively.

To monitor the heart, scanning technology or surgery is generally used. However, this is practicable only after a reliable diagnosis of the disorder is made. The disorder refers to some disturbance in the beats generated by the heart.

1.3.2 Rhythm Generation

The heart beats or the heart rhythms are recorded in the form of Electrocardiograph. The ECG is the electrical manifestation of the contractile activity of the heart that can be recorded fairly easily. Physicians interpret the morphology of the ECG waveform and decide, whether the heartbeat belongs to the normal sinus rhythm or to the appropriate class of disorder. The ECG contains certain data points that make it possible to extract useful information.

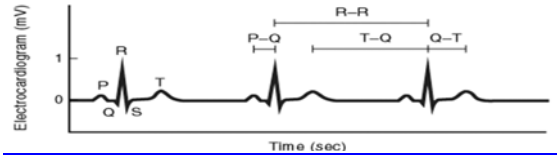

Figure 1.2 An ECG Waveform showing Useful Data Points

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The heart keeps on pumping steadily because of its natural "wiring". This amazing system of specialized nerve fibers conducts electrical impulses through the heart muscle tissue to regularly trigger the heartbeat. It is the polarization processes of the heart muscle that bring about the ECG waveform.

The generator for this system is the sinus node, a small area located in the heart's right atrium, or upper chamber. From the sinus node, a network of electrical fibers fans out to rhythmically jolt heart muscle into action. This sinus node, the heart's natural pacemaker, first stimulates the heart's two upper chambers, the atria to contract. A split second later, the impulse travels to another bundle of fibers, the atrio-ventricular node, and stimulates an even stronger contraction in the ventricles, the heart's main pumping chambers. Any disruption of this carefully orchestrated sequence can lead to a cardiac rhythm disorder, often with dire results.

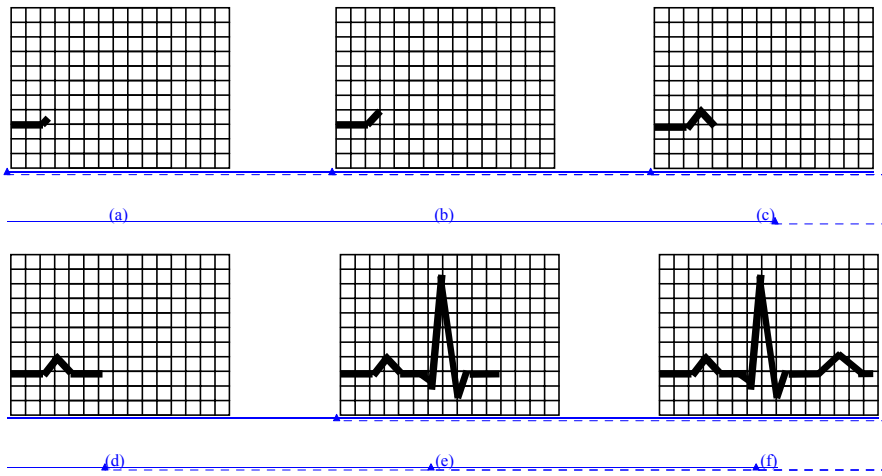

Figure 1.3 Polarization Processes in the Heart (a) Pacemaker Potential (b) Atrial Contraction (c) P-Wave: Atrial Decomposition (d) Conduction through Branch Bundles and Purkinje Fibers (e) QRS complex: Ventricular Depolarization and Atrial Repolarization (f) T- Wave: Ventricular Repolarization

1.4 Arrhythmia

Any disturbance in the regular rhythmic activity of the heart, that is, amplitude, duration, and the shape of rhythm is termed *arrhythmia*. There are many different types of

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arrhythmias, some much more serious than others. A minor arrhythmia may cause a single "skip" in the heartbeat. Almost all people feel this type of sensation at some point in their lives and this generally does not need special treatment. On the other end, much more dangerous arrhythmias may cause cardiac arrest, that is, sudden death.

1.4.1 Types of Cardiac Arrhythmias

Abnormalities of heart rate are classified as those that are too fast called tachycardia and those that are too slow called bradycardia. Arrhythmias are also classified by the region of the heart that gives rise to the rhythm. Normally, heartbeat arises from the sinus node; therefore, the normal heart rhythm is called normal sinus rhythm. Places where abnormal heart rhythms can start include the atria and the ventricles.

Several common types include sinus rhythms, premature contractions, heart blocks, junctional arrhythmias, tachyarrhythmias etc. Examples of each type are given respectively.

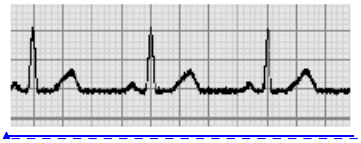

Figure 1.4 ECG Having Normal Sinus Arrhythmia

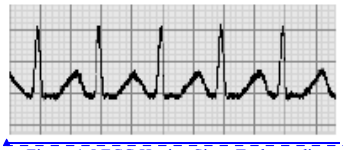

Figure 1.5 ECG Having Sinus Tachycardia

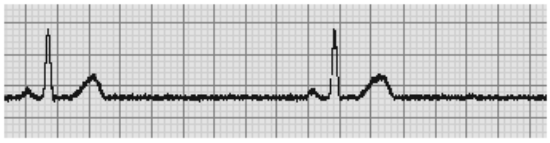

Figure 1.6 ECG Having Sinus Bradycardia

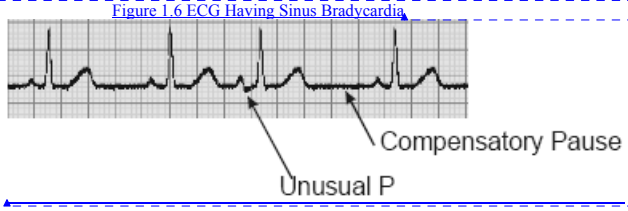

Figure 1.7 ECG Having Premature Atrial Contraction

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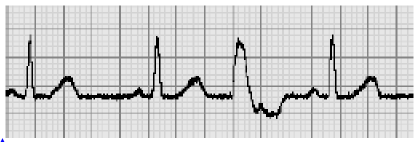

Figure 1.8 ECG Having Premature Ventricular Contraction

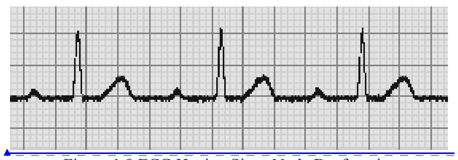

Figure 1.9 ECG Having Sinus Node Dysfunction

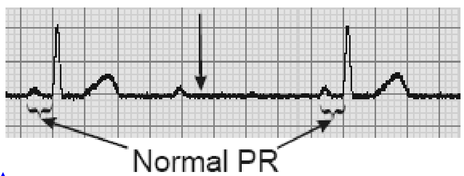

Figure 1.10 ECG Having AV Block

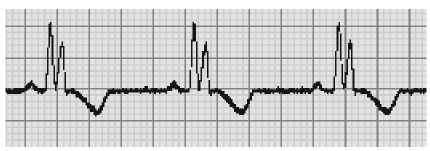

Figure 1.11 ECG Having Bundle Branch Block

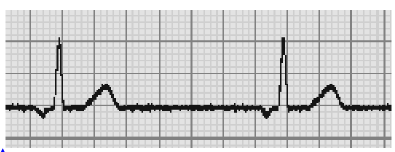

Figure 1.12 ECG Having Junctional beat

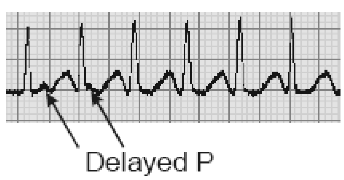

Figure 1.13 ECG Having Supra-ventricular Tachycardia

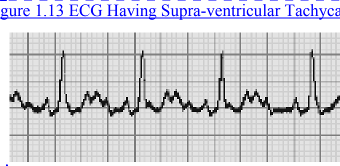

Figure 1.14 ECG Having Atrial Flutter

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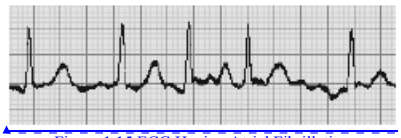

Figure 1.15 ECG Having Atrial Fibrillation

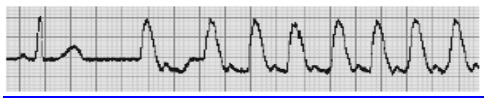

Figure 1.16 ECG Having Ventricular Tachycardia

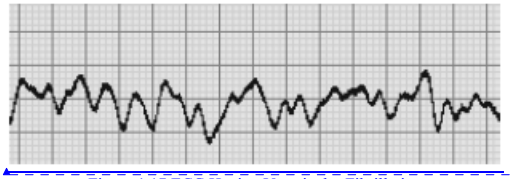

Figure 1.17 ECG Having Ventricular Fibrillation

1.4.2 Diagnosis of Arrhythmia

Many arrhythmias are detected during a routine physical examination. If there are physical exam findings suggestive of arrhythmia, evaluation of the heart's rhythm is carried out. The first test is almost always an ECG. This will catch some, but not all, arrhythmias. Special monitors are also used to detect arrhythmias that are only present occasionally. A Holter monitor is a small portable tape recorder that is worn for 24 hours and provides a recording of every heartbeat during an entire day.

Finally, some arrhythmias are diagnosed during an EP study. An EP study is a procedure where catheters, inserted through a large vein in the leg or the neck, are positioned inside of the heart. A very specialized cardiologist uses these catheters to provoke the onset of arrhythmia. EP studies can be combined with ablation procedures to improve or cure some arrhythmias.

1.4.3 Treatment

The treatment to arrhythmias includes medications, ablation and pacemaker therapies. The pacemakers are implantable devices usually for treatment of slow heart rhythms. Certain fast heart rhythms also may be treated with a device called an Implantable Cardio-verter Defibrillator (ICD).

1.5 Problem to be Solved

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Automation of arrhythmia diagnosis proves highly significant when the challenges between time and precision are to be met. The difficulty in the development of a reliable system is posed by the highly variable characteristics of the heart rhythms in individuals by nature.

In the use of implantable devices, the immediate detection of life threatening arrhythmias as ventricular tachycardia and ventricular fibrillation is a critical feature in arrhythmia diagnosis. Mostly, the diagnosis of VT/VF is mainly based on the measurement of heart-rates from ECG signals, and information on the origin of a rhythm is not available. So it is hard to reject supra-ventricular tachyarrhythmia from VT/VF. In clinical applications, it is critical to identify supra-ventricular tachyarrhythmias such as supra-ventricular tachycardia (SVT) and atrial fibrillation [1].

The exact definition of the rhythm origin is very important so that an implanted device may not give out unnecessary shocks to the patient which may prove fatal. Two issues are involved here: first is the correct decision about the problem, that is, any sudden rhythm shift must be categorized with acceptable accuracy. Second is the correct timing of the decision, that is, the decision may be made quickly as the reaction time is very less in case of deadly arrhythmias.

1.6 Problem Statement

To design and implement a reliable approach to arrhythmia diagnosis that discriminates the origin of individual beats, that is, either it is a supra-ventricular rhythm (SVR) or a ventricular rhythm (VR) and then classifies the type of arrhythmia to which it belongs; and eventually use the analysis results to test the algorithm against real time data to achieve regressive assessment of the work.

1.7 Motivation

The automation of arrhythmia diagnosis is an increasingly interesting and glaring field for research. The introduction of the concept of origin discrimination is a step towards meeting the tradeoffs between time and precision. Further, the use of heart rate variation is an edge over the commercially available systems wherein the measurements are made on the basis of heart rates only. The point to emphasize is that the heart rate may fluctuate over a variable range [2] attributed to the natural variety of cardiac contractions; however, the variation in the HR does not exceed a certain maximum. Hence, suitable thresholds can be

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extracted using this fact and a huge variety of arrhythmias can be diagnosed with appreciable accuracy and with lesser processing delay.

~~1.6~~1.8 S

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The challenges on time and precision are deemed very important as an ever glaring field of practical significance. Hence, it becomes evident that application of engineering techniques and methods can be applied to meet the existing trade-offs.

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Chapter 2

Research Objectives

2.1 Introduction

The goals and objectives as well as the set of deliverables planned for the project are explained in this chapter. These goals have been formulated after a careful and thorough review on the research perspectives and the practical implication of the work.

2.2 Project Goals

The aims and objectives guiding the project are:

- To produce an efficient and reliable algorithm for automated arrhythmia detection and classification which can improve upon the commercially available software systems.
- To describe the relationship between the waveforms and intervals of the ECG and the electrical activity of the heart.
- To incorporate more and more generality in the system so that it is extendible to a wider range of arrhythmias than is available in ambulatory equipment such as holters.
- To allow for the real time application of applied methods.
- To compare the performance with other systems in practice.
- To make the system platform independent for use on common platforms.

2.3 Deliverables

The deliverables associated with the project are:

- A compiled and installable version of the system's package including the executable files and dynamic link libraries.
- User manuals.
- Training data.

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2.4 Summary

The clearly defined set of goals and products have been intended at guiding the development process in an organized manner. A comprehensive consideration of the above has proven a great help in the software development process.

Chapter 3

Literature Review

3.1 Introduction

Automated systems for arrhythmia analysis have proliferated since the early 1960's and many are used clinically. In recent years, much research has been done to develop new algorithms or methods based on the microprocessor or minicomputer to allow physicians to derive more information for diagnosis. These works generally involve automatic rhythm analysis, classification, and diagnosis. These techniques extract features, which are either temporal or transformed representation of the ECG waveforms. On the basis of these features classification is performed by template matching hidden markov models, combination of wavelets and artificial neural network, and the neural networks alone. Several other methods have been employed in ECG analysis, such as Fourier descriptor for rhythm analysis and ECG data compression, auto regressive modeling and syntactic methods for automatic processing of ECG waveforms.

3.2 QRS Complex Detection in Arrhythmia Classification

Beat classification is an important step in designing an arrhythmia classification algorithm. Beat detection is the procedure preceding any kind of ECG processing and analysis. Large varieties of methods have been proposed which feature high percentages of correct detection.

Considerable research has been carried out in the development of QRS complex detectors. QRS complex detection algorithms typically incorporate a preprocessing filter, nonlinear transformation and decision rule algorithms [4].

Linear discriminators and quadratic discriminators were employed for classification, QRS onset/offset detection and P-wave onset detection. The best classification performance of 82% was achieved by employing the QRS width as an input feature to the linear discriminator classifier [5].

Friesen et al. have presented a comparison of nine QRS detection algorithms, based on amplitude and first derivative, first derivative only, first derivative and second derivative and digital filtering.

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3.3 Artificial Neural Network Arrhythmia Classifier

Artificial neural networks (ANN) are biologically inspired networks that are useful in application areas such as pattern recognition, classification etc. The decision making process of the ANN is holistic, based on the features of input patterns, and is suitable for classification of biomedical data. Typically, multilayer feed forward neural networks can be trained as non-linear classifiers using the generalized back propagation algorithm (BPA).

ANN is being used for QRS detection, rhythm identification and arrhythmia classification. A supervised ANN is developed to recognize and classify the nonlinear morphologies. ANN trained with back propagation algorithm, classifies the applied input ECG beat to appropriate class [6].

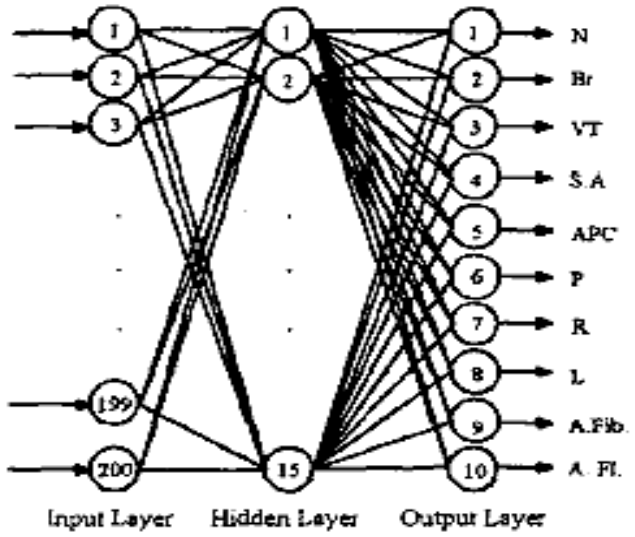

Figure 3.1 The Structure of ANN Arrhythmia Classifier

3.4 Fuzzy Hybrid Neural Network Classifier

A more efficient classifier is developed using a fuzzy equivalence relationship. The process of classification involves obtaining a fuzzy equivalence relationship matrix for each class of datum and then comparing a fresh input with each group for classification.

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The classification algorithm of the ECG beats, applying the fuzzy hybrid neural network and the features drawn from the higher order statistics is less sensitive to morphological variation of the ECG. The cumulants of the second, third and fourth orders are used for the feature selection. The hybrid fuzzy neural network consists of the fuzzy self-organizing sub-network connected in cascade with the multilayer perceptron used for recognition and classification of different types of heart beats [7].

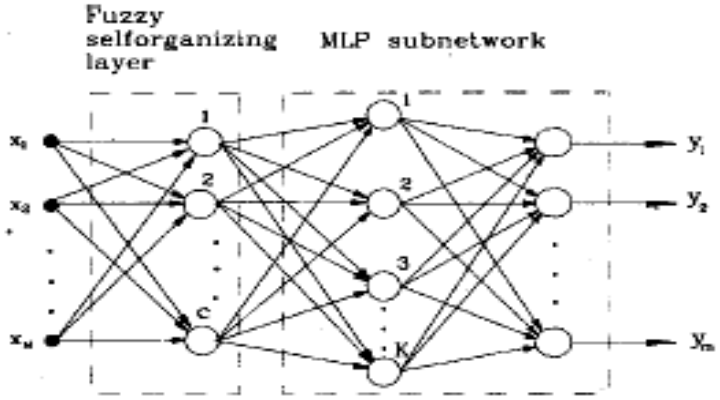

Figure 3.2 The Structure of Fuzzy Hybrid Neural Network

The main features of the classification algorithm are the simplicity, good recognition rate, fast performance and that the learned system works practically in real time.

3.5 Hidden Markov Models

This technique has been successfully applied to various automated recognition systems since the mid 1970's. Hidden Markov modeling is a statistical modeling technique that characterizes an observed data sequence by a probability density function which varies according to the state of an underlying markov chain.

Hidden Markov modeling has been applied to cardiac arrhythmia analysis with promising results. The hidden Markov modeling approach combines structural and statistical knowledge of the ECG signal in a single parametric model to classify the cardiac abnormalities [8].

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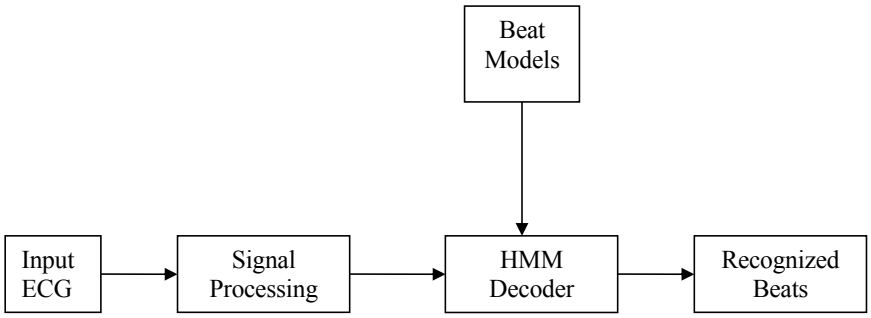

Figure 3.3 Hidden Markov Modeling Applied to ECG Analysis.

3.6 Wavelet Based Arrhythmia Classifier

Wavelet based classifiers are powerful for arrhythmia detection. Wavelet-based analysis of the ECG has two main advantages. One is that it provides the time-frequency distribution of the ECG and divides the signal into a set of time-frequency components that we can use to isolate abnormal signals. The other main advantage is that wavelet decomposition makes possible not only filtering out less important frequency components but also shortens the length of data that needs to be analyzed. With these advantages, reliable results for detecting and classifying arrhythmias are achieved

Wavelet transforms are used to extract features from the ECG. Wavelet transforms have been used in two ways for beat classification, that is, to detect the characteristic points of the ECG and to use the wavelet coefficients as the elements of the feature vectors [9].

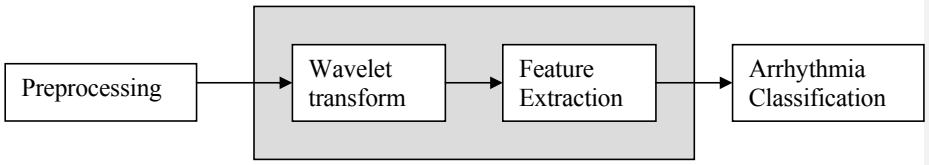

Figure 3.4 Block Diagram of Wavelet Based Arrhythmia Classifier

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3.7 Neural Wavelet Based Classifier

Neural networks and wavelet decomposition are used in combination to detect and diagnose the arrhythmias with increased accuracy. Back-propagation training algorithm performed on Discrete Wavelet Transformed (DWT) results to classify arrhythmias. The ability of the wavelet transform to decompose signal at various resolutions allows accurate detection of features from ECG. Back-propagation performs the gradient descent search to reduce the mean square error between the actual output of the network and the desired output through the adjustments of weights. Results show significant improvements in the sensitivity and specificity [10].

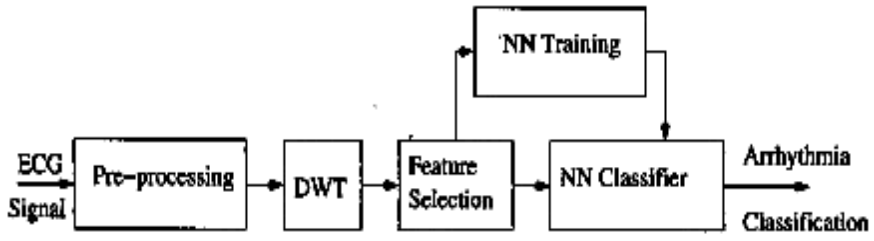

Figure 3.5 Block Diagram of Neural-Wavelet Based Arrhythmia Classifier

3.8 Support Vector Machine Classifier

The Support Vector Machine (SVM) classifier is known as an excellent tool for classification and regression problems with a good generalization performance. The recognition system uses the SVM working in the classification mode. Two different preprocessing methods for generation of features are applied. One method involves the Higher Order Statistics (HOS) while the second involves the Hermite characterization of QRS complex of the registered ECG waveform. Combining the SVM network with these preprocessing methods yields two neural classifiers, which have been combined into one final expert system. The combination of classifiers utilizes the least mean square method to optimize the weights of the weighted voting integrating scheme. The results of the performed numerical experiments for the recognition of heart rhythm types on the basis of ECG waveforms confirmed the reliability and advantage of the proposed approach [11].

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3.9 Arrhythmia Classifier Based On Time and Frequency Domain Analysis

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Many algorithms to detect shockable rhythms have been reported in the literature use information from the time domain and some others use information from the frequency domain. The performance of these algorithms is not ideal and each can be improved. The algorithms based on the frequency domain incorporate the short time Fourier transforms to compute the energy distribution of the ECG. Features are extracted from the energy distribution and are used in diagnostic classification algorithms .The fourier transform is well known to have a tradeoff in resolution between time and frequency, and the features are thus limited by the accuracy of the frequency distribution. Accurate methods of computing the time-frequency distribution (TFD) need to be determined for arrhythmia classification algorithms [12].

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3.10 Summary

The literature review presented includes the major and the most novel methods and techniques that have been in practice. This shall provide a manner for regressive assessment of the technique employed in this project.

Chapter 4

Requirement Analysis

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4.1 Introduction

The requirement analysis has been achieved after a thorough study of the existing system and its deficiencies as well as the utilities deemed necessary by the user. A certain well defined set of requirements has been derived which has to be met by the system.

4.2 Functional Requirements (FR)

FR1: Suitable conversion of data

FR2: Reliable and intelligent analysis of data

FR3: Rendering output report in printable and savable form

FR4: Reduction of processing delay to as minimum as possible

FR5: User friendly interface

FR6: System must be redistributable and environment independent

FR7: Optimal analysis power

4.3 Quality Requirements (QR)

QR1: To be application oriented

QR2: Provide complete set of interfaces

QR3: Be well documented

QR4: Be extensible

QR5: Be robust and fault tolerant

QR6: Provide verbose error messages

QR7: Allow integration of 3rd party soft wares

QR8: Allow monitoring

QR9: Be independent of other services

To meet the requirements identified in the above, the work of CARDIOGENIC has to be able to utilize other services. Certain functionality and properties are required from these services in order to achieve a successful outcome of our work.

Having identified requirements, the next step is to design architecture for the working package, which meets all requirements. Some requirements may not be enforced by the architecture, but have to be met at implementation level. The implementation plan for the system should cover the identified requirements completely.

4.4 Summary

The functional and non-functional requirements defined on the project are expected to be met by the system. It shall be a very important issue so as to make the system viable for future enhancement work and to render it as scalable as possible to allow for add-ons in future that will envisage improved functionality and efficieny.

Chapter 5

Design

5.1 Introduction

The design process is facilitated after the choice of a suitable process model. In this chapter, the system architecture details and the design issues to be met by the system have been discussed with a view to the practical implementation of the system.

5.2 Process Model

The process model used for the pursuit of system development is the "incremental evolutionary" model. This has been chosen since the modular division of the system demands that one module be built before the next and that the output of first module becomes the input to the other.

5.3 Modular Division

The system has been divided into two major modules. These are the *Origin Separator* and the *Arrhythmia Classifier*.

5.4 System Architecture

The system architecture is best illustrated in the diagram as shown below:

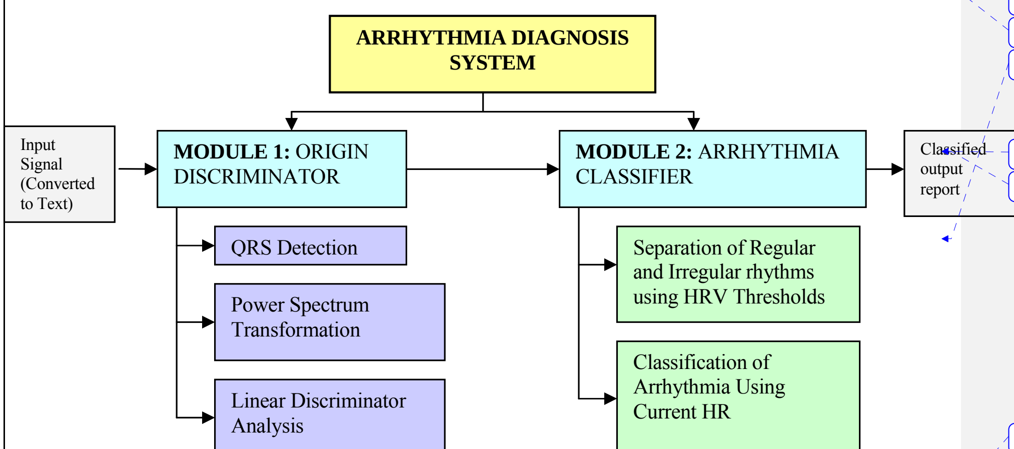

Figure 5.1 Arrhythmia Diagnosis System

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5.5 Module 1: *ORIGIN DISCRIMINATOR*

As shown in figure 5.1, the first module is the origin discriminator. This module takes the raw ECG recording as input.

5.5.1 Real Time Environment Operation

In real time environment, the raw input comes from the holter recorder. The holter is an ambulatory recorder that stays with the patient for an entire day to obtain full disclosure ECG. It records the ECG of the subject for a 24 hour period on a life card chip that is contained within the holter. When this recorded signal is loaded into a computer, the loader; which has an inbuilt analog-to-digital converter, performs the required conversion of the analog ECG data to digital form. The only constraint on the input data is that the digitized signal must be converted to text before any processing can be done on it.

The hardware for data acquisition, that is, the ambulatory holter recorders, the life card chip and the chip loader are manufactured by the Del Mar Reynolds Medical which is one of the leading multinational manufacturers in the cardiac diagnostic equipment.

5.5.1.1 Lifecard CF Digital Holter Recorder

Lifecard CF takes digital Holter recording to a higher level. 12-bit resolution and Ultrasharp™ technology deliver outstanding ECG quality for accurate analysis from the most challenging recording environments. The special recording features of the chip are that: only 3 electrodes for 3 channels of ECG are available, ECG display is continued during hook-up, 7-day continuous ECG recording is possible and 12-bit resolution for Crystal clear ECG is available.

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(a) (b)

(c) (d)

Figure 5.2 Different Views of Del Mar Reynolds Holter Recorder

However, at present, the system is developed using data from the standardized web-based research databases such as the MIT-BIH arrhythmia database and the American Heart Association (AHA) database. Though the evaluation of an algorithm on a database is not the ultimate answer to its utility in a clinical environment but it provides a standardized means of comparing the basic performance of one algorithm to another. These are employed for purely research purposes and the enhancement to real time data acquisition is discussed in chapter 10.

The schematic diagram of data flow expected for the real time operation and the current development are shown in figures 5.3 and 5.4 respectively.

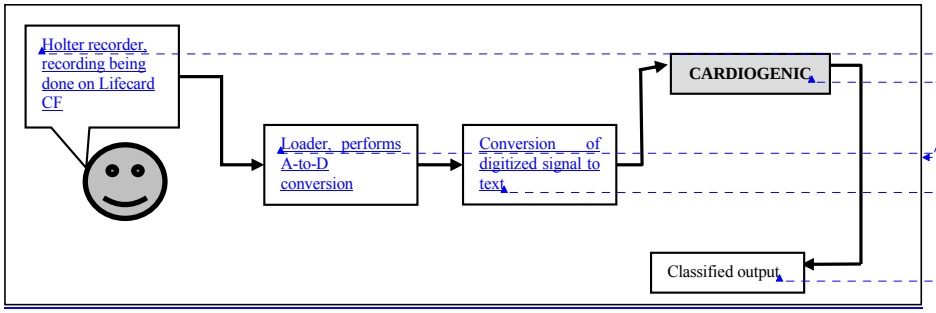

Figure 5.3 Schematic Diagram of Real Time Environment Operation

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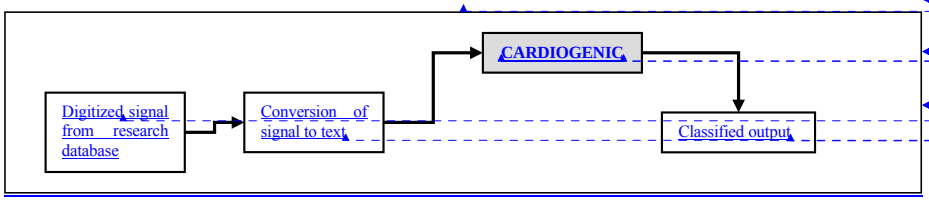

Figure 5.4 Schematic Diagram of Current Operation

5.5.2 Sub Modules

The origin discriminator comprises of three sub-phases: QRS Detection, Power Spectrum Transformation (PST) and Linear Discriminator Analysis (LDA). The input is first subjected to QRS detection followed by the power spectrum transformation and finally the LDA.

5.5.2.1 QRS Detection

In order to achieve reliable arrhythmia diagnosis, the most important and critical step is the correct and error free QRS detection in the ECG signals. The QRS complex, in addition to the P-wave and ST-segment, is the diagnostic critical wave in the ECG. However, the most reliable for diagnosis is the QRS complex and hence, the system has been designed with its detection at the very beginning.

Exiting data compression techniques for ECG signals fall in three categories, that is, the direct data transformation methods, transformation methods and parameter extraction techniques.

This thesis uses one of the transformation methods to analyze the spectral and energy distributions of the ECG signals and to detect and eliminate the redundancies. The transformation is conducted to each of the ECG beats. ECG beats are delineated by RR waves, and one RR wave starts from the peak sample of one QRS complex, and ends to the sample right before the peak of the next QRS complex.

The basic concept of the cycle-to-cycle compression is to represent a periodic signal by one cycle period and a count of the total number of cycles, and it is only valid for strict periodic signals where all the signal cycles are the same. Though the ECG waveforms do not bear such characteristic, ECG is considered a quasi-periodic signal [3].

It is noted that it is very important to accurately detect the QRS complex for cycle to cycle compression because it guarantees the adjacent cycles are statistically dependent to

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one another. While it is easier to detect the QRS complex from the normal ECG signals, it can be very difficult to accurately detect it from the abnormal ECG signals. Also, there can be various types of noises in the ECG signal. The noise sources include muscle noise, artifacts due to electrode motion, power line interference, base line wander and T-waves with high frequency characteristics similar to the QRS complex.

Software QRS detectors include one or more of three different types of processing steps, that is, linear digital filtering, nonlinear transformation or decision rule algorithm. The PAN-TOMPKINS technique has been employed for QRS detection. The developed algorithm developed uses all three types. However, to cater for the noise sources, digital filters are used to reduce their effect and to improve the signal-to-noise ratio. The figure 5.5 shows the block diagram of the algorithm.

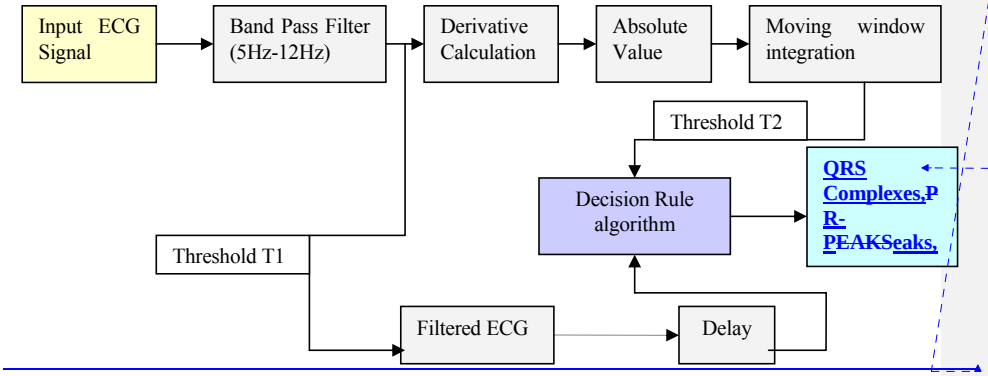

Figure 5.5 Block diagram of QRS detection algorithm

5.5.2.1.1 Band-pass filter

The band-pass filter shown in figure 5.5 is two filters cascaded, one is a low pass filter, and the other one is a high pass filter.

The transfer function of the second-order low pass filter is [3]

$$H(z) = (1-z^{-6})^2 / (1-z^{-1}) \quad (5.1)$$

And its difference equation is [3]

$$y(nT) = 2y(nT-T) - y(nT-2T) + x(nT) - 2x(nT-6T) + x(nT-12T) \quad (5.2)$$

where T is the sampling period. The cutoff frequency is about 12Hz and the gain is 36.

The filter processing delay is 6 samples.

The transfer function of the high pass filter is given by [3]

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$$H(z) = (-1 + 32z^{-16} + z^{-32})/(1 - z^{-1}) \quad 5.3$$

Its difference equation is

$$y(nT) = 32x(nT-16T) - [y(nT-T) + x(nT) - x(nT-32T)] \quad 5.4$$

The low cutoff frequency of this filter is about 5 Hz, the gain is 32.

5.5.2.1.2 Derivatives

Differentiation of the filtered signals is to provide the slope information of QRS complex since there are quick rise and fall times of the QRS complex in the ECG signals, taking the derivative of the ECG would make it easier to detect when the QRS complex occur. The transfer function of the five-point differentiation equation is given by [3],

$$H(z) = ((1/8)\ T)\ (-z^{-2} - 2z^{-1} + 2z^{1} + z^{2}) \quad 5.5$$

Its difference equation is given by [3],

$$y(nT) = ((1/8)T)\ [-x(nT-2T) - 2x(nT-T) + 2x(nT+T) + x(nT+2T)] \quad 5.6$$

It has 2 samples delay.

5.5.2.1.3 Nonlinear transform

Instead of squaring the output signal from the derivative filter point by point as stated in [13], this thesis rectifies the signals by taking their absolute values, thus reducing the gain sensitivities to improve the performance of the decision rule algorithm. The equation of this operation is

$$y(nT) = \sqrt{x(nT)^2} \quad 5.7$$

5.5.2.1.4 Moving Window Integration

The window size has to be taken properly, neither so wide that merges the QRS complex and T-wave together, nor so narrow that produces several peaks in the integration waveform. The proper window size will give waveform feature information besides the R-wave. It is calculated from [3],

$$y(nT) = (1/N)\ [x(nT - (N-1)T) + x(nT - (N-2)T) + \ldots + x(nT)] \quad 5.8$$

where N is the width of the integration window. Our system takes N as 30.

5.5.2.1.5 Decision Rule Algorithm

As seen in figure 5.5, the algorithm sets two thresholds T_1 and T_2 to make decisions. T_1 is set for the filtered ECG, and T_2 is set for the signals produced by the moving window

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integration. Thresholds T_1 and T_2 are running estimated from beat-to-beat in order to adapt to the rises and falls of R-waves' peak amplitudes. Processing delays are considered to estimate the average R-R intervals. If an R-wave is not detected after the maximum time interval, the algorithm will go back to a certain time interval to search for possible R wave candidate by using a new set of lower thresholds.

The above mentioned signal processing steps of the algorithm for a normal ECG signal are shown in Figure 5.6.

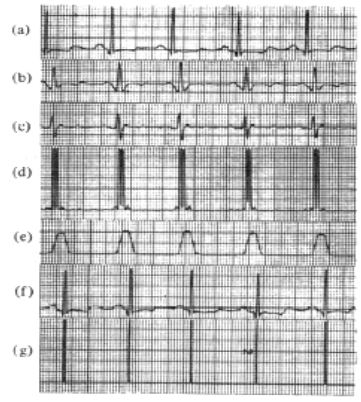

Figure 5.6 QRS detection algorithm processing steps for a normal ECG waveform. (a) Original signal (b) output of band pass filters (c) output of differentiator. (d) Output of squaring process (e) results of moving window integration. (f) Original ECG signal delayed by the total processing time. (g) output pulse stream.[13]

5.5.2.2 Power Spectrum Transformation

The power spectrum refers to the frequencies that contain the signal's power. The QRS complex of each beat is transformed to its power spectrum using the Fast Fourier Transform (FFT). The power spectrum is used to get the information about the point with maximum power in the ECG signal and the frequency at that point. A plot of frequency vs. power is obtained as a result of the power spectrum transformation. The power spectral analysis shifts the system to the frequency domain.

The power spectrum is normalized with a factor of 10. The frequency at the point of maximum power is recorded. It is observed that the frequency at maximum power for SVR is twice as much as that for VR that is, 1.2207 Hz and 0.6104 Hz respectively. This fact is used further in discriminating the origin of the ECG rhythms used.

$$f(\text{svr}) = 2f(\text{vr}) \qquad 5.9$$

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The algorithm for power spectrum transformation is given in the following.

Input: Vector *x*, containing ECG data

1. Take ~~Calculate FFTusing~~ next highest power of 2 ~~-~~greater than or equal to length(x).
2. Put input vector x through Hanning window.
~~1 and pad result with enough zeroes.~~
3. Take the magnitude of the FFT of x and pad result with enough zeroes to make it equal to the power of 2 calculated in step1.
~~.~~
4. Scale the FFT so that it is not a function of length of x.
5. Square the magnitude of FFT of x and multiply by 2.

~~2~~

~~3 Square the magnitude of FFT of x and multiply by 2~~

6. Get the unique DC and Nyquist component for

Figure 5.7 Algorithm for Power Spectrum Transformation

The output from the QRS detector is fed into the above mentioned algorithm. First of all, the FFT of the input vector is obtained. For achieving a fast FFT, the input vector, which is the vector containing the detected QRS values is padded with enough zeros to make its length a power of 2.

5.5.2.3 Linear Discriminator Analysis

The information of the frequencies obtained from the power spectrum is subject to the LDA for the origin discrimination. A linear discriminator is a classifier that uses a linear function of its inputs to base its decision on. That is, if the input feature vector to the classifier is a real vector $\vec{x}$, then the estimated output score or probability is

$$y = f(\vec{w} \cdot \vec{x}) = f\left(\sum_j w_j x_j\right), \quad j = 0,1,2,3, \ldots, n \qquad 5.10$$

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where $\vec{w}$ is a real vector of weights and *f* is a function that converts the dot product of the two vectors into the desired output. Often *f* is a simple function that maps all values above a certain threshold to "yes" and all other values to "no".

For a two-class classification problem, LDA is visualized as splitting a high-dimensional input space with a hyperplane: all points on one side of the hyperplane are classified as "yes", while the others are classified as "no". LDA is often used in situations where the speed of classification is an issue, since it is often the fastest classifier, especially when $\vec{x}$ is sparse. Figure 5.8 illustrates the simple function of LDA. The discriminator has *d* inputs each corresponding to the values of the components in an input vector. Each input value x_i is multiplied with its corresponding weight vector value w_i, so that the effective input at the output unit is the sum of all of these products, $\sum w_i x_i$. Each of the d-input units is linear, emitting exactly the value of its corresponding feature value. The single bias unit always emits the constant value 1.0. The single output unit emits,

+1 if $w_i.x_i+w_0 > 0$

-1 otherwise

Our system uses linear discriminator with two inputs, that is, power at 0.6104 and 1.2207 Hz. The discriminator calculates the weighted summation of the spectrum's power values and discriminates the origin as SVR = 0 and VR = 1 with a threshold of 0.5. The weights are calculated by a least square method so as to minimize squared error between the real diagnosis results and the classified outputs [2]. The training phase is accomplished by subjecting signals of known target values to the linear discriminator. The weights obtained after training are used for classifying origins of the test signals.

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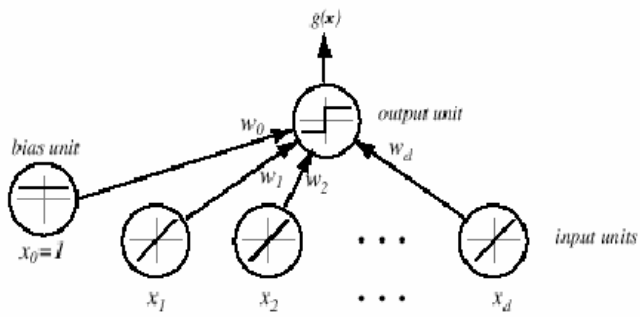

Figure 5.8 A simple LDA with *d* inputs.

The least square method is an adaptive technique which uses the gradient-based method of steepest descent. This method uses the estimates of the gradient vector from the available data. It incorporates an iterative procedure that makes successive corrections to the weight vector in the direction of the negative of the gradient vector which eventually leads to the minimum mean square error.

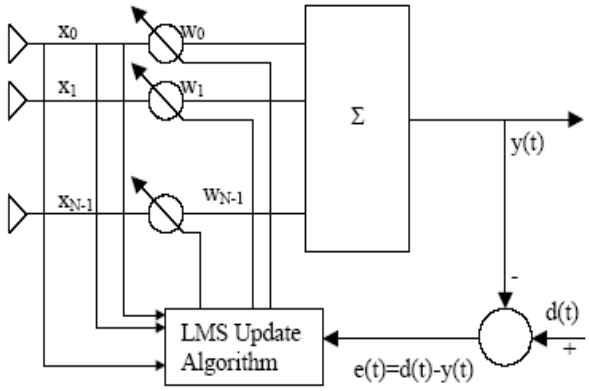

Figure 5.9 Block Diagram of LDA Algorithm

The algorithm developed for the LDA is as follows.

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Inputs: Vector *X [x0, x1, x2]*, power at frequencies 1.2207 and 0.6104Hz.

Weight vector *W [w0, w1, w2]*, to be generalized over training data.

1. define minimum acceptable error
2. define μ
3. initialize W
4. while *true*

y = 0

for i = 1: length (X)

y = y + X_i*W_i

if (y > 0)

then output = 1

elseif (y < 0)

then output = 0

Calculate error as tg – y

Check for any need for feedback by comparing with minimum error,

if (tg – y) > (minimum acceptable error)

then modify weight vector

for i = 1: length (W)

4.5.1 $W_i = W_i + (\mu * X_i *$ error);

else

Figure 5.10 Algorithm of LDA

Where μ is the step-size parameter and controls the convergence characteristics of the algorithm. The important equations used are:

$$\text{Output, } y = y + X_i * W_i \quad (5.11)$$

The error value calculated from the difference of the obtained output and the desired output (in training process) is given by:

$$\text{error} = \text{target value} - \text{output} \quad (5.12)$$

The weight update function is:

$$W_{i+1} = W_i + (\mu * X_i * \text{error}) \quad (5.13)$$

5.6 Module 2: *ARRHYTHMIA CLASSIFIER*

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After the separation of rhythm origin into supra-ventricular or ventricular by using the waveform of QRS complexes, the system classifies arrhythmia into 12 types according to the current heart rate (HR) and its variation from the preceding HR.

The heart rate is a non-stationary signal, and its variation can contain indicators of current disease or warnings about impending cardiac diseases. The indicators can be present at all times or can occur at random, during certain intervals of the day. However, to study and pinpoint abnormalities in large quantities of data collected over several hours is strenuous and time consuming. Hence, heart rate variation (HRV) measurement has become a popular and non-invasive tool for assessing the autonomic nervous system. Computer based analytical tools for the in-depth study and classification of data over day-long intervals can be very useful in diagnostics.

The 12 types of arrhythmias detected by the system include half of arrhythmias that belong to supra-ventricular arrhythmias and half that belong to ventricular arrhythmias. They are as follow

Supra-ventricular Arrhythmias

1. Premature atrial contraction (PAC)
2. Atrial fibrillation (AF)
3. Supra-ventricular run (Svrun)
4. Supra-ventricular tachycardia/flutter (SVT)
5. Sinus rhythm (SR)
6. Sinusbradycardia (Sbrady)

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Ventricular Arrhythmias

7. Premature ventricular contraction (PVC)
8. Ventricular Fibrillation (VF)
9. Ventricular run (Vrun)
10. Ventricular tachycardia/flutter (VT)
11. Junctional beat (JB)
12. Complete AV block (AVB)

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5.6.1 Heart Rate (HR)

Heart rate is a term used to describe the frequency of the cardiac cycle. It is considered one of the four vital signs. Usually it is calculated as the number of contractions of the heart in one minute and expressed as "beats per minute" (bpm)

Normal heart rate ranges from 60-100 bpm. In case of tachycardia heart rate becomes greater then 100bpm and in case of bradycardia ,heart rate becomes lesser than 60bpm.

5.6.1.1 Calculation of Heart Rate

The interval between two successive QRS complexes is defined as the R-R interval or N-N interval and the heart rate (beats per minute) is given as

$$HR = 60/R\text{-}R \qquad (5.14)$$

where HR is the instantaneous heart rate measured in bpm, R-R is the R-R interval measured in seconds (s). For example, if R-R is 0.8 s, the heart rate is 75 bpm. The R-R interval should be relatively constant from beat to beat. A changing R-R interval indicates irregular heart rate.

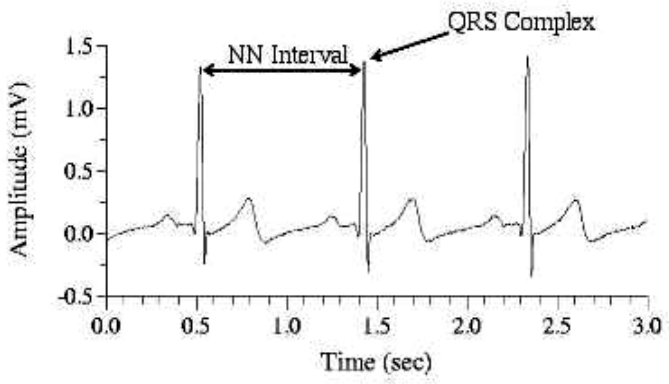

Figure 5.11 N-N interval or R-R interval shown on ECG

Since ECG data is sampled at a sampling rate of 250 Hz and our system detects the R-R interval with respect to samples so the heart rate is calculated as

$$HR = 60/\text{sampling interval}*\text{samples} \quad \text{bpm} \qquad (5.15)$$

Where sampling interval is reciprocal of sampling rate

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Sampling interval = 1/sampling rate 5.16

= 1/250

= 0.004 sec

5.6.2 Heart Rate Variation (HRV)

Heart rate variation refers to the beat-to-beat alterations in heart rate. Since abnormal ECG contains important pointers to the nature of diseases afflicting the heart, however, bio-signals being non-stationary, these pointers can occur at random on the time scale. Therefore, for effective diagnostics, the study of ECG patterns and HRV signals may have to be carried out over several hours. HRV is a non-invasive measurement of cardiovascular autonomic regulation. Specifically, it is a measurement of the interaction between sympathetic and parasympathetic activity in autonomic functioning.

5.6.2.1 Calculation of HRV

HRV is usually calculated by analyzing the time series of beat-to-beat intervals from ECG. We define the beat-to-beat HRV, as

HRV=| current HR – preceding HR | / current HR 5.17

~~5.6.2~~5.6.3 ~~S~~teps Involved in Arrhythmia Classifier

The arrhythmia classifier is based upon two major steps that further simplify the classification process. The separation of regular and irregular rhythms segregates the waveform to produce a more accurate diagnosis.

5.6.3.1 Separation of Regular from Irregular Rhythms by Using HRV

At first the arrhythmia classifier separates the physiological or regular rhythms from irregular ones by observing HR variation. While HRs fluctuate in normal beats, their variations do not exceed a certain maximum. In studies in [2], the maximum variation from the preceding beat (beat-to-beat variation) was less than 18 bpm with a 99% confidence level. Similar trends also appear in other regular pathological rhythms like tachycardia and flutter.

5.6.3.1.1 Applying Variance Thresholds

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The variance thresholds, ATHD and VTHD, dividing regular and irregular rhythms are set by the system for SVR and VR, respectively. An ATHD of 20% was chosen, which is more than the maximum variation in normal beats (18 beats @ 70 bpm), for accommodating typical arrhythmias seen in ICD/AED and pacemaker therapies. VTHD should be less than ATHD. Ventricular rhythm is adjusted by sympathetic cardiac nerves, not by SA node, and its HR variation is slow. The threshold for VR is set to VTHD=8%.

In the case of VR, if HRV is more than VTHD, the system recognized that beat as an irregular rhythm such as premature ventricular contraction (PVC) and fibrillation (VF). Otherwise, system referred it as a regular rhythm such as flutter, tachycardia (VT), junctional rhythm, or idioventricular rhythm.

Similarly in case of SVR case if HRV is more than ATHD, the system recognized the beat as an irregular rhythm such as premature atrial contraction (PAC) and fibrillation (AF). Otherwise, system referred it as a regular rhythm such as flutter, tachycardia, SVrun rhythm, or sinus bradycardia rhythm.

5.6.3.2 Classification of Beats by using Current HR

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After the separation of regular from irregular rhythms, the system classifies the beat to the appropriate class of arrhythmia. In cases of VR, idioventricular rhythms due to complete AV-block (AVB), junctional rhythm (JB), and tachycardia (VT and Vrun), are separated by HR boundaries of 40 and 120 bpm, respectively. Due to a similarity in the clinical treatment, flutter is classified as tachycardia. In cases of SVR, HR boundaries of 40 and 120 bpm are set to separate Sinus rhythm, sinus Bradycardia and tachycardia (SVT and SVrun) respectively.

5.6.4 HRV Analysis

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The time and frequency domain measures of HRV are analyzed by the system. The standard deviation of all normal R-R intervals (SDNN) and the difference between the maximum hourly HR variability (circadian rhythm) are computed as standard time-domain measures of HRV. Spectral power has been quantified by Fast Fourier Transform analysis.

5.6.4.1 Time Domain Analysis

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The basis of time domain methods is either the heart rate at any point in time or the intervals between successive complexes. The system analyzes the HRV statistically and geometrically.

5.6.4.1.1 Statistical Methods

From a series of instantaneous heart rates or cycle intervals, statistical time-domain measures are divided into two classes:

- Those derived from direct measurements of the instantaneous heart rate or RR intervals.
- Those derived from the differences between RR intervals

The system calculates the following variables using the statistical methods:

- Mean heart rate (HRmean)
- Max heart rate (HRmax)
- Min heart rate(HRmin)
- Mean heart rate variation(HRVmean)
- Mean RR interval (RRmean)

5.6.4.1.2 Geometrical Methods

The series of RR intervals can also be converted into a *geometric pattern*, such as the sample density distribution of RR interval durations, sample density distribution of differences between adjacent RR intervals, poincaré plot of NN or RR intervals, etc., and a simple formula is used which judges the variability based on the geometric and/or graphic properties of the resulting pattern. Three general approaches are used in geometric methods:

- A basic measurement of the geometric pattern (e.g. the width of the distribution histogram at the specified level) is converted into the measure of HRV.
- The geometric pattern is interpolated by a mathematically defined shape (e.g. approximation of the distribution histogram by a triangle, or approximation of the

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differential histogram by an exponential curve) and then the parameters of this mathematical shape are used.

- The geometric shape is classified into several pattern-based categories which represent different classes of HRV (e.g. elliptic, linear and triangular shapes of poincaré plots).

➢ Tachogram Analysis

The system analyzes the tachogram by using the geometrical measures. In tachogram the successive RR interval values are plotted against the 'beat number'.

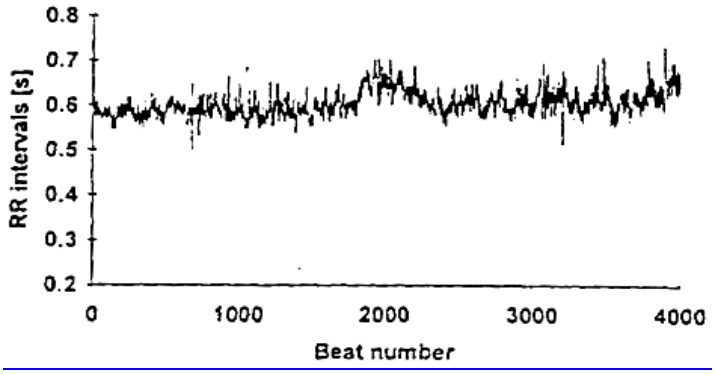

Figure 5.12 Tachogram

➢ Poincaré Plot Analysis

The system also performs the Poincaré plot analysis. It is an emerging quantitative-visual technique whereby the shape of the plot is categorized into functional classes that indicate the degree of the heart failure in a subject. The plot provides summary information as well as detailed beat-to-beat information on the behavior of the heart.

The Poincaré plot is a technique taken from nonlinear dynamics, portrays the nature of RR interval fluctuations. It is a plot in which each RR interval is plotted as a function of the previous RR interval.

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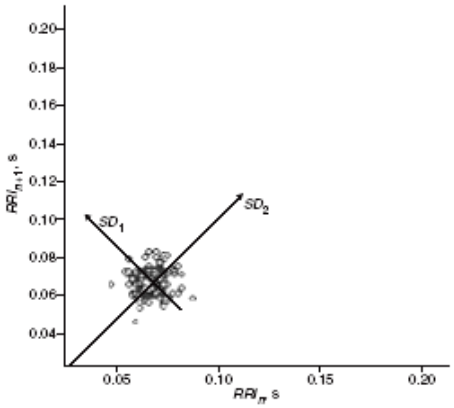

Figure 5.13 Poincaré Plot

The geometry of the Poincaré plot is essential and can be described by fitting an ellipse to the graph. The ellipse is fitted onto the so called line-of-identity at 45° to the normal axis. The standard deviation of the points perpendicular to the line-of-identity denoted by SD1 or minor axis describes short-term variability which is mainly caused by respiratory sinus arrhythmia (RSA). The standard deviation along the line-of-identity denoted by SD2 or major axis describes long term variability. Statistically, the plot displays the correlation between consecutive intervals in a graphical manner. Nonlinear dynamics considers the Poincaré plot as the two dimensional (2-D) reconstructed RR interval phase-spaces, which is a projection of the reconstructed attractor describing the dynamics of the cardiac system. The RR interval Poincaré plot typically appears as an elongated cloud of points oriented along the line-of-identity. The dispersion of points perpendicular to the line-of-identity reflects the level of short term variability. The dispersion of points along the line-of-identity is thought to indicate the level of long-term variability. The Poincaré plot may be analyzed quantitatively by calculating the standard deviations of the distances of the RR(i) to the lines y = x and y = -x+2*RRm, where RRm is the mean of all RR(i).The standard deviations are referred to as SD1 and SD2, respectively. SD1 related to the fast beat-to-beat variability in the data, while SD2 describes the longer-term variability of RR(i).

5.6.4.2 Frequency Domain Analysis

Various spectral methods for the analysis of the tachogram have been applied since the late 1960s. Power spectral density (PSD) analysis provides the basic information of how power

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(i.e. variance) distributes as a function of frequency. Independent of the method employed, only an estimate of the true PSD of the signals can be obtained by proper mathematical algorithms.

➢ Power Spectral Analysis

There are two methods to calculate PSD, classified as

- Non-parametric
- Parametric

The advantages of the non-parametric methods are:

- The simplicity of the algorithm employed (Fast Fourier Transform in most of the cases)
- The high processing speed

The advantages of parametric methods are:

- Smoother spectral components which can be distinguished independently of pre selected frequency bands
- Easy post-processing of the spectrum with an automatic calculation of low and high frequency power components and easy identification of the central frequency of each component
- An accurate estimation of PSD even on a small number of samples on which the signal is supposed to maintain stationarity. The basic disadvantage of parametric methods is the need to verify the suitability of the chosen model and its complexity (i.e. the order of the model).

The system calculates the power spectrum using non-parametric measures. In the frequency-domain analysis power spectral density (PSD) of the RR series is calculated. The PSD is analyzed by calculating powers and peak frequencies for different frequency bands. The commonly used frequency bands are very low frequency (VLF, 0-0.04 Hz), low frequency (LF, 0.04- 0.15 Hz), and high frequency (HF, 0.15-0.4 Hz). The most common frequency-domain parameters include the powers of VLF, LF, and HF bands in absolute and relative values, the normalized power of LF and HF bands, and the LF to HF ratio. For the FFT based spectrum powers are calculated by integrating the spectrum over the frequency bands.

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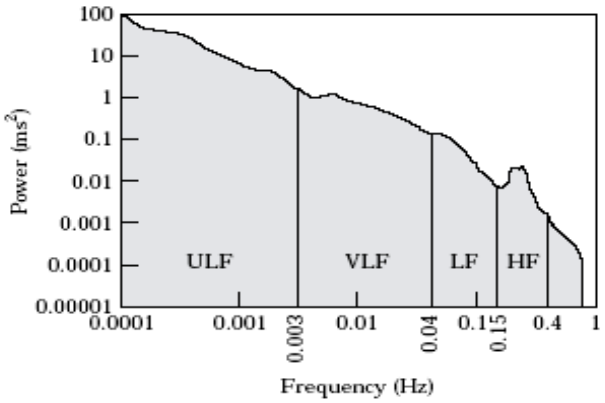

Figure 5.14 HRV Spectral Analysis

Finally, the classification flow is shown in figure 5.15.

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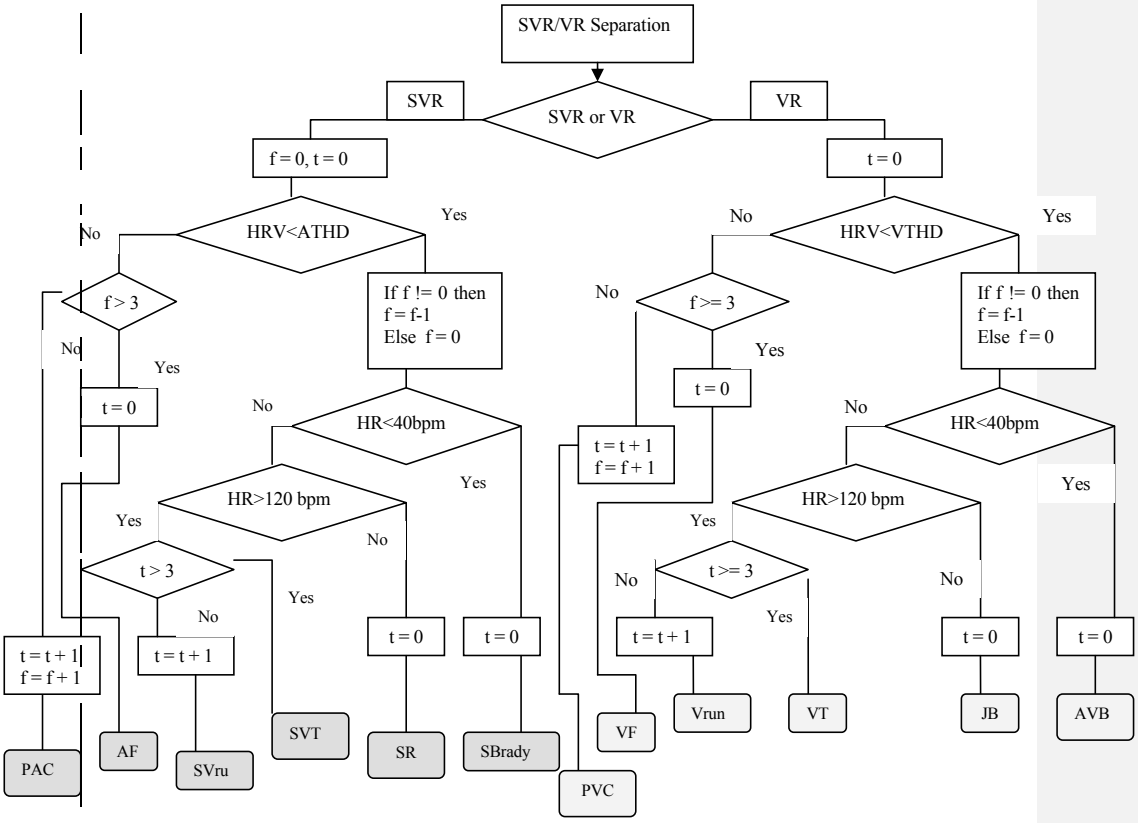


Figure 5.15 Flow Chart of Arrhythmia Classification

5.7 Summary

The design details are practically implemented in the system so that the set of requirements mentioned in chapter 4 can be met. The well defined modular structure under the incremental evolutionary model has proved a success to achieve the design goals.

Chapter 6

Graphical User Interface (GUI)

6.1 Introduction

For the development of the application like ECG beat recognition and arrhythmia classification the development of Graphical User Interface (GUI) was an important task. Hence a need for a user friendly GUI was required so that the user should not have any difficulty in carrying out the different operations required for ECG analysis without knowing what calculations and operations are carried out at the backend. A GUI is an ideal interface for this purpose. It provides a pictorial faceplate to streamline the extraction process. Users of all skill levels will already be familiar with GUI operation based on experience with the similar interfaces found on PCs, ATMs, etc. The GUI will lead users through the extraction process with a series of dialog boxes in conjunction with an accompanying text help file.

MATLAB is a technical computing program that is readily available and already widely used in many of the fields. It has the additional benefit of cross-platform compatibility. MATLAB® 7.0 was used throughout this process. However, this program is both backward and forward compatible for several versions, so that GUI will be accessible to the majority of MATLAB users regardless of program edition. Again, the development was undertaken iteratively.

The GUI was also revised with an eye to accessibility and ease of use. Several user-friendly features were added, such as the ability to browse for input files, zooming capability on the diagrams, and a wait-bar to provide feedback on the extraction's time to completion.

The Graphical User Interface is designed using the GUIDE toolbox (Graphical User Interface Development Environment) in Matlab. It comprises of following major portions.

1) Menu Operations
2) Training Signals drop down menu
3) Choose Signal and Path
4) Original signal Axes
5) Operations toolbar

6) R-Peaks and other operations Axes

7) Generate statistics

8) Check Arrhythmias detected

When turned on, the system starts with the following flash screen:

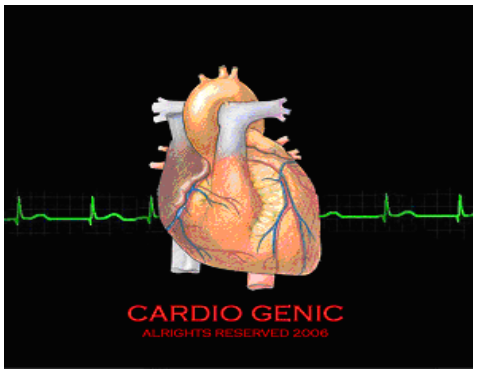


Figure 6.1 Flash Screen

Figure 6.2 shows the different parts of the GUI.

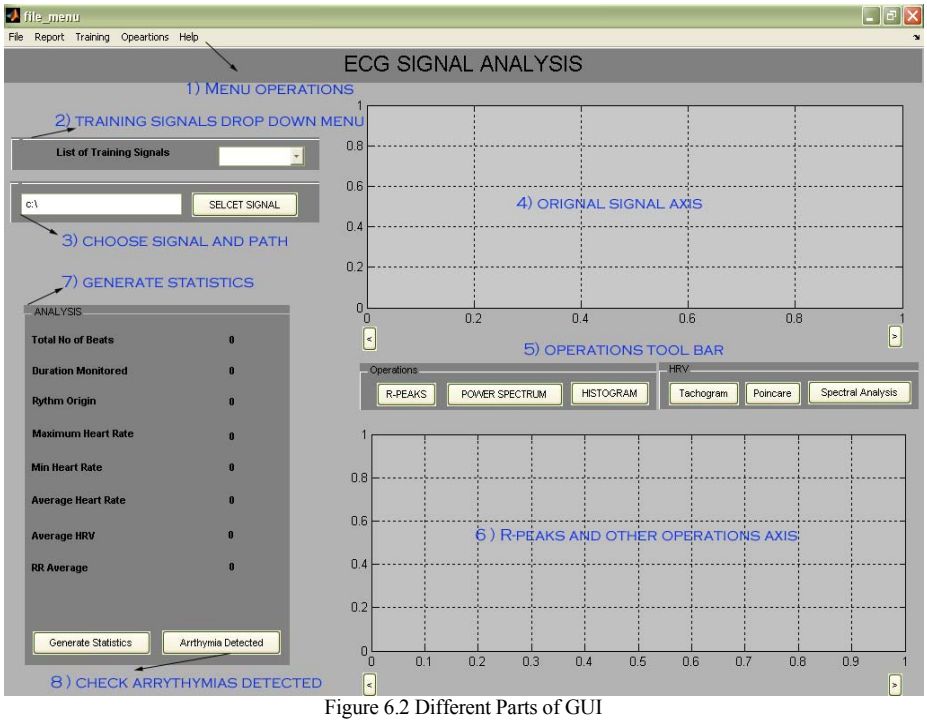


Figure 6.2 Different Parts of GUI

6.2 Operations in GUI

6.2.1 Menu Operations

The menu operations has following drop down menus.

a) File

File menu consists of Open and Quit options. Open is used to browse and open signals for analysis and it is an alternative to select signal button on the GUI. Quit is used to close the application.

b) Report

Report Menu consists of Save Report and View Report options. Save Report is used to save the report for currently analyzed case, and View Report is used to view previously saved reports.

c) Training

Training menu consists of Start training option. It allows the user to browser and select training archive in order to train the system before carrying out analysis on a signal.

d) Help

Help menu consists of version and types of arrhythmias.

6.2.2 Training signals drop down menu

Once the training of the system is complete the 'Training signals drop down menu' shows all the signals that were used in the training of the system.

6.2.3 Choose Signal and Path

Once the training of the system is complete the user can select the signal by browsing using a file browser once the signal is selected the complete path of the selected signal is shown.

6.2.4 Original Signal Axes

The original signal axes shows the original signal loaded after the user has selected the signal the user can zoom in and zoom out the signal using key board shortcuts ctrl+x, ctrl+X, ctrl+y, ctrl+Y, ctrl+z, ctrl+Z for zoom in and zoom out along x, y and z axes subsequently.

6.2.5 Operations Toolbar

The Operations toolbar consists of following buttons

a) R-Peaks: Shows the filtered ECG and R-Peaks marked by red circle.
b) Power Spectrum: Shows the power spectrum of the selected signal.
c) Histogram: Shows the Histogram of the power spectrum.
d) Tachogram: Shows the Tachogram graph of the ECG.
e) Poincaré: Shows the Poincaré graph against RR intervals and their frequency.
f) Spectral analysis: Shows the power spectrum of the HRV.

6.2.6 R-Peaks and Other Operations Axes

This axes draws the graphs for all the operations of the 'Operations toolbar'.

6.2.7 Generate Statistics

Displays the different statistics and calculations that are carried out during the analysis of the signal such as total number of beats, Duration of the signal, Signal Origin etc

6.2.8 Check Arrhythmias Detected

Shows the list of all the arrhythmias that are found after the analysis of the complete signal and in how many number of beats that arrhythmia has occurred.

The GUI is designed in such a way that it is friendly to the user. For this purpose the GUI has been tested extensively followed by error messages so that new users do not face any kind of problems during the use of application.

6.3 Sequence of Operations

Select *Start Training* from Training Menu

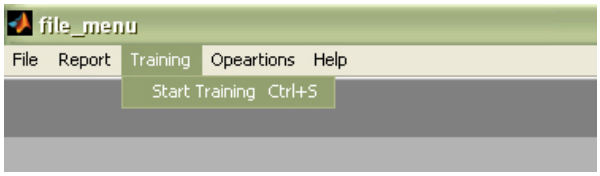


Figure 6.3 Initiating Training

Browse and select the training signals archive.

Figure 6.4 Selecting the Training Signals

The system automatically starts training and a progress bar shows the progress

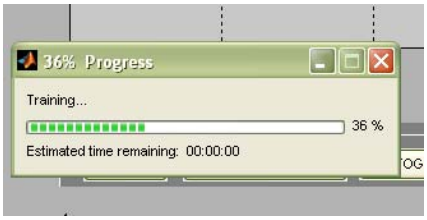


Figure 6.5 Training Progress

The Training signals are shown by a drop down menu.

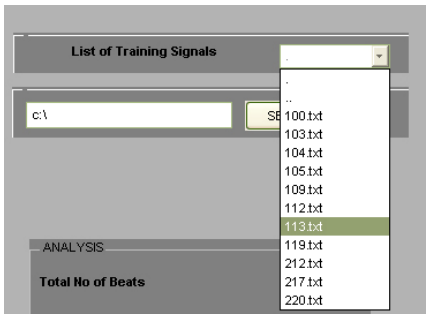


Figure 6.6 Training Drop Down Menu

Now select the signal that is to be analyzed by pressing *Select Signal* button

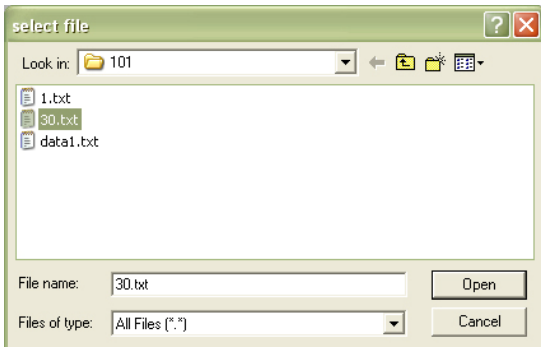


Figure 6.7 Selecting Signal for Analysis

The Original Signal selected is shown in axes1.

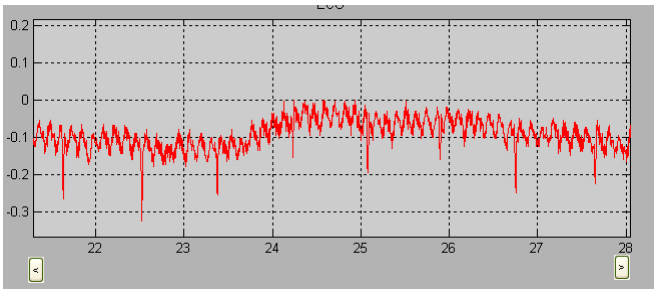


Figure 6.8 Original Signal without Filtering

Press *R-Peaks* button to show Filtered ECG and the R-Peaks shown by red circles.

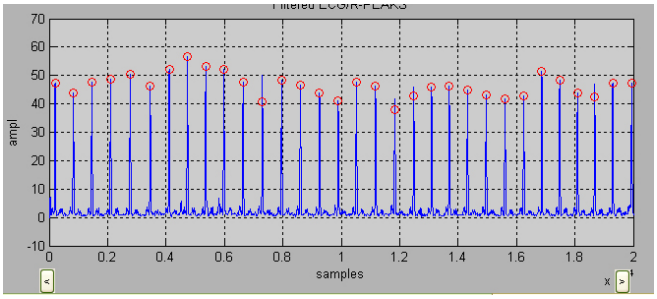


Figure 6.9 R-Peaks on Filtered ECG

Press *Power Spectrum* Button to show power spectrum of the signal.

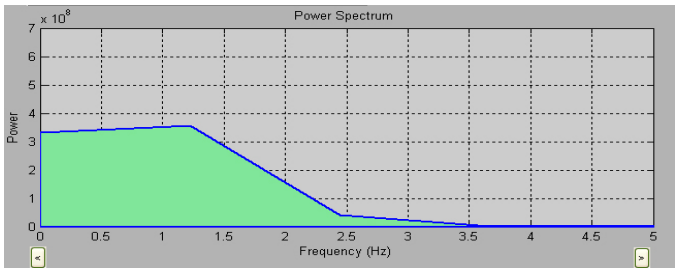


Figure 6.10 Power Spectrum

Press *Histogram* Button to show histogram of the power spectrum.

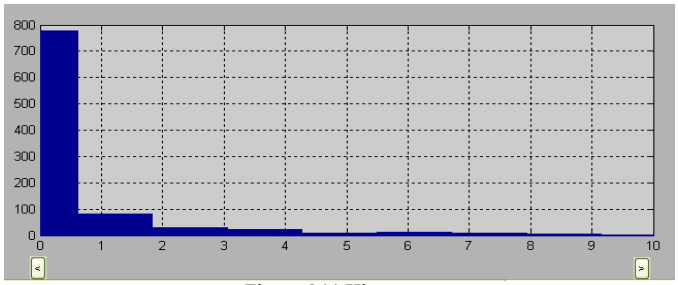


Figure 6.11 Histogram

Press the *Tachogram* button to show Tachogram graph which is plotted with RR intervals and beats.

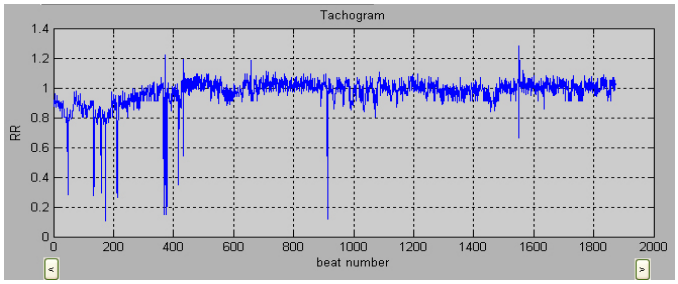


Figure 6.12 Tachogram

Press *Poincaré* button to show scatter graph of RR intervals.

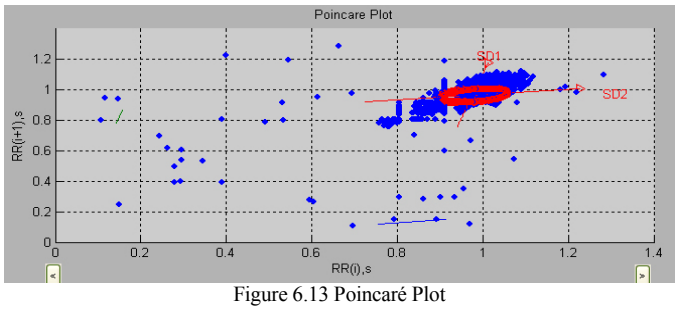


Figure 6.13 Poincaré Plot

Press *Spectral Analysis* button to show power spectrum of HRV.

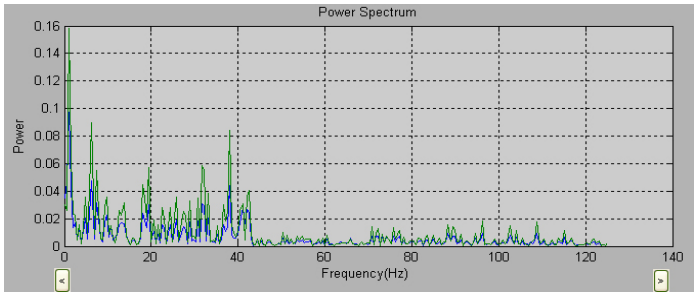


Figure 6.14 Power Spectrum of HRV

Press *Generate Statistics* button to generate the statistics as shown in the figure.

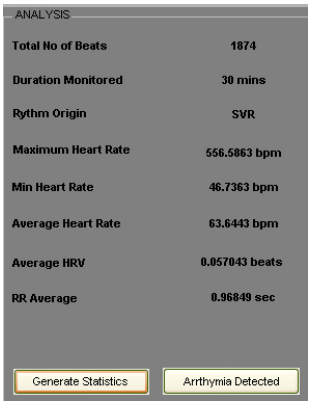


Figure 6.15 Calculated Statistics

Press *Arrhythmia Detected* Button to show the types of arrhythmias detected and number of times each arrhythmia occurred.

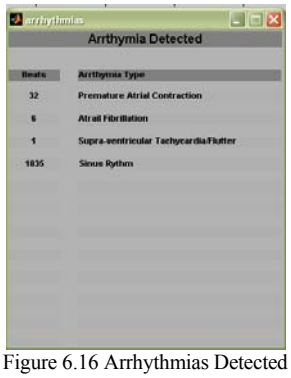


Figure 6.16 Arrhythmias Detected

Select save report from the Report menu to generate and save the report for current record.

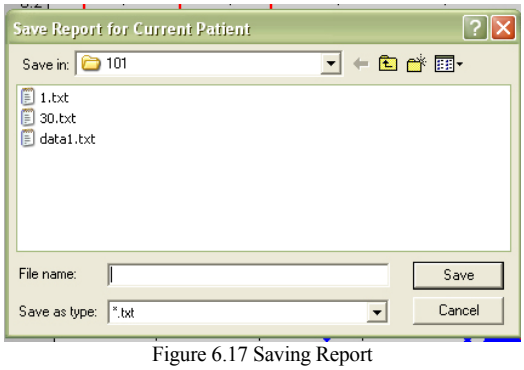
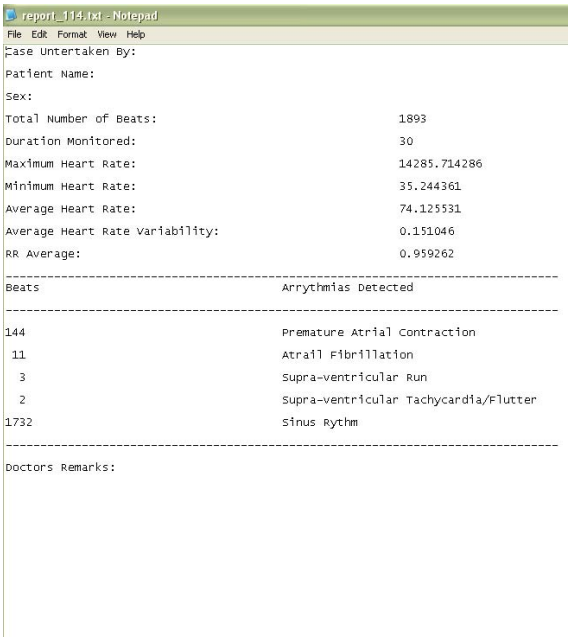


Figure 6.17 Saving Report

The saved report looks like this in the .txt file.


report_114.txt - Notepad

File Edit Format View Help

Case Untertaken By:

Patient Name:

Sex:

Total Number of Beats:	1893
Duration Monitored:	30
Maximum Heart Rate:	14285.714286
Minimum Heart Rate:	35.244361
Average Heart Rate:	74.125531
Average Heart Rate Variability:	0.151046
RR Average:	0.959262

Beats	Arrythmias Detected
144	Premature Atrial Contraction
11	Atrail Fibrillation
3	Supra-ventricular Run
2	Supra-ventricular Tachycardia/Flutter
1732	Sinus Rythm

Doctors Remarks:

Figure 6.18 Saved Report

The report is printable and can be printed after viewing.

6.4 Summary

One of the aims of the project is the development of a user friendly GUI which has been largely accomplished in the details mentioned details. The GUI incorporates all the issues that were expected and is very flexible to generalize even further with newer requirements.

Chapter 7

Training

7.1 Introduction

Training is an essential matter in machine learning techniques just as one has been applied in the system presented. The training process can be hardwired but it has not been done so for the sake of practical demonstration of the training process. Hardwiring shall be more useful when the software is commercialized.

7.2 Training Process

As the system aims at intelligent automation, therefore, it has been trained before it can be used for testing or validation. The data for training has been taken from the MIT-BIH arrhythmia database. It is important to mention here that one complete signal recording available on the database is 30 minutes long. However, the length of signals' recording consumed for training is only 1 minute. This is done because the system performance remains the same for both the recording times and hence it is preferable to use shorter recording to avoid processing overhead.

The origin discriminator is the module involved in training as it contains the LDA which requires training intelligence to separate origins correctly. The values of the frequencies at maximum power point and the origins of the ECG signals used for training have been calculated in order to monitor the training process. In total, 11 signals with approximately 790 QRS complexes have been used for learning. Table 7.1 shows these signals and their corresponding values.

Record No.	Frequency at maximum power (Hz)	Origin
100	1.2207	SVR
103	1.2207	SVR
104	0.6104	VR
105	0	VR
109	0.6104	VR
112	0	SVR
113	1.2207	SVR
119	0	VR

212	0	SVR
217	0.6104	VR
220	1.2207	SVR

Table 7.1 Signals used for Training

The value of the minimum acceptable error for the LDA is set at 1.0exp-20. The convergence parameter μ is set to 0.5. The weight vector obtained after training is:

Trained weight vector, W = [0.9 1.0 -1.0]

It is observed that extensive training improves the system performance to a higher degree.

7.3 Summary

The system is tested thoroughly and it is revealed that increasing the set size of the training data increases the test performance of the system.

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Chapter 8

Testing and Analysis

8.1 Introduction

This chapter explains the test performance of the system both in terms of unit testing and integration testing. The processing overheads encountered are also explained.

8.2 Unit Testing

The unit testing of the system has been done for both the modules. The accuracy and classification rates have been calculated separately.

8.2.1 Module 1: Origin Discriminator

25 records with approximately 1750 QRS complexes are subject to testing of the linear discriminator. The length of the records used is 1 minute attributed to the same reason as mentioned in chapter 7. The signals along with actual values of the frequencies and origins are shown in the Table 8.1. This table is used for reference when testing is carried out.

Record No.	Frequency at Maximum Power (Hz)	Origin
101	1.2207	SVR
102	0.6104	VR
106	1.2207	VR
107	0.6104	VR
111	1.2207	SVR
114	1.2207	SVR & VR
115	1.2207	SVR
117	1.2207	SVR
124	1.2207	SVR & VR
200	0	SVR & VR
201	0	SVR & VR
202	1.2207	SVR & VR
205	0	SVR & VR
208	0.6104	SVR & VR
209	0	SVR
210	0	SVR & VR
214	0	SVR & VR
215	0.6104	SVR & VR

219	1.2207	SVR & VR
221	0	VR
222	0	SVR
223	0	SVR & VR
230	0	SVR
234	0	SVR & VR

Table 8.1 Signals used for Testing

The accuracy calculated after testing is 88% with 210 misclassified QRS complexes. Table 8.2 gives the evaluation results for the linear discriminator. The misclassified QRS complexes are from record no. 106, 221 and 231. All the three misclassified records belong to VR but are classified as SVR.

Type of Ectopy	No. of QRS's used	Misclassification number	Rate of classification	Rate of misclassification
SVR and VR				

Table 8.2 Evaluation Results for LDA

The sensitivity and specificity are calculated using the grid in Figure 8.1.Sensitivity is taken to be the probability of a positive test among patients with disease and Specificity is considered as probability of a negative test among patients without disease.

	SVR	VR
POSITIVE	a	b
NEGATIVE	c	d

Figure 8.1 Sensitivity and Specificity Grid

From figure 8.1.

$$\text{Sensitivity} = a / (a+c) \qquad 8.1$$

$$\text{Specificity} = d / (b+d) \qquad 8.2$$

Table 8.3 lists the sensitivity and specificity in SVR detection. Sensitivity in VR detection is equivalent to specificity in SVR detection. SVR's were detected perfectly and the specificity is sufficiently high (>80%).

SVR sensitivity (VR specificity)	SVR specificity (VR sensitivity)

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Table 8.3 Sensitivity and Specificity

The graphic representation of the accuracy is,

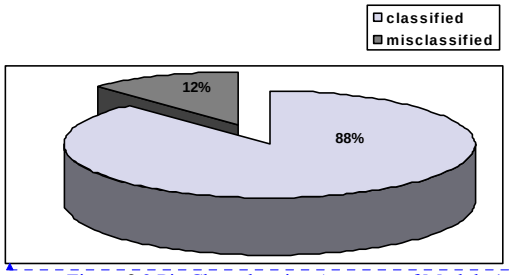

Figure 8.2 Pie Chart showing Accuracy of Module 1

8.2.2 Module 2: Arrhythmia Classifier

In the phase of arrhythmia classification, 20 records are consumed in testing. The arrhythmia classifier essentially requires the origin discriminator for its function; therefore, there is no difference between the unit testing of module 2 and the overall integration testing. The length of each record is 30 minutes, that is, each record comprises of approximately 2500 beats and each beat from each individual record is subject to arrhythmia classification. On the average, beat-to-beat accuracy has been found to be well above 75%. Table 8.4 shows the arrhythmia indications for tested records along with their numbers.

Record No.	Total No of Beats	Arrhythmia indication
101	1874	PAC 32 AF 6 SVT 1 SR 1835
102	2693	JB 1393 VF 455 PVC 845
107	4302	JB 30 Vrun 9 VF 4198
114	1892	SR 1732 SVT 2 SVrun 3 AF 11 PAC 144
115	1946	SR 1918

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		PAC 28
117	1537	SR 1520 PAC 11
124	1652	SR 1542 AF 44 PAC 66
200	3023	PAC 684 AF 1343 SVrun 54 SVT 28 SR 914
201	1974	AF 213 SVrun 43 SVT 6 SR 892 SBrady 18 PAC 775
202	2130	PAC 506 AF 91 SVrun 71 SVT 211 SR 1251
205	2662	PAC 81 AF 35 SVrun 9 SVT 13 SR 2524
208	3294	JB 1052 VT 12 Vrun 14 VF 1438 PVC 778
209	3020	PAC 299 AF 10 SVrun 36 SVT 229 SR 2446
214	2348	SBrady 1 SR 1526 SVT 1 SVrun 7 AF 166 PAC 647
215	3372	JB 2274

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		VT	17
		Vrun	127
219	2157	SBrady	35
		SR	1154
		PAC	791
		AF	177
222	2494	SR	1291
		SVT	110
		SVrun	136
		PAC	688
		AF	269
223	2601	SR	1814
		SVT	7
		SVrun	19
		AF	496
		PAC	265
230	2251	SR	2239
		PAC	12
234	2743	PAC	10
		SVrun	6
		SVT	19
		SR	2708

Table 8.4 Arrhythmia Classification for Test Signals

Table 8.5 gives the percent of accuracy w.r.t. the numbers of QRS complexes processed.

Total Number Of QRS Complexes	Number Of Correct Classifications	%age Classification	%age Misclassification
49965	39511	79%	21%

Table 8.5 Accuracy Results for Arrhythmia Classifier

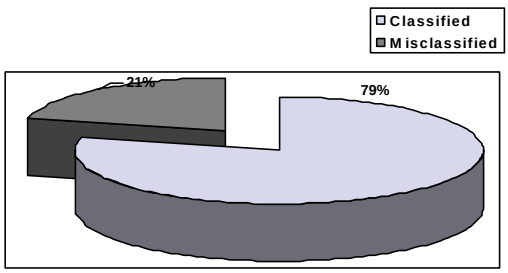

Figure 8.3 Pie Chart showing Accuracy of Module 2

8.3 Integration Testing

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The integration testing of the system along with the GUI has been accomplished successfully with a net accuracy percentage of 79%. The system has been developed in MATLAB 7.0 but its deployment is made possible as a standalone application and is explained in chapter 9.

8.3.1 Processing Overhead

Due to large data files, that is, 15.7 MB each, the system takes quite sometime to generate the output displays, e.g., on a Pentium 4, 1.5 GHz processor with 512 MB of RAM, it takes more than 4 minutes. This performance issue can be met with a faster processor having more RAM.

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Summary

It follows from the testing accuracy of the modules that increasing the training data and the use of appropriately high speed target machines shall render the system with a higher percentage of efficiency.

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Chapter 9

Deployment

9.1 Introduction

The deployment of the system as a standalone desktop application is a very important issue in practical software development projects. In the following, the deployment of the system as a standalone, environment independent desktop application is explained with a view of the possible deployment scenarios for the system.

9.2 Deployment process

The system has been developed in MATLAB 7.0 on a Windows XP platform. To make the system independent of the MATLAB environment, it has been converted to a standalone application using the MATLAB Compiler 4.0. Thus, the system is a redistributable software component which does not require MATLAB to be installed on the deployment machine. Anyhow, it is platform specific. Being a MATLAB Compiler-generated stand-alone, the system can be distributed to any target machine that has the same operating system as the machine on which the application was compiled. For example, if it is to be deployed to a Windows machine, it must be built on a Windows machine. However, any version of Windows will work. If the same application is to be deployed to a UNIX machine, it must use the MATLAB Compiler on the same UNIX platform and completely rebuild the application. To deploy an application to multiple platforms requires MATLAB and MATLAB Compiler licenses on all the desired platforms. The system can be distributed to a network drive so that it can be used on a network.

The guiding instructions for installing the system on target machine are encompassed in the user manuals that are provided with the package.

9.3 Summary

The system is deployable to any target machine that does not have the MATLAB 7.0 development environment installed on it; hence it is a standalone application. However, the system is still platform specific in the sense that it is deployable to a target machine that has

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the same operating system as the one where it has been compiled. An attractive feature is that the system is distributable to a network drive.

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Chapter 10

Future Enhancements and Conclusion

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10.1 Introduction

This chapter focuses on the conclusions drawn from the research work done and the future that is aimed on it.

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10.2 Conclusion

The arrhythmia diagnosis system, CARDIOGENIC, consists of SVR/VR discrimination using waveform analysis of QRS complexes and arrhythmia classification with HR variance. In our studies using human ECG data, the system provided high accuracy in the SVR/VR discrimination. Owing to the capability of distinguishing SVR from VR, diagnosis of specific arrhythmia can be carried out with a simple HR criterion.

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From the evaluation of the system, it can be concluded that CARDIOGENIC can be tailored to incorporate a wider range of arrhythmias along with its real time operation. The future work on the system shall make it compatible with the foreign manufactured hardware and hence it can lead to a research of practical importance in automated arrhythmia diagnosis which is a developing field in Pakistan. The project has met all of its planned objectives and hence it has been a success throughout.

10.3 Future Enhancements

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The research work and system implementation has been carried out in collaboration with Armed Forces Institute of Cardiology/National Institute of Heart Diseases (AFIC/NIHD), Rawalpindi. As mentioned in the previous chapters, the system; in its present form, is more likely to be validated with the standardized web-based data sources which accounts for its performance in comparison to the existing systems validated on the same criteria. The system performance is highly appreciable from a research perspective.

As a future enhancement to this work, it is intended that the system be validated with the ECG data available at AFIC so that the software can be of practical importance and of use to the collaborating organization. This will comply with an important objective of pursuing the project which is real time and practical use of the system. Though some effort to use

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data from AFIC has been made while beginning with the work but considering the time constraint on the system's development, it was left aside as a future enhancement.
Precisely, the data to be decompressed is the one from the holter recorders as discussed in chapter 5.

10.4 Summary

It is expected that the future aspects discussed shall steer the system to a commercial standard in the software industry.

Chapter 11

Achievements

11.1 Introduction

Certain research papers have been prepared in favor of the project and the research carried out. So far, the following achievements fall to our credit which have been presented as well.

11.2 National Platforms

- One paper titled "*Separating Origins in Arrhythmia Diagnosis*" has been presented in the 3rd International Conference on Electrophysiology held at AFIC/NIHD, Rawalpindi on 24th – 26th February 2006. This paper focused on the clinical details of the module 1 that is the origin discriminator.
- Another paper titled "*Designing Intelligent Algorithm for Origin Separation in Detection of Fatal Arrhythmias*" has been presented in the international symposium on "Meeting Health Challenges" held at Pakistan Institute of Medical Sciences (PIMS), Islamabad on 2nd - 5th March 2006. This paper focused on the clinical, engineering and algorithmic details of the origin discriminator.
- The system won the BEST STUDENT RESEARCH WORK award at the All Pakistan 3rd CIIT Workshop on Research in Computing, CWRC 2006 held at the COMSATS Institute of Information Technology, Wah Cantt on the 28th of April 2006.

11.3 International Platforms

- Our paper titled "*Arrhythmia Detection and Classification Using Rhythm Origin and Heart Rate Variation*", has been accepted and presented as a poster at the 3rd International Conference on Electronics and Computer (IKECCO2006) in Kyrgyzstan on the 12th – 13th April 2006. This conference had been sponsored by the University of Technology, Zurich, Switzerland.
- Our paper titled "*Significance of Rhythm Origin Separation in Arrhythmia Diagnosis*" has been accepted at the 3rd International Conference on Artificial

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Intelligence in Engineering and Technology (ICAIET2006) in Malaysia which is to be held on the 22nd – 24th November 2006.

- Paper titled "*Separation of Rhythm Origin in Arrhythmia Diagnosis*", has been accepted for presentation at the 2nd International Conference *"From Scientific Computing to Computational Engineering"*(IC-SCCE2006) which is to be held in Athens, Greece, from July 5th to July 8th, 2006.
- Our paper titled "*Detection and Classification of Cardiac Arrhythmias via Spectral Analysis and Origin Separation*" has been accepted for presentation at the IFMBE Young Investigator's Competition, World Congress on Medical Physics and Biomedical Engineering (WC 2006), Seoul, Korea. This platform is endorsed by the IEEE and is to be held from August 27 – September 1, 2006.
- Our paper *"Designing A Dual Threshold, Self Verifiable QRS Detection Algorithm For Arrhythmia Diagnosis"* has been accepted for oral presentation at the 18th International Conference on Systems Engineering *(ICSE2006)* which is to be held in Coventry, England, from Sept 5th to Sept 7th, 2006. This conference has been arranged by the IEEE section on Control Systems and Systems Engineering, UK.

11.4 Summary

The research work done has gained in stature and recognition as evidenced by the above. It is expected that it will gain even more recognition from reputable international platforms in the forthcoming month.

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APPENDIX A – FLOW CHARTS

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This appendix includes the flow charts of the algorithms discussed in chapter 5, that is, Design. The flow charts should be read in the order as they are given.

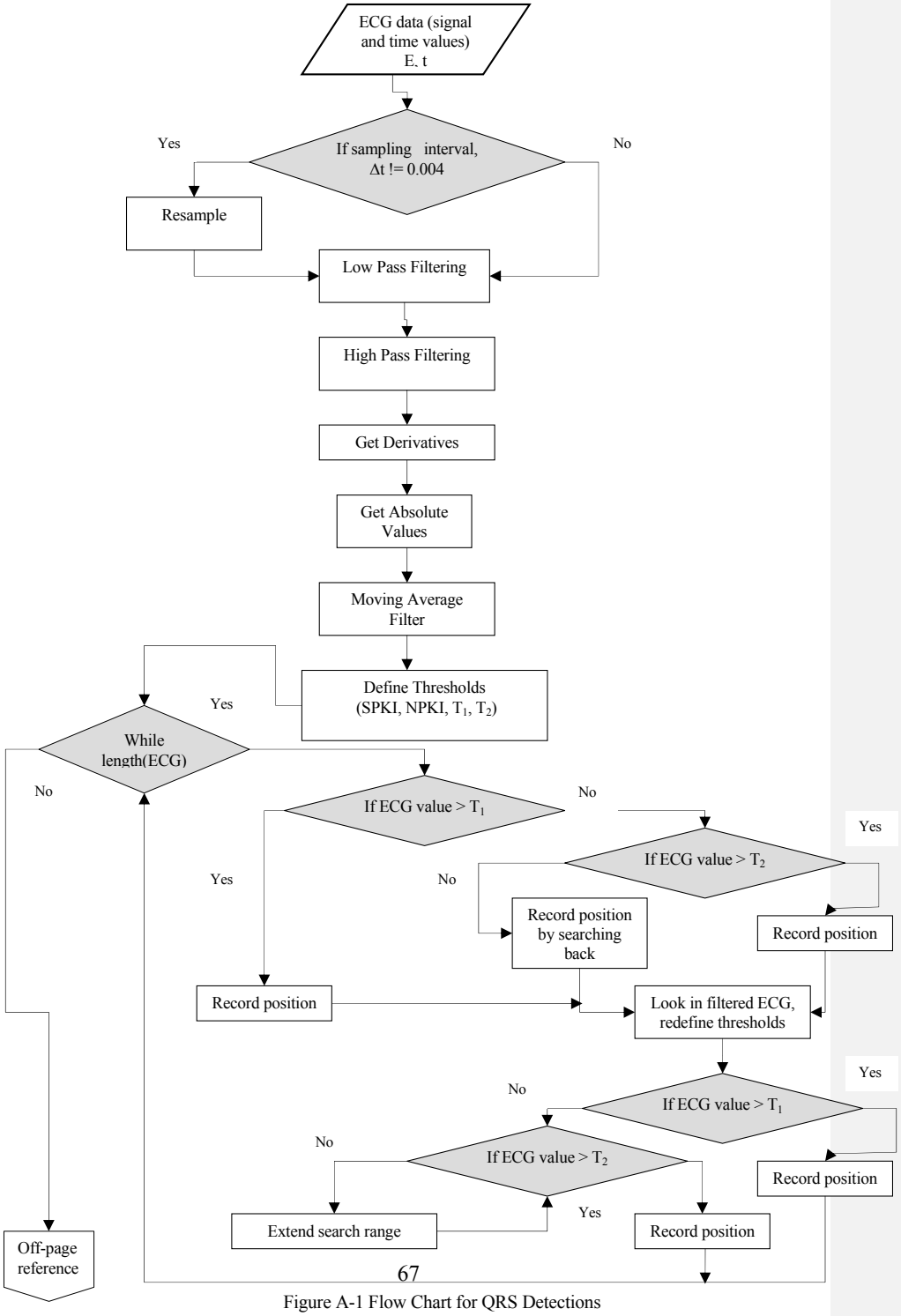

Figure A-1 Flow Chart for QRS Detections

Figure A-1 is given for regular heart rates, in case of irregular heart rate, the only change is in the threshold definition, that is,

For pre-processed ECG: $T_1 = 0.5T_1$

For filtered ECG: $T_2 = 0.5T_2$

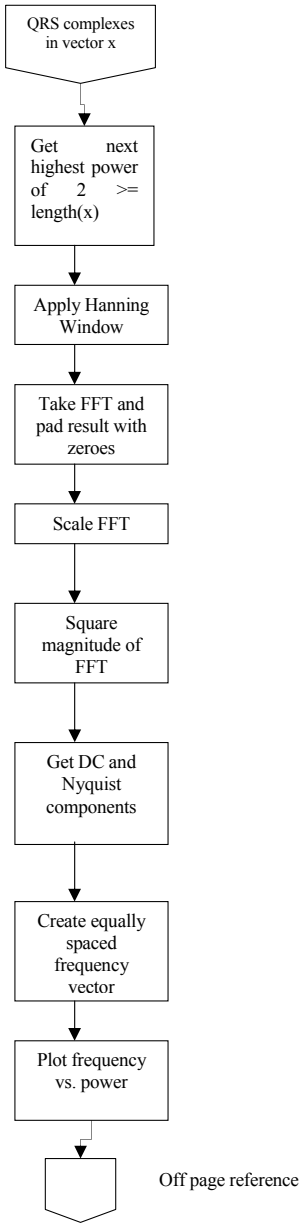

Figure A-2 Flow Chart for PST Algorithm

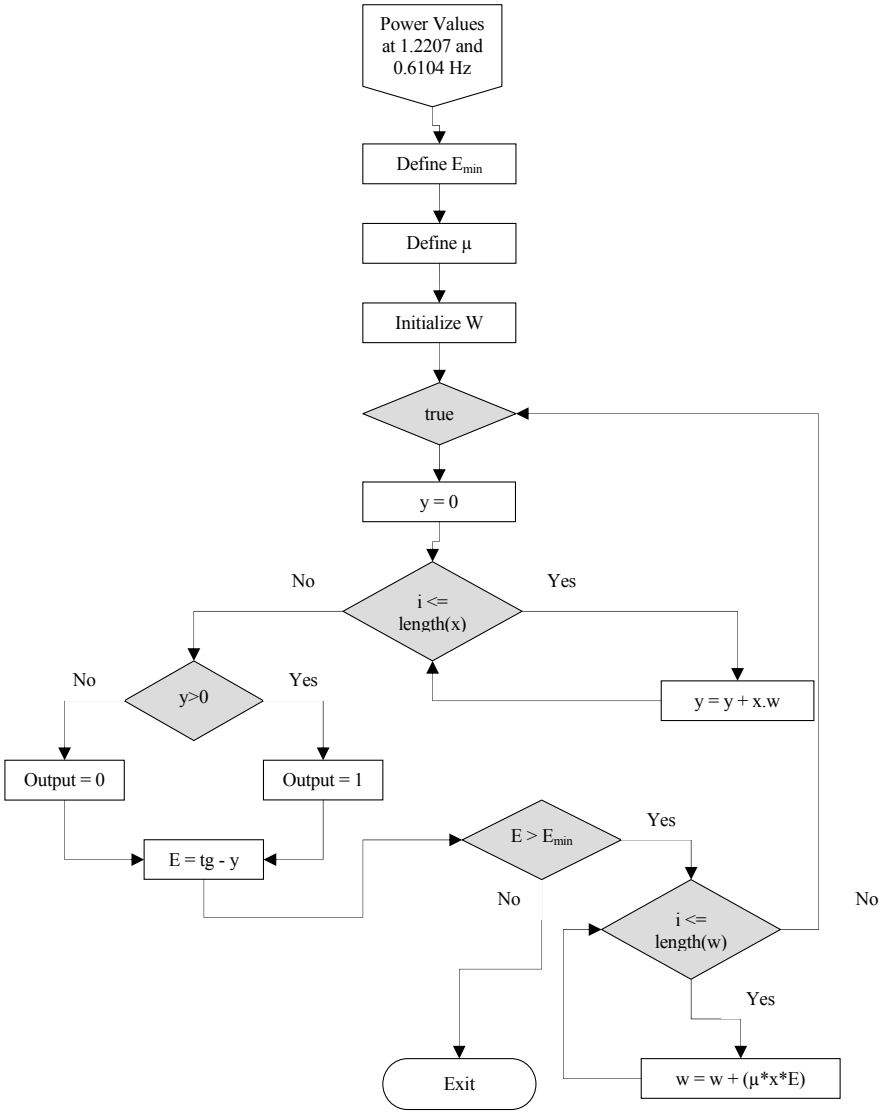

Figure A-3 Flow Chart for LDA Algorithm

APPENDIX B – USECASES AND SEQUENCE DIAGRAMS

This appendix includes the UML diagrams illustrative of the system. The diagrams should be read in the order in which they are given.

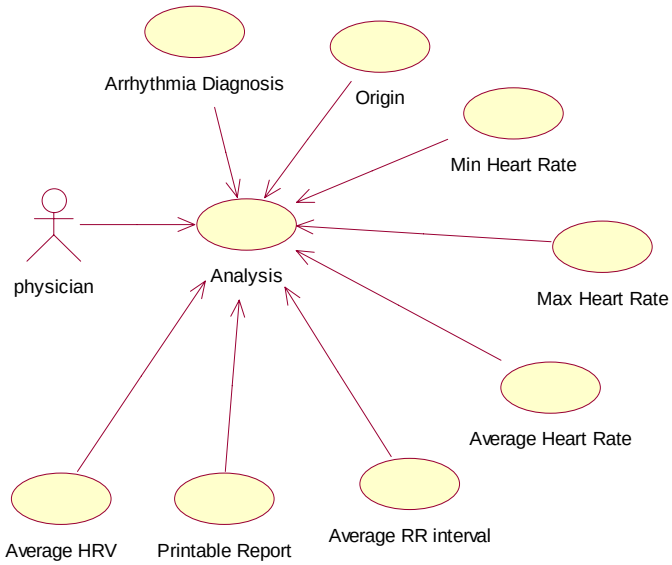

Figure B-1 Use Case View of System Functionality

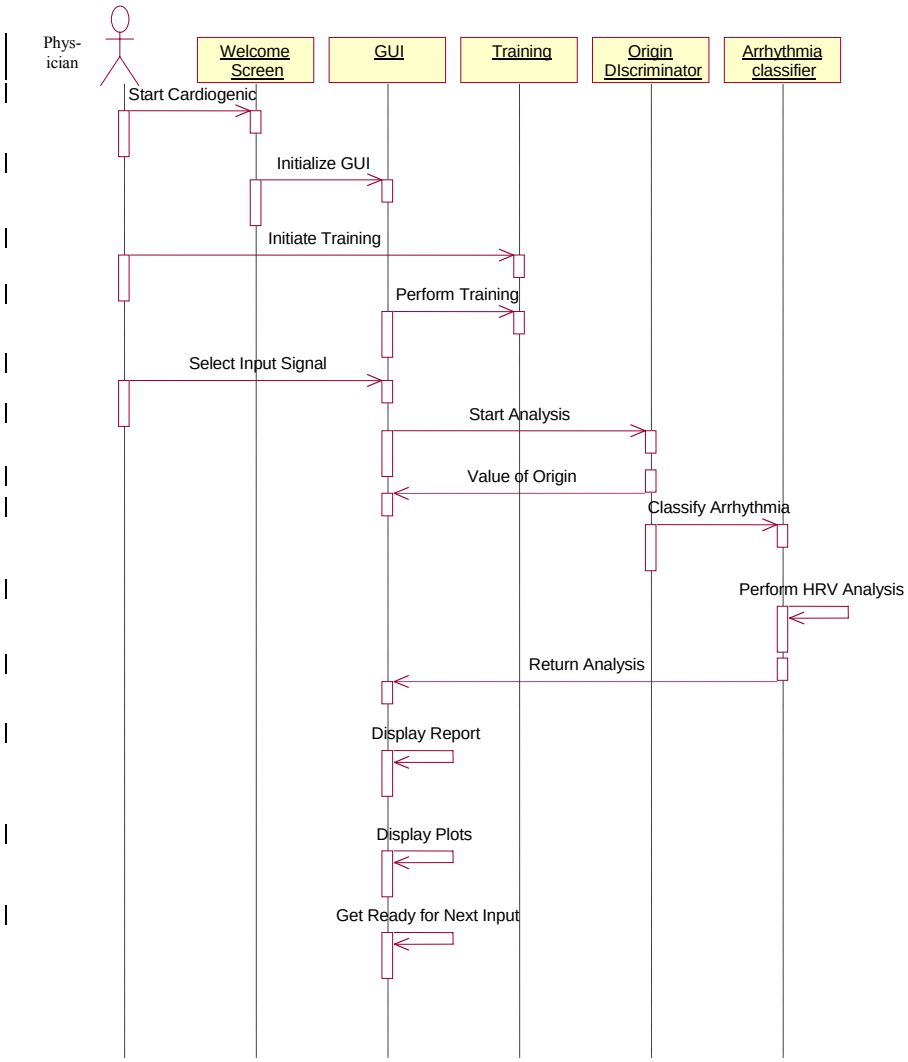

Figure B-2 Sequence Diagram for the Entire System

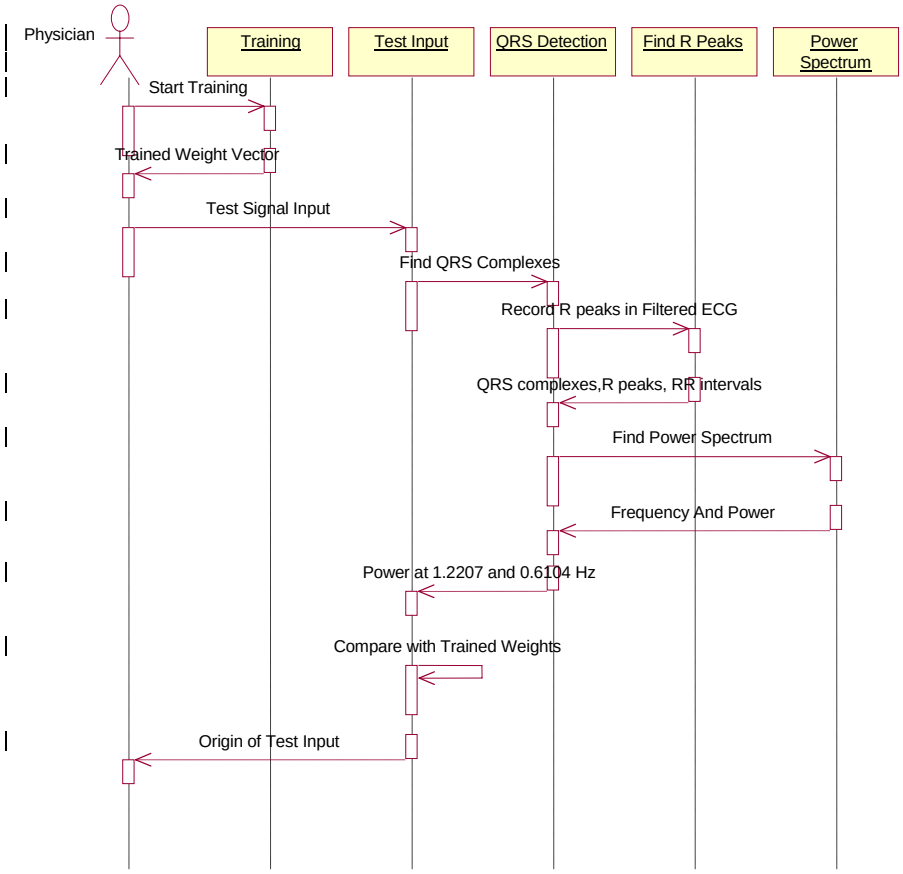

Figure B-3 Sequence Diagram for Origin Discriminator

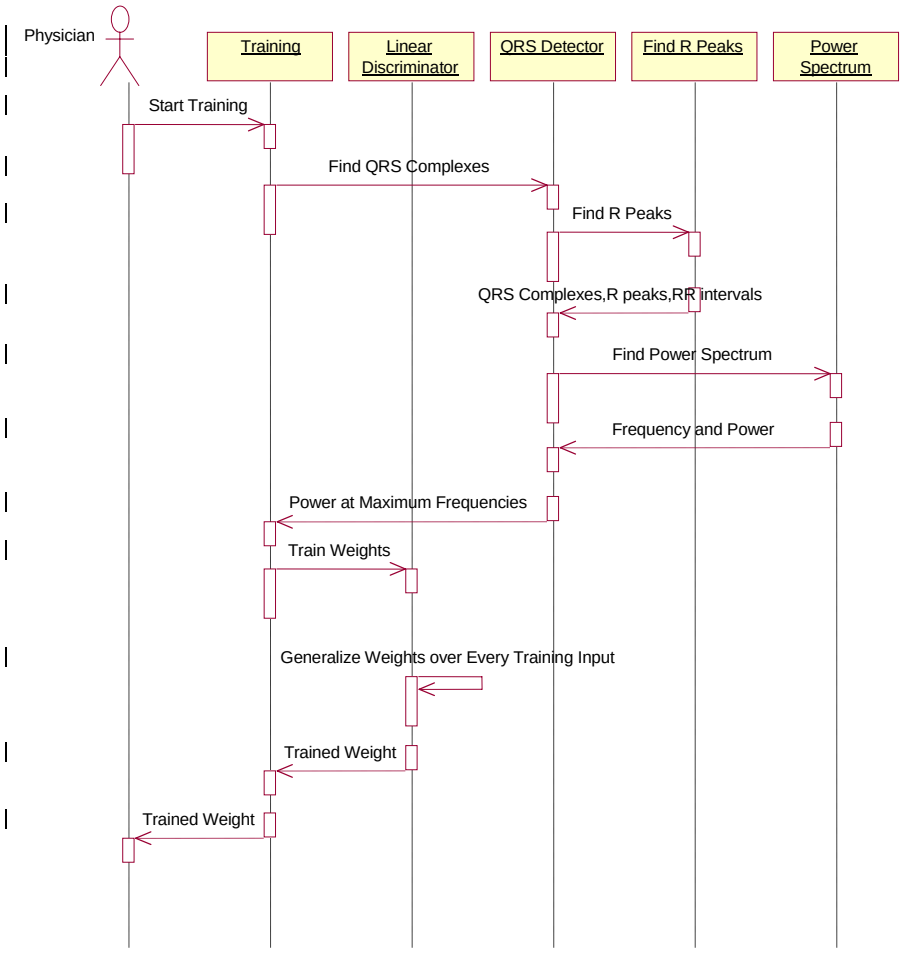

Figure B-4 Sequence Diagram for Training Process

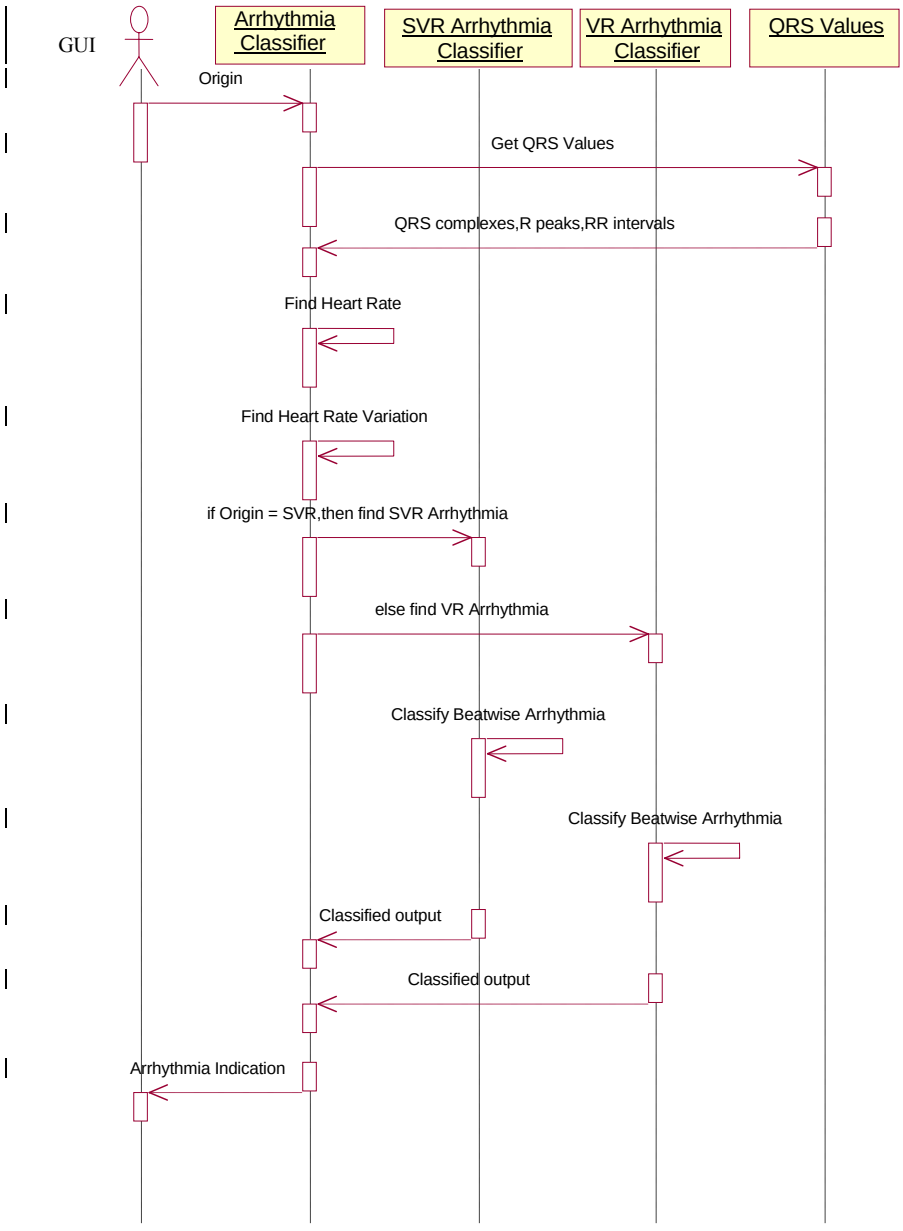

Figure B-5 Sequence Diagram for Arrhythmia Classifier

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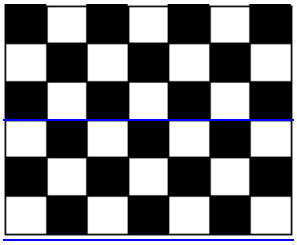

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