VECTORCARDIOGRAM ANALYSIS IN LABVIEW

A DISSERTATION

Submitted in partial fulfilment of the
requirements for the award of the degree
of
MASTER OF TECHNOLOGY
in
ELECTRICAL ENGINEERING
(With Specialization in Measurement & Instrumentation)

By

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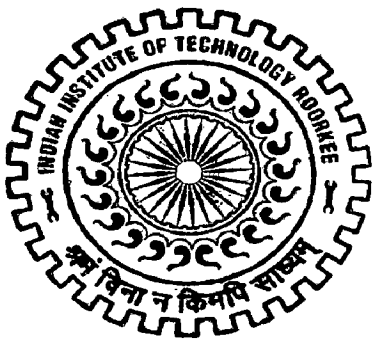

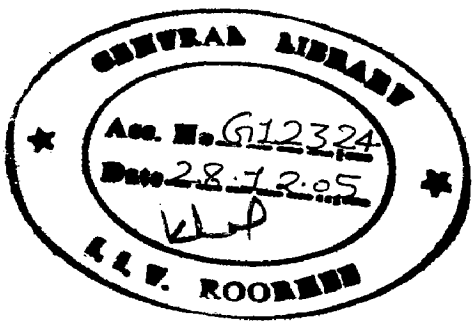

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JUNE, 2005

CANDIDATE'S DECLARATION

I hereby declare that the work that is being presented in this dissertation report entitled **"VECTORCARDIOGRAM ANALYSIS IN LABVIEW"** submitted in partial fulfillment of the requirements for the award of the degree of **Master of Technology** with specialization in **MEASUREMENT AND INSTRUMENTATION,** to the **Department of Electrical Engineering, Indian Institute of Technology, Roorkee**, is an authentic record of my own work carried out, under the guidance of **Dr. VINOD KUMAR**, Professor, Department of Electrical Engineering.

The matter embodied in this dissertation report has not been submitted by me for the award of any other degree or diploma.

Date: 30/06/05
Place: Roorkee

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This is to certify that the above statement made by the candidate is correct to the best of my knowledge.

30/6

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(DEEPU.C.VARUGHESE)

ABSTRACT

In the present Thesis some aspects of T-wave morphology are considered for characterization and quantification of heterogeneous repolarization, using the Vectorcardiogram (VCG) implemented in LabVIEW. The orthogonal Frank leads were synthesized from the standard 12 lead ECG. Several methods for synthesizing Frank VCG's from simultaneously recorded 12 standard leads has been studied and concluded that either a regression technique or a method based on the reverse of Dower's approach provide the best solution. From the derived X, Y, and Z leads, ventricular repolarization is defined from the start of S index to end of the cycle. The subtle changes in the course of T wave loop, possibly reflecting sudden changes in repolarization wave front is analyzed with the help of three parameters obtained from the VCG with two different methods of considering the zero point. The parameters are calculated for each cycle of waveform and averaged excluding the first and last detected R peaks. The obtained T–loop morphology parameters is studied on myocardial infarction and ischemia data sets and healthy database from the 'PTB diagnostic database' consisting of 15 lead data. The limits of the parameters of the healthy (42) from the myocardial patients (58) is distinguished. Cross-comparison between healthy and the myocardial groups is performed with t-test for all the parameters to evaluate the differences in mean between the two groups.

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CHAPTER 1
INTRODUCTION

The three-dimensional electrical vector is often of interest when studying cardiac diseases which involve conduction defects. The vector is the resultant of the electrical activity of the heart: a loop is formed by combining the instantaneous vectors during a cardiac cycle.

Loops from healthy individuals are usually quite similar in shape. Such loops start with a small extension frontally and continue left and downwards as the depolarization wave propagates to the apex. From the apex, the left ventricle dominates the loop due to its larger size compared to the right ventricle, making it turn backwards and upwards. The loop is closed after final extension in the right upward direction. Throughout the whole cycle, the loop is fairly planar due to symmetries of the heart. Although the size of the ventricles differs, the individual ventricle and atrium are rather symmetric canceling the contributions perpendicular to the symmetry plane. For certain pathologies, the VCG has proven to be a useful diagnostic tool.

A distinct feature of many pathologies is that the loop becomes less planar due to that the symmetry of the heart has changed. For example, in *myocardial infarction*, necrosis blocks the propagation through the diseased area. Due to the abrupt termination of the propagation wave in the infarct area, the VCG loop usually becomes angular in shape. Another example is *hypertrophy*, or increased heart mass, which may occur in one or both of the ventricles. Due to the increased mass, the electrical vector corresponding to the hypertrophic area increases. *Bundle branch block*, i.e., a blockage of the impulse in the conduction fibres which propagate the impulse to the ventricles, can be discerned from the VCG loop. Due to the block, the depolarization of the affected ventricle is delayed. Thus, in part, the blocked ventricle is not depolarized until the unblocked ventricle is repolarized. Since depolarization and repolarization result in electrical vectors with opposite directions, the block is discerned as a well-defined T wave with a direction opposite to that of the QRS complex. The result in the transversal plane is commonly an 8-shaped loop.

1.1 History of VCG

Augustus Waller was the first to record the electric activity of the human heart .He already observed that when recording the ECG on the surface of the thorax, the field behaves as if the source were a dipole. Recording the dipolar *x*-, *y*-, and *z*-components of the cardiac source is called the VCG. These three signals may be displayed as a function of time as a scalar display or one against another when they form vector loops.

Hubert Mann was the first to draw the VCG loop . He constructed it manually from the three signals of the Einthoven lead system. Because the Einthoven leads I, II, and III are in the frontal plane, only two of these are needed for the construction of the frontal plane loop. Hubert Mann later constructed a mirror galvanometer with three coils in 120◦ intervals around a mirror that projected the monocardiogram on photographic film. After J.B. Johnson of the Western Electric Company invented the low-voltage cathode ray tube , it became possible to display bioelectric signals in vector form in real time. This invention allowed the VCG to be used as a clinical tool. Wilson and Johnston were the first to add the third dimension to the VCG with a lead system called Wilson tetrahedron. This lead system was uncorrected; i.e., it did not consider the effect of the surface of the thorax and its internal inhomogeneities of the resistivity to the recording of the VCG. Several other uncorrected lead systems were later constructed by various scientists for recording the VCG.

Each cell in the heart can be represented as a dipole with differing direction throughout the heartbeat. A collection of all these small dipoles close to each other can be represented as a single dipole. The electric field of the heart can then be studied as a field of a single dipole, *the cardiac vector*. The line drawn by the tip of the cardiac vector is the vectrocardiogram (VCG). During a heart beat the tip draws a collection of three-dimensional loops, called a VCG complex or a VCG loop. Measurements on the body surface give information about the dipole inside. The measured signals are used to reconstruct the cardiac vector and the VCG. Represented as a function of time, the scalar signals are called electrocardiograms (ECG). The first electrocardiogram was obtained in 1889 by A. Waller. The technique was further developed by Einthoven, who chose to measure the heart activity as potential differences between the right arm, the left arm and

the left leg. Einthoven classified the complex into P-, QRS- and T-waves, where the P-wave is the depolarization of the atria, the QRS-wave is the depolarization of the ventricles and the T-wave is the repolarization of the ventricles.

Around 1904 Einthoven et al. derived the cardiac vector. Measuring three signals and drawing them into "Einthovens triangle" gives an approximate direction of the vector. Hubert Mann described, with the help of Einthovens signals, the change of length and direction of the cardiac vector as a function of time. This procedure takes long time and was therefore not adopted until 1930 when it became possible to present the obtained loops on an oscilloscope.

To get an accurate picture of the cardiac vector, it was necessary to use a system of measuring the ECG which gave orthonormal coordinates. One problem was that the potentials were measured relative to each other. First of all, potentials had to be measured relative to a common ground. But this still did not give a correct result since the body's electrical resistance had not been taken into account [Arvill; Webster, 1992; Jern, 1987; Pahlm, 1994]. Frank [Frank, 1953,-55], among others, looked for a solution to this problem by performing experiments on a torso model. The model was created of a homogeneous conducting medium, which can be considered a simplification of the human body. Frank came to a solution by measuring the potentials on the surface of the torso model containing a known dipole, and correcting the signals by introducing electrical corrections (like adding resistors between electrodes) and increasing the number of electrodes (leads). These corrections are related to the shape of the torso, the variability of the dipole location and the anatomic variation of the orientation of the heart [Frank 1956]. One of the major problems with Frank's system is that the measurement is very dependent of the placement of the electrodes. Frank's experiments have been compared with experiments on real patients. Humans give fuzzier and wavier lines, which is the result of the simplification made on the torso model. This method is considered to obtain the orthonormal coordinates of the tip of the cardiac vector at each instant. These coordinates are denoted x, y and z, (x is from right to left, y from the head to the feet and z is from the front to the back). Placing all measured points in an orthogonal coordinate system results in a vectorcardiogram.

1.2 Clinical Importance of VCG:

Scalar electrocardiograms have been used for the past hundred years and are still used today. Since the sixties computers have been used to analyze the ECG and to calculate the VCG. The enthusiasm for the VCG has oscillated a lot during the past 50 years, but the interest has been very large since the beginning of the nineties. [Pahlm, 1994]. Today's medical equipment uses the VCG for calculations and storage, but present the 12 lead ECGs (which are derived from the VCG). The reason why ECGs still are popular is that they are easy to present and doctors are trained to interpret them. Today's medical equipment usually does not show the VCG loops because there is no good way of presenting them. In some systems, VCG loops are presented as projections in different planes (x-y plane, x-z plane and y-z plane—the projection seen from the side, from the front and from above).

A study was made [Dellborg, 1991] to investigate vectorcardiography as a method for monitoring acute myocardial ischemia (lack of oxygen in the heart muscle) and came to the conclusion that the VCG has a great advantage of being able to collect information about all changes of the QRS complex in a single parameter. Dellborg also found that dynamic VCG is a user-friendly clinical tool and a good tool for studying ischemic heart disease in man. At the present moment though, it is difficult to say if VCG provides a better diagnosis than ECG. It is more probable that optimal performance can be obtained by combining VCG and scalar ECG (each method presenting information neglected by the other).

1.3 VCG Acquisition

Generally, two different approaches can be taken to acquiring the VCG, namely, direct recording or synthesis from another lead set. The common lead configuration for direct recording is the orthogonal Frank lead system. Initially, the VCG was used due to the less costly recording equipment, however, it soon became evident that the loop representation over some valuable clinical features. The electrode positions used for the Frank lead system (X, Y and Z) are defined to be uniform on the three anatomic planes of the torso, i.e., the horizontal, frontal and (left) sagittal planes.

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Since modern equipment has been developed for acquisition of the 12-lead ECG, methods have been developed for synthesizing the VCG from the ECG . The methods are based on linear projection with certain constraints. Studies have been made which evaluate different methods by comparing the synthesized loop to the corresponding VCG recording using, e.g., the Euclidean distance. It was found that methods based on regression or the inverse Dower matrix exhibit the better performance in synthesizing the Frank VCG.

1.4 Organization of the Report:

This chapter gives an overview and clinical importance of Vectorcardiogram.

Second chapter deals with Cardiac Physiology and diseases associated with it.

Third chapter deals with the VCG Lead system and different methods for deriving VCG from 12 lead ECG.

Fourth chapter discusses about the T- Loop morphology in Myocardial disorders and the parameters associated with it including the flowchart programmed in LabVIEW.

Fifth chapter gives on overview of Spectrum analysis that can be performed on VCG signal and further study is needed to enhance the results.

Sixth Chapter discusses the results that are obtained from the PTB Diagnostic database and analyzing the T loop parametric changes. Points of detection R index,S index, endpoints of major axes on the different planes of QRS and T loops are shown.

Second last chapter deals with the conclusions and scope for future work.

The closing chapter gives all references, which are used to carry out the dissertation.

CHAPTER 2

CARDIAC PHYSIOLOGY

2.1 Cardiovascular function

The heart serves as a pump which provides the body with blood. The heart is a four-chambered double serial pump, the walls of which consist of tough musculature, the *myocardium*. The right atrium receives blood from the head and the torso via the superior vena cava, and from the abdomen and legs from the inferior vena cava. From the right atrium, the blood is passed into the right ventricle from where it is pumped to the pulmonary artery and on to the lungs. From the lungs, the oxygenated blood returns through the pulmonary veins to the left atrium and the left ventricle, and is then pumped out again to the body through the aorta.

The pumping of the heart is caused by alternating contractions and relaxations of the myocardium - the heart beat. The atria contract slightly earlier than the ventricles, allowing the ventricles to be filled so as to result in a better performance. Valves between the atria and the ventricles as well as between the ventricles and the arteries maintain unidirectional blood flow. The heart itself is provided with blood through the coronary arteries which originate from the aorta.

2.2 Conduction path

The cardiac cycle is normally initiated by an electrical impulse that originates from the sinoatrial (SA) node, and which firrst spreads in the atrial muscle. Passing through the atrioventricular (AV) node, the impulse is slightly delayed to allow the atria to finish their contraction before the ventricles contract. From the AV node, the impulse continues through conduction fibres in the septum to the bundle of His. Here, the signal is divided into the right and left bundle branch. The Purkinje fibres ramifies the impulse around the apex of the heart, towards the atria.

Normally, the heart rate is controlled by the autonomic nervous system. However, both the SA node, the AV node and the Purkinje fibres may initialize impulses spontaneously. The spontaneous heart rate varies in these cells from 70 beats per minute for the SA node,

to approximately 30 beats per minute for the Purkinje fibres. In this way, a block of the impulse does not prevent basic ventricular activity.

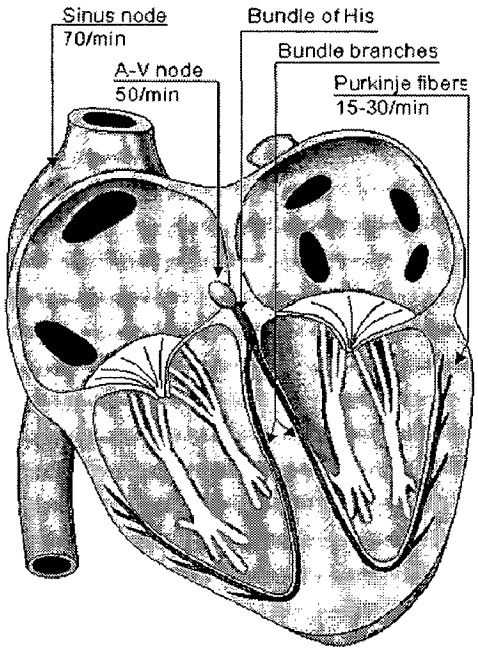

Figure 2.1 Electrical Conduction system of the Heart

The impulse initiating the heart beat is transmitted as a change in electrical potential of the cell membranes along the propagation path. All myocardial cells exhibit a potential difference across the cell membrane which is due to a difference in ion balance between the exterior and the interior of the cell. Muscular cells, as well as some other cells such as nerve cells, possess the ability of being excitable, i.e., transmitting an impulse.

Deviations in the cell membrane permeability may cause a change in the membrane potential. The result is an inward flow of positively charged potassium ions. As a result, the cell becomes neutral or slightly positively charged, i.e., it is *depolarized.* A fraction of a second later, the cell resumes its usual negative potential again, i.e., it is *repolarized.* From the onset of the depolarization phase and extending into the repolarization phase, a refractory period prevents the cell from initiating a new depolarization phase. The entire event is referred to as an *action potential* and is caused by, e.g., electrical stimulation or a change in chemical concentration. The depolarization corresponds to the contraction phase, while repolarization corresponds to the relaxation phase.

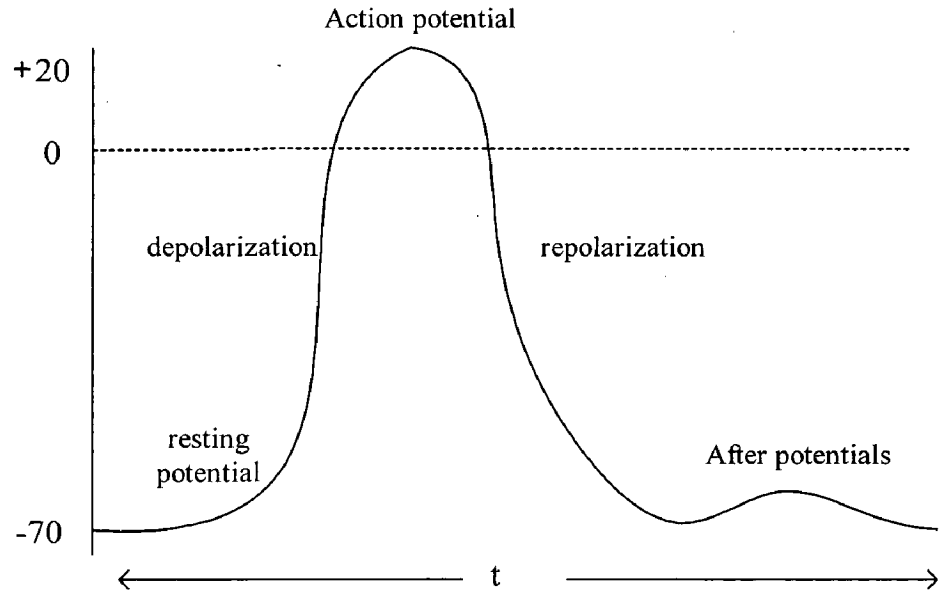

Figure 2.2 Waveform of the action Potential

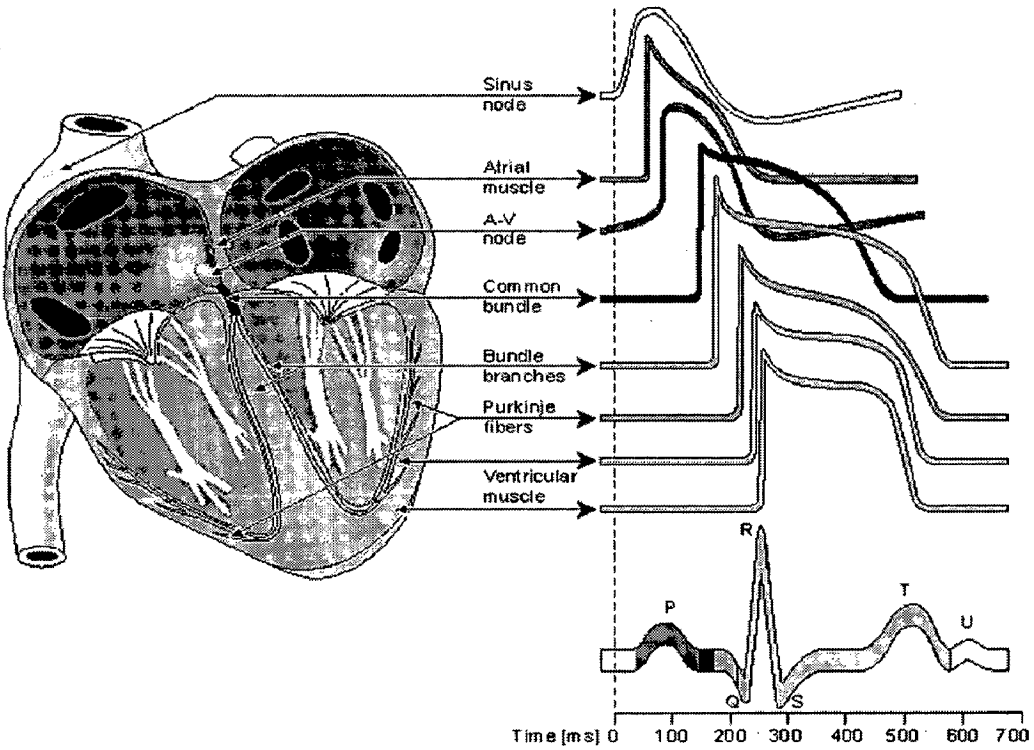

Figure 2.3 Propogation of Electrical Impulse inside the Heart

The cardiac cycle can be divided into several features. The main features are the P wave, PR interval, QRS complex, Q wave, ST segment and T wave. Each of these components represents the electrical activity in the heart during a portion of the heart beat.

The different events of the cardiac cycle are reflected in the ECG. The atrial depolarization is reflected by the *P wave*, and is followed by the ventricular depolarization, the *QRS complex*. The atrial repolarization is simultaneous with the ventricular depolarization and is therefore not visible in the ECG. Finally, the ventricular repolarization is reflected by the *T wave*.

- The P- wave represents the depolarization of the atria.
- The PR- interval represents the time of conduction from the onset of atrial activation to the onset of ventricular activation through the bundle of His.

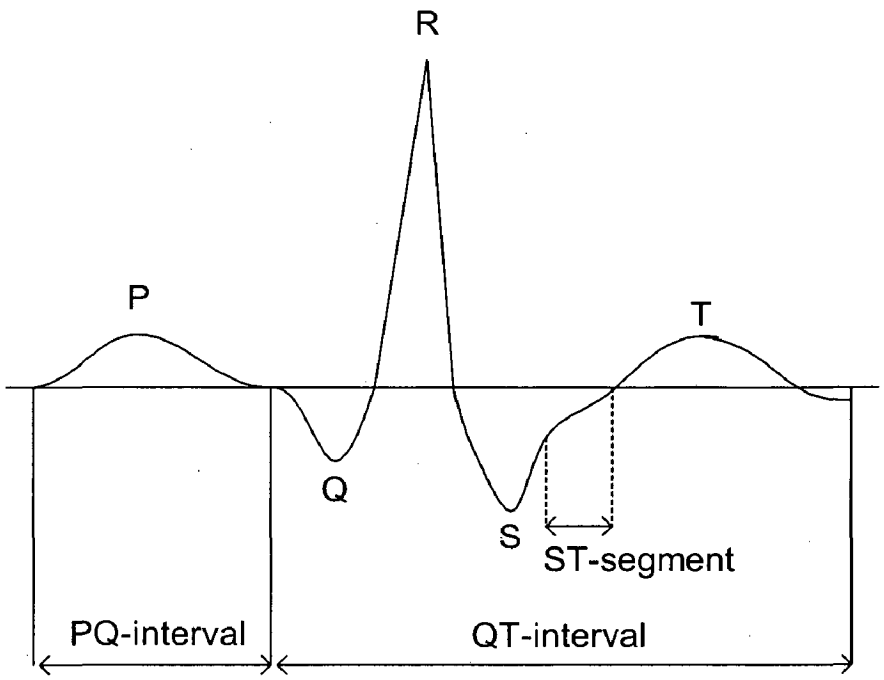

Figure 2.4 ECG waveform

- The QRS complex is a naming convention for the portion of the waveform representing the ventricular activation and completion.

- The ST segment serves as the isoelectric line from which the amplitudes of other waveforms are measured and also is important in identifying pathologies, such as myocardial infarctions (elevations) and ischemia (depressions).
- The T wave represents the ventricular repolarization.

2.3 Cardiac diseases

Ischemia and *arrhythmias* constitute two main types of cardiac disease. The former is related to insufficient blood supply to the heart muscle itself, due to blocks in the coronary arteries. The latter type involves deviation in rhythm from the normal resting heart rate, which is usually 60-100 beats per minute (bpm) and initiated by the SA node. The two types of diseases are related in such a way that, e.g., ischemic heart disease may cause certain types of arrhythmias.

2.3.1 Ischemic heart disease

Ischemic heart disease is one of the most common causes of death in the industrialized world; A diet including large amounts of cholesterol combined with smoking are contributing risk factors to ischemic heart disease. Ischemic heart disease is caused by insufficient blood supply to the heart itself. Being a muscle, the heart must be supplied with oxygen in order to function; this is taken care of by the coronary arteries and their ramifications. In more or less all adults, areas of thickening, denoted atherosclerotic plaques, slowly develop in the arteries. In many individuals, these plaques never cause any symptoms, while in others they may obstruct the coronary arteries. The obstruction may lead to difficulties in supplying the heart with sufficient amount of blood during exercise or mental stress.

Three different manifestations can be derived from ischemic heart disease. The least severe is *angina pectoris* which is defined as discomfort or chest pain. In *myocardial infarction*, ischemia results in necrosis (death) of a certain part of the myocardium. An infarction is most often associated with pain, usually lasting from a few hours to a couple of days. For most patients who suffer from an infarction, circulation is maintained and the only remaining sign is found in the ECG. For a small percentage of these patients, a state

of shock with pallor, coolness and low blood pressure occurs; in these cases mortality is around 70 %.

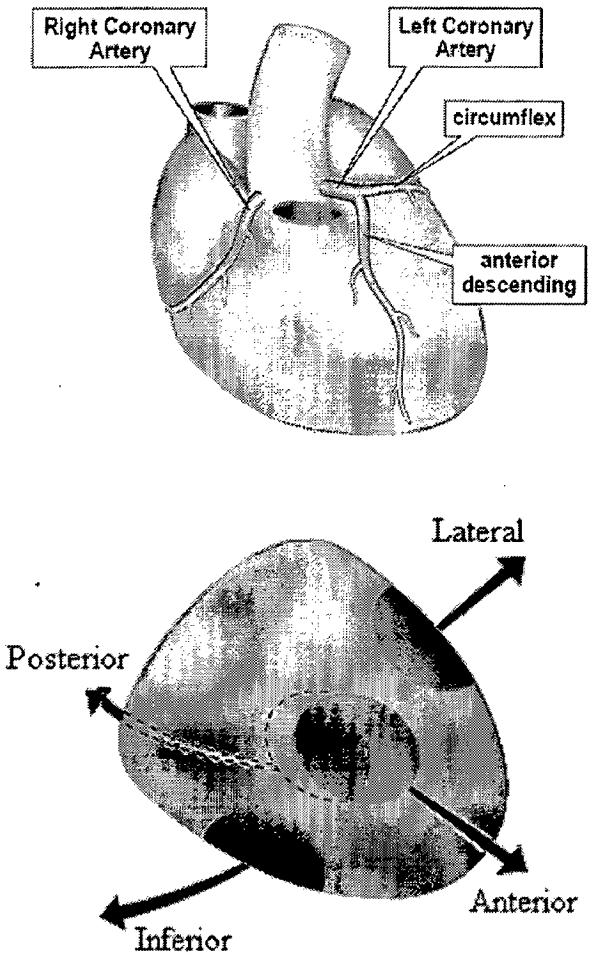

Figure 2.5 Coronary Artery Anatomy

Myocardial infarction (infarct) occurs when a coronary artery supplying the left ventricle become occluded, so an area of the heart is without blood supply. Commonly, thick left ventricle suffers myocardial infarction. The left ventricle is the thickest chamber of the heart; so if the coronary arteries are narrowed, the left ventricle (which uses the greatest blood supply) is the first to suffer from an obstructed coronary. The Q wave makes the diagnosis of infarction. A lateral infarction is caused by an occlusion of the circumflex branch of the Left Coronary Artery. An anterior infarction is due to an occlusion of the Anterior descending branch of the Left Coronary Artery.

Traditionally, ischemia has been diagnosed by primarily analyzing the STT segment amplitude at a specific interval, e.g., 60 ms after the J-point, or at the T wave maximum. However, with the ever-increasing computer power, diagnosis of ischemia can be based on more advanced signal processing algorithms. During the last decade, a number of approaches have been developed which focus on measures reflecting global changes in the ST-T segment. It is believed that such approaches make algorithms more sensitive to ischemia as well as more robust to noise.

CHAPTER 3

DERIVED LEADS

When computers were first used for interpretation of the ECG, there were two competing approaches. In one, the conventional 12 lead ECG was used with each lead being recorded singly. In the other, 3 orthogonal lead ECG was used by Pipberger and co-workers(1961). At that time, research groups used physically large computer systems which were not necessarily fast, certainly by today's standard. For this reason, it was of interest to use the minimum number of leads necessary to obtain a satisfactory diagnostic interpretation. In 1961, Pipberger had shown that the 3- orthogonal lead ECG contained as much clinical information as the 12- lead ECG. In addition, the concept of telephonic transmission of ECGs was being implemented widely at that time and the 3- orthogonal lead ECG could also be transmitted more quickly than the 12- lead ECG. However the X,Y, Z leads were not well understood by physicians. Possibly for these reasons, Dower et al(1980) had the idea that the 12- lead ECG could be derived from the three orthogonal leads X,Y,Z .

3.1 Spatial Vector:

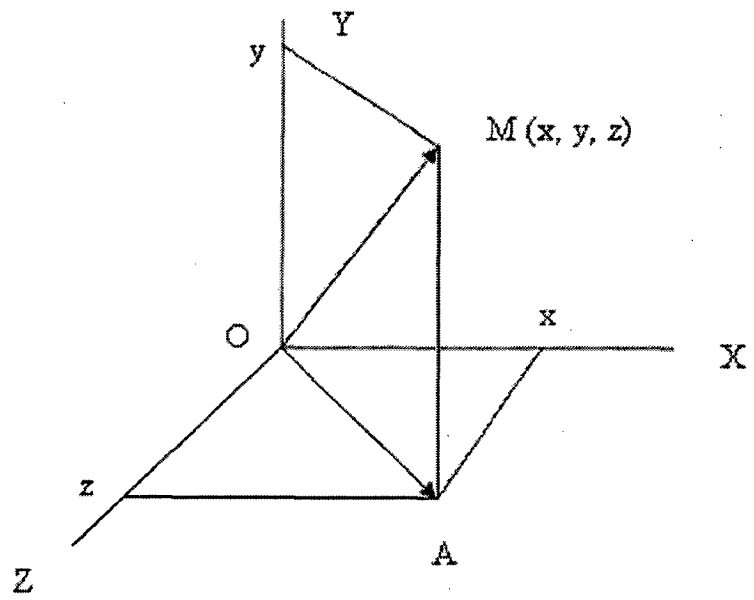

Figure 3.1: A spatial Vector OM and its components x,y,z in a 3- dimensional coordinate system

The magnitude of the vector can be calculated from using triangle OMA,

$$OM^2 = OA^2 + AM^2 = x^2 + y^2 + z^2 \qquad \text{--------- (3.1)}$$

OM can have components in the three mutually perpendicular directions X,Y, Z. If the components x, y, z can be measured at a particular instant in the cardiac cycle, then a resultant OM can be calculated. This is the basis of an orthogonal lead system.

3.2 VCG Lead System

There are both *uncorrected* and *corrected* VCG lead systems. The uncorrected VCG systems do not consider the distortions caused by the boundary and internal inhomogeneities of the body. The uncorrected lead systems assume that the direction of the spatial line connecting an electrode pair yields the orientation of the corresponding lead vector. Currently it is known that this assumption is inaccurate

The goal of the corrected lead system is to perform an *orthonormal* measurement of the electric heart vector. In an orthonormal measurement both of the following requirements are fulfilled:

1. The three measured components of the electric heart vector are *orthogonal* and in the direction of the coordinate axes (i.e., the lead vectors are parallel to the coordinate axes, which are usually the body axes). Furthermore, each lead field is uniform throughout the heart.
2. Each of the three components of the electric heart vector is detected with the same sensitivity; that is, the measurements are *normalized.*

3.2.1 Uncorrected lead systems

For historical reasons, it is worth noting that the earliest attempts at designing orthogonal lead systems were made on the basis of constructing leads such that lines joining the electrodes were essentially mutually perpendicular. This is most easily understood by considering the cube system introduced by Grishman (1952). However, as experience

gained and mathematical modeling advanced, it was found that these lead systems did nto accurately measure the required components.

3.2.2 Corrected orthogonal lead systems

As a result of considerable modeling, both mathematical and physical, such as using model torsos filled with conducting solution, corrected orthogonal lead systems gradually were introduced. The most notable and the one which is generally used wherever vectorcardiography is currently studied using an orthogonal lead system, is that of Frank .

In 1956 Ernest Frank published a vectorcardiographic lead system that was based on his previously published data of image surface. Because the image surface was measured for a finite, homogeneous thorax model, the volume conductor model for the Frank VCG-lead system was also the same.

3.2.2.1 Electrode Location Requirements

To measure the three dipole components, at least four electrodes (one being a reference) are needed. Frank decided to increase the number of electrodes to seven, in order to decrease the error due to interindividual variation in the heart location and body shape.

It is important that the electrode location can be easily found to increase the reproducibility of the measurement. The reproducibility of the limb electrodes is very good. However, the arm electrodes have the problem that the lead fields change remarkably if the patient touches the sides with the arms, because the electric current flows through the wet skin directly to the thorax. This problem has a special importance to the left arm, since the heart is closer.

3.2.2.2 Determination of the Electrode Location

Based on the above requirements Frank devised a lead system, now bearing his name, which yields corrected orthogonal leads. Electrode numbers and positions were chosen very deliberately, and were based upon his image surface model. Specifically, he chose the points designated A, E, I, and M on the left, front, right, and back, respectively. He

also chose point C between points A and E because it is close to the heart. In addition, a point on the neck and one on the left foot were included [17].

3.2.2.3 Frank's image space

In developing his 3- orthogonal lead system, Frank made use of the concept of image space.

The potential Vp at any point P on the surface of the body or a torso model is given by

$$Vp = HxLx + HyLy + HzLz \quad \text{---------(3.2)}$$

Where Hx, Lx etc are heart and lead vector components respectively.

If a torso model is used and the heart vector is represented by a physical dipole, it is feasible to study the potential measurements on the surface of the torso model. If the dipole is oriented in X direction then the potential at point P is given by Vp = HxLx because Ly and Lz are zero. If the physical dipole is given unit strength, i.e Hx = , then the equation reduces to Vp = Lx. In other words the X component Lx of the lead vector corresponding to the point P can be measured. Similarly Ly and Lz can be obtained. Thus the components of the lead vector L can be determined. In effect, the components Lx, Ly and Lz provide the coordinates of another point in space.

If this procedure is repeated for a large number of points P on the surface of the torso model, then in a similar fashion, a complete set of points corresponding to the tips of the lead vectors for the various points P will be obtained. There is mapping for each point P on the physical torso surface onto an imaginary or image surface determined by the coordinates of the various lead vectors Lp. This was image surface described by Frank. For example, a line joining two points on the image surface corresponding to two points on the torso model would map out the actual lead direction for the bipolar lead so constructed. Thus the magnitude and direction of the lead vector could be obtained experimentally.

3.3 The derived 12- lead ECG according to Dower

In the same way as the orthogonal lead vectors can be constructed using the image surface of Frank, it is possible to study the standard 12- lead ECG. If the potentials from leads X,Y and Z are available, it is then possible using the image space to construct a system whereby a percentage of each of the potentials measured by these leads is added together to produce the potential that would be measured by lead I. In general, it can be shown that any lead can be derived from the X,Y,Z leads using the following form of equation:

$$V \text{ derived} = aX + bY + Cz \quad \text{---------- (3.3)}$$

Where a,b,c are coefficients derived from the image surface.

It would be expected that a derived lead I would be in large measure consist of a hight percentage of the potential of lead X with perhaps a small percentage of the potentials of leads Y and Z to take account of the differences in lead vector magnitude and direction between lead I and lead X.

$$\text{Lead I} = 0.632X - 0.235Y - 0.059Z \quad \text{---------- (3.4)}$$

Similarly, coefficients a, b, c can be provided for the calculation of any of the 12 leads. A comparison of the derived 12 lead ECG with the actual 12 lead ECG recorded from a patient in large measure shows good similarity.

3.4 Vector loop presentation

3.4.1 Nomenclature

The American Heart Association Committee on electrocardiography (1975) published a set of recommendations for vectorcardiographic terminology. The committee recommended that the lead Z be directed positively to the posterior thorax, although this does mean that the scalar presentation of lead Z is essentially opposite to that of Lead V2. This certainly caused much confusion when describing scalar lead appearances and for this reason, lead Z is directed positively to the anterior to be similar to V2.

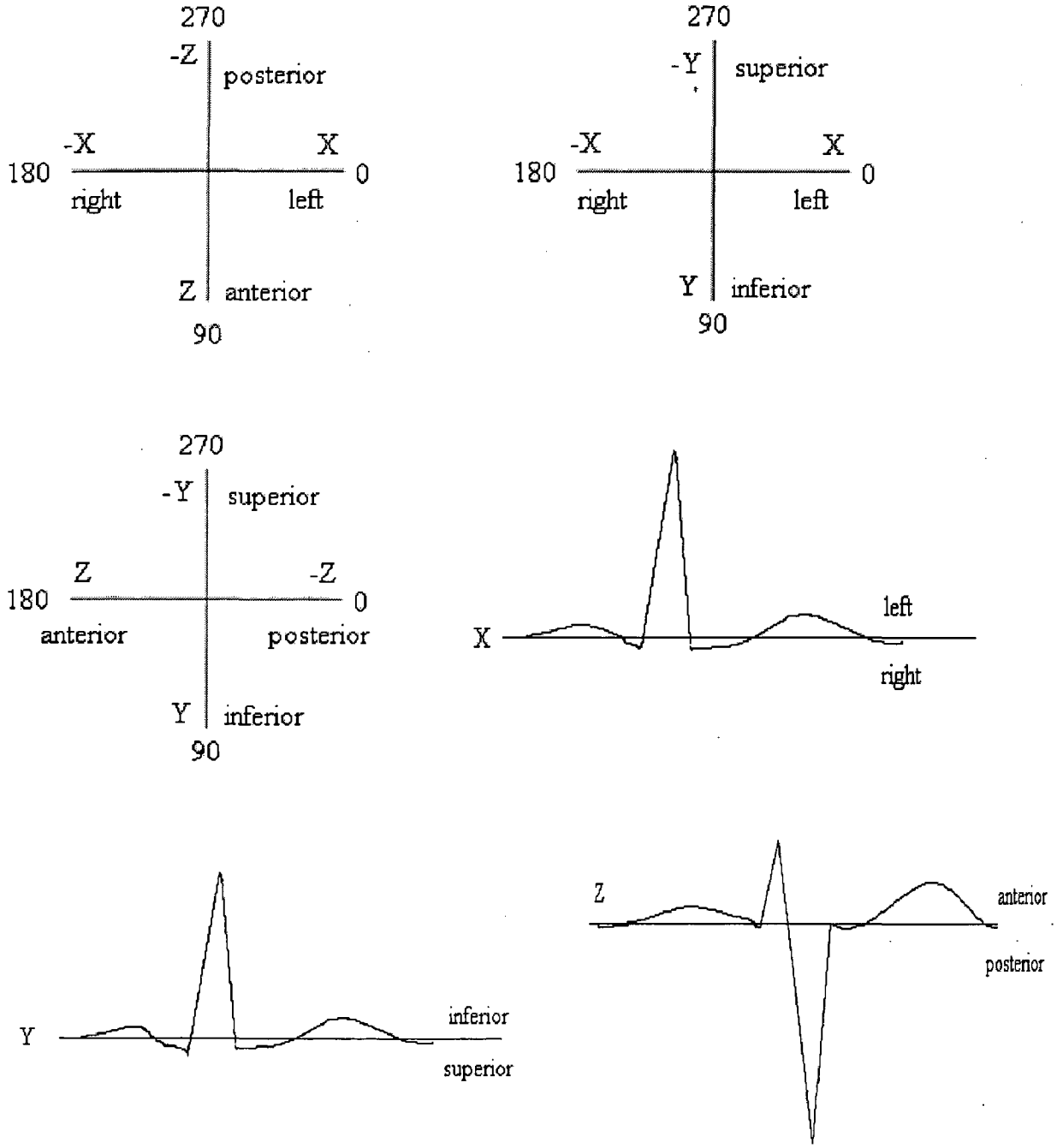

Figure 3.2: Polarity of leads X,Y and Z and angular reference frame of the frontal, transverse and left sagittal planes

3.4.2 Display techniques

About 20 years ago the most common method of displaying the vectorcardiogram was via an oscilloscope. Pairs of leads such as X and Y were used to deflect the electron beam

horizontally and vertically respectively and in this case frontal plane loop would be generated. Consider that leads X and Y are recorded simultaneously. At any instant of time, an amplitude for each lead is known i.e an (x,y) coordinate pair of values is available. These could be plotted simply on XY axes. If this is repeated throughout the cardiac cycle, then a complete frontal plane set of P, QRS and T loops can be generated.

If the leads X, Y, Z are considered in pairs, then by plotting lead XY, the frontal plane vectorcaridographic loop can be obtained. Similarly XZ plots produce the transverse plane loop and ZY plots produce the sagittal plane vectorcardiographic loop. Nowadays with the widespread availability of computer technology, these plots can be produced in a straightforward fashion. Using LabVIEW, the leads X, Y and Z are plotted against each other by clustering and using the function XY graph.

It is important that vectorcardiographic loops have some indication of the speed of inscription as this can contain diagnostic information. Some indication must be given to make it quite clear to the viewer in which direction the different planar loops are inscribed. On the loop at 10, 20, 30... ms dots are marked to indicate that the loop is clockwise or anticlockwise visually [9].

3.5 Derived X,Y,Z leads

3.5.1 Methods of derivation:

Different methods of deriving the X,Y,Z leads from the `12- lead ECG have been proposed. Rubel et al. (1991) compared eight different published approaches and concluded that either a regression technique or a method based on the reverse of Dower's approach provided the best solution. Other methods include, at the one extreme using leads I, aV_f, V_2 as substitutes for X,Y and Z to more complex mathematical techniques for deriving regression equations for each mapping nevertheless are simple to apply [10].

3.5.1.1 Simple Analogues

Although it is quite feasible to suggest that I, aV_f and V_2 are in sense equivalent to X, Y and Z, these leads are not mutually perpendicular according to Frank's image surface.

However very similar approaches have been used. For example, Bjerle and Arvedson (1886) suggested [18]:

$$X = 1.06\ V_6 \qquad \text{-------} (3.5)$$

$$Y = 1.88\ V_f = 1.25\ aV_f \qquad \text{-------} (3.6)$$

$$Z = 0.532\ V_2 - 0.043\ V_6 \qquad \text{-------} (3.7)$$

Marquette Electronics used the following transformation:

$$X = I \qquad \text{-------} (3.8)$$

$$Y = aV_f \qquad \text{-------} (3.9)$$

$$Z = 0.8\ (V_1 + V_2/2) \qquad \text{-------} (3.10)$$

3.5.1.2 Regression techniques

If the 12- lead and the corresponding X,Y, Z lead ECG are available from a large number of patients, then it becomes feasible to the statistical techniques to calculate regression coefficients that allow the X,Y,Z leads to be expressed in the following form:

$$X = aI + bII + cV_1 + dV_2 + eV_3 + fV_4 + gV_5 + hV_6 \quad \text{--- (3.11)}$$

This approach was used by Kors et al. (1990) who provided the following coefficients

Leads	I	II	V_1	V_2	V_3	V_4	V_5	V_6
X	0.38	-0.07	-0.13	0.05	-0.01	0.14	0.06	0.54
Y	-0.07	0.93	0.06	-0.02	-0.05	0.06	-0.17	0.13
Z	0.11	-0.23	-0.43	-0.06	-0.14	-0.20	-0.11	0.31

Table 3.1 Regression coefficients

CHAPTER 4

T-Loop Morphology

Abnormalities of either ventricular depolarization or repolarization on standard 12-ECG can predict cardiovascular mortality. QRS duration is a measure of ventricular depolarization, whereas the ST segment and T wave of the QT interval represent the repolarization phase. Depolarization abnormalities reflect ventricular structural abnormalities (i.e., damage and hypertrophy), whereas repolarization abnormalities represent heterogeneities associated with electrical instability and sudden death. Combining both phenomena should optimize prognostication [1].

4.1 Background

Ventricular repolarization abnormalities play an important role in the determination of arrhythmia and sudden cardiac death and may reflect subclinical myocardial ischemia changes. The process of repolarization at rest is routinely quantified from standard 12-lead electrocardiogram (ECG), either as time-domain indexes such as the QT interval and its derivations, or as abnormalities of the ST segment or of the T wave. While these indexes have all been reported to be associated to some degree with incident coronary heart disease events, they have some limitations. The widely used QT interval reflects only the temporal aspect of the repolarization, and as currently defined and measured from the 12-lead ECG, QT dispersion has major conceptual and technical limitations. Theoretical and experimental studies suggest that ventricular repolarization occurs in a nonlinear and inhomogeneous fashion. As a consequence, spatial measures of repolarization that take into account T-wave complexity using the T-wave vector (axis) should be more accurate and useful surface ECG markers of repolarization abnormalities than simple scalar intervals from the ECG, such as the QT interval or QT dispersion. Clinical studies have shown that the T wave axis reflects changes associated with autonomic adaptive or maladaptive influences, systemic hypertension, coronary occlusion, and microalbuminuria in individuals without diabetes mellitus. Moreover, it has recently been shown that spatial T wave axis deviation has good measurement

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Z	0.11	-0.23	-0.43	-0.06	-0.14	-0.20	-0.11	0.31

Table 3.1 Regression coefficients

3.5.1.3 Matrix Inversion Technique or Inverse Dower's approach

Dower et al (1980) in Vancouver had the idea that the 12 lead ECG could be derived from the three orthogonal leads X,Y,Z. Using the concept of Frank's image space, Dower derived the coefficients for deriving the 12 – lead ECG from frank leads X,Y and Z. If this matrix of transfer coefficients is denoted by T then the equation for deriving the 12 – lead ECG, E, from the vector leads X,Y,Z denoted by V can be expressed in matrix form as follows [18]

$$E = TV \qquad \text{------- (3.12)}$$

By using mathematical manipulation, it is possible to turn the equation around so that tha vector leads V can be derived from a knowledge of the 12 lead ECG.

$$V = N^{-1} T^{t}E \qquad \text{------- (3.13)}$$

Because the matrix of values has been inverted, this approach has been called the 'inverse Dower' technique

Leads	I	II	V_1	V_2	V_3	V_4	V_5	V_6
X	0.156	-0.01	-0.172	-0.074	0.122	0.231	0.239	0.194
Y	-0.227	0.887	0.057	-0.019	-0.106	-0.022	0.041	0.048
Z	-0.022	-0.102	0.229	0.310	0.246	0.063	-0.055	-0.102

Table 3.2 Inverse Dower Coefficients

CHAPTER 4

T-Loop Morphology

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properties and is repeatable, a finding that supports its use in clinical and epidemiological research.

The QT duration measured in any ECG lead is dependent on projection of T wave loop toward recording electrodes located on preferential planes. The projection phenomenon contributes to interlead differences in QT duration, i.e. QT dispersion, which have been considered as a measure of repolarization heterogeneity. It has been demonstrated that low amplitude and wide T wave loops increase QT dispersion, whereas high-amplitude and narrow loops are associated with lower QT dispersion .Since QT dispersion is an inaccurate reflection of T wave loop morphology there is a need for new methods quantifying T wave loop abnormalities. The shape of T wave loop reflects the sequence of ventricular repolarization and quantifying it may provide insight into regional heterogeneity of repolarization in myocardium. By evaluating T wave loop features (angle, T wave axis) we may identify patients at risk for sudden cardiac death [4].

The principal component analysis (PCA) is an example of new methods aiming to determine the shape of T wave loop and to quantify its complexity. This method calculates eigenvectors of repolarization, which represent overall spatial dimensions of T wave loop. The ratio of first to second eigenvector provides insight into the length and width of T wave loop, but it does not inform about specific shape of T wave loop. In particular, the PCA method is unable to identify small deflections in T loop shape, which might reflect sudden changes in repolarization wave front in the myocardium most likely reflecting fractionated (heterogeneous) repolarization [2].

Abnormalities of ventricular repolarization in the electrocardiogram (ECG), such as ST depression, T-wave inversion and QT prolongation, have repeatedly been shown to carry prognostic value for cardiac morbidity and mortality. More recently, there has been renewed interest in vectorcardiographic parameters characterizing T-loop morphology to quantify ventricular repolarization. A vectorcardiographic parameter that has been found to be a strong and independent risk indicator for cardiac events is the frontal T axis, which reflects the main orientation of electrical heart activity during repolarization [8].

4.2 T – Loop parameters

Particular aspects of T-wave morphology are considered for characterization and quantification of heterogeneous repolarization, using the vectorcardiogram (VCG) [1].

Three parameters were obtained from the VCG with two different computation methods:

1. Maximum angle between QRS and T loop axes
2. T axis elevation and azimuth angle difference
3. Ratio of maximum to mean T vector magnitudes

4.2.1 Maximum angle between QRS and T loop axes

The parameter that has recently been proposed as a marker of ventricular repolarization is the spatial QRS-T angle. The spatial QRS-T angle is defined as the angle between the directions of ventricular depolarization and repolarization. Thus, unlike the T axis, it also takes depolarization into account, akin to the concept of the ventricular gradient. The spatial QRS-T angle, was recommended years ago by vector cardiographers as such a combined measurement and more recently, the global angle between QRS wave and T wave was demonstrated to risk stratify post myocardial infarction patients [3].

It yields adequate information for detection of myocardial infarction and ischemia and their location in the myocardium. While the infarction is mostly connected with the QRS loop morphology and axes direction, the ischemia of the myocardium impairs its recovery process. When a portion of the ventricular wall is so affected, the normal sequence of ventricular repolarization will be disturbed and the resultant potential altered. Consequently, abnormalities are manifested in the T vectors.

The QRS and T axes are determined from the zero point of the VCG (the isoelectric point in ECG) to the most remote point of the respective loops.

The angles (AF, AH and ALS) in the three VCG planes - frontal, horizontal and left sagittal - are computed, and the maximum value is the considered parameter MA:

$$MA = \max(AF, AH, ALS) \qquad \text{-------} (4.1)$$

The estimation of the T loop axis in relation to the QRS complex axis provides adequate information for detection of myocardial infarction and ischemia and their location in the myocardium.

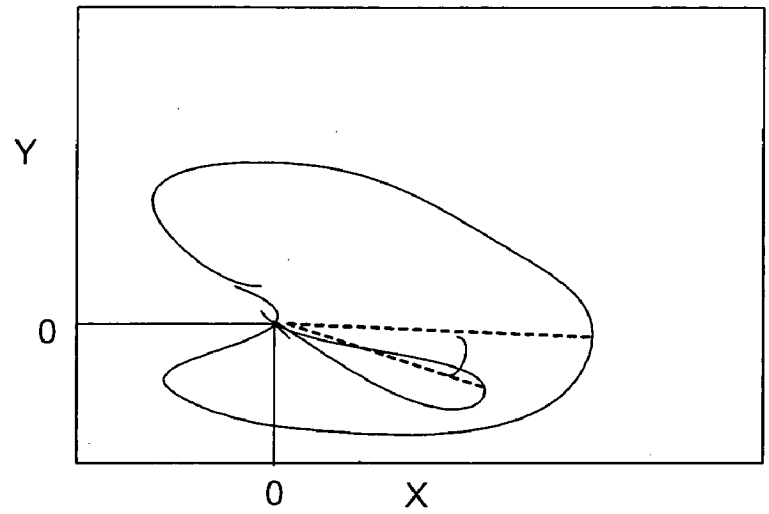


Figure 4.1: Angle between the QRS and T loop axes in the frontal plane

The orientation of QRS and T loops are close to each other in normal case, whereas they are far from each other in some heart diseases.

3.1.2 T axis elevation and azimuth angles difference

The difference between frontal plane elevation and azimuth angles (DEA) is calculated along the entire segment of repolarization, providing a description of how complex the T-wave loop morphology is.

DEA is defined as the mean absolute value of the difference between the frontal plane Elevation (α) and Azimuth (β) of all loop samples (n)

$$DEA=\text{mean}(\text{abs}(\alpha_n - \beta_n)) \qquad \text{-------} (4.2)$$

The Elevation with respect to the frontal plane is the angle between the loop axis at sample *n* and the Z axis in the left sagittal plane, while the Azimuth is the angle between the loop axis at the same sample and the X axis in the frontal plane [1].

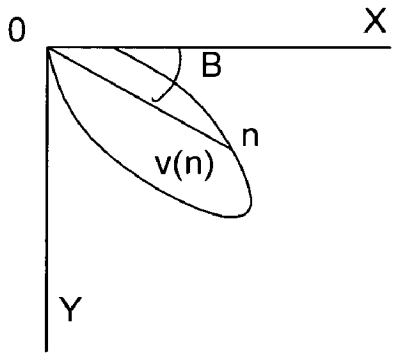


Figure 4.2: Definition of the elevation angle used for DEA parameter

4.1.3 Ratio of maximum to mean T vector magnitudes

The ratio of maximum to mean vector magnitudes (RMMV) is adopted from the work of Kallert et al.:

$$RMMV = V_{max} / \text{mean}\,(V_n), \qquad \text{-------}\ (4.3)$$

Where V_n is the magnitude of the spatial vector of the T loop at sample n.
The magnitude is calculated by the Pythagorean formula:

$$V = \sqrt{(Vx^2 + Vy^2 + Vz^2)} \qquad \text{-------}\ (4.4)$$

Where V_x, V_y and V_z are the magnitudes of the projection of the spatial vector on the orthogonal axes.

4.2 Modified T Parameters

The ST segment elevation in the electrocardiogram leads is a recognized reliable indicator of myocardial injury, ischemia or infarction. For this reason, T loop parameter measurements with respect to the zero point of the VCG (the isoelectric point in the ECG) may lead to errors. This is especially valid when the zero point is not included in the T loop.

The size and direction of an ellipse (the form QRS and T loops mostly look like) is best characterized by its major axis, where T loop Eigen values parameter is defined as a ratio of the major to the minor axis of the ellipse [7].

In order to capture the information of major axes, the two points of the loop with the maximum distance are detected and the point nearest to the zero point of the VCG loop is then considered (zero*) for the following analysis. Then three modified parameters (MAm, DEAm, RMMVm) are measured with respect to the new considered zero* point. The difference between the two measurement techniques is illustrated in Figure.

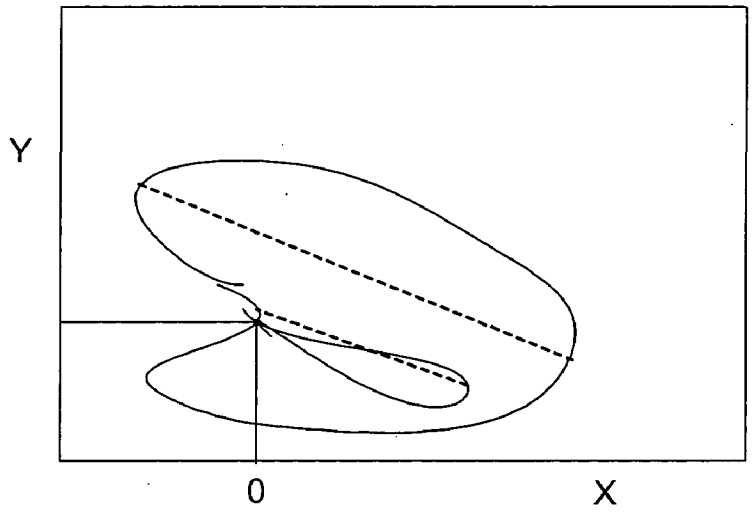
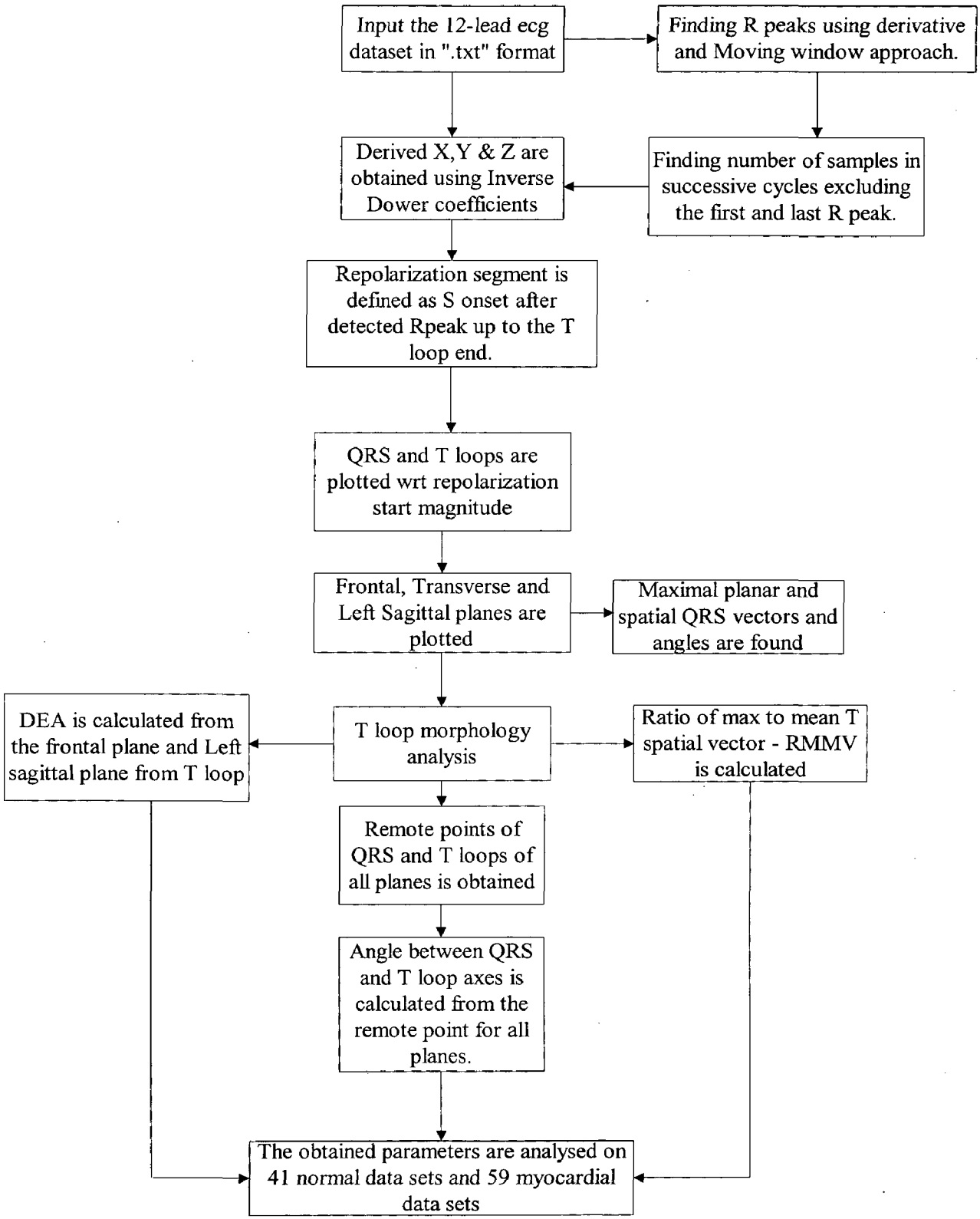


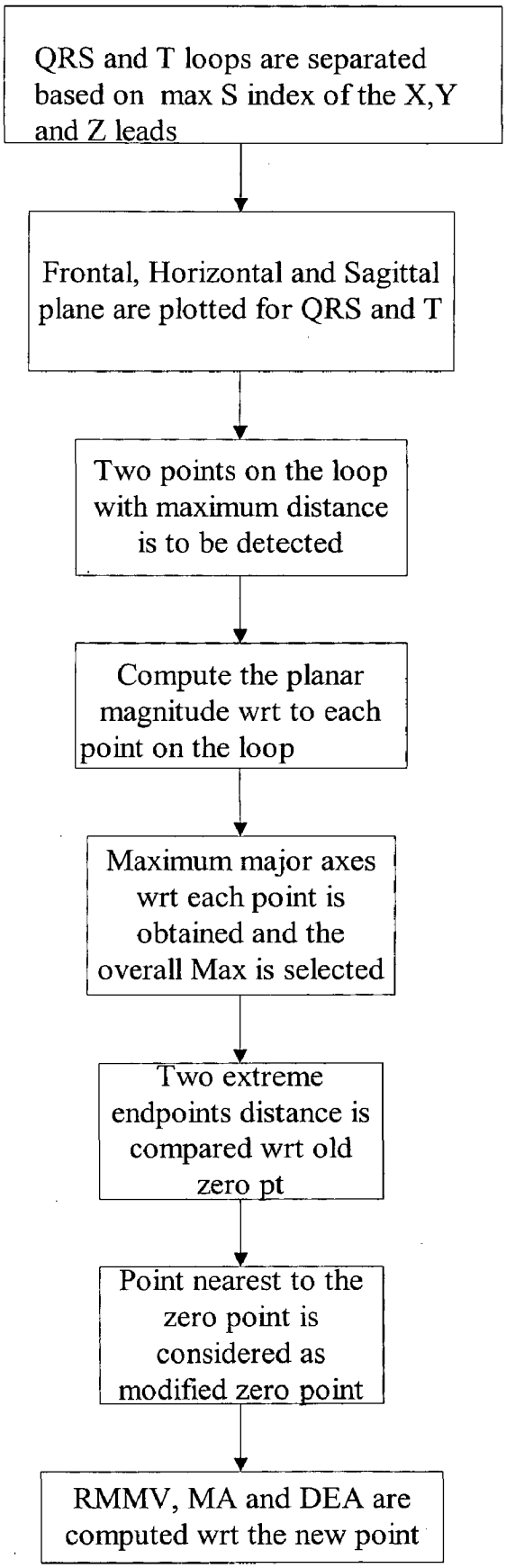
Figure 4.3: Angle between the QRS and T loop axes wrt new zero point in the frontal plane.

In the next pages are the flowcharts of the detection of points in X,Y and Z leads and T loop morphology analysis implemented in LabVIEW are given.

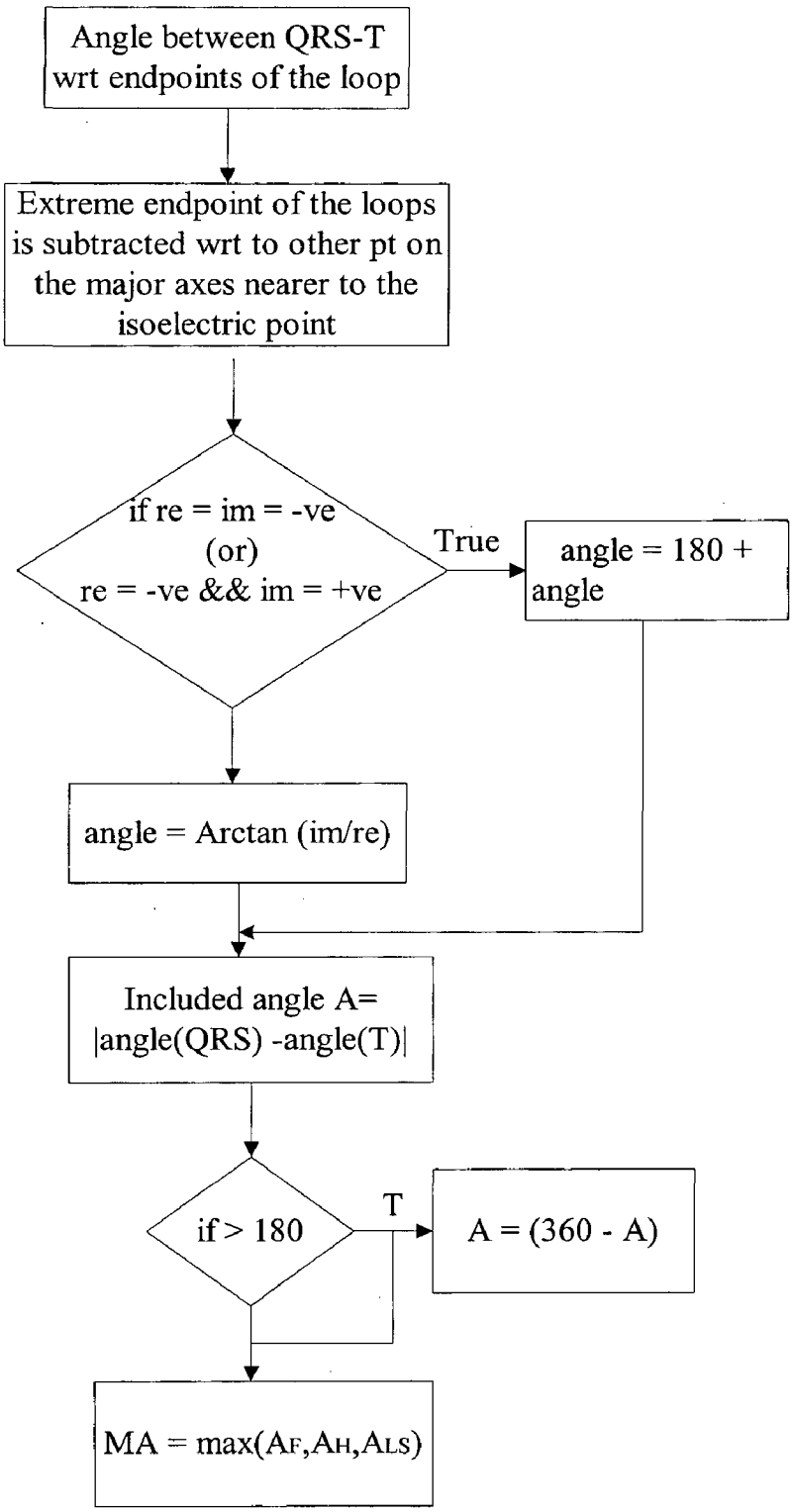
T – Loop Morphology Analysis Flow diagram



Modified Zero point Flow chart



Calculation of MA between QRS and T loop axes:



CHAPTER 5

SPECTRUM ANALYSIS OF VCG SIGNAL

VCG can be seen as a vector on a 3-D orthogonal leads system (e.g., the Frank system) and it is able to describe completely the characteristics of the cardiac electromagnetic vector in 3-D space. In order to specify clearly the VCG characteristics in frequency domain and on a 3-D or 2-D space, the vector properties of every frequency component should be studied.

In order to detect myocardial ischemia, ECG waveform analysis in the time domain has been widely used in clinical practice, but significant changes of the ECG waveform cannot be recognized unless the serious myocardial ischemia occurs. Up to now, many studies of the ECG spectrum have been done to detect myocardial ischemia. It has been found that detecting myocardial ischemia by the ECG spectrum is more sensitive than that by the ECG waveform. ECG spectrum analysis is gradually being applied in clinical practice [11].

5.1 Theory of the Two-Dimensional Spectrum Technique

If a signal is a quasi-period signal, it consists of an essential frequency component and many multiplying (harmonic) frequency components. By decomposing the signal with the FFT algorithm, three parameters (the amplitude, the frequency and the initial phase) of every frequency component can be accurately calculated.

Let two signals, $x(t)$ and $y(t)$, be perpendicular to each other on a 2-D plane, P, formed by axes of $x(t)$ and $y(t)$. Assume $xt)$ and $y(t)$ be quasi-periodic. Because $x(t)$ and $y(t)$ are orthogonal on the P plane, their i^{th} component signals are also orthogonal on the P plane and constitute a 2-D vector. This vector can be drawn on the P plane and is called a Lissajous figure in mathematics. This figure is an ellipse and shows vector properties of the i^{th} VCG frequency components in the frequency domain and also on the 2-D plane, P.

Six parameters of the ellipse are shown in Figure 5.1.

1) long-axis $a(A, A')$;
2) short-axis $b(B, B')$;

3) δ (the angle between long-axis and X-axis);

4) θ (the angle between initial point and X-axis);

5) r *(the* eccentricity of an ellipse);

6) SIGN (the rotational direction of ellipse).

This set of six parameters can describe complete properties of the ellipse (i.e., a Lissajous figure) of the i^{th} frequency component.

Assume the coordinates of orthogonal ECG components at a harmonic frequency of f_i are:

$$x_i(t) = A_{xi} \sin (2\pi f_i t + \alpha_{xi}) \qquad \text{------ (5.1)}$$

$$y_i(t) = A_{yi} \sin (2\pi f_i t + \alpha_{yi}) \qquad \text{----- (5.2)}$$

Where A_{xi}, A_{yi}, α_{xi}, α_{yi} and f_i are obtained by the FFT algorithm.

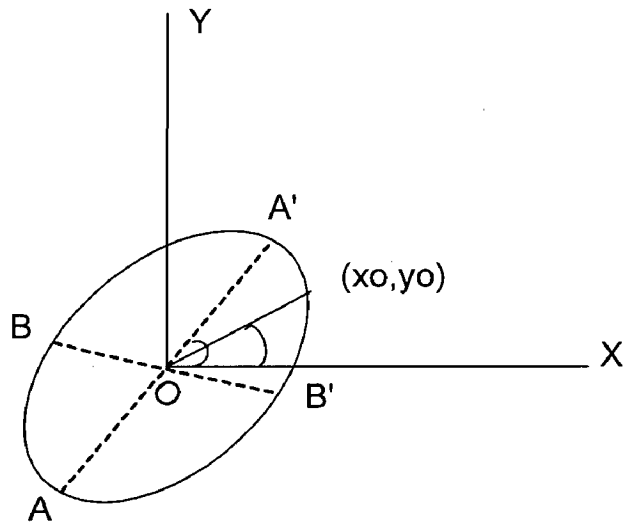


Figure 5.1 Lissajous figure

The definitions of the six ellipse parameters can be defined as follows:

The long axis and short axis are the maximum and minimum from the array of values obtained by $2\sqrt{(x^2 + y^2)}$ from the zero point of the lissajous figure.

$$\text{Longaxis 'a'} = \max\left(2\sqrt{(x^2 + y^2)}\right) \quad \text{-------- (5.3)}$$

$$\text{Shortaxis 'b'} = \min\left(2\sqrt{(x^2 + y^2)}\right) \quad \text{-------- (5.4)}$$

The angle between long axis and X-axis is calculated since the point A'(x_1, y_1) is known from the index of maximum value.

Similarly the angle between initial point (x_0, y_0) and X-axis.

$$\delta = \arctan(y_1 / x_1) \quad \text{and} \quad \theta_0 = \arctan(y0 / x_0) \quad \text{--- (5.5)}$$

The centrifugal rate i.e the eccentricity of the ellipse is given by

$$r = b/a = \text{long axis/ short axis} \quad \text{-------- (5.6)}$$

The rotational direction of the ellipse is:

SIGN = -1 means the ellipse rotates counter clockwise

SIGN = +1 means the ellipse rotates clockwise

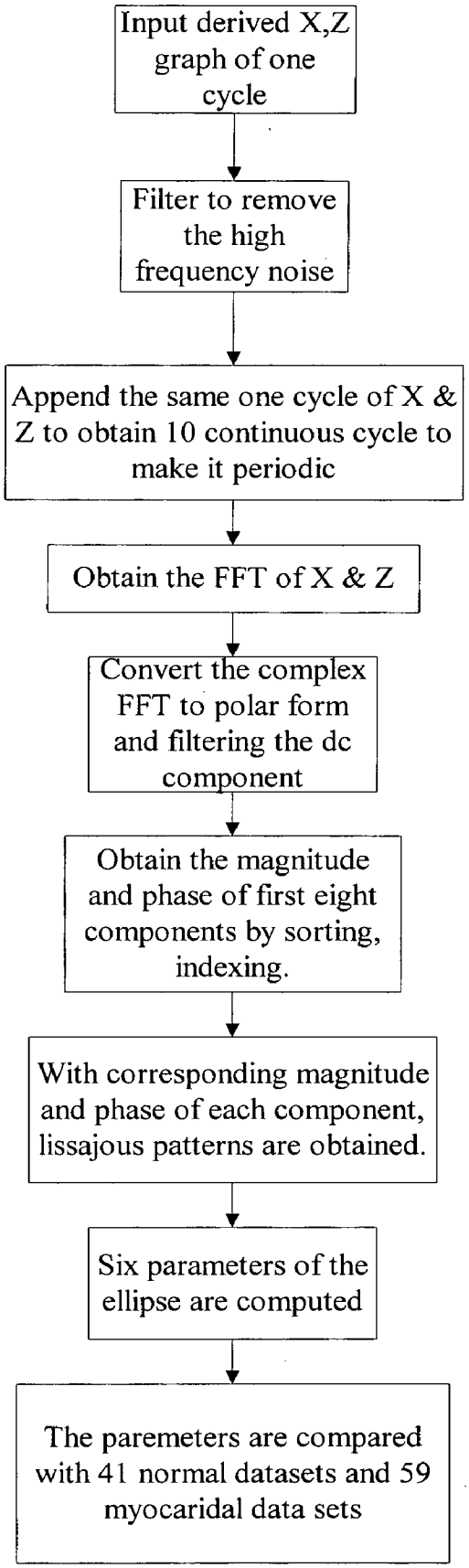
Where SIGN = $\{ 1 , \partial(\Delta\theta)/\partial t \text{ at } (x_0, y_0) < 0$ where $\partial(\Delta\theta)/\partial t = \sin(\alpha_{yi} - \alpha_{xi})$

$$\{ -1, \partial(\Delta\theta)/\partial t \text{ at } (x_0, y_0) > 0 \quad \text{------ (5.7)}$$

Thus, parameters of the ellipse can be calculated. The rotary direction of the ellipse stands for the phase leading or lagging for the two channels of ECG signals. These parameters completely describe the characteristics of the 2-D spectrum of the VCG signal. Lissajous figures are plotted and all the 6 parameters variations are noted for both healthy and myocardial subjects, but further study and analysis is needed to effectively use for the myocardial ischemic disease.

Given in the next page is the implementation of the same in LabVIEW.

5.2 2-D Spectrum Analysis of VCG signal Flow diagram



CHAPTER 6
RESULTS AND DISCUSSION

6.1 Database

The parameters RMMV(Ratio of maximum to mean T vector) , MA (Maximum angle between QRS-T axes) and DEA(Frontal plane elevation and Azhimuth angle difference) and the modified parameters are investigated on "PTB Diagnostic Database" containing 15 lead data (12 lead ECG signal + 3 lead Frank) downloaded from the physionet website.

The database is composed by 100 15-lead data sampled at 1000 samples per second with the leads in the following order I, II, III, aV_r, aV_l, aV_f, V_1, V_2, V_3, V_4, V_5, V_6, X, Y and Z.
The population consists of 72 males and 28 females aged from 22 to 86 years old.
The following two groups have been selected for this study:
42 healthy subjects
58 patients with myocardial diseases.
Following two groups of X, Y and Z leads; and Frontal, Transverse, Horizontal planes of Normal and myocardial subjects is shown below.
Healthy subject:

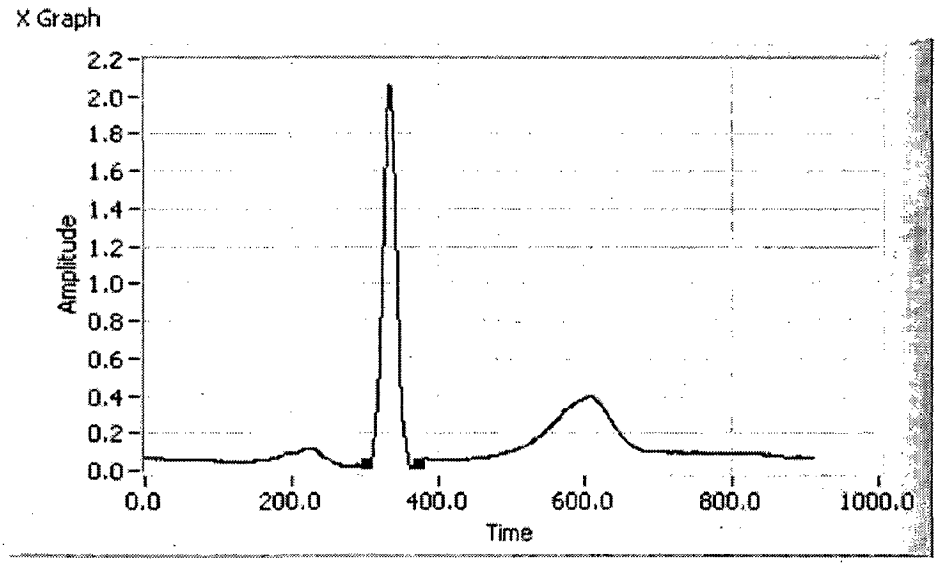


Figure 6.1 Waveform of X lead from the healthy subject

Q and S points are marked as dots in the graph.

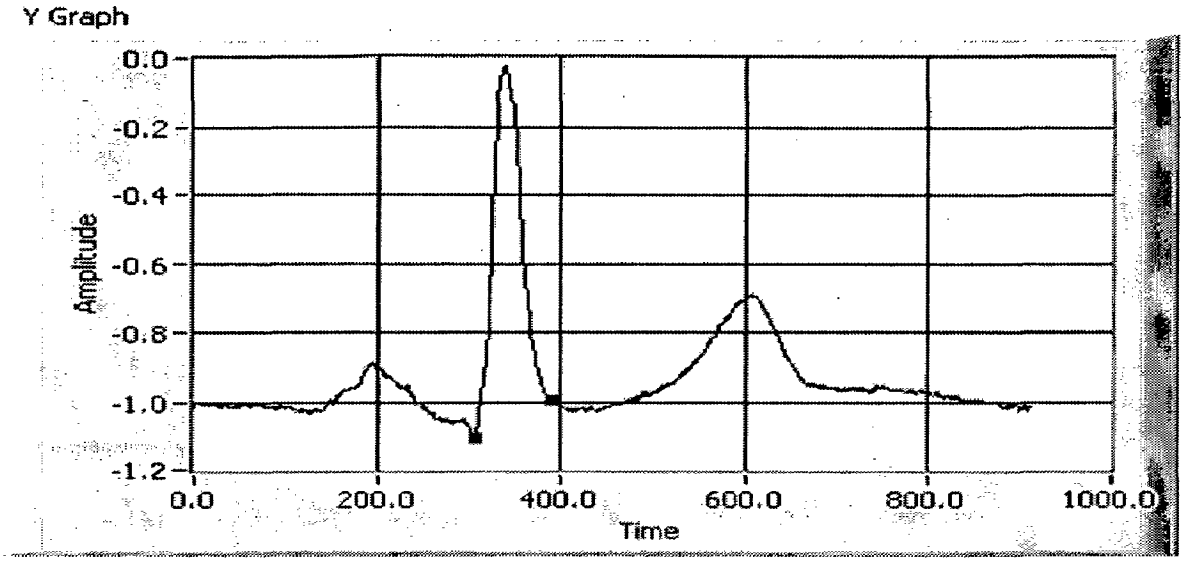


Figure 6.2 Waveform of Y-Lead

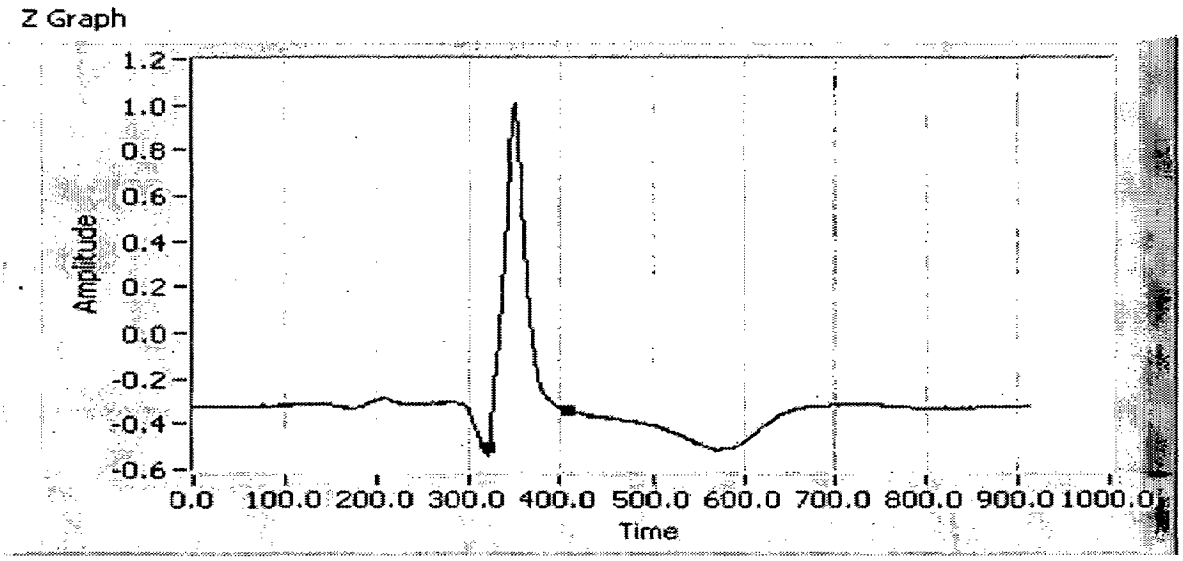


Figure 6.3 Waveform of Z-Lead

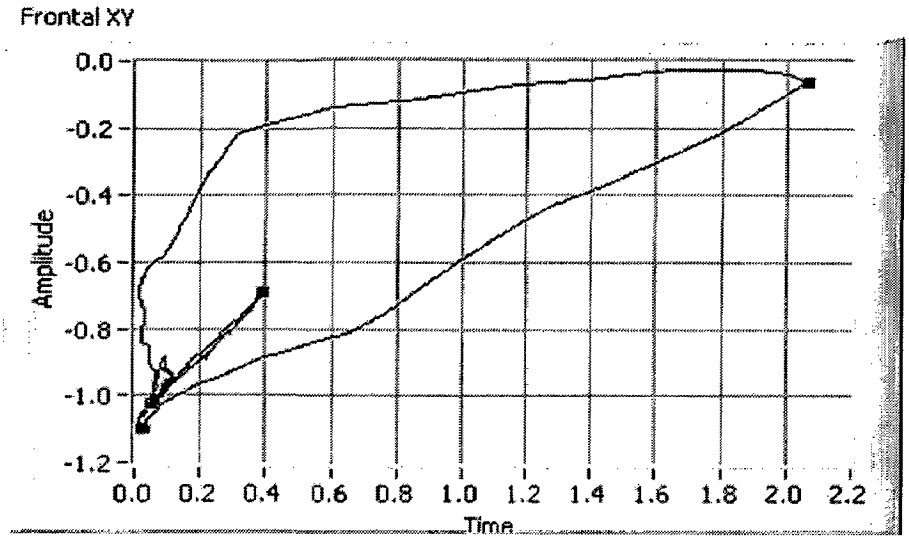


Figure 6.4 Frontal Plane showing QRS and T loops with endpoints of the major axes.

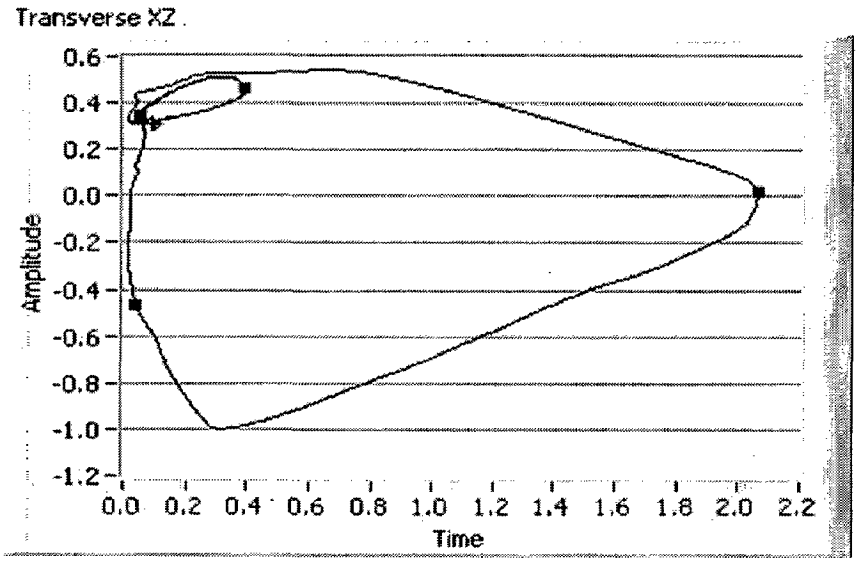


Figure 6.5 Transverse Plane showing QRS and T loops with endpoints of the major axes

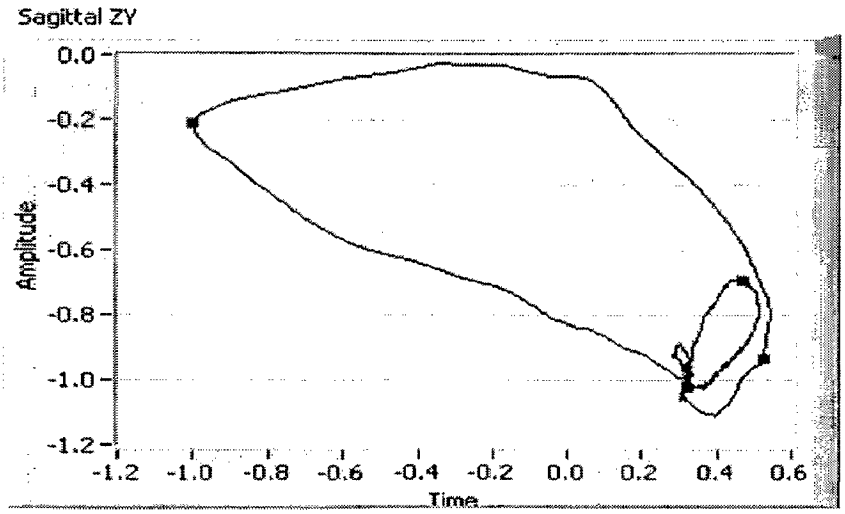


Figure 6.6 Left sagittal Plane showing QRS and Tloops with endpoints of the majoraxes.

Myocardial Patient:

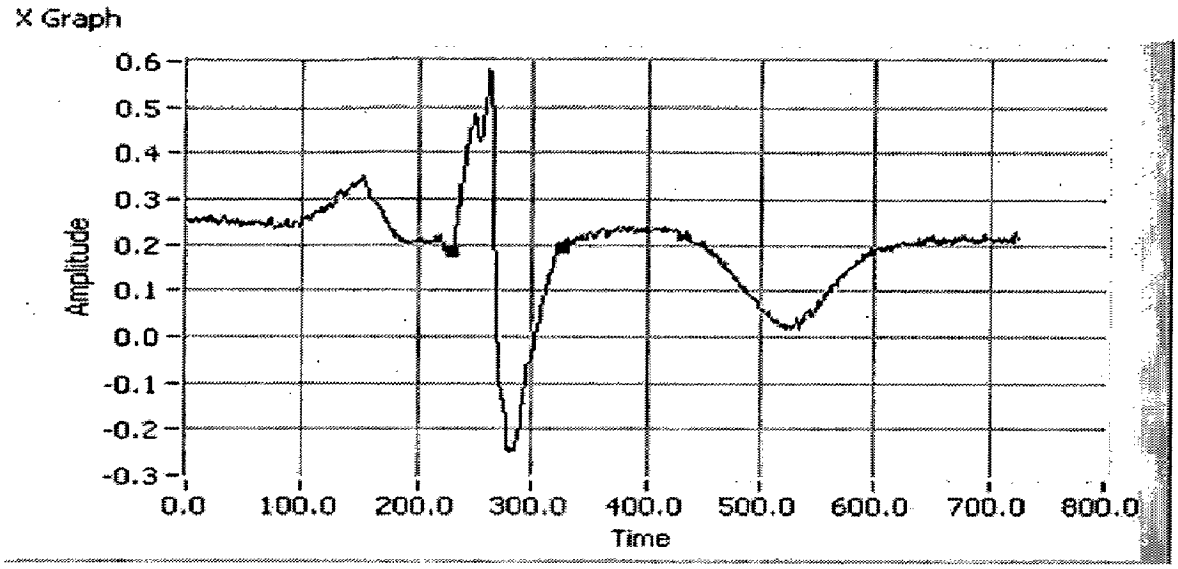


Figure 6.7 Waveform of X lead of the myocardial patient

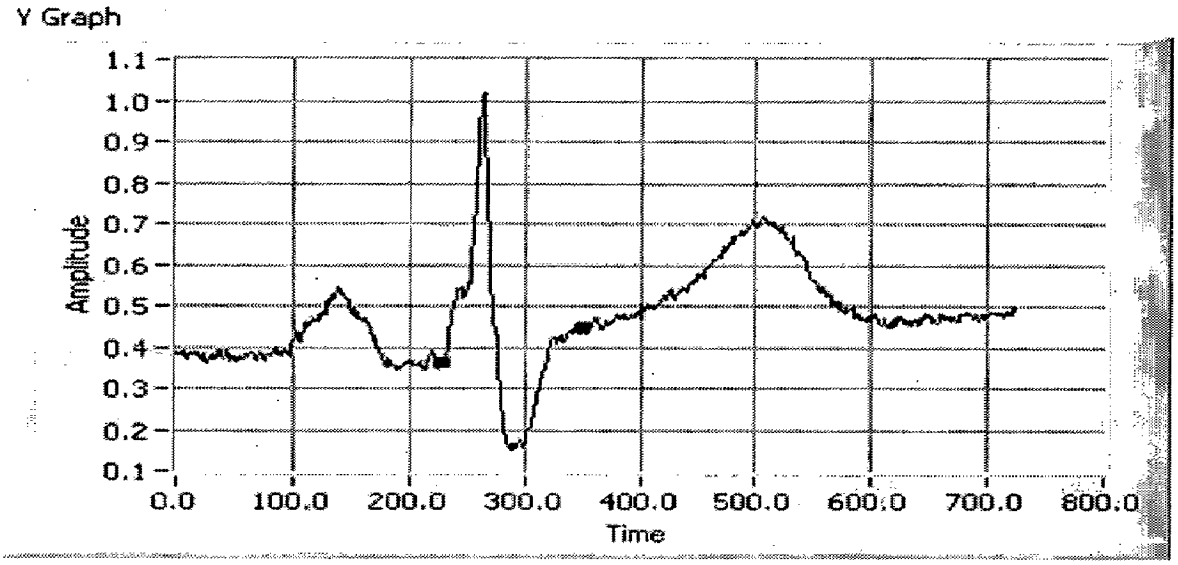


Figure 6.8 Waveform of the Y lead of the myocardial patient

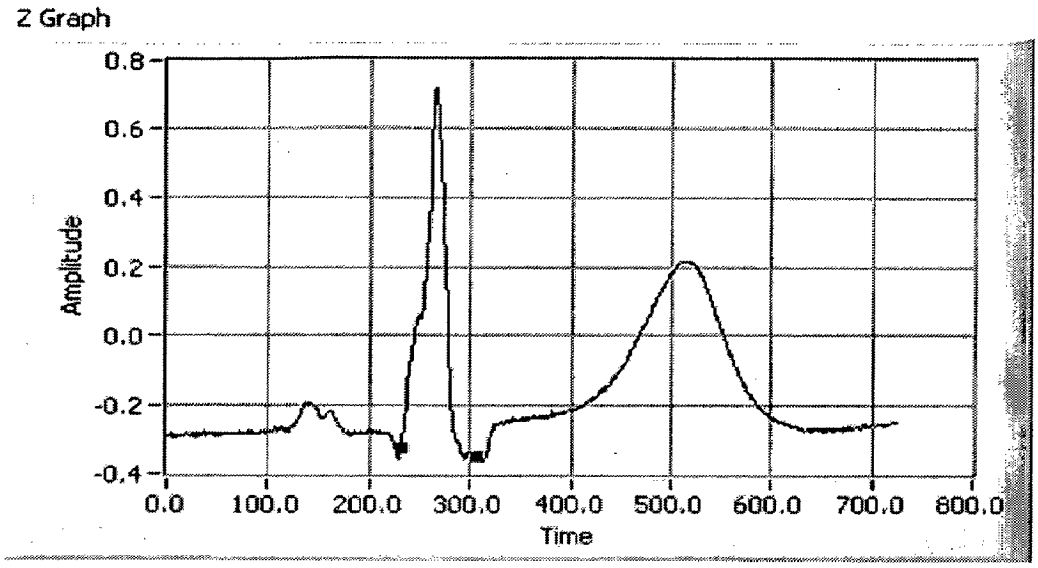


Figure 6.9 Waveform of the Z lead of the myocardial patient

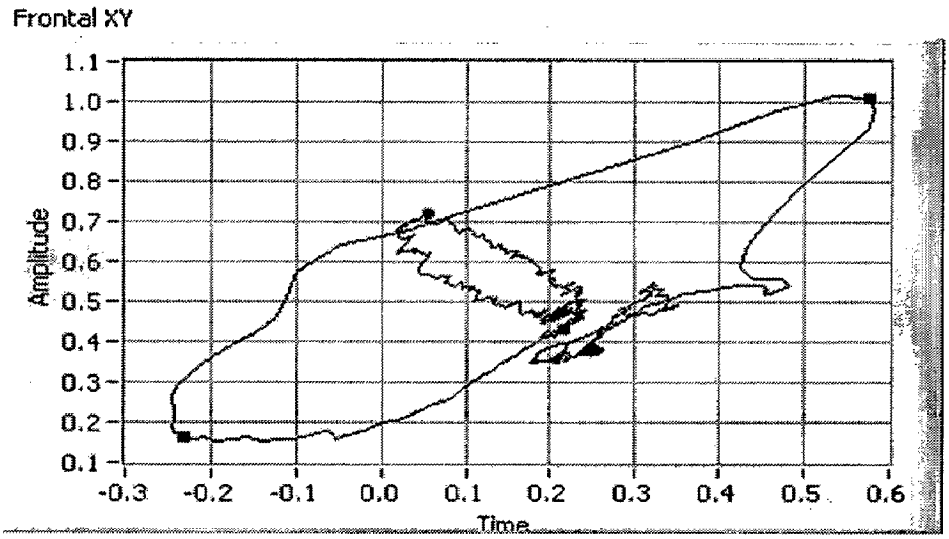


Figure 6.10 Frontal plane showing end points of majoraxes of QRS and T loops

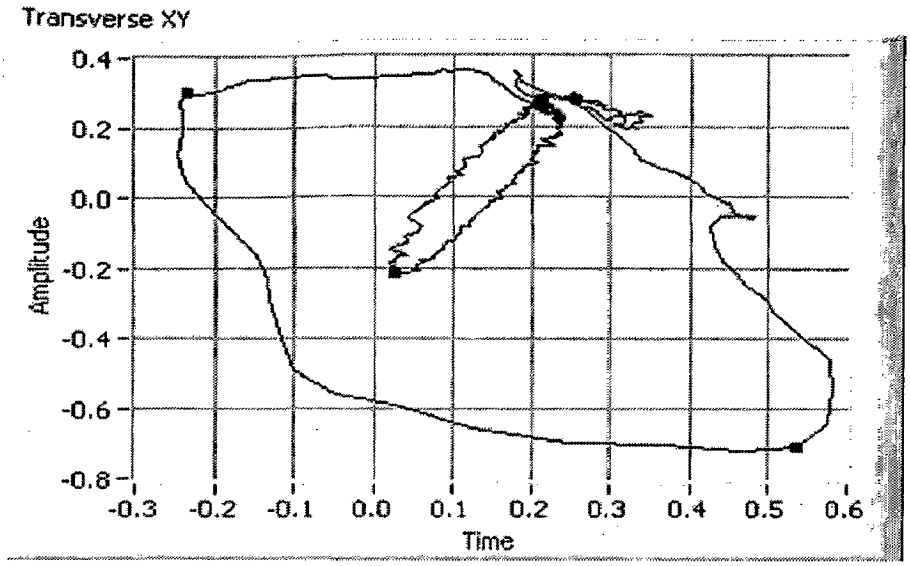


Figure 6.11 Transverse Plane of the Myocardial subject

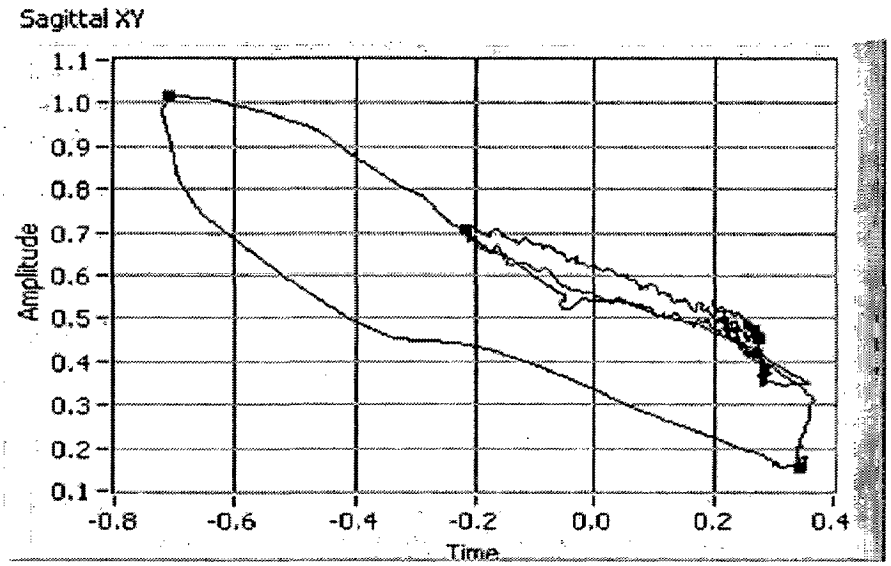


Figure 6.12 Left Sagittal Plane of the Myocardial patient

6.2 T-Loop Parameters

The 3 original parameters (RMMV, MA, DEA) and the 3 modified parameters (RMMVm, MAm, DEAm) have been computed in the 2 groups of patients selected from the database.

Following are the values of the modified parameters for the database computed.

Normal:

Patient	RMMVavg (mv)	MA-avg (degrees)	DEA-avg (degrees)
104	3.32	81.26	33.9
105	3.41	20.08	67.96
116	3.35	119.8	30.28
117	3.18	118.5	8.82
121	2.75	55.18	68.67
122	3.96	118.49	25.56
131	2.97	118.65	52.22
155	2.75	74.88	51
156	2.81	74.17	11.01
165	3.21	112.42	34.78
166	3.22	106.37	62.38
169	2.77	65.36	41.95
170	2.85	20.11	34.05
174	4.21	50.63	49.97
180	3.72	8.87	17.46
182	3.26	91.5	28.14
184	2.98	106.78	28.18
185	3.58	19.06	72.09
214	2.72	119.5	29.25
229	3.42	112.71	22.78
233	3.69	111.47	19.58
234	4.18	66.1	24.92
235	2.84	99.05	56.13
236	3.65	114.69	28.72
237	3.75	114.03	10.58
240	2.59	19.03	50.71
241	3.17	96.55	31.52
242	3.85	118.97	40.72
243	2.97	36.34	40.03
244	4.06	46.72	41.1
245	3.09	124.6	19.26
246	3.59	119.21	32.54
247	2.82	95.53	53.64
248	3.4	65.84	30.04
251	2.77	117.35	46.87
266	3.24	119.8	36.87
267	4.01	71.44	46.37
275	3.12	90.12	52.11
276	2.78	86.75	46.94
277	4.78	54.56	74.14
279	4.41	91.46	19.62

Myocardial:

Patient	RMMVavg (mv)	MA-avg (degrees)	DEA-avg (degrees)
2	2.01	134.64	46.52
4	1.97	158.58	71.78
5	1.64	143.42	76.94
6	2.18	159.89	63.45
7	1.65	146.28	65.49
8	1.83	139.47	43.98
9	2.55	143.95	79.02
11	1.82	159.73	43.62
13	1.53	166.86	76.85
14	1.47	137.64	49.11
16	2.57	161.56	51.32
17	2.56	170.95	80.49
19	2.15	193.34	72.94
20	1.54	161.98	63.13
21	2.22	170.04	19.85
22	2.4	148.62	42.89
24	2.25	131.83	38.92
25	1.8	161.65	80.23
26	2.33	167.91	60.98
27	2.04	144.54	53.16
28	1.86	138.42	38.59
30	2.29	149.7	60.16
31	2.42	129.25	57.04
32	1.59	142.35	68.4
33	2.2	132.99	37.05
35	2.11	147.92	24.05
36	2.29	151.33	42.73
37	2.39	129.79	48.74
38	1.86	128.5	64.81
39	1.86	153.92	64.59
42	2.14	135.84	57.62
44	1.99	154.54	50.16
46	1.87	175.8	70.06
45	1.92	169.02	56.94
47	1.53	158.28	90.82
48	1.9	156.82	72.69
49	2.58	146.1	24.12
50	2.17	165.23	44.36
51	1.64	166.65	41.03
52	2.49	130.21	54.34
54	2.67	131.12	33.06
55	2.21	153.38	71.91
57	2.37	130.01	29.33

				59	2.31	164.37	26.92
				60	1.72	140.64	51.82
				61	1.54	126.89	66.49
				62	1.42	129.87	66.7
				65	2.51	162.66	78.47
				68	1.24	215.73	57.27
				69	1.31	177.3	35.81
				70	2.62	191.15	23.98
				72	2.57	125.79	25.36
				73	2.36	197.33	19.82
				74	1.93	186.58	56.51
				75	2.26	126.28	48.11
				125	1.28	204.11	91.25

The mean values and the standard deviation of the 6 parameters in the group of healthy subjects and myocardial disorders are reported in Table 6.1.

6.2.1 Modified RMMVavg for healthy subjects and myocardial:

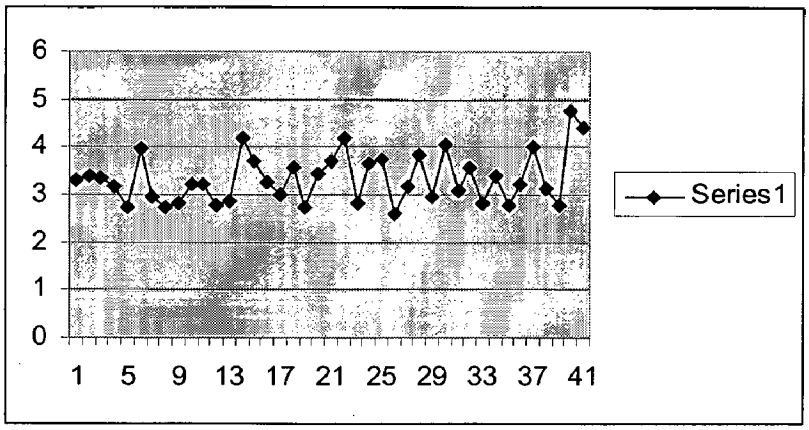


Figure 6.13 Values of RMMVavg of the healthy subjects

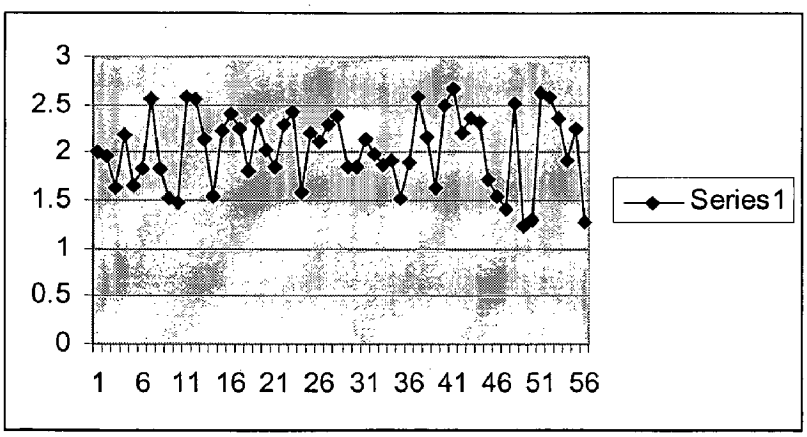


Figure 6.14 Values of RMMVavg of the myocardial patients

6.2.2 Modified MAavg for healthy and myocardial:

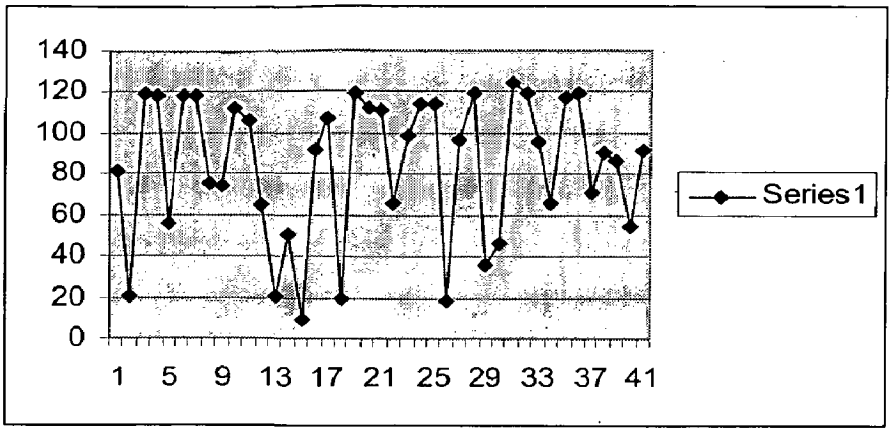


Figure 6.15 Values of MAavg of the healthy subjects

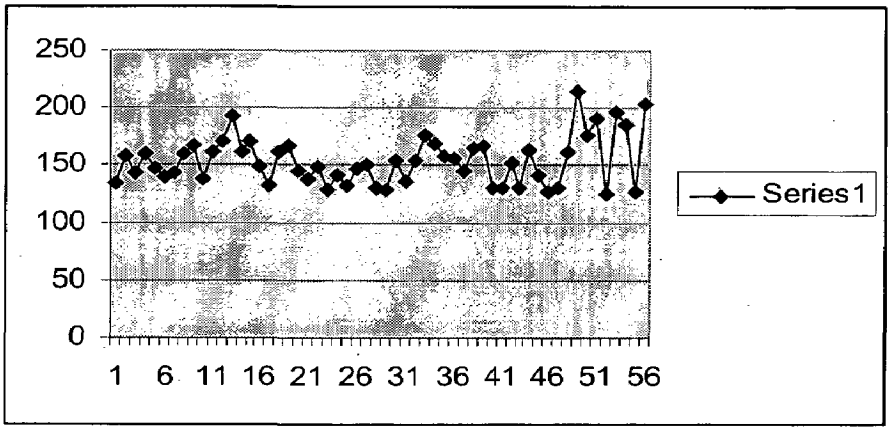


Figure 6.16 Values of MAavg of the myocardial patients

6.2.3 Normal and Abnormal ranges of the T loop parameters:

Parameter	Healthy	Myocardial
RMMV	3.53 ± 0.71	2.43 ± 0.63
MA	80.29 ± 34.29	94.18 ± 48.38
DEA	63 ± 15.95	60.41 ± 20.73
RMMVm	3.43 ± 0.52	1.98 ± 0.41
MAm	74.47 ± 35.22	148.62 ± 38.65
DEAM	85 ± 21.5	88.89 ± 26.79

Table 6.1 Mean ± SD of the 6 T Loop parameters

6.3 Statistical Analysis

6.3.1 Cross Comparison (t – test)

The *t*-test is the most commonly used method to evaluate the differences in means between two groups. The *t* test compares one variable between two groups. The *p*-level reported with a *t*-test represents the probability of error involved in accepting our research hypothesis about the existence of a difference. Technically speaking, this is the probability of error associated with rejecting the hypothesis of no difference between the two categories of observations (corresponding to the groups) in the population when, in fact, the hypothesis is true. The P value is the probability, with a value ranging from zero to one. If the P value is small, we can conclude that the difference between sample means is unlikely to be a coincidence. Instead, we can conclude that the populations have different means. The accepted P level is to be less than 0.0001 for the parameter to statistically significant.

Cross-comparison between healthy and the other group performed with t-test for all the parameters is shown in Table 6.2 .

From this statistical analysis it is concluded that the mean values of the three modified parameters of the healthy subjects group are statistically different from those of myocardial group. In addition, the modified parameters proved to yield higher discriminative power.

Parameter	Healthy vs Myocardial disorder
RMMV	P < 0.0001 -- statistically significant
MA	P = 0.1247 -- not significant
DEA	P = 0.8326 -- not significant
RMMVm	P < 0.0001 -- statistically significant
MAm	P < 0.0001 -- statistically significant
DEAm	P = 0.2033 --not significant

Table 6.2 Values of P for the T-parameters

6.4 Analysis on CSE database.

Further analysis is done on the CSE database on 64 subjects with unknown details of their health based on the previous analysis done using PTB Diagnostic database and comparing with the normal ranges.

Patient	RMMV (mv)	MA (degrees)	DEA (degrees)
cse1	1.655	155.902	35.203
cse2	1.952	173.933	32.623
cse3	1.824	70.665	41.326
cse4	2.437	85.83	34.163
cse5	1.824	157.536	79.403
cse7	2.331	112.088	30.667
cse8	3.016	32.591	28.696
cse9	1.919	107.198	15.435
cse10	1.411	53.069	8.904
cse11	3.082	99.998	25.765
cse12	2.389	169.095	105.27
cse13	3.111	177.271	117.256
cse14	1.244	173.626	31.817
cse15	4.422	120.279	15.658
cse16	1.677	141.594	24.644
cse17	2.063	174.934	86.529
cse18	2.418	163.49	20.437
cse19	2.598	133.681	15.941
cse20	1.538	178.86	86.437
cse21	4.044	145.779	9.356
cse22	1.402	107.334	110.04
cse23	3.383	74.939	71.607
cse24	4.62	165.839	144.767
cse25	1.878	164.524	46.748
cse26	2.069	19.067	45.076
cse27	2.458	178.437	7.642
cse28	2.557	168.585	74.069
cse29	1.316	177.46	74.366
cse30	1.983	10.911	66.714
cse31	2.702	11.418	93.045
cse32	1.672	93.278	13.779
cse33	1.504	156.418	56.41
cse34	3.241	154.764	17.776
cse35	1.525	147.666	65.255
cse36	3.612	173.736	65.832
cse37	2.983	114.829	10.967
cse38	2.556	171.717	29.251
cse39	1.743	173.085	51.262
cse40	1.409	176.318	137.605
cse41	2.107	75.409	69.13

cse42	2.686	60.762	60.205
cse43	2.685	162.535	82.409
cse44	3.036	110.433	13.888
cse45	1.594	168.331	42.166
cse46	1.904	161.606	72.375
cse47	1.71	163.51	65.134
cse48	2.614	85.799	112.301
cse49	2.329	132.241	32.299
cse50	2.821	53.42	61.225
cse51	1.893	144.451	25.719
cse52	1.746	136.9	38.71
cse53	1.654	34.218	20.714
cse54	1.362	148.095	14.942
cse55	3.045	100.308	18.742
cse56	1.756	173.835	31.889
cse57	1.305	90.994	135.788
cse58	2.861	116.543	14.939
cse59	2.894	79.486	23.133
cse60	3.122	47.465	16.311
cse61	2.72	106.488	23.639
cse62	2.007	95.437	37.448
cse63	1.247	128.915	34.734
cse64	2.748	107.381	67.58
cse65	1.562	175.735	93.018

Total number of subjects: 64

From the analysis:

Patients that are termed as abnormal: (RMMV<2.5 && MA > 135) or (MA > 170)

1,2,5,12,13,14,16,17,18,20,25,27,28,29,33,35,38,39,40,43,45,46,47,51,52,54,56,65.

Total: 29

Patients that are termed as normal: (RMMV >2.4 && MA < 130)

4,8,11,15,19,23,31,37,42,44,48,50,55,58,59,60,61,64 --------- Total: 18

Patients having very low RMMV but normal MA:

3, 10, 22, 32,53,57,63 ------- Total: 7

Patients that are on border line:

7,21,24,26,30,34,36,41,49,62

The following result has to be authenticated by a cardiologist to confirm the analysis.

The derived Vectorcardiogram obtained by Inverse Dower is observed to be in close proximity with the Frank VCG and T loop parameters are computed with respect to all the planes. The modified T loop parameters is found to be statistically significant than the initial parameters where only RMMV is found to be significant. In the modified parameters DEA is found to be not statistically significant but MA (maximum angle between QRS-T loop axes) and RMMV is found to contain higher discriminative power and is a strongest predictor of cardiac mortality in the elderly compared to classical cardiovascular risk factors and other ECG risk indicators. Spatial characteristics of ventricular repolarization has been shown to assess ECG qualities that are different from conventional ECG parameters, thus have additional value. The normal limits of healthy vs myocardial subjects is analyzed from the work carried on the database of 100 subjects consiting of 41 healthy and 58 mycardial. Thus, the performance of an ECG computer program for myocardial diseases can be improved by incorporating both ECG and VCG classificatory knowledge with the same 12 lead system.

Further work can be carried out by considering the geometric factors for different individuals based on weight and height to obtain the derived VCG and to further analyze the T loop complexity with few more parameters together with a neural network for decision making on a large database to validate this promising computerized approach, which could be supplementary to other methods attempting to determine heterogeneity of repolarization based on scalar ECG measurements.

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APPENDIX A

PTB DIAGNOSTIC DATABASE

Physikalisch-Technische Bundesanstalt (PTB), the National Metrology Institute of Germany, has provided this compilation of digitized ECGs for research, algorithmic benchmarking or teaching purposes to the users of PhysioNet. The ECGs were collected from healthy volunteers and patients with different heart diseases by Professor Michael Oeff, M.D., at the Department of Cardiology of University Clinic Benjamin Franklin in Berlin, Germany. His current address is:

Department of Cardiology
Klinikum Brandenburg
14770 Brandenburg, Germany

The ECGs in this collection were obtained using a non-commercial, PTB prototype recorder with the following specifications:

- 16 input channels, (14 for ECGs, 1 for respiration, 1 for line voltage)
- Input voltage: ±16 mV, compensated offset voltage up to ± 300 mV
- Input resistance: 100 Ω (DC)
- Resolution: 16 bit with 0.5 μV/LSB (2000 A/D units per mV)
- Bandwidth: 0 - 1 kHz (synchronous sampling of all channels)
- Noise voltage: max. 10 μV (pp), respectively 3 μV (RMS) with input short circuit
- Online recording of skin resistance
- Noise level recording during signal collection

The database contains 549 records from 294 subjects (each subject is represented by one to five records). Each record includes 15 simultaneously measured signals: the conventional 12 leads (i, ii, iii, avr, avl, avf, v1, v2, v3, v4, v5, v6) together with the 3 Frank lead ECGs (vx, vy, vz). Each signal is digitized at 1000 samples per second, with 16 bit resolution over a range of ± 16.384 mV. On special request to the contributors of the database, recordings may be available at sampling rates up to 10 KHz.

Within the header (`.hea`) file of each ECG record is a detailed clinical summary, including age, gender, diagnosis, and where applicable, data on medical history, medication and interventions, coronary artery pathology, ventriculography, echocardiography, and hemodynamics. The diagnostic classes of the subjects are summarized below:

Diagnostic class	Number of subjects
Myocardial infarction	148
Cardiomyopathy/Heart failure	18
Bundle branch block	15
Dysrhythmia	14
Myocardial hypertrophy	7
Valvular heart disease	6
Myocarditis	4
Miscellaneous	5
Healthy controls	54