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To the Graduate Council:

I am submitting herewith a thesis written by Martha Carol Snyder Hansard entitled "Investigation of the Applicability of a Digital Filtering Technique for the Feature Enhancement of Electrocardiogram Waveforms." I recommend that it be accepted for nine quarter hours of credit in partial fulfillment of the requirements for the degree of Master of Science, with a major in Computer Science.

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INVESTIGATION OF THE APPLICABILITY OF A DIGITAL FILTERING TECHNIQUE FOR THE FEATURE ENHANCEMENT OF ELECTROCARDIOGRAM WAVEFORMS

A Thesis

Presented to

the Graduate Council of

The University of Tennessee

In Partial Fulfillment

of the Requirements for the Degree

Master of Science

---

by

Martha Carol Snyder Hansard

March 1973

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ABSTRACT

Computers have been established as useful aids in electrocardiogram (ECG) analysis. However, full potential has not been reached and research is still needed to make computer ECG analysis as accurate as the cardiologist's. Also, computerized analysis must be fast enough to enable real-time response to changing events in an ECG.

This research investigates the use of a digital filtering technique as a feature extractor for a pattern recognition system to detect points used for measurements in ECG analysis. The point of onset of the P-wave was used as a test of the method. The P-wave is important in ECG analysis because it signifies the firing of the heart's pacemaker and accurate detection is instrumental in measuring the true rhythm of the heart's electrical system. Detection of the P-wave onset point was used as a test of the method because even cardiologists often disagree as to the exact point and no direct means of computerized detection exists at present.

The conclusion of the research is that the digital filtering technique is an applicable feature extractor for the detection of ECG waveforms. However, the findings were that the filtering frequencies must be set for each individual case and normalizing factors peculiar to each ECG must be used to optimize the filtering frequencies for wave enhancement. The normalizing factor used in this research was the R-R interval. Further research is needed to "fine tune" the filters and define the exact normalizing factors necessary for feature extraction in

all cases. In implementing the technique, initial information about the normalizing factors would be fed into the system by either a technician or another simple computer program to set the filtering frequencies. As the filtering frequencies are set to enhance certain features, the information from these filter outputs themselves would be used to further tune the filters to gather other information.

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CHAPTER I

INTRODUCTION AND PROBLEM DEFINITION

I. INTRODUCTION

Electrocardiograms (ECGs) have been used for over half a century to record the heart's electrical activity. Since the ECG was first detected and recorded, a vast amount of knowledge has been empirically obtained to correlate the detected electrical activity with the heart's various stages in performance of its functions in the body. Even greater amounts of knowledge have been and are being made available by the use of the digital computer in electrocardiography.

II. BACKGROUND

It has long been known that the messages used by the human body to coordinate activities are of an electrical nature (1). Use of this knowledge awaited the development of measuring devices. In the early 1900's it became possible to measure the electrical potential difference across the heart and the art of electrocardiography was born. The deverlopment of instrumentation has not traveled a smooth road. Many of the problems encountered involved the question of precision versus practicality. Also, there were significant differences among machines in the produced waveforms. Many of the problems have been solved by new technology, especially the advent of advanced electronic techniques.

Some of the subjective problems are still open to discussion. Development of instrumentation has progressed to such a degree that a great deal of difference exists between the instruments used in the early 1900's and those now in use in the 1970's. Reliability and repeatability among today's machines are at an advanced stage of development.

The ECG is a time plot of the voltage difference measured between two points on the surface of the body (2). The voltage difference is present because of the electrical activity in the body which may be considered a volume conductor. Although the total potential difference is present and measured, cardiologists have empirically related the ECG to the difference of potential across the heart.

The physician and researcher's role in the development of electrocardiography has by no means been small. Physicians have recognized and accepted the ECG as the best means of examining the heart short of surgery. Examination of the heart using the ECG is indirect; but direct examination is too dangerous for the patient. Therefore, the medical profession has been concerned with circumventing the indirectness of the measurement which the ECG affords. The ECG would be valueless to the physician if he could not detect the heart's normalities and abnormalities from it. An empirical approach has been taken; and today a large amount of knowledge has been acquired about how the ECG portrays the heart (3).

The heart performs its functions by a series of rhythmic contractions and relaxations (4). The electrical activity in the heart signals these functions. The ECG measures the change in potential difference

produced by electrical currents of depolarization and repolarization in the heart tissue. Depolarization immediately precedes contraction and repolarization immediately precedes relaxation.

A typical ECG is shown in Figure 1. The ECG may be divided into three main parts--the P-wave, the QRS-complex and the T-wave (2, 5,6). These three parts are interactive and are by no means exhaustive of the divisions which can be made by the cardiologist.

A simplified diagram of the heart's electrical tissue is shown in Figure 2. The electrical activity of the heart and the ECG are related in the following manner. Electrical activity begins at the sinoatrial (SA) node, also known as the pacemaker (2,6). The current spreads through the muscle of the atria to the atrioventricular (AV) node. This electrical activity, initiated by the depolarization of the SA node, produces contractions of the muscles surrounding the atria and is represented in the ECG by the P-wave. Repolarization of the atria follows depolarization; however, the repolarization is masked by the QRS-complex and is rarely manifested in the ECG. The AV node initiates the depolarization of the bundle of His. Depolarization travels through the bundles of His and the Purkinje tissue resulting in the contraction of the ventricular muscles and the QRS-complex in the ECG. Relaxation of the ventricular muscles is obtained upon repolarization which produces the T-wave in the ECG.

The cardiologist is able to relate this representation of the electrical activity to the functions of the heart. There is a great deal of variation even among ECG's in normal hearts. Typical sources

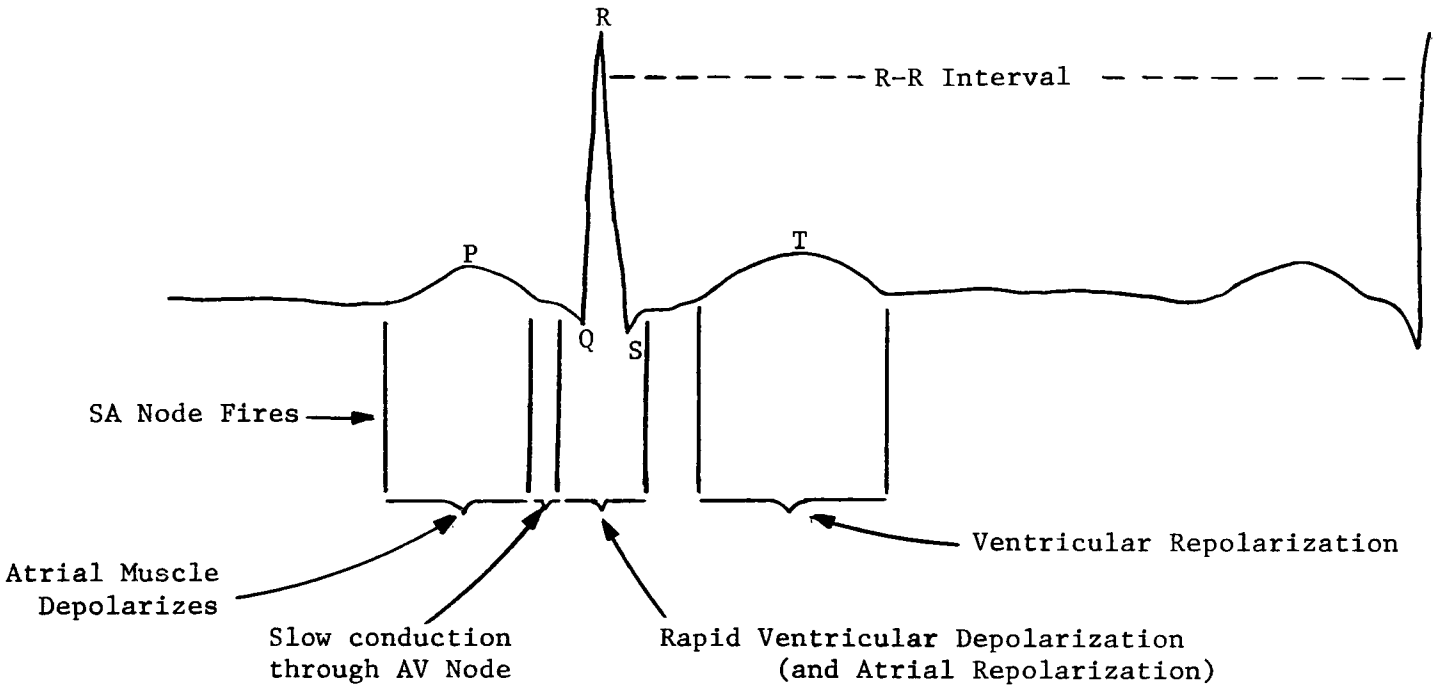


Figure 1. Major divisions of a typical ECG.

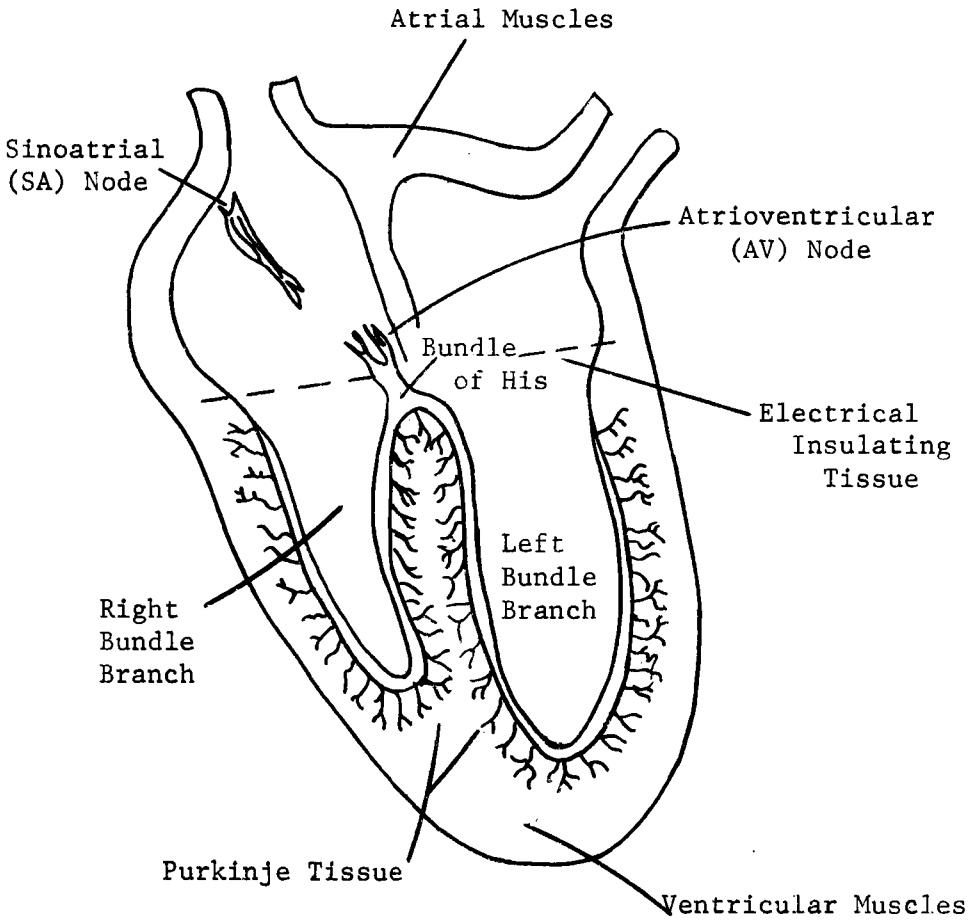


Figure 2. Simplified diagram of the heart's electrical system.

of variation are due to body build, position of the heart, and non-homogeneous tissue. Abnormalities in heart functions usually manifest themselves in the ECG as aberations in rhythm or shape and contour of specific waves (5).

Cardiologists may use as many as twelve leads which record the potential difference at different angles across the heart. Recently the Frank lead system has been shown to contain as much information as the classical lead system (7). This system uses three orthogonal leads and allows for some compensation of differences in individuals. Reference is made to Figure 3. The potential generated within the heart is attenuated by different amounts in the X, Y, and Z directions; and, because of torso inhomogeneity, these signals are not generated along axes which are exactly 90° to one another (2). The Frank lead system compensates for the attenuation and inhomogeneities. Another reason for the use of this lead system is that the leads are recorded simultaneously so that significant events are recorded in all leads at the same time.

ECG analysis may be divided into three parts: (1) wave detection and identification, (2) rhythm analysis, and (3) contour analysis (5). The first part involves the detection and identification of the various waveforms. Normally the QRS-complex is identified first because it dominates the total ECG. The second and third parts of analysis are dependent upon the first because they involve rhythm and wave contour analysis. Rhythm analysis concerns not only the heart rate but also the time and ordering relationships among the various waves. Contour

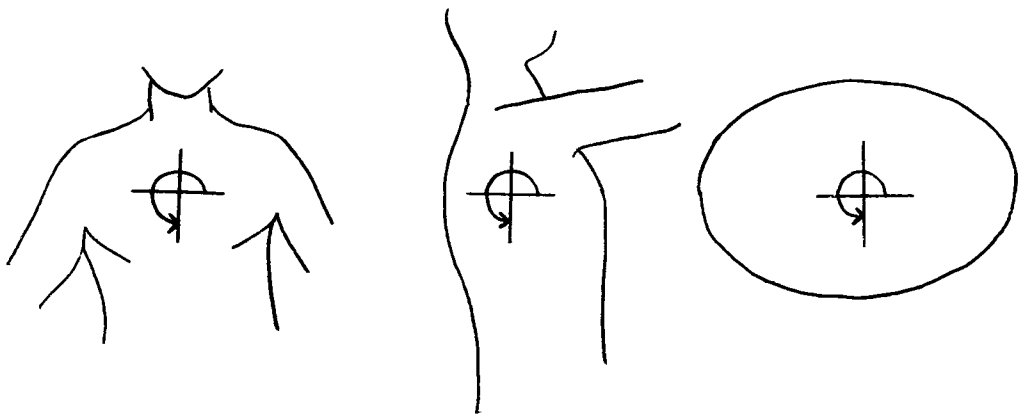


Figure 3. Three orthogonal (Frank) lead system.

analysis deals with the shapes of the various waves.

A cardiologist generally spends only a small amount of time in analyzing a normal ECG (5). However, many hours may be required in the analysis of an abnormal ECG. With scanning of the total population for early detection of heart disease as a goal of many physicians, the cardiologist needs many of his tasks relieved. Computers are the logical aid to the over-worked cardiologist. However, computational techniques are required to be as accurate as the cardiologist.

III. USE OF COMPUTERS

One of the first uses of computers in biomedical research was in the field of electrocardiography (1). Computers have been used successfully for data acquisition and filing. This has been accomplished by digitizing the ECG and storing the data for easy retrieval. However, the greatest potential lies in the use of the computer as a diagnostic aid. Diagnosis consists of analyzing the ECG and then relating the analysis to the condition of the person under study. The primary role of the computer has been in the realm of analysis with the final diagnosis being made by the cardiologist. There are currently six major efforts for computerized ECG analysis (8).

Dr. Cesar A. Caceres and his associates at the Medical Systems Development Laboratory in Washington, D. C. are probably the most experienced group in computerized ECG analysis techniques. Their work has been done using a Control Data Corporation (CDC) 160A, with some buffering from a Digital Equipment Corporation (DEC) PDP-8. Their program

was originally written in assembler language but several groups have rewritten the program in FORTRAN and a newer and improved version, "Version D," is now available from the Health Care Technology Division of the National Center for Health Services Research and Development. In February 1970, the Medical System Laboratory was closed and the work was transferred to the Health Care Technology Division. The program uses the classical twelve lead system requiring 3.72 seconds of each channel and data may be accepted from single or three channel equipment. The sampling rate is 500 samples per second.

Dr. Ralph E. Smith at the Mayo Clinic in Rochester, Minnesota has developed a well known program for practical ECG analysis. The program initially was written for the International Business Corporation Machine (IBM) 7090 but later the IBM 1800 was used. The program, written in FORTRAN, uses the three lead Frank system and is available at no cost from Dr. Smith at the Mayo Clinic. The sampling rate used is 250 samples per second.

Dr. Leon Pordy at the Mount Sianai Hospital of New York has developed a program for the IBM 1800 and the IBM 360 series. The program uses ten seconds of Frank leads and five seconds of the classical twelve lead system sampled at 250 samples per second.

Dr. Hubert V. Piperger at the Veterans Hospital in Washington, D. C. has written a program in machine language for the CDC 3200. His program uses the Frank lead system sampled at 100 samples per second and has a different approach from the other programs written to date.

Dr. Homer R. Warner at the Latter Day Saints Hospital in Salt Lake City, Utah, has worked with remodeling and restructuring parts of other ECG programs for use in the Medlab system on the CDC 3000 series computers. The Medlab system is used on-line for patient care. The ECG analysis part of the system has been limited by available core space due to demands for core by other tasks of the system. ECG analysis has developed to give an immediate response to the operator/technician/physician. The program uses the Frank lead system sampled at 200 samples per second.

Dr. Ovar Arvedson at the University of Umea in Umea, Sweden, has contributed the newest program to ECG analysis. His program has been written for the CDC 3200, but he is also working with IBM for development of the program on the 360 series of machines. The program requires at least a 32K machine and Dr. Arvedson has developed special input devices. The program uses the Frank lead system sampled at 200 samples/second and Marquette Electronics is working with Dr. Arvedson to bring the program to the United States in the near future.

A quote from Dr. Caceres seems to best describe the state of development of computers in ECG analysis:

> The multiplicity of approaches and the fact that we still do not have an economic working system tells us that we are still in the research and development phases in the area (9).

Indeed, the computer has not reached the stand alone stage and human supervision is still required. Computers have achieved a permanent place as an aid in ECG analysis but research must still be done to reach the full potential. Some adventurous and exciting attempts have

even been made in using the computer for complete diagnosis (3). Dr. Caceres goes on to state that the simulation of human logic has been the approach taken thus far and that mathematical and statistical analytic techniques might be more fruitful.

IV. OBJECTIVES

The objective of this research was to measure the effectiveness of a new method for the detection and identification of waveforms in the ECG. Detection of the P-wave was the primary goal. The P-wave was chosen for this research because there has been no reliable means of detecting the onset of the P-wave and even cardiologists often disagree over the exact point. P-wave detection has been accomplished by an indirect approach which finds other waves and then looks for a probable P-wave location. The approach developed here enables direct detection of the P-wave onset. The onset of the P-wave is very important because it depicts the true rhythm of the heart by signifying firing of the SA node, the heart's pacemaker (10).

Timing of the electrical signals from the SA node to the bundle of His is used in detecting blockages in the heart's electrical system. Detection of the onset of the P-wave is the first half of timing the interval from the SA node to the bundle of His. The second half, detection of the firing of the bundle of His has only been accomplished by intra-heart electrodes (4). The method tested in this research for detecting the onset of the P-wave is not restrictive and might work successfully for detecting the firing of the bundle of His. Therefore,

direct detection of the P-wave onset is important for two reasons. It gives the P-P interval and it is the first half of detecting blockages in the total electrical system from the SA node to the bundle of His.

Furthermore, this research has general importance in furthering the development of computerized ECG analysis. The method which is measured would greatly increase the speed of accurately detecting and identifying waveforms. In addition, the method could be implemented on mini-computers and take advantage of their fast response times and low costs. The method would be well suited for intensive care monitoring and scanning of a large population. Finally, since many biomedical signals are similar to ECG signals (4), the same method used for analyzing ECG signals may be applicable for use in analysis of other biomedical data.

V. DESCRIPTION OF THE METHOD

The method investigated in this research is a critical part of a pattern recognition system. The pattern recognition system would distinguish points of interest in the ECG signal such as the P-wave onset point from the other points in the signal. There are three aspects of pattern recognition--measurement, feature extraction, and classification (12). Measurement has been fairly well standardized in the present ECG machine (1). If the feature extraction part works properly, the classification part may be simple and fast. Specifically, a linear classifier was used. The research reported in this thesis deals with the feature extraction aspect of the pattern recognition system.

A filter may be defined as a device or material for suppressing or minimizing waves or oscillations of certain frequencies. In effect, a filter may be noted for the frequencies it passes through or the frequencies it screens out. Filters have been used in obtaining ECGs to screen out noise (1,3). The type of filter used in this research was designed to permit certain frequencies to pass through the filter and to attenuate other frequencies, i.e., a band pass filter. The intent was to enhance certain frequencies in the ECG signal. The hypothesis was that proper filtering of the ECG signal would enhance certain frequencies which could be associated with particular waves in the ECG. Hence, the filtering process would be a logical feature extractor for a pattern recognition system.

The filtering technique used in this research is based on the theory of functional approximation (13). An unknown function, the ECG, is theoretically approximated over a finite interval by an infinite series using the Generalized Fourier Technique. If the ECG signal is represented by the function F(t), the function may be represented as

$$F(t) = \sum_{i=1}^{\infty} b_i \Phi_i(t)$$

where $\Phi_i(t)$ is a set of orthogonal functions and b_i are coefficients which describe the function over the interval of definition. The b_i used are the magnitudes of the filtered signal at time t.

The orthogonal functions used were defined on the interval (-aT,aT) as

$$A(N,t) = \frac{1}{\sqrt{T}} \begin{cases} 2^{N/2} & \text{for } -aT < t < t_o \\ -2^{N/2} & \text{for } t_o \leq t < aT \\ 0 & \text{elsewhere} \end{cases}$$

$$B(N,t) = \frac{1}{\sqrt{T}} \begin{cases} 2^{N/2} & \text{for } -aT \leq t \leq -aT/2 \\ -2^{N/2} & \text{for } -aT/2 \leq t < aT/2 \\ 2^{N/2} & \text{for } aT/2 \leq t < aT \\ 0 & \text{elsewhere} \end{cases}$$

where $a = 2^{-N-1}$ and $N = 0, 1, 2 \ldots$ is defined to be the order of the family of orthogonal functions.

The transfer functions for A(N,t) and B(N,t) in the frequency domain are, respectively

$$A(jw) = \frac{-2j}{w\sqrt{T}} \left[\cos(wT/2) - 1\right]$$

and

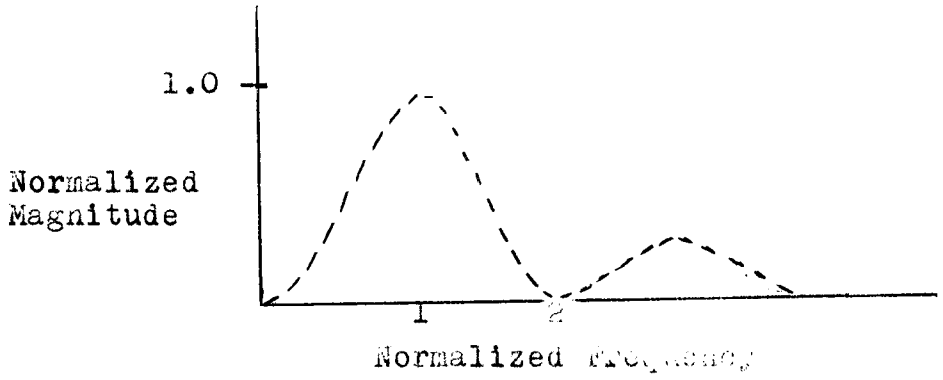
$$B(jw) = \frac{2}{w\sqrt{T}} \left[\sin(wT/2) - 2\sin(wT/4)\right] ,$$

where $w = 2\pi f$ and f is the input frequency of the signal. The filter output, σ, is given by

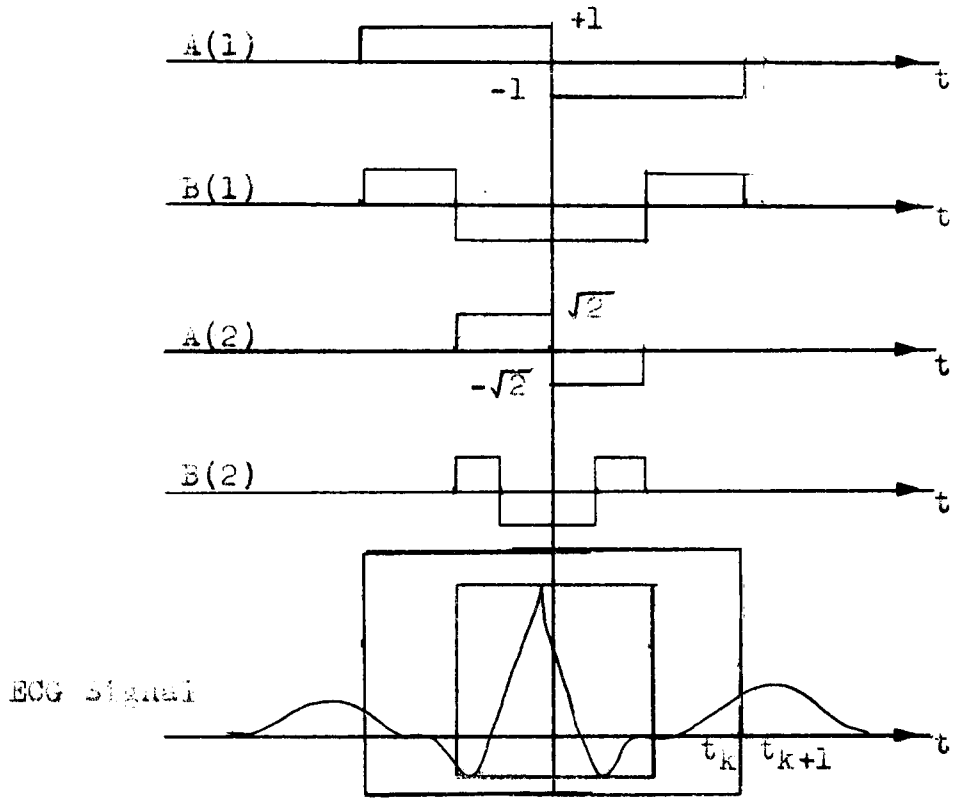
$$\sigma^2(jw) = A^2(jw) + B^2(jw) .$$

The frequency response for this type of filter is shown in Figure 4a.

The filter accepts all frequencies and enhances frequencies which are determined by a window size. Figure 4b shows the A and B components of two different window sizes and how they correspond to the ECG signal in time t_k. Each filter output corresponds to the ECG output at the



(a) Normalized filter frequency response.



(b) Filter window and ECG signal relationship.

Figure 4. Graphical description of digital filter.

midpoint of the window. The observability of the ECG through the window is a function of time. Therefore, in time t_{k+1} the window will not see the leftmost point in the t_k window but will see a point to the right of the t_k window which was not seen in the t_k window. In other words, as the ECG signal progresses in time, t, to the right, a point is dropped on the left of the window and another is picked up on the right of the window.

The normalized frequency of one is determined by the window width. For example, if the window width is 0.05 seconds, the normalized frequency of one corresponds to a center frequency of 1/0.05 or 20 Hertz. The maximum filter output will occur at 92.091 percent of the center frequency or 18.192 Hertz. This value is obtained by determining the value of w for which $\sigma^2(jw)$ is a maximum. This point of maximum amplitude can be shifted to any frequency by increasing or decreasing the width of the window which is a function of the sampling rate. Filter frequency will be referred to as the frequency which the filter enhances and is calculated as

$$\text{Filter frequency} = \frac{1}{\text{time of window}} \times 0.92091$$

where the time of the window is the product of the sampling rate and the number of points in the window, and equal to T in the equations given for A(jw) and B(jw).

The filtering technique exhibits several innovations. The first is that the orthogonal functions have a value of +1 or -1 over the interval of interest and zero elsewhere. Therefore, multiplications

over the interval reduce to summations. The second innovation is the use of a moving window approach. The epochs, which are defined as intervals of data, are overlapping rather than the customary non-overlapping epochs. The advantage is that the epochs are time representative. Events happening in the filtered signal are shown by the filter output in the time sequence in which they occur.

VI. ORGANIZATION

The research is divided into two parts. The first part, discussed in Chapter II, deals with the work done in recognizing the R-wave peak. Recognition of the R-wave peak was the preliminary test of the filtering method and is important in the P-wave onset measurement because the R-R interval can be used as a normalizing factor for generalizing the method over all cases. The second part, discussed in Chapter III, concerns the work done in recognizing the onset of the P-wave.

CHAPTER II

DETECTION OF R-WAVE PEAK

I. DESCRIPTION OF DATA

ECG data was obtained from a standard three lead derivation designed to compensate for body inhomogeneities. This Frank lead output was recorded at The University of Tennessee Memorial Research Center in Knoxville, Tennessee. All cases received were abnormal. Six cases were used in the research. The three lead outputs were recorded simultaneously on a four channel FM tape recorder. The fourth channel was grounded and used in the recording system to minimize noise in the other channels during playback by using a differential technique. Each channel output was filtered and digitized using a DEC PDP-11 in the Electrical Engineering Department of The University of Tennessee. The sampling rate was set at 100 samples per second to allow for detection of frequencies up to 50 Hertz which was the highest frequency of interest.

Each channel output was filtered through five filters with filtering frequencies of 11.49, 5.75, 2.87, 1.44, and 0.72 Hertz respectively. These filters correspond to window sizes of 8, 16, 32, 64, and 128 points. As described in Chapter I, the filter frequency is a function of the number of points in the window and the sampling rate.

$$\text{Filter frequency} = \frac{1}{\text{points in window} \times \text{sampling rate}} \times 0.92091.$$

Varying the number of points in the window or the sampling rate will change the filter frequency. Theoretically, an infinite number of

filtering frequencies would describe the channel output. However, five filters were used in this research. The first filter was chosen with a window size of eight and each succeeding filter had a window size twice the previous one. It is important to note that the filtering frequencies were fixed throughout the investigation of the R-wave peak. Use of static filter settings in this initial research helped further define the applications of the filtering process used in Chapter III.

Three second intervals of digitized channel and filter outputs were selected from each case and punched on paper tape. The format of the tape was binary with each channel output from one to three followed by the five filter outputs for each channel. This tape was then read on a Tally 424 reader and transferred to magnetic tape using an IBM 1800 at Oak Ridge Associated Universities Medical Division in Oak Ridge, Tennessee.

The data format of the magnetic tape had a block length of 180 words and each record contained eighteen words (three channel outputs and five filter outputs per channel). The first block of each case contained identifying information in words one through six. The identifying information was set number, sampling rate, total sampling time, and date received. The last record of each case contained 9999 in the first word of the record. This became the standard format. All further analysis was performed on the IBM 1800 and magnetic tape was used as the input device.

II. THE CLASSIFIER

A linear discriminant function classifier was chosen as the means of measuring the effectiveness of the filtering as a feature extractor. A "linear" classifier was used because any further use of this feature extraction method would be with a simple classifier. If the feature extraction method proved to be effective, a linear classifier would be sufficient.

A classifier is a machine that, when presented with a pattern, classifies that pattern into a certain class or group of patterns. A linear classifier attempts to find linear surfaces which separate the classes. When training the linear classifier to recognize patterns, one is actually finding the surfaces which separate the classes.

Linear discriminant functions are defined as follows (12): Given m pattern classes c_1, c_2, c_3, . . . c_m, discriminant functions are functions $g_i(\underline{X})$ such that, given an augmented pattern $\underline{X} = (x_1, x_2, \ldots x_d, 1)$ belonging to class c_i, $g_i(\underline{X}) > g_j(\underline{X})$ for all $j \neq i$.

The two class situation can be simplified in the following manner: Let $g(\underline{X}) = g_2(\underline{X}) - g_1(\underline{X})$, then $g(\underline{X}) > 0$ when $\underline{X}$ belongs to class c_2 and $g(\underline{X}) < 0$ when $\underline{X}$ belongs to class c_1. The surface separating the two classes has the equation $g(\underline{X}) = 0$. The discriminant function has the form

$$g(\underline{X}) = \underline{W} \cdot \underline{X}$$

where $\underline{W} = (w_1, w_2, \ldots w_d, w_{d+1})$. Training the machine involves presenting the machine with patterns belonging to a known class and determining the weights, $\underline{W}$.

For proper classification $\underline{W} \cdot \underline{X} > 0$ for class c_2 and $\underline{W} \cdot \underline{X} < 0$ for class c_1. After each attempt which fails to properly classify $\underline{X}$, the weight vector $\underline{W}$ is modified and a new weight vector is calculated as

$$\underline{W}_{k+1} = \underline{W}_k + \rho \underline{X}$$

where $\underline{W}_k$ is the weight vector used to test proper classification and $\underline{W}_{k+1}$ is the new vector which properly classifies $\underline{X}$. ρ is used to scale the weights.

III. THE TRAINING ALGORITHM

The two class training algorithm used in the research is as follows:

1. Initialize $\underline{W} = \underline{0}$.
2. Present a pattern $\underline{X}$ of known class.
3. Compute $g(\underline{X}) = \underline{W} \cdot \underline{X}$.
4. If $\underline{X}$ belong to class c_2,
   a. if $g(\underline{X}) > 0$, leave $\underline{W}$ unchanged and go to step 2.
   b. If $g(\underline{X}) \leq 0$, compute a new $\underline{W} = \underline{W} + \rho \underline{X}$, go to step 2.
5. If $\underline{X}$ belongs to class c_1,
   a. if $g(\underline{X}) < 0$, leave $\underline{W}$ unchanged and go to step 2.
   b. if $g(\underline{X}) \geq 0$, compute a new $\underline{W} = \underline{W} - \rho \underline{X}$, go to step 2.

A listing of the training program TRAN2 is given in Appendix A.

IV. DESCRIPTION OF THE PATTERNS

Generally, the easiest wave to identify is the R-wave; and the peak of the R-wave can usually be identified by the untrained eye.

Therefore, the algorithm trained on the peak of the R-wave. Class c_1 consisted of non-peak points and class c_2 consisted of peak points.

Since visual observation revealed that the fourth and fifth filters contained little information, these filter outputs were omitted for each channel. Plots of one case are shown in Figures 5 through 7. It can be seen that filters one, two, and three have peaks or dips which correspond to the R-wave peak in time. Filters four and five do not have any significant and consistent time correspondence to the R-wave peak. The patterns used in training consisted of the three channel outputs, three filter outputs for each channel and the augmented one.

VI. RESULTS OF FIVE TRAINING DATA SETS

The first data set used in training chose class c_2 patterns as patterns where the R-wave peak occurred in channel two. For example, if the R-wave peak occurred at pattern 54 in channel two and pattern 55 in channels one and three, pattern 54 was in class c_2 and pattern 55 was in class c_1. This first attempt at training was very optimistic by continuously training on one case of data until the percent correctly classified was greater than 99 percent or 50 iterations had been completed on the case under consideration.

Partial results of this training are shown in Table 1. A test is considered to be one time through all the cases used. The total percent correct is calculated from the last iteration performed on each case. Fifteen tests were completed with no significant change in the

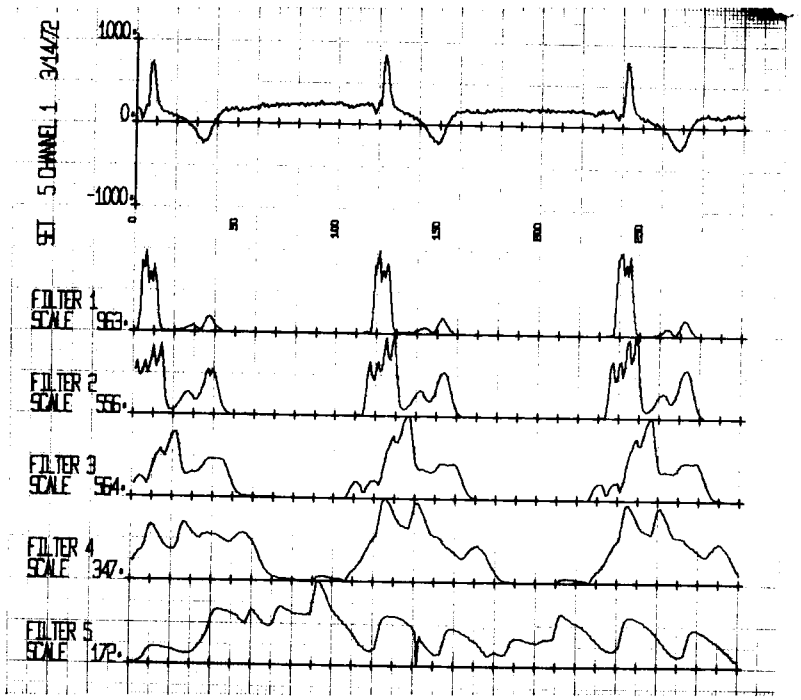


Figure 5. Plot of channel one and five filter outputs for an abnormal, unsmoothed ECG with fixed filter frequencies.

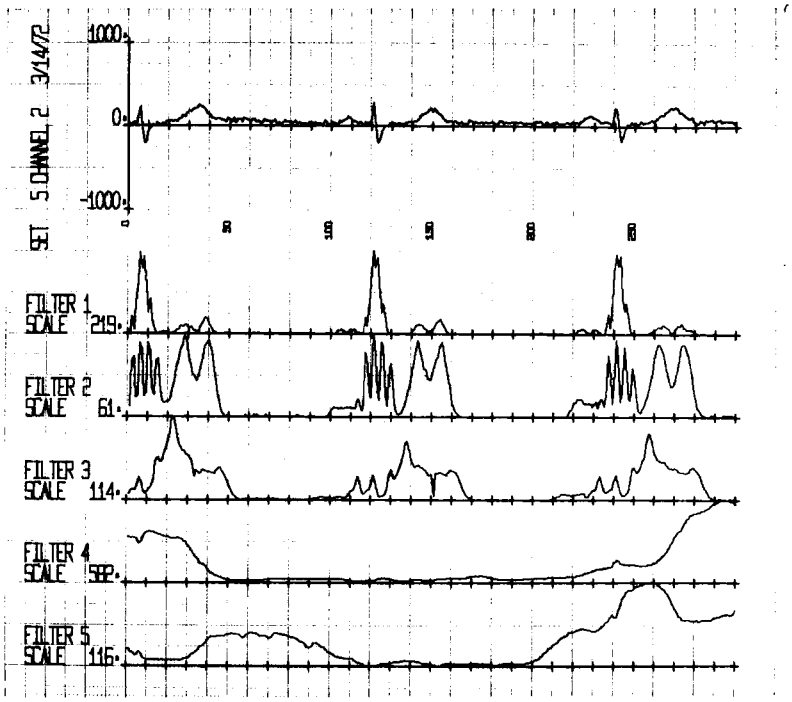


Figure 6. Plot of channel two and five filter outputs for an abnormal, unsmoothed ECG with fixed filter frequencies.

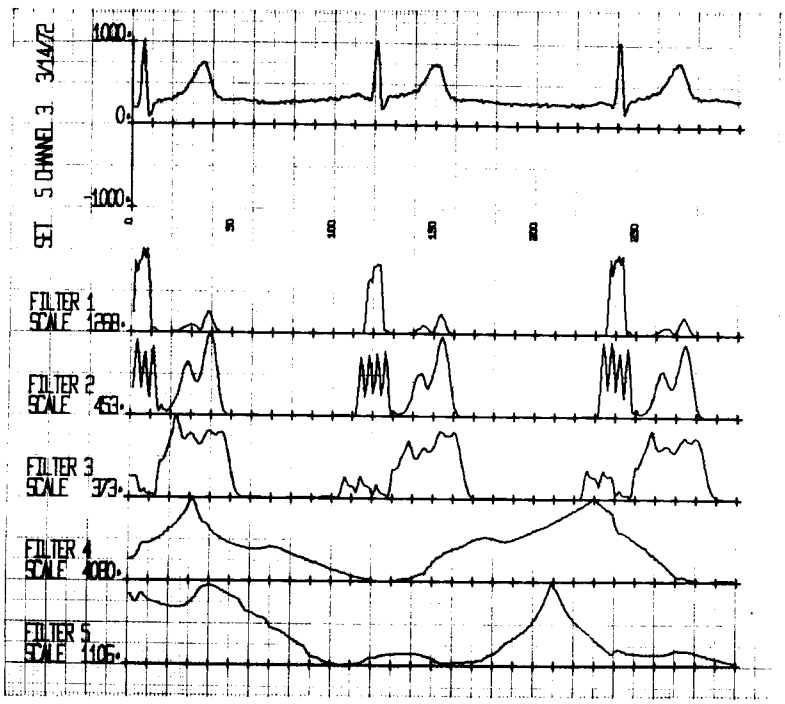


Figure 7. Plot of channel three and five filter outputs for an abnormal, unsmoothed ECG with fixed filter frequencies.

Table 1. Results of first training data set with class c_2 patterns having R-wave peaks in channel two (training was by case).

Case or Set	Test 1		Test 2		Test 3		Test 4		Test 5	
	Itera-tions	Percent Correct	Itera-tions	Percent Correct	Itera-tions	Percent Correct	Itera-tions	Percent Correct	Itera-tions	Percent Correct
2	6	100	5	100	3	100	3	99.6	5	99.6
3	25	99.6	8	99.6	16	99.6	19	99.6	21	99.6
5	50	97.3	50	97.0	50	98.8	50	97.0	50	97.0
6	50	98.2	50	98.3	50	98.3	50	98.0	50	98.3
9	6	99.6	5	100	4	100	4	100	4	100
10	50	98.6	50	98.6	50	97.3	50	96.6	50	97.0
Total		98.7		98.9		98.7		98.5		98.6

results from the first five.

The second attempt at training was similar to the first except patterns with peaks in channel one were used as the patterns belonging to class c_2. The training method was changed to eliminate iterations through one set at a time. An iteration is defined as one time through all the patterns and the correct percentage was calculated from the last iteration with no change in weights. After 390 iterations, the percent correct was 84.1 percent.

The third training data set used patterns with a R-wave peak in any channel as class c_2 patterns. If the R-wave peak occurred in channel one at pattern 30, at pattern 31 in channel two and at pattern 29 in channel three, all these patterns were used as class 2 patterns. After 504 iterations, the percent correct was 84.8 percent.

The fourth training data set used patterns in class c_2 as the average point among the peaks in the three channels. For example, if pattern 19 had a peak in channel one, pattern 21 had a peak in channel two and pattern 20 had a peak in channel three, the class c_2 pattern would be pattern 20. After 378 iterations, the percent correct was 89.2 percent.

The fifth and final attempt of training with this data consisted of transforming the data by taking a moving average of every three points. The peak patterns of class c_2 were chosen to be those averaged points where the most peaks appeared in the three channels. For instance, averaging patterns 41, 42, 43 resulted in a new pattern number 41. If the three channels showed peaks in each of the patterns 41,

42, and 43, then the new averaged pattern 41 was in class c_2. If only two channel peaks were represented in the averaged pattern, placement into a class was reserved until a greater or lesser number of peaks in an averaged pattern was observed. If two averaged patterns contained the same number of peaks, only the first was chosen for class c_2. After 17 iterations, the percent correct on the 18th iteration with no change of weights was 85.8 percent.

CHAPTER III

DETECTION OF THE P-WAVE ONSET

I. DESCRIPTION OF THE DATA

One hundred six cases of ECGs were received from The University of Tennessee Medical Unit in Memphis. This data contained the digitized X, Y, and Z channel outputs of one heart cycle per case. Sampling was at 500 samples per second. The data had been averaged and smoothed and each case was considered normal (14). Furthermore, a committee of doctors had determined the onset of the P-wave in each case and this point was included in the data. Conversion of the data into a format useable on the IBM 1800 is described in Appendix B.

II. THE FILTER SIMULATOR

The method described in this chapter for the detection of the P-wave onset differs from the method for detection of the R-wave peak described in Chapter II. The major difference is that, in detecting the P-wave onset, the filter frequencies were not fixed. The statement of the problem changed from "Will the filtering method be an applicable feature extractor for ECG waveforms and what are some characteristics of the filter output?" to "What filter frequencies optimize the feature extraction of specific waves--namely the P-wave--and how may the optimum filter frequencies be found?" A program, CHANE, described in Appendix C was written to simulate the filtering process

and allow for variation of the filtering frequencies.

As stated previously in Chapter I, the filter frequency is a function of the time of the window. The time of the window may be calculated as the product of the number of points in the window and the sampling rate, 0.002 seconds. Therefore, the window size may be referred to as a certain number of points or as an interval of time. With a window size of 320 points and a sampling rate of 0.002 seconds, the filter frequency is 1.4389 Hertz.

$$1.4389 = \frac{1}{320 \times 0.002} \times 0.92091 \ .$$

Therefore, frequencies of 1.4389 Hertz would be enhanced in the ECG signal. The same filter frequency could be achieved with a sampling rate of 100 per second and a window size of 64 points. Also, it should be noted that this window size would be in a set of window sizes similar to the set in Chapter I, but the minimum window size would be 40 points--40, 80, 160, 320, 640.

Each filter output corresponds to a channel output which occupies the midpoint of the window. The filter output will be denoted as σ. $\sigma(160)$ is defined to be the filter output corresponding to the 160th point in the window. $\sigma(160)$ is calculated as

$$\sigma^2(160 = (\sum_{i=1}^{160} x_i - \sum_{i=161}^{320} x_i)^2 + (\sum_{i=1}^{80} x_i - \sum_{i=81}^{240} x_i + \sum_{i=241}^{320} x_i)^2 \ .$$

After the initial calculation of $\sigma(160)$, the calculation of $\sigma(161)$ proceeds rapidly given

$$\alpha(160) = \sum_{i=1}^{160} x_i - \sum_{i=161}^{320} x_i$$

and

$$\beta(160) = \sum_{i=1}^{80} x_i - \sum_{i=161}^{240} x_i + \sum_{i=241}^{320} x_i \quad .$$

$\sigma^2(160)$ then becomes

$$\sigma^2(160) = \alpha^2(160) + \beta^2(160)$$

and $\sigma(161)$ is derived as

$$\sigma^2(161) = (\alpha(160)-x_1+2x_{161}-x_{431})^2+(\beta(160)-x_1+2x_{81}-2x_{241}+x_{321})^2 \quad .$$

Since the data for each case represented exactly one heart cycle, the points could be placed head to tail to obtain as many filter outputs as points in the heart cycle. This, of course, assumes a repetitious heart cycle. Such an assumption is valid because of the previous smoothing and averaging of this data.

III. OBSERVATION OF THE FILTER PLOTS

Observations were made of the filter plots resulting from program CHANE, the simulator program. A marker was plotted to indicate where the P-wave began, as interpreted by the physicians. Reference is made to Figure 8. It was observed that use of the proper window size caused the minimum filter output point and the P-wave onset marker to coincide. A method was needed to systematically choose the window size such that the minimum filter output would detect the onset of the P-wave. The method which is described in sections IV and V establishes a linear functional relationship between the optimum P-wave detection filter frequency and the heart cycle frequency.

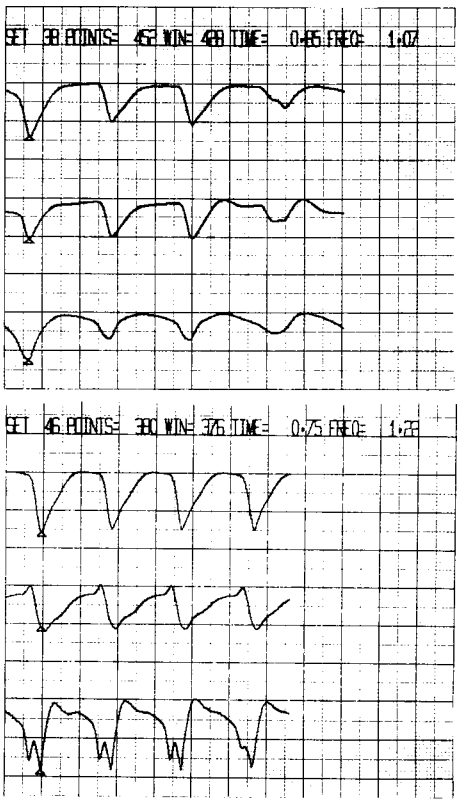


Figure 8. Sample plot of filter output for two cases with filter frequency varied by changing the window size.

Triangle marker is on P-wave onset point chosen by physicians.

IV. FILTER FREQUENCY AS A FUNCTION OF THE HEART CYCLE FREQUENCY

Since each person has his own norm and one easily obtained measurement of each person's ECG is the R-R interval, partial normalizing of the optimum P-wave detection filter frequency could be obtained readily. This method of normalizing is by no means exhaustive. There are many other attributes which are peculiar to each individual ECG which might be used.

In this research, the relationship of the optimum filter frequency as a function of the heart cycle frequency is investigated. However, this functional relationship implies other functional relationships because the filter frequency and the heart cycle frequency are themselves functions of other measurements. This means that the functional relationship is not only between the heart cycle frequency and the filter frequency, but also between any of the measurements on which the heart cycle frequency and the filter frequency depend. The heart cycle frequency can be obtained from a measure of the R-R interval. The filter frequency is dependent upon the sampling rate and the window size. Also, the filter which optimizes the detection of the P-wave onset has its frequency closely associated with the frequency of the P-wave. Therefore, other terms may be used to discuss the same relationship used in this research. For example, the P-wave frequency is a function of the R-R interval.

V. METHOD OF INVESTIGATION

The simulator program was run on automatic selection of set number and window size, and the results of the filtering process were recorded. All window sizes from the largest number less than the total points in the heart cycle and divisible by four to 16 were used in the filtering.

Filtering frequencies which had a minimum filter output within 0.004 seconds of the point chosen by the physicians were plotted against the heart cycle frequency. The plots are shown in Figures 9, 10, and 11. Each plot is the result of filtering the three channel outputs. Linear regression was used to fit a line through the points and thus obtain the filter frequency as a linear function of the heart cycle frequency. Confidence intervals were drawn about the mean using a Student's T value of 96 percent. The confidence intervals state that with 96 percent confidence the mean should be within the confidence bands. Visual observations of the plots reveal that neither the regression line nor the confidence intervals adequately apply to the points plotted.

The plots show three groups of points at three different levels in the filter frequency. Divisions occur in channel one at 1.80 cycles/second and at 5.20 cycles/second in the P-wave filtering frequency. In channel three divisions occur at 1.75 cycles/second and 4.5 cycles/second. In channel two there were few cases in which any filter frequency had its minimum output within 0.004 seconds of the P-wave onset

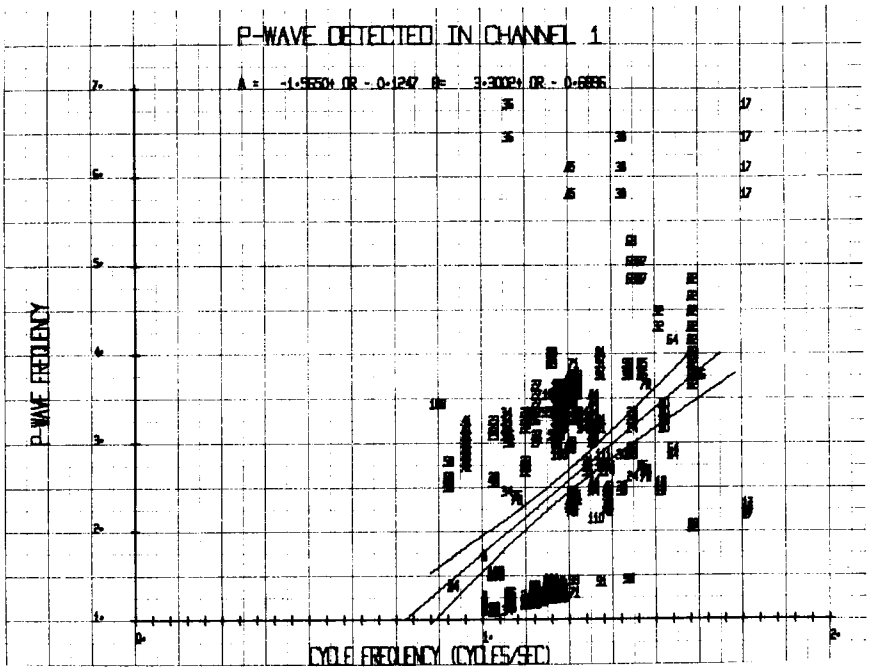


Figure 9. Heart cycle frequency plotted against channel one filter frequency in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the mean at 96 percent.

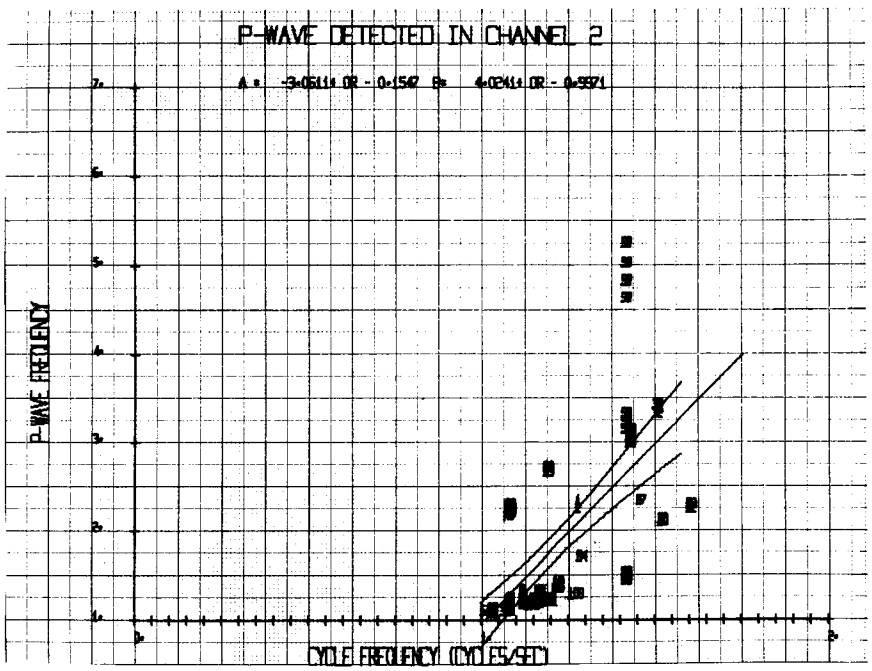


Figure 10. Heart cycle frequency plotted against channel two filter frequency in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the mean at 96 percent.

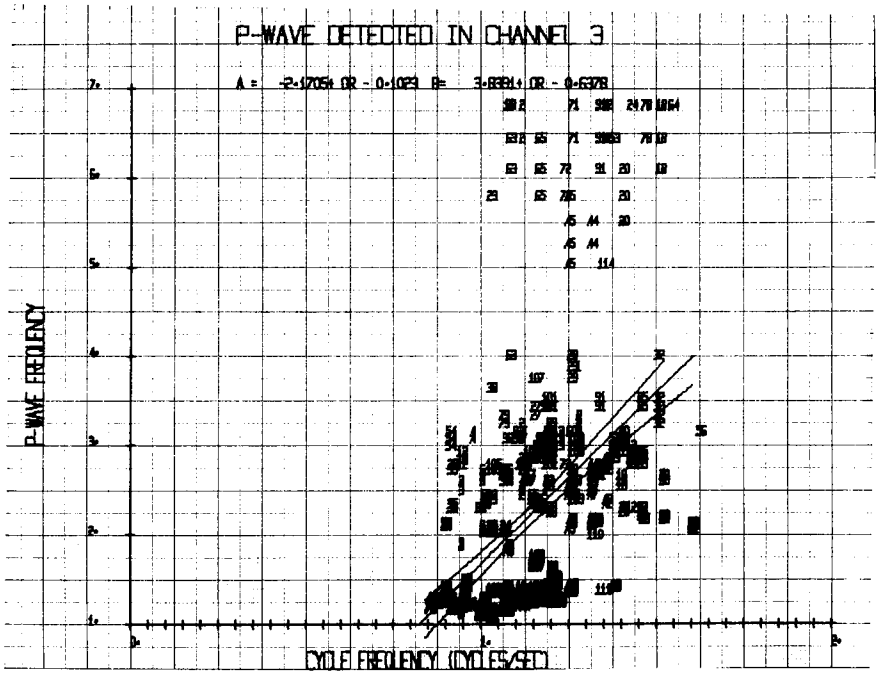


Figure 11. Heart cycle frequency plotted against channel three filter frequency in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the mean at 96 percent.

point. Therefore, channel two was not considered further. Observations of the groups revealed that in some cases there were as many as three different levels of filter frequencies in which the minimum filter output came within 0.004 seconds of the P-wave onset point.

Since the objective in plotting the heart cycle frequency against the filter frequencies which detected the P-wave onset within 0.004 seconds was to establish a linear relationship, the failure of one regression line to adequately describe the relationship was seen as an indication that possibly more than one linear relationship existed. The procedure used was to divide the filtering frequencies of channels one and three into three groups. The upper groups contained few cases and all but four cases were represented in the two lower groups so the upper groups were discarded. Further research used only the two lower groups of channels one and three and linear relationships were established within these groups between the filtering frequency and the heart cycle frequency.

In Figure 12, the plot of the lower group of points that were shown in Figure 9, page 35, is enlarged and a regression line is drawn through the points. Confidence intervals are drawn about the individuals at 95 percent. These bands use the Student-t value at 95 percent but are wider than those in Figure 9 because they mean that with 95 percent confidence all the individual points will fall within the bands.

Figure 13 is a similar treatment of Figure 9 but the plot of the middle group of points in Figure 9 is enlarged and a linear relationship

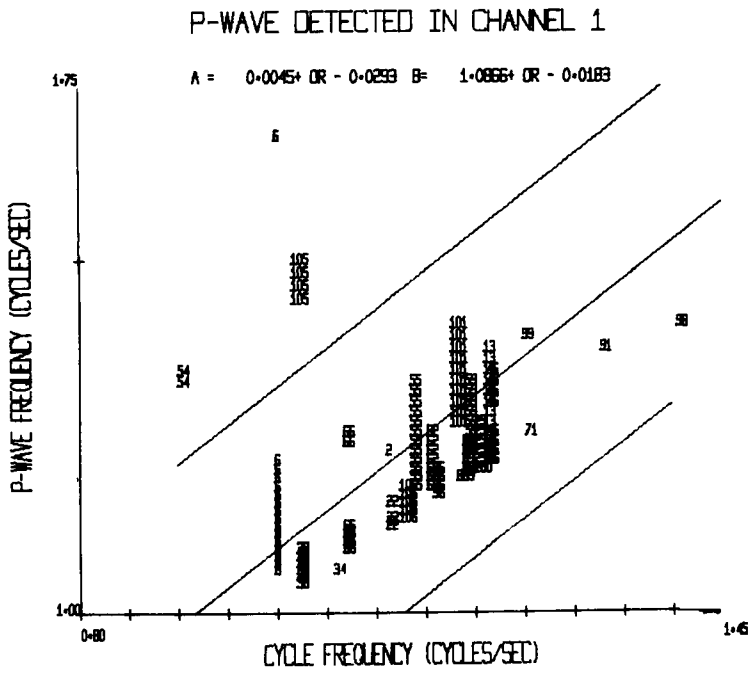


Figure 12. Heart cycle frequency plotted against P-wave filter frequencies of channel one less than 1.80 cycles/second in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the individuals at 95 percent.

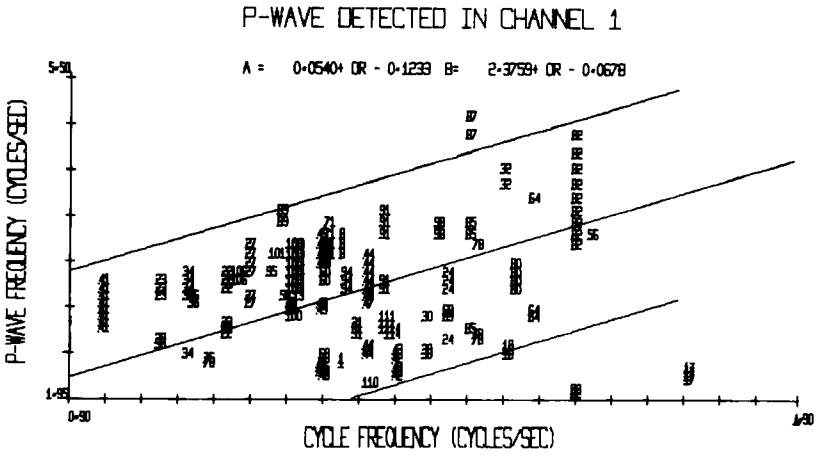


Figure 13. Heart cycle frequency plotted against P-wave filter frequencies of channel one greater than 1.80 cycles/second and less than 5.20 cycles/second in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the individuals at 95 percent.

between the filtering frequency and heart cycle frequency is established within this group. The channel three plot in Figure 11, page 37, has the two lower groups enlarged in Figures 14 and 15. From each of these groups, a different linear relationship was established between the optimum P-wave detection filtering frequency and the heart cycle frequency.

The coefficients A and B for $Y = A + BX$, where Y is the optimum P-wave filtering frequency and X is the heart cycle frequency, were thus obtained from linear regression in each group. The results are as follows:

1. Channel one--P-wave frequency less than 1.80 cycles/second.

   A = 0.00452

   B = 1.08662

2. Channel one--P-wave frequency greater than 1.80 cycles/second and less than 5.20 cycles/second.

   A = 0.05406

   B = 2.37590

3. Channel three--P-wave frequency less than 1.75 cycles/second.

   A = 0.00862

   B = 1.17900

4. Channel three--P-wave frequency greater than 1.75 cycles/second and less than 4.5 cycles/second.

   A = 0.04403

   B = 2.19897

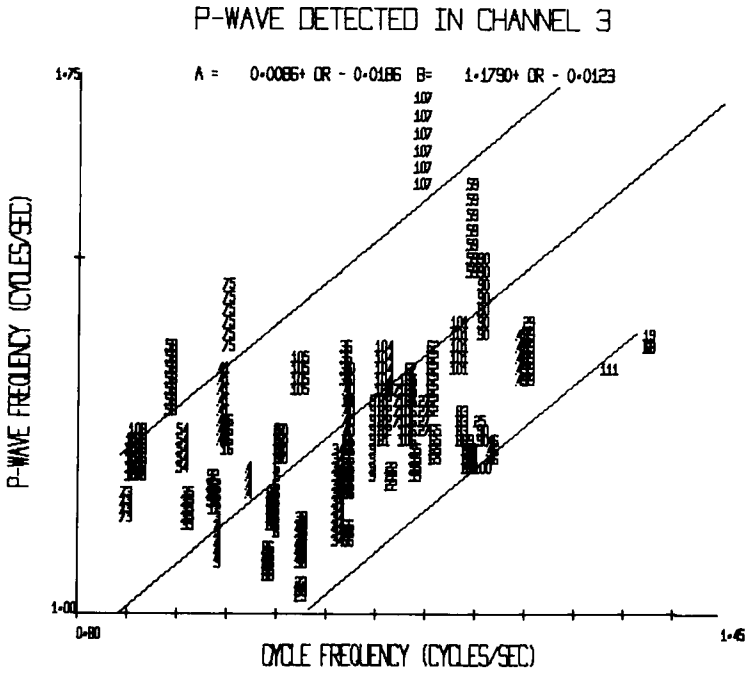


Figure 14. Heart cycle frequency plotted against P-wave filter frequencies of channel three less than 1.75 cycles/second in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the individuals at 95 percent.

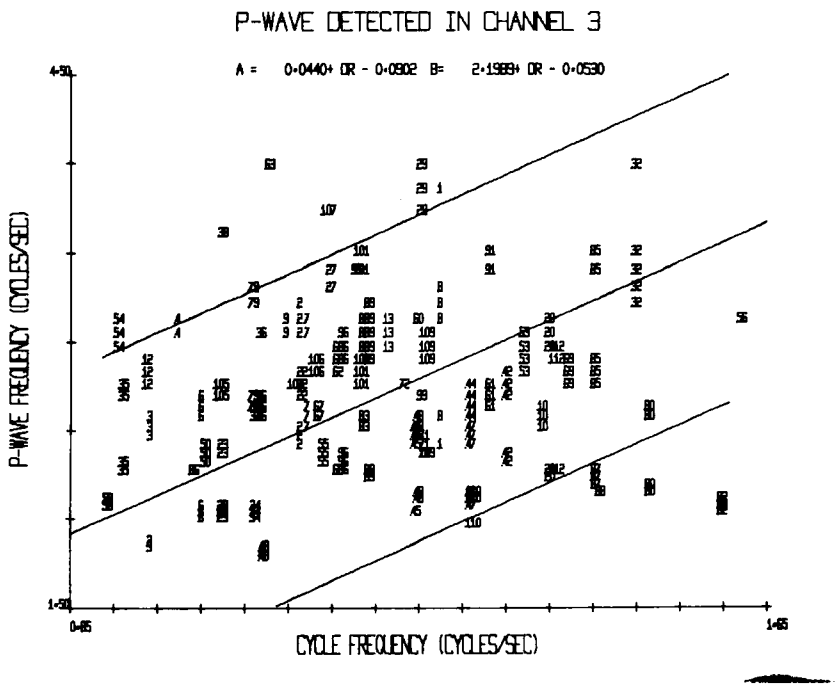


Figure 15. Heart cycle frequency plotted against P-wave filter frequencies of channel three greater than 1.75 cycles/second and less than 4.5 cycles/second in which P-wave onset detected within 0.004 seconds.

Linear regression line is drawn with confidence intervals about the individuals at 95 percent.

The hypothesis that A = 0 at a confidence of 99 percent may not be rejected. However, the hypothesis that B = 0 may be rejected. Therefore, the Y-intercept could possibly be 0; whereas the slope is definitely not zero and the P-wave frequency is a function in each group, or possibly a multiple, of the heart cycle frequency.

The final test of the hypothesis that the proper filter frequency for detecting the P-wave onset is a function of the heart cycle frequency was to use the derived equations to choose the window size for filtering the ECG signal and determine how effective the equations were. Program CHANE was again used with the window size determined by the regression lines. Results of this test are shown in Table 2.

In Table 2 the coefficients which were used are given and underneath them are the results of the filters for channels one and three. The results for each case are the time interval between the minimum filter output and the onset of the P-wave. The negative sign indicates the minimum filter output was to the left of the onset of the P-wave. At the end of the results for each case, a summary is given. The row entitled "L6" is the number of cases for each channel and regression line in which the minimum filter output was within 0.006 seconds of the P-wave onset. Other summary lines give the mean interval, standard deviation of the interval, minimum interval, maximum interval, and percent of cases within 0.006 seconds for each line and output from filtering channels one and three. The final result indicates that 51 cases or 48 percent of the total cases had at least one minimum filter output within 0.006 seconds of the P-wave onset point chosen by the physicians.

Table 2. Results of regression equations used to set filter frequencies for detection of P-wave onset point.

	Line 1 A = 0.00452 B = 1.08662		Line 2 A = 0.05406 B = 2.37590		Line 3 A = 0.00862 B = 1.17900		Line 4 A = 0.04403 B = 2.19897	
Case	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3
1	-0.336	0.046	0.048	0.039	-0.350	0.053	-0.350	0.232
2	-0.006	-0.008	-0.026	0.016	0.372	0.006	-0.036	0.004
3	0.458	-0.008	-0.042	-0.016	0.442	0.002	-0.052	-0.034
4	0.410	-0.026	-0.048	-0.044	0.392	-0.006	-0.056	-0.058
6	-0.002	-0.006	0.092	-0.016	0.002	-0.002	-0.030	-0.026
7	0.378	-0.009	-0.032	-0.018	0.366	0.006	-0.036	-0.004
8	0.182	0.018	-0.012	-0.004	0.184	0.034	0.206	0.018
9	0.386	-0.004	-0.039	-0.019	0.374	0.004	-0.046	-0.036
10	-0.304	-0.152	-0.036	-0.024	-0.316	-0.176	-0.030	-0.036
11	0.406	-0.012	-0.118	0.206	0.392	0.380	-0.126	0.201
12	0.431	-0.038	-0.176	-0.038	0.416	-0.009	0.222	-0.052
13	0.002	-0.182	-0.008	-0.004	0.192	-0.156	0.228	-0.019
15	-0.131	-0.079	-0.182	-0.102	-0.148	-0.065	-0.078	-0.102
16	0.408	-0.044	-0.046	-0.056	0.394	-0.024	-0.064	-0.032
17	-0.022	-0.030	0.100	0.032	-0.009	-0.016	0.036	0.090
18	0.060	0.068	0.058	-0.032	0.064	0.079	0.228	0.152
19	-0.333	0.009	-0.019	0.156	0.184	0.016	-0.028	0.156
20	-0.328	0.039	-0.088	0.014	0.319	0.056	-0.022	0.004
22	0.208	0.008	0.000	-0.004	0.208	0.026	0.184	0.121
23	0.445	0.018	-0.019	0.008	0.430	0.036	-0.024	0.028
24	0.048	-0.106	0.006	-0.305	0.052	-0.134	-0.002	-0.093
25	0.366	0.009	0.000	-0.124	0.190	0.026	-0.082	-0.012
27	0.382	-0.004	-0.014	0.012	0.378	0.014	-0.026	-0.096
28	0.456	-0.068	-0.164	-0.076	0.436	-0.042	-0.172	-0.060
29	0.344	-0.004	-0.030	-0.170	0.353	0.012	-0.039	-0.156
30	-0.104	0.076	-0.052	-0.048	-0.086	0.090	0.014	0.158
31	-0.347	0.356	0.019	-0.102	0.354	0.344	0.009	-0.114
32	-0.266	0.064	0.053	0.173	-0.280	0.062	0.042	0.000
34	0.417	0.002	0.009	-0.002	0.403	0.008	-0.006	-0.018
35	0.442	-0.036	-0.042	-0.034	0.424	-0.014	0.232	-0.053
36	0.388	-0.030	0.346	-0.004	0.374	-0.014	0.344	-0.039
37	0.417	-0.036	-0.048	-0.065	0.402	-0.009	-0.052	-0.070
38	0.008	0.002	-0.006	-0.012	0.022	0.016	0.228	0.028
39	0.420	0.224	-0.030	-0.290	0.406	0.224	-0.034	0.136
40	0.406	-0.009	-0.126	0.038	0.394	-0.004	-0.136	0.028
41	-0.019	-0.032	-0.026	-0.034	0.424	-0.018	-0.042	-0.053

Table 2 (continued).

Case	Line 1 A = 0.00452 B = 1.08662		Line 2 A = 0.05406 B = 2.37590		Line 3 A = 0.00862 B = 1.17900		Line 4 A = 0.04403 B = 2.19897	
	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3
42	-0.110	0.046	-0.092	-0.078	0.106	0.060	-0.078	-0.068
43	0.428	-0.048	-0.039	-0.044	0.410	-0.034	-0.056	-0.064
44	0.046	0.039	-0.002	0.022	-0.342	0.056	-0.008	0.012
45	-0.346	0.053	-0.114	-0.107	-0.360	0.074	-0.026	-0.092
46	0.004	0.018	-0.118	-0.126	0.009	-0.009	0.214	0.121
47	0.058	-0.116	-0.002	0.074	0.262	0.078	0.042	0.026
48	0.060	-0.140	0.056	0.065	0.065	0.068	0.044	-0.008
49	0.358	0.000	0.006	0.062	0.346	0.014	-0.008	0.056
50	-0.019	-0.030	-0.050	0.008	-0.006	-0.009	-0.056	-0.008
51	-0.424	0.036	-0.126	0.110	0.248	0.034	-0.128	0.106
52	-0.032	-0.034	0.154	0.009	-0.022	-0.030	-0.048	-0.050
53	0.028	-0.142	-0.019	-0.076	0.229	0.042	-0.030	0.004
54	-0.058	-0.050	-0.062	-0.058	-0.046	-0.030	-0.070	-0.048
55	0.002	0.006	-0.022	-0.012	0.006	0.024	-0.028	0.016
56	0.076	0.052	0.006	0.187	0.050	0.065	0.024	0.022
58	0.448	-0.048	-0.028	-0.076	0.428	-0.032	-0.048	-0.058
59	0.002	0.187	-0.009	-0.032	0.012	-0.009	0.194	-0.034
60	0.039	0.019	-0.078	0.084	-0.364	0.028	0.028	0.074
61	-0.328	0.039	0.036	-0.084	-0.342	0.056	0.026	0.014
62	0.004	-0.006	-0.145	-0.004	0.016	0.004	-0.065	0.024
63	-0.186	0.032	-0.150	0.079	0.220	0.019	-0.136	0.088
64	-0.106	-0.107	-0.014	-0.282	0.300	-0.142	-0.022	0.042
65	0.184	0.008	-0.134	0.120	0.184	0.026	-0.130	0.112
66	0.016	-0.004	-0.028	0.079	0.006	0.006	-0.032	0.074
67	0.372	-0.018	-0.114	0.012	0.358	0.004	-0.120	-0.006
68	-0.333	0.024	-0.042	0.048	0.328	0.039	-0.048	0.114
69	0.206	0.039	0.022	-0.072	0.215	0.056	-0.064	-0.065
70	-0.032	-0.284	0.186	-0.060	-0.024	-0.314	0.158	-0.060
71	-0.162	-0.170	-0.116	-0.120	0.184	0.042	-0.110	-0.104
72	0.192	-0.166	-0.038	0.014	0.194	0.052	-0.116	0.004
73	-0.060	-0.053	0.210	0.096	-0.042	-0.032	0.198	-0.044
75	0.414	-0.053	0.375	-0.058	0.394	-0.036	-0.064	-0.072
76	0.416	-0.166	0.019	-0.384	0.224	-0.131	0.009	-0.380
78	0.222	0.068	-0.006	0.106	0.228	0.079	-0.012	0.172
79	-0.236	-0.276	-0.194	-0.036	-0.254	-0.305	-0.180	-0.050
80	-0.079	-0.092	0.006	-0.030	-0.277	-0.072	0.004	-0.030
81	0.372	-0.009	-0.032	-0.034	0.356	0.006	-0.044	-0.046
82	-0.030	-0.042	0.002	0.204	-0.014	-0.022	0.004	0.200
83	0.014	0.006	-0.026	-0.006	0.022	0.024	-0.034	0.006

Table 2 (continued).

	Line 1 A = 0.00452 B = 1.08662		Line 2 A = 0.05406 B = 2.37590		Line 3 A = 0.00862 B = 1.17900		Line 4 A = 0.04403 B = 2.19897	
Case	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3	CHAN1	CHAN3
84	0.410	-0.012	-0.076	-0.075	0.394	0.006	-0.072	-0.082
85	0.210	0.039	-0.008	-0.002	0.212	0.046	-0.016	0.016
86	0.458	0.002	-0.039	0.016	0.442	0.018	-0.046	0.104
87	0.093	-0.098	-0.034	-0.036	0.068	0.079	-0.028	0.038
88	0.086	0.074	0.070	-0.028	0.076	0.090	0.058	0.046
89	0.009	0.014	-0.014	-0.004	0.016	0.009	-0.014	-0.014
90	0.370	-0.009	-0.093	0.116	0.360	-0.002	-0.012	0.110
91	-0.342	0.018	0.002	-0.009	-0.342	0.016	-0.009	-0.019
92	0.431	-0.034	-0.038	-0.076	0.430	-0.012	-0.053	-0.060
93	0.222	-0.192	-0.168	-0.190	0.224	-0.215	0.403	-0.187
94	0.360	0.024	-0.004	0.039	0.347	0.042	-0.012	0.030
95	0.356	-0.243	-0.022	-0.028	0.342	-0.272	-0.114	-0.144
96	0.009	0.016	-0.114	-0.006	0.018	0.032	-0.026	-0.018
97	0.214	-0.144	-0.092	0.026	0.224	-0.120	-0.022	0.032
98	0.170	-0.170	-0.018	0.028	0.172	-0.194	-0.032	0.016
99	-0.006	0.000	-0.018	-0.076	0.338	0.016	-0.026	-0.056
100	0.370	0.016	-0.012	-0.028	0.358	0.032	-0.009	-0.039
101	-0.004	-0.014	-0.016	0.004	0.004	0.002	-0.102	-0.012
102	-0.134	0.039	0.018	0.052	-0.118	0.056	-0.019	0.039
103	0.009	-0.164	-0.082	-0.096	0.229	0.028	-0.014	-0.016
104	0.374	-0.022	-0.032	-0.034	0.360	-0.004	0.246	-0.018
105	-0.048	-0.038	-0.044	-0.052	-0.034	-0.022	-0.060	-0.034
106	0.008	-0.006	-0.014	-0.006	0.019	0.006	-0.026	-0.019
107	0.354	-0.034	-0.030	0.046	0.340	-0.026	-0.042	0.039
108	-0.092	-0.088	-0.078	-0.102	-0.118	-0.076	-0.173	-0.110
109	0.190	0.032	-0.124	0.004	0.190	0.046	-0.042	-0.006
110	0.098	0.078	-0.022	0.173	0.076	0.093	0.060	-0.082
111	-0.128	0.024	0.022	0.204	0.036	0.014	0.009	0.058
112	0.204	0.042	0.036	0.022	0.208	0.060	-0.036	0.009
114	-0.330	0.032	0.028	0.084	-0.342	0.036	0.262	0.078
115	0.400	-0.006	0.214	-0.044	0.386	0.014	-0.034	-0.046
SUMMARY OF P-WAVE DETECTION								
Total	106	106	106	106	106	106	106	106
Mean	0.107	-0.020	-0.020	-0.013	0.154	0.003	-0.003	-0.001
SD	0.250	0.089	0.090	0.098	0.230	0.094	0.114	0.086
Min	-0.424	-0.284	-0.194	-0.384	-0.364	-0.314	-0.350	-0.380
Max	0.458	0.356	0.375	0.206	0.442	0.380	0.403	0.232
L6	9	15	13	13	5	16	4	9
Perc	8	14	12	12	4	15	3	8

Total number of cases where P-wave detected within 0.006 seconds is 51 for 48 percent.

CHAPTER IV

SUMMARY AND CONCLUSIONS

I. R-WAVE DETECTION

Results of the first training method applied to individual cases indicate that the filtering technique is a good feature extractor for some cases. In this research weights could be found for the linear discriminant classifier which classified two cases 100 percent correctly. However, four cases were never 100 percent correctly classified in 50 iterations. When the training method was changed, it became obvious that the weights used to correctly classify the R-wave peak in one case did not generalize to other cases. The conclusion which can be drawn is that the filtering process must be normalized so that features will be extracted for all cases, not just a few. The normalization must compensate for some of the differences in individual ECGs.

The results of training on all the patterns seem rather good even with no attempt at normalization. Evaluation of the results must consider the fact that the cases were all considered abnormal and no smoothing or averaging had been done. However, it should be noted that there were only three to five R-wave peaks in the three second intervals used for each case. If all patterns were classified as non-peak points, the percent correct would be between 98 percent and 99 percent. To be significant, correct classification would have to closely approach 100 percent. Although knowledge of the difference in cases was gained in the

research on detection of the R-wave peak, total success was not achieved. Refinement of the feature extractor was needed.

II. P-WAVE DETECTION

Conclusions reached in the investigation of the R-wave detection helped formulate the approach to the detection of the P-wave onset point. The approach varied the filter frequencies to determine an optimum frequency for detecting the P-wave onset. In the P-wave detection research, it was shown that a proper filtering frequency could detect the onset of the P-wave in particular cases. An attempt was made to generalize the optimum filtering frequency to all 106 cases used in the research. The normalizing method investigated the use of the heart cycle frequency to determine the optimum P-wave detection filtering frequency. The filter frequency was determined as a linear function of the heart cycle frequency. The heart cycle frequency was chosen as a normalizing factor because it is an aspect of the ECG which is peculiar to each case and it is a readily obtainable measurement. Other attributes of an individual ECG could also be likely candidates for use as normalizing factors but only the R-R interval was used in this research. The results indicate that this is the right approach to be taken for optimizing the filter settings. In 48 percent of the cases, the P-wave onset point could be detected by the filter outputs within 0.006 seconds of the point indicated by physicians.

III. RECOMMENDATIONS FOR FUTURE RESEARCH

Several comments can be made on how the filter outputs were used and the normalizing method. First, it was observed that the minimum filter output coincided with the P-wave onset point. Use of the filters does not restrict detection to be the minimum output value. Indeed, the maximum or even some combination of filter outputs might possibly improve the detection. This aspect of filter output useage could benefit from future research.

Secondly, other normalizing factors which are easily measured in an individual ECG could be used singularly and in combinations to improve the results. The filters could be "fine tuned" to produce results which would generalize to all cases.

A minicomputer system which could be used to implement the procedure is shown in Figure 16. In this figure, the feedback of information from the technician or the minicomputer is of particular interest. The technician might initially set the filters by using a rough approximation of the R-R interval or the minicomputer might contain a simple program to calculate the R-R interval and initially set the filters. Setting the filters would be accomplished by changing the sampling rate while the number of points in the window remained constant. The exact amount of information required to initially set the filters is left for future research. After the filters are initialized to enhance certain waveforms, the information gathered could be looped back from the minicomputer to further "fine tune" the filters. In this manner a complete

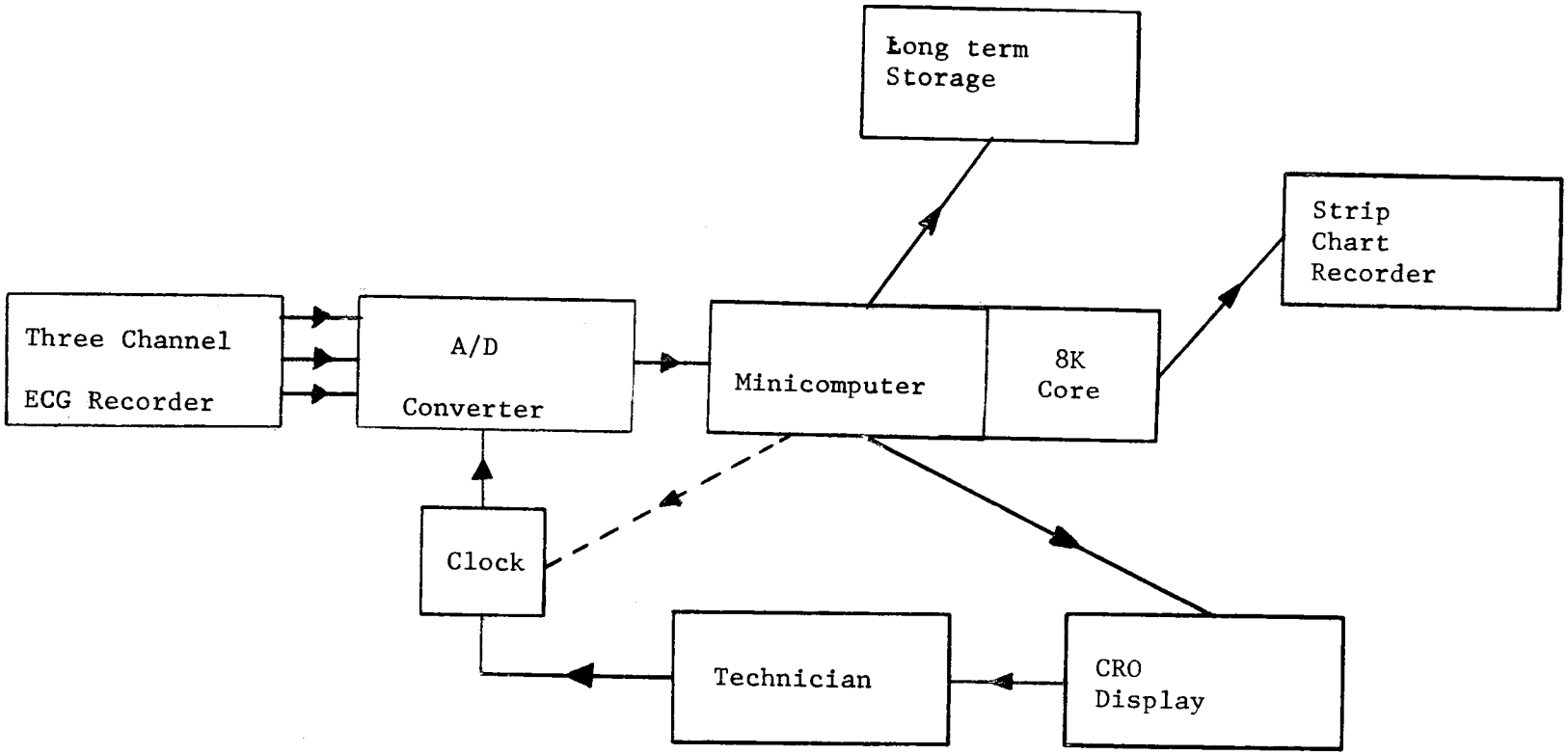


Figure 16. Typical minicomputer system for implementation of ECG filtering process for detection of waveforms.

system would evolve to analyze the ECG. A total system has not been developed in this research but a basic foundation has been laid and the direction indicated for future research.

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APPENDIXES

APPENDIX A

THE LINEAR CLASSIFIER TRAINING PROGRAM

Program TRAN2 was used to train the two class linear discriminant function classifier described in Chapter II. The program was written in FORTRAN for use on the IBM 1800. TRAN2 reads the patterns from magnetic tape using the MAGOP subroutine. Each pattern is thirteen words long. The first twelve words contain the three channel outputs and three filter outputs per channel. The thirteenth word contains the set and pattern number within the set. If the thirteenth word is positive, the pattern belongs to class c_1. If the thirteenth word is negative, the pattern belongs to class c_2. Options via data switches are provided for intermediate printing of results and misclassed patterns. Also, the program is terminated by data switch 15 or 99.9 percent correct classification.

```
**********LISTING OF PROGRAM TRAN2**********


C*************************************************************
C
C    TRAN2 TRAINS A LINEAR MACHINE TO PICK OUT ORS PEAKS
C      CLASS 2 IS PEAKS  GREATER THAN 0
C      CLASS 1 IS NON PEAKS LESS THAN 0
C    PATTERNS CONSIST OF THE 3 CHANNEL OUTPUTS AND THREE
C           FILTER OUTPUTS FROM EACH CHANNEL
C    EACH SET OF DATA IS READ FROM TAPE AND STORED ON DISK
C    THE MACHINE TRAINS UNTIL
C           1. SUCCESS IS GREATER THAN 99.9 PERCENT
C           2. DATA SWITCH 15 IS ON
C    DATA SWITCH 14 USES LAST WEIGHTS TO TEST TRAINING--NO
C           ADJUSTMENT IS MADE TO WEIGHTS
C    DATA SWITCH 12 CONTROLS PRINTING OF MISCLASSED PATTERNS
C    DATA SWITCH 13 ON TO CHANGE CONSTANT
C
C*************************************************************
      INTEGER MBUF(325),N(131),NP(13,10),TRY,NPTS(13),
     1 CORR(4)
      DIMENSION W(13)
      EQUIVALENCE (NP(1,1),N(1))
      DEFINE FILE 1(100,320,U,IKD1)
      TRY = 1
      LIM = 0
      WRITE (1,74)
      CALL TWAIM
      READ (6,75) CON
      READ (2,3) W
```

```
3       FORMAT (8F10.0/5F10.0)
        READ (2,41) X
        N(131) = 130
C*****OPEN DISK FILE
        CALL DOPEN (MBUF,1,13,24)
95      CONTINUE
C*****READ PATTERNS FROM MAGNETIC TAPE
96      CALL MAGOP (0,1,N(131),INDIC)
        DO 2 I=1,10
        DO 90 J=1,13
90      NPTS(J) = NP(J,I)
        IF(NPTS(13)-9999) 1,80,1
1       CONTINUE
        LIM = LIM + 1
C*****STORE PATTERNS ON DISK
2       CALL DPUT (MBUF,LIM,NPTS)
        GO TO 96
80      ITER = 0
65      ITER = ITER + 1
        WRITE (3,912) ITER,CON
912     FORMAT ('0BEGIN ITERATION',I6,' CONSTANT = ',F8.4)
102     DO 103 I=1,4
103     CORR(I) = 0
        DO 61 M=1,LIM
        CALL DGET (MBUF,M,NPTS)
        SUM = W(13)
        DO 6 I=1,12
6       SUM = SUM + W(I)*FLOAT(NPTS(I))
        IF(NPTS(13)) 7,7,8
8       IF(SUM) 14,9,9
C*****CLASS 1 FIND
14      I=3
```

```
      GO TO 60
C*****CLASS 1 FAILURE
9     I=1
      IF(IFLAG) 60,70,60
70    DO 10 J=1,12
10    W(J) = W(J)-CON*FLOAT(NPTS(J))
      W(13) = W(13)-CON
      GO TO 60
7     IF(SUM) 11,11,13
C*****CLASS 2 FIND
13    I=4
      GO TO 60
C*****CLASS 2 FAILURE
11    I=2
      IF(IFLAG) 60,71,60
C*****CHANGE WEIGHTS
71    DO 12 J=1,12
12    W(J) = W(J) + CON*FLOAT(NPTS(J))
      W(13) = W(13) + CON
60    CORR(I) = CORR(I) + 1
      IF(I-2) 942,942,61
C*****IF DATSW 12 ON WRITE MISCLASSED PATTERNS
942   CALL DATSW (12,IK)
      GO TO (89,61),IK
89    WRITE (3,97) NPTS(13)
97    FORMAT (1H ,'MISS CLASSED PATTERN ',I6)
61    CONTINUE
C*****PRINT TABULATION AFTER EACH ITERATION
      WRITE (3,5)
5     FORMAT (1H ,'ITERATION   CLASS   FINDS   ',
     1 FAILURES   PERC.RITE')
      PERC = FLOAT(CORR(3)+CORR(4))/FLOAT(LIM)*100.
```

```
      WRITE (3,15) ITER,CORR(3),CORR(1),CORR(4),CORR(2),PERC
15    FORMAT (1H ,3X,I3,8X,'1',2(4X,I5)/' ',14X,'2',2(4X,I5),
     1 8X,F6.1)
      WRITE (7,31) ITER,W
31    FORMAT (' ITERATION' I3,'  WEIGHTS ARE'/' ',13F9.3)
C   CONTROL FOR NEW CONSTANT
      CALL DATSW (13,I)
      GO TO (72,73),I
72    WRITE (1,74)
74    FORMAT ('ENTER CONSTANT XX.XXXX')
      CALL TWAIM
      READ (6,75) CON
75    FORMAT (F7.4)
C   CONTROL FOR END
73    CALL DATSW (15,I)
      GO TO (40,63),I
63    IF(PERC-99.9) 85,40,40
85    IFLAG = 0
C   CONTROL FOR NO WEIGHT CHANGE TESTING
      CALL DATSW(14,I)
      GO TO (87,65),I
87    IFLAG = 1
      GO TO 65
40    WRITE (2,41) W
41    FORMAT (8F10.4/5F10.4)
      CALL EXIT
      END
```

APPENDIX B

DATA CONVERSION FOR THE IBM 1800

The ECG data which was received from The University of Tennessee Medical Unit in Memphis was written on a seven track magnetic tape using an IBM 360 with a Disk Operating System. The seven track tape was written with even parity and an unblocked record length of 81 Binary Coded Decimal characters. The first record of each case contained the set or case number, the total number of points in the heart cycle and the point number of the P-wave onset. This identification record was written with a FORTRAN format of (5X,I5,15X,I5,15X,I5). Each succeeding record contained the ECG channel outputs and was written with a format of (15I5).

The seven track tape was read on an IBM 360 with an Operating System (OS) using these same FORTRAN formats and written on a nine track tape with odd parity. The nine track tape had a blocked record length of 80 bytes and 45 records to the block.

A check of the IBM 1800 operating system, Time-Sharing Executive, revealed that the option for IBM 360 magnetic tape label compatibility had been chosen so the IBM 360 tape could be physically read on the IBM 1800. The IBM 1800 has the capability for FORTRAN reads and writes but the two machines were incompatible because of FORTRAN array addressing arrangements. The IBM 1800 stores FORTRAN arrays from high core locations to low core locations. The IBM 360 does just the reverse.

For example, an array dimensioned as 80 on the IBM 1800 would place element one in core location say 100, whereas the IBM 360 would place the same element in core location 20. Also, the IBM 1800 allowed no control for blocking records in FORTRAN.

It was felt that the best procedure would be to read the IBM 360 tape by the MAGOP routine on the IBM 1800. MAGOP is the standard magnetic tape routine used on the IBM 1800 at Oak Ridge Associated Universities. It uses unformated core image reads and writes. By printing the values as they appeared in core under an I8 format, a conversion program could be designed to convert the data to a format useable on the IBM 1800. For example, two values on the IBM 360 tape which were read and printed on the IBM 1800 under an I8 format would appear as

-2832 16480 -3520 28820 16448

These words in memory could be seen to be the two's complement form of the hexidecimal numbers

F4F0 4060 F240 60F4 4040

When reversed, these numbers appeared as

4040 60F4 F240 4060 F4F0

which, using five bytes to the number, is the Binary Coded Decimal form of the numbers -42 and -40.

The conversion program literally reversed the blocks and by stripping and masking, obtained the data that was compatible with the IBM 1800. Another program was written to put the data into the standard format described in Chapter II. Added to the identifying information

in the first record were total number of points in the case and point number of the P-wave onset.

APPENDIX C

DESCRIPTION OF THE SIMULATOR PROGRAM

Program CHANE with link CHANF was used to simulate the filtering process in the detection of the P-wave onset described in Chapter III. To a large extent the form of the program was dictated by the configuration of the IBM 1800. The IBM 1800 has 24K, 16 bit words of memory. The operating system, Time-Sharing Executive, occupies approximately 12K of core. For speed of calculation, it was desirable to hold all three channel outputs from one case in memory. The 12K of variable core was not adequate to contain the entire simulator program. Therefore, the program was linked. Linking a program on the IBM 1800 simply means that the program is divided logically into as many parts or links as necessary to fulfill the task. Variables in COMMON are shared by all links. Each link must fit into the available variable core. The links are stored in object form on disk until called by the operating system to be loaded into variable core and executed.

The initial simulator program differed from the one listed below; however, the filtering process remained the same. The program as first written plotted the filter outputs on the IBM 1627 Plotter. Also, as use of the program progressed, it became desirable to insert many options for operator interaction. Data switches on the console and the IBM 1053 Typewriter were used to enable options.

The operator could select the set number of the data to be loaded from tape and could manually select the window size. The operator

could also select an initial set and the program would automatically apply all window sizes applicable and proceed to the next highest set number. It is noted that the window size had to be divisible by four because of the four equal intervals of summation.

As the visual plotter outputs became unnecessary, the program was changed to output the results of the filtering process onto another magnetic tape. The results were also printed on the second IBM 1053 Typewriter. In addition, as the means for picking the best window size became available, the program was used to evaluate the results of the method.

**********LISTING OF PROGRAM CHANE**********

```
C*************************************************************
C
C    PROGRAM CHANE IS LINK 1 OF THE FILTER SIMULATOR.  IT
C       HANDLES INPUT FROM TAPE OF SET DATA AND SETS UP
C       CONTROL FOR SELECTING WINDOW SIZE.
C    OPTIONS ARE BY DATASWITCH
C       DATSW 0--ON FOR AUTOMATIC SELECTION OF WINDOW SIZE
C       DATSW 3--ON FOR CALCULATIONS OF WINDOW FROM EQUATIONS
C       DATSW 4--ON FOR NO RESULTS PRINTED ON TYPEWRITER
C       DATSW 6--ON FOR FULL FILTER CALCULATIONS
C       DATSW 8--ON FOR WRITING ALL RESULTS TO MAG TAPE
C
C*************************************************************
      REAL Y(4)
      INTEGER CHAN(3,600),IBUF(18,10),MBUF(181),WIN(4)
      COMMON SAMP,FREQ1,IPTS,IPWAV,ISET,CHAN,IFLAG,
     1 IWIN,IBEGN,LIMST,WIN
      EQUIVALENCE (MBUF(1),IBUF(1,1))
      MBUF(181) = 180
      SAMP=.002
C*****TEST FOR AUTOMATIC CALCULATIONS OF WINDOW AND SET
      CALL DATSW(0,IK)
      GO TO (31,100),IK
C*****TEST FOR FIRST AUTOMATIC
31    CALL DATSW (2,IK)
      GO TO (107,32),IK
C*****IF ON FIRST AUTOMATIC, READ SET LIMIT
```

```
107    WRITE (1,108)
108    FORMAT ('ENTER SET LIMIT FOR THIS RUN--LLL')
       CALL TWAIM
       READ (6,2) LIMST
       GO TO 100
C*****IF ON AUTOMATIC, READ FIRST SET AVAILABLE
32     CALL MAGOP (1,1,MBUF(181),INDIC)
       IFLAG = 1
       IF(INDIC-1) 35,99,35
35     ISET = MBUF(1)
       GO TO 92
C*****  INITIALIZE AND FIND SPECIFIED SET
C*****NOT AUTOMATIC, GET DESIRED SET
100    WRITE (1,1)
1      FORMAT ('SSS')
       CALL TWAIM
       IFLAG = 0
       READ(6,2) ISET
2      FORMAT (I3)
C*****IF DESIRED SET IS NEGATIVE, PRINT HEADER ON TYPEWRITER
       IF(ISET) 97,99,96
97     WRITE (7,970)
970    FORMAT ('SET  CHAN  CYC PTS  CYC FREQ  WIND SIZE  '
      1'P-FREQ  P-PT  MIN  MINPT  DIFF  DELT')
       GO TO 100
C*****NOT AUTOMATIC, READ FIRST RECORD
96     CALL MAGOP (1,1,MBUF(181),INDIC)
       IF(INDIC-1) 91,95,91
C*****REWIND IF EOF
95     CALL MAGOP (1,5,MBUF(181),INDIC)
       GO TO 96
C*****TEST FOR SET DESIRED
```

```
91      IF(MBUF(1)-ISET) 93,92,93
C*****SKIP OVER IF NOT DESIRED SET
93      DO 3 I=1,1000
        CALL MAGOP (1,1,MBUF(181),INDIC)
        DO 44 J=1,10
        IF(IBUF(1,J)-9999) 44,96,44
44      CONTINUE
3       CONTINUE
C*****DETERMINE WINDOW SIZE
92      IPTS = MBUF(7)
C*****CALCULATE FREQUENCY
        FREQ1 = .92091/(SAMP*FLOAT(IPTS))
        CALL DATSW(6,IK)
        GO TO (1101,1102) , IK
C*****SET UP FOR FULL RANGE OF CALCULATIONS
1101    IWIN = IPTS/4*4
        IBEGN = 16
        GO TO 502
1102    CALL DATSW(3,IK)
        GO TO (501,502),IK
C*****SET UP FOR LINEAR FUNCTION CALCULATIONS OF WINDOW
501     IFLAG = -1
        Y(1) = .00452+1.08662*FREQ1
        Y(2) = .05406+2.37590*FREQ1
        Y(3) = .00862+1.17900*FREQ1
        Y(4) = .04403+2.19897*FREQ1
505     DO 632 J=1,4
        WIN(J) = .92091/(SAMP*Y(J))
632     WIN(J) = (WIN(J)+2)/4*4
        IWIN = WIN(1)
        IBEGN = 1
502     CONTINUE
```

```
      IPWAV = MBUF(8)
      NUM = 0
C*****READ CHANNEL OUTPUTS FROM TAPE
      DO 4 I=1,1000
      CALL MAGOP (1,1,MBUF(181),INDIC)
      DO 4 J=1,10
      IF(IBUF(1,J) - 9999) 6,94,6
6     NUM = NUM + 1
444   CHAN(1,NUM) = IBUF(1,J)
      CHAN(2,NUM) = IBUF(7,J)
      CHAN(3,NUM) = IBUF(13,J)
4     CONTINUE
C*****WRITE SET INFORMATION ON TYPEWRITER
94    WRITE (1,5) ISET,IPTS,IPWAV
      IF(ISET-LIMST) 997,99,99
997   CONTINUE
5     FORMAT ('SET',I4,' HAS',I4,'POINTS, ',
     1 'P-WAVE ON POINT',I4)
      CALL LINK(CHANF)
C*****TO TERMINATE, REWIND TAPES
99    CONTINUE
      CALL MAGOP (0,8,I,J)
      CALL MAGOP (0,6,I,J)
      CALL MAGOP (1,5,MBUF(181),INDIC)
      CALL EXIT
      END
```

```
**********LISTING OF PROGRAM CHANF**********


C*************************************************************
C
C    CHANF IS LINK 2 OF FILTER SIMULATOR PROGRAM.  IT
C       CALCULATES THE FILTER OUTPUT AND WRITES RESULTS TO
C       MAGNETIC TAPE.
C
C*************************************************************
      REAL ALPHA(3),BETA(3)
      INTEGER IA(21),MAX(3),CHAN(3,600),HOLD(3,1200)
      COMMON SAMP,FREQ1,IPTS,IPWAV,ISET,CHAN,IFLAG,
     1 IWIN,IBEGN,LIMST,KWIN(4)
      EQUIVALENCE (Q,IA(6))
C*****ENTER DESIRED WINDOW SIZE IF IFLAG = 0
      IF(IFLAG) 80,94,80
94    WRITE (1,7)
7     FORMAT ('WWW')
      CALL TWAIM
      READ(6,8) IWIN
8     FORMAT (I3)
      IF(IWIN) 80,100,80
100   CALL LINK(CHANE)
80    CONTINUE
      FREQ= .92091/(SAMP*FLOAT(IWIN))
      MM = 3
342   MM = 1
343   NUM = 0
C*****CALCULATE INITIAL ALPHA AND BETA VALUES
```

```
      N=IWIN/4
      M=IWIN/2
      LOW = IPTS-M+1
C*****FIRST FOURTH
      DO 11 I=1,N
      NUM = NUM + 1
      LOW = LOW + 1
      DO 11 J=1,MM
      HOLD(J,NUM) = CHAN(J,LOW)
11    ALPHA(J) = ALPHA(J) + FLOAT(CHAN(J,LOW))
      DO 12 J=1,MM
12    BETA(J) = ALPHA(J)
      K=N-1
C*****SECOND FOURTH
      DO 13 I=1,K
      NUM = NUM + 1
      LOW = LOW + 1
      DO 13 J=1,MM
      HOLD(J,NUM) = CHAN(J,LOW)
      BETA(J) = BETA(J)-FLOAT(CHAN(J,LOW))
13    ALPHA(J) = ALPHA(J) + FLOAT(CHAN(J,LOW))
      NUM = NUM + 1
      DO 14 J=1,MM
      MAX(J) = 0
      HOLD(J,NUM) = CHAN(J,1)
      BETA(J) = BETA(J)-FLOAT(CHAN(J,1))
14    ALPHA(J) = ALPHA(J) + FLOAT(CHAN(J,1))
      K=N+1
C*****THIRD FOURTH
      DO 15 I=2,K
      NUM = NUM + 1
      DO 15 J=1,MM
```

```
      HOLD(J,NUM) = CHAN(J,I)
      ALPHA(J) = ALPHA(J)-FLOAT(CHAN(J,I))
15    BETA(J) = BETA(J)-FLOAT(CHAN(J,I))
C*****LAST FOURTH
      DO 16 I=1,N
      NUM = NUM + 1
      K=K+1
      DO 16 J=1,MM
      HOLD(J,NUM ) = CHAN(J,K)
      ALPHA(J) = ALPHA(J) - FLOAT(CHAN(J,K))
16    BETA(J) = BETA(J) + FLOAT(CHAN(J,K))
      K=K+1
C*****FILL IN REST OF HOLD
      DO 17 I=K,IPTS
      NUM = NUM + 1
      DO 17 J=1,MM
      HOLD(J,NUM) = CHAN(J,I)
17    CONTINUE
      DO 18 I=1,M
      NUM = NUM + 1
      DO 18 J=1,MM
18    HOLD(J,NUM) = CHAN(J,I)
      I1 = 1
      I2 = M+1
      I3 = IWIN+1
      I4 = N+1
      I5 = N*3+1
C*****CALCULATE SUCCEEDING ALPHAS AND BETAS
      DO 20 I=1,IPTS
      DO 23 J=1,MM
      P=SQRT(ALPHA(J)*ALPHA(J) + BETA(J)*BETA(J))
      ALPHA(J)=ALPHA(J)+FLOAT(-HOLD(J,I1)+2*HOLD(J,I2)
```

```
     1-HOLD(J,I3))
      BETA(J)=BETA(J)+FLOAT(-HOLD(J,I1)+HOLD(J,I3)
     1 +2*HOLD(J,I4)-2*HOLD(J,I5))
23    HOLD(J,I1) = P
      I1 = I1 + 1
      I2 = I2 + 1
      I3 = I3 + 1
      I4 = I4 + 1
      I5 = I5 + 1
20    CONTINUE
C*****FIND MINIMUM
      DO 199 J=1,MM
      Z=32000
      ALPHA(J) = 0
      BETA(J) = 0
      DO 403 I=1,IPTS
      Q = HOLD(J,I)
      IF(Z-Q) 403,403,301
301   K=I
      Z=Q
403   CONTINUE
C*****FIND DISTANCE FROM P-WAVE ONSET
402   I1 = K-IPWAV
      IF(I1) 43,43,40
40    I2 = IPTS-K+IPWAV
      IF(I2-I1) 42,43,43
42    I1 = -I2
43    Q=I1*SAMP
C*****DATSW 4 ON WRITE RESULTS TO TYPEWRITER
      CALL DATSW(4,N)
      IF(N-1) 31,32,31
31    WRITE (7,303) ISET,J,IPTS,FREQ1,IWIN,FREQ,
```

```
     1 IPWAV,Z,K,I1,Q
303   FORMAT (I3,4X,I1,5X,I3,6X,F6.4,5X,I3,3X,F8.4,3X,I2,
     1 1X,F6.0,2X,I3,2X,I5,2X,F6.3)
32    Z = ABS(Q)
C*****DATSW 8 ON TO WRITE ALL RESULTS TO TAPE
      CALL DATSW(8,IK)
      IF(IK-1) 277,276,277
277   IF(Z-.0041) 276,199,199
276   IA(1) = ISET
      IA(2) = IPTS
      IA(3) = IWIN
      IA(4) = J
      IA(5) = K
      IA(6) = I1
      IA(7) = IPWAV
      IA(21) = 20
      CALL MAGOP(0,3,IA(21),INDIC)
199   CONTINUE
405   IF(IFLAG) 107,94,101
C*****LINEAR CALCULATIONS OF WINDOW SIZES
107   IBEGN =IBEGN + 1
      IF(IBEGN-4) 108,108,109
109   IFLAG = 1
      GO TO 100
108   IWIN = KWIN(IBEGN)
      GO TO 343
C*****CONTINUATION FOR RANGE OF WINDOW SIZES
101   IWIN = IWIN-4
      IF(IWIN-IBEGN) 100,80,80
      END
```

VITA

Martha Carol Snyder Hansard was born in Kingsport, Tennessee on January 1, 1947. She attended Kingsport City Schools and graduated from Dobyns-Bennett High School in 1965. She enrolled in The University of Tennessee in the fall of 1965 and graduated with a Bachelor of Science degree in Mathematics in March 1969. In May of that year she accepted a position as a Statistical Quality Control Analyst with the Univac Division of Sperry Rand Corporation in Bristol, Tennessee. In the fall of 1970 she enrolled in the Computer Science Graduate Program at The University of Tennessee, and began teaching and consulting for The University of Tennessee Computer Center. In the fall of 1971 she accepted a graduate assistantship with The University of Tennessee to teach and program in the Statistical Applications Group. In December 1971 she accepted a position as a Programmer/Analyst with Oak Ridge Associated Universities in Oak Ridge, Tennessee. She received a Master of Science with a major in Computer Science from The University of Tennessee in March 1973. She is presently Supervisor of Computer Applications at Oak Ridge Associated Universities.