Computer-Aided Electrocardiogram Analysis

by

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DEDICATION

The author dedicates this thesis particularly to his parents, and , for their patience, teaching, encouragement and love, without which this degree would not have been sought.

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Chapter I

INTRODUCTION

The electrocardiogram is a very essential and basic tool for any cardiac study, and it has been used by physicians and cardiologists throughout the medical profession for a number of years. However, analyzing an electrccardiogram is a very tedious task and it is not surprising that there is considerable variation among cardiologists' interpretations. Studies[1], that have been made to assess the human variability in vector-cardiogram and electrocardiogram interpretations, show that cardiologists not only disagree as to whether a given tracing should be labeled as normal or abnormal, but also often failed to agree with all their own previous diagnosis on reviewing the records a second time. Thus computer-assisted interpretations should be able to decrease the degree of variations and increase the efficiency fo cardiologists. Computer-aided measurement and analysis of electrocardiographic data offer several advantages over traditional human interpretations. These advantages[2] include:

1. Rapidly measuring and analyzing all quantitative aspect of an electrocardiogram with accuracy, repeatibility, and consistency unattainable by man;

2. Accurate, consistent and more objective diagnostic criteria based on the analysis of large volumes of electro-cardiogram data using complex mathematical methods;

3. Elimination of time consuming and tedious processing of electro-cardiographic records by human operators, thus decreasing the probability of human errors;

4. Enabling long-distance(remote) diagnostic processing through common voice grade telephone network;

5. Contribution to clinical research by easy retrieval and analysis of electro-cardiographic data from various sources; and finally

6. Enable better control of heart diseases through computer-assisted mass electro-cardiogram screening.

Electronic technology, particularly micro-processors, has made it feasible to produce a low-cost electro-cardiogram computer system, which is affordable by most physicians and hospitals. Thus, the objective of this project is to develop the software for a computer-assisted

electrocardiogram system, aiming at portability, reliablity, ease of operation, generality, hardware independence and providing as much useful information for the user as possible.

Chapter II

FUNDAMENTALS OF ELECTROCARDIOGRAPHY

2.1 ELECTRICITY OF HEART

The contraction of any muscle is associated with an electrical change called 'depolarization' which can be detected by electrodes attached to the surface of body. Amplification and filtering through electronic circuitry provide signals which can be recorded for detailed analysis. Electrocardiogram is the term used for the electrical changes of heart muscles.

The heart has four chambers, the left and right atrium and left and right ventricles. However from an electrical point of view, it can be thought of as having two chambers because the two atria contract together and followed by contraction of both ventricles.

A cardiac cycle of an electrocardiogram can be subdivided into several waveforms , each of which represents the polarization or depolarization of a particular portion of the heart muscle. A figure of the heart and its associated muscles is shown in Figure 1[18].

The depolarization process of the heart originates in a special area of the right atrium called the "sino-atrial"(SA) node . The cells of the SA node demonstrate a

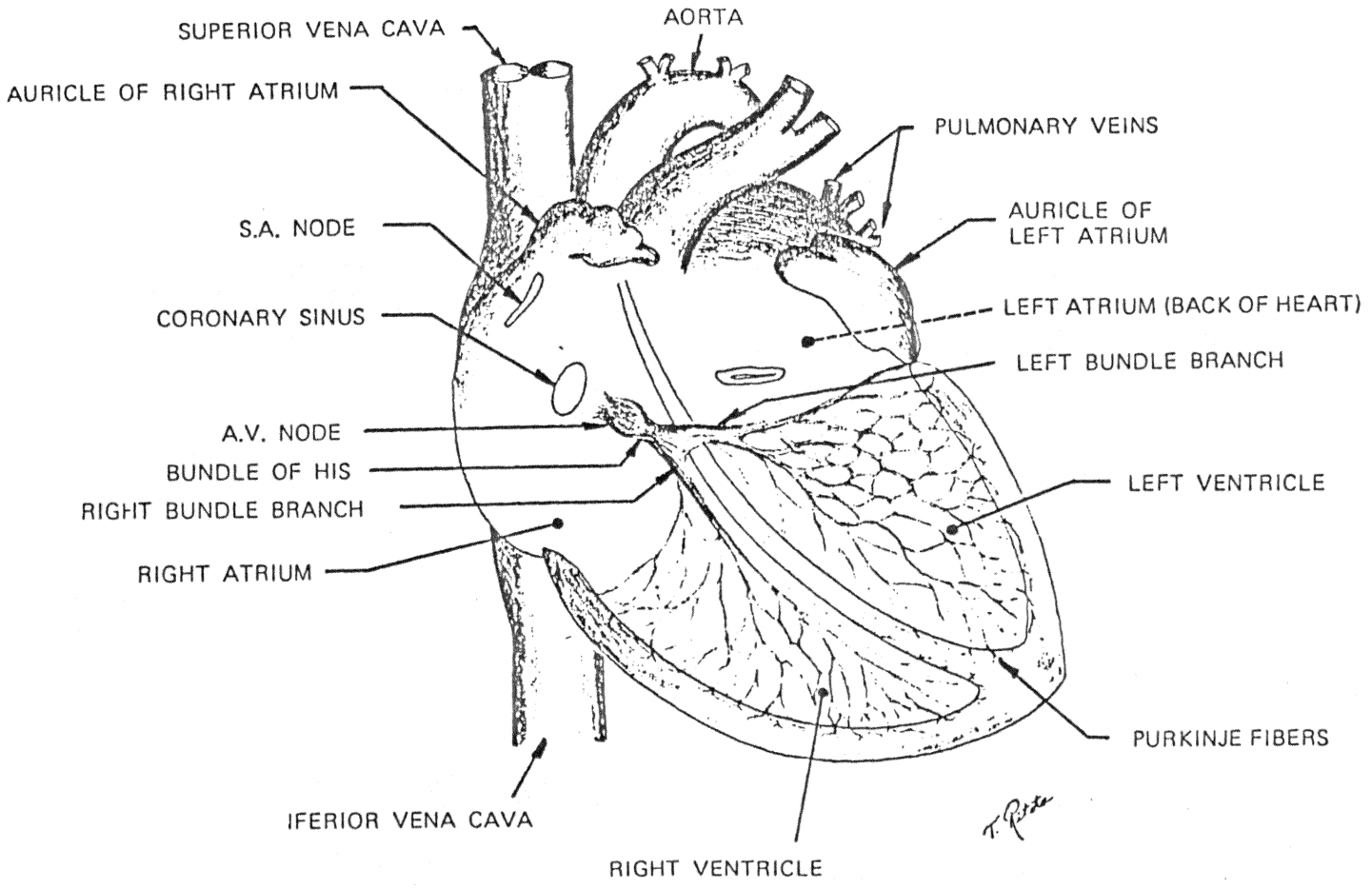


Figure 1: Chambers of Heart and its associated muscles [18]

spontaneous depolarization, which normally occurs sixty to one hundred times a minute in a rhythmic fashion. Because of this capability the SA node is designated as the pacemaker. However, all areas of the heart muscles have the ability to serve this function(an inherent property of the cardiac tissues) but only assume this role under abnormal conditions, such as the failure of the SA node.

The process of depolarization is propagated over the walls of the atria and taken over by the atrioventricular(AV) node, which is located near the top of the ventricular septum. Here, after depolarization of the atria, a physiological delay in conduction takes place, resulting in a brief electrical "delay". The AV node, in turn, activates the bundle of HIS which divides into two branches serving the right and left ventricles. Through these branches the stimulus finally passes to the Purkinje fibers, which directly activate the inner-most layer of the myocardium.

Depolarization proceeds from the inner surface of the ventricular walls toward the outer surface, as well as from the apex to the base of ventricles. When one region of the ventricles becomes electronegative with respect to the remainder it acts as a dipole. The equivalent dipoles for

each region can be represented by vectors and their resultant is the mean electrical axis of the the heart.

Contraction of the atria causes the deflection in the ECG called 'P' wave, which is small in magnitude due to the relatively small electrical change in the small muscle mass of the atria. The contraction of ventricles lead to the wave called 'QRS' complex. The 'T' wave is caused by the repolarization of the ventricles. The 'U' wave, usually too small in magnitude to be detected, is caused by the after-potential following the repolarization of the ventricles. Table 1 illustrates the sequence of events in a complete cardiac cycle.

2.2 BASIC ELECTRO-CARDIOGRAPHIC PATTERN AND DEFINITION

A standard electrocardiagram waveform is shown in detail in Figure 2[18]. The normal ECG waveform for a single cardiac cycle can be sub-divided into four main waveforms(or deflections), which are designated as the P wave, QRS complex, T wave and the usually undetected U wave. The letters were chosen arbitrarily and have no additional meaning. When the heart is at rest the electrocardiogram displays a straight horizontal line, called the-electric line or baseline. The alternating current developed by the

TABLE I

Electrical Events

Sequences of Electrical Events of Cardiac Cycle	ECG Representation
1. Impulse from SA node	Not visible
2. Depolarization of the atria	P wave
3. Depolarization of AV node	Iso-electric
4. Repolarization of the atria	Usually obscurred by QRS complex
5. Depolarization of ventricles a) Intraventricular septum b) right and left ventricles	QRS complex a) Initial Portion b) Central and terminal portions
6. Activated state of the ventricles immediately after depolarization	ST segment: iso-electric
7. Repolarization of ventricles	T wave
8. After potential following repolarization of the ventricles.	U wave (usually too small to be visible)

myocardial activity is recorded as upward or downward deflections superimposed on the baseline.

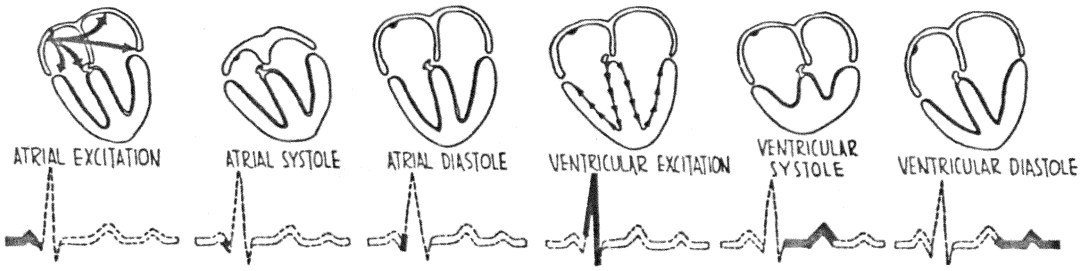
2.2.1 P Wave

The P wave results from depolarization of the atrial muscle. The initial portion of the P wave reflects right atrial depolarization and the terminal portion indicates left atrial depolarization. The duration of the P wave is normally less than one-tenth of a second and decreases somewhat as the heart rate increases.

2.2.2 QRS Complex

The QRS complex represents ventricular depolarization and normally follows the 'P' wave and P-R segment. The QRS complex may have different configurations as shown in Figure 3[2]. Variations in the number and appearance of the component waveforms in the QRS complex depend upon the lead system used for recording and the electrical axis of the heart. The direction of the electrical axis in normal individuals generally follows the long axis of the heart and is influenced by the anatomical position of the heart.

If the initial deflection of the QRS complex in a given lead is negative, it is called a Q wave. The first positive



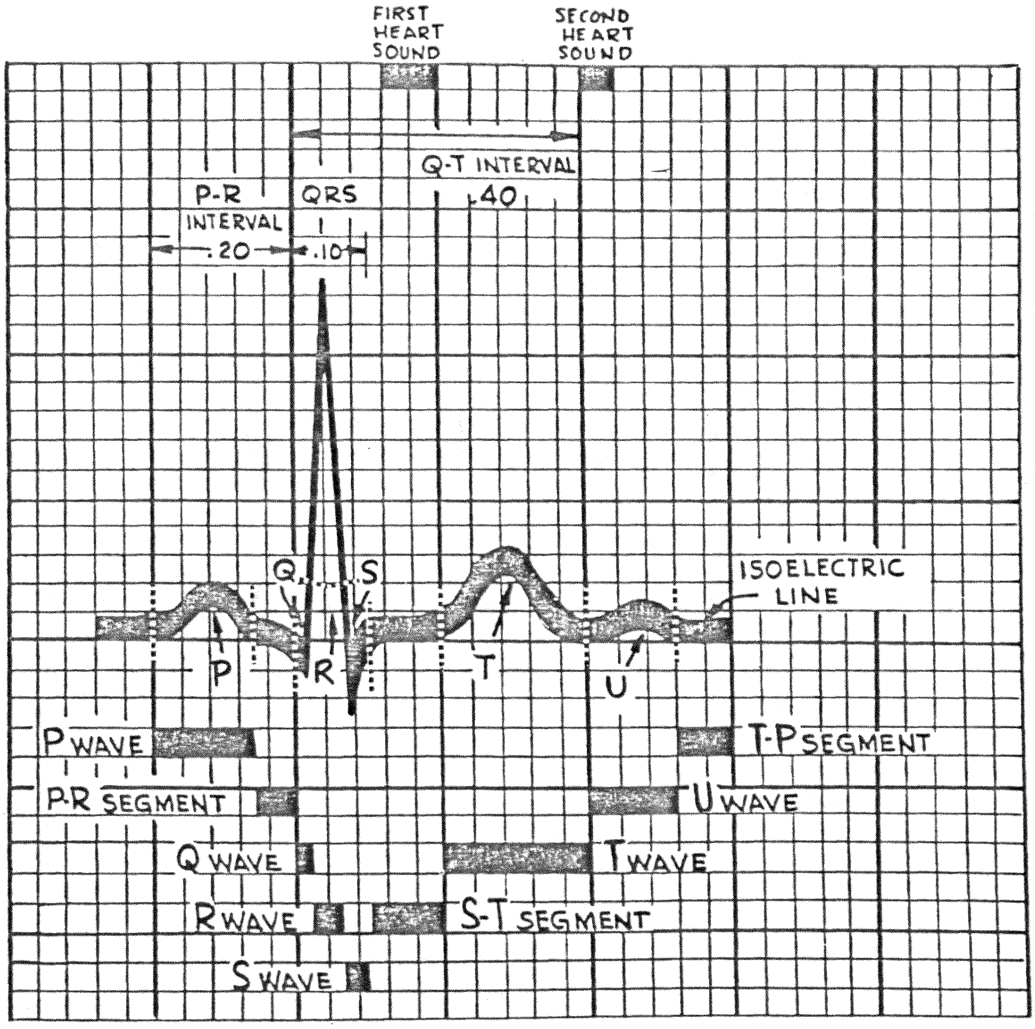


Figure 2: Detail Electrocardiogram Waveform [18]

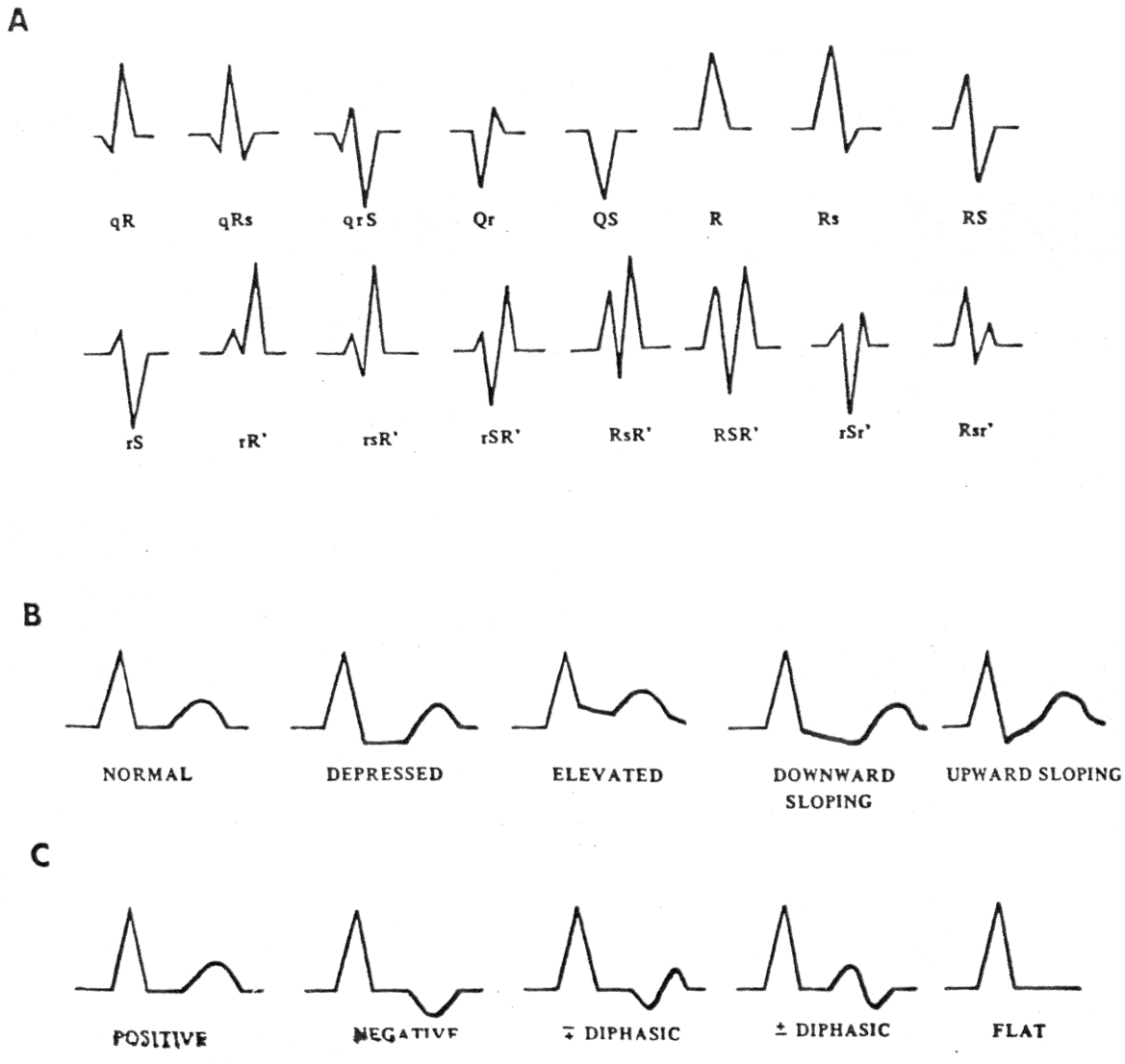


Figure 3: Different QRS Configurations[2]

deflection in the QRS complex is called an R wave; which may or may not preceded by a Q wave. Any negative wave in the QRS complex preceded by an R wave is called an S wave. The QRS complex in the form of single negative waves are referred to as QS waves or QS complexes. Secondary positive deflections after the S wave are labeled R'(R prime) and secondary negative waves after the R' are S'(S prime). By convention, capital letters are used when the wave is the major deflection or is at least one-half the amplitude of the major deflection. Minor waves, with less than one-half of the amplitude of the major deflection, are written in lower case letters.

The duration of the normal QRS complex is from 0.05 to 0.10 msec, dependent on placement of recording leads (in the precordial leads, the duration is slightly longer than in limb leads). The amplitude and duration of Q,R and S waves varies from lead to lead and from subject to subject. The normal Q wave duration should not exceed 0.04 sec and the normal Q wave amplitude should not exceed 25 per cent of the amplitude of the succeeding R wave.

2.2.3 T Wave

The T wave is the deflection representing ventricular repolarization and follows the QRS complex and S-T segment.

The polarity of this deflection is usually, but not always, concordant with that of the major QRS deflection. The amplitude of the T wave is correlated in a general way with the height of the preceding QRS. Therefore, unlike the latter complex, in which absolute voltage values are emphasized, evaluation of T amplitudes is based on the QRS/T ratios. The beginning of the T wave cannot be determined in many cases because of the gradual transition of the preceding S-T segment to the T wave.

2.3 SEGMENTS AND INTERVALS

The portions of the iso-electric line between deflections(waves) are called segments and are designated by the letters of the preceding and following waves. A segment together with the preceding or following wave, or both, is called an interval.

2.3.1 PR or PQ Segment

The portion of the electrocardiographic tracing from the end of P wave to the onset of the QRS complex is called a P-Q segment or P-R segment. It is normally iso-electric and sometimes slightly below the baseline. A considerable depression of this segment is called an atrial repolarization wave (Ta).

2.3.2 S-T Segment

The S-T segment lies between the end of the QRS complex and the beginning of the T wave and corresponds to the period of complete ventricular depolarization. It is elevated or depressed in comparison with that portion of the baseline between the termination of the T wave and the beginning of the P wave.

2.3.3 T-P Segment

The T-P segment lies between the end of the T wave and the beginning of the P wave in the adjacent cycle. This segment is almost always iso-electric and is usually taken as a reference for a baseline. However, sometimes this segment is not equal to the baseline because of the appearance of the small U wave between the T and P waves.

2.3.4 J Point

The point where the QRS ends and S-T segment begins is referred to as the J(junction) point.

2.3.5 QRS Interval

This is the measurement of the total ventricular depolarization time. It is measured from the onset of the Q wave (or R if no Q is visible) to the termination of the S wave. The upper limit is normally about 0.11sec.

2.3.6 PQ or PR Interval

The P-Q or P-R interval is measured from the beginning of the P wave to the beginning of the QRS ccmplex. This interval represents the time required for depolarization of the atria, as well as that involved in the physiological delay in the conduction of the impulse by the atrioventricular node. The P-Q interval is from 0.10 to 0.20 second and is partly dependent on the heart rate; a higher heart rate is accompanied by a shorter P-Q interval.

2.3.7 Q-T interval

The Q-T interval is measured from the onset of the Q wave to the end of T wave and denotes the time of electrical systole (time elapsing during depolarization and repolarization of the ventricules). The Q-T interval varies with the heart rate and can be calculated according to the following formula : Q-T (seconds)=o.4 * SQRT(R-R interval) (seconds)

2.3.8 R-R Interval

The R-R interval denotes the time between successive R peaks.

2.3.9 P-P Interval

The P-P interval is measured from the onset of a P wave to the onset of the immediate following P wave.

2.4 LEAD SYSTEMS

There are a number of different lead systems for acquiring electrocardiograms. The Frank Orthogonal Lead system and the Twelve Lead system are most commonly used for clinical purposes and are the only systems discussed in this thesis.

2.4.1 The Frank Orthogonal Lead System

The Frank orthogonal lead system is widely used because it provides orthogonality, relatively simple and uncritical electrode placement, and a small number of electrodes. Only eight electrodes are required, seven electrodes are used directly and an eighth electrode connected to the right leg for a ground or reference electrode. The signal from the heart is picked up by the electrodes and fed through a resistor network, as shown in Figure 4[1]. The output signals, Vx,Vy,Vz can be seen from the following equations:

$$Vx=0.610Va + 0.17Vc - 0.781Vi$$

$$Vy=0.655Vf + 0.345Vm - 1.000Vh$$

$$Vz=0.133Va + 0.736Vm - 0.264Vi - 0.374Ve - 0.231Vc$$

The body locations to which these leads are connected are shown below:

1. I -right mid-axilary line in the fouth intercostal space.
2. A - left mid-axilary line in the fouth intercostal space.
3. E - anterior midline.
4. M - posterior midline.
5. C - V3 precordial lead location
6. F - left leg.
7. H - back of neck.

The placement of electrodes and the resistor network is shown in figure 4[1]. The value of the resistance 'R' is not critical but is normally chosen between 5000 to 10,000 Ohms.

2.4.2 Twelve Lead System

The twelve lead system is widely used in clinical practice because it has been used for a long time and the medical professionals are very familiar with it. The twelve lead system consists of three sub-systems.

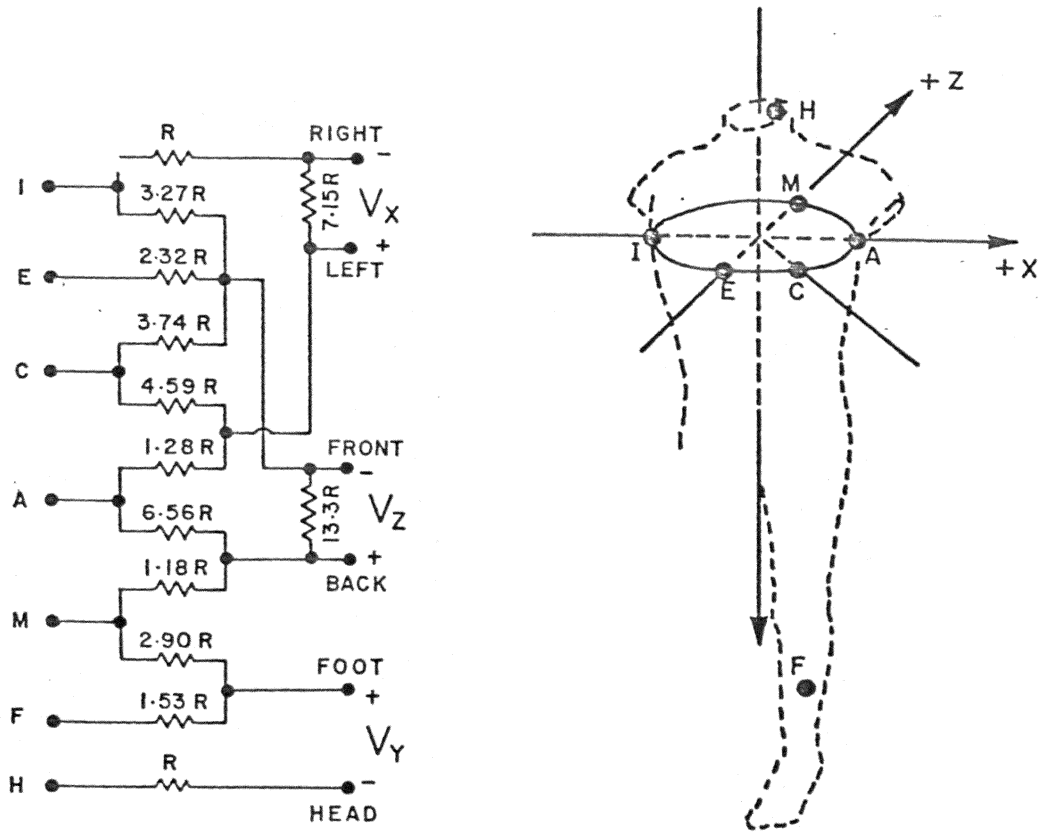


Figure 4: Frank Orthogonal Leads and the associated Network[1]

Three bipolar extremity leads (standard limb leads) L1,L2,L3 : The bipolar leads measure the difference of potential between three pairs of electrodes placing in the following manner:

1. Lead L1 : left arm (+) to right arm (-)

2. Lead L2 : left leg (+) to right arm (-)

3. Lead L3 : left leg (+) to left arm (-)

These bipolar leads are arranged in such a manner that the amplitude of deflection in L1 plus that of L3 is equal to that in lead L2 according to Einthoven's Law[1].

Three unipolar extremity leads, VR, VL and VF are measured with respect to a central terminal in the electronic apparatus for measurement. The locations on the body for these extremity leads are:

1. VR : right arm

2. VL : left arm

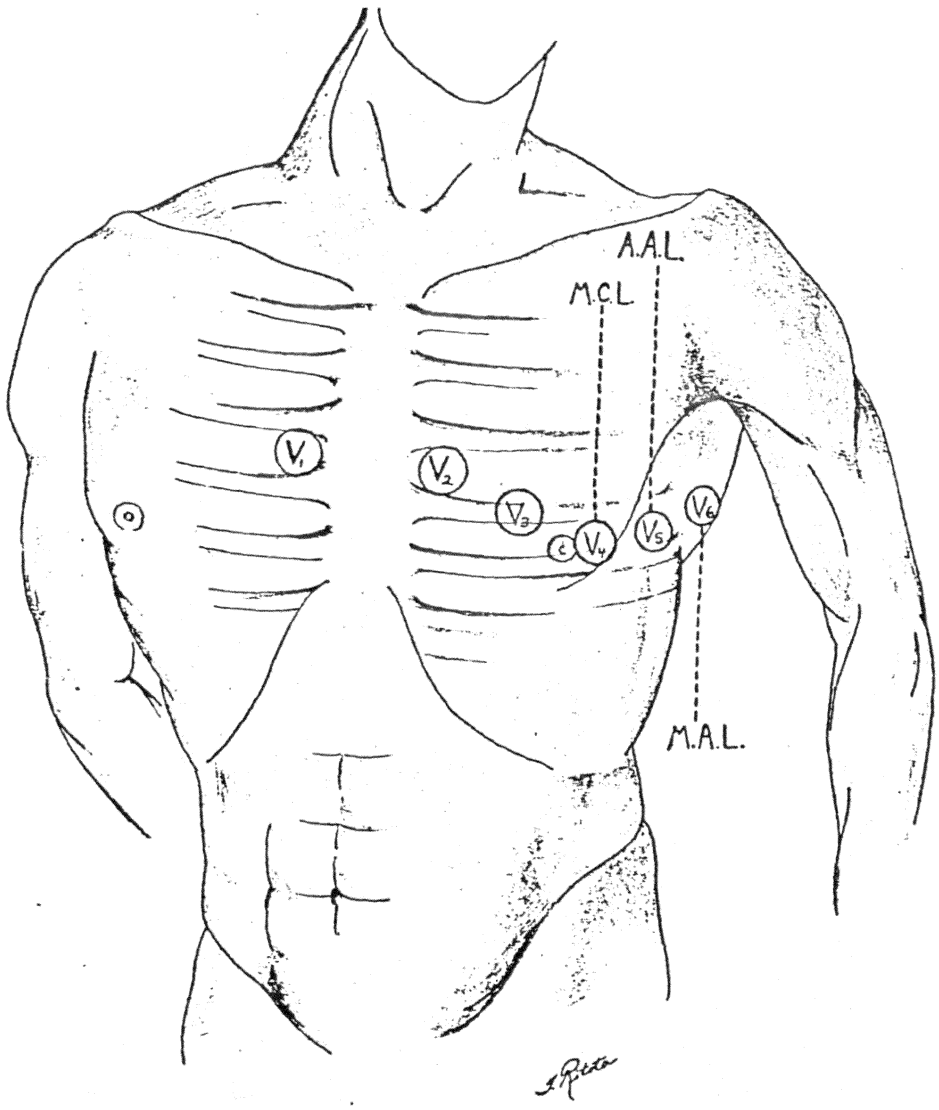
3. VF : left leg

Six unipolar precordial(chest) leads,V1 to V6, are similar to the unipolar extremity leads and are also measured with respect to a central terminal. The body location for electrode placement for the precordial leads are shown as follows (see Figure 5[18]):

1. V1 : 4 th intercostal space on the right edge of sternum
2. V2 : 4 th intercostal space on the left edge of sternum
3. V3 : midway between V2 and V4
4. V4 : 5 th intercostal space in the midclavicular line
5. V5 : anterior axillary line at the level of V4
6. V6 : midaxillary line at the level of V4.

2.5 AXIS CONFIGURATION

The axis co-ordination of the ECG system is defined a little bit different from the traditional geometrical co-ordinate system. The ECG co-ordinate system is shown in Figure 6[1].



The landmarks for the precordial leads (V leads)

V_1 – 4th interspace at the right border of the sternum
V_2 – 4th interspace at the left border of the sternum
V_3 – left parasternal line midway between V_2 and V_4
V_4 – 5th interspace in the left midclavicular line
V_5 – in the anterior axillary line at the level of V_4
V_6 – in the midaxillary line at the level of V_4 and V_5
MCL – Midclavicular line. AAL – Anterior axillary line. MAL – Midaxillary line.

Figure 5: Electrode Placement for Precordial Leads[18]

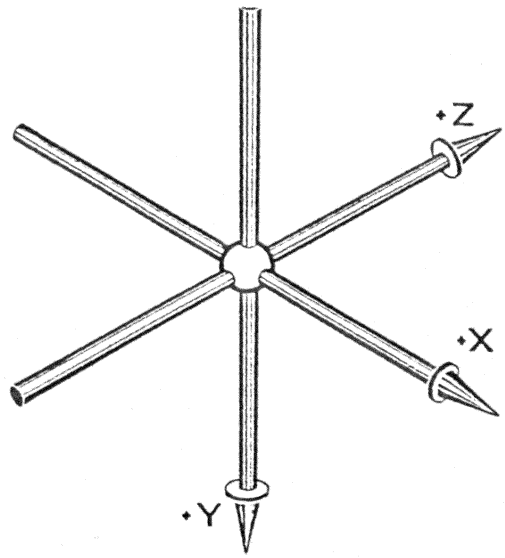


Fig. 1-109. A system of rectangular (orthogonal) coordinates representing the axes of leads used for recording the spatial heart vector.

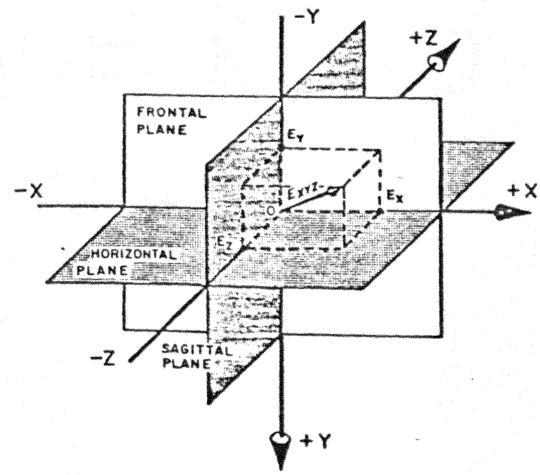


Figure 6: ECG X-Y-Z Configuration[1]

2.6 GENERAL HARDWARE CONFIGURATION

In general, ECG computer-assisted system will consist of the following items:-

1. Electrodes
2. Resistive network
3. Instrumentation amplifier(s)
4. Filtering amplifier(s)
5. Gain amplifier(s)
6. ECG output (strip chart and/or CRT output)
7. Multiplexer
8. A/D and D/A convertors
9. Computer and or processors
10. Computer analysis outputs

A block diagram for a generalized computer-assisted ECG system is shown in Figure 7.

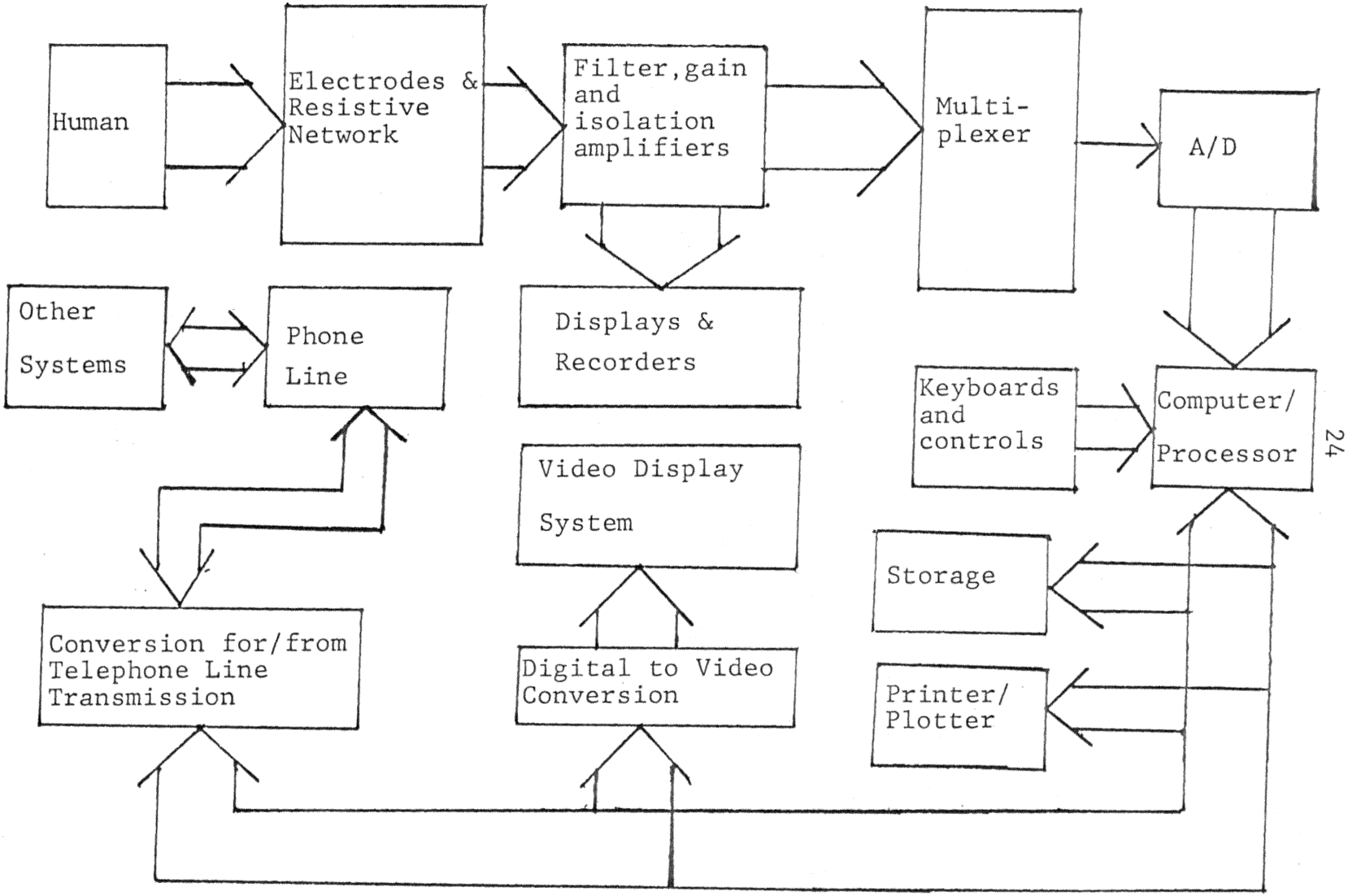


Figure 7: Block Diagram for Computer Assisted ECG System

Chapter III

SAMPLING AND FILTERING

3.1 SAMPLING AND DATA PRECISION

The ECG data[1] used in this thesis was acquired at a sampling rate of 500 samples per second. Other sampling rates (i.e. 250, 125) have been obtained by taking alternate samples from the 500 samples per second data. The algorithm presented in this thesis was tested for sample rate of 500, 250 and 125 samples per second. For sampling rates of 500 and 250, the result are satisfactory. The sampling rates of 125 or less tend to vary the magnitude of the waveform and cause delay, thus affecting the derivation of important parameters. As discussed in the literature review section of this thesis, results obtained by Alan S. Berson[3] showed that significant changes occur in both the amplitude and duration when the sampling rate drops below 167 hertz. This is as expected since information theory requirements dictate sampling rates which are at least twice the maximum frequency component present in the signal. Since past

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[1]The ECG data is obtained as tape from Veterans Administration Hospital in Washington, D.C. The data tape is recorded on 9 tracks at 800 bpi, at 16 bits per word and 60 words per physical record. Alphmeric data is recorded in ASCII and negative integers are expressed in two's complement.

researchers have shown that the maximum useful frequency component of the ECG signal is 100 hertz, thus a sampling rate of 200 samples per second will be sufficient for most application.[3,9]

The accuracy of the analysis process is also impacted by the number of bits used to develop the digital equivalent of the analog signals. The number of data bits used in this study is sixteen bits. Due to the nature and format of the data obtained from Veteran Administration Hospital in Washington, D.C., it was not feasible to study the effect of word length in this thesis. However, it has been shown[2,3,9] that eight bits will be practically sufficient for almost all ECG analysis application. A conversion accuracy table and the signal to noise ratio is shown in Table 2[2] for reference.

3.2 FILTERING AND SMOOTHING

The ECG signal from the electrode is passed through an analog filter and instrumentation amplifiers to remove as much of the noise component as possible before analog to digital conversion. The ECG signals are further conditioned by digital filtering before analysis.

TABLE 2

Data Bit Precision and Accuracy Table [2]

ACCURACY OF CONVERSION			SIGNAL-TO-NOISE RATIO
Number of Bits	Number of Levels of Quantization	% Accuracy	(Giving Equivalent Accuracy)
5	32	3 %	30 dB
6	64	1.5%	36 dB
7	128	0.8%	42 dB
8	256	0.4%	48 dB
9	512	0.2%	54 dB
10	1024	0.1%	60 dB

replaced by the weighted combination of the values of adjacent sampling points, thus reducing the effect of random variation at any single data point. This can be expressed mathematically as below:

$$Yo = \sum_{k=-N}^{N} W(k) * Y(k) \qquad \text{where}$$

Yo = the filtering sample point

N = number of adjacent sample points taken into consideration

W(k) = the weighting function for the k th point preceeding or following the filtering point

Y(k) = K th point preceeding or following the filtering point

Another technique, similar to the moving average is smoothing by parabolic curve fitting. In this approach, a parabola is fitted into a number of ECG sampling points by means of a least squares method. The expression for the parabolic curve is:

```
Y   = A + B*X + C*X**2
```

The filtering point is the center sample tc fit in this parabolic curve for X=0, thus resulting in the following equation[2]

$$Yo = A$$

$$= \frac{3}{4n(n^2-4)} \sum_{i=-m}^{m} (3n^2-7-20i^2)Y(i)$$

where $m=(n-1)/2$

The parabolic curve can be combined with the moving average, as follows:

$$Yo = A = \sum_{i=-m}^{m} W(i) * Y(i)$$

$$= \frac{3}{4n(n^2-4)} \sum_{i=-m}^{m} (3n^2-7-20i^2)Y(i)$$

Thus the weighting function can be re-written as below:

$$W(i) = \frac{3(3n^2-7-20i^2)}{4n(n^2-4)}$$

where n is the number of points to which the parabola is fitted and i is the i th weighing function.

Since n is programmable, the length of the interval to be smoothed is also programmable. If the smoothing interval is 1/60 second, then almost all of the sixty cps and its harmonics can be eliminated. Since the basic data rate was 500 samples per second, the smoothing interval should not be longer than 20 milli-second and the sampling frequency

should not below 250 samples per second, otherwise, a considerable change will occur in the resulting waveform.

Testing of these combined filter technique indicates that an additional criterion is needed to provide more accurate results. A threshold of 25 micro-volts is set for the difference between the old sample point value and the filtered(smoothed) point value. If the new value differs by more than 25 micro-volt from the old value, the old value will be kept; otherwise, the new (filtered) value will replace the old sample value. By doing this, the high frequency component in the ECG data will be retained while the random noise will be eliminated to a great extent.

3.3 SUMMARY

In this thesis, the maximum sampling rate for the ECG signal is 500 samples per second, from which the 250, 125 and 67 samples per second are derived by taking alternate samples from the previous sampling rate. The different sampling rates are tested by the author in the algorithm and significant changes in both the quality of the signal and the analysis result of the algorithm occur at sampling rates of 125 samples per second or less. For example, the amplitude of the QRS peak changes by approximately ten per

cent when the sampling rate drops below 125 samples per second. These results obtained by the author show close agreement with the work done by Berson[3]. Thus a sampling rate of 250 or 500 samples per second are used for the evaluation of the analysis algorithm.

Digital filtering of the input ECG signal is necessary to remove extrenous noise. A simple moving average technique and a parabolic smoothing technique have been tested by the author

and found to be inadequate for processing ECG data. Thus, a parabolic smoothing technique with a preset threshold, for attenuating the high frequency effect, is applied. This technique removes most of the random noise , particularly the 60 cycles component and its harmonics. This method is chosen in the algorithm for ECG analysis because of its simplicity and satisfactory results.

Chapter IV

QRS RECOGNITION

A number of different approaches have been used in recognizing QRS pattern and determining the onset and offset of the QRS waveform. However, there is no unique set of rules and designers have come up with different strategies. In the past, a lot of QRS recognition in the ECG waveform is based on special characteristics of the QRS wave. These include:

1. point of steepest positive slope
2. point of steepest negative slope
3. maxima
4. minima
5. corner
6. zero crossing
7. first slope difference
8. second slope difference
9. different order of derivatives

Figure 8[2] illustrates some of these features in the ECG waveform. These characteristics and other special points are all usable in ideal cases, but in practice, noise and sensitivity to main interferences and artifacts make it difficult to use them effectively. For example, a large T wave and a regular QRS can easily be confused if only the characteristics shown above are used.

As discussed in the literature review of this report, the IBM-Bonner[4,5] program uses a segment difference method to distinguish QRS from other waves. In this approach, several passes are required to eliminate false detection and to locate the onset and offset of the QRS wavefcrm. It also requires a special type of analog pre-processor to set flags for special points such as maxima and minima for the computer to use.

Several other approaches have been used successfully in QRS recognition. In this thesis, four QRS recognition techniques were tested and evaluated before finalyzing an approach. They are:

1. Spatial Magnitude

2. Spatial Velocity

3. Spatial GF transform

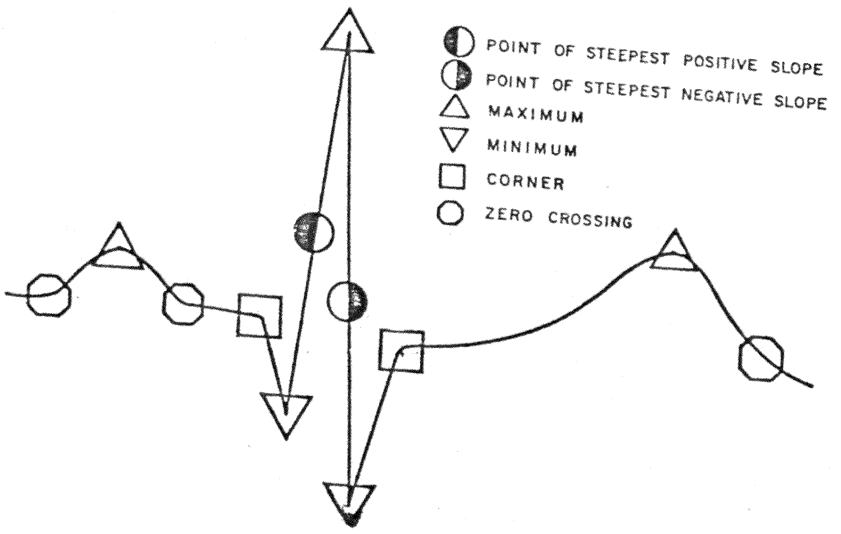

Figure 8: QRS characteristics and Features[2]

4. Spatial Operational Transform

Since each technique has distinct advantages and disadvantages, their attributes will be summarized in the following sections.

4.1 SPATIAL MAGNITUDE

The spatial magnitude method has only been used in the X,Y,Z Frank orthogonal lead technique and it is not valid for twelve lead data. In this method, the X,Y,Z data are used to calculate the Spatial Magnitude (SM) function which can be expressed mathematically as :

```
SM = SQRT (X**2 + Y**2 + Z**2)
```

The spatial magnitude (SM) function is the root mean square of the X,Y,Z signal. During ventricular depolarization, the spatial magnitude function takes on values far in excess of those occuring during other portion of the cardiac cycle. Thus a threshold can be set to determine the location of the QRS onset and offset. A schematic illustration of spatial magnitude function is shown in Figure 9[2]. Ideally, a threshold for P and T wave can also be set for recognition of them. However, the setting of the threshold is very difficult in most cases

because of the great variation in the function magnitude for different subjects. It is difficult to set a distinct threshold to determine the onset and offset cf any waveform in the ECG signal. The spatial magnitude function is an effective means to determine the general (rough) location of the QRS. The T and P waves can also be located by this method, but the result is usually not accurate enough to determine the exact location.

4.2 SPATIAL VELOCITY

Spatial velocity approach is similar to spatial magnitude method, except that it uses a different function for evaluation. This method is also used primarily for X,Y,Z Frank orthogonal leads. The spatial velocity (SV) can be expressed in the following form:

```
SV = F * SQRT (DELX**2+DELY**2+DELZ**2)
```

where

```
F    = sampling frequency

DELX= X(i+1)-X(i)

DELY= Y(i+1)-Y(i)

DELZ= Z(i+1)-Z(i)
```

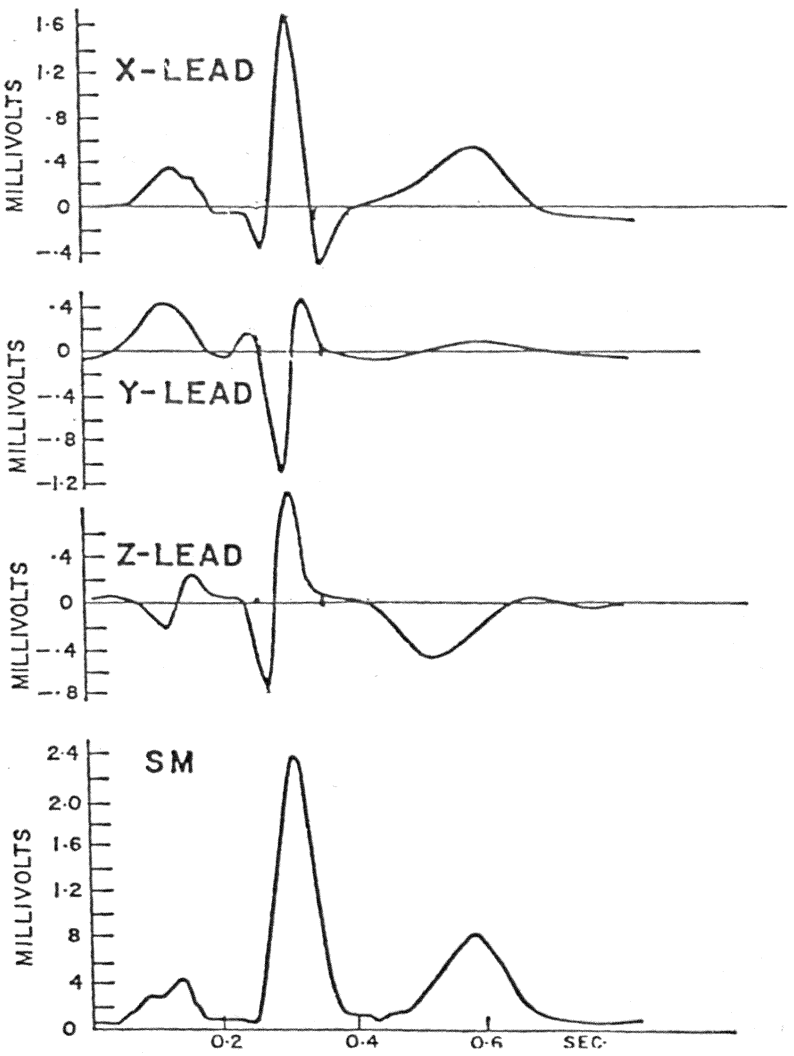

Figure 9: Spatial Magnitude Function[2]

The function SV is defined as the rate of spatial change per unit time. A schematic illustration of the spatial velocity function is shown in Figure 10[2].

Similar to the spatial magnitude approach, a threshold is set for each individual waveform (QRS, P, and T). Typical values are 2mv/sec for P wave, 7mv/sec for the QRS and 3mv/sec for the T wave. The threshold tends to be different for different subjects; thus causing difficulties in computer program implementation. Also, the onset and offset of the wave cannot be located accurately with either the spatial magnitude or velocity.

4.3 SPATIAL GF TRANSFORM

The GF transform[6] is based on the first difference principle and several mathematical expressions have been developed for it. The following GF transform is proposed by Ivaturi S. N. Murthy and Mandayam R. Rangaraj.[6] The GF transformation is accomplished in two parts, the first being the G transformation defined as follows:

$$G(n) = \sum_{i=1}^{N} (X(n-i+1)-x(n-i))**2*(n-i+1)$$

where n is the window within which N past differences are computed, squared and weighted by weighting factor (n-i+1).

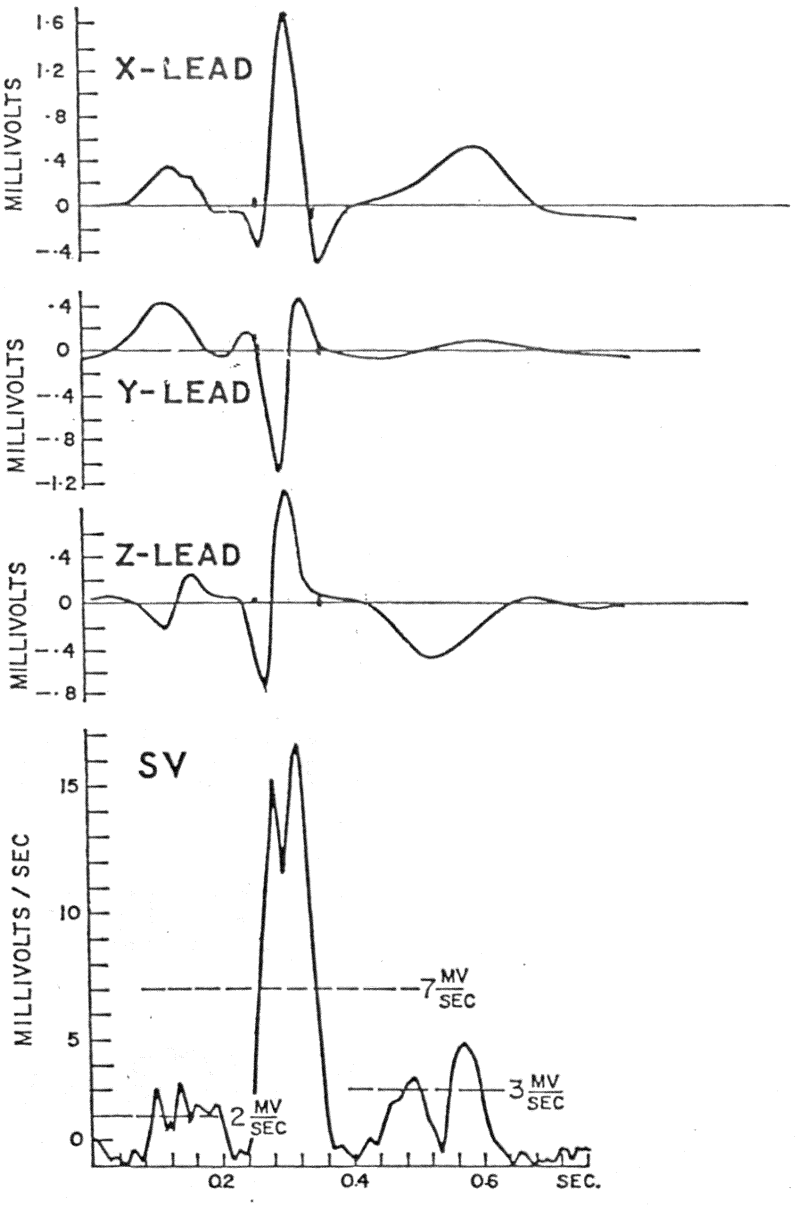

Figure 10: Spatial Velocity Function[2]

With the G transformation, a large twin peak will be obtained at QRS locations, but often there will be ripples due to artifacts in ECG and interferences. To eliminate the multiple peaks, another transformation is performed on the G function. This transform results in the GF function which has only a single peak occurring at the end of each QRS wave. This can be expressed mathematically as:

$$GF = \frac{1}{M} \sum_{j=0}^{M-1} G(n-j)$$

Substitution of the G transform results in the following expression

$$GF(n) = \frac{1}{M} \sum_{j=0}^{M-1} \sum_{i=+1}^{N} (X(n-i-j+1)-X(n-i-j))^2 (n-i+1)$$

In the above expression, M and N is chosen to be eight and the sampling period is ten milli-seconds. For a higher sampling rate, more ripple will appear in the transformation. Thus, a sampling period of approximately 6 milli-seconds to 12 milli-seconds is found tc be appropiate for this particular application. For m=n=8, this transformation provides a single ripple free positive peak for each cardiac cycle with the maximum value cccuring right at the end of each QRS complex. This method is capable of locating the end of QRS accurately and is not sensitive to artifacts or main interferences. This method can also

distinguish between tall T wave versus pre-mature ventricular beat(PVC) and regular QRS'. This method can be applied on any single lead regardless of the lead system used and requires no adjustments to pre-set thresholds for different subjects. This simplifies programming and will operate at lower sampling rates . Thus less computation time is required.

The GF transforms obtained from the X, Y, and Z leads are combined and normalized to locate a single location as the ending of the QRS wave for all three leads. The three GF transform functions are combined to give the spatial GF transform (SGF) in the following manner: (Figure 11 shows the SGF transform function)

$$SGF(i) = SQRT((n_1 GF_x(i))^2+(n_2 GF_y(i))^2+(n_3 GF_z(i))^2)$$

where GFx max * n1

=GFy max * n2

=GFz max * n3

=max (GFx,GFy,GFz)

In this function, n1,n2 and n3 provide weighting factors to normalize the three GF transforms (GFx, GFy, GFz). The

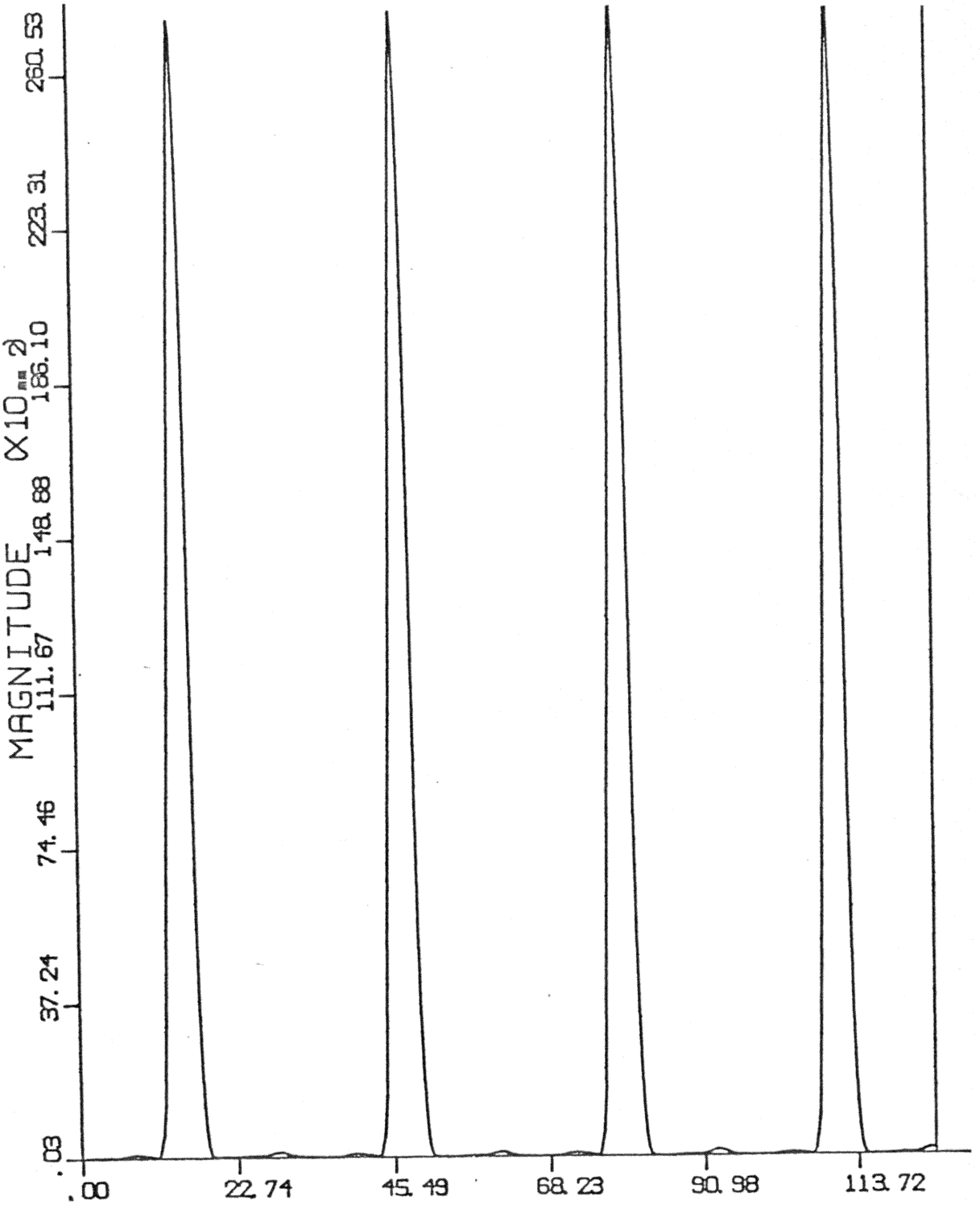

Figure 11: Spatial GF Transform

resultant SGF function will produce a single positive peak for each cardiac cycle with maximum value occurring at the end (offset) of the QRS wave for all X, Y, and Z leads.

4.4 SPATIAL OPERATIONAL TRANSFORM

This operation transform is similar to the G transform except that it is intended to find the onset of the QRS wave It was orginally proposed by Richard Kitney, Chris Turner and Alistair McDonald in their paper, "Assessment of QRS shape and measurement of interval as a basis of ECG Rhythm Analysis"[7]. In their paper the operational transform is expressed as:

```
OW(n) = (ECG(n)-ECG(n-2))**2
```

This transform will result in a monophasic waveform with a twin peak occurring during each QRS cardiac cycle. The transform can be performed on any single lead and is insensitive to interferences or artifacts in the ECG. The OW(n) function produces two relatively large values during the QRS period. This approach is ordinarily used to locate the occurence of the QRS wave, it has been modified to locate the the onset of the QRS wave by noting the last minima of OW(n) prior to its onset of the twin peaks. Usually, this method accurately detects the onset of the

QRS. Similar to the GF transform, the operation transforms for X, Y, and Z leads can be combined by means of a spatial relationship with normalization of the three leads to result in a single QRS onset point. This results in the Spatial Operational Transform (SPOW) which can be expressed as:

$$SPOW(n) = SQRT((n_1 OW_x(n))^2+(n_2 OW_y(n))^2+(n_3 OW_z(n))^2)$$

where OWx max *n1

=OWy max * n2

=OWz max * n3

=max (OWx,OWy,OWz)

Figure 12 shows the spatial operaticnal transform function.

Thus the SPOW function provides a systematic way of locating the onset of QRS for all three leads. This flexible transform is accurate at low sampling rates and reduces the computation time.

The onset and offset of the QRS wave can be accurately determined by the spatial operational transform (SPOW) and the spatial GF transform (SGF). In additicn, these two

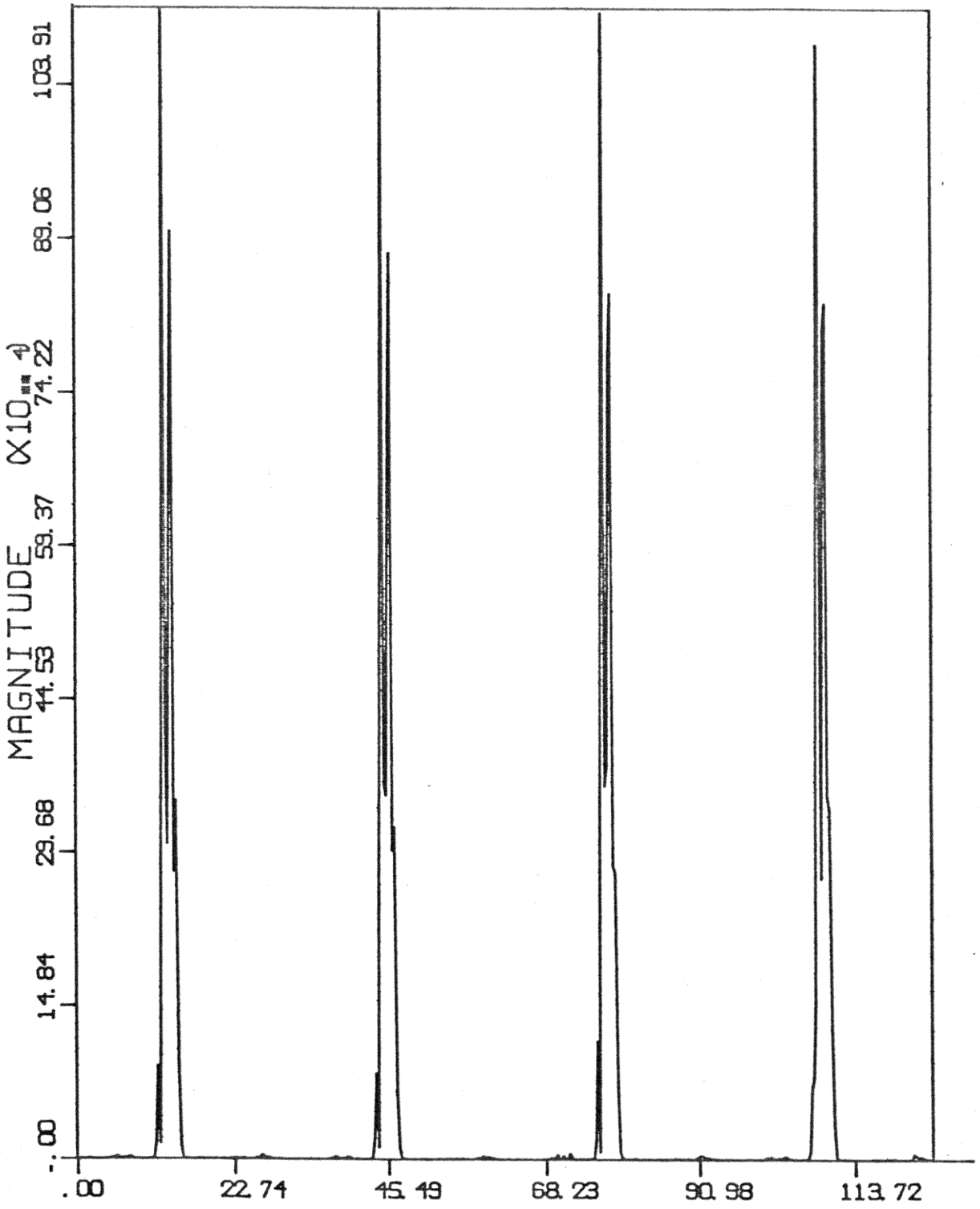

Figure 12: Spatial OWn Transform

functions are relatively insensitive to artifacts and noises. They are also effective in identifying the QRS wave when abnormally large T wave and small QRS wave occur.

4.5 RECOGNITION OF Q, R, AND S WAVES IN QRS

Once the onset and offset of the QRS wave are located, it is much easier to locate the corresponding Q, R, and S waves and assign morphological properties. A program can be developed to recognize Q, R, and S waves from their definitions. The first negative deflection within QRS is called the Q wave; the first positive deflection is called an R wave; and any negative deflection following the R wave is called the S wave. However, there are certain degrees of variation in the appearance of the Q, R and S waves. Most variations which occur in the QRS wave are shown in Figure 3[2]. If a twin peak occurs at the R location, the one with higher magnitude is called R wave and the other is called R' wave. Also, a lower case letter, r, can be assigned for a secondary R wave with small amplitude. Similar assignments hold true for the Q and S waves.

After the onset, offset, Q, R, and S wave are located, a program can be written to evaluate the attributes of the QRS wave. Typical quantities of interest are the spatial vector, elevation, azimuth, frontal plane vector, frontal

plane vector angle, horizontal plane vector, horizontal plane angle, vertical plane vector, vertical plane angle, the initial and terminal QRS vector and their corresponding angles, the area for QRS wave, the half area location for QRS, the mean and the maximum spatial vector, the axis of the heart and the various different segment and time intervals.

4.6 SUMMARY

The different methods (spatial magnitude, spatial velocity, spatial GF transform, spatial Own transform), as discussed in this chapter are tested individually for processing the ECG signal. The spatial magnitude and spatial velocity methods cannot provide an accurate identification of the onset and offset of the QRS waveform. These methods also involve too many other constraints and require changes for different cases of ECG signals from a variety of subjects.

The GF transform, recently proposed by Murthy and Rangaraj[6], is tested and examined to determine its feasibility, limitations and constraints. Also, the GF transform which is originally proposed for single ECG lead application, is modified in this thesis to accept the three lead data (the Frank Orthogonal Lead System) and provide a

single QRS offset point without affecting the accuracy of the original transform. In testing the GF transform, one important constraints is found: the sampling interval of the data for this transform should not be less than four milli-seconds, particularly in the case of noisy ECG signal. When the sampling interval is less than four milli-seconds, the ripple which appears in the G transform will also appear, at least in part, in the GF transform. Thus the position for the QRS offset cannot be uniquely identified and the accuracy of the GF transform is also constrained to the interval between consecutive data points. Therefore, a six milli-seconds interval is used for implementing the transform. Apart from the problems discussed above, the transform provides good results and is useful for EKG analysis.

The operational transform, originally proposed by Kitney, Turner and McDonald [7], is used for locating occurence of QRS on single lead ECG signal. In this thesis the transform is tested and modified to accept the three lead system and to identify accurately the QRS onset point for the three lead system. The Operational transform has constraints similar to the GF transform and the interval between consecutive data pcints should not be too small(less than 4 milli-seconds). In the modification done in this thesis,

the spatial operational transform is formed and the last minima prior to the onset of the large ripples is designated as the onset of the QRS wave.

Finally, the modified GF transform (spatial GF transform) and the modified operational transform(spatial operational transform) are implemented in the algorithm and tested. They are able to identify consistently and correctly the onset and offset of QRS for the three lead system (Frank Orthogonal Lead System). Figure 13 shows a composite diagram of the three lead signal corresponding to the spatial operational transform and the spatial GF transform as obtained in the developed alogorithm in this thesis.

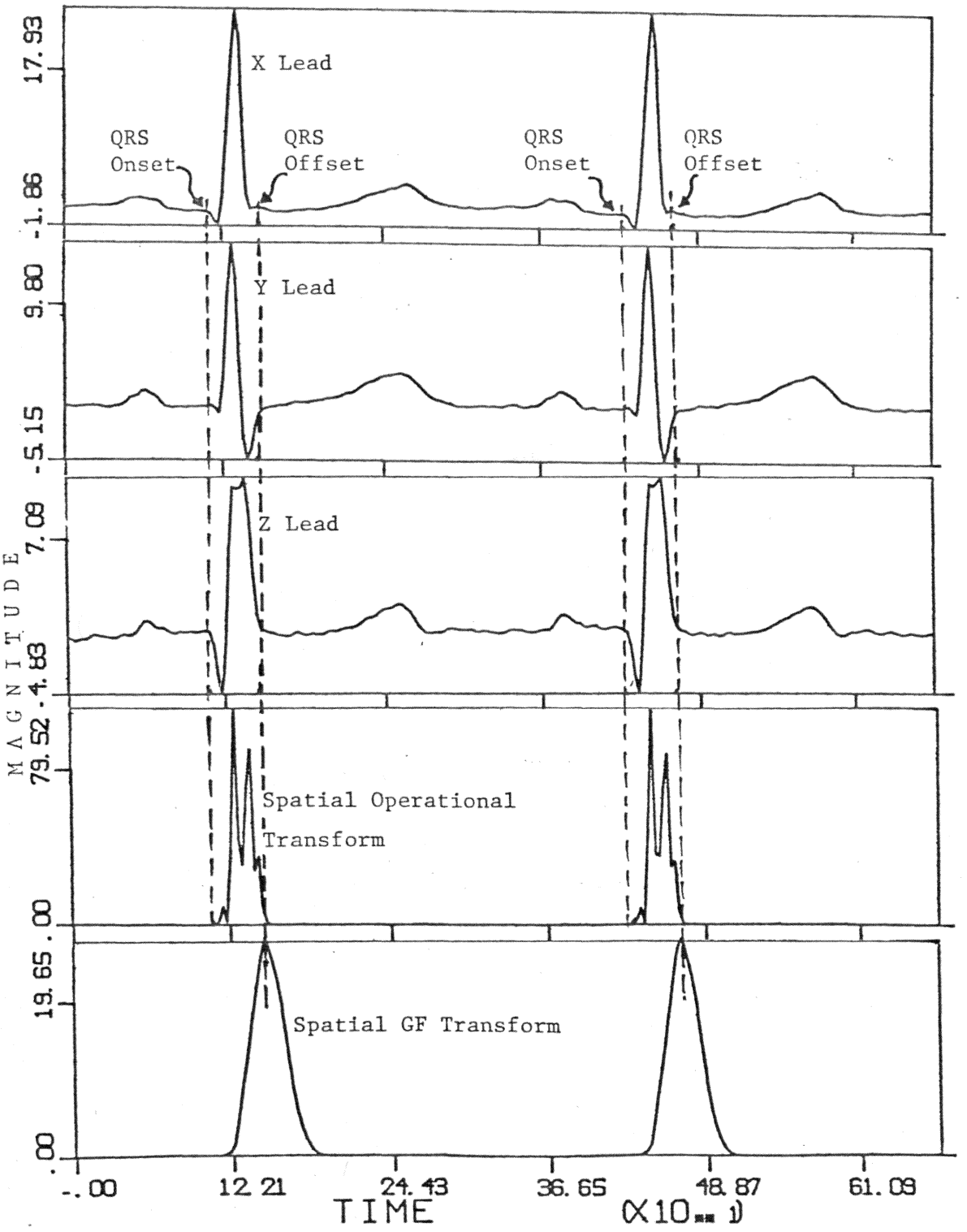

Figure 13: Timing Diagram for SPGF and SPOW transforms

Chapter V

RECOGNITION OF P AND T WAVES

Once the QRS wave has been located, the P and T wave can be located and analyzed. Under normal conditions, the magnitude of the T wave is smaller than that of the QRS, and the P wave is smaller than the T wave. It is much harder to locate the onset and offset of the P and T waves because in many instances, the signal to noise ratio is quite small. Also, if the ST segment is elevated, the onset of the T wave is hidden inside the ST segment. Thus it is almost impossible to determine the onset for the T-P segment. Many researchers use this segment as the baseline or the iso-electric point, but it is contaminated by the presence of the U wave. The drifting of the baseline of the ECG signal also increases the difficulty of locating the onset and offset for the P and T waves. It should be noted that for QRS recognition, baseline drift does not create a significant problem since the mathematical expression is based on slope differences rather than an absolute value. Therefore, before attempting P and T wave recognition, it is appropriate to discuss normalization for baseline drift.

5.1 NORMALIZATION

In the process of normalization, the points before the beginning of each QRS are used as a reference for the baseline or iso-electric point. A linear interpolation technique is applied to the ECG signal. The differences between the five points before the first QRS begins and the first five points before the following QRS begins are calculated. Five points are chosen instead of a single point in order to reduce the variation caused by random noise. The signal from the beginning of the P wave to the beginning of the next QRS are normalized under a straight line interpolation method (i.e. fitting the baseline drifting into a straight line and normalize all the data in between the two points by assuming a straight line drift of the baseline). This straight line interpolation is done repeatively for every cardiac cycle. Figure 14[2] illustrates how this straight line interpolation method applies. After the values of the ECG signals are normalized, they can be used for recognition of the P and T waves.

5.2 T WAVE RECOGNITION

Before discussing T wave recognition, it should be noted that similar procedures can be used for both the P and T

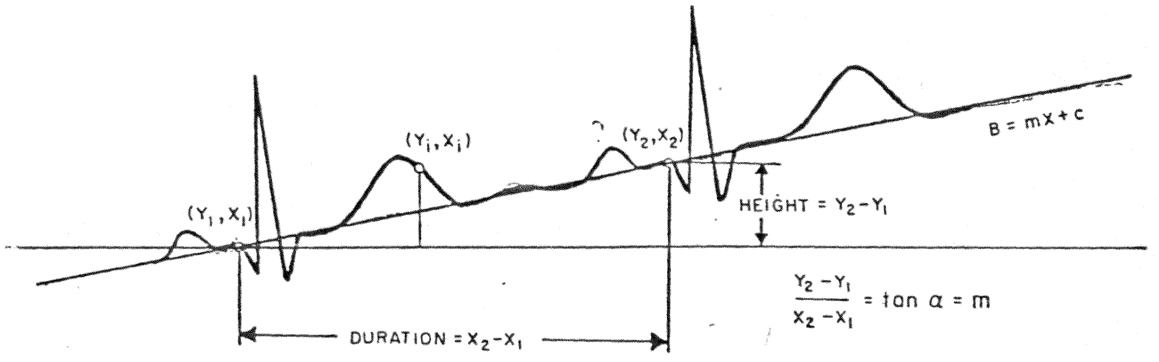

Figure 14: Illustration of Baseline Normalization[2]

wave. The differences between T and P wave are that T wave has a larger amplitude and a higher frequency component than P wave and the T wave occurs right after QRS while P wave occurs before QRS or after T wave. In the T wave pattern recognition, the peak, onset and offset must be identified. A search region and a set of criteria are established for the identification of each case. The search region for locating the peak(s) of T wave starts at the end of QRS plus 40 milli-seconds and reaches to the end of QRS plus 150 milliseconds. This search region has a total interval of 110 milli-seconds and is chosen such that the T peak can be located within this period for all cases. Several passes are required to establish the location of a T wave peak. The criteria are:

1. Absolute magnitude of the sample point must be greater than 50 micro-volts.

2. The sample point must be either a maxima or minima.

3. The 5 points before and after (a total of ten points) the sample point must have the follow relationship. The former five points must have a positive slope and the latter five points must all have negative slopes or vice versa.

4. All T wave peaks that satisfy the above criteria must exceed a threshold which equals half the magnitude of the maximum T peak found above.

After the T peak(s) is (are) located, the search for onset and offset can be initiated. Sometimes, the onset of T wave is not distinguishable if the ST segment is rising gradually. In this case, it is difficult to identify where the ST segment ends and where the T begins. In such case, the onset of T will assume to be undefined and a different set of criteria is required to locate the onset of T wave. When all these criteria are met then that particular sample point will be considered as T onset. The search region for T onset lies between the end of the corresponding QRS plus 20 milli-seconds to the corresponding T peak minus 20 milli-seconds. The criteria for locating T onset are:

1. The absolute differences between the sample point and 3 consecutive points before and after must not be greater than 10 micro-volts.

2. The absolute value for the consecutive sample points must be less than 20 micro-volts.

For locating the end of T wave, the search region lies between the corresponding T peak plus 20 milli-seconds to

the T peak plus 140 milli-seconds. Thus, a total of 110 milli-seconds is used to search for the end of the T wave. The criteria for locating end of T wave is:

1. A total of 10 points (5 points before and 5 points after the sample point) must have a slope and slope difference(with respect to the central sample point) which are less than 10 micro-volts.

Once the location for peak(s), onset and offset (end) of the T wave are identified, they can be used to determine other important functions, plots and parameters.

5.3 P WAVE RECOGNITION

In recognizing the P peak, the approach used is same as the recognition of the T peak. The only difference is the search region. The search region for P peak is between the offset of previous T wave plus 20 milli-seconds to the onset of the corresponding QRS minus 20 milli-seconds. The criteria are the same as shown in the T peak recognition. They are:

1. Sample point greater than 50 micro-volts
2. The sample point must be either a maxima or a minima.

3. Five points before and five points after the sample point must have a positive going slope for the former and negative going slope for the latter or vice-versa.

4. The value must exceed a threshold of at least half the magnitude of the maximum P peak found above.

Locating the onset and offset of P wave is also similar to locating the onset and offset of T wave. The search region for the P onset is defined between P peak minus 100 milli-seconds to P peak minus 20 milli-seconds. The P offset search region is defined between P peak plus 20 milli-seconds to QRS onset minus 30 milli-seconds. The criteria for P onset and offset are the same :

1. A total of ten points (5 points before and 5 points after the sample point) must all have a slope and slope difference (with respect to the central sample point) less than 10 micro-volts.

Once the locations for the onset, peak and offset of T and P waves are found, they can be used to find the following parameters:

1. Initial spatial vectors

2. Initial Azmuith

3. Initial spatial elevation

4. Frontal plane vectors

5. Horizontal plane vectors

6. Sagittal plane vectors

7. Spatial angles between vectors

8. Angles between vectors on the three (frontal, sagittal, and horizontal) planes. Half area vector and angle

9. Maximum area vectors and angles.

10. Intervals for each waves and between waves.

11. Spatial velocity for all vectors.

Chapter VI

GENERATION OF CLINICAL PARAMETERS

After the QRS, P and T waves are identified, these data can be used to generate the clinically important parameters and functions. This chapter will summarize and describe the more important parameters and functions. They are as follows:

1. QRS interval
2. Q interval and magnitude
3. R interval and magnitude
4. S interval and magnitude
5. Overall QRS interval
6. P interval and magnitude
7. T interval and magnitude
8. PR or PQ segment interval and average magnitude
9. ST segment interval and average magnitude
10. T-P segment interval and average magnitude
11. J point

12. PQ or PR interval

13. Q-T interval

14. P-P interval

15. R-R interval

16. Initial, maximal and terminal QRS spatial vectors and spatial angles

17. Initial, maximal and terminal P spatial vectors and spatial angles

18. Initial, maximal and terminal T spatial vectors and spatial angles

19. Frontal, Horizontal and Sagittal plane QRS vectors and angles (initial, maximal and terminal)

20. Frontal, Horizontal and Sagittal plane of P vectors and angles (initial, maximal and terminal)

21. Frontal, Horizontal and Sagittal plane of T vectors and angles (initial, maximal and terminal)

22. Mean vectors for all waves in frontal, horizontal and sagittal planes

23. Heart axis

24. Plots for all vectors on frontal, horizontal and sagittal planes

25. Overall QRS, P-R, Q-T, T and P interval

26. Overall heart rate and rhythm

Parameters one to fifteen can be found directly from the time domain ECG signal, once the critical patterns and locations are identified. They are derived mainly from the intervals and magnitudes for different waves and segments and provide clinically important information for cardiologists. Parameters sixteen through twenty-three are spatial and vector representation of the ECG signal and require special calculations or data manipulation. These parameters will be discussed in the following sections.

6.1 SPATIAL REPRESENTATION OF HEART VECTOR

The cardiac dipole, generated by the working heart, has a certain direction from - to +, and a certain magnitude (pole strength) and thus can be represented as a vector, called the heart vector. The QRS, P and T wave can be represented in terms of spatial vectors, with spatial magnitude and spatial angles (azimuth and elvation) in a three-dimensional space. They can also be represented in three two-dimensional planes.

The cardiac dipole or heart vector is represented as a sum of three vectors directed along the Cartesian axes provided by the corrected orthogonal lead system. The spatial vector can be expressed as:

$$ECG\overrightarrow{xyz} = Ex\ \vec{i} + Ey\ \vec{j} + Ez\ \vec{k}$$

where ECGxyz is the spatial vector and Ex, Ey and Ez are the scalar components along the X, Y and Z axes respectively.

A minimum of three parameters are required to specify a unique spatial vector in the vectorcardiogram. They are:

1. Spatial Magnitude (M)
2. Azimuth (H) (the angle between the projection of the heart vector onto the horizontal and the left half of X-axis)
3. Elevation (V) (the angle between the caridac vector and the horizontal plane)

The relationships between the scalar Ex, Ey and Ez and the magnitude, azimuth and elvation are shown below:

```
M = SQRT (Ex**2 + Ey**2 + Ez**2)
```

Ex = M * cos(v) * cos (H)

Ey = M * sin(V)

Ez = M * cos(V) * sin(H)

With the X, Y and Z scalar component given, the spatial vectors, azimuth and elevation can be found from the following relationships:

Spatial Magnitude = M

= SQRT(Ex**2+Ey**2+Ez**2)

Azimuth = H

$$= \text{arc cos} \left(\frac{E_x}{\sqrt{E_x^2 + E_y^2}} \right)$$

$$= \text{arc tan} \left(\frac{E_z}{E_x} \right)$$

Elevation = V

$$= \text{arc cos} \left(\frac{\sqrt{E_x^2 + E_z^2}}{\sqrt{E_x^2 + E_y^2 + E_z^2}} \right)$$

$$= \text{arc tan} \left(\frac{E_y}{\sqrt{E_x^2 + E_z^2}} \right)$$

Figure 15[1] illustrates the spatial vector with its azimuth and elevation.

6.2 PLANE PROJECTION OF HEART VECTOR

The heart spatial vector can be represented in terms of its projection upon a plane. Such a projection is obtained by vector addition of two mutually perpendicular vectors directed along the Cartesian axes. In studying of vectorcardiogram, the spatial vectors are usually represented in three planes, the frontal, horizontal and sagittal planes. (Figure 16[1] shows schematically the vector representation of ECG on the frontal plane.) The three plane projections can be obtained by the following relationships:

1. Frontal Plane

$$Exy = SQRT\ (Ex**2 + Ey**2)$$

$$F = \arctan\left(\frac{E_y}{E_x}\right)$$

2. Sagittal Plane

$$Eyz = SQRT\ (Ey**2 + Ez**2)$$

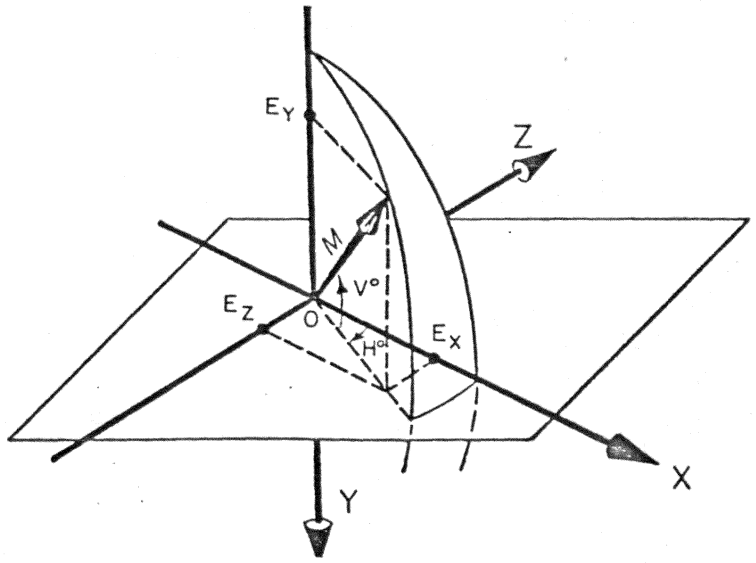

Figure 15: Spatial Vector Co-ordinates[1]

$$S = \arctan\left(\frac{E_y}{E_z}\right)$$

3. Horizontal Plane

$$Exz = SQRT\ (Ex**2 + EZ**2)$$

$$H = \arctan\left(\frac{E_z}{E_x}\right)$$

6.3 MEAN HEART VECTOR

Mean heart vectors are also important in clinical analysis of electrocardiograms and vectorcardiograms. The P, QRS and T mean vector are of particular importance and are defined as follows:

1. The mean P vector ($\vec{A}_P$) results from instantaneous vectors of auricular depolarization.

2. The mean QRS vector ($\vec{A}_{QRS}$) results from instantaneous vectors of ventricular depolarization.

3. The mean T vector ($\vec{A}_T$) results from the instantaneous vectors of ventricular repolarization.

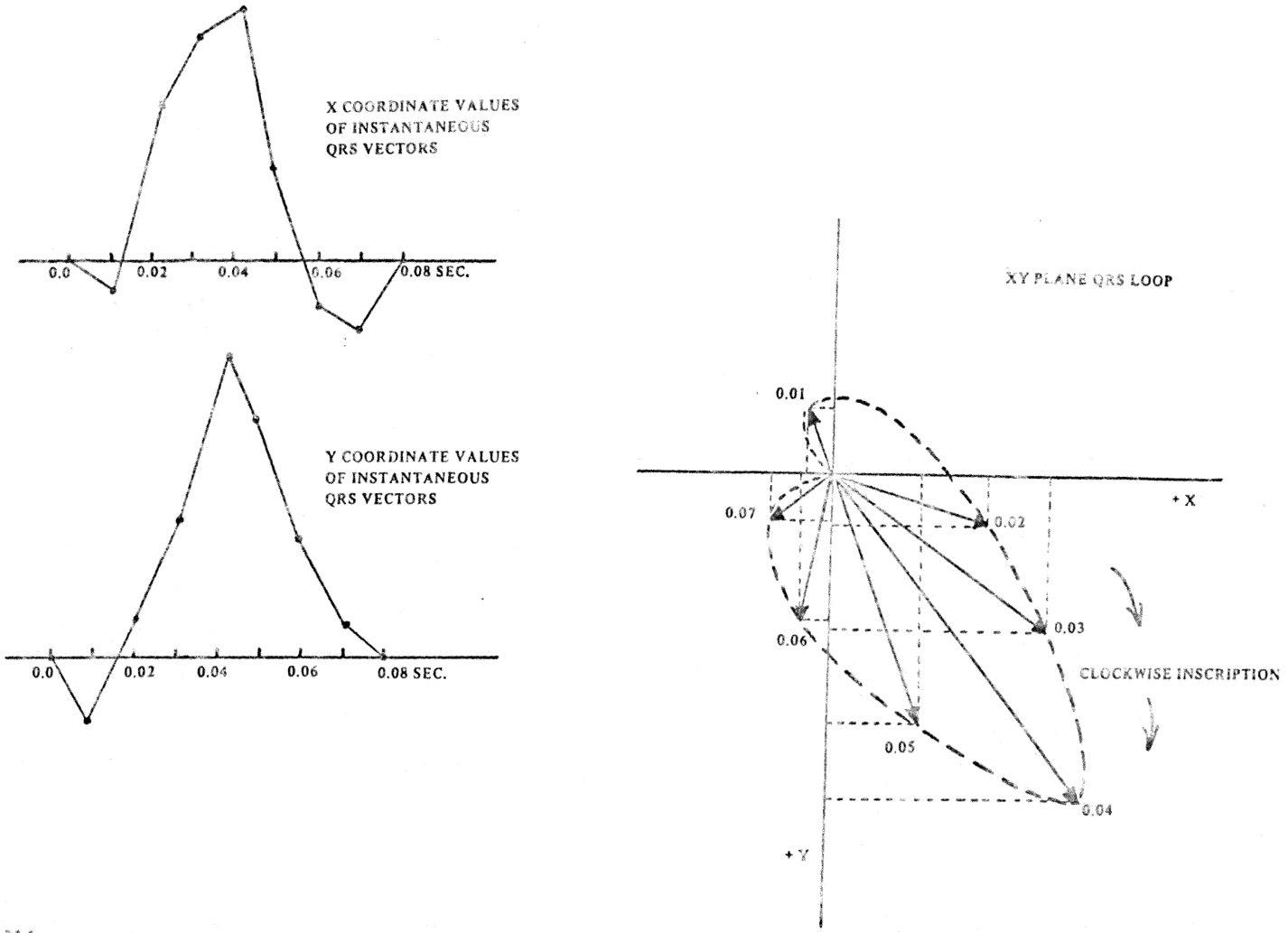

Figure 16: Frontal Plane Projection of ECG Vectors [1]

The mean vector for any single axis can be described mathematically. For example, the mean QRS vector on X-axis can be expressed as:

$$A_x^{QRS} = \int_Q^S Ex\ (t)\,dt$$

$$\approx \frac{1}{n} \sum_{i=1}^{n} \left(E_x^{QRS} \right)_i$$

The mean QRS vector on the frontal plane is:

$$\left| A_{xy}^{QRS} \right| = SQRT\ \left(\left| A_x^{QRS} \right| **2 + \left| A_y^{QRS} \right| **2 \right)$$

and the angle for the mean QRS vector on the frontal plane is:

$$F = \arctan \left(\frac{\left| A_y^{QRS} \right|}{\left| A_x^{QRS} \right|} \right)$$

The mean spatial QRS vector on a three dimensional space is as follows:

$$\left| A_{xyz}^{QRS} \right| = SQRT\ \left(\left| A_x^{QRS} \right| **2 + \left| A_y^{QRS} \right| **2 + \left| A_z^{QRS} \right| **2 \right)$$

and the angle azimuth can be calculated as follows:

$$H = \arctan\left(\frac{\left|A_z^{QRS}\right|}{\left|A_x^{QRS}\right|}\right)$$

$$\text{and} \quad V = \arctan\left(\frac{\left|A_y^{QRS}\right|}{\sqrt{\left|A_x^{QRS}\right|^2 + \left|A_z^{QRS}\right|^2}}\right)$$

(as shown in spatial vector in section 6.1)

Thus the mean vectors for all waves can be found in any plane and also in three-dimensional space.

There are three other mean vectors which are of significant value in analyzing ECG. They are:

1. The initial mean QRS vector (i.e. septal depolarization) is generated within the first ten to fifteen milli-seconds of ventricular depolarization.

2. The ventricles' free wall depolarization vector is the mean direction and magnitude of heart vectors generated between the end of septal depolarization and the onset of the ventricles' base depolarization.

3. The mean terminal QRS vector (ventricles' base depolarization) is generated within the last ten or fifteen milli-seconds of the QRS vector.

All these mean vectors provide important information for detail cardiac studies, and can be found with this algorithm.

Chapter VII

LITERATURE REVIEW

7.1 INTRODUCTION

This chapter presents a summary of a review of the available EGG analysis programs. The efforts of other researchers are summarized and discussed. The review can be partitioned into four parts:

1. A brief study on sampling, bandwidth and data precision;

2. A summary of methodology in IBM-Bonner, Argus and FSM programs;

3. An evaluation of ECG programs for repeatibility and comparisons of measurements from different programs;

4. Conclusions and suggestions for future programs.

7.2 SAMPLING, BANDWIDTH AND DATA PRECISION

In any automatic ECG system, the first step is to obtain data and convert it from analog to digital format. Then the data is processed by the computer for feature extractions, pattern recognition and diagnostic processes. In most systems, a high sampling rate and a large number of data bits are used to assure good quality data. However, in many

instances, excessive bandwidth, sampling rate and bit precision result in considerable cost for memory and computation time. Studies by Ruttiman[9] and Alan S. Berson address this problem[3]. In a study by Berson, 433 adult male ECG records were analyzed using different combinations of analog and digital bandwidth, sampling rate and data bits. The analog bandwidths used are 50, 100 and 200 hertz, while the digital bandwidths are 40 and 50 hertz . The sampling rates used are 500, 250, 167 and 100 hertz. Duration for P, QRS and T, and the amplitude for Qz, Rx, Rz and QRSmax are measured and compared with values obtained with a 200 hertz bandwidth analog filter and 500 hertz sampling rate. The percentage of records with differences outside 8 milliseconds for duration and 0.1 millivolt for amplitudes are recorded for both single (individual) beat and best(average) beat measurements. The results indicate that both amplitude and duration are seriously affected when the sampling rate drops below 167 Hertz regardless of the bandwidth used.

Also, measurements on the 'average' beat are more accurate and reliable than those on the single beat. There is a problem with this approach and it will be discussed in the later part of this chapter. For a single beat or an average beat, there appears to be little difference in the

results for 100 Hertz bandwidth with 250 Hertz sampling rate and 200 Hertz bandwidth with a 250 Hertz sampling rate. It is obvious from the results that the bandwidth cannot be reduced to 50 Hertz or below (e.g., for the purpose of eliminating power-line noises) without significant impact on accuracy.

In Berson's paper, the effect of bit precision on the measurements is investigated. The error is minor for eight or more bits. The error becomes significant when the number of data bits is six or below. It appears that the duration, particularly for the P wave, is more sensitive to data bit precision than amplitudes. In specific wave recognition algorithms, the precision does not cause much error if the program can be tailored to meet with the particular precision. In general, a system with 100 hertz bandwidth, 200-250 hertz sampling rate and 8-10 bits of precision can provide accurate and reliable results.

In almost all ECG analysis systems, uniform sampling has been used. sampling interval. It may be possible to use variable interval sampling to improve accuracy and filtering and decrease storage and computer time requirements. Variable lenth encoding schemes such as Huffman coding can also be used. Other sampling or coding techniques are

worth investigating for improving efficiency and accuracy of ECG analysis systems.

7.3 METHODOLOGY

Since there is no standard set of rules to establish the criteria for analysis programs, a number of pattern recognition techniques have been used for ECG analysis. The criteria and strategies used by designers are often quite differ from one another. For example, the IBM-Bonner, Argus and FSM (by Florenz,Rolnitz and Bigger) programs employ different approaches and will be reviewed and discussed.

7.3.1 IBM-Bonner Program

In Bonner's program[4,5], an analog pre-processor is required to set flags on data of particular interest(e.g. maxima, minima). About ten percent of the sampled points are flagged. The points are filtered digitally with twenty points before and twenty points after the point being filtered. The filter is also designed for QRS waveform by calculating the difference between the filtered and old values. If the difference exceeds the pre-set threshold (0.025 millivolt), the old value is retained, otherwise the filtered value is used to replace the old value. The Bonner program uses a maximum segment slope difference method for

locating QRS search. The program utilizes the fact that the slope in QRS is greater than slope of any other waves, then finds a pre-assumed number of maximum 'segment slope difference'. The program finds the location of QRS and then by a 'segment adding' method to locate the onset and offset of the QRS. Segments before and after the identified QRS are added after passing a number of criteria (e.g. no QRS of duration more than 220 milli-second can have a segment added.) The slope of the the segment to be added is one of the criteria used in adding segments to the QRS progressively. Each QRS found in the first pass is compared to others; if significant differences exist, they are flagged. (e.g. any single QRS with width wider than 1.3 times of average QRS width is flagged.) Also a point rating method is used to assign to each of the similar or un-similar QRS. The type with the highest rating is considered as the dominant rhythm.

For the recognition of P wave and T wave, Bonner also used a point rating method to assign points to a particular wave under a series of criteria tests. Thus , by means of the rating result, the particular wave under test is identifies as to corresponding waveform. In the diagnosis portion of Bonner's program, the method used is based on 'cardiologist diagnosis type' approach: a set of rules are

set in the program for each case of disease basing on the parameters obtained from the pattern recognition program.

Since its introduction, a number of improvements have been made in the IBM-Bonnor program. However, the repeatibility has not reached a level needed to gain the wide acceptance of the medical world. Apparantly, the medical professionals are not yet ready to rely completely on this new technology.

7.3.2 Argus Program

ARGUS(Automatic Routine for Generating and Updating System)[8,11,12,19,20] was developed in the Biomedical Computer Laboratory of Washington in St. Louis, Missouri. Argus consists of three processors: (1) Aztec Processor, (2) Primitive Processor, and (3) Cycle Processor. Each processor is a complex system in itself and they will be briefly introduced here for some basic understanding of the work involved. Figure 17[20] shows a block diagram of the ARGUS.

Before the ECG signal is fed into the Aztec processor, it is treated by an analog filter and then a digital pre-processor. The digital pre-processor and the Aztec processor provide successive data reduction to the input

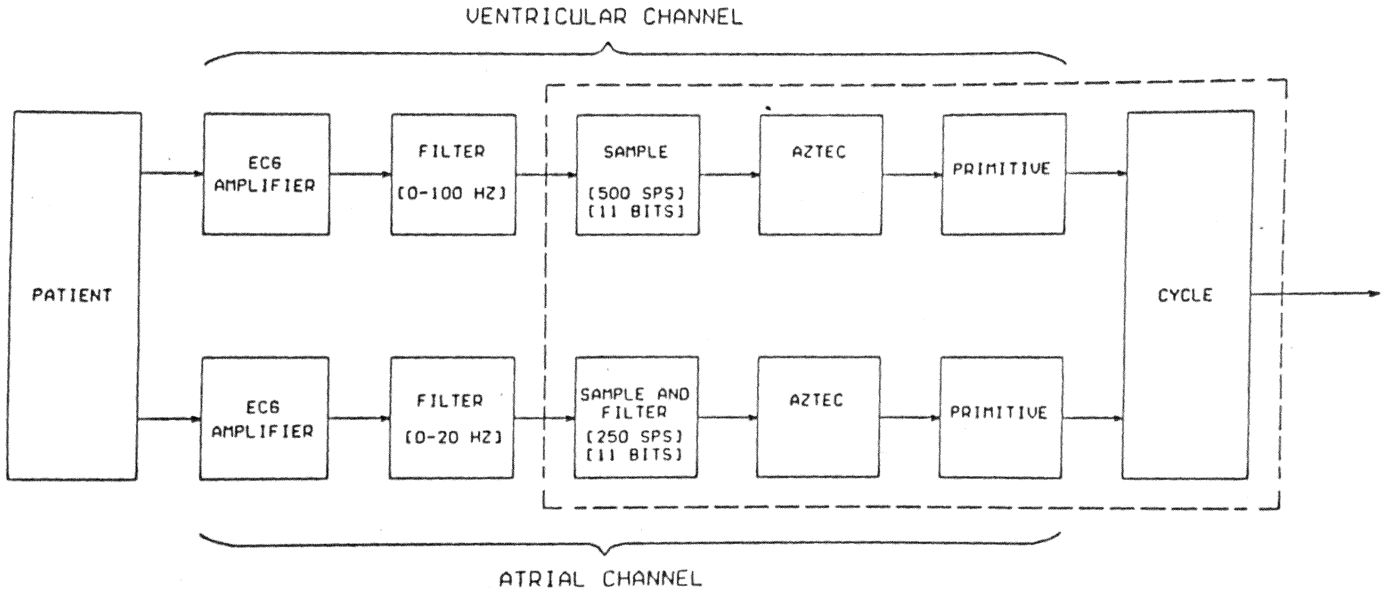

Figure 17: Block Diagram of ARGUS [20]

signal. Aztec then converts the output of the digital pre-processor to a set of symbols with indicators for slope, line, direction and magnitude; then actions are taken for accumulating the line time or the slope time. Then this set of symbols are fed into the Primitive processor. The Primitive processor is a quite complicated processor which can be subdivided into three sub-processors.

1. Grade sub-processor : Identifying flat sections (Gf) of the record and to catagorize slope into three classes(Gi, Gm, Gl).

2. Wiggle sub-processor : Identifying and grouping together symbols that represent vacillating portions of the record.

3. Ventricular sub-processor : Combining results in Grade and Wiggle sub-processors to identify QRS onset and offset.

The Cycle processor uses the information provided by the two Primitive processors(one for QRS and one for P and T) and reconstructs the signal and identifies location for all onset, offset and rhythm for QRS, P and T waves.

In the PVC program for the ARGUS, four basic morphologic parameters (area, height, duration and offset) are used for

cataloging QRS's. These four parameters are illustrated in Figure 18[19].

In ARGUS' PVC program, the four basic parameters are used to assigned a given QRS into one of the sixteen QRS families. Age, feature boundaries and pathology are some revelant parameters assigned as a new QRS member enters into a family. The program initializes the first normal family, on the basis of duration and population of that particular family under consideration. Then new members are added to the family if the four basic morphologies are included within the feature boundaries of the existing families, otherwise a new family is formed.

In the first pass of the program, QRS'S are classified into one of the three classes: normal, borderline and abnormal QRS. In the second pass, the borderline and abnormal QRS'S will be candidates for a pre-mature ventricular contraction beats. In a recent version of the program, an additional morphologic parameter is added to improve identification between QRS wave and large T waves. This parameter is apparant triangular area(ATA), which equals to duration times height divided by two(ATA = Duration X Height/2). The ATA is compared with the original morphological(actual) QRS area, which is the absolute value

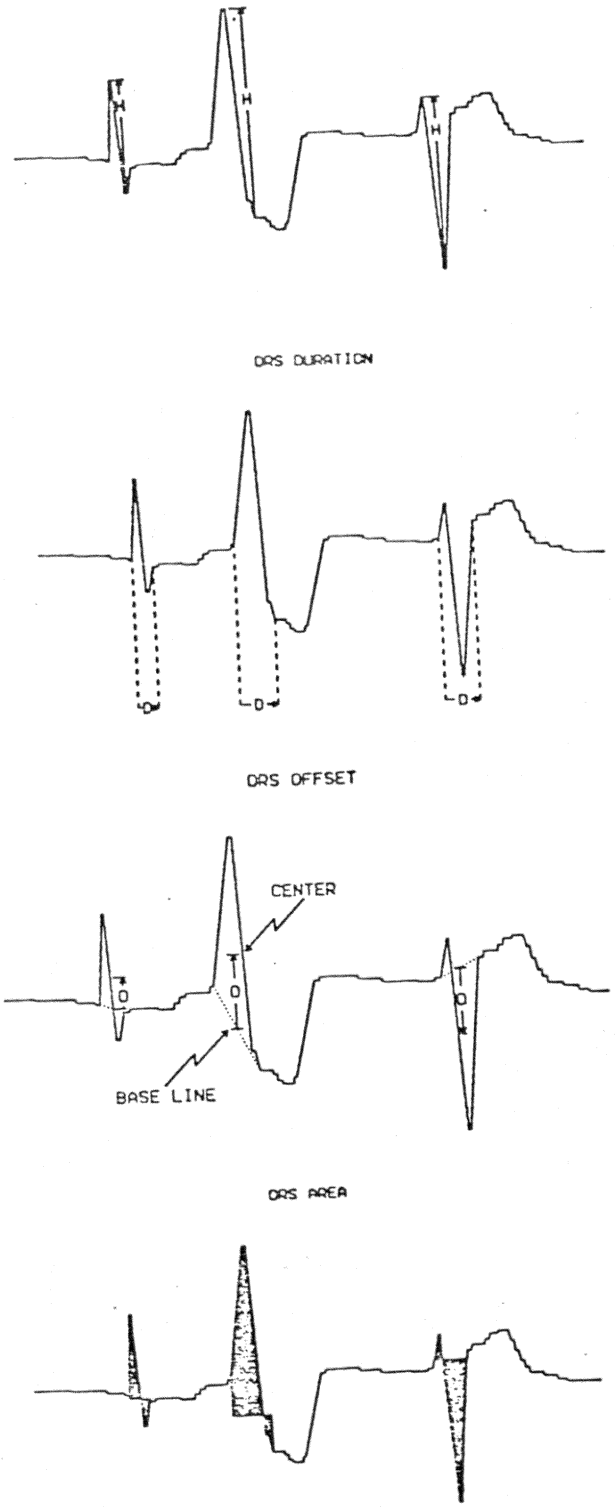

Figure 18: "ARGUS" Four Basic Morphologic Parameters[19]

of the rectified area. If the QRS area (actual) is greater than or equal to ATA, this implies the wave under test is T wave; if the reverse is true, then the wave under test is a QRS wave or pre-mature ventricular beat.

A lot of improvements and changes in ARGUS have been made to up-date and provide better and more efficient results. In the future, ARGUS can increase in its processing capabilities and more rapid data access and less 'human interactivity' in order to achieve a fully automated and efficient ECG analysis system.

7.3.3 Finite State Machine Approach

This system was designed by Bigger, Florenz and Rolnitz[13]. It uses cascaded program modules implemented as finite state machines (FSM), i.e. rapid state table driven algorithms. This program is particularly useful for analysis of prolonged ECG recordings. However the finite state machine (FSM) principle is also applicable for other uses of ECG, for example, pattern recognition.

With the FSM design feature, the pattern recognition section of an ECG system can operate as fast as one hundred and twenty times real time. In the FSM program, the ECG signal is encoded by a digital first-crder linear

interpolator pre-processor as a stream of symbols, US,UL,DS and DL to indicate direction and time in an aperture pre-set by programmer or user. The FSM accepts symbols and quickly detects ECG components, by identifying patterns such as (US,US,US,.....) or (DS,DS,DS,.....). In this manner, the inputs are reduced from many data points to a set of symblic inputs for the Finite State Machine. In the process, one of the following three actions can be taken place for each change in state:

1. accumulate line time;

2. accumulate slope time;

3. call "queries" (another cascaded FSM module) with line/slope data as input.

The queries FSM module uses a state table of sixteen input symbols, which are generated earlier, twenty-one states and thirteen actions for shape and rhythm analysis when a QRS is detected. The FSM program contains features which allow the operator to interact with the computer. Further improvements for the FSM program can be made in the shaping algorithm to give more accurate results. The finite state machine principle can be applied to other feature extraction, recognition and diagnostic processes and should reduce computer-time and the storage spaces.

7.4 DISCUSSION

Each of the programs discussed in this chapter uses a different approach to detect particular waves in the ECG. In Bonner's program, a set of criteria for every individual case is set by the cardiologists for diagnosis. This approach is similar to an intelligent tree-decision method while the ARGUS utilizes a set of cascaded and parallel processors, each of which is similar to a finite state machine. The finite state machine method provides a fast table driven algorithm to attack the ECG pattern recognition and rhythm status monitoring. It may be possible to modify the FSM approach for feature extraction, pattern recognition, as well as, diagnosis.

In ARGUS approach, the method is more systematic than Bonner's program. Bonner's program uses a 'step by step' tree-decision approach which seems to require more computer time than the method used by ARGUS. By linear discriminant functions (multi-dimensional variables), the ECG in ARGUS is classified and diagnosis is provided.

Most designers have used an inter-active approach in the diagnosis portion to make sure the computer does the analysis correctly. (e.g., When the computer comes across some undecided cases, that particular waveform will be

plotted in the output terminal for the operator's judgement or commands.) A great deal of effort has been devoted to minimizing the human inter-action because it is tedious and costly. Also, the operator judgement may very likely vary from one to another. Also, the operators usually require a lot of training or else they have to be cardiologists. In order to minimize the cost of ECG diagnosis, the system must be as automatic, self-explanatory, efficient and accurate as possible.

A number of other approaches have been used by other designers and each has their distinct advantages and disadvantages. New methods and approaches are still being investigated by researchers to improve accuracy, reliability and efficiency. Another promising approach utilizes a polynomial to represent the ECG data. By varying the coefficients of the polynomial, different forms of ECG pattern can be obtained. Thus , it should be easier for the computer to recognize and diagnose ECG features by observing the changes in the set of polynomials and coefficients. It is also possible that each of the coefficients is a parameter in the ECG representing a particular heart problem. In this approach, a very large sample of different combinations of normal and abnormal ECG records must be used in order to establish some co-relation between the

variations of coefficients, order of polynomials and the different types of heart abnormalities.

Some of the other approaches used in ECG analysis are summarized as follows:

1. Karhunen-Loeve Expansion: feature extraction by expanding the ECG waveform into a series of orthonoraml basis functions.

2. Fourier Series(or other series representation) : Transforming the ECG waveform into Fourier Series with diagnosis based on coefficients.

3. Linear Discriminant Functions: A set of multi-dimensional variables are assigned to the waveform and then cataloged into classes by means of distances between each class.

4. Power Spectrum: Similar to Fourier transform, but uses the power spectrum of the ECG waveform. One advantage of this approach is that it is less sensitive to sampling rate than most other approaches and requires lower sampling rate.

5. Operational Waveform Transformation: The ECG waveform is transformaed into another

representation which is more suitable for recognition and analysis. For example, the transforamtion of OWn=(ECG-ECGn-2)**2, is characterized by monophasic and twin-peaked properties.

6. Statistical Methods: The ECG waveform is analyzed by autocorrelation, cross-relation, histcrgram and joint probabilities.

The above list is not complete and is include to the variety of approaches that have been used.

7.5 REPEATIBILITY AND VARIABILITY

In automating ECG diagnosis, one of the goals is high repeatibility. In human diagnosis, there is always a certain degree of inconsistency. It has been shown that computer diagnosis can have a higher degree of consistency than human techniques. However, computer methods are still highly affected by the format of the input data. For example, a different sampling rate of the same analog recording may give different results. Also, a shift in the starting point of the ECG data set may affect the results significantly.

One of Piperberger's papers[14] investigated the variability of diagnostic statements and repeatibility of some available ECG programs. For example, the ECAN-D program had only sixty per cent of the QRS diagnostic statements repeated, while the remaining forty percent was either omitted or altered. The results obtained by Bailey et al[15] is shown in Table 3, Table 4 and Table 5. These results showed the repeatibility of P wave diagnostic statements is less than for QRS and ST-T waveforms. In most programs, the QRS repeatibility is the highest due to its larger amplitudes, greater slopes and other distinct characteristics. As Piperberger pointed out[14], some of the programs do not need further evaluation because of their low rate of repeatibility in certain or all waveform recognition or diagnosis. Among ECAN-D, IBM-Bonner, Mayo-Smith and VA program, it seems VA program gives the best performance in repeatibility of diagnostic statements.

Detail evaluation of ECG computer programs is a lengthy and very difficult process. Although tests of repeatibility are essential, the evaluation should also include diagnostic outputs, wave recognition and measurements. If computer analysis of ECG's is to become a standard laboratory procedure, then there should be no systematic or significant differences in measurement outputs from the analysis of

TABLE 3

Repeat Variability of ECG Analysis (from Bailey et al [15])

	Diagnostic Statements	
ECAN D Program	Repeated	Altered or Omitted
QRS	60%	40%
P	42%	58%
ST-T	59%	41%
IBM-Bonner Program		
QRS	79%	21%
P	72%	28%
ST-T	66%	34%
Mayo-Smith Program		
QRS	76%	24%
P	50%	50%
ST-T	67%	33%

TABLE 4

Repeat Variability of diagnostic statements for Veterans Adminstration program (from Bailey et al [15])

	QRS	P
Change of posterior probabilities by less than 10%	99.6%	93.5%
Change of more than 10% but less than 20%	0.4%	2.1%
Change of more than 20%	0	4.4%
	100%	100%

TABLE 5

Sensitivity and Specificity for diagnosis of acute myocardial infarct (from Harris et al. [17])

	Sensitivity	Specificity	Performance Score
Reader 3	79%	92%	85.5%
VA Computer Program	72%	91%	81.5%
Reader 2	73%	89%	81%
Reader 4	74%	87%	80.5%
IBM Computer Program	63%	97%	80%
Reader 1	87%	56%	71.5%

identical ECG input data. In a paper by Jos L. Willems and Jos Pardaens[16], the differences in the measurement results obtained by four different ECG computer analysis programs are compared.

The AVA[16] and TNO[16] are compared with each other using XYZ orthogonal leads as input data. Then TNO, ECAN-D[16] and HP-5[16] are compared with the staandard twelve leads input. Ten basic measurements were used for comparison. These are QRS duration, the amplitudes and durations of the Q, R and S waves, the amplitudes of point J, the T wave and the QT intervals. Each program is also tested for reproducibility using odd and even data points, a technique orginally used by Bailey et al[15].

Table 6 through table 11 contain data comparing the results of different programs. These results indicate substained differences in time measurements, and in reporting small Q and R waves. The differences in amplitudes were quite small. From these results, it is apparent that AVA and TNO produce significantly greater measurement reliability and reproducibility than either ECAN-D or HP-5. There are two techniques in AVA and TNO programs to which the better performance can be attributed to. First, AVA and TNO locate fiducial points on

TABLE 6

Comparison of Q waves in 168 ECG records by TNO, HP-5 and ECAN-D programs. [16]

	TNO-ECAN-D (250)			HP-5-ECAN-D			TNO-HP-5 (250)			TNO--TNO (250)(500)		
LEAD	A	B	C	A	B	C	A	B	C	A	B	C
I	54	21	12	32	7	34	35	38	4	72	3	21
II	66	22	20	49	8	37	49	38	8	81	7	13
III	67	17	18	69	16	16	71	21	14	75	9	11
aVR	58	19	21	69	33	10	66	6	36	63	14	5
aVL	48	19	10	36	7	22	37	26	6	59	8	13
aVF	69	16	13	56	11	26	54	29	13	82	3	12
V1	19	3	10	25	9	4	19	1	15	20	2	0
V2	15	3	14	17	4	12	14	4	7	18	0	2
V3	16	5	12	15	6	13	17	3	4	20	1	2
V4	28	12	18	20	5	26	24	17	1	37	3	7
V5	74	17	12	45	3	41	47	39	1	90	1	11
V6	85	23	9	49	8	45	57	52	0	107	1	8

A: Q wave reported by both; B: Q wave listed by first, but not by 2nd program; C: inverse of B; in the remaining of 168 cases no Q or QS wave was detected by both.

TABLE 7

Agreement in % on presence of Q or QS wave. [16]

LEAD	4-7	6-7	5-6	3-4	4-5	5-8	6-16
I	80.4	76.8	75.0	85.7	83.3	86.5	88.7
II	73.8	73.2	72.6	87.5	91.1	88.3	88.3
III	79.2	81.0	79.2	88.1	88.1	92.0	89.3
aVR	76.2	74.4	75.0	88.7	85.1	87.1	82.7
aVL	82.7	82.7	81.0	87.5	91.7	89.6	94.6
aVF	82.7	78.0	75.0	91.1	92.9	90.8	93.5
V1	92.3	92.3	90.5	98.8	97.6	97.5	95.8
V2	89.9	90.5	93.5	98.8	97.6	97.5	97.0
V3	89.9	88.7	95.8	98.2	97.1	96.9	96.4
V4	82.1	81.5	89.3	94.0	94.6	95.7	96.4
V5	82.7	73.8	76.2	92.9	93.5	91.4	89.3
V6	81.6	68.5	69.0	94.6	91.1	93.9	88.1

3: TNO (500 Hz); 4 and 5: TNO (250 Hz) respectively using odd and even sample points; 8: TNO (300 Hz); 16 and 6: HP-5 using odd and even data points; 7: ECAN-D

TABLE 8

Comparison of QRS duration results (n=168) [16]

LEAD	Mean values TNO (250)	ECAN-D	HP-5	Paired differences 2-1	2-3	3-1
I	83.3	91.5	77.5	8.2 * (2.1)	14.1 (2.2)	-5.6 (1.7)
II	88.9	96.7	87.5	7.8 (2.1)	9.2 (2.4)	-1.4 (1.8)
III	87.7	94.3	82.4	-6.6 (2.0)	11.9 (2.0)	-5.3 (1.5)
aVR	80.9	83.4	73.7	2.5 (2.0)	9.7 (1.9)	-7.1 (1.6)
aVL	84.2	92.1	77.2	7.9 (2.0)	14.8 (2.2)	-7.0 (1.4)
aVF	86.3	90.5	81.5	4.2 (1.7)	9.0 (1.7)	-4.8 (1.6)
V1	91.9	95.6	91.9	3.7 (1.2)	3.8 (1.4)	0.0 (1.4)
V2	96.4	103.2	101.4	6.8 (1.2)	1.9 (1.1)	4.9 (1.1)
V3	94.7	100.5	99.2	5.8 (1.1)	1.3 (0.9)	4.4 (0.9)
V4	89.7	96.4	95.2	6.8 (1.2)	1.2 (1.1)	5.6 (1.0)
V5	88.3	94.3	87.4	6.0 (1.3)	6.9 (1.4)	-0.9 (1.4)
V6	87.2	88.6	82.4	1.4 (1.3)	6.1 (1.2)	-0.5 (1.2)

* figures between parentheses are standard errors of paired mean differences

simultaneously recorded leads, while the other programs perform single lead analysis. Second,, AVA computes measurements for each complex and then averaged from morphologically similar complexes, and TNO program computes an average complex from all similar complexes and then measurements are extracted from this average complex. In contrast, the ECAN-D and HP-5 programs use the so-called 'best complex' for analysis.

The evaluations by Bailey et al,[15] and Jos L. Willems and Jos Pardaens[16] and others showed that analysis results have different degrees of accuracy. Different evaluations procedures may result in different conclusions. As suggested in Dr. Piperberger's paper[14], there are some principal considerations for planning comprehensive ECG system evaluation. They are as follows:

1. Use an adequate data base consisting of ECG records with a cross-section of normal and a great variety of abnormal.

2. Standard recording equipment and procedures should be developed in close do-operation with ECG system designers.

3. Standardize program output terminology problem.

TABLE 9

Frequency distribution of Q, R and S waves detected in X. Y. Z leads by the AVA and TNO computer program (N=250) [16]

	AVA - TNO (500)			AVA - TNO (250)			Odd - Even TNO (250 Hz)		
WAVE	A	B	C	A	B	C	A	B	C
Q - X	144	22	3	137	29	4	136	5	7
- Y	147	21	13	149	19	10	148	11	11
- Z	221	3	6	221	3	5	223	3	0
R - X	248	0	1	248	0	1	249	0	0
- Y	240	1	7	240	1	5	245	0	1
- Z	249	1	0	249	1	0	248	1	0
S - X	163	19	4	166	16	1	164	3	5
- Y	123	9	17	121	11	13	127	7	6
- Z	10	5	4	9	6	1	8	2	3

A: measurement results reported by both; B: by first program but not by 2nd; C: by 2nd but not by 1st; D: by none (D=250 - (A + B + C)

TABLE 10

Mean Values from AVA (3.6) and TNO ECG computer programs in 250 VCG records. [16]

Duration	AVA	TNO (500)	TNO (250)
QRS	98.8	107.4	106.7
Q - X	21.8 (14.0)	23.7 (13.8)	22.3 (12.4)
- Y	28.0 (19.9)	28.2 (17.1)	28.7 (17.5)
- Z	37.2 (33.3)	40.7 (36.3)	39.5 (35.2)
R - X	50.6	59.3	60.9
- Y	52.1	57.6	57.9
- Z	57.0	63.0	63.2

TABLE II

Mean Differences in Duration and Amplitude between AVA and TNO programs.[16]

Measurement	3-1[x]	4-1	4-3	4-5	8-5
Q duration					
- X	-0.2	-1.5	-1.4	0.2	-0.3
	(0.6)	(0.6)	(0.3)	(0.3)	(0.3)
- Y	-2.8	-2.4	-0.4	-0.4	-0.1
	(1.1)	(1.0)	(0.8)	(0.8)	(0.8)
- Z	3.1	2.0	-1.1	-0.0	-0.1
	(0.5)	(0.4)	(0.4)	(0.4)	(0.5)
R duration					
- X	8.6	10.3	1.7	-0.4	-0.4
	(0.7)	(0.7)	(0.6)	(0.5)	(0.6)
- Y	5.6	5.9	0.3	0.5	-1.3
	(0.6)	(0.6)	(0.5)	(0.5)	(0.6)
- Z	6.0	6.2	0.2	0.6	-0.3
	(0.5)	(0.5)	(0.3)	(0.4)	(0.5)
R amplitude					
- X	0.2	-13.5	-13.7	-0.7	9.0
	(3.2)	(3.4)	(1.7)	(1.6)	(1.7)
- Y	-4.1	-19.9	-15.8	-1.9	10.6
	(2.3)	(2.5)	(2.1)	(1.8)	(1.8)
- Z	-4.6	-14.8	-10.2	1.2	4.0
	(1.8)	(2.0)	(1.4)	(1.4)	(1.4)
J amplitude					
- X	4.7	5.2	-0.4	-0.9	-1.9
	(1.7)	(1.8)	(0.7)	(1.8)	(1.8)
- Y	0.3	-0.1	-0.4	0.5	-0.1
	(1.6)	(1.5)	(1.1)	(4.1)	(3.7)
- Z	-12.1	-13.0	-0.1	0.5	1.4
	(1.2)	(1.2)	(0.6)	(1.7)	(2.1)

x 1=AVA results; 3=TNO(500 Hz); 4, 5=TNO(250 Hz) results derived respectively from odd and even data points; 8=TNO results using a sampling interval of 3 msec.

4. Careful analysis of repeatibility in diagnostic outputs, measurement routines and wave recognitions.

5. The use of objective yardsticks for comparative evaluation of programs.

Chapter VIII

SUMMARY AND CONCLUSIONS

8.1 SUMMARY

The objective of this thesis is to investigate ECG computer analysis programs and develop a more effective system. In the development of the work, the author has examined the property of various sampling rates and their effects on the recognition and analysis results. The conclusion drawn is in close agreement with the work done by earlier researchers. Sampling rates of 250 and 500 samples per second are used as standards throughout the development of the algorithm. A brief study of the simple smoothing or filter techniques is made and a parabolic smoothing technique with pre-set threshold as described in chapter three is implemented in the final algorithm. This smoothing technique eliminate most of the 60 cycle interference and its harmonics. However, investigation of other sampling techniques should be continued which may provide better smoothing results, for example, a non-uniform sampling rate.

The various recognition methods for QRS onset and offset are studied and four of the methods are tested experimentally in this thesis with the available ECG

records. The GF transform is studied and tested experimentally for its constraints, limitations, accuracy and flexibility in implementation. The GF transform, originally proposed for single lead ECG record, is modified in this thesis to obtain the spatial GF transform for accepting the three lead system(Frank Orthogonal Lead System) and providing the QRS offset point uniquely for the three lead system without degrading the accuracy of the original transform. This modified transform is then implemented in the algorithm and it is able to identify consistenly and correctly the offset location of QRS in the Frank Orthogonal Lead System.

The operational transform, originally proposed by Kitney, Turner and McDonald, was intended for locating the general occurrence of the QRS events in single lead ECG record. In the development of this thesis, this transform was modified for obtaining not only the general QRS occurrence but also the exact QRS onset point for the three lead systems.

Thus, the modified GF and operational transforms are implemented in the algorithm for locating onset and offset points of QRS in the Frank Lead system and the results shows complete agreement with visual (traditional) inspection of the analog ECG records. In the implementation of these two

modified transforms, studies are made to access their constraints and flexibility. The only constraint is that both transforms require an interval between successive data input points which is greater than four milli-seconds in order to locate QRS accurately. The implementation of these two modified transforms is done in standard Fortran IV. The exact computation time for locating QRS onset and offset in other systems, like ARGUS and Bonner's programs, are not available, but the principle and the approaches (as discussed in the literature review) show that the two modified transforms here require fewer instructions to implement in a computer program.

The approach shown in this thesis is easy to implement and insensitive to noise and artifacts. The algorithm is also able to overcome the problems associated with distinguishing a small QRS waveform from a large T wave and a pre-mature ventricular contraction from a regular QRS waveform. The algorithm has been tested with over three hundred cardiac cycles of electrocardiogrm from ten different subjects(normal and abnormal), and the results show that all the 300 cardiac cycles are identified correctly with reasonable accuracy. Though the subject sample is small, but it illustrates the feasibility of implementation of the approach. In addition, the originator

for the GF transform has tested the GF transform for a much larger sample with good results.

In the development of this thesis, an algorithm for detecting and locating the exact location of P and T waves in the three lead system is implemented. The approach used for this is quite similar to that implemented in Bonner's program. The developed algorithm also contains of a plotting algorithm to generate the different clinical parameter plots in addition to the various quantitative parameters.

The present algorithm has been designed in a modular format such that it can be expanded or shortened without much difficulty. The algorithm is divided into small functional modules and user can interpret or translate each module to meet the need of particular systems. The modular concept also enables this algorithm to have additional add-on features (e.g. three-dimensional display, diagnosis for particular heart diseases) by the addition of an appropriate module. It is written in ANSI Fortran to facilitate portability, ease of understanding and implementation on other systems. Another merit of the the algorithm is ease with which it can be applied

The algorithm provides all the clinically important parameters required for detail cardiac studies. The vector display and analysis also enable the user to have a vectorcardiogram anlaysis. In the present system, the display of vectors is only two-dimensional for any plane of view; however, high quality three-dimensional display of vectors is possible. Further development of the system can address the high quality three-dimensional display with specified rotation and any viewing angle. The parameters provided by this algorithm can be used for ECG analysis and as input to other modules such as 3-D display and/or diagnosis for LVH. In some applications, the output can be stored instead of the actual ECG data, thus reducing memory space required.

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Appendix A

COMPUTER PROGRAM LISTINGS AND DOCUMENTATION

```
C
C
C*************************************************************************C
C*********************                                      ******************C
C*********************COMPUTER-AIDED ELECTROCARDIOGRAM ANALYSIS******************C
C*********************                                      ******************C
C*********************        WRITTEN            BY         ******************C
C*********************                                      ******************C
C*********************    PETER          H.          PAN     ******************C
C*********************                                      ******************C
C*********************                   IN                 ******************C
C*********************                                      ******************C
C*********************         DECEMBER        1979         ******************C
C*********************                                      ******************C
C*************************************************************************C
C
C
C
C THIS PROGRAM IS A COMPUTER-AIDED ANALYSIS ALGORITHM FOR LOCATING ACCURATELY
C THE ONSETS AND OFFSETS OF P, T AND QRS WAVES AND ALSO VARIOUS CLINICAL PARA-
C METERS ARE CALCULATED AND AVAILABLE FOR PLOT ROUTINES.  THE ALGORITHM CONSISTS
C A SET OF SUBROUTINES, EACH OF WHICH DOES A PARTICULAR FUNCTION, FOR EXAMPLE,
C FILTERING AND SMOOTHING.  IN THIS ALGORITHM, IT ALSO CONSISTS OF SOME TESTING
C SUBROUTINES WHICH SERVE FOR THE PURPOSES OF EXAMINING THE VARIOUS PROPERTIES
C OF DIFFERENT METHODS.  THE ALGORITHM MAY CONSISTS OF SUBROUTINES DO THE
C SAME JOB IN DIFFERENT APPROACHES, SUCH THAT THE AUTHOR CAN STUDY THE EFFECT
C AND RESULTS PRODUCED BY DIFFERENT APPROACHES.
C
C
C
C
C SPV IS SPATIAL VELOCITY FUNCTION
```

```
C SPM IS SPATIAL MAGNITUDE FUNCTION
C SPOW IS SPATIAL OPERATIONAL TRANSFORM
C SPGF IS SPATIAL GF TRANSFORM
C X, Y AND Z ARE THE DATA OF THE THREE LEAD IN FRANK ORTHOGONAL LEAD SYSTEM
C G AND GF ARE THE CORRESPONDING TRANSFORM FOR DERIVING SPATIAL GF TRANSFORM
C
C
C
C
      DIMENSION IA(15060),X(5000),Y(5000),Z(5000)
      DIMENSION SPV(3000),SPM(3000),SPW(3000),SPOW(3000),SPGF(3000)
      DIMENSION XW(3000),YW(3000),ZW(3000)
      DIMENSION XX(500),YY(500),ZZ(500)
      DIMENSION SMAX(100),IX(100)
      DIMENSION DELSXN(300),DELSYN(300),DELSZN(300)
      DIMENSION G(3000),GF(3000),GFX(3000),GFY(3000),GFZ(3000)
      DIMENSION QBEG(100),QEND(100)
      DIMENSION QP(5,10),RP(5,10),SP(5,10)
      DIMENSION QM(5,10),RM(5,10),SM(5,10)
      DIMENSION TPEAK(10),TMAG(10),TON(10),TOFF(10)
      DIMENSION FACNX(10),FACNY(10),FACNZ(10)
      DIMENSION IPBEG(10),IPEND(10),IPEAK(10),PPEAK(10),PPLOC(10)
      DIMENSION IQBEG(10),IQEND(10)
      DIMENSION XM(500),H(500),V(500)
      INTEGER * 2 IA
      DO 10 I=1,15060,60
      K=I+59
      READ (5,20) (IA(J),J=I,K)
   20 FORMAT(60A2)
   10 CONTINUE
      L=0
      DO 30 I=24,15000,3
```

```
      J=I+1
      K=I+2
      L=L+1
      X(L)=IA(I)*1.0
      Y(L)=IA(J)*1.0
      Z(L)=IA(K)*1.0
   30 CONTINUE
      DO 50 I=1,500
      XK=(I-1)*1.0
   50 CONTINUE
   55 FORMAT(' FIRST',5F12.2)
      DO 60 I=5,2000
      X(I)=-0.0909*(X(I-4)+X(I+4))+0.0606*(X(I-3)+X(I+3))
     C     +0.1688*(X(I-2)+X(I+2))+0.2337*(X(I-1)+X(I+1))
     C     +0.2554*X(I)
      Y(I)=-0.0909*(Y(I-4)+Y(I+4))+0.0606*(Y(I-3)+Y(I+3))
     C     +0.1688*(Y(I-2)+Y(I+2))+0.2337*(Y(I-1)+Y(I+1))
     C     +0.2554*Y(I)
      Z(I)=-0.0909*(Z(I-4)+Z(I+4))+0.0606*(Z(I-3)+Z(I+3))
     C     +0.1688*(Z(I-2)+Z(I+2))+0.2337*(Z(I-1)+Z(I+1))
     C     +0.2554*Z(I)
   60 CONTINUE
      DO 65 I=1,501,2
      XK=(I-1)*1.0
      XX(I)=X(I)*1.0
      YY(I)=Y(I)*1.0
      ZZ(I)=Z(I)*1.0
   65 CONTINUE
      XWMAX=0.0
      YWMAX=0.0
      ZWMAX=0.0
      DO 89 I=1,2
```

```
      XK=(I-1)*1.0
      XW(I)=(X(3)-X(1))**2
      YW(I)=(Y(3)-Y(1))**2
      ZW(I)=(Z(3)-Z(1))**2
      IF (XW(I).GT.XWMAX) XWMAX=XW(I)
      IF (YW(I).GT.YWMAX) YWMAX=YW(I)
      IF (ZW(I).GT.ZWMAX) ZWMAX=ZW(I)
      SPW(I)=SQRT(XW(I)**2+YW(I)**2+ZW(I)**2)
   71 FORMAT(1X,2F12.2)
   89 CONTINUE
      DO 90 I=3,1500
      XW(I)=(X(I)-X(I-2))**2
      YW(I)=(Y(I)-Y(I-2))**2
      ZW(I)=(Z(I)-Z(I-2))**2
      IF (XW(I).GT.XWMAX) XWMAX=XW(I)
      IF (YW(I).GT.YWMAX) YWMAX=YW(I)
      IF (ZW(I).GT.ZWMAX) ZWMAX=ZW(I)
      SPW(I)=SQRT(XW(I)**2+YW(I)**2+ZW(I)**2)
      XK=I*1.0-1.0
   90 CONTINUE
      READ (10,*)N,T,M,PERIOD
   95 FORMAT( ' THIS IS ',5F12.2)
   96 FORMAT(' BEFORE CALL SMOOTH')
      CALL SMOOTH (X,Y,Z,N)
  300 FORMAT(' AGTER SMOOTH BEFORE SPMV')
      CALL SPMV(X,Y,Z,M,T,SPM,SPV)
  320 FORMAT(' AFTER SPMV BEFORE SLPE')
      CALL SLOPE(X,DELSXN,M,GXMAX,GFX)
  330 FORMAT(' AFTER SLOPE BEFORE SLOPE Y')
      CALL SLOPE(Y,DELSYN,M,GYMAX,GFY)
      CALL SLOPE(Z,DELSZN,M,GZMAX,GFZ)
  350 FORMAT(' BEFORE SPACE')
```

```
      CALL SPACE(GFX,GFY,GFZ,GXMAX,GYMAX,GZMAX,16,500,SPGF)
      DO 365 I=1,500
      XSI= I*3.-2.
  360 FORMAT(5X,2F12.3)
  365 CONTINUE
      CALL SPACE(XW,YW,ZW,XWMAX,YWMAX,ZWMAX,1,1500,SPOW)
      DO 101 IJ=1,500
      I=IJ*3-2
      XI=IJ*3.0-3.0
  102 FORMAT(5X,2F12.3)
  101 CONTINUE
      SPOMAX=SQRT(XWMAX**2+YWMAX**2+ZWMAX**2)
      GFMAX=2.0*GXMAX
      CALL FIND(SPGF,SPOW,GFMAX,SPOMAX,QBEG,QEND)
      CALL NORMA(X,Y,Z,QBEG,QEND,FACNX,FACNY,FACNZ)
      GO TO 150
  150 DO 98 I=1,1270,2
      WRITE(1,99)I,X(I),Y(I),Z(I)
   99 FORMAT(1X,'X,Y,Z',I10,3F12.3)
   98 CONTINUE
  980 CONTINUE
      CALL QRSPGM(X,QBEG,QEND,QP,RP,SP,QM,RM,SM,FAC)
  220 FORMAT( ' AFTER CALL QRSPAGM OF X')
      CALL QRSPGM(Y,QBEG,QEND,QP,RP,SP,QM,RM,SM,FAC)
      CALL QRSPGM(Z,QBEG,QEND,QP,RP,SP,QM,RM,SM,FAC)
      CALL TWAVE(QBEG,QEND,X,TPEAK,TMAG,TON,TOFF,IQBEG,IQEND)
      CALL TWAVE(QBEG,QEND,Y,TPEAK,TMAG,TON,TOFF,IQBEG,IQEND)
      CALL TWAVE(QBEG,QEND,Z,TPEAK,TMAG,TON,TOFF,IQBEG,IQEND)
      CALL PWAVE(IQBEG,IQEND,IPBEG,IPEND,IPEAK,PPEAK,TOFF,PPLOC,X)
      CALL VECT(XM,H,V,QBEG(1),QEND(1),X,Y,Z)
      VX=1.0
      VY=0.0
```

```
      VZ=0.0
      CALL VCPOT(X,Y,Z,IQBEG(1),IQEND(1),IPBEG(1),IPEND(1),TON(1),TOFF(1
     *))
      CALL CONV(X,Y,Z,VX,VY,VZ,IQBEG(1),IQEND(1))
      DO 400 I=16,500
      J=I*3-2
      XJX=J*1.0-46.
      WRITE(19,310)XJX,X(J),Y(J),Z(J)
      WRITE(20,310)XJX,SPOW(I),SPGF(I),SPOW(I)
  310 FORMAT(5X,4F12.2)
  400 CONTINUE
      STOP
      END
C
C
```

```
C
C THE SMOOTH SUBROUTINE DOES THE JOB OF TAKING THE X, Y AND Z DATA AND APPLYING
C A PARABOLIC SMOOTHING TECHNIQUE WITH A PRE-SET THRESHOLD FOR ATTENUATING THE
C EFFECT CAUSED IN HIGH FREQUENCY COMPONENTS IN THE SIGNAL.
C
C
C W(I) IS THE I TH WEIGHING FUNCTION FOR THE PARABOLIC SMOOTHING CURVE
C N IS THE NUMBER OF WEIGHING FUNCTION APPLIED FOR THE SMOOTHING
C
C
      SUBROUTINE SMOOTH(X,Y,Z,N)
      DIMENSION W(21),X(5000),Y(5000),Z(5000)
      IN=(N-1)/2
      IST=(N-1)/2+1
      DO 10 I=1,IN
      W(I)=(3.*(3.*N**2.-7.-20.*I**2.))/(4.*N*(N**2.-4.))
      J=N-I+1
      W(J)=W(I)
   10 CONTINUE
      W(21)=(3.*(3.*N**2.-7.))/(4.*N*(N**2.-4.))
      DO 20 I=5,3000
      TEX=W(21)*X(I)
      TEY=W(21)*Y(I)
      TEZ=W(21)*Z(I)
      DO 30 L=1,IN
      J=I-L
      K=I+L
      TEX=TEX+W(L)*(X(J)+X(K))
      TEY=TEY+W(L)*(Y(J)+Y(K))
      TEZ=TEZ+W(L)*(Z(J)+Z(K))
   30 CONTINUE
C
```

```
C
C THE FOLLOWING DOES THE PRE-SET THRESHOLD SECTION OF THE SMOOTHING
C
   33 DIFX=ABS(X(I)-TEX)
      DIFY=ABS(Y(I)-TEY)
      DIFZ=ABS(Z(I)-TEZ)
      IF(DIFX.LT.50.0) X(I)=TEX
      IF(DIFY.LT.50.0) Y(I)=TEY
      IF (DIFZ.LT.50.0) Z(I)=TEZ
   20 CONTINUE
      RETURN
      END
C
```

```
C
C
C
C THE SPMV SUBROUTINE IS A TESTING SUBROUTINE USED DURING THE STUDY OF DIFFERENT
C APPROACHES FOR RECOGNIZING THE ONSET AND OFFSETS OF QRS WAVE.  THIS SUBROUTINE
C INPUTS X, Y, Z RECORDS, THE SAMPLING INTERVAL AND THE NUMBER OF POINTS FOR
C THE RECORDS; AND PRODUCES THE SPATIAL MAGNITUDE AND SPATIAL VELOCITY FUNCTIONS
C
C
C
      SUBROUTINE SPMV(X,Y,Z,M,T,SPM,SPV)
      DIMENSION X(3000),Y(3000),Z(3000),SPM(3000),SPV(3000)
      DIMENSION XAGSPV(3000)
      NM1=M-1
      DO 100 I=1,NM1
      DELX = (X(I+1) - X(I))**2
      DELY = (Y(I+1) - Y(I))**2
      DELZ = (Z(I+1) - Z(I))**2
      SUM = SQRT(DELX+DELY+DELZ)
      SPV(I) = SUM/T
      XAGSPV(I) = SPV(I)*2.5
      SPM(I) = SQRT(X(I)**2+Y(I)**2+Z(I)**2)
      XI=(I-1)*1.0
   99 FORMAT(1X,4F12.2)
  100 CONTINUE
      SPM(M) = SQRT(X(M)**2+Y(M)**2+Z(M)**2)
      RETURN
      END
C
C
```

```
C
C THE SEARCH SUBROUTINE WAS INVESTIGATED FOR LOCATING THE MAXIMA, THIS ROUTINE
C IS NO LONGER USED.
C
C
      SUBROUTINE SEARCH(DUMMY,IX,SMAX,PERIOD,M)
      DIMENSION DUMMY(5000),IX(100),SMAX(100)
      N=M-1
      CRIT=0.1/PERIOD
      DO 10 I=1,100
      IX(I)=0
      SMAX(I)=0.01
   10 CONTINUE
C
      K=1
      DO 92 I=1,400
      J=I-1
      IF (SMAX(K) .GT. DUMMY(I)) GO TO 40
   20 IF((J-IX(K)) .GT. 50) GO TO 50
      IX(K)=J
      SMAX(K)=DUMMY(I)
      SMAX(K-1)=TEMP
      GO TO 90
   50 K=K+1
   51 SMAX(K)=DUMMY(I)
    1 IX(K)=J
      GO TO 90
C
C
C WHEN DUMMY IS SMALLER THAN PREVIOUS MAX.
C
C
```

```
   40 IF ((J-IX(K)) .LT. 51) GO TO 90
      TEMP=SMAX(K)
      K=K+1
      IK=K-1
      IX(K)=J
      SMAX(K)=DUMMY(I)
      SMAX(K-1)=TEMP
      GO TO 90
   90 CONTINUE
      IJ=K-1
   92 CONTINUE
  100 CONTINUE
      RETURN
      END
C
C
```

```
C
C
C THE SLOPE SUBROUTINE PERFORMS THE FUNCTION OF FINDING THE GF TRANSFORM AND
C ALSO FINDING THE MAXIMUM VALUE OF THE CORRESPONDING GF TRANSFORM
C
C
      SUBROUTINE SLOPE(DUM,DELSN,M,GMAX,GF)
      DIMENSION DUM(3000),DELSN(200)
      DIMENSION G(3000)
      DIMENSION GF(3000),GDUM(600)
      IN=M-2
      DO 100 I=1,5
      DELSN(I)=-(DUM(I+2)-2.0*DUM(I+1)+DUM(I))
  100 CONTINUE
  101 DO 110 I=1,550
      J=I*3-2
      GDUM(I)=DUM(J)
      XX=I*1.0
      XJ=J*1.0
  110 CONTINUE
      MN=8
      DO 200 N=9,500
      G(N)=0.00
      DO 250 I=1,MN
      GP=(GDUM(N-I+1)-GDUM(N-I))**2
      GP=GP/100.0
      G(N)=G(N)+(GP*(MN-I+1))/10.0
  250 CONTINUE
      XN=N*3.0-2.0
  200 CONTINUE
      GMAX=0.0
      DO 300 I=16,500
```

```
      GF(I)=0.0
      DO 350 J=1,7
      GF(I)=GF(I)+1.0/8.0*G(I-J)
  350 CONTINUE
      GF(I)=1.0/8.0*G(I)+GF(I)
      IF (GF(I).GT.GMAX) GMAX=GF(I)
      XX=I*3.0-2.0
  300 CONTINUE
      RETURN
      END
C
C
```

```
C
C
C THE SUBROUTINE SPACE CALCULATE THE SPATIAL OWN TRANS AND THE
C SPATIAL GTRANS WITH THE FEATURE OF EQUALIZING THE THREE AXIS
C WITH A FACTOR MULTIPLY TO EACH OF THE THREE AXIS SIGNAL BEFORE
C THE SPATIAL FUNCTION IS FOUND
C
C
C
      SUBROUTINE SPACE(A,B,C,AMAX,BMAX,CMAX,IBPT,IEPT,SP)
      DIMENSION A(3000),B(3000),C(3000),SP(3000)
      SPMAX=BMAX
      IF (AMAX.GT.BMAX) SPMAX=AMAX
      IF (CMAX.GT.SPMAX) SPMAX=CMAX
      AFAC=SPMAX/AMAX
      BFAC=SPMAX/BMAX
      CFAC=SPMAX/CMAX
C
      WRITE(3,99)SPMAX,AMAX,BMAX,CMAX
   99 FORMAT(10X,4F15.2)
      WRITE(3,98)AFAC,BFAC,CFAC
   98 FORMAT(10X,3F15.2)
C
      DO 100 I=IBPT,IEPT
      ASUM=(A(I)*AFAC)**2
      BSUM=(B(I)*BFAC)**2
      CSUM=(C(I)*CFAC)**2
      SP(I)=SQRT(ASUM+BSUM+CSUM)
  100 CONTINUE
      RETURN
      END
C
```

```
C THE FOLLOWING SUBROUTINE FIND DOES THE JOB OF FINDING THE MAX VALUE
C OF SPATIAL GF TRANSFORM AND THE LAST MINIMA OF THE SPATIAL ADJUST OWN
C TRANSFORM BEFORE THE RIPPLES OCCUR.
C
C
C THE MAX POINT OF SPATIAL GF TRANSFORM FUNCTION IS THE POINT WHERE THE
C QRS ENDS; AND THE DESCRIBED MINIMA OF THE SPATIAL ADJUST OWN TRANSFORM
C FUNCTION IS THE POINT WHERE THE QRS BEGINS.
C
C
      SUBROUTINE FIND(SPGF,SPOW,GFMAX,SPOMAX,QBEG,QEND)
      DIMENSION SPGF(3000),SPOW(3000),QBEG(100),QEND(100)
      DIMENSION IFUN(1000),ITABLE(2,5)
      THRES1=GFMAX*0.4
      K=1
      I=13
   10 IF (I.GT.496) GO TO 100
      I=I+3
   20 IF (SPGF(I).GT.THRES1) GO TO 50
      GO TO 10
   50 IF ((SPGF(I)-SPGF(I+1)).GE.0.0) GO TO 10
   55 IF (SPGF(I+2)-SPGF(I+1)) 60,60,70
   60 QEND(K)=(I+1)*3.0-2.0
      WRITE(1,600)K,QEND(K)
  600 FORMAT(1X,'QRS NUMBER',I5,' ENDS AT SAMPLE ',F10.2)
      K=K+1
      GO TO 10
   70 I=I+1
      GO TO 50
C THE FOLLOWING FIND MA
C THE FOLLOWING FIND THE BEGINNING POINT FOR QRS"S
C
```

```
  100 K=1
      THRES1=SPOMAX*5.0/100.0
      THRES2=SPOMAX*1.0/100.0
      I=46
  110 IF (SPOW(I).LT.THRES2) GO TO 150
      IF (I.GT.1499) GO TO 300
      I=I+2
      GO TO 110
C
  150 I=I+2
  160 IF (SPOW(I).GT.THRES1) GO TO 200
      IF(I.GT.1499) GO TO 300
      I=I+2
      GO TO 160
C
C
  200 J=I
  210 IF ((SPOW(J)-SPOW(J-1)).LT.0.0) GO TO 250
      J=J-1
      GO TO 210
C
C
  250 QBEG(K)=J*1.0
      WRITE(1,290) K,QBEG(K)
  290 FORMAT(1X,'QRS NUMBER',I5,' BEGINS AT SAMPLE',F10.2)
      K=K+1
      I=I+10
      GO TO 110
C
  300 RETURN
      END
C
```

```
C
C
C
C THE FOLLOWING SUBROUTINE QRSPGM DOES THE JOB OF FINDING THE LOCATION
C , TIME AND MAGNITUDE OF THE Q, R AND S WAVE FROM THE GIVEN SETS
C OF QBEG'S AND QEND'S.
C
C
      SUBROUTINE QRSPGM (VAR,QBEG,QEND,QP,RP,SP,QM,RM,SM,FAC)
      DIMENSION VAR(1500),QBEG(100),QEND(100)
      DIMENSION QP(5,10),RP(5,10),SP(5,10)
      DIMENSION QM(5,10),RM(5,10),SM(5,10)
      TMAX=VAR(QBEG(1))
      TMIN=TMAX
      DO 10 I=1,5
      M=QBEG(I)
      MM=QEND(I)
      WRITE(1,11)M,QBEG(I),VAR(M),QEND(I),VAR(MM)
   11 FORMAT(1X,'M=',I10,'  QBEG(I) = ',F10.2,3X,3F12.2)
      IF (VAR(M).GT. TMAX) TMAX=VAR(M)
      IF (VAR(M) .LT. TMIN) TMIN=VAR(M)
      DO 9 K=1,10
      QM(I,K)=0.0
      RM(I,K)=0.0
      SM(I,K)=0.0
      QP(I,K)=0.0
      RP(I,K)=0.0
      SP(I,K)=0.0
    9 CONTINUE
   10 CONTINUE
      WRITE(1,15) TMAX,TMIN
   15 FORMAT(1X,'TMAX= ',F10.2,5X,'TMIN= ',F10.2)
```

```
      IF ((TMAX-TMIN) .LT. 100) GO TO 20
   20 DIFF=(TMAX-TMIN)*0.5
      FAC=(DIFF+TMIN)
      WRITE(1,21)DIFF,FAC
   21 FORMAT(1X,'DIFF,FAC',2F12.2)
C
C
      DO 150 I=1,5
      K=0
      IQ=0
      IR=0
      IS=0
      IQBEG=QBEG(I)+1
      IQEND=QEND(I)-1
      DO 100 J=IQBEG,IQEND
      IDEF=100
      FIR=VAR(J)-VAR(J-1)
      SEC=VAR(J+1)-VAR(J)
      IF(FIR.GT.0.0) GO TO 40
      IF(SEC.GT.0.0) GO TO 41
      GO TO 101
   40 IF (SEC.LT.0.0) GO TO 42
      GO TO 101
   41 IDEF=0
      GO TO 50
   42 IDEF=1
      GO TO 50
   50 SUM=ABS(FIR) + ABS(SEC)
      IF (SUM .GT. 25.0) GO TO 60
      GO TO 101
C
C
```

```
   60 IF (VAR(J) .GT. 0.0) GO TO 80
      IF (IDEF .GT. 0) GO TO 101
      IF (IR .GE. 1) GO TO 70
      K=K+1
      IQ=IQ+1
      QM(I,K)=VAR(J)
      QP(I,K)=J*1.0
      GO TO 99
C
C
C FOLLOWING IS S ROUTINE
   70 K=K+1
      IS = IS+1
      SP(I,K) =J*1.0
      SM(I,K)=VAR(J)
      GO TO 99
C
C FOLLOWING IS MAXIMA ROUTINE THAT IS R ROUTINE
   80 K=K+1
      IR =IR+1
      RM(I,K)=VAR(J)
      RP(I,K)=J*1.0
C
   99 K=K
      WRITE(1,955)QM(I,K),RM(I,K),SM(I,K),I,K
  955 FORMAT('  MAG THEN POSITION ',3F10.2,2I10)
      WRITE(1,955) QP(I,K),RP(I,K),SP(I,K),I,K
      WRITE(1,956)IR,IQ,IS
  956 FORMAT(1X,'IR,IQ,IS',3I10)
  101 CONTINUE
  100 CONTINUE
  150 CONTINUE
```

```
  200 RETURN
      END
C
C
C
```

```
C
C
C
C THIS SUBROUTINE IS A TEST ROUTINE FOR STUDY OF METHODS USED IN LOCATING
C THE P ONSET AND OFFSET; BUT THIS APPROACH HAS FOUND OUT TO BE INADEQUATE FOR
C THE PURPOSE.
C  THE SUBROUTINE PWAVE DOES THE JOB OF LOCATING THE ONSET AND OFFSET
C  OF THE PWAVE FROM THE GIVEN DATA OF QBEG AND QEND.
C
C  THIS SUBROUTINE UTILIZES THE SLOPE DIFFERENCE RATIO TO CALCULATE
C  THE LOCATION OF THE OFFSET OF THE P WAVE
C
C
      SUBROUTINE PAVE(QBEG,QEND,IPBEG,VARE,VAL,FAC)
      DIMENSION IQBEG(10),IQEND(10),QBEG(10),QEND(10)
      DIMENSION VARE(1500)
      DO 10 I=1,5
      IQBEG(I)=QBEG(I)
      IQEND(I)=QEND(I)
   10 CONTINUE
      DO 21 K=1,3
      DO 20 I=3,120
      J=IQBEG(K)-I
      VARE(J)=VARE(J)-FAC
   20 CONTINUE
   21 CONTINUE
      K=0
      DO 99 K=1,3
      K=K+1
      DO 100 I=3,120
      J=IQBEG(K)-I
      DEM=VARE(J+1)
```

```
      XNUM=VARE(J-1)
      FACT=XNUM/DEM
      FACT=ABS(FACT)
      IF(FACT .LT. 2.0) GO TO 50
      J=J-2
      XDEM=VARE(J+1)
      XXNUM=VARE(J-1)
      FACT=XDEM/DEM
      XFACT=XXNUM/XNUM
      XFACT=ABS(XFACT)
      WRITE(1,30)DEM,XDEM,XNUM,XXNUM,J
   30 FORMAT(1X,'DEM,XDEM,XNUM,XX',4F12.2,I10)
      RESU=FACT+XFACT
      IF(RESU.GT. 3.5) GO TO 40
      GO TO 50
   40 IPBEG=J+2
      WRITE(1,45)IPBEG
   45 FORMAT(1X,'IPBEG IS ',1X,I10)
   50 CONTINUE
  100 CONTINUE
   99 CONTINUE
      RETURN
      END
C
C
C
C
```

```
C
C
C
C     THE FOLLOWING SUBROUTINE TWAVE DOES THE JOB OF FINDING THE PEAAKS
C     IN THE SEARCH REGION FOR T WAVE.  IT ALSO DOES THE JOB OF FINDING
C     THE ONSET AND OFFSET OF THE T WAVE.  IN ADDDITION IT DOES THE WORK
C     OF FINDING THE T WAVE CONFIGURATION, FOR EXAMPLE POSITIVE OR NEGAT
C     -IVE OR BIPHASIC AND SO ON.
C
C
C
      SUBROUTINE TWAVE(QBEG,QEND,VARE,TPEAK,TMAG,TON,TOFF,IQBEG,IQEND)
      DIMENSION QBEG(10),QEND(10),VARE(2000),TPEAK(10),TMAG(10)
      DIMENSION TON(10),TOFF(10),IQBEG(10),IQEND(10)
      DIMENSION FACN(10),ITLOC(100)
      DIMENSION TEP(10),TEM(10)
      DIMENSION IPEAK(10)
      DO 10 I=1,5
      IQBEG(I)=QBEG(I)
      IQEND(I)=QEND(I)
   10 CONTINUE
C
C
      DO 20 I=1,4
      IV1=IQBEG(I)
      IV2=IQBEG(I+1)
      IV4=IV2-1
   20 CONTINUE
      DO 200 K=1,4
      IL=1
      IX=1
      II=0
```

```
      IJ=IQEND(K)+20
      IK=IQEND(K)+150
      DO 100 I=IJ,IK,2
      IF(ABS(VARE(I)).GT. 50.0) GO TO 30
      GO TO 100
   30 IF (ABS(VARE(I+2)).GT. ABS(VARE(I))) GO TO 100
      IF (ABS(VARE(I-2)).GT. ABS(VARE(I))) GO TO 100
      II=II+1
      ITLOC(II)=I
      DO 40 JJ=1,3
      BACK=ABS(VARE(I-JJ))-ABS(VARE(I-JJ-2))
      IF(BACK.LT.0.0) GO TO 100
      FORW=ABS(VARE(I+JJ))-ABS(VARE(I+JJ+2))
      IF (FORW.LT.0.0) GO TO 100
      GO TO 40
   40 CONTINUE
      TEP(IX)=I*1.0
      TEM(IX)=VARE(I)
      IX=IX+1
  100 CONTINUE
      IX=IX-1
      PMAX=0.0
      DO 150 IXX=1,IX
      IF (ABS(TEM(IXX)).GT.PMAX) PMAX=ABS(TEM(IXX))
  150 CONTINUE
      CRIT=PMAX/2.0
  151 FORMAT(' PMAX,CRIT',2F12.3)
      DO 160 IXX=1,IX
      IF (ABS(TEM(IXX)).GT.CRIT) GO TO 170
      GO TO 160
  170 TPEAK(IL)=TEP(IXX)
      TMAG(IL)=TEM(IXX)
```

```
      IL=IL+1
  160 CONTINUE
      IL=IL-1
      WRITE(1,175) (K,TPEAK(I),TMAG(I),I=1,IL)
  175 FORMAT(' TLOC,MAG FOR NO. ',I5,'  SAMPLE',2F15.5)
      IDIFF=TPEAK(I)-IQEND(K)
      DO 300 IZ=1,IDIFF
      IPEAK(I)=TPEAK(I)
      A=ABS(VARE(IPEAK(I)-IZ))
      B=ABS(VARE(IPEAK(I)-IZ-1))
      C=ABS(VARE(IPEAK(I)-IZ-2))
      IF ((A.LT.10.).AND.(B.LT.10.).AND.(C.LT.10.))GO TO 310
      GO TO 300
  310 TON(K)=TPEAK(I)-IZ-2
      GO TO 350
  300 CONTINUE
  350 WRITE(1,185)K,I,TON(K),VARE(TON(K))
  185 FORMAT(1X,'K,I,TON(K)',2I5,2F12.2)
      IAB=TPEAK(I)+10
      IBC=IQBEG(K+1)-10
      DO 450 ICD=IAB,IBC
      DO 400 IBA=1,5
      DIFA=ABS(VARE(ICD-IBA)-VARE(ICD))
      DIFB=ABS(VARE(ICD+IBA)-VARE(ICD))
      IF ((DIFA.GT.10.).OR.(DIFB.GT.10.)) GO TO 440
  400 CONTINUE
      TOFF(K)=ICD*1.0
      WRITE(1,410)K,TOFF(K),VARE(TOFF(K))
  410 FORMAT(' K,TOFF,VARE(TOFF(K))',I10,2F12.3)
      GO TO 500
  440 CONTINUE
  450 CONTINUE
```

```
  500 CONTINUE
  200 CONTINUE
      RETURN
      END
C
C
```

```
C
C
C
C THIS PROGRAM NORMALIZE THE X,Y,Z DATA FOR EACH CARDIAC PERIOD, ONE BY ONE
C USING FIVE POINTS BEFORE THE ONSET OF THE FIRST CARDIAC CYCLE AND THE FIVE
C POINTS OF THE SECOND CARDIAC CYCLE AND REDUCING AS MUCH OF THE RANDOM
C VARIATION AS POSSIBLE.
C
C
C
      SUBROUTINE NORMA(X,Y,Z,QBEG,QEND,FACNX,FACNY,FACNZ)
      DIMENSION X(2500),Y(2500),Z(2500),QEND(10),QBEG(10)
      DIMENSION FACNX(10),FACNY(10),FACNZ(10),IQBEG(10),IQEND(10)
      DO 10 I=1,5
      IQBEG(I)=QBEG(I)
      IQEND(I)=QEND(I)
   10 CONTINUE
      DO 20 I=1,4
      IV1=IQBEG(I)
      IV2=IQBEG(I+1)
      IV4=IV2-1
      XAVG=X(IV1)-X(IV2)
      YAVG=Y(IV1)-Y(IV2)
      ZAVG=Z(IV1)-Z(IV2)
      DO 50 IV5=1,5
      XAVG=XAVG+X(IV1-IV5)-X(IV2-IV5)
      YAVG=YAVG+Y(IV1-IV5)-Y(IV2-IV5)
      ZAVG=ZAVG+Z(IV1-IV5)-Z(IV2-IV5)
   50 CONTINUE
      FACNX(I)=XAVG/6.0
      FACNY(I)=YAVG/6.0
      FACNZ(I)=ZAVG/6.0
```

```
      WRITE(1,25)I,FACNX(I),FACNY(I),FACNZ(I)
   25 FORMAT(' FAC OF X,Y,Z',I10,3F12.2)
   20 CONTINUE
      IIT=IQBEG(1)-1
      XD=(IQBEG(2)-IQBEG(1))*1.0
      DO 90 IV=1,IIT
      XN=(IQBEG(1)-IV)*1.0
      X(IV)=X(IV)-FACNX(1)*XN/XD-X(IQBEG(1))
      Y(IV)=Y(IV)-FACNY(1)*XN/XD-Y(IQBEG(1))
      Z(IV)=Z(IV)-FACNZ(1)*XN/XD-Z(IQBEG(1))
   90 CONTINUE
      DO 60 I=1,4
      IV1=IQBEG(I)
      IV4=IQBEG(I+1)-1
      IV5=IV4+1
      XD=(IQBEG(I+1)-IQBEG(I))*1.0
      DO 60 J=IV1,IV4
      XN=(IV4-J)*1.0
  200 FORMAT(1X,'X(J)',I5,5F12.2)
      X(J)=X(J)-FACNX(I)*XN/XD-X(IV5)
      Y(J)=Y(J)-FACNY(I)*XN/XD-Y(IV5)
      Z(J)=Z(J)-FACNZ(I)*XN/XD-Z(IV5)
   60 CONTINUE
   30 RETURN
      END
C
C
C
```

```
C
C THE FOLLWOING IS PWAVE SUBROUTINE WHICH FINDS THE P PEAK MAGNITUDE
C AND ITS LOCATION AND ALSO THE ONSET AND OFFSET OF THE P WAVE IF
C AVAILABLE AND IDENTIFY THE CONFIGURATION OF THE P WAVE
C
C
      SUBROUTINE PWAVE(IQBEG,IQEND,IPBEG,IPEND,IPEAK,PPEAK,TOFF,PPLOC,
     1VARE)
      DIMENSION IQBEG(10),IQEND(10),IPBEG(10),IPEND(10),TOFF(10)
      DIMENSION IPEAK(10),PPEAK(10),VARE(2000)
      DIMENSION ITOFF(10),PPLOC(10)
      DO 600 IJ=1,2
      IF (IJ.EQ.2) GO TO 50
      IM=4
      IPB=IQBEG(1)-100
      IPE=IQBEG(1)-10
      GO TO 90
   50 DO 500 IM=2,4
      ITOFF(IM-1)=TOFF(IM-1)
      IPB=ITOFF(IM-1)+10
      IPE=IQBEG(IM)-10
   90 DO 400 I=IPB,IPE
      IF (ABS(VARE(I)).GT.50.0) GO TO 100
      GO TO 400
  100 DO 300 K=1,6
      J=K-1
      BACK=ABS(VARE(I-J))-ABS(VARE(I-J-1))
      IF (BACK.GT.0.0) GO TO 200
      GO TO 400
  200 FORW=ABS(VARE(I+J))-ABS(VARE(I+J+1))
      IF (FORW.GT.0.0) GO TO 300
      GO TO 400
```

```
300 CONTINUE
    IF (IJ.EQ.2) GO TO 250
    IA=1
    GO TO 260
250 IA=IM
260 PPEAK(IA)=VARE(I)
    PPLOC(IA)=I*1.0
    WRITE(1,150)PPEAK(IA),PPLOC(IA),IA,IJ
150 FORMAT(1X,'FROM PWAVE,PEAK,LOC',2F12.2,2I5)
400 CONTINUE
    I1=PPLOC(IA)+10
    I2=IQBEG(IA)-15
    DO 450 I3=I1,I2
    DO 440 I4=1,5
    DIFA=ABS(VARE(I3-I4)-VARE(I3))
    DIFB=ABS(VARE(I3+I4)-VARE(I3))
    IF ((DIFA.GT.10.0) .OR. (DIFB.GT.10.0)) GO TO 450
440 CONTINUE
460 IPEND(IA)=I3
    WRITE(1,470)IPEND(IA),IA,VARE(IPEND(IA))
470 FORMAT(1X,'PEND,I,MAG',2I5,F12.2)
    GO TO 490
450 CONTINUE
490 IST=PPLOC(IA)-10
    DO 650 I3=1,80
    JK=IST-I3
    DO 640 LL=1,5
    DIFA=ABS(VARE(JK-LL)-VARE(JK))
    DIFB=ABS(VARE(JK+LL)-VARE(JK))
    IF ((DIFA.GT.10.0) .OR. (DIFB.GT.10.0)) GO TO 650
640 CONTINUE
    IPBEG(IA)=JK
```

```
      WRITE(1,645)IPBEG(IA),IA,VARE(IPBEG(IA))
  645 FORMAT(1X,'PBEG,I,MAG',2I5,F12.2)
      GO TO 500
  650 CONTINUE
  500 CONTINUE
  600 CONTINUE
      RETURN
      END
C
C
C
```

```
C THE VEC SUBROUTINE DOES THE JOB OF FINDING THE SPATIAL AMPLITUDE (M),
C THE AZIMUTH (H), AND THE ELEVATION (V) FOR A GIVEN STARTING POINT AND
C ENDING POINT OF ANY PARTICULAR WAVEFORM.
C THIS SUBROUTINE ALSO FINDS THE VECTOR AMPLITUDE AND VECTOR ANGLE FOR
C FRONTAL, LEFT SAGITTAL AND HORIZONTAL PLANE.
C THE ASSIGNMENT OF THE VARIABLE ARE SHOWN BELOW
C      SPM1=MAXIMUM SPATIAL AMPLITUDE
C      FM1 =MAXIMUM FRONTAL PLANE AMPLITUDE VECTOR
C      SM1 =MAXIMUM SAGITTAL PLANE AMPLITUDE VECTOR
C      HM1 =MAXIMUM HORIZONTAL PLANE MAGNITUDE VECTOR
C      SAH =THE AZIMUTH(H) OF THE MAXIMUM VECTOR(SPM1)
C      SAV =THE ELEVATION(V) OF THE MAXIMUM VECTOR(SPM1)
C      FA1 =THE ANGLE OF THE MAXIMUM FRONTAL PLANE VECTOR(FM1)
C      SA1 =THE ANGLE OF THE MAXIMUM SAGITTAL PLANE VECTOR(SM1)
C      HA1 =THE ANGLE OF THE MAXIMUM HORIZONTAL PLANE VECTOR(HM1)
C
C
C
      SUBROUTINE VECT(M,H,V,ST,EN,X,Y,Z)
      REAL MMAX
      DIMENSION X(3000),Y(3000),Z(3000)
      DIMENSION XM(500),H(500),V(500)
      DIMENSION FXY(500),FANG(500)
      DIMENSION SYZ(500),SANG(500)
      DIMENSION HXZ(500),HANG(500)
      MMAX=0.0
      FMAX=0.0
      SMAX=0.0
      HMAX=0.0
      J=0
      K=ST
      L=EN
```

```
      WRITE(4,15)
   15 FORMAT(10X,'SP VECTOR',3X,'HOR. ANG',4X,'VERT.ANG',3X,'FRONT VECT'
     X,2X,'FRONT ANG')
      DO 10 I=K,L
      J=J+1
      XM(J)=SQRT(X(I)**2+Y(I)**2+Z(I)**2)
      CALL SETT(Z(I),X(I),ANG)
      H(J)=ATAN(Z(I)/X(I))*360.0/2.0/3.1416+ANG
      V(J)=ATAN(Y(I)/(SQRT(X(I)**2+Z(I)**2)))*360./2./3.1416
      IF(XM(J).LT.MMAX) GO TO 50
      MMAX=XM(J)
      JMAX=J
   50 FXY(J)=SQRT(X(I)**2+Y(I)**2)
      CALL SETT(Y(I),X(I),ANG)
      FANG(J)=ATAN(Y(I)/X(I))*360.0/2.0/3.1416+ANG
      IF (FXY(J).LT.FMAX) GO TO 55
      FMAX=FXY(J)
      IFMAX=J
   55 SYZ(J)=SQRT(Y(I)**2+Z(I)**2)
      CALL SETT(Y(I),Z(I),ANG)
      SANG(J)=ATAN(Y(I)/Z(I))*360.0/2./3.1416+ANG
      IF (SYZ(J).LT.SMAX) GO TO 60
      SMAX=SYZ(J)
      ISMAX=J
   60 HXZ(J)=SQRT(X(I)**2+Z(I)**2)
      CALL SETT(Z(I),X(I),ANG)
      HANG(J)=ATAN(Z(I)/X(I))*360.0/2./3.1416+ANG
      IF(HXZ(J).LT.HMAX) GO TO 70
      HMAX=HXZ(J)
      IHMAX=J
   70 WRITE(4,20)XM(J),H(J),V(J),FXY(J),FANG(J)
   20 FORMAT(5X,5F12.3)
```

```
   10 CONTINUE
      WRITE(4,25)
   25 FORMAT(10X,'SP MAX',6X,'HOR ANG',5X,'VERT. ANG',2X,'FRONT MAX  MAX
     XFRONT ANG')
      WRITE(4,30)XM(JMAX),H(JMAX),V(JMAX),FXY(IFMAX),FANG(IFMAX)
   30 FORMAT(5X,5F12.3)
      SPM1=MMAX
      SAH=H(JMAX)
      SAV=V(JMAX)
      FM1=FMAX
      FA1=FANG(IFMAX)
      SM1=SMAX
      SA1=SANG(ISMAX)
      HM1=HMAX
      HA1=HANG(IHMAX)
      WRITE(4,100)
  100 FORMAT(22X,'SPATIAL')
      WRITE(4,110)
  110 FORMAT(19X,'H',12X,'V',7X,'FRONTAL',5X,'SAGITTAL',4X,
     C'HORIZONTAL')
      WRITE(4,120) SPM1,FM1,SM1,HM1
  120 FORMAT(' MAGNITUDE',10X,F12.3,4X,3F12.3)
      WRITE(4,150)SAH,SAV,FA1,SA1,HA1
  150 FORMAT(' ANGLE',6X,5F12.3)
      RETURN
      END
C
C
C
```

```
C THIS SUBROUTINE SETT IS USED TO DETERMINE WHICH QUADRANT OF THE ANGLE
C DOES THE VECTOR LIES WITH GIVEN OPPOSITE SIDE OVER ADJACENT SIDE FOR
C FINDING INVERSE OF TANGENT.
C
C
C
      SUBROUTINE SETT(TNUM,TDEM,ANG)
      IF (TNUM.GT.0.0 .AND. TDEM.GT.0.0) ANG=0.0
      IF (TNUM.GT.0.0 .AND. TDEM.LT.0.0) ANG=180.0
      IF (TNUM.LT.0.0 .AND. TDEM.LT.0.0) ANG=-180.0
      IF (TNUM.LT.0.0 .AND. TDEM.GT.0.0) ANG=0.0
      RETURN
      END
C
C
C
C
C
```

```
C THE CONV SUBROUTINE CONVERTS THE THREE DIMENSIONAL (X,Y,Z) ARRAYS INTO
C TWO DIMENSIONAL ARRAY FOR PLOTTING ON THE GRAPHIC TERRMINAL. THIS ALS
C PROVIDES A CHOICE FOR THE ANGLE OF VISION AS SPECIFIED IN THE PARAMETE
C VX,VY,VZ,THE VISION OF THE AXIS X,Y,Z.
C
C
      SUBROUTINE CONV(X,Y,Z,VX,VY,VZ,IS,IE)
      DIMENSION C(4,4),AR(4),D(4),PARX(500),PARY(500)
      DIMENSION X(3000),Y(3000),Z(3000)
      A1=SQRT(VX**2+VY**2)
       B1=SQRT(VX**2+VY**2+VZ**2)
      DO 10 I=1,4
      DO 10 J=1,4
      C(I,J)=0.0
   10 CONTINUE
      C(1,1)=-VY/A1
      C(1,2)=-VX*VZ/A1/B1
      C(1,3)=-VX/B1
      C(2,1)=-VX/A1
      C(2,2)=-VY*VZ/A1/B1
      C(2,3)=-VY/B1
      C(3,2)=A1/B1
      C(3,3)=-VZ/B1
      C(4,3)=B1
      C(4,4)=1.0
C
C
C
      L=0
      AR(4)=1.0
      DO 50 K=IS,IE
      L=L+1
```

```
      AR(1)=-Z(K)
      AR(2)=X(K)
      AR(3)=-Y(K)
      DO 20 J=1,4
      D(J)=0.0
      DO 20 I=1,4
      D(J)=D(J)+AR(I)*C(I,J)
   20 CONTINUE
      PARX(L)=D(1)/D(3)
      PARY(L)=D(2)/D(3)
      WRITE (4,30)PARX(L),PARY(L),X(K),Y(K),Z(K)
   30 FORMAT(1X,'ARR ',5F12.2)
   50 CONTINUE
      RETURN
      END
C
C
C
```

```
C THE FOLLOWING SUBROUTINE VCPOT TAKES THE QRS, P AND T WAVE ONSET AND
C OFFSETS AND PROVIDE DATA APPROPIATE DATA FILE FOR VECTOR PLOT IN A TWO
C DIMENSIONAL PLANE.
C
C
      SUBROUTINE VCPOT(X,Y,Z,IQS,IQE,IPS,IPE,TS,TE)
      DIMENSION X(3000),Y(3000),Z(3000)
      DIMENSION A(5),B(5)
      DO 11 I=1,5
      A(I)=0.0
      B(I)=0.0
   11 CONTINUE
      B(1)=100.0
      B(2)=-100.0
      A(4)=100.0
      A(5)=-100.0
      IPQ=0
      IPP=0
      IPT=0
      ITS=TS*1.
      ITE=TE*1.
      DO 10 I=IQS,IQE
      YY=-Y(I)
      WRITE(14,20)X(I),YY
   20 FORMAT(5X,2F12.3)
      IPQ=IPQ+1
   10 CONTINUE
      DO 30 I=IPS,IPE
      YY=-Y(I)
      WRITE(14,20)X(I),YY
      IPP=IPP+1
   30 CONTINUE
```

```
      DO 40 I=ITS,ITE
      YY=-Y(I)
      WRITE(14,20)X(I),YY
      IPT=IPT+1
   40 CONTINUE
      WRITE(16,50)IPQ,IPP,IPT
   50 FORMAT(5X,3I10)
      DO 60 I=1,5
      WRITE(14,20)A(I),B(I)
   60 CONTINUE
      RETURN
      END
C
C
C
C
```

```
C THE FOLLOWING PROGRAM IS A GENERAL PURPOSE PLOTTING PROGRAM FOR PLOTTING
C THE DATA OBTAINED IN THE ALGORITHM ABOVE.  THIS IS AN INTERACTIVE GRAPHIC
C PACKAGE.  INSIDE THIS ALGORITHM, THE MLTPLT SUBROUTINE PORTION IS WRITTEN BY
C STEVE ALIFF AT V.P.I. & S.U.
C
C
C THE FOLLOWING INTERACTIVE GRAPHIC PACKAGE PROVIDES A EFFICIENT AND EFFECTIVE
C MEANS TO STUDY AND EXAMINE THE ECG SIGNAL AND ITS VECTORS AND VECTOR LOOPS AS
C OBTAINED FROM THE ABOVE ALGORITHM.
C
C
      DIMENSION X(500), Y(500), TITLE(20), FORMIN(18), NOPTS(5), XLAB(
     1, YLAB(3)
      REAL*8 LEGEND(5)
      CALL NEWPAG
      CALL SYSCAL (14HCP SLEEP 1 SEC,14,IRC)
      WRITE (6,80)
      READ (5,90) TITLE
      WRITE (6,100)
      READ (5,110) XLAB
      WRITE (6,120)
      READ (5,110) YLAB
      WRITE (6,130)
      READ (5,*) XAXL,YAXL
      WRITE (6,140)
      READ (5,150) FORMIN
      WRITE (6,160)
      READ (5,*) NOCURV
      WRITE (6,170)
      WRITE (6,180)
      WRITE (6,190)
      WRITE (6,200)
```

```
      READ (5,*) I
      GO TO (10,20,30), I
10    IOPT=-4
      GO TO 40
20    IOPT=0
      GO TO 40
30    IOPT=4
40    CONTINUE
      WRITE (6,240)
      READ (5,*) I
      GO TO (42,44,46), I
42    ILOG=-1
      GO TO 48
44    ILOG=0
      GO TO 48
46    ILOG=1
48    CONTINUE
      DO 60 I=1,NOCURV
      IF (IOPT.EQ.0) GO TO 50
      WRITE (6,210) I
      READ (5,220) LEGEND(I)
50    CONTINUE
      WRITE (6,230) I
      READ (5,*) NOPTS(I)
60    CONTINUE
      DO 70 I=1,500
      READ (1,FORMIN,END=75) X(I),Y(I)
      WRITE (2,*) X(I),Y(I)
      IF (ILOG.GE.0) Y(I)=ALOG(Y(I))
      IF (ILOG.GT.0) X(I)=ALOG(X(I))
70    CONTINUE
75    CONTINUE
```

```
      CALL NEWPAG
      CALL SYSCAL (14HCP SLEEP 1 SEC,14,IRC)
      WRITE (6,250)
      READ (5,*) I
      GO TO (76,77,78), I
76    FAC=0.0
      CALL NEWPAG
      CALL SYSCAL (14HCP SLEEP 1 SEC,14,IRC)
      GO TO 79
77    FAC=-1.0
      GO TO 79
78    FAC=1.0
      CALL NEWPAG
      CALL SYSCAL (14HCP SLEEP 1 SEC,14,IRC)
79    CONTINUE
      CALL HDCPY (FAC)
      CALL MLTPLT (X,Y,XLAB,YLAB,3,3,NOPTS,IOPT,0,NOCURV,XAXL,YAXL,TITLE
     1,LEGEND,80)
      CALL PLOT (0.0,0.0,-4)
      STOP
C
80    FORMAT (34H * ENTER TITLE (80 CHARS. MAX.) - )
90    FORMAT (20A4)
100   FORMAT (41H * ENTER X-AXIS LABEL (12 CHARS. MAX.) - )
110   FORMAT (3A4)
120   FORMAT (41H * ENTER Y-AXIS LABEL (12 CHARS. MAX.) - )
130   FORMAT (39H * ENTER X & Y AXIS LENGTHS (INCHES) - )
140   FORMAT (49H * ENTER FORMAT OF DATA, INCLUDING PARENTHESES - )
150   FORMAT (18A4)
160   FORMAT (35H * ENTER NUMBER OF CURVES (<= 5) - )
170   FORMAT (16H 1 - POINT PLOT,)
180   FORMAT (15H 2 - LINE PLOT,)
```

```
190    FORMAT (10H 3 - BOTH.)
200    FORMAT (22H * ENTER 1, 2, OR 3 - )
210    FORMAT (42H * ENTER LEGEND (8 CHARS. MAX.) FOR CURVE ,I2)
220    FORMAT (A8)
230    FORMAT (30H * ENTER NO. OF PTS. IN CURVE ,I2)
240    FORMAT(' 1 - ARITHMETIC PLOT,'/' 2 - SEMI-LOG PLOT,'/
      1' 3 - LOG-LOG PLOT.'/' * ENTER 1, 2, OR 3 - ')
250    FORMAT(' 1 - PLOT ON SCREEN,'/' 2 - PLOT TO FILE,'/
      1' 3 - BOTH.'/' * ENTER 1, 2, OR 3 - ')
       END
C
```

```
C      - SUBROUTINE MLTPLT -
C
C         PLOTS UP TO 4 CURVES ON ONE GRAPH. PATTERNED AFTER EZPLOT
C      BY VPI COMPUTING CENTER.
C
C      USES SUBROUTINES -
C         1. SCALE (VPI)
C         2. PLOT (VPI)
C         3. AXIS (VPI)
C         4. SLINE (VPI)
C         5. SYMBOL (VPI)
C
C      *** WRITTEN ON 9/14/77 BY STEVE ALIFF FOR THE AOE DEPT.
C
C      ------------------------------------------------------------------
       SUBROUTINE MLTPLT (X,Y,LABELX,LABELY,IXN,IYN,NXP,IOPT,LSD,NPLTS,XA
      1LNTH,YALNTH,TITLE,LEGEND,ITCH)
C      ------------------------------------------------------------------
C
C      X,Y  -  THE ARRAYS OF X & Y VALUES (COORDINATES)
C              (MAXIMUM OF 500 SINGLE PRECISION VALUES, 500 DOUBLE)
C
C      LABELX,LABELY  -  THE X & Y AXIS LABELS (A4 HOLLERITH)
C
C      IXN,IYN  -  THE # TIMES A4 DIVIDES LABELX & LABELY RESPECTIVELY
C
C      NXP(I)  -  THE # POINTS IN CURVE I
C
C      IOPT  -  0 < IOPT <= 25  ->  LINE & POINT PLOT
C               -25 <= IOPT < 0  ->  POINT PLOT
C               =0  ->  LINE PLOT
C      (THE VALUE OF IOPT DETERMINES THE PLOT SYMBOL FOR THE FIRST
```

```
C      PLOT. < E.G. -11 -> ASTERISK >  IOPT MUST ALLOW FOR IOPT =
C      IOPT + NPLTS <= 25 )
C
C      LSD  -  AN INTEGER; LSD=0  ->  SINGLE PRECISION X & Y
C                          LSD>0  ->  DOUBLE PRECISION X & Y
C
C      NPLTS  -  # CURVES ON THIS PLOT (<=4)
C
C      XALNTH,YALNTH  -  X & Y AXIS LENGTHS
C
C      LEGEND  -  LEGEND FOR EACH PLOT (A8 HOLLERITH)
C
C      TITLE  -  TITLE FOR THE PLOT (A4 HOLLERITH)
C
C      ITCH  -  THE NUMBER OF CHARACTERS IN THE TITLE
C
C      ------------------------------------------------------------------
       DIMENSION X(500), Y(500), LABELX(IXN), LABELY(IYN), TX(500), TY
      10), TITLE(ITCH), NXP(NPLTS)
       REAL*8 LEGEND(NPLTS)
C      ------------------------------------------------------------------
C      SET PRECISION OPTION
C      ------------------------------------------------------------------
       K=1
       IF (LSD) 10,30,20
10     WRITE (6,80)
20     K=2
C      ------------------------------------------------------------------
C      SCALE X & Y ARRAYS TO INCHES
C      ------------------------------------------------------------------
30     CONTINUE
       NSUM=0
```

```
      DO 40 J=1,NPLTS
40    NSUM=NSUM+NXP(J)
      CALL SCALE (X,NSUM,XALNTH,XMIN,DX,K)
      CALL SCALE (Y,NSUM,YALNTH,YMIN,DY,K)
      KXN=-IXN
      L=KXN*4
      M=IYN*4
C     ------------------------------------------------------------------
C     SET THE ORIGIN AND PLOT THE AXIS
C     ------------------------------------------------------------------
      CALL PLOT (1.0,1.0,-3)
      CALL AXIS (0.0,0.0,LABELX,L,XALNTH,0.0,XMIN,DX)
      CALL AXIS (0.0,0.0,LABELY,M,YALNTH,90.0,YMIN,DY)
C     ------------------------------------------------------------------
C     CLOSE THE 'BOX'
C     ------------------------------------------------------------------
      CALL PLOT (XALNTH,0.0,3)
      CALL PLOT (XALNTH,YALNTH,2)
      CALL PLOT (0.0,YALNTH,2)
C     ------------------------------------------------------------------
C     PLOT THE CURVES
C     ------------------------------------------------------------------
      JOPT=IABS(IOPT)
      IF (JOPT.GT.25) WRITE (6,90)
      IF (JOPT.GT.25) JOPT=11
      INDEX=0
      DO 60 N=1,NPLTS
      ISTOP=NXP(N)
      DO 50 I=1,ISTOP
      INDEX=INDEX+1
      TX(I)=X(INDEX)
      TY(I)=Y(INDEX)
```

```
50     CONTINUE
       CALL SLINE (TX,TY,ISTOP,K,IOPT,JOPT)
       JOPT=JOPT+1
60     CONTINUE
C      ------------------------------------------------------------------
C      PLOT THE LEGEND (FOR A PLOT WITH POINTS )
C      ------------------------------------------------------------------
       IF (IOPT.EQ.0) GO TO 70
       JOPT=IABS(IOPT)
       IF (JOPT.GT.25) JOPT=11
       DO 70 I=1,NPLTS
       CALL SLINE (XALNTH+0.5,YALNTH-0.5*FLOAT(I),1,1,-1,JOPT)
       CALL SYMBOL (XALNTH+1.0,YALNTH-0.5*FLOAT(I),0.13,LEGEND(I),0.0,8)
       JOPT=JOPT+1
70     CONTINUE
C      ------------------------------------------------------------------
C      PLOT THE TITLE
C      ------------------------------------------------------------------
       CALL SYMBOL (0.5,YALNTH+0.5,0.13,TITLE,0.0,ITCH)
C      ------------------------------------------------------------------
C      MOVE THE ORIGIN TO AVOID OVERPLOTTING AND RETURN
C      ------------------------------------------------------------------
       CALL PLOT (XALNTH+4.0,-1.0,-3)
       RETURN
C
80     FORMAT (1H0,66H*** WARNING ***-NEGATIVE VALUE FOR PRECISION OPTION
      1   ENCOUNTERED /30HASSUMED TO BE DOUBLE PRECISION)
90     FORMAT (1H0,49H*** WARNING ***-VALUE FOR IOPT IS GREATER THAN 25/1
     18H  VALUE 11 ASSUMED)
       END
C**********END**********C
C**********END**********C
```

```
C**********END**********C
C**********END**********C
```

********** END OF APPENDIX **********

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COMPUTER-AIDED ELECTROCARDIOGRAM ANALYSIS

by

Peter H. Pan

(Abstract)

An electrocardiogram analysis package is developed in this thesis. A parabolic smoothing with pre-set threshold for attenuating effects on high frequency components is applied. The GF and operational transforms are studied and examined for their constraints, applications and flexibility. These transforms are modified and adapted for recognition of QRS onset and offset in the Frank Orthogonal Lead System. An algorithm, based on a set of pre-set criteria and thresholds, is developed. Various clinical parameters, such as segment or wave intervals, magnitude, vectors and areas etc. are generated together with data files for two-dimensional and three-dimensional graphic analysis. In addition, a general purpose graphic package is supplied to create the vector displays for ECG analysis. The algorithm has been tested on three hundred cardiac cycles taken from ten patients with both normal and abnormal ECG characteristics.