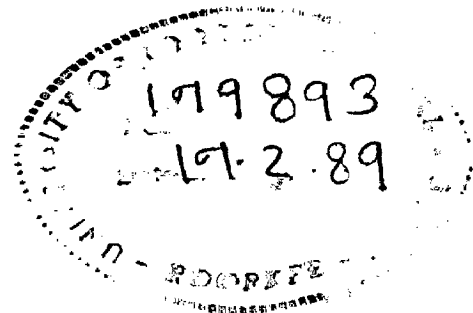# DATA COMPRESSION OF ELECTROCARDIOGRAM

A DISSERTATION

Submitted in partial fulfilment of the
requirements for the award of the degree
of
MASTER OF ENGINEERING
in
ELECTRICAL ENGINEERING
(Measurements and Instrumentation)

by

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CANDIDATE'S DECLARATION

I hereby certify that the work which is being presented in the dissertation entitled - **DATA COMPRESSION OF ELECTROCARDIOGRAM** in partial fulfilment of the requirement for the award of degree of **MASTER OF ENGINEERING** in **ELECTRICAL ENGINEERING** with specialization in **MEASUREMENTS AND INSTRUMENTATION**, submitted in the Department of Electrical Engineering, University of Roorkee, Roorkee is an authentic record of my own work carried out during a period of about seven months from January 1988 to the beginning of August 1988, under the supervision of **Dr. S.C. Saxena**, Professor, Electrical Engineering Department, University of Roorkee, Roorkee.

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

Ishan Ranjan

Dated : August 8,1988
(ISHAN RANJAN)

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

Dated : August 8 , 1988

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ABSTRACT

ECG is a graphic record of the electrical potentials produced by the electro-mechanical activity of the heart. ECG signal and its parameters have to be detected from various time varying or nonstationary physiological signals in a random noise background. The need for ECG data compression arises in order to extract relevant information from a plethora of cardiographic data. ECG data compression entails the reduction in the number of sample points required to generate an ECG wave without losing its information content. Data compression is required for various reasons which include the economic utilization of storage space for ECG data bases and the transmission of digitized ECGs to an ECG processing and recording establishment via telephone etc.

There are various techniques of data compression which are applied in different clinical situations. In this study the "Template method" for data compression has been used. It is a logical method and involves making a template of an ECG wave which then forms the basis of comparison with subsequent data waves ; record being stored only when the deviation is greater than a certain limit say ten percent.

The "Moving Sign Slope Method" which is a straight line approximation of an ECG wave has also been discussed. Data compression here is achieved by approximating the ECG wave by straight lines within a corridor which is defined by the maximum allowable limit.

The results show that the Template method yields the greatest advantage in a clinical situation where the ECG wave is not deviating significantly from its initial characteristics, whereas the Moving Sign Slope Method, being a statistical method of data compression results in large data compression with good wave reconstruction fidelity only when the allowable error limit is one percent.

ACKNOWLEDGEMENT

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CHAPTER - 1

INTRODUCTION

1.1 INTRODUCTION

The electrocardiogram (ECG) is a graphic record of the electrical potentials produced by the electromechanical activity of the heart. The heart is unique among the muscles of the body in that it possesses the property of automatic rhythmic contraction. The impulses that precede contraction arise in the conduction system of the heart. These impulses result in the excitation of the muscle fibres throughout the myocardium. Impulse generation and conduction produce weak electrical currents that spread throughout the body. By placing electrodes at various defined positions on the body and connecting these electrodes to an amplifying and processing unit, the ECG is recorded. This complete system is called the ECG machine. The connections of the machine are such that an upright deflection indicates positive potential and a downward deflection indicates a negative potential.

The constant changes in the electrical potential, as the electrical impulse passes through the conduction system and activates the myocardium, are clearly reflected in the various segments of the ECG wave.

The pattern in any one lead of the normal ECG, will depend on the relationship of it's lead axis to the cardiac vector. The patterns will therefore vary with the change in electrode positions with respect to the heart. The ECG consists of atrial and ventricular depolarization and repolarization complexes whose magnitude and duration reflect the state of the heart.

The accuracy of cardiac diagnosis as a result of advances in electrocardiographic research has greatly increased the need for data compression.

Apart from this, the other various reasons for data compression of ECG signals include the economic utilization of storage space for ECG data bases; the transmission of digitized ECGs to an ECG processing and recording establishment via telephone and to enhance the efficiency of computerized analysis of the digitized ECGs.

1.2 ELECTO CARDIOGRAM

The ECG wave segments are related to the electrical activity of the heart as follows :

P wave : The deflection produced by atrial depolarization

Q wave : The initial negative deflection resulting from ventricular depolarization. It precedes the first position deflection (R).

R wave : The first positive deflection during ventricular depolarization.

S wave : The first negative deflection of ventricular depolarization that follows the first positive deflection (R).

T wave : The deflection produced by ventricular repolarization.

U wave : A deflection (usually positive) seen following the T wave and preceding the next P wave. The exact cause of this wave is unknown. It is currently thought to be the result of the slow repolarization of the intraventricular (Purkinje)conduction system.

1.3   NORMAL INTERVAL VALUES

1.3.1   R-R Interval

The R-R interval is the distance between two successive R waves. If the ventricular rhythm is regular, the interval in seconds (or fraction of a second) between the peaks of 2 successive R waves divided into 60 (seconds) will give the heart rate per minute. If the ventricular rhythm is irregular, the number of R waves in a given period of time (e.g. 10 seconds) can be counted and then converted into the number per minute. For example, if 20 R waves are counted in a 10 second interval, the ventricular rate is counted as 120 per minute.

1.3.2   P-P Interval

In regular sinus rhythm, the P-P interval will be the same as the R-R interval. However when the ventricular rhythm is irregular or when atrial and ventricular rates are different but regular, the P-P interval should be measured from the same point on two successive P waves and atrial rate per minute be computed in the same manner as the ventricular rate.

1.3.3   P-R Interval

This measures the AV conduction time. It includes the time required for atrial depolarization and part of atrial repolarization (the late phase of atrial repolarization occurring during ventricular depolarization), plus the normal delay of excitation (approximately 0.07 second) in the AV node. It is measured from the on set of the P wave to the beginning of the QRS complex. The normal value is in the range of 0.12-0.20 seconds. This must be correlated with the heart rate, normally, the slower the heart rate, the longer the P-R interval. A P-R interval of 0.2 second may be of no clinical significance

with a heart rate of 100. The values vary also with age and the build of the body.

1.3.4 QRS Interval

This is the measurement of 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 of the normal value is 0.1 second.

Ventricular Activation Time (VAT) : It is the time taken by an impulse to traverse the myocardium from the endocardial to the epicardial surface.

Q-T Interval : This is measured from the onset of the Q wave to the end of the T wave. It measures the duration of the electrical systole. The Q-T intervals varies with the heart rate and must be corrected.

Q-U Interval : This measures the interval from the beginning of the Q wave to the end of U wave. It has no known clinical significance.

1.4 ELECTROCARDIOGRAPHIC LEADS

Twelve lead ECG system is a generally used to monitor the performance of the heart. The twelve leads of the ECG system are as follows :

Bipolar Standard Leads

The bipolar standard leads (I, II and III) are the original leads selected by Einthoven to record the electrical potentials in the frontal plane. Electrodes are applied to the left arm, right arm and left leg. Proper skin contact must be made by rubbing electrode paste on the skin. The LA (left arm), RA (right arm) and LL (left leg) leads are then attached to their respective leads

as shown in Fig. 1.1. By turning the selector switch to I, II and III position, the record of three standard leads (I, II and III) is obtained.

All electrocardiographic machines also have a right leg electrode lead. This acts as a ground wire and plays no role in the recording of the ECG. In areas where there is electrical interference, it may be necessary to run a ground wire from the bed or the machine to an appropriate ground.

Electrical Potential

The bipolar leads represent a difference of electrical potential between two selected sites as shown in Fig. 1.2.

Lead I   -   Difference of potential between the left arm and the right arm (LA - RA)

Lead II   -   Difference of potential between the left leg and the right arm (LL - RA)

Lead III   -   Difference of potential between the left leg and left arm (LL - LA).

The relation between the three leads is expressed algebrically by Einthoven equation:

Lead II = Lead I + Lead III

This is based on Kirchoff law, which states that the algebric sum of all the potential differences in a closed circuit equals zero. If Einthoven had reversed the polarity of lead II (i.e. RA - LL), the three bipolar lead axes would results in a closed circuit and leads I + II + III would equal zero. However, since Einthoven did not make this alteration in the polarity of lead II axis, the equation becomes,

LA Lead

RA Lead

LL Lead

ECG

Move selector switch
to desired lead
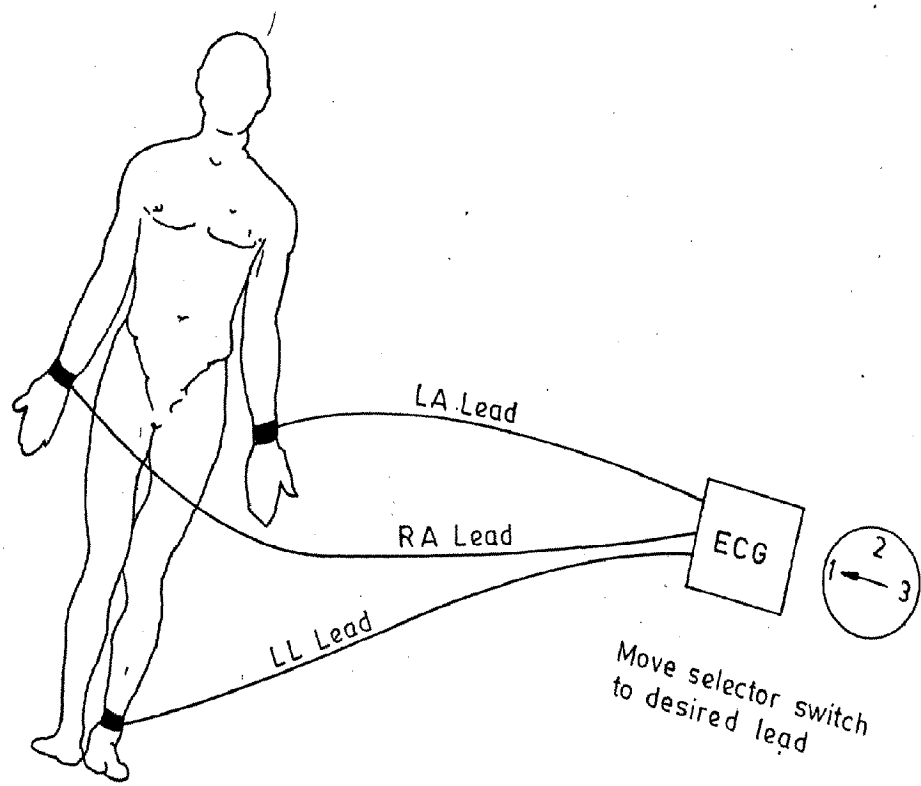
Fig.1.1 Illustration of the Standard Leads

To ECG Recorder

− I +

RA Lead

RA Lead

I

II III

RA Lead

LA Lead

−

II

+

To ECG
Recorder

To ECG
Recorder

−

III

+

LL Lead

LL Lead
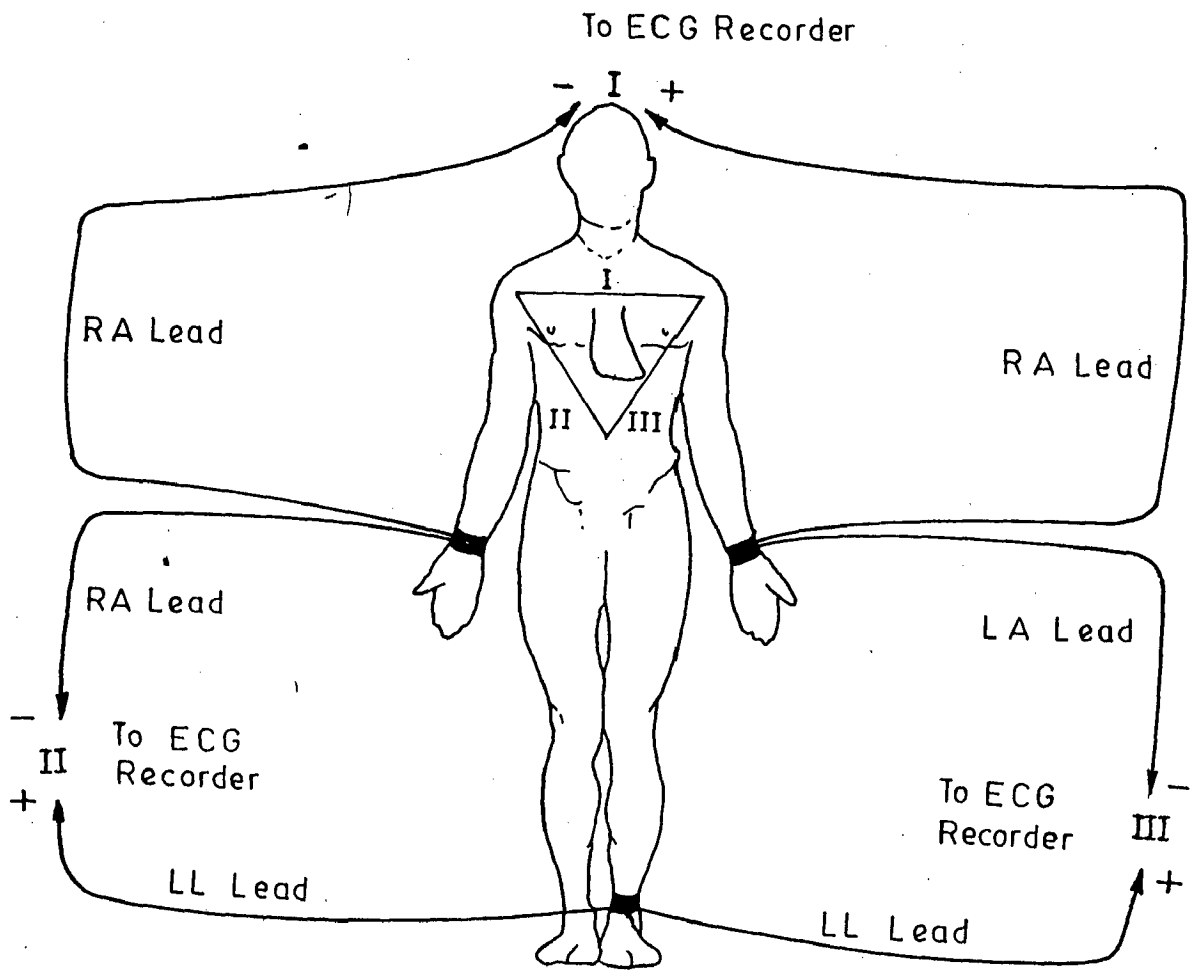
Fig. 1.2 Bipolar Leads-Difference of Potential Between
Selected Sites

I - II + III = 0   Hence   II = I + III

The electrical potential as recorded from any one of the locations will be the same no matter where the electrode is placed on the extremity. The electrodes are usually applied just above the wrists and ankles. If an extremity has been amputated, the electrode can be applied to the stump.

Bipolar Chest Leads

Bipolar chest leads record the difference of potential between any given position on the chest (C) and one extremity. Before unipolar electrocardiography was introduced, the left leg (R) was used as the indifferent electrode and the leads were called CF leads; less commonly, the right arm (CR leads) or the left arm (CL leads) were used. It was assumed that the left leg (or right or left arm) were so remote from the heart that it would act as an indifferent electrode and not interfere with the chest potential. However, it is now realized that the potentials in the extremeties can appreciably alter the pattern of the chest lead. For this reason CF, CR and CL leads are not frequently used these days.

Unipolar Leads

(i)   Extremity leads

(ii)   Precordial (chest) leads

(iii)   Esophageal leads

Unipolar leads (VR, VL, VF, multiple chest leads 'V' and esophageal leads 'E') were introduced into clinical electrocardiography by Wilson in 1932.

The frontal plane unipolar leads (VR, VL, VF) bear a definite mathematical relationship to the standard bipolar leads (I, II, III).

$$VR = -\frac{I + II}{3}$$

$$aVR = -\frac{I + II}{3} \times \frac{3}{2} = -\frac{I + II}{2}$$

$$VL = \frac{I - III}{3}$$

$$aVL = \frac{I - III}{3} \times \frac{3}{2} = \frac{I - III}{2}$$

$$VF = II + III/3$$

$$aVF = \frac{II + III}{3} \times \frac{3}{2} = \frac{II + III}{2}$$

The unipolar nonaugmented extremity leads have been replaced by the augmented extremity leads aVR, aVL and aVF. The only difference between leads VR, VL and VF and leads aVR, aVL and aVF is the difference in amplitude. In routine electrocardiographic practice, the augmented leads have replaced the nonaugmented unipolar extremity leads as they are easier to read.

The precordial leads (V_1 - V_6) record potentials in the horizontal plane without being influenced by actual potentials from an 'indifferent' electrode used in recording bipolar chest leads as shown in Fig. 1.3. A unipolar precordial lead (or esophageal lead) does not record only the electrical potential from a small area of the underlying myocardium but records all of the electrical events of the entire cardiac cycle as viewed from a selected lead site. Figure 1.4 shows the location of all the leads of the twelve lead system.

Although it is possible to use any lead (or multiple simultaneous leads if equipment is available) in a specialized area such as coronary care unit, it is more common to use a modified bipolar chest lead. In this work we have selected the lead II for processing and data compression. Lead II has been selected out of all the available leads, as only in lead II the P wave
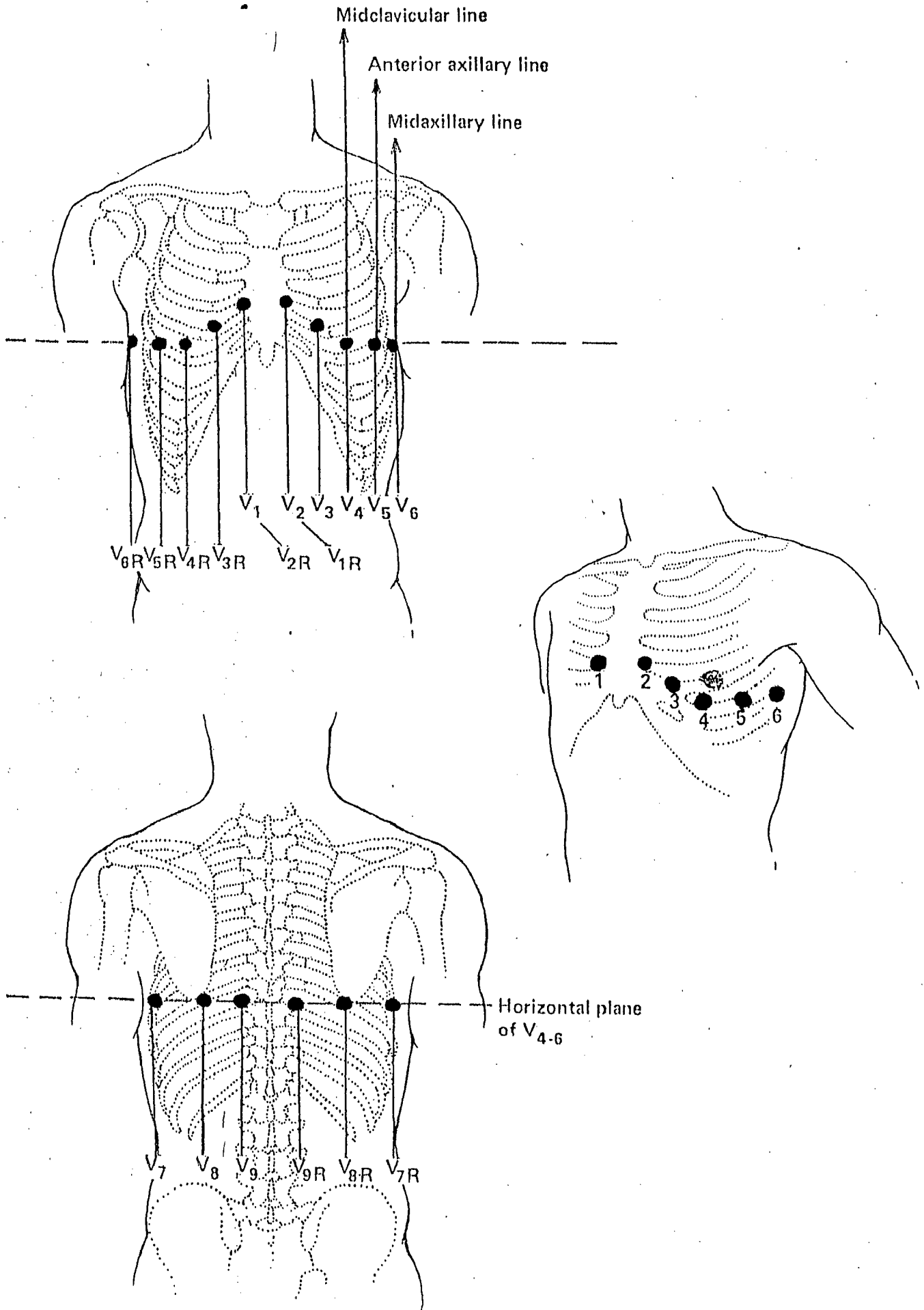
Fig. 1.3 Illustration of Precordial Leads and their Location
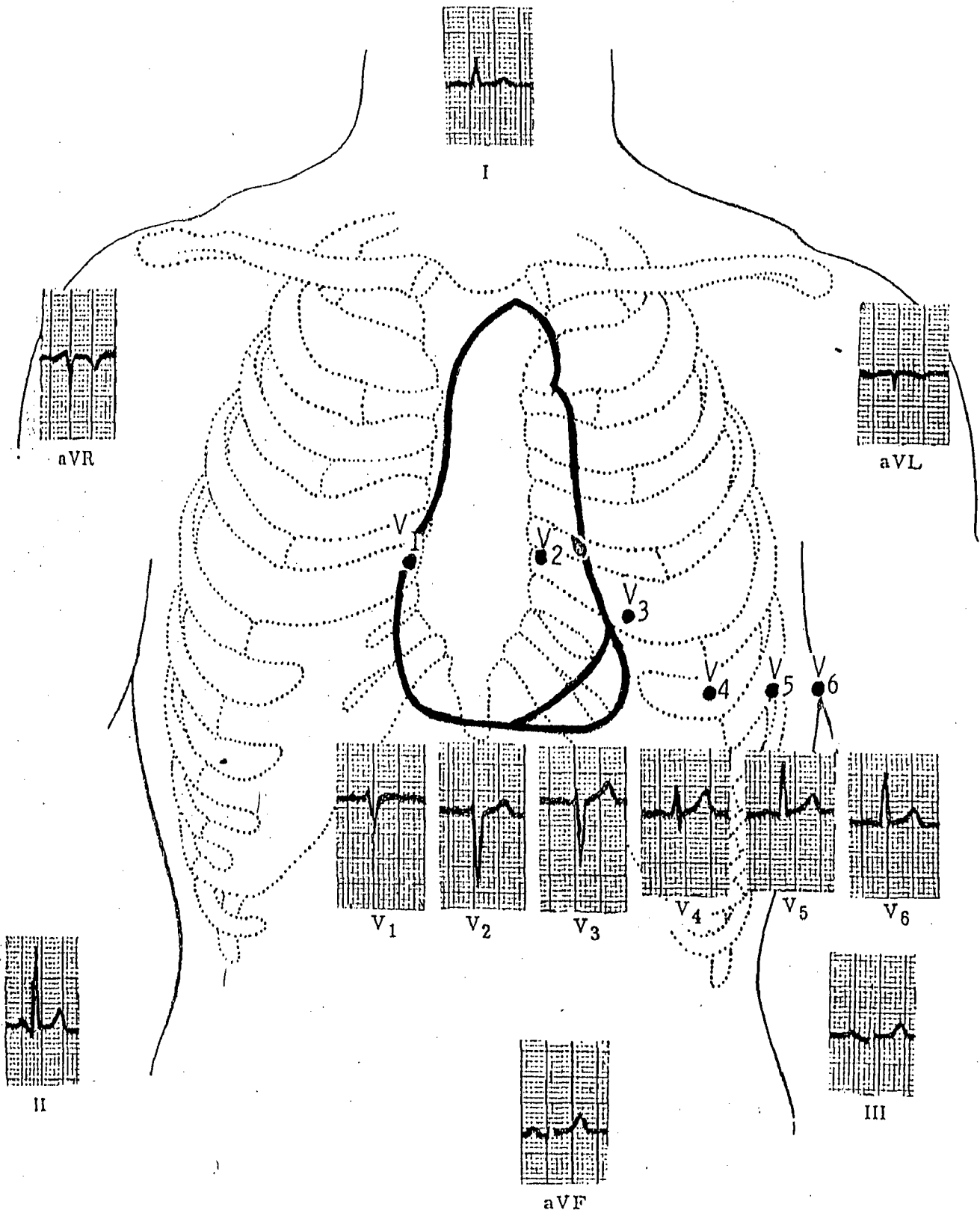
Fig. 1.4 Location of all the Leads of the Twelve Lead System

is indicative of atrial depolarization prominently and all other wave segments

i.e. the QRS complex and T wave duration clearly marked, as is evident

in Fig. 1.5.

1.5 DATA COMPRESSION OF ECG SIGNALS

The increased use of computer assisted interpretation of the electrocardiogram has resulted in a great deal of interest in compact digital representation of ECG records. An efficient and reversible compression of ECG tracings is necessary to store large quantities of cardiograms in machine readable form, enabling serial comparison, and the desire to transmit cardiograms in digital form via telephone lines to centralized processing centres.

Many schemes have been proposed in literature to achieve data compression/reduction of the ECG. However all the schemes can be grouped as under :

(A)   ECG data compression by reduction of sampling rate or shortening of word length :

Although this is the most obvious solution to reduce the bit rate, it is rarely applicable. It is mostly applied in patient monitoring where ECG signal is replaced by a signal at unequal samples, only measured if a certain threshold of first or second factor's derivative is crossed.

(B)   ECG data compression using orthogonal functions :

In this case, only the weightings or coefficients of the individual orthogonal functions would have to be transmitted. Well known of the orthogonal basis functions is the Karhunen-Loeve expansion or a Chebyshev transforms.

$$aVR = -\frac{I + II}{2}$$

$$I = 2/3\ (aVL - aVR)$$

$$aVL = \frac{I - III}{2}$$

$$II = 2/3\ (aVF - aVR)$$

$$III = 2/3\ (aVF - aVL)$$

$$aVF = \frac{II + III}{2}$$
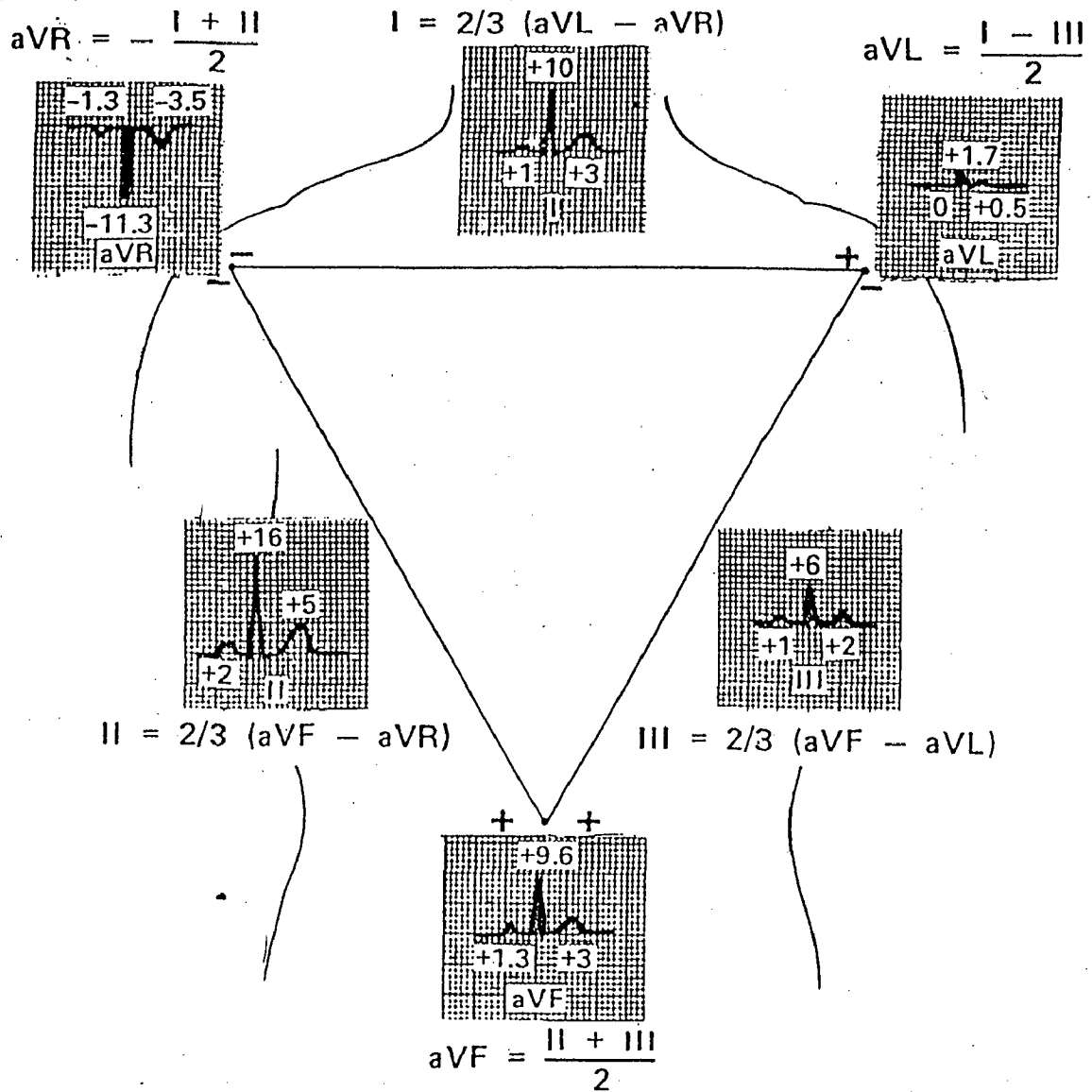
Fig. 1.5 The ECG Waves Recorded in Various Leads

(C) Data compression by prediction and interpolation :

A technique for transmission of the difference between a predicted value and the actual value of the ECG is used for data compression. The scheme that is employed predicts a new ECG sample from several of the preceding samples. The predicted value will rarely be exactly the same as the actual value and the difference between the two, which represents the prediction error, would have to be transmitted. This error is usually very small, but in some cases it may also be vary large. In order to efficiently transmit the error signal, an optimal encoding scheme is required.

(D) ECG data compression using template method :

This involves transmission of the signal or it's storage only if the ECG has chahged by a predetermined value . The time when it occurs and the value of it have to be transmitted. Again application of an optimal encoding scheme would minimize the bit rate.

(E) ECG data compression by straight line approximation :

This scheme involves representation of a set of samples lieing within a predetermined zone by a sample with constant slope. The transmission of vertex points occurs only if the straight line approximation is not being met by the previous slope line.

Out of the five different data reduction strategies discussed in literature, this work focuses on two different data reduction strategies - the first one's the template method, the scheme in which the transmission and storage takes place only if the ECG changes by a predetermined value and the second one is the straight line approximation method, in which the ECG curve is approximated by straight lines.

In the former scheme, data compression would be tremendous if the data ECG waves do not vary a great deal from the template ECG wave. On the other hand, in the straight line approximation method, the extent of data compression shall depend more on the nature of the ECG wave. Thus both the schemes find use in markedly different situations.

In this work an attempt has been made to put into clear focus the advantages of data compression of ECG signals with regards to storage and transmission. The first chapter deals with the introduction to the ECG signal and the need for ECG data compression. Chapter II deals with a review of the various data compression techniques and their relevance to the ECG wave. Chapter III presents a template method for ECG data compression. The philosophy of the method alongwith the algorithm so developed are discussed. Chapter IV outlines a straight line approximation scheme for data compression. The conclusions and suggestions for future work are given in Chater V.

CHAPTER - II

DATA COMPRESSION TECHNIQUES AND
THEIR RELEVANCE TO ECG SIGNALS

2.1   USEFULNESS OF THE ELECTROCARDIOGRAM

The ECG is of particular value in the following clinical conditions:

(1)   Atrial and ventricular hypertrophy

(2)   Myocardial infraction : unipolar extremity leads, multiple chest leads, esophageal leads, and vectorcardiograms have greatly increased the incidence of correct diagnosis.

(3)   Arrhythmias : Not only can more exact diagnoses be made, but unipolar and intracardiac electrocardiography have also contributed substantially to the basic understanding of the origin and conduction of abnormal rhythms.

(4)   Pericarditis

(5)   Systemic diseases that affect the heart

(6)   Effect of cardiac drugs

(7)   Disturbances in electrolyte metabolism, especially potassium abnormalities.

The ECG is a laboratory test only and is not a sine qua non of heart disease diagnosis. A patient with an organic heart disorder may have a normal ECG, and a perfectly normal individual may show nonspecific electrocardiographic abnormalities. Very often a patient is assigned the status of a cardiac invalid solely on the basis of some electrocardiographic abnormality. On the other hand, a patient may be given unwarranted assurance of the absence of heart disease solely on the basis of a normal ECG. The ECG must always be interpreted in conjunction with the clinical findings.

2.2 NEED FOR DATA COMPRESSION

Computer application to electrocardiography has resulted in great attention being focussed on 'compression of digital electrocardiograms'. Compression of digital electrocardiograms is desirable for the following three reasons:

(A) The need for increased efficiency in computerized rhythm analysis systems:

Data compression is very effective in increasing the efficiency of automatic computer analysis of electrocardiograms [4 - 9]. Especially under the constraints of a microprocessor based system, efficient data compression preprocessing is critical to real time operation.

(B) The need for economic use of storage space for data bases/permanent storage of digitized ECGs:

Compact data storage with easy retireval, without introducing clinically significant distortion, is important for reducing data storage expense. Long term digital recording and 'real time analysers' are also greatly benefited from the greatly increased storage space capacity on the instruments/recorders magnetic tape [13].

(C) The need to transmit multichannel digitized electrocardiograms to an'ECG processing and recording centre over telephone lines:

To trahsmit digitized ECG data from various locations to a centre for further processing a data compression scheme is necessary if public telephone lines are to be used economically [10 - 11].

The need for digital transmission of cardiograms is the result of noise problems associated with transmitting analog ECGs in the form of FM signals

over the telephone lines, while the advantages of a digital data base are all based on the use of a computer to analyze the data subsequently. Although, in theory, man can certainly perform these functions, the volume of repetitive work required is prohibitive in many cases.

The main problems associated with the digital transmission and storage of ECGs is the large number of bits required to represent the signals adequately. The proposed solutions to this problem are all based on taking advantage of both the large sample to sample redundancy and the large beat to beat redundancy associated with cardiograms. The techniques range from sending and storing samples only when a change exceeding a certain threshold occurs, through using time series to represent the beats, to using a transform which reduces quantization redundancy, and using Huffman encoding techniques.

2.3   TECHNIQUES FOR DATA COMPRESSION

Electrocardiograms offer different possibilities for considerable data reduction required for digital transmission, compact recording as in ambulatory monitoring or long term storage for serial comparison of waveforms.

Depending upon the end use for which the compressed ECG waveform has to be put, the data compression of ECG waveforms can be divided into the following categories:

2.3.1 Reduction of Sampling Rate or Shortening of Word Length [23]:

The limitation of the bandwidth of the ECG signal, permitting a corresponding reduction of the sampling data   is an important step in the technique of data compression. However, there is a major inherent disadvantage in

that the signal is degraded. The reduction of sampling rate is achieved by digital bandpass filters of a nonrecursive type to guarantee a linear phase response:

The signal degradation resulting from the bandwidth limitation achieved by four digital lowpass filters with half amplitude bandwidths of 100,80,60 and 40 Hz was investigated by 'Barr and Spach'[15]. The filters were of the nonrecursive type to guarantee a linear phase response, and Kaiser data windows were carefully designed to avoid undesirable ringing.

The effects of signal degradation were assessed by noticing the changes induced in a set of 16 clinically meaningful measurements. The effected changes were compared with the spontaneously occuring beat to beat variations of the measurements as observed within the 10s record. In this way, the effects of bandwidth limitation were assessed in terms of the actual measurement variability noted from each patient's original recording.

In general, the results show an increased degradation of the amplitude measurements with decreased bandwidth, whereas time measurements were not adversely affected. For a bandwidth of 100 Hz, only one of 220 recorded the fidelity criterion, and the mean differences of amplitude computed from the pair beats of the original versus the filtered records, were less than 10 μV. Hence it was concluded that a bandwidth of 100 Hz was sufficient for practical quality demands, permitting the selection of a minimum rate of 200 Hz.

Further compression of ECG without any loss of signal fidelity can be achieved by removal of some of the redundancy within the sequence of the digitized signal samples which is due to one or both of two reasons:

(i)   neighbouring samples are not statistically independent, and

(ii)   quantized signal amplitudes do not occur with equal probability.

The redundancy resulting from these two different reasons can be partly removed by separate techniques, which could be used together to achieve a large data compression ratio.

In general results show an increased degradation of the amplitude measurements with decreased bandwidth, whereas, the time measurements were not adversely affected.

The amount of data to be stored in a data base depends on the data retrieval requirements.

This method is also rather inefficient for contour classification. In case if only average complex has to be stored and not the original signal (Jokinen 1980) then reduction of sampling rate or shortening of word lendth would be a useful technique.

2.3.2 Compression of ECG Using Orthogonal Functions [24] :

The ECG is expressed as series of orthogonal functions. A number of methods have been used in literature to represent the ECG like sine and cosine, fourier components, unit impulses, and matched exponentials.

It has been found that most of these orthogonal functions can be used to identify signular patterns and specific leads. Each segment of an ECG wave is specified by an orthogonal function. In this case only the weighting factors or coefficients of the individual orthogonal functions would have to be transmitted.

Other techniques for the compression make use of a series of orthogonal basis functions for the reconstruction of waveform. Well known is the Karhunen-

Loeve (K - L) expansion or a Chebyshev transform. The first methods yielding eigenvectors, was evaluated by Womble (1977) together with reduction by spectral techniques. Such methods do not take into account the semantic information comprised in the ECG. For that reason, they are most helpful in detecting trends in intervals of sudden changes in waveshapes in individual patients.

As is known, if one wants to represent waveforms from separate random processes with a global truncated series, the optimal series (with respect to mean-square-error, entropy, etc) is the Karhunen-Loeve series with the functions in the series determined from a global covariance function. The global covariance function is merely the weighted sum of the covariance functions of the separate random processes, where the weightings are merely the a priori probability that a waveform will be from that particular random process.

Womble, used the Karhunen-Loeve expansion for simulating an ECG wave in order to achieve data compression. First each beat was assigned to a broad class of beats by correlating it's spatial velocity curve with representative curves. Next, each beat was segmented into two parts, P and QRST, and ultilizing K-L functions desired for each broad class of cardiograms, a separate Karhunen-Loeve (K - L) expansion was performed on each segment.

The Karhunen-Loeve series coefficients were considered to be a time series. They were transformed to minimize quantization redundacny via Huffman encoding techniques.

The K-L expansion is a statistical approach and requires a knowledge of the covariance function of each random process involved. The required covariance functions are estimated from a data base of electrocardiograms

containing representative samples of each random process. The K-L expansion performs much better for highly nonstationary signals for which various inflections and waves occur in narrow time slots. In fact it has been shown that, as the periodic signals approach a stationary random process, the K-L functions approach sine waves and K-L expansion becomes the fourier expansions.

Since for the global K-L expansion, the P wave location is more variable than that of the QRST, one would expect the expansion to perform much better on the QRST segment of the beat.

However for contour classifications they are rather inefficient since, tiny Q waves may be missed by the fact that their signal power is less than the distortions allowed, if integrated over the duration of the wave. This is the reason why most long term storage systems store either samples of the dominant beat or even the entire recording eventually sampled at 200 Hz. Another reason is the fact that the technical means for inexpensive storage and retireval have gradually diminished the need for data reduction algorithms that are always more or less increasing the signal entropy.

2.3.3 Template Method for Data Compression of ECG Signals [25]:

This method for data compression achieves data reduction by first making a template of the data ECG waves - find amplitude, duration and from them slope etc of various complexes that make up the ECG wave.

This method for data compression/reduction of electrocardiograms offers good possibilities for considerable data required for compact recording as in ambulatory monitoring or long term storage for seriesl comparison of ECG waves.

The template method for data compression of ECG signals has found use in monitoring of the cardiogram when refrigeration prior to surgery is done. With the use of refrigeration in surgery, it is essential to be aware of the changes that will occur electrocardiographically.

Bradycardia and prolongation the Q-T interval are to be expected at temperatures of 30°C. Various atrial arrhythmias and heart block are also not uncommon. Fatal arrhythias, cardiac arrest, or ventricular fibrillation may occur during the induction period of refrigeration or at temperatures for below 30°C. It is, therefore, essential to have constant electrocardiographic monitoring.

In ECGs interpretation, the wave shapes are classified either by logical methods or by statistical techniques. The main problem in ECG interpretation that arises is that a one to one relationship is not always being able to be established between the ECG and the state of the heart. Hence a template which is specific to a particular person and is made by considering an initial span of 10 ECG waves of an individual is able to use the ECG parameter values logically - rather than basing it's decision on the state of the heart statistically only. This feature, that provides diagnostic power - computed by nonlinear combinations of the syntactic basic components such as products and ratios or time intervals which may be related to biological events and phenomenon is very important.

The template method involves making a detailed template of the ECG so that an ECG wave can be characterised by the parameters. These parameters are useful for compact storage and as far as serial comparison is concerned, they are used for comparing two ECG waveforms.

Data reduction in this method is dependent on the type of ECG wave template and the clinical situation for which it is being used. In a normal monitoring situation, where the patients ECG is not subjected to great fluctuations, data compression/reduction is very large as the variation in the template once made is not very large.

However in an ambulatory situation where the ECG is changing rather rapidly different templates are stored as variation of contour parameters keeps changing rather rapidly and so data reduction effected is not very large.

Serial comparison of ECG is possible in this type of data reduction technique. For serial comparison, the ECG wave is quantified into clinically important parameters and wave contours are classified and compared.

This technique can also be used for data transmission but the entire template information shall have to be transmitted.

2.3.4 Data Compression by Interpolation and Prediction with Entropy Encoding [10]:

One motivation for the study of data compression stems from the need of teletransmission of cardiograms over the telephone network. With the currently often used practice of employing frequency modulation (FM) techniques to transmit cardiograms in analog form, the quality of the received signal varies in an unpredictable way and may sometimes be quite low while these occasional equal impairments do not seriously hamper the visual cardiographic interpretation, they do lower the reliability of computer interpretation programs.

A solution to this problem is provided by digitizing the cardiograms at the site they are taken and transmitting the digitized signals, because the latter are mush less succeptible to noise than analog signals. The improved

reliability is gained at the expense of an increased bandwidth required for the transmission of digitized signals, unless some data compression technique is applied.

The compression of the cardiogram without loss of diagnostic information is based on the fact that consecutive samples of the digitized electrocardiogram carry redundant information that can be removed with reasonable computing efforts.

Several compression techniques have been reported yielding data reduction ratios from 3:1 upto 12:1 by exploiting different properties of the signal.

Compression of the ECG without any loss of signal fidelity can be achieved by removal of some of the redundancy within the sequence of the digitized signal samples which is due to one or both of two reasons : (i) neighbouring samples are not statistically independent, and (ii) the quantized signal amplitudes do not occur with equal probability [16]. The redundancy resulting for these two different reasons can be partly removed by separate techniques, which may be employed together to achieve a large data compression ratio.

In dealing with the first redundancy source, a considerable simplification of the problem is obtained if, instead of endeavouring to establish statistical independence of suitably transformed signal samples, it is attempted to remove only the intersample correlation. This approach is based on the linear mean-square estimation theory and provides the optimal solution under the assumption of stationary signal statistics [17]. Within this framework, the rational for the minimum mean-square error (MMSE) predictor is as follows. Instead of storing the n-th sample, x_n, one may without loss of information, retain the difference e (the residual error) between x_n and its prediction, $\hat{x}_n$, based on

a linear combination of previous samples.

$$\hat{x}_n = \sum_{k=1}^{p} a_k \cdot x_{n-k}$$

The number of samples employed for the prediction is p, and the weight parameters a_k are chosen so that the expected mean-square error $\sigma_e^2 = E(x-\hat{x})^2)$ is minimum. The reason for seeking this minimum is the fact that since the residuals have zero mean, σ_e^2 is equivalent to the variance of the residual sequence, e_n, which thus enables one to encode e_n with a shorter digital word length than would be required for the original sequence, x_n, without any loss in precision. Furthermore, under certain conditions, e.g., if the residuals have a Gaussion distribution, minimum variance implies minimum entropy [18].

The structure of the compression and reconstruction system employing a second order predictor, a quantizer, and an encoder is shown in Figure 2.1. It is important to note that the predictors located in the compressor and the reconstructor loop are identical, and that both predictors, in the case of errorless transmission, estimate the ECG signal from identical sample sequences, $\tilde{x}_n$. At the compressor side, this is achieved by the inclusion of a feedback around the quantizer, which ensures that the error in the reconstructed and quantized signal, $(x_n - \tilde{x}_n)$, is not an accumulation of previous quantization errors and is precisely the quantization error of the residual e_n,

$$q_n = x_n - \tilde{x}_n = x_n - \hat{x}_n - \tilde{e}_n = e_n - \tilde{e}_n$$

The optimal MMSE predictor weights a_k were computed from the Yule-Walker equations by using the serial autocorrelations, which were estimated from the total set of (100 Hz bandwidth limited), ECG signal records sampled at 200 Hz [19]. The sample sequance at this rate was obtained by a 4-point Lagrange interpolation from the original sampling rate of 500 Hz to an intermediate
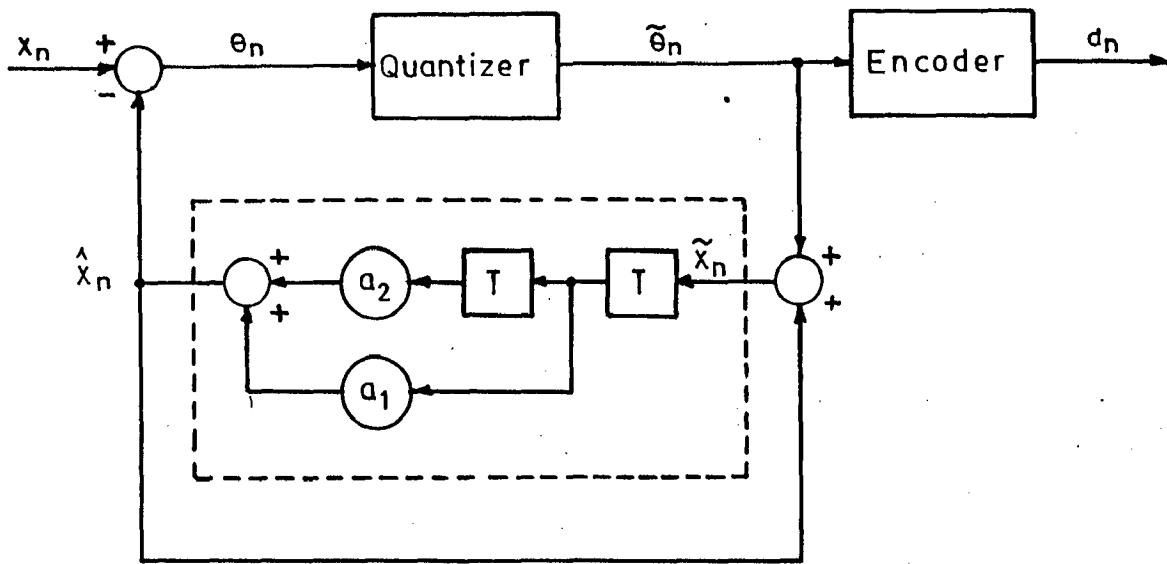
(a)
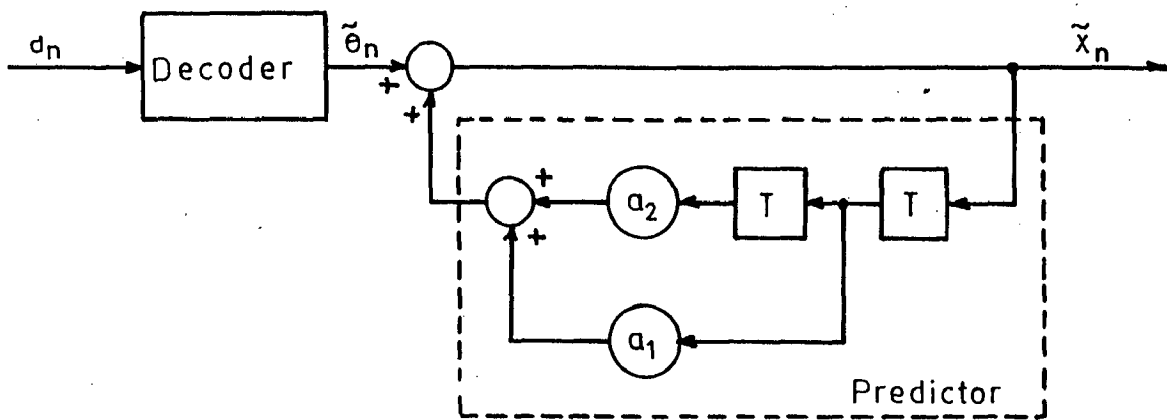
(b)

Fig. 2.1 Block Diagram of the Compression
(a) and Reconstruction (b) Scheme
Using Second Order Prediction

sampling rate of 1000 Hz, and subsequent 5:1 decimation to the desired rate of 200 Hz with a Kaiser windowed lowpass filter of 100 Hz (-6 dB) bandwidth [19].

The analysis of the results obtained by MMSE predictors in terms of residual error variance revealed that for a sampling rate of 200 Hz, the same set of weights a_k could be used for all three orthogonal leads without practical loss of efficiency, the variance was not substantially decreased by implementing predictors of higher than second order, and the variance reduction achieved relative to the variance of the original ECG signal sequence was on the average 14 dB.

Linear mean - square estimation theory was used in an analogous way to design an MMSE interpolator [17]. However, in this case the estimator, $\hat{x}_n$, consists of a linear combination of past and future samples, which, in practice, can always be achieved by introducing an appropriate time delay. Using the result from the MMSE predictor, a second order estimator is sufficient, and thus adequate performance of an interpolator using only one past and one future sample can be expected, resulting in the estimator,

$$\hat{x}_n = a x_{n-1} + b x_{n-1}$$

In Figure 2.2 the structure of the compression scheme using interpolation is shown together with the associated reconstruction procedure based on the transmission of the residuals $e_n = x_n - \hat{x}_n$, leading to

$$x_{n+1} = (1/a)(x_n - e_n) - (b/a)x_{n-1}$$

Minimizing the expected mean-square of the residuals, e_n, provides the following conditions : (a) equal weights $a = b$ are required, (b) the autocorrelations estimated from the 220 patient sample suggest weight values close to 0.5, and
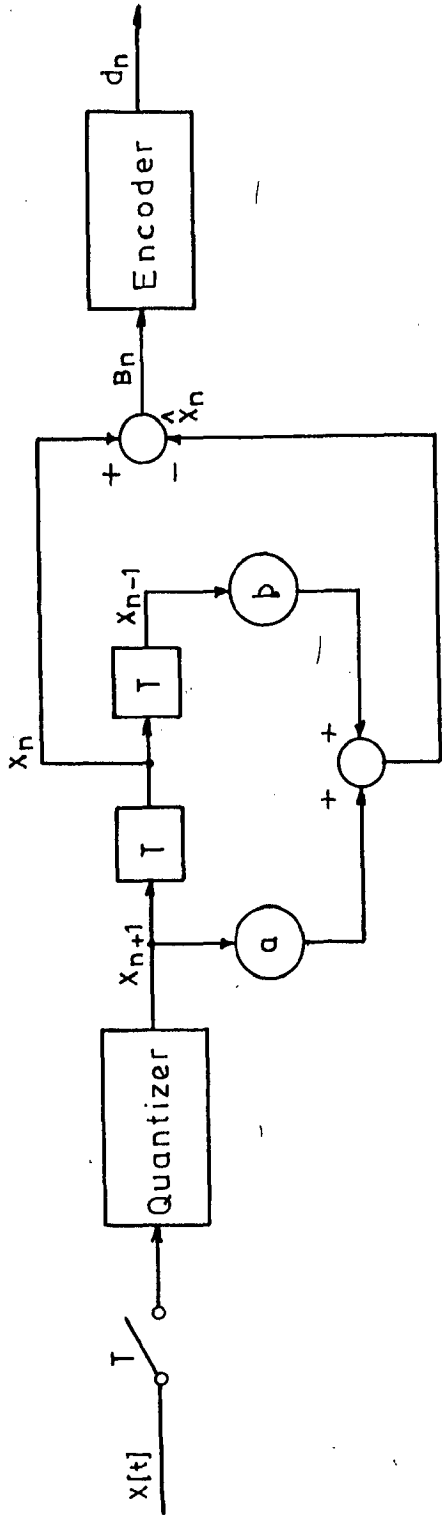
(a)
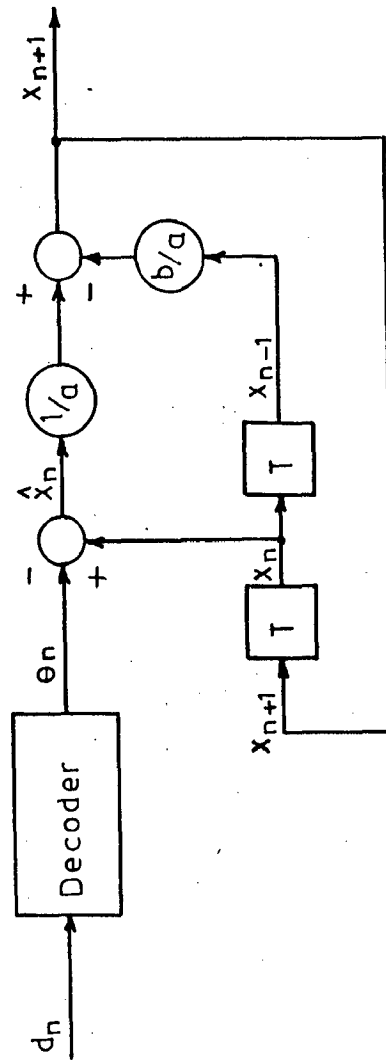
(b)

Fig. 2.2 Block Diagram of the Compression (a) and Reconstruction (b) Scheme
Using Interpolation

(c) at a sampling rate of 200 Hz, and the selection of a = b = 0.5, the variance of the residuals is 20 dB below the variance of the original ECG signal sequence.

Of great practical significance is the fact that the optimal values for the interpolation weights are very close to 0.5. This then permits, by specifying them exactly 0.5, extremely simple implementations of the equations (3) and (4), where all the necessary multiplications can be executed by arithmetic shift operations. Furthermore, truncation errors can be completely avoided by selecting the word length of the arithmetic processor two bits longer than the one required for adequate representation of the signal samples, x_n. These features make the instrumentation of the interpolation scheme simple, and attractive for implementation by microprocessors.

A disadvantage of the interpolation scheme is that the effects of occasional bit errors on the reconstructed signal sequence, which may occur during the transmission of e_n, do not vanish after a sufficient number of properly transmitted samples. This is in sharp constrast to the characteristics of the MMSE predictor scheme for which exponential decay of transmission error effects can be shown. Therefore, in applications where the transmission is not otherwise protected, as in a computer dish/memory transfer, adequate error control is mandatory (21), which then will add some overhead in terms of storage space and/or transmission time. This overhead will have to be taken into consideration for evaluating the relative merits of interpolation versus prediction in a particular application.

Addressing the other source of redundancy arising from the nonuniform distribution of the residuals, optimal removal can be achieved by entropy encoding, where short codewords are assigned to frequently appearing residuals and longer ones to those occurring less frequently. A widely known procedure to construct such variable-length codes was proposed by Huffman [22], resulting

for independent signal samples in average codeword lengths that are lower bounded by the entropy of the original sequence. In the present application, the residuals are merely uncorrelated rather than statistically independent, and thus some deviation from that lower bound must be accepted.

The corresponding entropy estimates for prediction were 3.6, 3.0, 2.6 bits/ sample, and for interpolation 3.1, 2.4, 2.1 bits/sample, for quantization step sizes of 10, 20, 30 μV, respectively. It is evident that the residual distributions arising from interpolation are narrower peaked and more symmetrical than the corresponding ones obtained from prediction, resulting in the smaller entropy values.

The root mean-square (RMS) quantization noise introduced with the step of Δ = 10, 20 and 30 μV measured 2.9, 5.8 and 8.7 μV, respectively, corresponding to distortion relative to the ECG signal RMS amplitudes of 2, 3.5 and 5 %. The effects of this quantization noise, in terms of changes induced in the wave measurements, were assessed by the same protocol utilized for the bandwidth evaluation. The results revealed that the amplitude measurements were not adversely affected, whereas, the time measurements showed increased discrepancies with a progressive increase of the quantization step size. This was most prominent for the P duration, where for Δ= 20 and 30 μV the fidelity criterion was exceeded in 0.4 and 4.5 percent of the records, respectively, while the corresponding numbers for the QRS duration were 1.8 and 2.3 percent. Hence, with the selection of a quantization step size of 30 μV and a sampling rate of 200 Hz, the proportion of records with measurement errors larger than the amount of intrapatient beat-to-beat variability can be kept below 5 percent.

In order to reduced the size of the look-up table for the practical implementation of variable-length codes, Huffman encoding was extended only over ± 5 and ± 8 quantization levels for the quantization step sizes of 30 and 20 uV, respectively. Residuals falling outside these central ranges were tagged with a prefix to which an ordinary 8-bit binary encoding of the quantization level was attached.

The better performance of the interpolation versus the prediction procedure is most evident by the smaller 96 percent range of average wordlengths, demonstrating the better uniformity of compression achieved within the patient sample.

For a quantization step size of $\Delta = 30 \, \mu V$, the interpolation scheme requires an average wordlength of 2.21 bits/sample, and 98 percent of the leads can be encoded with less than 3 bits/sample. For a sampling rate of 200 Hz, the corresponding transmission rates for 3 simultaneous leads are 1,327 bits/s, and 1,800 bits/s, respectively, which could be reliably accommodated by presently available synchronous MODEMs operating over the switched telephone network at 2,4000 bits/s. For the linear quantization with $\Delta = 20$ uV, the average wordlength and upper range are 2.57 and 3.63 bits/sample, respectively, resulting in transmission rate of 1.542 and 2.180 bits/s, which are still feasible for MODEMs.

The average wordlength encoding from the prediction scheme would still permit a real-time transmission, yet, due to the much larger 96 percent range, a substantial number of records could not be transmitted at a rate below 2,400 bits/s, thus rendering this method less attractive for synchronous transmission.

In conclusion, the extremely simple fixed-point algorithm suitable for microprocessor implementation, and the relativelyuniform compression performance make the interpolation scheme for significant reduction in transmission rates and storage space of digitized ECG records very attractive.

2.3.5 Straight Line Approximation for Data Compression of Electrocardiograms [2]

This technique is based on the principle of coherent averaging as applied to biological and especially ECG signals. ECG signals have repetitive and recurrent wave form complexes. The shapes of these complexes indicate the state or condition of the process by which they are generated.

In this technique, we assume that we know more or less accurately the presence of the wave forms in case of a particular ECG; the different ventricular wave shapes like supraventricular beats or ventricular extrasystoles. It is still possible, however that certain parts of the waveforms are very much distorted by noise or other disturbances. In the case of fetal abdominal ECG, there might be the always present electromyographic noise and the maternal ECG complexes, in ECGs obtained during physical exercise this might be artefacts and muscle noise. For resting ECGs, it can also happen that the patient is not quite, so that, because of disturbances, the Q waves can not be seen clearly. In such cases of distorted, but recurrent wave shapes, if they are always typified, coherent averaging technique like straight line approximation may help to improve the Signal Noise Ratio.

In this technique, an approximation of the electrocardiogram is obtained by piecewise linear segments inside the corridor formed by the two functions which are less than and greater than the given waveform by amounts ε as shown in Figure 2.3. This concept yields polygonal curves for which the
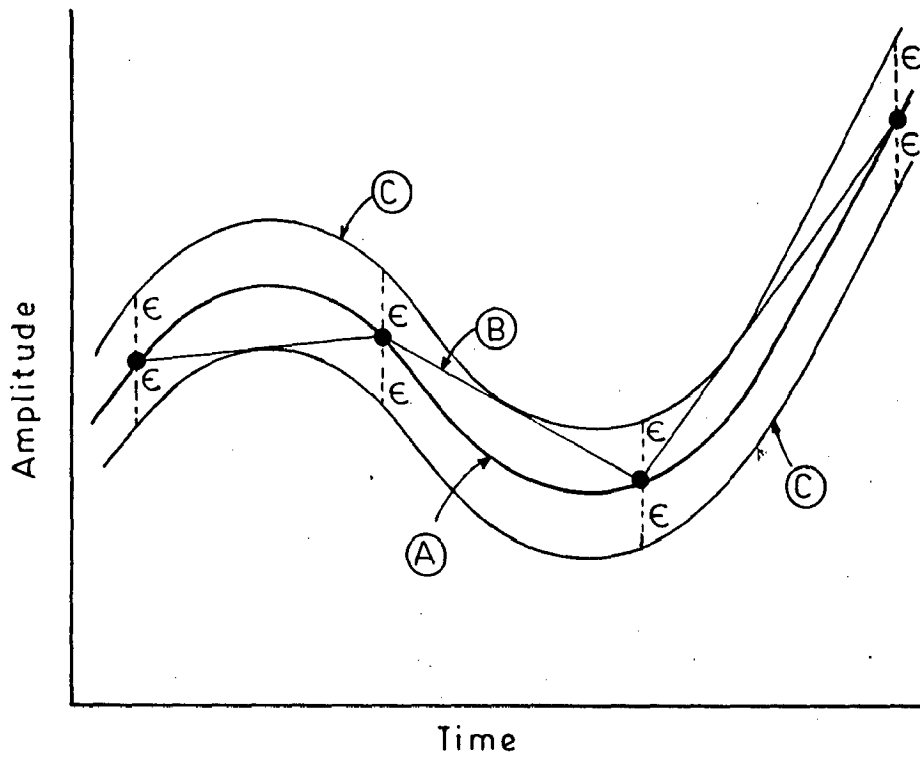
Fig. 2.3 Basic Concept of Moving Sign Slope Method
(A) Original Signal (B) Straight Line
Approximation of Original Signal
(C) Approximation Error Boundary

maximum difference between the given waveform and the polygonal approximation is equal to or less than a specified value.

In this, data can be compressed in real time i.e. as the waveform is generated. The operation is scan along i.e. the output is generated as the input is received.

The compression ratio achieved is dependent upon the type of ECG being monitored and the sampling rate at which it is being monitored. The compression achieved is of the order of 10 to 1 [2].

The original waveform can be reconstructed with adjustable fidelity depending on the requirements of the host system.

CHAPTER - III

TEMPLATE METHOD FOR DATA COMPRESSION OF ECG SIGNALS

3.1 PURPOSE OF ECG PROCESSING AND TEMPLATE MAKING

3.1.1 Stages in Processing

The purpose of the processing of signals is generally the reduction of a redundant transducer output flow to a few parameters that have to be relevant to the problem to be solved, in order to prepare a subsequent template and take appropriate human action. It means that only those parameters need to be extracted from the transducer output signal, that give insight into the signal source or the underlying physiological process. The processing of the information consists of three stages:

(i)   the measurement and signal recording;

(ii)   the improvement of the signal to noise ratio (SNR), signal transformation and data reduction; and

(iii)   the recognition, interpretation and decision.

These three phases of the signal handling might be intermingled when attacking a specific problem. Ideally, however, the information processing is carried out in a closed loop, with still a fourth stage added to it :

(iv)   the treatment

All the processing can be done most effectively, only when as much a priori knowledge as possible has been assembled and is used concerning the ECG and the heart. It is inappropriate to try to describe generally the process-ing of ECG signals as if for this purpose there could exist any general scheme. Before a useful scheme can be designed, the signal must have been analysed

with respect to its different aspects, its source and pathways must have been characterized and the physiological relevance of the resulting information taken into account. All this needs to be done prior to the template formation.

3.1.2 Disturbance and Variability in ECG Signals

It is almost never possible to classify unambiguously the ECG signal from a physiological process by means of its extracted parameters, because these parameters can seldom be determined with a sufficiently high accuracy. The cause of this insufficient accuracy is twofold: biological variability is an inherent property of such signals and disturbance is always superimposed on the signals. The main causes are the variability with time and the inter-individual variability of the information sources. Nearly every physiological signal is generated by a process of which the properties and parameters change as a function of time. The fetus may, for instance, move rather freely below the electrodes during the recording of its ECG so that the FECG parameters vary continuously as a function of time.

Also, from a statistical point of view, the signals are rarely ergodic. In terms of the signal-space concept, the variability between physiological heart processes of different individuals (e.g. the de-and repolarization processes of heart muscles) results in the signal-space, in an overlap of the regions which describe signals drawn from normal and abnormal processes, so that a certain percentage of diseases cannot be recognized intrinsically. On account of the nonstationarity of the processes, the observation time of the signals has to be short even when all controllable circumstances that influence the process and thus the signal, are kept constant.

The second reason for the sometimes very low accuracy of the determination of the parameters is the superimposed disturbance. Together with the

ECG signals one always records small and sometimes large amounts of random noise and other disturbing signals. The latter is especially true for the fetal ECG. The noise, being the random component, is fortunately mostly uncorrelated with the signal. Other disturbing signals may, however, be largely correlated with the signal that we wish to process and may even originate from the same or related sources. Since the observation time must be kept short on account of the nonstationarity of the process, it is difficult to eliminate the influence of the disturbance sufficiently in order to obtain a small enough noise variance.

3.1.3 Parameter Estimation and Interpretation

ECG signal parameters have thus to be estimated for a time varying or nonstationary signal in a random noise background. The signal itself often has a stochastic character too, some aspects of the signal may be stochastic or the signal may have a completely random character. In the latter case the processing reveals parameters that are only describable in statistical terms.

It sometimes happens that nearly nothing is known about the process except for the signals flowing to the outside world, and that these signals can only be measured by passive methods, without the possibility of influencing the process at will. Computation of significant parameters is then possible by a mutual comparison of several information streams. By comparing different signals of the same nature (transversal, from many patients for instance) or by observing one signal as a function of time (longitudinal, e.g. in follow-up studies) it is possible to extract only those parameters that change significantly.

This can often be done by classical signal analysis or transformation methods. Of course it is then important to search for the physiological meaning

of such changes since this is the ultimate aim of the analysis and the proce-
ssing.

When several signals can be obtained from the same source it is always
meaningful to search for interrelations even when the signals are statistically
not comparable at all. It could happen that the determination of the relation-
ship leads to insight into their common origin or even reveals the fact that
one of the signals is responsible for the generation of the other. Understanding
of the process is sometimes much facilitated when it is feasible to simulate
the process with an input signal. This increases the possibility that the recorded
signals can be interpreted since all other (mostly unknown) inputs and distur-
bing factors can be eliminated by using a suitable 'probe' signal which can
unequivocally be identified. This of course has to do with the fact that we
have then enlarged our a priori knowledge about the system by offering
it controlled inputs; it permits the application of system analysis and control
theory for such problems.

All physiological systems are nearly non-linear, but even when these
systems could be describable in linear terms, physiological information about
the process would still be needed, in order to make possible an interpretation
in physiological terms.

It can thus be stated, that any processing method for the estimation
of relevant parameters must be firmly grounded in physiological and physical
knowledge about the process.

The action stage is only reached when we dispose of reliable parameters.
With independent parameters it is possible to build up a parameter space in
which the development of the process can be characterized as a function of
time. A trend in a patient's condition can then be represented by a shifting

or wandering condition vector in this space. The trajectory can be extrapolated and predicted when the bandwidth of this vector space-walk is limited. The smaller the bandwidth, the longer the time course over which it can be extrapolated. Then it is possible to extend one's attention beyond the therapeutic situation towards the prevention.

3.1.4 Point Processes - Their Significance in ECG Data Interpretation and Compression

Since the 'state' of the heart processes can often be followed by studying the types of events generated by them as a function of time, special attention to such events is given and they are treated as point processes.

Many physiological systems like the heart are accompanied by signals which are related to a coupled series of events. Such events may be the cause for the generation of wave forms, repetitive or not, with stable or variable shapes. From a signal processing point of view it is sometimes desired not to analyze such signals with respect to their wave contours, but to look at their occurrence in time and their behaviour as point processes.

In many cases it is quite a complicated work to derive from such waveforms a series of pulses, which are one-to-one coupled with the physiological event. Once the pulse series has been obtained, the whole series of events can be traced and the signal can be treated as a point process. If the wave shape properties are known and if their variability is of no special interest, it is easy to reconstruct the physiological signal with the help of pulse series and the statistics about the wave forms. In physiology and medicine there are many such examples of point processes with coupled wave forms. In a way we could say that almost all physiological signals are generated in this manner, although it is not always possible to extract the point process

properties from such signals, the electrocardiogram, if generated in a non-pathological way, is coupled to the processes generated in S-A and A-V nodes.

Although the analysis of the physiological heart signal into its different aspects has many important advantages - e.g. the wave shape of the ECG for the diagnosis of infractions or hypertrophy and the occurrence of the different wave types for a rhythm classification - it is in practice often extremely difficult to separate these aspects. For patient monitoring one is not so much interested in the different wave shapes, but mainly in the repetitive aspects of the various waves. In the case of diagnosing a patient with help of his vectorcardiogram during physical exercise one is interested, in the midst of the many thousands of different heartbeats, in the depression of only a small part of the ST-T complex. In both instances, however, the proper determination of the presence of the wave form seems easy, if performed by humans e.g. from paper recordings, but it appears to be highly complex if done by instruments or computers.

As soon as one has a more specific knowledge about the ECG signals, apart from the general knowledge, e.g. from what source it is generated, one can optimize the methods or processing algorithms. It is of high importance to assemble as much prior information as can be done about waveshapes and - not less important - about the noise and disturbance properties.

3.2 ALGORITHM FOR TEMPLATE METHOD OF ECG DATA COMPRESSION

The template matching technique used for the data compression is an exhaustive method for first finding the various distinctive parameters that characterize an ECG wave and then finding means to compress the data points required to characterize this wave. The template method is a logical

and adaptive method where data compression achieved is dependent on the clinical situation for which it is being used.

The algorithm that has been developed for making the template is discussed as below :

First the approximate R-R interval from the pulse rate is found out, say RRI. Record of a length which is approximately equal to ten times RRI is made. This record of about ten ECG waves is going to be used to make the ECG template.

Now scan a length of 1.5 times RRI which would contain 1 or 2 R peaks - to get global positive maxima, thereby getting one R peak. Scanning rate is of 200 samples/second i.e. a time interval of 0.005 seconds.

Once the R peak has been found out, find slopes on either side of the R peak by taking a point three divisions in time ahead and previous of the R peak. Also find change in slope at R peak. Change in slope should not be less than 100°. This change in slope has been characterized in degrees, if there is a decrease in magnitude of R wave a correction factor of the order of $1.5/R_{max}$. is applied to offset the change in slope due to decrease of magnitude.

If change in slope is greater than 100°, R peak has been found. However, if change in slope is less than 100°, the scanning is done for 1.5 times RRI interval again in order to find R wave. If R peak is not found in five scannings, then the process is stopped and alarm sounded.

But if change in slope is greater than 100°, scanning for the next R peak in the region of one and a half times RRI is done to find the next R peak. The distance between the first R peak and the second R peak gives the exact RR interval.

Once the exact distance between two successive R peaks has been found. From the second R peak, start scanning ahead of it applying area Al criteria according to which the current area is computed till the current area sum does not become equal to the 2 percent of total area. At this station, the start of S segment is found. Now applying the area Al criterion again and moving ahead, the end of S segment is also computed. The peak in this region gives the S peak which is found out through the "MIN" subroutine.

Similar to the area criterion applied for computing the ordinates of S segment, which is a forward part of QRS complex, the Q segment is computed with the help of negative incremental area being 2 percent of current sum. Thus the entire QRS complex is identified, start of Q segment, end of Q segment, start of R complex, end of R complex, and beginning of S segment and its end. The slope values of the various segments are also computed along with the maximum amplitude.

With R_{max} - global maxima point of the entire ECG wave identified, scanning ahead of QRS complex for local maxima gives rise to the T peak. Applying area criterion Al similar to the one to calculate RAPF, the end of T wave segment is identified. If area criterion A, similar to the above is applied previous to peak of T wave, the start of T wave is identified.

Similarly, as the R_{max} - global maxima point has once been identified, scanning points previous to the QRS complex for local maxima, identifies the P wave peak. Now applying the area criterion forward of the P peak the end of P wave can be identified. At the same time, applying area criterion backward of P peak, the start of P wave is identified.

Thus all the clinically significant parameters - the QRS complex, its duration, maximum and minimum points are identified. Same is the case

with the T and P wave whose duration and maximum points help to characterise the wave.

Once the template has been made, the data ECG waves have to be compared with the template. Comparison between two ECG waves is possible only if they both are quantified - in terms of measurable parameters.

The quantification of the ECG wave is achieved by characterizing the various wave segments by their duration, amplitude and slope values.

The data ECG waves are characterised in the same way as the original template ECG. The basis of comparison between the two waves are the slopes that defines the various wave segments. If the slope values that identify a wave segment change by more than 10 percent of the slope value of the template, the data ECG wave is stored.

The data compression achieved in this method apart from being dependent on the clinical situations, also depends on the basis of parameter quantification of the ECG wave. This is a logical and comprehensive method of data compression that incorporates all possible features. It's flow chart is shown in Fig. 3.1.

3.3   FEATURES OF THE ECG TEMPLATE PROGRAM

(i)   The program developed for ECG template formation is a comprehensive program which takes into account all types of possibilities.

(ii)   The program is flexible in the sense that in order to save computing time various steps could be omitted. Although this would tend to make the computations less accurate, but the nature of the ECG wave is such, that large inaccuracy is not introduced.

DATA
X = APPROX, VALUE OF PULSE DURATION IN SEC.
LENGTH OF ECG WAVE AVAILABLE
= 10 * X SEC
STORE AMPLITUDES IN A1 TO A(2000 * X)
DT = SCANNING INTERVAL IN SEC
CFS = LIMITING 2° SLOPE TO FIND BASE
LINE CORRECTION
M = COUNTER TO CHECK PRESENCE OF R PEAK
APF 1 = PERCENT INCREMENT AREA AS CONVERGENCE
CRITERIA FOR R PEAK
APF2 = PERCENT INCREMENT AREA AS CONVERGENCE
CRITERIA FOR Q-& S PEAKS

INITIALIZE
M = 0
X = 0.600
I3 = 1.5 * X
DT = 0.005
CFS = 0.00044
APF1 = 2.
APF 2 = 1.
I1 = 1
I2 = I1 + I3

GENERATE 9 MORE WAVES IF ONLY ONE WAVE
AVAILABLE FOR TESTING THE PROGRAMME
(CONTINUOUS 10 WAVE RECORD ESSENTIAL
IN REAL PROBLEM)

A(N1 + I) = A(I)

FIND 1st R PEAK THROUGH 'MAX'
RMAX1 = 1st R PEAK AMPLITUDE
IM1 = 1st R PEAK STATION

FIND SLOPE OF LINES JOINING AMAX AND
POINTS THREE STATIONS AHEAD AND BEHIND IM1

T1 = 10. * (AMAX - A(IM - 3))/(0.015 * 25.0)
TH1 = ATAN (T1)
T2 = 10. * (A(IM + 3) - AMAX)/(0.015 * 25.0)
TH2 = ATAN (T2)

```
┌─────────────────────────────┐
│      CHANGE OF SLOPE        │
│     DTHA = (TH1 - TH2)      │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│  MODIFIED CHANGE OF SLOPE   │
│  DTHAM = (DTHA) * (1.5/AMAX)│
└─────────────────────────────┘
              │
              ▼
┌───────────────────────────────────┐
│  FIND 2nd  R PEAK THROUGH 'MAX'   │
│  RMAX 2 = 2nd  R PEAK AMPLITUDE   │
│  IM 2 = 2nd  R PEAK OCCURENCE     │
│  XE = EXACT  R-R INTERVAL IN SECONDS │
└───────────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│  APPLY BASE LINE CORRECTION │
│      A(I) = A(I) - COX      │
└─────────────────────────────┘
              │
              ▼
┌───────────────────────────────────┐
│            N1 = IM2               │
│            M = N1                 │
│            II = 1                 │
│            N1 = M-II              │
│            DAL = 0.002            │
│            NZY = 0                │
│  CALCULATE AREA UNDER FORWARD HALF OF │
│  THE  R WAVE (RAPF) AND AREA UNDER THE │
│  S WAVE SEGMENT (SAN) THROUGH "ACOM". │
└───────────────────────────────────┘
              │
              ▼
┌───────────────────────────────────┐
│  FIND THE  S PEAK IN THE DEFINED  │
│     RANGE THROUGH 'MIN'           │
│     SMIN = AMIN ; INS = IM        │
└───────────────────────────────────┘
              │
              ▼
┌───────────────────────────────────┐
│            NZY = 0                │
│            II = -1                │
│            N1 = M - II            │
│  CALCULATE THE BACKWARD HALF AREA │
│  UNDER THE  R WAVE (RAPB) AND AREA UNDER │
│  THE  Q WAVE SEGMENT (QAN) THROUGH 'ACOM' │
└───────────────────────────────────┘
              │
              ▼
┌───────────────────────────────────┐
│     FIND THE  Q PEAK IN THE       │
│   DEFINED RANGE THROUGH 'MIN'     │
│     QMIN = AMIN ; INQ = IM        │
└───────────────────────────────────┘
```

```
┌─────────────────────────────────┐
│  FIND    P PEAK  IN  THE  RANGE │
│        I1 = I2 - 24   TO        │
│        I2 = IM2 + NR - 6        │
└─────────────────────────────────┘
                 │
                 ▼
┌─────────────────────────────────┐
│ FIND AREA  SAP UNDER  P WAVE THROUGH │
│   'AREA' IN THE RANGE N1   TO  NS │
└─────────────────────────────────┘
                 │
                 ▼
┌─────────────────────────────────────────┐
│ COMPUTE UPHILL SLOPE  SLUP OF LINE JOINING │
│      STARTING  PT. AT  N1   TO  P PEAK  & │
│      DOWN SLOPE  SLDP OF LINE JOINING     │
│          P PEAK   TO END PT. NS           │
└─────────────────────────────────────────┘
                 │
                 ▼
┌─────────────────────────────────┐
│  FIND    T PEAK  IN  THE  RANGE │
│      I1 = IM1 + NR + 36   TO    │
│          I2 = I1 + 40           │
└─────────────────────────────────┘
                 │
                 ▼
┌─────────────────────────────────┐
│ FIND AREA SAT UNDER   T WAVE THROUGH │
│   'AREA' IN THE RANGE  NS  TO  N1 │
└─────────────────────────────────┘
                 │
                 ▼
┌─────────────────────────────────┐
│  FIND UP-SLOPE SLUT  OF LINE    │
│     JOINING  NS TO  T PEAK  &   │
│  DOWN-SLOPE OF LINE JOINING     │
│     T PEAK   TO END PT. NS      │
└─────────────────────────────────┘
                 │
                 ▼
              ( END )
```

Fig. 3.1(a) Flowchart for 'MAIN PROGRAM'

```
┌─────────────────────────────┐
│   R  AREA FORWARD OF/        │
│   BACKWARD OF   R PEAK       │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│  SA   AREA UNDER THE PEAK    │
│   INITIAL VALUE = 0.         │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│    INPUT FIRST SAMPLE        │
│    STORE AS   N1             │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│    INPUT NEXT SAMPLE         │
│     N = N1 + II              │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│   COMPUTE AREA BETWEEN       │
│   TWO SUCCESSIVE SAMPLES     │
│  A1 = 0.5 * (A(N) + A(N1)) * DT │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│    UPDATE TOTAL AREA         │
│     SA = SA + A1             │
└─────────────────────────────┘
              │
              ▼
┌─────────────────────────────┐
│  COMPUTE CURRENT AREA AS A   │
│  PERCENTAGE OF TOTAL AREA    │
│   APC = A1 * 100.0/SA        │
└─────────────────────────────┘
              │
              ▼
         ◇ APC - APF1 ≥ 0 ◇ ──── Yes
              │ No
              ▼
         ◇ APC ≥ 0 ◇ ──── Yes
              │ No
```

Area between two successive samples:

$$A1 = 0.5 * (A(N) + A(N1)) * DT$$

Update total area:

$$SA = SA + A1$$

Current area as a percentage of total area:

$$APC = A1 * 100.0/SA$$

Decision: $APC - APF1 \geq 0$

Decision: $APC \geq 0$

┌─────────────────┐   ┌─────────────────┐
│   DEFINE   │   │   DEFINE   │
│   RAP = SA   │   │   RAP = SA - A1 │
└─────────────────┘   └─────────────────┘

┌─────────────────────────┐   ┌─────────────────────────┐
│ COMPUTE TIME FOR   │   │ COMPUTE TIME FOR   │
│ START/END OF S/Q   │   │ START/END OF S/Q   │
│ SEGMENT RESPECTIVELY   │   │ SEGMENT RESPECTIVELY   │
│   II = N   │   │   II = N1   │
│ AN1 = N - IM2   │   │ AN1 = N1 - IM2   │
│ TZRS = AN1 * DT   │   │ TZRS = AN1 * DT   │
└─────────────────────────┘   └─────────────────────────┘

N1 = N1 - II
S/Q   AREA
COMPUTATION

SA = AREA UNDER THE PEAK
INITIAL VALUE = 0.
II = +1 (INCREMENT)
OR -1 (DECREMENT) IN STATION NO.

INPUT FIRST SAMPLE
STORE AS N1 = IM2

INPUT NEXT SAMPLE
N = N1 + II

COMPUTE AREA BETWEEN TWO
SUCCESSIVE SAMPLES
A1 = 0.5 * (A(N) + A(N1)) * DT

COMPUTE
DA = A(N) - A(N1)

ABS (DA)
- DAL ≯ 0

No

Yes

USE A COUNTER NZY
TO CHECK IF TWO
INCREMENTAL AREAS
ARE LESS THAN A SMALL
LIMITING VALUE
NZY = NZY + 1

NZY - 2
≯ 0

Yes

No

NEGATIVE PEAK
PROBABLY MISSING

UPDATE TOTAL AREA
SA = SA + A1

COMPUTE CURRENT AREA AS A
PERCENTAGE OF TOTAL AREA
APC = A1 * 100.0/SA

50

Fig. 3.1(b) Flowchart for 'ACOM' Subroutine

51

Fig. 3.1(c) Flowchart for 'AREA' Subroutine

Fig. 3.1(d) Flowchart for 'MAX' Subroutine

```
┌─────────────────────────┐
│   INPUT FIRST SAMPLE     │
│  STORE AS  A(I1) = AMIN  │
│        J = I1            │
└─────────────────────────┘

┌──────────────────────────────────────┐
│      SET RANGE FOR STATIONS            │
│ I2 = UPPER LIMIT ;  I1 = LOWER LIMIT   │
└──────────────────────────────────────┘

┌─────────────────────────┐
│    INPUT NEXT SAMPLE     │
│        J = J + 1         │
└─────────────────────────┘

        A(J) - AMIN ⟩ 0      Yes
              │ No

┌─────────────────────────┐
│  UPDATE MINIMUM VALUE    │
│      AMIN = A(J)         │
└─────────────────────────┘

┌─────────────────────────┐
│   STORE STATION OF       │
│   MINIMUM VALUE          │
│      IM = J              │
└─────────────────────────┘

       ( RETURN )
```

Fig. 3.1(e)   Flowchart for 'MIN' Subroutine

DATA
X = PULSE DURATION OF ONE PULSE IN SEC
AT THE TIME OF MONITORING
RR = EXACT R-R INTERVAL IN SEC
A( ) = ORDINATES OF ECG WAVE
DT = STATION INTERVAL IN SEC
APF1 = PERCENT INCREMENT AREA AS CONVERGENCE
CRITERION FOR P-AND T-PEAKS
SRU, SPU, STU = UP SLOPES OF R-, P-, T-WAVES
SRD, SPD, STD = DOWN-SLOPES OF R-, P-, T-WAVES
ER, EP, ET = ACCEPTABLE PERCENTAGE ± ERRORS
(DERIATIONS) IN SLOPES OF R-, P-, T-WAVES
EX = ACCEPTABLE PERCENTAGE ± DEVIATION
IN PULSE DURATION, RR
CFR = FACTOR FOR REDUCING ORDINATES OF
SUCCESSIVE ECG WAVES, USED ONLY IN TRIAL RUN
OF PROGRAMME WHEN MORE WAVES ARE
GENERATED FROM A GIVEN SINGLE WAVE

↓

INITIALISE
CF = 1.0
DT = 0.005
APF1 = 2.
ER = 20.
EP = 30.
ET = 40.
EX = 0.
CFR = 0.1

↓

COMPUTE AVERAGE SLOPES
SR, SP, ST FOR
R-, P-, T-WAVES

↓

COMPUTE UPPER LIMITING SLOPE
VALUES : SRP, SPP, STP &
LOWER LIMITING VALUES : SRN, SPN, STN

↓

COMPUTE UPPER & LOWER LIMITING
VALUES OF R-R INTERVAL : RRP, RRN

↓

FIND IXL : THE UPPER LIMIT FOR
ORDINATES IN ONE ECG WAVE

↓

```
┌──────────────────────────────────────────────┐
│        GENERATE 2 MORE ECG WAVES WHILE        │
│  TESTING PROGRAMME WHEN ONLY ONE TRIAL WAVE   │
│      IS GIVEN AS DATA INPUT : A(N1 + I)        │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│     DETECT 1st R-PEAK THROUGH 'MAX' :          │
│       RMAXC AND OCCURRENCE : IM1               │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│   COMPUTE AVERAGE SLOPE OF R-WAVE :            │
│   SRC & COMPARE WITH STANDARD VALUE :          │
│   SR PRINT IF OUTSIDE GIVE LIMITS              │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│     DETECT 2nd R-PEAK THROUGH 'MAX' :          │
│       RMAXC AND OCCURRENCE : IM2               │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│     FIND EXACT R-R INTERVAL : RRC              │
│   COMPARE WITH STANDARD VALUE : RR             │
│   PRINTING OUTSIDE GIVEN LIMITS                │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│        FIND T-PEAK IN RANGE :                  │
│   I1 = IM1 + 36   TO   I2 = I1 + 50            │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│              FOR  T-WAVE :                     │
│         COMPUTE UP SLOPE : STUC                │
│           DOWN SLOPE : STDC                    │
│          AVERAGE SLOPE : STC                   │
│      COMPARE WITH STANDARD VALUE : ST          │
│        PRINT IF OUTSIDE GIVEN LIMITS           │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│              FOR P-WAVE :                      │
│         COMPUTE UP SLOPE : SPUC                │
│           DOWN SLOPE : SPDC                    │
│          AVERAGE SLOPE : SPC                   │
│      COMPARE WITH STANDARD VALUE : SP          │
│        PRINT IF OUTSIDE GIVEN LIMITS           │
└──────────────────────────────────────────────┘
                       │
                       ▼
┌──────────────────────────────────────────────┐
│     CONTINUE PROCESSING SUBSEQUENT            │
│           ECG WAVE SETS                        │
└──────────────────────────────────────────────┘
```

Fig. 3.1(f) Flowchart for 'ECG-CONTINUOUS MONITOR' Program

In order to save computation time, slopes on either side of R peak need not be found out as the QRS complex slope itself is fairly indicative.

Secondly while applying the area criterion, the percentage area of the current computed area sum even could be increased to say 7 percent without much loss of accuracy, but it would result in a saving of computation time.

(iii)   The area criterion has been used in this program to characterize initially the various wave segments of the ECG wave. The area under the curve is a measure of the myocardial activity.

Although area has been initially used to characterize the wave segments, it is the slope which is the most important parameter as a means of comparison.

In a way, both slope and area are a measure of similar components – amplitude of the wave and its duration though in different ways. One is the division of amplitude and its time duration, while the other is the product.

Depending upon the clinical situation for which it is being used, either · of the two parameters i.e. slope or area could be used to characterize the ECG wave segments.

Thus this template incorporates the merits of both the slope and area criterion.

(iv)   The other significant feature of the ECG template program is the baseline correction that is incorporated into it.

The baseline shift is a problem inherent of the ECG wave. In order to apply the baseline correction, the program searches for a sector of station points which have a low amplitude of less than 0.05 mV and the relative angle/slope between them is not greater than 2°. The double criteria helps to

overcome most of the baseline shift error which is subtracted uniformly from the entire wave.

(v)   The area criterion is initially applied in the program to find out both and T and the P wave segments of the ECG signal. The slopes - upward slope and downward slope are used to quantify the wave segments for comparison.

The slope is an important parameter in such a clinical situation as it represents a means of coherent averaging and it is not subjected to sudden variations caused by muscle artifacts or other noise sources.

(vi)   The program computes a large number of features to ·quantify and characterize the ECG wave and depending upon the requirements of the situation, the parameters can be identified.

3.4   RESULTS

The algorithm developed for the ·'template method' computes a number of parameters -/e.g. the station point of R peak, duration of QRS complex, area under various segments, the slope values of the R wave, P wave and T wave and their times of occurrence.

The first R peak has a magnitude of 1 mV and occurs at station point thirty five. The upslope and downslope of the first R peak is $\tan^{-1}$ (17.6) and $\tan^{-1}$ (17.3) respectively. The second R peak occurs at station point of one hundred fifty four (154). The R-R interval is of 119 station points which makes the R-R interval approximately equal to 0.6 seconds as the time duration between two successive station points corresponds to 0.005 seconds. The baseline shift is zero.

The R-R interval is next scanned for the S peak and area ahead .of R peak under the QRS curve is computed. The RAPF area (RAPF = Area of

R which is positive and forward of R peak) is about eight times the area under the S segment of the QRS complex defined as SAN area (SAN = Area of S wave which is negative). RAPF is equal to 0.013 and SAN equal to - 0.0014. The S peak has a magnitude of - 0.12 mV and occurs at station point 159. The duration of S segment is equal to 0.055 - 0.125 = 0.030 seconds.

The Q peak has a magnitude of 0.10 mV and occurs at station point one hundred and forty nine (149). The total area under the R segment of the QRS complex is (.0113 + .0113) which is equal to 0.0226, while the area under the Q segment is 0.007. The duration of Q segment is equal to 0.015 sec.

Table 3.1 Characteristics of the QRS Complex

Segment	Area	Peak	Duration
R segment	0.0226 mm^2	1.0 mV	0.050 sec.
S segment	0.0014 mm^2	-0.12 mV	0.030 sec.
Q segment	0.0007 mm^2	-0.10 mV	0.015 sec.

Similarly, the area under the P wave is 0.0085 mm^2, the peak occurs at station point one hundred and thirty two (132) and its duration is 0.01 seconds. The Q wave has an area of 0.0487 mm^2, which is twice that of the area under the QRS complex. The T wave's peak occurs at station point two hundred and five (205) and its duration is 0.325 seconds, which is greater than 50 percent of the total ECG wave duration of 0.6 seconds. The entire ECG wave is characterized in Table 3.2.

The upslope and downslope of the various wave segments of the ECG wave are computed on the basis of the start of wave segment and its end, and the station point at which the peak occurs. This forms the criteria for

Table 3.2 : Characteristics of ECG wave

Segment	Area	Peak	Duration
(i) QRS Complex			
R segment	0.00226 mm^2	1.0 mV	0.050 sec
S segment	0.0014 mm^2	-0.12 mV	0.030 sec
Q segment	0.0007 mm^2	-0.10 mV	0.015 sec
(ii) P Wave	0.0085 mm^2	0.17 mV	0.010 sec
(iii) T Wave	0.0487 mm^2	0.30 mV	0.325 sec

template matching and a variation of more than 10 percent in any of the slopes used to characterize the wave causes the entire ECG wave to be stored.

The data compression results for 50 continuous ECG data waves is shown in Table 3.3. Out of the 50 ECG waves inputed, only 10 needed to be stored, so the data compression achieved is of the order of 5:1. The template algorithm was checked by generating ECG wave data with missing complexes as shown in Table 3.4.

3.5 DISCUSSION

In the template method, the parameters of the sample ECG wave are able to reconstruct to a great degree the original ECG wave. Data compression achieved is large, as compared to the original input data as the wave is quantified with the help of area, peaks and time duration of various wave segments.

The slope, used as a measure of comparison between sample and data ECG waves gives good results - both for original waveform reconstruction fidelity and also as a means of storage.

This method, being a "logical" method for data compression takes into account all the prior knowledge of an individual's ECG wave and this then

Table 3.3 : Data Compression Results-Continuous Monitoring

```
RR interval   = 0.6 Sec.

R  peak          1.0              I = 35
S  peak         -0.12            I = 40
S  end           0.0             I = 45
Q  peak         -0.1             I = 30
Q  start         0.0             I = 27

QR slope         44.0  mV/s
RS slope        -44.8  mV/s

P  peak          0.17            I = 13
P  start         0.0            .I = 2
P  end           0.0             I = 21

Pu slope         3.2  mV/s
Pd slope        -4.4  mV/s

T  peak          0.3             I = 87
T  start         0.0             I = 47
T  end           0.0             I = 111

Tu slope         1.5  mV/s
.Td slope       -2.5  mV/s
              TEMPLATE

RR interval   = 0.6 Sec.

R  peak          0.95            I = 37
S  peak         -0.17            I = 42
S  end           0.0             I = 47
Q  peak         -0.15            I = 32
Q  start         0.0             I = 29

QR slope         44.0  mV/s
RS slope        -44.8  mV/s

P  peak          0.12            I = 16
P  start         0.0             I = 5
P  end           0.0             I = 24

Pu slope         2.2  mV/s
Pd slope        -3.0  mV/s

T  peak          0.25            I = 86
T  start         0.0             I = 46
T  end           0.0             I = 110

Tu slope         1.25  mV/s
Td slope        -2.1  mV/s
            WAVE 2
```

1

```
RR interval  = 0.6 Sec.

R    peak           0.9              I = 39
S    peak           absent
S    end            0.0              I = 44
Q    peak           -0.1             I = 34
Q    start          0.0              I = 31

QR  slope           40.0  mV/s
RS  slope          -36.0  mV/s

P    peak           0.2              I = 14
P    start          0.0              I = 4
P    end            0.0              I = 23

Pu  slope           4.0  mV/s
Pd  slope          -4.4  mV/s

T    peak           0.36             I = 85
T    start          0.0              I = 48
T    end            0.0              I = 109

Tu  slope           2.7  mV/s
Td  slope          -3.0  mV/s
               WAVE 9




RR interval  = 0.6 Sec.

R    peak           0.84             I = 39
S    peak          -0.10             I = 45
S    end            0.0              I = 50
Q    peak           absent
Q    start          0.0              I = 34

QR  slope           30.7  mV/s
RS  slope          -33.6  mV/s

P    peak           0.19             I = 15
P    start          0.0              I = 5
P    end            0.0              I = 24

Pu  slope           3.8  mV/s
Pd  slope          -4.2  mV/s

T    peak           0.54             I = 84
T    start          0.0              I = 45
T    end            0.0              I = 109

Tu  slope           2.8  mV/s
Td  slope          -4.0  mV/s
               WAVE 16
```

2

RR interval = 0.6 Sec.

```
R    peak         1.0 .          I = 41
S    peak        -0.11           I = 46
S    end          0.0            I = 51
Q    peak        -0.1            I = 36
Q    start        0.0            I = 33

QR   slope        44.4 mV/s
RS   slope       -44.0 mV/s

P    peak         0.27           I = 15
P    start        0.0            I = 5
P    end          0.0            I = 24

Pu   slope        5.4 mV/s
Pd   slope       -6.0 mV/s

T    peak         0.53           I = 80
T    start        0.0            I = 41
T    end          0.0            I = 108

Tu   slope        2.7 mV/s
Td   slope       -3.6 mV/s
                WAVE 22
```

RR interval = 0.6 Sec.

```
R    peak         1.0           , I = 34
S    peak        -0.08            I = 39
S    end          0.0             I = 44
Q    peak        -0.05            I = 29
Q    start        0.0             I = 26

QR   slope        42.0 mV/s
RS   slope       -43.2 mV/s

P    peak         0.12            I = 16
P    start        0.0             I = 5
P    end          0.0             I = 24

Pu   slope        2.2 mV/s
Pd   slope       -3.0 mV/s

T    peak         0.2             I = 80
T    start        0.0             I = 40
T    end          0.0             I = 104

Tu   slope        1.0 mV/s
Td   slope       -1.7 mV/s
                WAVE 31
```

3

```
RR interval  = 0.6 Sec.

R   peak           0.85            I = 45
S   peak          -0.10            I = 50
S   end            0.0             I = 55
Q   peak          -0.06            I = 40
Q   start          0.0             I = 37

QR  slope          36.4 mV/s
RS  slope         -38.0 mV/s

P   peak           0.35            I = 16
P   start          0.0             I = 5
P   end            0.0             I = 24

Pu  slope          6.4 mV/s
Pd  slope         -8.7 mV/s

T   peak           0.72            I = 71
T   start          0.0             I = 31
T   end            0.0             I = 102

Tu  slope          3.6 mV/s
Td  slope         -4.6 mV/s
                WAVE 36


RR interval  = 0.6 Sec.

R   peak           1.04            I = 43
S   peak          -0.12            I = 49
S   end            0.0             I = 55
Q   peak          -0.06            I = 37
Q   start          0.0             I = 35

QR  slope          36.6 mV/s
RS  slope         -38.7 mV/s

P   peak           0.17            I = 16
P   start          0.0             I = 5
P   end            0.0             I = 24

Pu  slope          3.2 mV/s
Pd  slope         -4.4 mV/s

T   peak           0.3             I = 96
T   start          0.0             I = 56
T   end            0.0             I = 120

Tu  slope          1.5 mV/s
Td  slope         -2.5 mV/s
                WAVE 45
```

```
RR interval   = 0.6 Sec.

R   peak            1.0           I = 39
S   peak           -0.10          I = 50
S   end            0.0            I = 55
Q   peak           -0.1           I = 30
Q   start          0.0            I = 27

QR  slope          24.4  mV/s
RS  slope         -20.0  mV/s


P   peak            0.17          I = 16
P   start           0.0           I = 5
P   end             0.0           I = 24

Pu  slope           3.2  mV/s
Pd  slope          -4.4  mV/s


T   peak            0.3           I = 86
T   start           0.0           I = 46
T   end             0.0           I = 110

Tu  slope           1.5  mV/s
Td  slope          -2.5  mV/s
              WAVE 49



RR interval   = 0.6 Sec.

R   peak            1.1           I = 42
S   peak           -0.12          I = 50
S   end             0.0           I = 55
Q   peak           -0.11          I = 35
Q   start           0.0           I = 34

QR  slope          34.5  mV/s
RS  slope         -30.8  mV/s

P   peak            0.19          I = 16
P   start           0.0           I = 5
P   end             0.0           I = 24

Pu  slope           3.4  mV/s
Pd  slope          -4.7  mV/s


T   peak            0.52          I = 86
T   start           0.0           I = 46
T   end             0.0           I = 107

Tu  slope           2.6  mV/s
Td  slope          -4.5  mV/s
              WAVE 50
              TOTAL WAVES = 50
              END
```

5

Table 3.4 : Data Compression Results-Template Method

```
RR interval   = 0.8 Sec.

R   peak         1.0          I = 45
S   peak        -0.12         I = 50
S   end          0.0          I = 55
Q   peak        -0.1          I = 40
Q   start        0.0          I = 37

QR  slope        44.0 mV/s
RS  slope       -44.8 mV/s

P   peak         0.17         I = 16
P   start        0.0          I = 5
P   end          0.0          I = 24

Pu  slope        3.2 mV/s
Pd  slope       -4.4 mV/s

T   peak         0.3          I = 96
T   start        0.0          I = 56
T   end          0.0          I = 120

Tu  slope        1.5 mV/s
Td  slope       -2.5 mV/s


RR interval   = 0.8 Sec.
Baseline shift = 0.05
R   peak         0.95         I = 45
S   peak        -0.17         I = 50
S   end          0.0          I = 55
Q   peak        -0.15         I = 40
Q   start        0.0          I = 37

QR  slope        44.0 mV/s
RS  slope       -44.8 mV/s

P   peak         0.12         I = 16
P   start        0.0          I = 5
P   end          0.0          I = 24

Pu  slope        2.2 mV/s
Pd  slope       -3.0 mV/s

T   peak         0.25         I = 96
T   start        0.0          I = 56
T   end          0.0          I = 120

Tu  slope        1.25 mV/s
Td  slope       -2.1 mV/s
```

```
RR interval  = 0.8 Sec.

R   peak              0.9              I = 45
S   peak              absent
S   end               0.0              I = 50
Q   peak             -0.1              I = 40
Q   start             0.0              I = 37

QR  slope            40.0  mV/s
RS  slope           -36.0  mV/s

P   peak              0.2              I = 15
P   start             0.0              I = 5
P   end               0.0              I = 24

Pu  slope             4.0  mV/s
Pd  slope            -4.4  mV/s

T   peak  .           0.36             I = 94
T   start             0.0              I = 57
T   end               0.0              I = 118

Tu  slope             2.7  mV/s
Td  slope            -3.0  mV/s




RR interval  = 0.8 Sec.

R   peak              0.84             I = 44
S   peak             -0.10             I = 50
S   end               0.0              I = 55
Q   peak             absent
Q   start             0.0              I = 39

QR  slope            30.7  mV/s
RS  slope           -33.6  mV/s

P   peak              0.15             I = 15
P   start             0.0              I = 5
P   end               0.0              I = 24

Pu  slope             3.8  mV/s
Pd  slope            -4.2  mV/s

T   peak              0.54             I = 95
T   start             0.0              I = 56
T   end               0.0              I = 122

Tu  slope             2.8  mV/s
Td  slope            -4.0  mV/s
```

RR interval = 0.8 Sec.

```
R   peak          1.0           I = 45
S   peak          -0.11         I = 50
S   end           0.0           I = 55
Q   peak          -0.1          I = 40
Q   start         0.0           I = 37

QR  slope         44.4 mV/s
RS  slope         -44.0 mV/s

P   peak          0.27          I = 16
P   start         0.0           I = 6
P   end           0.0           I = 25

Pu  slope         5.4 mV/s
Pd  slope         -6.0 mV/s

T   peak          0.53          I = 93
T   start         0.0           I = 54
T   end           0.0           I = 122

Tu  slope         2.7 mV/s
Td  slope         -3.6 mV/s
```

RR interval = 0.8 Sec.

```
R   peak          1.0           I = 45
S   peak          -0.08         I = 50
S   end           0.0           I = 55
Q   peak          -0.05         I = 40
Q   start         0.0           I = 37

QR  slope         42.0 mV/s
RS  slope         -43.2 mV/s

P   peak          0.12          I = 16
P   start         0.0           I = 5
P   end           0.0           I = 24

Pu  slope         2.2 mV/s
Pd  slope         -3.0 mV/s

T   peak          0.2           I = 96
T   start         0.0           I = 56
T   end           0.0           I = 120

Tu  slope         1.0 mV/s
Td  slope         -1.7 mV/s
```

Con

```
RR interval    = 0.8 Sec.

R   peak            0.85          I = 45
S   peak           -0.10          I = 50
S   end             0.0           I = 55
Q   peak           -0.06          I = 40
Q   start           0.0           I = 37

QR  slope          36.4  mV/s
RS  slope         -38.0  mV/s

P   peak            0.35          I = 16
P   start           0.0           I = 5
P   end             0.0           I = 24

Pu  slope           6.4  mV/s
Pd  slope          -8.7  mV/s

T   peak            0.72          I = 96
T   start           0.0           I = 56
T   end             0.0           I = 127

Tu  slope           3.6  mV/s
Td  slope          -4.6  mV/s



RR interval    = 0.8 Sec.

R   peak            1.04          I = 43
S   peak           -0.12          I = 49
S   end             0.0           I = 55
Q   peak           -0.06          I = 37
Q   start           0.0           I = 35

QR  slope          36.6  mV/s
RS  slope         -38.7  mV/s

P   peak            0.17          I = 16
P   start           0.0           I = 5
P   end             0.0           I = 24

Pu  slope           3.2  mV/s
Pd  slope          -4.4  mV/s

T   peak            0.3           I = 96
T   start           0.0           I = 56
T   end             0.0           I = 120

Tu  slope           1.5  mV/s
Td  slope          -2.5  mV/s
```

RR interval = 0.8 Sec.

R	peak	1.0		I = 39
S	peak	-0.10		I = 50
S	end	0.0		I = 55
Q	peak	-0.1		I = 30
Q	start	0.0		I = 27

| QR | slope | 24.4 mV/s |
| RS | slope | -20.0 mV/s |

P	peak	0.17		I = 16
P	start	0.0		I = 5
P	end	0.0		I = 24

| Pu | slope | 3.2 mV/s |
| Pd | slope | -4.4 mV/s |

T	peak	0.3		I = 96
T	start	0.0		I = 56
T	end	0.0		I = 120

| Tu | slope | 1.5 mV/s |
| Td | slope | -2.5 mV/s |

RR interval = 0.8 Sec.

R	peak	1.1		I = 42
S	peak	-0.12		I = 50
S	end	0.0		I = 55
Q	peak	-0.11		I = 35
Q	start	0.0		I = 34

| QR | slope | 34.5 mV/s |
| RS | slope | -30.8 mV/s |

P	peak	0.19		I = 16
P	start	0.0		I = 5
P	end	0.0		I = 24

| Pu | slope | 3.4 mV/s |
| Pd | slope | -4.7 mV/s |

T	peak	0.52		I = 96
T	start	0.0		I = 56
T	end	0.0		I = 117

| Tu | slope | 2.6 mV/s |
| Td | slope | -4.5 mV/s |

5

forms the basis of template formation.

It is better than the statistical methods of data compression like 'Reduction of Sampling Rate' which treat the ECG wave as a bunch of statistic without taking into consideration the underlying physiological basis of the ECG. However, the Tamplate Method of ECG Data compression finds use in clinical situations where the ECG wave does not change rapidly. However the limitation of this method is the comparitively large processing time required for template formation.

CHAPTER - IV

MOVING SIGN SLOPE METHOD FOR DATA REDUCTION
OF ELECTROCARDIOGRAMS

4.1 MOVING SIGN SLOPE METHOD

The moving sign slope method for data reduction of electrocardiograms is a statistical method for data reduction/compression and takes into account the nature of the curve.

The method for data reduction produces compressed data in real time. It is based on the rectified minimum perimeter polygonal approximation. Compression òf digital electrocardiograms (ECG) has received great attention since computer application to electrocardiography for various reasons data.

Data compression is very effective in increasing the efficiency of automatic computer analysis of electrocardiograms [4 - 9]. Especially under the constraints of a microprocessor based system, efficient data compression preprocessing is critical to real-time operation [4, 5, 9]. Compact data storage with easy retrieval, without introducing clinically significant distortion, is important for reducing data storage expense. Long-term digital recording and 'real-time analyzers' can benefit, too, from greatly increased storage capacity on their instrument's magnetic tape [13]. To transmit digitized ECG data from various locations to a center for further processing, a data compression scheme is necessary if public telephone lines are to be used economically [10-11].

The basic concept underlying the design of MSS is that an excellent approximation of an electrocardiographic waveform is obtainable from the piecewise linear segments inside the corridor formed by the two functions

which are less than and greater than the given waveform by amounts ε, as shown in Fig. 2.3.

The concept yields polygonal curves for which the maximum difference between the given waveform and the polygonal approximation is equal to or less than a specified value.

4.2   THE MSS ALGORITHM

To describe polygonal approximations and associated algorithms, the following notation are used. The original sampled data to be approximated are presented by the discrete sequence $DS(k)$ where the integer k is the sample number. The allowable error in the approximation is $\pm \varepsilon$, having been specified by the user. The polygonal approximations will allow reconstruction of the original data within $\pm$ and a specific reconstruction is symbolized by the discrete sequence $\widehat{DS}(k)$, for which

$$| DS(k) - \widehat{DS}(k)| \leqslant \varepsilon , \quad k = 1,2,3, \dots$$

The approximations themselves are described by a subset of the original data in which a number of the original data points have been deleted. If the original data consists of the (sample number, amplitude) pairs

$$(k, DS(k)), \quad k = 1,2,3, \dots)$$

the approximation of represented by the reduced data

$$(k, w(k)), \quad k = S_1, S_2, \dots)$$

The (sample number, amplitude) pairs of the reduced data are termed vertices of the original data. Between any two vertices, the data are approximated by points on the straight line joining the two vertices.

A.   Sideline Criterion, MSS-I

Let   k = s   denote the sample number of the most recently found vertex of the data. When beginning the approximation algorithm, the first data point is a vertex. Upon receipt of a later sample at k   j, the normalized slopes of the two lines joining   w(s)   to   w(j) + ε   and   w(j) - ε   are calculated. These slopes are

$$m(j, \varepsilon) = \frac{DS(j) + \varepsilon - DS(s)}{j - s}$$

and

$$m(j, -\varepsilon) = \frac{DS(j) - \varepsilon - DS(s)}{j - s}$$

and indicated in Fig. 4.1.

As each new data point is processed, the current smallest value of $m(j, \varepsilon)$ is stored as m_1, and the current largest value of $m(j, -\varepsilon)$ is stored as m_2. The numbers m_1 and m_2 thus store the maximum and minimum slopes that a line from   s   may have and still pass within $\pm \varepsilon$ of subsequent data points. Whenever,

$$m_2 > m_1$$

as is the situation in Fig. 4.1 k = j + 1, the immediately previous data point is selected as a vertex.

B.   Centreline Criterion, MSS-2

The MSS-1   algorithm can exhibit approximation error in excess of ε , but these errors must be less than 2ε . This is illustrated in Fig. 4.2(a). In this figure the error associated with the line segment joining vertices   s   and   j + i - 1   is more than ε , but since δ is less than ε, the error (the length of δ + ε) is less than 2ε . Some improvement in efficiency is obtained with the
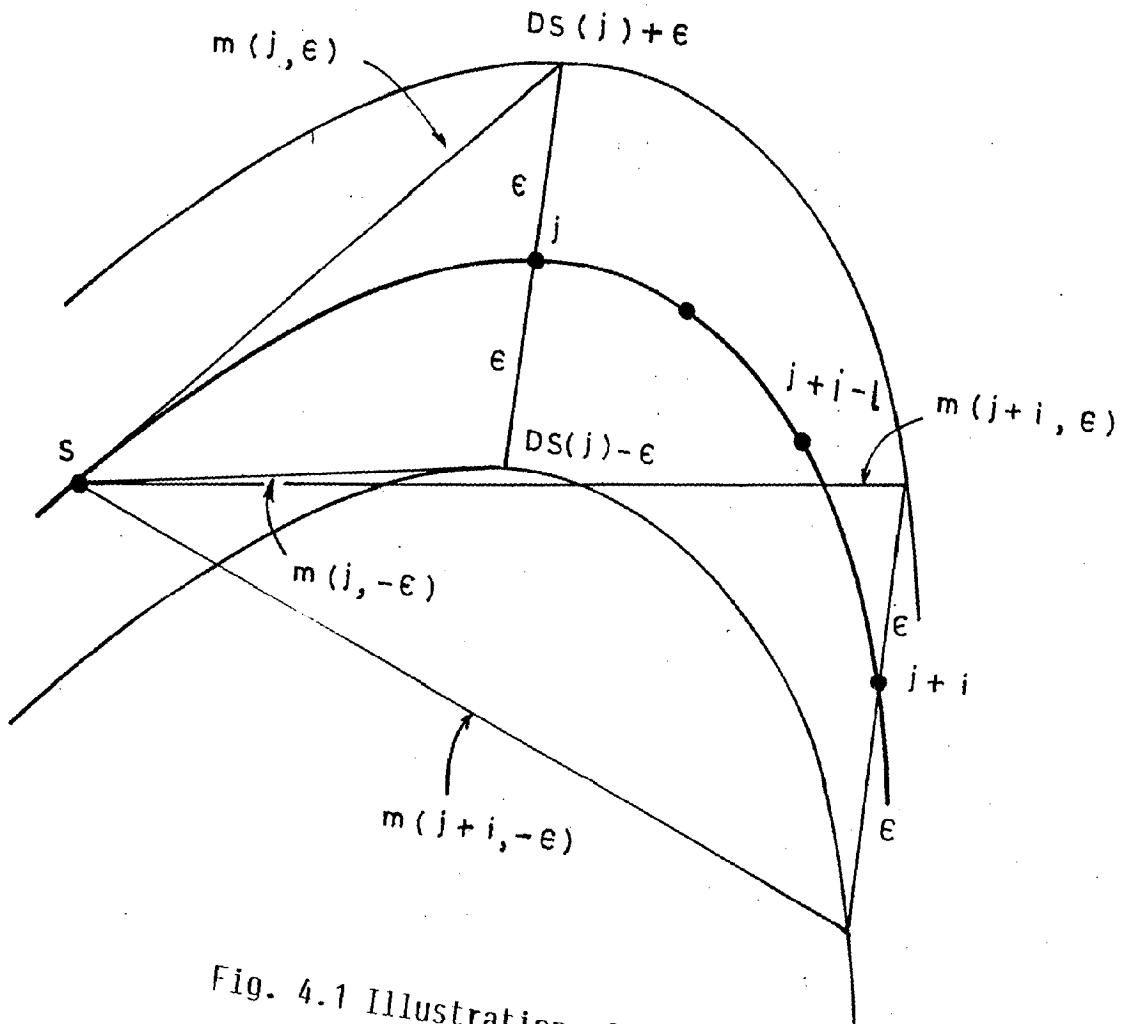
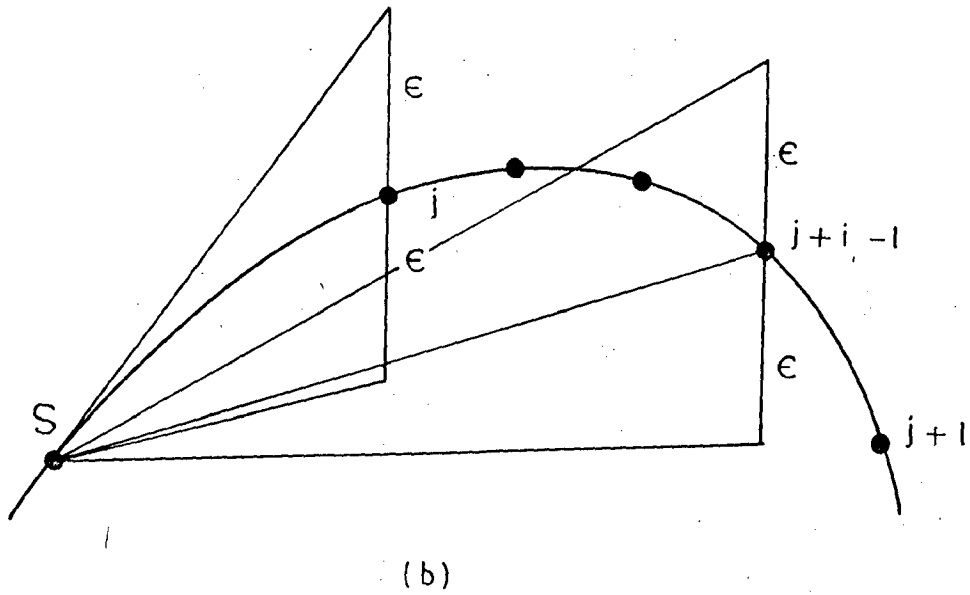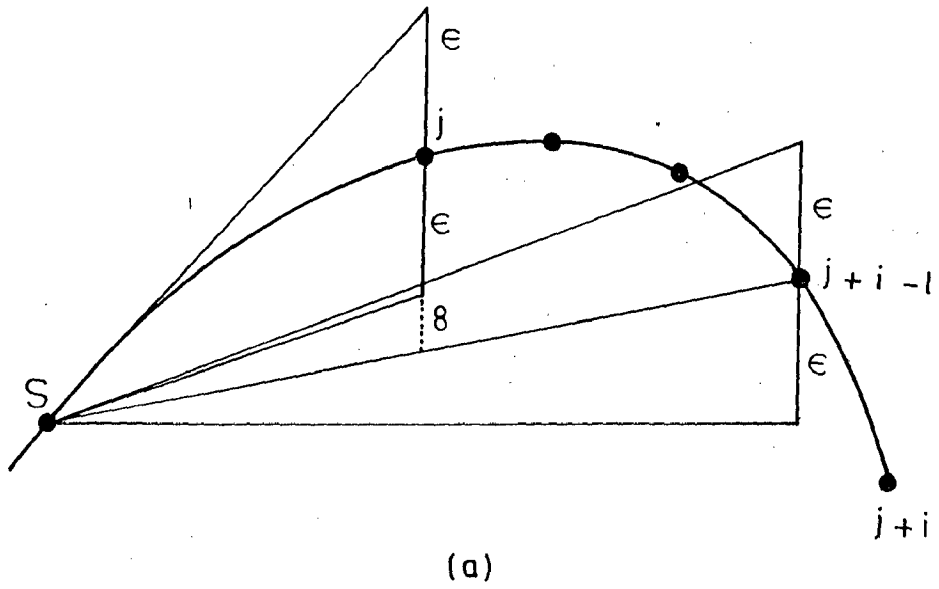Fig. 4.1 Illustration of MSS-1 Algorithm

(a)

(b)

Fig. 4.2 Distinction between MSS-1 and MSS-2 performances
(a) MSS-1 compression (b) MSS-2 compression

MSS-2 algorithm in which the slope calculation

$$m(j, 0) = \frac{DS(j) - DS(s)}{j - s}$$

is made in addition to the slopes $m(j, \epsilon)$ and $m(j, -\epsilon)$ which are used in determining the current maximum and minimum slopes of an approximating line from $k = s$ to $k = j$ with errors less than $\pm \epsilon$.

The tests for whether or not the points between $k = s$ and $k = j + i$ may be adequately approximated by a straight line between $(s, m(s))$ and $((j + i), m(j + i))$ are

$$m(j + i, 0) < m_1$$

and

$$m(j + i, 0) > m_2$$

as indicated in Fig. 4.3. If either of these conditions is violated, then the immediately previous data point is a vertex.

Figure 4.2 shows distinctions in the performance of MSS-1 and MSS-2. In Fig. 4.2(a), using MSS-1, the vertex at $k = j + i - 1$, selected after the vertex at $k = s$ exhibits an error of magnitude $\delta + \epsilon < 2\epsilon$ at an intermediate data point at $k = j$. The error of the approximation of intermediate data points has magnitude less than or equal to with MSS-2, as shown in Fig. 4.2(b).

4.3 FEATURES OF THE MSS PROGRAM

The MSS program whose flowchart is shown in Fig. 4.4 has the following advantages:

(i) The data can be compressed in real time, i.e. as the wave form is generated. The operation is scan-along, i.e., the output is generated as the input is received.
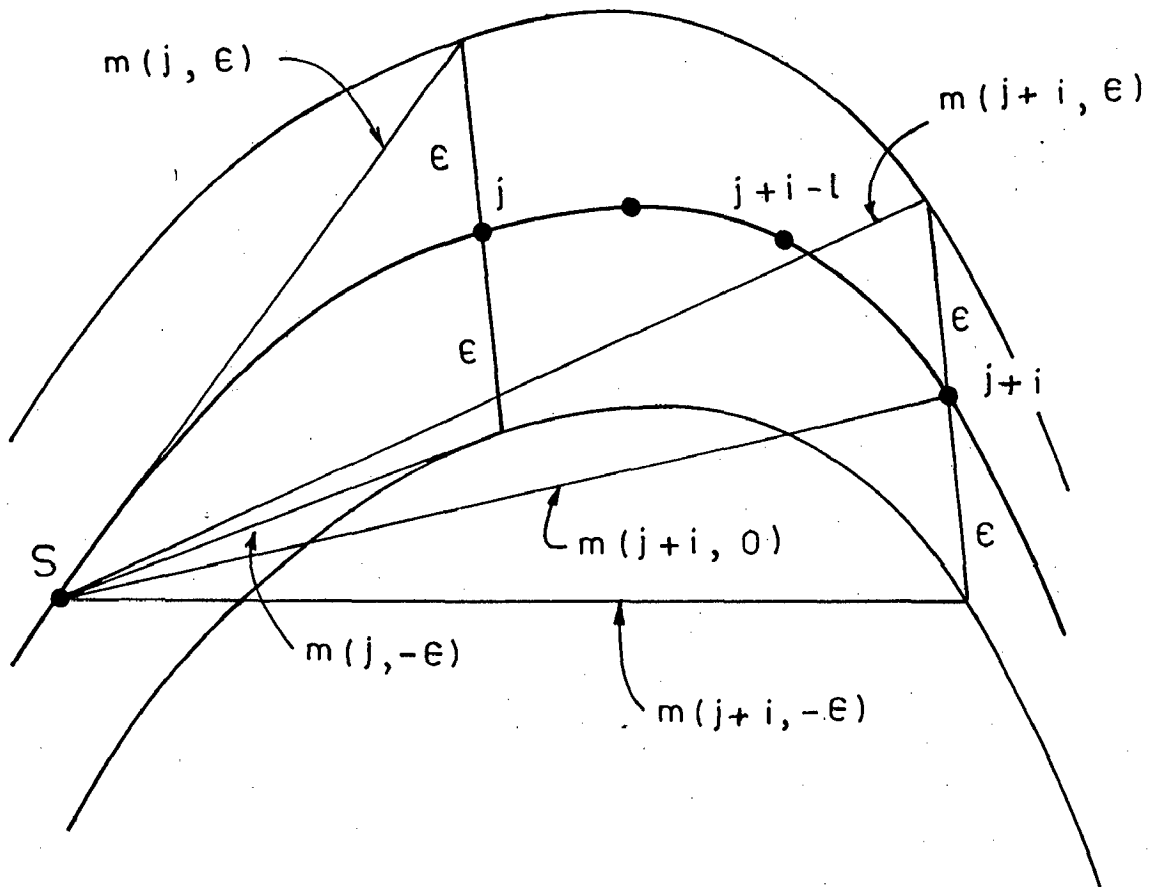
Fig. 4.3 Illustration of MSS-2 Algorithm

Fig. 4.4(a) MSS-1 Flow Diagram

```
                    ┌──────────────────────────┐
                    │  INPUT FIRST SAMPLE      │
                    │  STORE AS  DS(0)         │
                    └──────────────────────────┘
```

CALCULATE UPPER SLOPE

$$\text{SLOP1} = \frac{DS(J) - DS(ICOUNT) + EPS}{STA(J) - STA(ICOUNT)}$$

$ONEM \leq SLOP1$

UPDATE MIN. UPPER SLOPE
$ONEM = SLOP1$

CALCULATE LOWER SLOPE

$$\text{SLOP2} = \frac{DS(J) - DS(ICOUNT) - EPS}{STA(J) - STA(ICOUNT)}$$

$TWOM \geq SLOP2$

UPDATE MAX. LOWER SLOPE
$TWOM = SLOP2$

CALCULATE CENTER SLOPE

$$\text{SLOP0} = \frac{DS(J) - DS(ICOUNT)}{STA(J) - STA(ICOUNT)}$$

$ONEM > SLOP0 > TWOM$

SAVE DATA POINT;
$J = J - 1,\ DS(J)$
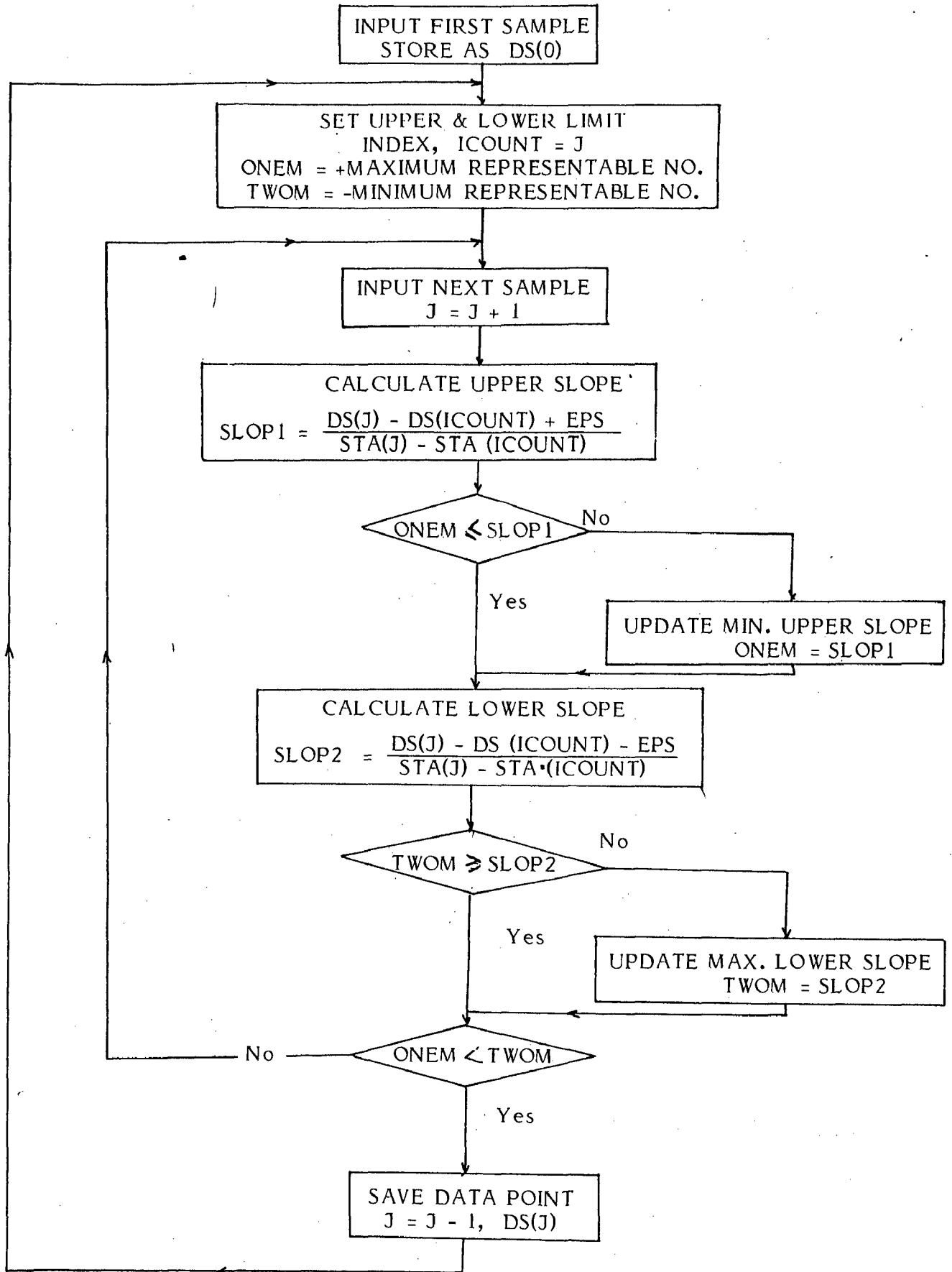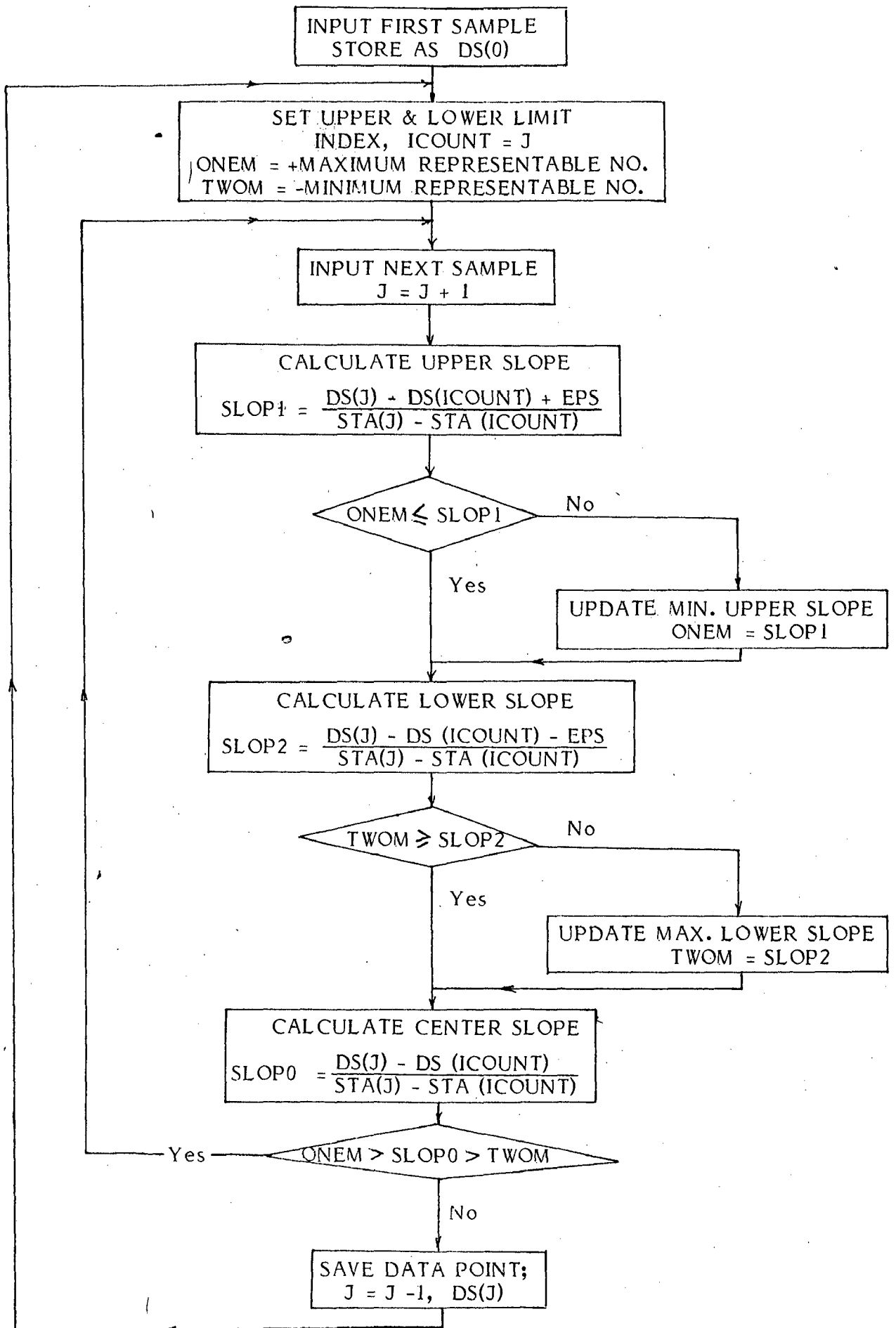
Fig. 4.4(b) MSS-2 Flow Diagram

(ii)   The original waveform can be reconstructed with adjustable fidelity depending on the requirements of the host system, by changing the allowable error limit.

(iii)   Reconstruction procedures are simple and straight forward.

(iv)   The compressed data are a subset of the digitized waveform data. Therefore, the original signal can be reconstructed perfectly in the limiting case of $\epsilon = 0.01$

4.4   RESULTS

The moving sign slope technique for data compression of ECGs signals was applied to the ECG signal available with various acceptable EPS values (giving rise to the corridor in which straight line approximation is possible). The results of the MSS-I Program as applied to the input ECG wave as shown in Fig. 4.5 for different allowable error limits are as follows:

Case I

No. of data points = 120

EPS = ϵ = 0.01

No. of compressed signal points = 14

Compressed signal data :

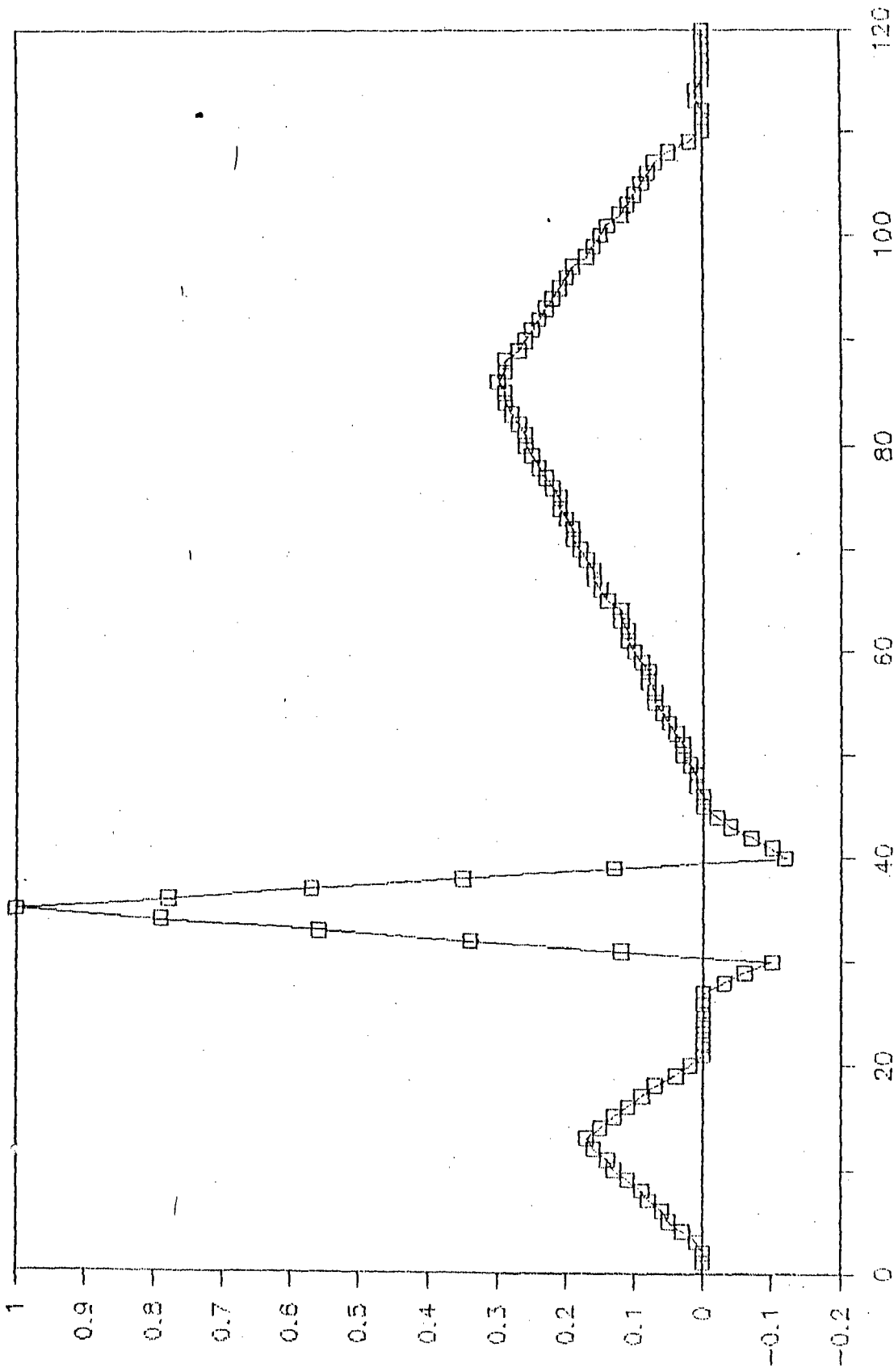| CSTA (I)	CS (I)
4	0.03
13	0.17
21	0.00
27	0.00
30	-0.10
35	1.00
39	0.13
40	-0.12
45	0.00
65	0.14
86	0.30
101	0.14
108	0.50
110	0.00

Data Compression Ratio = 120/14 = 8.5

Fig. 4.5 Input ECG Wave

Case 2

No. of data points = 120

EPS = ε = 0.02

No. of compressed signal points = 9

Compressed signal data :

CSTA (I)	CS (I)
14	0.15
23	0.00
28	-0.03
30	-0.10
35	1.00
40	-0.12
47	0.01
88	0.29
112	0.00

Data Compression Ratio = 120/9 = 13

Case 3

No. of data points = 120

EPS = ε = 0.03

No. of compressed signal points = 9

Compressed signal data :

CSTA (I)	CS (I)
14	0.15
24	0.00
29	-0.06
30	-0.10
35	1.00
40	-0.12
49	0.02
89	0.27
115	0.00

Date Compression Ratio = 120/9 = 13

Case 4

No. of data points = 120

EPS = ε = 0.04

No. of compressed signal points = 8

Compressed signal data :

CSTA (I)	CS (I)
15	0.13
26	0.00
30	-0.10
35	1.00
40	-0.12
53	0.05
90	0.26
117	0.00

Data Compression Ratio = 120/8 = 15

Case 5

No. of data points = 120

EPS = ε = 0.05

No. of compressed signal points = 7

Compressed signal data :

CSTA (I)	CS (I)
16	0.11
30	-0.10
35	1.00
40	-0.12
57	0.08
92	0.24
119	0.00

Data Compression Ratio = 120/7 = 17

Case 6

No. of data points = 120

EPS = ϵ = 0.06

No. of compressed signal points = 6

Compressed signal data :

CSTA (I)	CS (I)
17	0.09
30	-0.10
35	1.00
40	-0.12
69	0.17
94	0.22

Data Compression Ratio = 120/6 = 20

Case 7

No. of data points = 120

EPS = ϵ = 0.07

No. of compressed signal points = 5

Compressed signal data :

CSTA (I)	CS (I)
17	0.09
30	-0.10
35	1.00
40	-0.12
98	0.29

Data Compression Ratio = 120/5 = 24

Case 8

No. of data points = 120

EPS = ϵ = 0.08

No. of compressed signal points = 5

Compressed signal data :

CSTA (I)	CS (I)
18	0.70
30	-0.10
35	1.00
40	-0.12
93	0.23

Data Compression Ratio = 120/5 = 24

The results for the MSS-II Program as applied to the input ECG wave for different 'allowable error limits' EPS are as follows :

Case 1

No. of data points = 120

EPS = ϵ = 0.01

No. of compressed signal points = 7

Compressed signal data :

CSTA (I)	CS (I)
13	0.17
27	0.00
35	1.00
30	0.13
86	0.30
97	0.19
108	0.05

Data Compression Ratio = 120/7 = 17

Case 2

No. of data points = 120

EPS = ϵ = 0.02

No. of compressed signal points = 6

Compressed signal data :

CSAT (I)	CS (I)
13	0.17
28	-0.08
35	1.00
39	0.13
88	0.29
108	0.05

Data Compression Ratio = 120/6 = 20

Case 3

No. of data points = 120

EPS = ϵ = 0.03

No. of compressed signal points = 4

Compressed signal data :

CSTA (I)	CS (I)
13	0.17
28	-0.03
35	1.00
109	0.02

Date Compression Ratio = 120/4 = 30

Case 4

No. of data points = 120

EPS = ϵ = 0.04

No. of compressed signal points = 3

Compressed signal data :

CSTA (I)	CS (I)
14	0.15
20	-0.06
35	1.00

Data Compression Ratio = 120/3 = 40

Case 5

No. of data points = 120

EPS = ϵ = 0.05•

No. of compressed signal points = 3

Compressed signal data :

CSTA (I)	CS (I)
14	0.15
29	-0.06
35	1.00

Data Compression Ratio = 120/3 = 40

Case 6

No. of data points = 120

EPS = ϵ = 0.06

No. of compressed signal points = 2

Compressed signal data :

CSTA (I)	CS (I)
15	0.13
35	1.00

Data Compression Ratio = 120/2 = 60

Case 7

No. of data points = 120

EPS = ϵ = 0.07

No. of compressed signal point = 2

Compressed signal data :

CSTA (I)	CS (I)
15	0.13
35	1.0'

Data Compression Ratip = 120/2 = 60

Case 8

No. of data points = 120

EPS $= \epsilon = 0.10$

No. of compressed signal points = 2

Compressed signal data :

CSTA (I)	CS (I)
16	0.11
35	1.00

Data Compression Ratio = 120/2 = 60

As is seen from the graphs plotted with the compressed signal obtained by using the MSS-1 and MSS-2 programs, although data compression ratio tends to increase with increase in value of allowable error limit,, the reconstruction fidelity of the original ECG wave is greatly reduced with the MSS-1 program, EPS equal to 1 percent gives the best curve fit alongwith a data compression ratio of 14, as shown in Fig. 4.6(a) to 4.6(f).

However with the MSS-2 program the best wave reconstruction is again obtained with EPS equal to 1 percent although data compression achieved is greater of the order of 17, as shown in Fig. 4.7 (a) to 4.7(h).

4.5 DISCUSSION

The Moving Sign Slope (MSS) method for data compression, both the sideline and the centre line cases give a much higher data compression ratio of the order of 17:1 as compared to other statistical data compression methods.

However the wave reconstruction fidelity criteria is satisfied much better as the allowable error limit decreases-both for MSS-1 and MSS-2 programs and is best for EPS equal to 1 percent. This method for data

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Fig. 4.6 MSS-1 Results

COMPRESSED SIGNAL DATA —MSS

(c)

COMPRESSED SIGNAL DATA —MSS

(d)

Fig. 4.6 MSS-1 Results

COMPRESSED SIGNAL DATA —MSS

NO. OF POINTS =5, EPS=0.07

(e)

COMPRESSED SIGNAL DATA —MSS

NO. OF POINTS =5, EPS=0.08

(f)

Fig. 4.6 MSS-1 Results

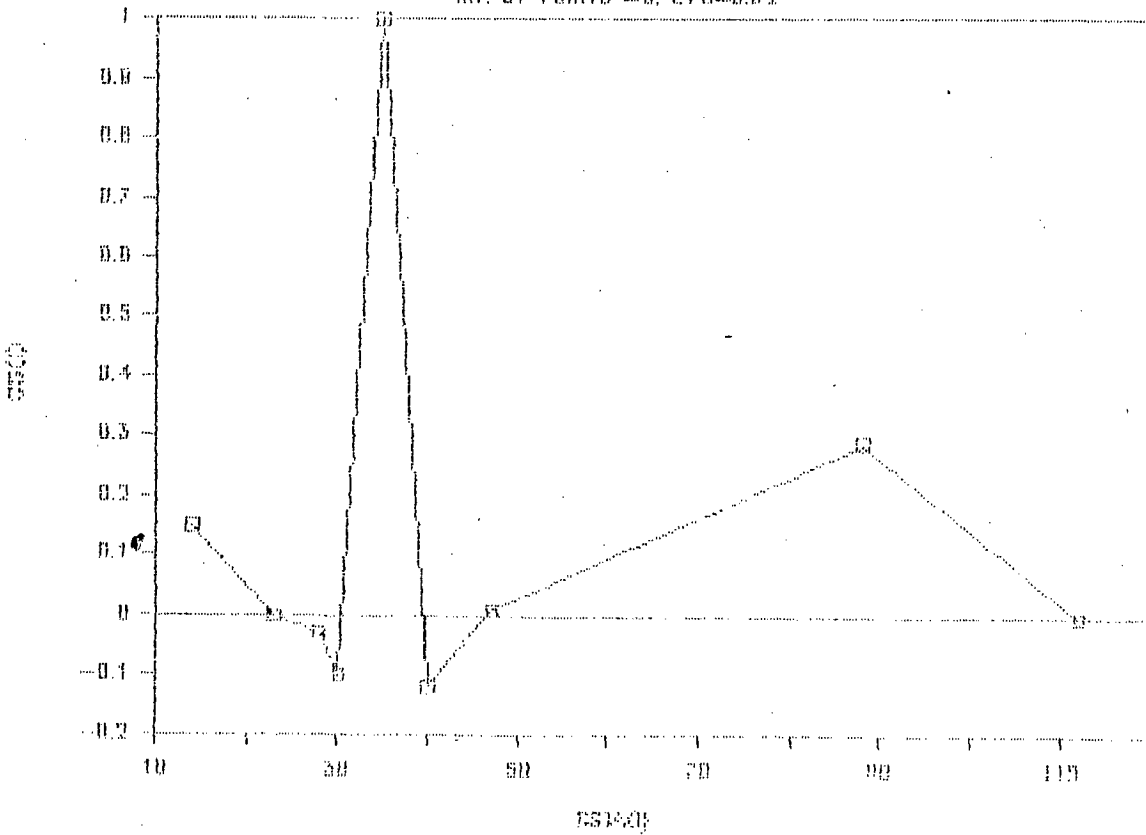## COMPRESSED SIGNAL DATA — MSS2

NO. OF POINTS = 7, EPS = 0.01
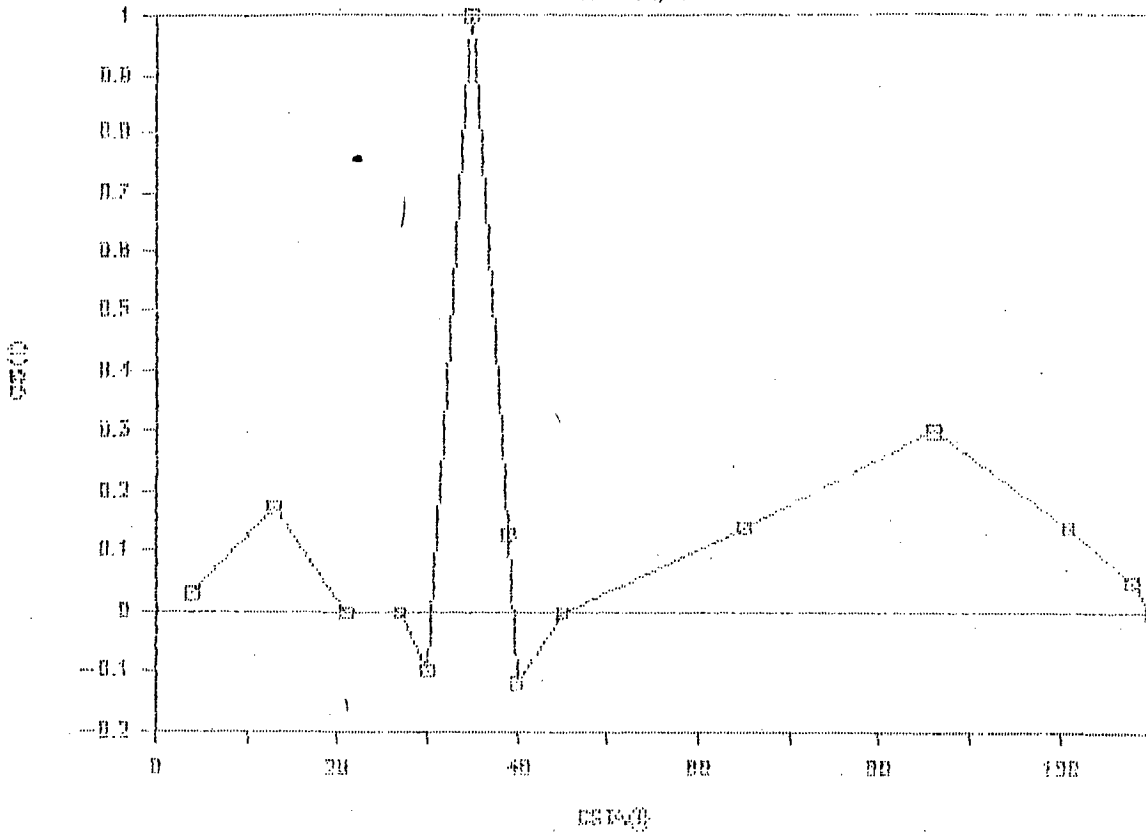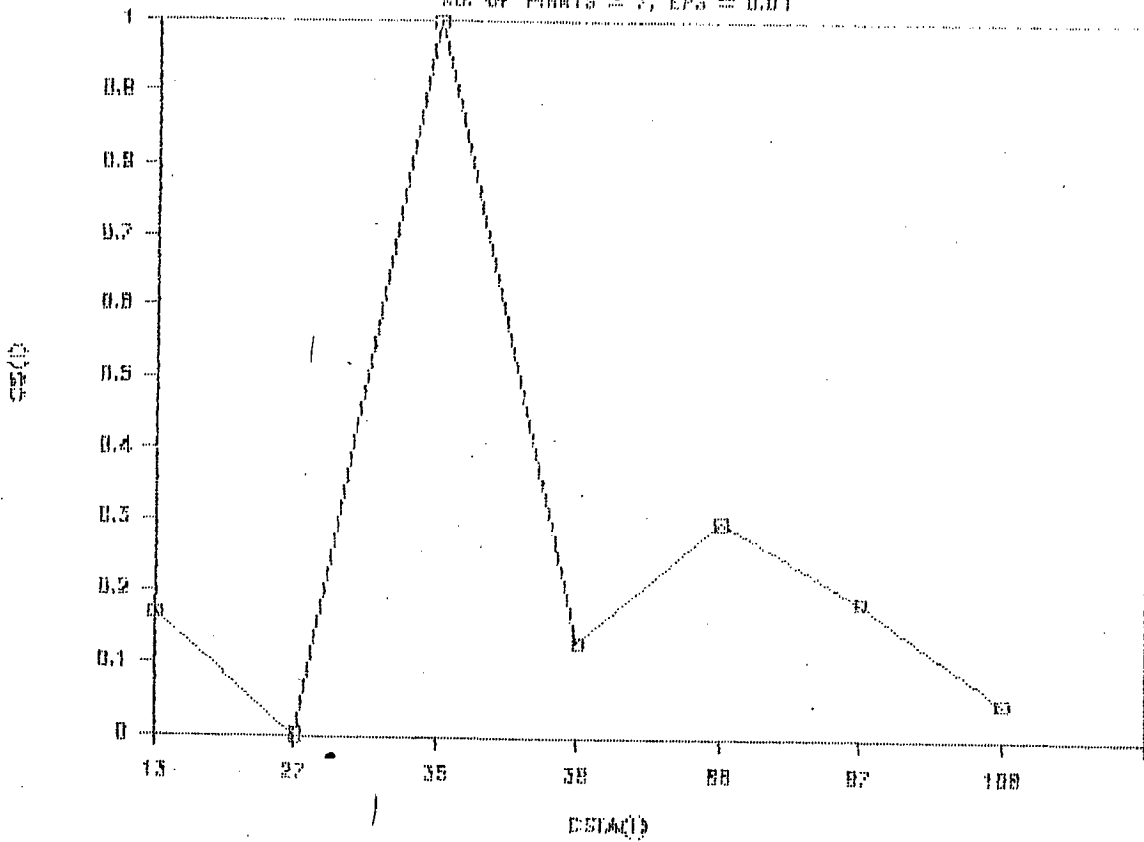
(a)

COMPRESSED SIGNAL DATA — MSS2
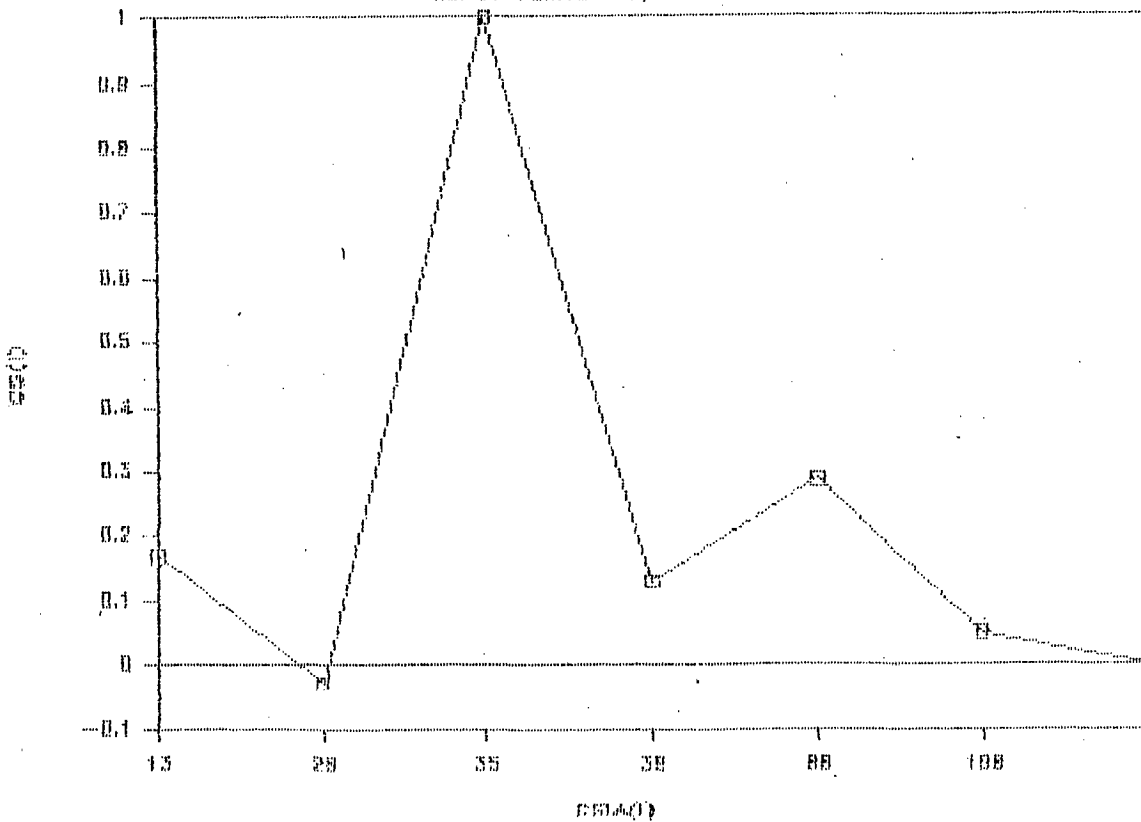
NO. OF POINTS = 8, EPS = 0.02

(b)
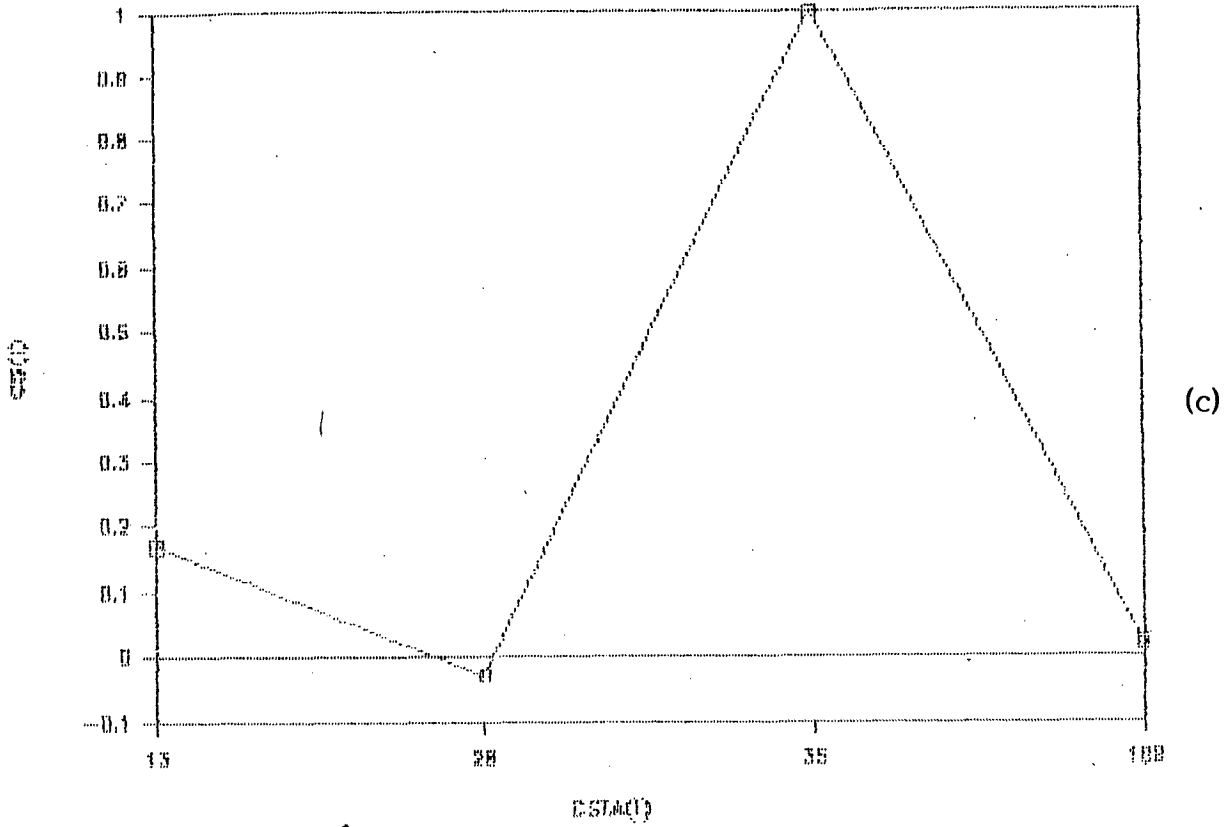
Fig. 4.7 MSS-2 Results

COMPRESSED SIGNAL DATA — MSS2
NO. OF POINTS = 3, EPS = 0.05

(e)
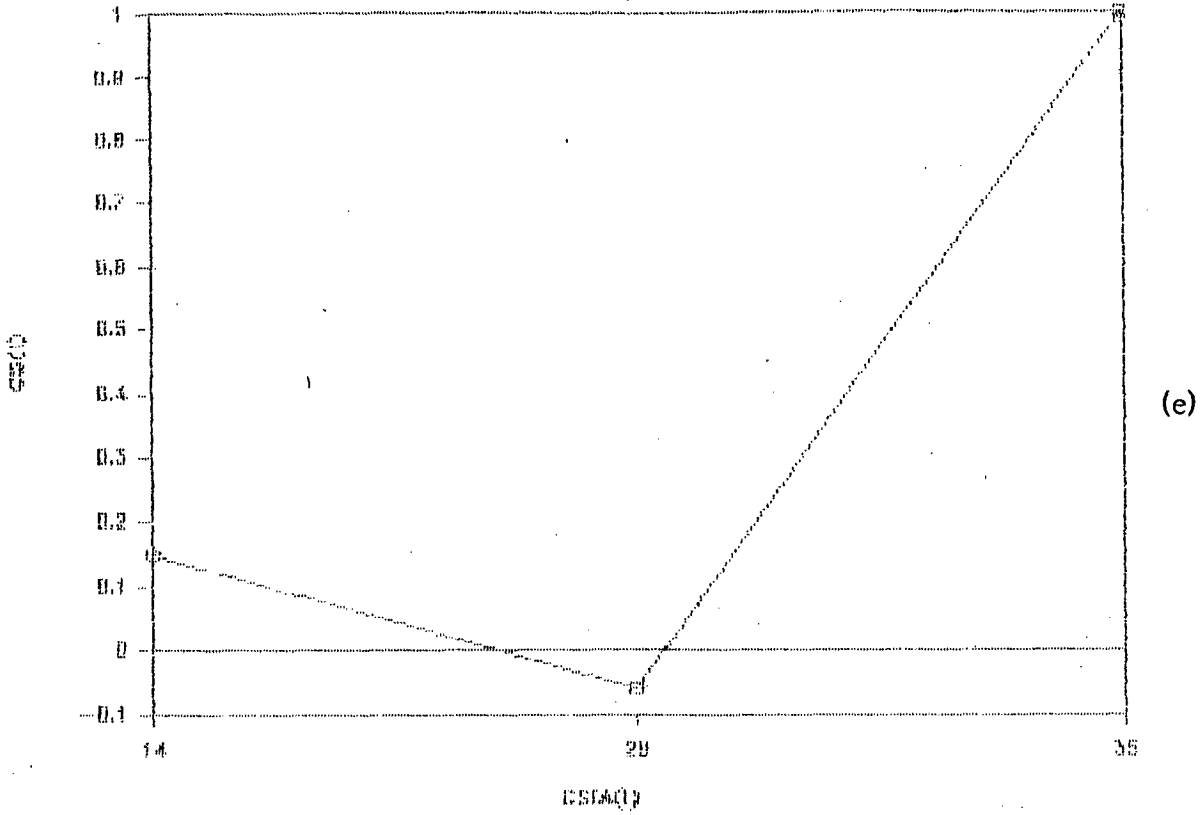
COMPRESSED SIGNAL DATA — MSS2
NO. OF POINTS = 2, EPS = 0.00

(f)

(g)
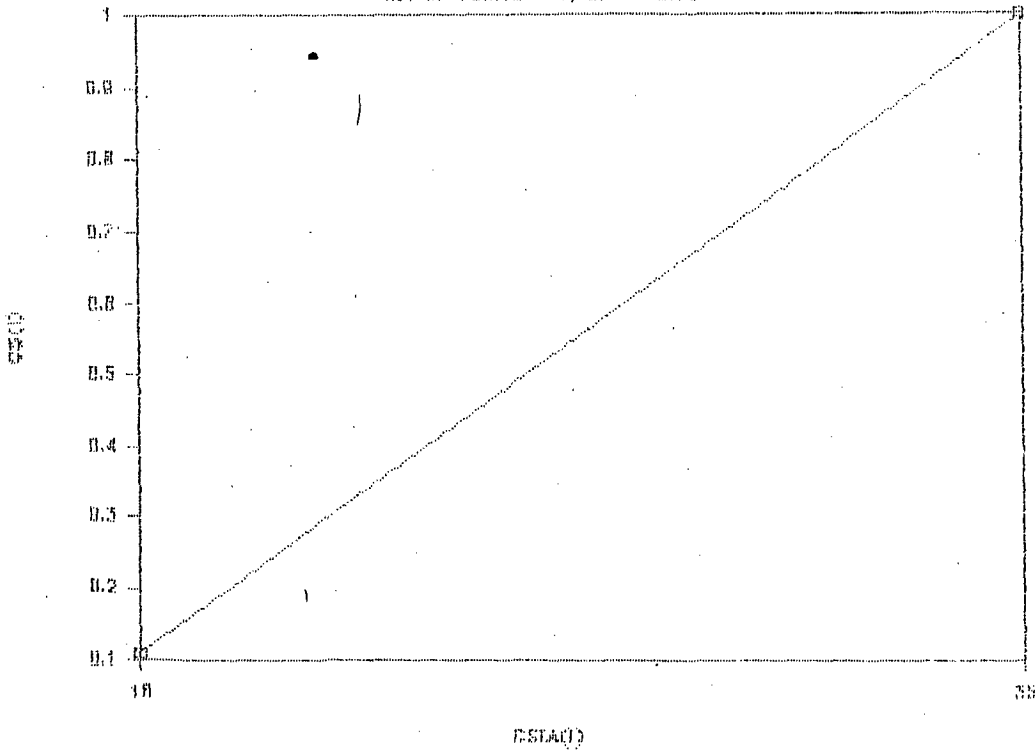
(h)

Fig. 4.7 MSS-2 Results

compression finds use in clinical situations where the data is to be stored. It is useful for storage and transmission of cardiograms. The processing time is very small as compared to other methods.

However the disadvantage of this method is that it has to store ECG data value repetitively while a logical method would save on this.

CHAPTER - V

CONCLUSIONS AND SUGGESTIONS FOR FUTURE WORK

5.1   CONCLUSIONS

In this dissertation work, two techniques and their algorithms of data compression of ECG signals have been developed. One of them, the template method for data compression involves making a template of the sample ECG wave by quantifying it with respect to it's wave segment parameters - QRS slopes, R peak and it's duration, PQ upslope and downslope etc. which are clinically significant. In this method, the slopes are the means of comparing the template ECG wave with the data ECG waves.

The moving sign slope method for data compression of ECG signals is used to approximate with a straight line within a corridor of the ECG wave which is determined by the maximum allowable error limit.

The attempt has been to bring into sharp focus the importance of "data compression" in clinical electrocardiography along with it's limitations. The conclusions of the work are as under :-

(i)   The algorithm developed for the template method is quite comprehensive as it is computing slope as well as area of various segments of the ECG wave. The slope has been used as a parameter for comparison between sample ECG wave and data waves.

(ii)   The template algorithm also takes into account the baseline shift and applies a correction for it. By computing the area under various wave segments, the total work done by the myocardial muscles during various phases of the ECG signal is aptly represented.

(iii)   The data compression achieved for the data used was 5:1 in case of

the template method. However, lack of available data base did not allow the algorithm to be tested for a larger number of cases.

(iv) The clinically significant parameters of slope, duration and peaks of various segments formed the basis of comparison of ECG waves with the template wave for data compression in the time domain.

(v) The template method for data compression of ECG signals is to be used in a specific clinical condition.

(vi) The Moving Sign Slope (MSS) method for data compression of ECG signals gives a straight line fit for data points lieing in the corridor which is governed by the maximum permissible error ε. Data compression achieved is much larger as compared to the data compression achieved by the template method for the sample ECG wave used, although their reconstruction is not too good.

(vii) The ECG signal required various clinical parameters to completely describe and reconstruct it and so data reduction techniques can be applied to only specific cases or clinical situations.

5.2 SUGGESTIONS FOR FUTURE WORK

Further work can be done to widen the scope and suggestions in this regard are as below :-

(i) The algorithm may be tested on a much larger no. of patients i.e. subjected to a much larger data base. Clinical tests will help to improve on the values of the thresholds and settings assumed here on the basis of published data and to modify the initial parameters accordingly.

(ii) On line data compression methods require that the processing time be as small as possible of the order of one or two ECG waves and this can be achieved by computing parameters that singularly describe an ECG wave.

(iii) The ECG wave to be properly described requires both logical and statistical information. So data compression techniques applied in parallel would help to achieve more reliable data compression.

(iv) The area criterion and slope of the ECG signals need to be logically linked with ECG wave generation processes in order to obtain a more physically true picture about the clinical relevance of these parameters.

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```
C        PORGRAM SAPA1
         DIMENSION STA(1000),DS(1000),CSTA(500),CS(500)
         OPEN(UNIT=1,FILE='MSS.DAT')
             OPEN(UNIT=2,FILE='MSS.OUT',STATUS='NEW')
C        N VALUES OF STATIONS(STA) AND DIGITIZED SIGNAL (DS),INPUT
C        M VALUES OF COMPRESSED SIGNAL (CSTA=STATION NUMBER,CS=SIGNAL ORDINATE)
         READ(1,*)N
C        EPS=TOLERANCE LIMIT
         READ(1,*)EPS
10       READ(1,*)(STA(I),I=1,N),(DS(I),I=1,N)
C        ICOUNT DEFINED AS VERTEX
         ICOUNT=1
             M=0
16       CONTINUE
C        J IS COUNTER FOR STATIONS
         J=ICOUNT
         ONEM=1.0E+20
             TWOM=-1.0E+20
12       J=J+1
         IF(J.GT.N)GO TO 14
C        M1=1.0 CORRESPONDS TO 45 DEGREES
             Q1=DS(J)-DS(ICOUNT)
             Q2=STA(J)-STA(ICOUNT)
             SLOP1=(Q1+EPS)/Q2
             WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'ONEM=',ONEM
             IF(ONEM.GT.SLOP1)ONEM=SLOP1
             WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'ONEM=',ONEM
             SLOP2=(Q1-EPS)/Q2
             WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'TWOM=',TWOM
             IF(TWOM.LT.SLOP2)TWOM=SLOP2
             WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'TWOM=',TWOM
             IF(ONEM.GT.TWOM)GO TO 12
             M=M+1
         WRITE(2,*)M
             J=J-1
             CSTA(M)=STA(J)
             CS(M)=DS(J)
             ICOUNT=J
             GO TO 16
14           CONTINUE
         WRITE(2,100)M
100      FORMAT(5X,37HNUMBER OF POINTS IN COMPRESSED SIGNAL,I5///,
     1   5X,36H COMPRESSED SIGNAL DATA IS AS FOLLOWS//)
             WRITE(2,105)
105      FORMAT(7X,7HCSTA(I),10X,5HCS(I))
             DO 17I=1,M
             WRITE(2,101)CSTA(I),CS(I)
101      FORMAT(5X,F10.4,5X,F10.4)
17       CONTINUE
         STOP
             END
```

```
C          PORGRAM MSS2
           DIMENSION STA(1000),DS(1000),CSTA(500),CS(500)
           OPEN(UNIT=1,FILE='MSS.DAT')
               OPEN(UNIT=2,FILE='MSS.OUT',STATUS='NEW')
C          N VALUES OF STATIONS(STA) AND DIGITIZED SIGNAL (DS),INPUT
C          M VALUES OF COMPRESSED SIGNAL (CSTA=STATION NUMBER,CS=SIGNAL ORDINATE)
           READ(1,*)N
C          EPS=TOLERANCE LIMIT
           READ(1,*)EPS
10         READ(1,*)(STA(I),I=1,N),(DS(I),I=1,N)
C          ICOUNT DEFINED AS VERTEX
           ICOUNT=1
               M=0
16         CONTINUE
C          J IS COUNTER FOR STATIONS
           J=ICOUNT
           ONEM=1.0E+20
               TWOM=-1.0E+20
12         J=J+1
           IF(J.GT.N)GO TO 14
C          M1=1.0 CORRESPONDS TO 45 DEGREES
               Q1=DS(J)-DS(ICOUNT)
               Q2=STA(J)-STA(ICOUNT)
           SLOPO=Q1/Q2
               SLOP1=(Q1+EPS)/Q2
               WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'ONEM=',ONEM
               IF(ONEM.GT.SLOP1)ONEM=SLOP1
               WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'ONEM=',ONEM
           SLOP2=(Q1-EPS)/Q2
               WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'TWOM=',TWOM
               IF(TWOM.LT.SLOP2)TWOM=SLOP2
               WRITE(2,*)'Q1=',Q1,'Q2=',Q2,'TWOM=',TWOM
               IF(ONEM.GT.SLOPO)GO TO 13
13         IF (SLOPO.GT.TWOM)GO TO 12
               M=M+1
           WRITE(2,*)M
               J=J-1
               CSTA(M)=STA(J)
               CS(M)=DS(J)
               ICOUNT=J
               GO TO 16
14             CONTINUE
           WRITE(2,100)M
100        FORMAT(5X,37HNUMBER OF POINTS IN COMPRESSED SIGNAL,I5///,
          1 5X,36H COMPRESSED SIGNAL DATA IS AS FOLLOWS//)
               WRITE(2,105)
105        FORMAT(7X,7HCSTA(I),10X,5HCS(I))
               DO 17I=1,M
               WRITE(2,101)CSTA(I),CS(I)
101        FORMAT(5X,F10.4,5X,F10.4)
17         CONTINUE
           STOP
               END
```

```
        COMMON A(2000)
        OPEN(UNIT=1,FILE='ECG.DAT')
        OPEN(UNIT=2,FILE='ECG.OUT',STATUS ='NEW')
C       TEMPLATE PROGRAM
        IM=1
            M=0
            I3=180
        AMIN=0.0
        AMAX=0.0
        X=0.6
        DT=0.005
        CFS=0.000444
        READ(1,*)APF1,APF2
        APF1=2.
        APF2=1.
        IJ=X/DT
            I1=4
            I2=I1+I3
            IJ=119
            WRITE (*,*) IJ
            IJ4=IJ*10
            WRITE (*,*) IJ4
        READ(1,*) (A(I),I=1,IJ4)
            WRITE (*,*) (A(I),I=1,IJ4)
100     CONTINUE
        DO 29 N=1,9
            WRITE(*,*) N
        N1=N*IJ
        DO 28 I=1,IJ
28      A(N1+I)=A(I)
29      CONTINUE
11      FORMAT(1X,'R peak',9X,F5.2,8X,'I =',I3)
12      FORMAT(1X,'R peak missing in 5 records')
        IF (IM-3)30,30,31
30      I1=20
        I2=I1+I3
31      CALL MAX (I1,I2,IM,AMAX)
        WRITE(2,11) AMAX,IM
        T1=10.*(AMAX-A(IM-3))/(0.015*25.)
        TH1=ATAN(T1)
        T2=10.*(A(IM+3)-AMAX)/(0.015*25.)
        TH2=ATAN(T2)
        DTHA=TH1-TH2
        DTHAM=DTHA*(1.5/AMAX)
        IF(DTHAM-100.)32,35,35
32      M=M+1
        IF(M-5)33,34,34
34      WRITE (2,12)
33      CALL MAX(I1,I2,IM,AMAX)
        WRITE(2,*)T1,T2,DTHAM,M
C       FIND NEXT R PEAK
35      I1=IM+100
        I2=IM+I3
        IM1=IM
            WRITE(2,*)I1,I2,IM1,AMAX
            CALL MAX(I1,I2,IM,AMAX)
        IM2=IM
        IX=IM2-IM1
```

```
        WRITE(2,*)I1,I2,IM2,AMAX
      XE=0.005*IX
      SUM=0.
      IC=0
      J1=IM-60
      J3=IM-30
39    IF(0.10-A(J1))40,41,41
40    J1=J1+1
      GO TO 39
41    J2=J1+1
      IF(J2.GT.J3)GO TO 43
      DCA=ABS(A(J2)-A(J1))
      IF(DCA.GT.CFS)GO TO 42
      SUM=SUM+A(J1)
      IC=IC+1
42    J1=J2
      GO TO 39
43    COX=SUM/IC
      DO 45I=IM1,IM2
45    A(I)=A(I)-COX
        N1=IM2
        M=N1
      II=1
        N1=M-II
        DAL=0.002
      NZY=0
        WRITE(2,*)I1,I2,IM,AMAX
      CALL ACOM(N1,II,DAL,NZY,APF1,APF2,DT,RAPF,SAN,TZRSS,TZRSE,I1,I2,
     1 IM2)
C     TO FIND S
C     S START AND S END
        WRITE(2,*)RAPF,SAN,TZRSS,TZRSE,I1,I2
      CALL MIN(I1,I2,IM,AMIN)
C
      SMIN=AMIN
      INS=IM
23    FORMAT(1X,'S peak',9X,F5.2,8X,'I =',I3)
      WRITE(2,23) SMIN,INS
25    FORMAT(1X,'RAPF=',F10.4,' SAN=',F10.4,' TZRSS=',F10.4,' TZRSE=',
     1 F10.4)
      WRITE(2,25) RAPF,SAN,TZRSS,TZRSE
      NZY=0
      II=-1
      N1=M-II
      CALL ACOM(N1,II,DAL,NZY,APF1,APF2,DT,RAPB,QAN,TZRQE,TZRQS,I1,I2,
     1 IM2)
      IS=I1
        I1=I2
        I2=IS
      CALL MIN(I1,I2,IM,AMIN)
        WRITE(2,*)I1,I2,IM,AMIN
      QMIN=AMIN
      INQ=IM
26    FORMAT(1X,'Q peak',9X,F5.2,8X,'I =',I3)
      WRITE(2,26) QMIN,INQ
88    FORMAT(1X,'RAPB=',F10.4,'QAN=',F10.4,'TZRQS=',F10.4,'TZRQE=',
     1 F10.4)
      WRITE(2,88) RAPB,QAN,TZRQS,TZRQE
```

```
C          TO FIND P-PEAK    |
           NR = TZR QS*RDT
           I2 = IM2 + NR-6
           I1 = IZ- 24
           (ALL MAX (I1,I2,IP, PMAX)
           WRITE (2,51) PMAX, IP
51         FORMAT (1X, 'P-PEAK', 9X, F10.4, 8X, 'I=', I3)
C          TO FIND AREA UNDER P-WAVE
           N1 = I2
           II = -1
           CALL AREA (N1,II,APF1,DT,AP,NS)
           WRITE (2,52) N1,NS,AP
           SPU = RDT*(A(IP)-A(N1)/(IP-N1)
           SPD = RDT*(A(NS)-A(IP))/(NS-IP)
           WRITE (2,53) SPU, SPD 2X,
52         FORMAT (1X,'START STN=', I5,2X 'LAST STN.=', I5,2X,
                     'AREA OF P-WAVE=', F10.4)
53         FORMAT (1X, 'UP-SLOPE P-WAVE=', F8.3, 2X,
                     'DOWN-SLOPE P-WAVE=', F8.3/)
C          TO FIND T-PEAK
           NR = TZ RSE * RDT
           I1 = IM1 + NR + 36
           I2 = I1 + 40
           CALL MAX (I1, I2, IT, TMAX)
           WRITE (2,54) TMAX, IT
54         FORMAT (WX, 'T-PEAK', AX, F10.4, 8X, 'I=', I3)
C          TO FIND AREA UNDER T-WAVE
           N1 = I1
           II = 1
           CALL AREA (N1, II, APF1, DT, AT, NS)
           WRITE (2,55) NS, N1, AT
           STU = RDT*(A(IT)-A(NS))/(IT-NS)
           STD = RDT*(A(N1)-A (IT))/(N1-IT)
           WRITE (2,56) STU, STD
55         FORMAT (IX,1START STN.=', IS,2X, 'LAST STN.=', I5,2X,
                     AREA OF T-WAVE=', F10.4)
56         FORMAT (IX, 'UP-SLOPE T-WAVE=', F8.3, 2X,
                     'DOWN-SLOPE T-WAVE=', F8.3/)
           CLOSE(UNIT=1)
           CLOSE(UNIT=2)
           END
           SUBROUTINE ACOM(N1,II,DAL,NZY,APF1,APF2,DT,RAP,AN,TZRS,TZRE,
      1 I1,I2,IM2)
C          R AREA FORWARD/BACKWARD OF R PEAK
           COMMON A(2000)
           SA=0
 3         N1=N1+II
           N=N1+II
           A1=0.5*(A(N)+A(N1))*DT
                SA=SA+A1
           APC=A1*100.0/SA
           IF(APC-APF1)4,4,3
 4         IF(APC)7,8,8
 7         RAP=SA-A1
           I1=N1                |
           AN1=N1-IM2
```

```
            TZRS=AN1*DT
            GO TO 11
  3         AN1=N-IM2
            I1=N
            TZRS=AN1*DT
               RAP=SA
  11        N1=N1-II
C        S/Q AREA COMPUTATION
            SA=0
  13        N1=N1+II
            N=N1+II
            A1=0.5*(A(N)+A(N1))*DT
                DA=A(N)-A(N1)
                IF (ABS(DA)-DAL)5,5,6
  5         NZY=NZY+1
            IF (NZY-2)6,9,9
  9         WRITE(2,51)
  51        FORMAT(1X,'NEGATIVE PEAK IS PROBABLY MISSING')
  6         SA=SA+A1
            APC=A1*100.0/SA
            IF(APC-APF2)14,14,13
  14        IF(APC)18,18,17
  17        AN=SA-A1
            AN1=N1-IM2
            I2=N1
            TZRE=AN1*DT
               GO TO 21
  18        AN=SA
            AN1=N-IM2
            I2=N
            TZRE=AN1*DT
  21        RETURN
            END
            SUBROUTINE MAX(I1,I2,IM,AMAX)
            COMMON A(2000)
            AMAX=A(I1)
            J=I1
  1         J=J+1
            IF(J-I2)2,2,4
  2         IF(AMAX-A(J))3,1,1
  3         AMAX=A(J)
            IM=J
            GO TO 1
  4         RETURN
            END
            SUBROUTINE MIN(I1,I2,IM,AMIN)
            COMMON A(2000)
            AMIN=A(I1)
               IM=I1
            J=I1
  1         J=J+1
            IF(J-I2)2,2,4
  2         IF(A(J)-AMIN)3,1,1
  3         AMIN=A(J)
            IM=J
            GO TO 1
```

```
4       RETURN
        END
        SUBROUTINE AREA (N1,II,SPF1,DT,SA,NS)
C       FIND P/T AREA
        COMMON A(2000)
        DAM = 0.001
        SA = 0
3       N1 = N1 + II
        N  = N1 + II
        DA = A(N) - A(N1)
        IF (ABS (DA) - DAM) 6, 6,5
5       NS = N1
6       A1 = 0.5 * (A (N) + A (N1)) * DT
        SA = SA + A1
        APC= A1 * 100./SA
        IF (APC - APF1) 4,4,3
4       N1 = N
        RETURN
        END
```

```
C       CONTINUOUS MONITORING OF E.C.G.
        OPEN (UNIT = 1, FILE ='ECGC. DAT')
          OPEN (UNIT = 2, FILE ='ECGC.OUT:STATUS = 'NEW')
          COMMON A (2000)
          READ (1,*) ER,ER,ET,EX
          READ (1,*) DT,RR,APF1,
          READ (1,*) X,CFR,
          RDT = 1./DT
          IX = RDT * X          IX2 = IX*1.2
          IJ = RDT *RR          IX8 = IX*0.8
          IJ3 = 3*IJ
          CF = 1.
          READ (1,*) (A(I), I=1, IJ)
          SR = 0.5 * (SRU-SRD)
          SP = 0.5 * (SPU-SPD)
          ST = 0.5 * (STU-STD)
          ER = 0.01*ER
          EP = 0.01*EP
          ET = 0.01*ET
          EX = 0.01*EX
          SRF = SR*(1.+ER)
          SRN = SR*(1.-ER)
          SPP = SP*(1.-EP)
          SPN = SP*(1.-EP)
          STD = ST*(1.+ET)
          STN = ST*(1.-ET)
          RRP = RR*(1.+EX)
          RRN = RR*(1.-EX)
          IF (IX2-IJ) 21,22,22
21      IXL = IJ
        GO TO 23
22      IXL = IX2
23      DO 24 N = 1,2
        NI= N*IJ
          CF = CF-CFR
          DO 24 I= 1,IJ
          A(N1+I) = CFR*A(I)
24      CONTINUE
        I3 = IXL+IXL/2
          I1 = 4     *
          I2 = I1+I3
          CALL MAX (I1,I2,IM1,RMAX C)
          WRITE (2,*) RMAXC,IM1
        SRUC = 0.3333* (RMA X C - A(IM1-3))*RDT
        SRDC = 0.5* (SRUC-SRDC)
          IF (SRC.LT.SRP.AND.SRC.GT.SRN)GO TO 25
          SRR = SRC/SR
          WRITE (2,1)SRUC/SRDC/SRC,SRR
      1 FORMAT (1X,'UPR SLOPE = 'F8.3,2X,'DN R-SLOPE =',F8.3,2X,'AV. R SLOPE
          =',F8.3,2X,'RATIO = ',F6.3)
C       FIND SECOND R-PEAK
25      I1 = IM1+I3
        I2 = IM1+I3
          CALL MAX (I1,I2,IM,2,RMAXC)
          WRITE (2,*) RMAXC, IM2
          RRC = IM2-IM1
          RRR = RRC/RR
          IF (RRC.LT.RRP.AND.RRC.G TO RRN)GO TO 29
          ERR = 100. - 100.*RRR
          AERR = ABS (ERR)
        WRITE (2,2) RRC
2       FORMAT (IX,'R-R INTERVAL = ',F6.3)
        IF (ERR) 26,27,28
26      WRITE (2,3) AERR
        60 TO 29
27      WRITE (2,4)
```

```
         GO TO 29
28       WRITE (2,5) ERR
 3       FORMAT (1X,'PULSE SLOW BY = ',F6.2,2X,'PERCENT')
 4       FORMAT (1X,'NO CHANGE IN PULSE RATE')
 5       FORMAT (1X,'PULSE FAST BY = ',F6.2,2X,'PERCENT')
C        TO FIND T-PEAK AND AVERAGE
C        T SLOPE 'ST'
29       I1 = IM1+36
         I2 = I1+50
             CALL MAX (I1,I2,IT,TMAXC)
             WRITE (2,*) TMAXC,IT
             N1 = I1
             II =1
             CALL AREA (N1,II,APF1,DT,AT,NS)
             WRITE (2,*) NS,N1,AT
             STUC = RDT * (A(IT) - A(NS))/IT-NS)
         STDC = RDT * (A(N1) - A(IT))/N1-IT)
             STC  = 0.5 * (STUC - STDC)
             WRITE (2, *) STUC, STDC, STC
             IF (STC.LT.STD.AND.STC.GT.STN) GO TO 31
             STR = STC/ST
             WRITE (2,6) STUC,STDC,STC,STR
 6       FORMAT (1X, 'UP T-SLOPE =', F8.3,2X'DNT
                 SLOPE =',F8.3,2X,1AVT SLOPE =',
                            F8.3,2X,'RATIO =', F6.3)
         TO FIND P PEAK AND AVERAGE P SLOPE'S
31       I2 = IM2-I1
         I1 = I2-24
             CALL MAX (I1,I2,IP,PMAXC)
             WRITE (2,*) PMAXC, IP
             N1 = I2
             II = -1
             CALL AREA (N1,II,APF1,DT,AP,NS)
             WRITE (2,*) N1,NS,AT
             SPUC = RDT*(A(IP)- A(N1))/(IP-N1)
             SPDC = RDT*(A(NS)- A(IP))/(NS-IP)
             SPC = 0.5* (SPUC-SPDC)
             WRITE (2,*) SPUC,SPPC,SPC
             IF (SPC.LT.SPP.AND.SPC.GT.SPN)
             GO TO 32
             SPR = SPC/SP
             WRITE (2,7) SPUC,SPPC,SPC,SPR
 7       FORMAT (1X,'UP P-SLOPE=' F8.3,2X,
                 'DNP-SLOPE=', F8.3,2X, 'AV P-SLOPE
                            =',F8.3,2X, 'RATIO = ', F6.3)
C        BEGIN PROCESSING NEXT EC6 WAVE
C        SET
32       IF (CF) 51,51,23
51       CLOSE (UNIT=1)
         CLOSE (UNIT=2)
             END
```