

HIGH SPEED MINICOMPUTER ANALYSIS OF CARDIAC ARRHYTHMIA
IN 24 HOUR HOLTER TAPE RECORDINGS

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ABSTRACT

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David H. Wong

Ambulatory recording of cardiac activity is a powerful diagnostic method for the detection and analysis of cardiac arrhythmia. Its use has been limited, however, by the size and complexity of the task of analysing 24 hours of ECG tape recording produced for each patient under surveillance. Although analog and hybrid-tape scanning systems have been developed to facilitate this task, each tape analysis still requires considerable operator intervention. This thesis describes a fully digital system designed to provide accurate quantitative information about the occurrences of arrhythmia, while significantly reducing operator overhead. The system is based on the use of a small minicomputer that reads and analyse the ECG tapes directly at 60 times real time, without the use of a hardware preprocessor. A unique aspect of the software design is the partitioning of the analysis into on-line and off-line functions in order to achieve the analysis speed potential of hybrid systems, while retaining the flexibility and precision that characterize digital systems. The results of extensive testing have demonstrated a high degree of reliability in beat detection and classification.

Preface

The topic of cardiac arrhythmia has been receiving growing attention in recent years. This is primarily due to the increased understanding of its origin and relation to more serious cardiac diseases. Arrhythmia detection has been based primarily on two electrocardiogram (ECG) analysis methods: hospital monitoring and ambulatory monitoring. While both methods promise to provide the large sample of ECG which is necessary for effective arrhythmia analysis, the latter method has the added advantage of being able to collect the data under the patient's normal living conditions. As ambulatory monitoring involves the recording of the ECG over a 12 to 24 hour period of time, a vast amount of data is collected for each patient. The analysis of a sample of this size is an appreciable task, especially in consideration of the complexity of some arrhythmias. Traditionally, the analysis is performed by trained operators with the aid of an analog/hybrid computer. The analysis process, however, has neither the efficiency nor the accuracy needed for wide acceptance and application of this technique. The analysis system developed in the research undertaken here was designed to overcome the drawbacks in contemporary ambulatory ECG recording analysis systems. The results obtained by this system compare very favourably to existing systems and it also has the potential of possible implementation based on a low cost microprocessor system.

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I wish to express my sincere gratitude and appreciation to Dr. Terrill Fancott, who provided the guidance, and much of the ideas that directed the development of this project. I also wish to thank Ms. H. Irving, who provided valuable ideas in the initial implementation of the development system, J. Blaison, who built the A/D interface, and G. Boast, for his valued advice and assistance in the software implementation. The development of the detection algorithms was done under the guidance of Dr. J. Lemire, Head of Cardiology of Notre Dame Hospital. Dr. R. Fontaine of Poulenc Ltd. advised me on the user requirements of the system, and supplied the tapes for the development and testing of the system. Mr. Kevin O'Mara gave me much appreciated assistance in the design of the system tests.

I wish to dedicate this thesis to my wife, who cheered as the pile of notes thickens on my desk and grieved as the pile increased even faster in the wastebasket.

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CHAPTER ONE

GENERAL INTRODUCTION

THE HOLTER SYSTEM

In 1957, the Holter Foundation Laboratory of Helena, Montana, reported on the development of a system which allows observation of a patient's heart activities over an extended period of time [Holter, 1961]. It was designed to provide a means of observing waveshape variations of the electrocardiogram (ECG) of an ambulant subject. A system based on these techniques is marketed by Avionics Research Products, Los Angeles, California. The Holter-Avionics ambulatory ECG recording system consists of a portable, self-contained electrocardiographic amplifier and a low speed precision tape recorder. The recorder is capable of recording up to 24 hours of ECG information under the patient's normal daily activities [Hinkle, 1967]. A log is carried by the patient to record all the changes in activities and other manifestations for the duration of the ECG recording. The analysis is performed on a composite scanner which superimposes complexes of the recorded ECG on an oscilloscope screen at 60 times real time.

The Holter-Avionics system opened up a new dimension in heart care. It provides a means of observing the activities of the heart over an extended period of time. With it, significant pathological changes brought on by daily activities which would

otherwise pass unnoticed can be detected. This is particularly useful for the detection of cardiac arrhythmia which is not normally apparent in conventional electrocardiograms. It has therefore proven to be a valuable technique in anti-arrhythmia drug research, where long term effects of the drug can be observed [Sheffield, 1976; Harrison, 1976]. It has also been successfully applied in the early detection of several cardiac diseases [Reiffel, 1977; Romero, 1976].

HOLTER MONITORING AND CARDIAC ARRHYTHMIA

The topic of cardiac arrhythmia has been the subject of intense interest in recent years. The term cardiac arrhythmia refers to a heart condition characterized by changes in the regular rhythm of the heart that are usually abrupt and, except in cases of severely degraded heart condition, transient. This interest is due to an increased understanding of their relationship to sudden death and as foretokens of more serious cardiac disorders [Macfarlane, 1974; Romero, 1976]. This, coupled with the development of medication for suppressing and even terminating arrhythmias, have made their detection very desirable.

It has been observed that approximately 20% of routine ECG contains arrhythmia [Macfarlane, 1974]. In short record ECG analysis only a small amount of data, usually 5 to 25 seconds, is collected, making detection of all but very severe arrhythmias improbable. The actual condition is difficult to assess in this

time span because it cannot be determined whether the record is shorter or longer than the average occurrence of the arrhythmia manifestation [LeBlanc, 1975; Prody, 1975; Geddes, 1971]. In order to effectively detect arrhythmia, records much longer than those obtained in routine ECG are needed. The Holter recording system is capable of providing the size of sample required for an accurate analysis of cardiac arrhythmia.

A 24 hour ECG recording contains approximately 100,000 to 150,000 heartbeats. The analysis of a sample of this size and complexity is an appreciable task, especially since practical considerations dictate that the analysis should be performed in less than 30 minutes. Most contemporary Holter tape analysis systems achieve this speed by employing special purpose analog hardware preprocessors for the detection and identification of abnormal heartbeat complexes. While hardware circuits promise higher analysis speed, they have the inherent disadvantage of being inflexible. This lack of flexibility is due, to a great extent, to the inability of the analog circuit to adapt to the many non-pathological variations in the data. In view of the complexity and variability of the ECG signal, this drawback of the hardware preprocessor cannot be ignored when designing an analysis system. A programmed analysis system, on the other hand, can overcome these problems and can potentially provide a higher accuracy in the detection and classification of abnormal cardiac activities. A fully software detection method can be programmed to accommodate and adapt to the inter and intra

patient variations that hardware devices are unable to track effectively. This implies, however, that special detection and classification algorithms must be designed to meet very tight time constraints. The work presented here concerns the implementation of such an analysis system on a standard minicomputer without any special preprocessing hardware. This design philosophy takes advantage of the high accuracy and flexibility that characterize digital systems, and the low cost of a standard minicomputer.

COMPUTER INTERPRETATION OF ELECTROCARDIOGRAMS

Conventional ECG interpretation by an electrocardiographer is an extremely time consuming process. Due to this fact, the use of electrocardiograms in preventive medicine has been very limited. Heart attacks and related heart diseases remain the single biggest killer in Canada, accounting for 34 percent of all deaths each year [Wylie, 1978]. Most of these death could have been prevented if the cardiac disorders had been discovered at an early stage. Medical facilities as they presently exist, however, cannot provide the throughput necessary for mass screening of ECG's. Traditional ECG analysis with human interpreters measuring the minute repetitious detail is a tiring and time consuming task.

The basic objective in automated ECG analysis/diagnosis is to relieve some of the cardiologists' burden. In its present evolutionary stage, it is inherently incapable of replacing

physicians, but it can provide a highly visual report which could convey a summary of cardiac activities of the patient to him at a glance. It is more accurate for this purpose than the human observer, because unlike human beings, it excels in repetitive and lengthy tasks. An automated analysis system also has the potential of reducing costs to a point where arrhythmia analysis can become part of a regular checkup procedure. The Holter system could be especially valuable in this respect as it could be used to detect arrhythmias that are precursors to dangerous cardiac abnormalities. The analysis process is, however, more complicated than that of conventional ECG due to the sheer volume of data.

Contemporary ECG interpretation systems can be categorized into two types: static ECG diagnosis and dynamic ECG analysis. The static ECG has been used to diagnose cardiac diseases based short records; while dynamic ECG analysis includes intensive care unit (ICU) monitoring and ambulatory ECG recording analysis. Short records (5 to 25 seconds) of ECG recorded on paper or magnetic tape, obtained during visits to clinics or hospitals, have been used for diagnosis of cardiac diseases. This type of diagnosis requires detailed examination of the heart beat complexes, timing and morphological differences. In ICU monitoring, the main interest is in the recognition of transient cardiac abnormalities.

SHORT RECORD DIAGNOSIS

Short record ECG analysis attempts to determine the condition of the patient's heart by careful measurement and classification of the shape of individual heart beat complexes. This could be termed as an entirely static measurement that provides an assessment of the condition of the patient's heart at the point in time of the ECG recording. The initial efforts in automated ECG processing were concentrated in the diagnosis of short records. The technique involves digitizing 5 to 25 seconds of ECG on 12 leads under carefully controlled conditions, and performing an off-line analysis process by computer. The first experimental programs were developed in the early sixties [Caceres, 1969; Piberger, 1962]. A detailed discussion on automated ECG diagnosis systems is presented in Appendix A.

ICU MONITORING

Later developments in cardiac analysis used much longer samples of ECG in an attempt to make a dynamic measurement of the heart's condition that would provide an assessment of the stability of the patient's heart. Unstable cardiac activity is denoted in general by the term arrhythmia, but within this general category there are many specific manifestations of instability, some benign, others indicating a dangerous condition potentially leading to a cardiac failure. The fundamental difference between a "static" analysis of the heart's condition and an arrhythmia analysis is the nature of information required.

A single heart beat could potentially provide sufficient information for a static analysis, whereas, to be significant, arrhythmia analysis depends on a scan of many consecutive beats, with detection of changes from beat to beat and the classification of irregular beat shapes.

One way to obtain the amount of data required for arrhythmia analysis is to hospitalize the patient for monitoring. Conventional ECG monitoring with medical personnel watching oscilloscopes was, however, found to be inaccurate and it was observed that proper quantification was almost impossible in the case of patients with frequent arrhythmias [Romhilt, 1973]. The initial successes in short record ECG diagnosis have prompted researchers to develop automated systems for rhythm monitoring in the hospital environment (see Appendix B for further discussions on automated ECG monitoring systems). Such systems have proven to be extremely valuable in intensive and coronary care units as early warning systems to catastrophic cardiac disorders such as cardiac arrest [Hulting, 1976, 1977].

AMBULATORY ECG MONITORING

Hospitalization for rhythm monitoring presents both economic and diagnostic problems. In economic terms, it is expensive for the public health system, and too inconvenient and impractical to be used for routine checkups. The diagnostic problem is that it places the patient in an atypical environment. Since the occurrence of arrhythmia is often highly dependent on stress and

exertion [Fortuin, 1977], monitoring under clinical conditions does not necessarily produce the same conditions which may be causing the cardiac rhythm disorder of the patient. This is especially true if the patient is only marginally ill.

Since arrhythmia is actuated by certain activities causing physical or emotional stress, it is desirable to use samples which are taken under a variety of conditions. One method which has gained increasing acceptance is the exercise electrocardiogram, where the patient is subjected to steadily increasing physical stress [Fortuin, 1977]. Another method, which has been growing in popularity since the late fifties, is the Holter ambulatory ECG recording technique. While the Holter method provides an excellent sample of information for arrhythmia analysis, it has the disadvantage of involving a large volume of data in the analysis process.

The only commercial equipment in widespread use today, the Avionics Electrocardioscanner, is capable of processing the tapes at 60 to 120 times real time, [Holter, 1961]. This equipment, however, requires constant operator supervision and interaction, and is far from ideal in its ability to produce an accurate summary of cardiac activities over the 24-hour period. The analysis procedure relies heavily on visual scanning by the operator and the quality and quantification of the analysis is directly related to the skill of the operator [Hinkle, 1967; Romero, 1976]. A large variety of artifact has been observed in these recordings, some of which are erratic disturbances which

obliterate the ECG signal, while others mimic arrhythmias which could lead to erroneous diagnostics [Karsnow, 1976]. The job of distinguishing these artifacts rests entirely on the operator [Romero, 1976]. The detection parameters of the system are fairly rigid, making tailoring to suit individual needs difficult and often impossible. It is for this reason that interest has been shown in the computer analysis of these tapes.

The most common approach to the problem is to share the analysis task between two or more processing elements. In general, the digitization and beat detection function is performed by a hardware preprocessor, which then passes the partially processed data to a digital computer for ectopic beat and arrhythmia classification. The preprocessor may be digital or analog, but it is usually hardwired.

The design philosophy behind a hybrid system is a division of tasks between processing elements. This division is imposed by the fundamental conflict between the requirement of speed of analysis, and the amount of processing required to achieve accurate classification. By assigning the tasks of beat detection and measurement to a preprocessor, the remaining tasks of classification and report generation can be accommodated without difficulty by a small processor. Such a task division, however, has the inherent disadvantage of leaving a critical part of the pattern recognition process to a hardware preprocessor, and therefore suffers from the same problems of inflexibility as the analog scanner.

A number of experimental and commercial digital/hybrid ambulatory ECG analysis systems, representing a wide range of complexity and capability, are presently available. A commercial system marketed by Cardio-Dynamics Laboratories Inc., California, uses a random sampling process called the "Dyna-Gram Analysis" [Hansmann, 1975]. The frequency of occurrence of arrhythmia is estimated statistically from the information contained in the random samples. One fact that is immediately obvious is that the tape is not completely analysed; only portions of it are sampled. Since arrhythmia does not occur randomly, the practical value of such an analysis technique is not certain. The Stanford University system, [Fitzerald, 1975], utilizes a hardware circuit for detection of the heart beat onset and offset points. The analysis requires 2 passes of the analog tape, where the first pass collects information about the heart beats and the second pass utilizes the data obtained during the first pass to locate areas of abnormalities. The Argus/H system from Washington University, [Nolle, 1974], uses extensive operator editing to assist the computer in the classification of beats. A system based on finite state machine concepts was developed in Columbia University, [Florenz, 1977], however, the system's performance is not clear at this time. The University of Edinburgh system employs a special purpose hybrid computer, [Neilson, 1974], which detects abnormal beats by cross correlation with an average model. The U. S. Air Force system uses a hybrid computer in conjunction with the Avionics composite electrocardioscanner [Walter, 1973]. The Université de Paris system relies on a

special purpose hardware circuit to provide the digital computer with the width, polarity, amplitude and timing information of the heart beat complexes [Coulmel, 1975].

The degree of success varies greatly, but no one system is superior in all aspects. The major problem lies in the inflexibility of the hardware detectors, the very problem that plagues the Holter-Avionics system. TABLE 1.1 summarizes several ambulatory ECG analysis systems, their features and degrees of success. An effort was made to include all the present experimental and commercial systems found in publications. The generally accepted approaches to quality control are even less developed in this area than in real time monitoring, so the results must be judged accordingly.

SYSTEM	COMPUTER	# OF PASSES	MAX. SPEED (X REAL TIME)	EVALUATION
Stanford [Harrison, 1976]	PDP-12 special purpose hardware	2	60	98%
Washington [Nolle, 1974]	IBM SYSTEM 7 special purpose hardware	2	60	89%
Edinburgh [Neilson, 1974]	hybrid	1	60	96%
USAF [Walter, 1973]	hybrid	1	60	---
Universite de Paris [Coulmel, 1975]	special purpose hardware	1	60	---
<u>COMMERCIAL SYSTEMS</u>				
Cardio- Dynamic [Hansmann, 1974]	special purpose random sampling	3	60	---
Avionics [Hinkle, 1967]	hybrid	1	120	---
Medilog [Cashman, 1974]	hybrid	1	25	---

NOTE: The evaluation results quoted are those claimed in the published reports.

TABLE 1.1 : Ambulatory ECG Recording Analysis Systems.

DIFFERENCES BETWEEN AUTOMATED ECG DIAGNOSIS
AND ARRHYTHMIA ANALYSIS

From the point of view of the designer of an automated ECG interpretation system, the difference between ECG diagnosis and arrhythmia analysis is in the nature of the information required. ECG diagnosis requires detailed examination of all waveforms of the ECG signal, morphological differences, and exact timing information. The diagnosis process is usually based on a small amount of data (10 to 30 heart beats). The origin of the disorder (type of disease) is determined from the shape of the heart beat and various other parameters. On the other hand, arrhythmia analysis attempts to obtain an overall view of the condition (stability) of the heart by scanning a large amount of information. It does not require examination of minute details nor correlation of the information to specific cardiac diseases. Due to the difference in the nature of the two interpretation systems, the accuracy of arrhythmia analysis systems can be substantially higher (approaching 100%) than that of a short record diagnosis system.

DIFFICULTIES IN DIGITAL ECG ANALYSIS SYSTEM IMPLEMENTATION

In classical textbook examples, arrhythmias are easy to recognize, as they usually have features that are distinctly different from those of a normal ECG. Experience has shown, however, that while these arrhythmias do occur, they do not always follow the rules found in textbooks. There are three major contributing factors of ECG waveform variability, namely:

intersubject variability,
record-to-record variability, and
beat-to-beat variability.

A bewildering variety of QRS shapes can and do occur in the ECG's of many patients, even in seemingly "normal" ECG's of different patients. The variability of the "normal" QRS complexes, against which abnormality must be detected, together with transient superimposed muscles tremors, baseline drifts and other artifacts, render simple hardware devices like those used in many monitoring system unsuitable for the task of reliable wave detection.

Unlike the human interpreter, the computer requires lengthy and explicit instruction to recognize and classify the various ECG waveforms. The lack of definitive standards between normal and abnormal ECG has further complicated the problem. This is not unique to the area of computer interpretation of ECG's, as even experienced cardiologists themselves are sometimes faced with this problem, especially in marginal cases [Caceres, 1973].

This factor presents the greatest difficulty in automated ECG analysis. Vast gaps in the diagnostic criteria are left open to the interpretation of individual researchers [Glantz, 1978].

LIMITATIONS OF THE COMPUTER IN ECG INTERPRETATION

The basic limitations of ECG programs are the result of the difficulties mentioned above. These are primarily a result of the lack of objectivity in clinical electrocardiographic criteria for both measurement and diagnosis. In addition, the lack of reasonably exact correlation between electrocardiographic waveforms and cardiac diseases, have further complicated the problem. Recent medical research has revealed, for example, that arrhythmia occurs in apparently normal healthy people and is even more pronounced in athletes. This further narrows the distinction between normal and abnormal ECG waveforms. To be fully effective the ECG must be correlated with various other symptoms for an accurate diagnosis.

ECG analysis systems will necessarily be evolving systems. As they provide new knowledge to medical science, the correlation between the ECG and cardiac abnormalities will be better understood. This in turn can be used to improve the diagnostic criteria of automated ECG analysis systems.

GOAL OF THIS THESIS

Since the introduction of the Holter-Avionics system, several studies have been made on its limitations [Hinkle, 1967; Romero, 1976; Reiffel, 1977]. Most of these limitations have been mentioned previously in this chapter. The major drawbacks are the inability to obtain precise quantitative information, and extensive operator interaction required. A number of researchers have attempted to implement ambulatory ECG recording analysis systems based on small digital computers with analog preprocessors. The inflexibility of analog devices for this type of application is, however, well demonstrated in the Holter-Avionics system. An entirely digital system where all the analysis algorithms are implemented in software, can potentially provide far more versatility and flexibility.

In this project an attempt was made to design and implement an entirely digital ambulatory ECG analysis system within the following constraints:

- A) analysis at high speeds (50-100 times real time),
- B) construction with commercially available equipment, no special hardware is to be used,
- C) implementation based on a small digital computer system,
- D) quantitative record of all arrhythmias occurring during the period of the recording with a very low error rate,
- E) presentation of the analysis result in a compact printed report.

() The system described here differs from the existing systems, TABLE 1.1, principally in constraint B: it is the only system reported which does not require some kind of special purpose hardware preprocessing device. This was a significant factor in the development of the system, as it placed much tighter constraints on the time allowed for processing of the signal. In return, however, it has the advantage of being easily adaptable and extendable, both qualities that will be desirable when the results of clinical use are obtained. Furthermore, it is potentially realizable with standard microprocessor system components, whose cost would be significantly lower than existing systems.

CHAPTER TWO

THE ECG SIGNAL

INTRODUCTION

The design problems of an ambulatory ECG recording analysis system are intimately related to the nature of the signal to be analysed. This chapter describes the different characteristics of the ECG signal and how they are related to the physical structure of the heart. The pathological conditions that the system is designed to detect along with their representation in ECG waveforms, are discussed in the following sections.

THE STATIC ELECTROCARDIOGRAM

The electrocardiogram is a recording of the electrical activity of the heart expressed as a function of time. The recording is made by means of electrodes taped to the surface of the body in various locations, including the region of the heart, the back and the limbs. The signals captured by these electrodes are the projections of the electrical polarization and depolarization of the heart muscle as it contracts and relaxes rhythmically [Marriott, 1972]. This activity represents a volume change in polarity which propagates from a single node through the Atria and the Ventricles of the heart. It can be modelled for the purposes of analysis as a three dimensional vector which rotates about a point, varying in amplitude and radial velocity.

An electrode on the surface of the body detects a projection of this vector, developing the corresponding variation in the voltages of the ECG signals captured. The signal amplitude as a function of time is therefore quite dependent on the position of the electrode, the size of the patient, and the position of the patient's heart, among other factors. Other muscle activities also produce signals similar in nature, and the quality of contact between the electrode and the patient's skin is a critical factor. A poorly placed electrode will pick up noise from any source of electromagnetic radiation, in particular, from the electrical system of a building, whose frequency of alternation is in the same range as the frequencies which characterize cardiac activity.

THE DYNAMIC ELECTROCARDIOGRAM

The dynamic ECG (for example, hospital monitoring, ambulatory recording) is a method of observing the beat to beat variations in the ECG waveshape. It is a measurement of the stability of the heart's function over a period of time. Its main concern is variation in the shape and timing of the heart beats in the context of a series of beat rather than the usual waveshape of the heart beats. Since the primary objective is to determine the significant transient variations in the ECG, a single lead (refer to Appendix C for discussion on ECG lead systems) is sufficient for this type of monitoring.

ANATOMY OF THE HEART

In order to understand the anatomical basis of the ECG, it is first necessary to briefly examine the structure of the heart, Figure 2.1.

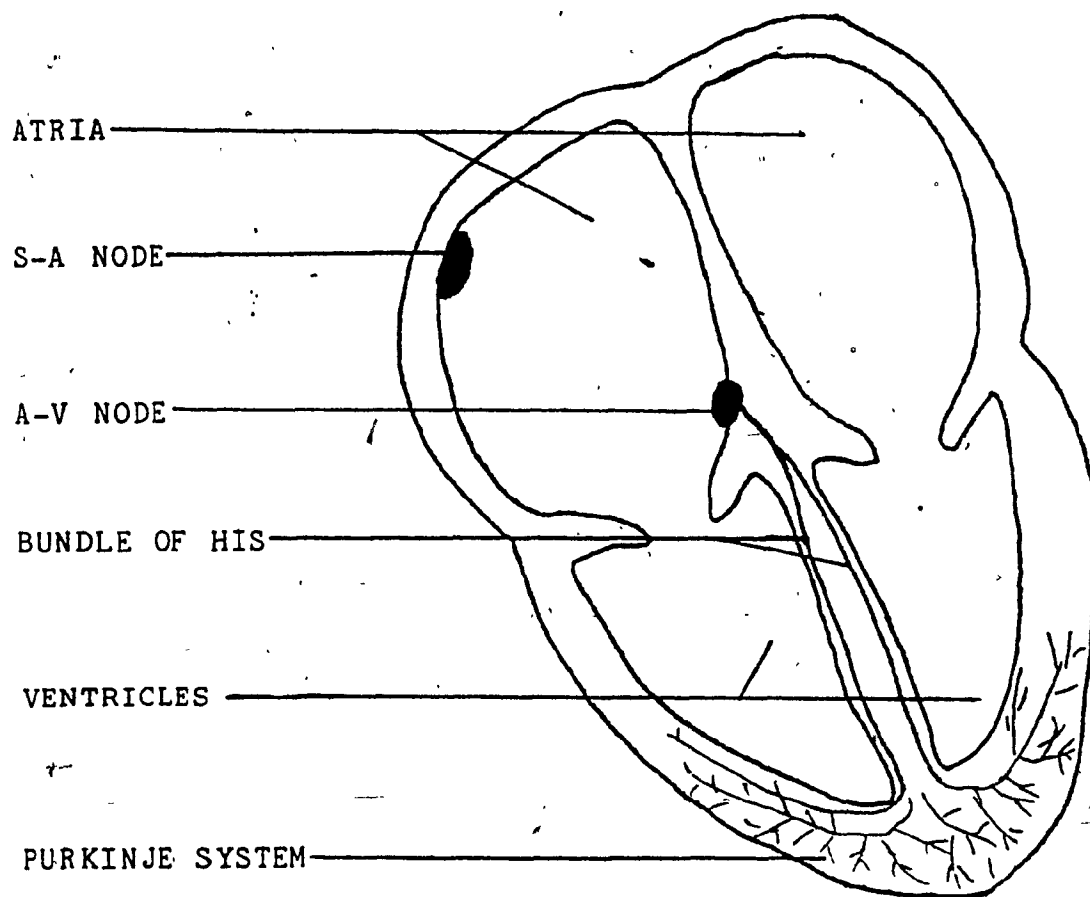


Figure 2.1 : Anatomy of The Heart.

The heart is divided into two primary sections: the upper two Atria, and the lower two Ventricles. A node of specialized tissue located in the upper part of the right Atrium, the Sino-Atrial (S-A) node or "pace-maker" of the heart, generates pulses which regulate the heart rate. It is autonomous, that is, it is capable of the independent rhythmic generation of impulses. The electrical wave from this node spreads through the two Atria, causing them to contract and force blood into the Ventricles. At the lower part of the two Atria there is another node of specialized tissue, the Atrio-Ventricular (A-V) node. It is stimulated by the electrical waves originating from the Atria and relays these impulses to the Ventricles via a nerve network called the Bundle of His. This nerve bundle fans out into a network of fibers called the Purkinje System. The electrical waves are propagated to the Ventricular walls by this system causing their contraction and the pumping of the blood [Bellet, 1972].

() RELATIONSHIP OF THE ECG TO HEART ANATOMY

In each cardiac cycle there are up to five major deflections referred to as the P, Q, R, S, and T waves, Figure 2.2. These waves correspond to the activation and relaxation of the cardiac muscles.

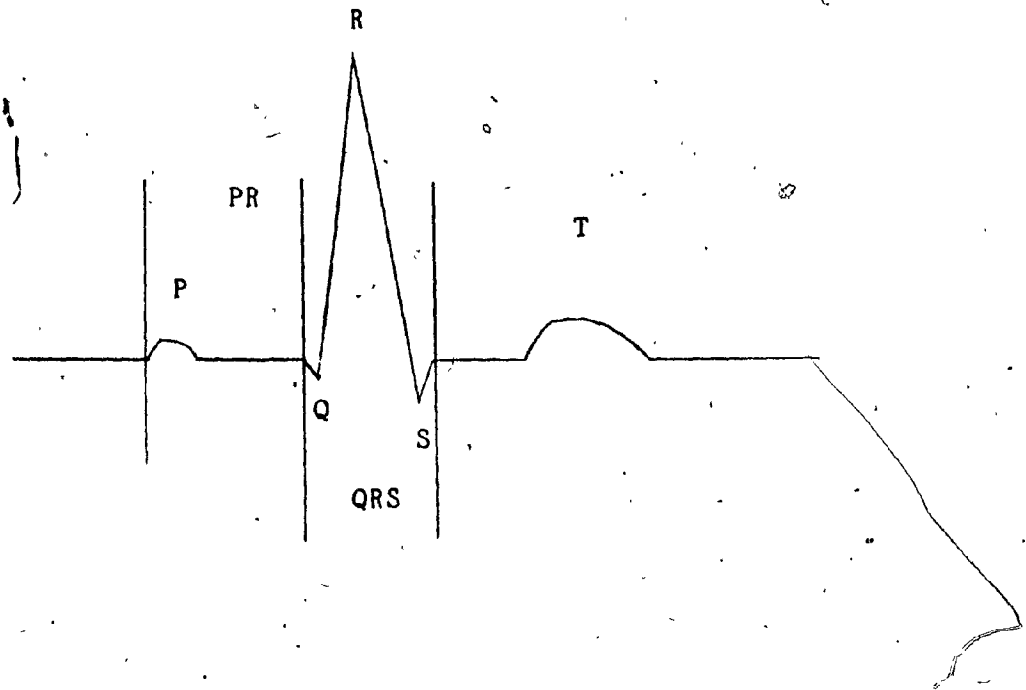


Figure 2.2 : ECG Waveform.

P WAVE - The contraction of the Atria initiated by the S-A node.

PR INTERVAL - The beginning of the P to the beginning of the QRS. This is the time it takes from the beginning of Atrial contraction to the beginning of Ventricular contraction.

QRS COMPLEX - The contraction of the Ventricles, depolarization of both Ventricles.

T WAVE - The relaxation of the Ventricular muscles, Ventricular recovery.

R-R INTERVAL - The time interval between two successive heart beats.

Heart rate (beats per minute) = $60 \text{ seconds} / \text{R-R seconds}$

TABLE 2.1 : ECG Deflection Definitions.

Figure 2.3 relates the various deflections to their origins in the heart.

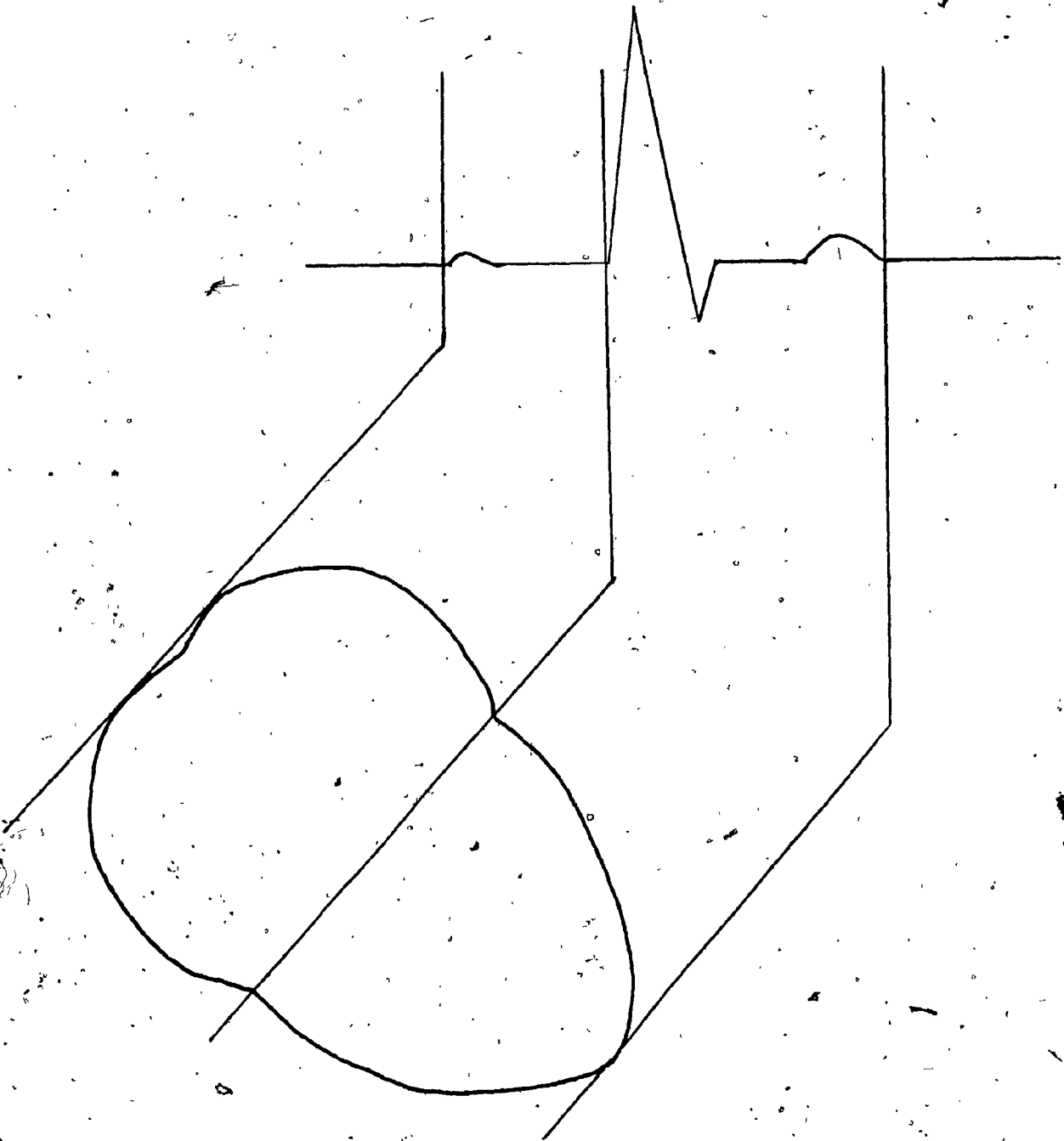


Figure 2.3 : Relations Between ECG Deflections and Heart Anatomy.

CARDIAC ARRHYTHMIA

The normal heart function in a resting adult is characterized by a cycle of electrical activity whose rate is 60 to 100 times a minute. Any transient change in this operation, in the origin of the discharge, in the conduction paths or in the rate of operation, is classified as a cardiac rhythm disorder, or arrhythmia.

Although the term arrhythmia refers in principle to cardiac rhythm disorder, it has acquired a conventional usage as a global term to refer to any non permanent abnormal cardiac activity, even those that do not involve rhythm variation. Another term that has gained conventional usage as a synonym of arrhythmia is ectopic rhythm, although, in its strict definition, it refers only to an abnormal origin of impulses [Bellet, 1972]. In this report the current conventional usage of these terms is adopted, and the two are used interchangeably to refer to any non permanent abnormal cardiac activity.

Disorders of cardiac rhythm are common and are responsible for a significant part of the mortality of cardiac diseases. The medical profession has classified close to 700 types of arrhythmia, ranging in importance from benign fluctuations in the heart rhythm to complete break down of the heart function leading to death in minutes [Shaub, 1965; Bellet, 1972]. It is impractical to program an automatic system for such a large number of variations. The purpose of ambulatory ECG monitoring

() is, however, to observe the marginally ill or apparently healthy patients and detect arrhythmias that are precursors of more serious cardiac abnormalities. This reduces the number of arrhythmias of interest to a relatively small subset of the 700, making it economically feasible for implementation as an automatic clinical system. Several types of arrhythmia which are pertinent in the analysis of ambulatory ECG recordings are described in the following sections.

THE PREMATURE BEAT

Premature beats are the most common of arrhythmias. They are usually the initial evidence of cardiac abnormality. Medical studies have shown that serious, life-threatening cardiac disorders are frequently preceded by premature beats and other less serious arrhythmias. Their frequency of occurrences correlated with the patient's symptoms are of tremendous value in the early diagnosis of cardiac diseases [Bellet, 1972; Marriott, 1972].

Premature beats may occur as isolated beats or in clusters, they may appear infrequently, or they may be recurrent. There are two basic types of premature beats, classified according to the site of the originating impulse: Premature Atrial contraction and Premature Ventricular Contraction.

The Premature Ventricular Contraction (PVC) or Ventricular Extrasystole (VE) as suggested by its name is a beat that arises prematurely before the next expected discharge of the dominant pacemaker of the heart. It may arise from any region of the Ventricles or the conduction system, i.e. the A-V node or His Bundle. The wave shape of the PVC is abnormal. The QRS is widened, frequently slurred in a bizarre fashion, and may have a different direction (Figure 2.4). It is not preceded by a P wave and most of the time it is followed by a full compensatory pause, a cycle long enough to compensate for the prematurity of the extrasystole. The T wave following the PVC is usually large and

() opposite in direction to the deflection of the QRS.



0 Figure 2.4 : Premature Ventricular Contraction Waveshapes.

Another type of Premature Ventricular Contraction is the Ventricular fusion beat Figure 2.5. This is an ectopic impulse arising from the Ventricles which occurs after the S-A node has initiated a beat. The contraction of the Ventricles is triggered partly by the ectopic impulse and partly by the Sinus impulse. The resultant waveform is a Premature Ventricular Contraction and a normal beat fused together. The QRS complex of a Ventricular fusion beat is usually slurred and notched.



Figure 2.5 : Ventricular Fusion Beat.

The Premature Atrial Contraction (PAC) or Supraventricular Extrasystole (SVE) is caused by an impulse whose origin arises prematurely from an ectopic focus in the Atria outside the S-A node. In an Atrial premature beat, the QRS complex is of normal duration (width) and shape, Figure 2.6. It is preceded by a premature ectopic P wave and is followed by less than a full compensatory pause [California Heart Association, 1968; Bellet, 1972].

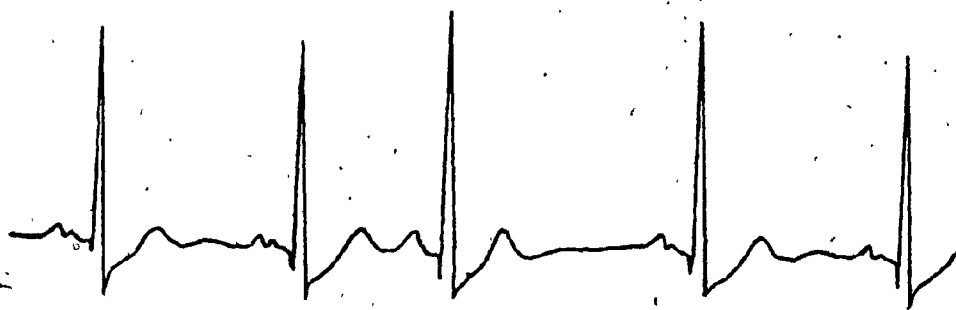


Figure 2.6 : Supraventricular Extrasystole.

There are some exceptions to the Atrial rules: a premature atrial contraction may be followed by a full compensatory pause and there may be aberrant Ventricular conduction so that the QRS is of abnormal shape. The P wave may be buried in some other wave or artifact and cannot be seen, although this is more of a limitation of the present recording technique rather than an exception to the rules. Similarly, there are corresponding exceptions to the wave shape of the Premature Ventricular Contraction. A PVC may not be followed by a full compensatory pause, it may be preceded by a P wave, and the shape of its QRS complex may not be aberrant. In both cases the exceptional features seldom occur together.

() Ventricular Bigeminy is a rhythm where every second beat is a Premature Ventricular Contraction, Figure 2.7. It indicates a more advanced stage of cardiac disorder than isolated Premature Ventricular Contractions. When the ratio of premature beat to normal beat is two to one, Figure 2.8, it is called Trigeminy.

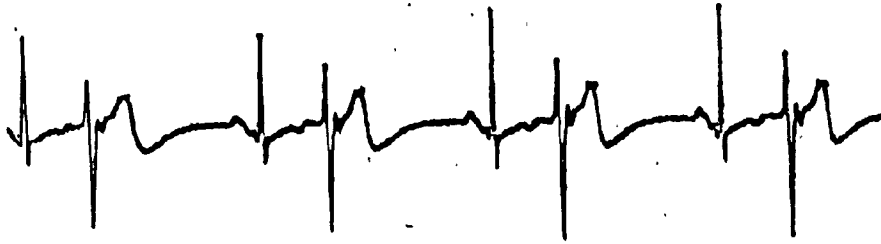
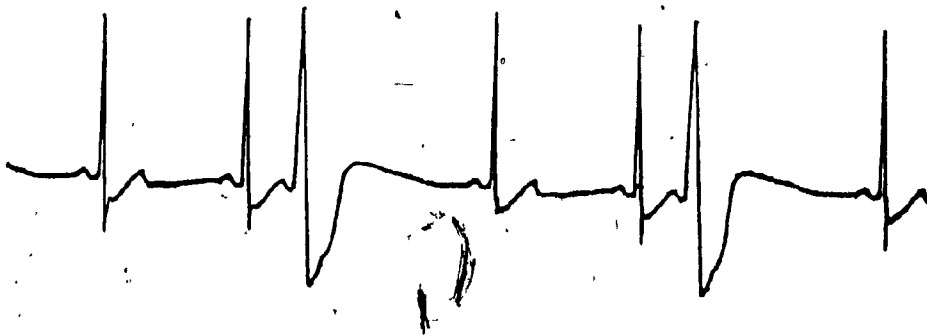


Figure 2.7 : Ventricular Bigeminy.



0 Figure 2.8 : Ventricular Trigeminy.

In general, occasional Premature Atrial Contractions are relatively benign. The condition becomes more serious as the frequency of occurrence increases. Premature Ventricular contractions are usually of a more serious nature, and they are associated with a definitely increased risk of sudden death [Bellet, 1972]. They may forewarn the possible occurrence of Ventricular Tachycardia and Ventricular Fibrillation.

There are no definitive rules as to the significance of the various types of premature beats. Clinical observations have shown that clear differentiation between those occurring in abnormal hearts is often difficult. Premature beats that occur mainly in the presence of cardiac diseases have been observed in hearts that are considered to be normal. The significance of these beats must be determined by a complete clinical evaluation of the patient.

OTHER ARRHYTHMIAS

Ventricular Tachycardia is a type of arrhythmia that indicates serious cardiac abnormality. It is a series of successive Premature Ventricular Contractions, Figure 2.9.

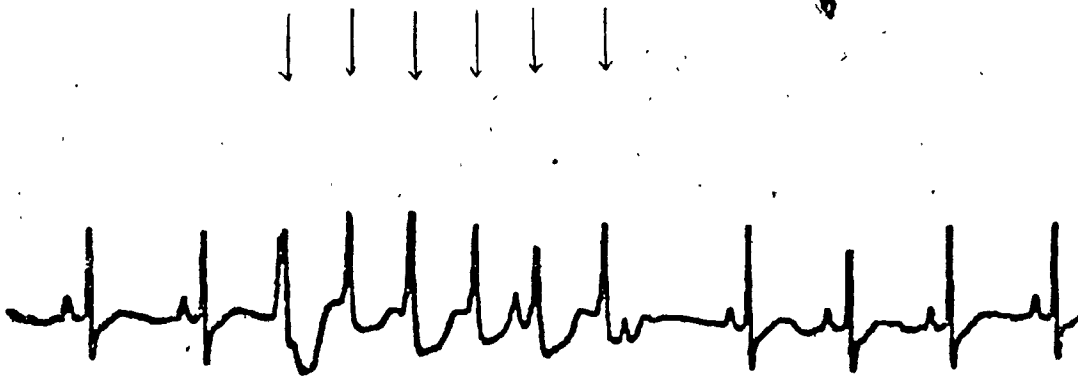


Figure 2.9 : Ventricular Tachycardia.

Sinus Arrhythmia is an irregularity in the rate of the pacemaker which the whole heart beats in response to, Figure 2.10. Sinus Arrhythmia per se is a normal phenomenon; its presence neither establishes nor rules out the presence of cardiac abnormality. The interest in this case is its effect on the calculation of the average heart rate rather than the physiological significance (see chapter 4).

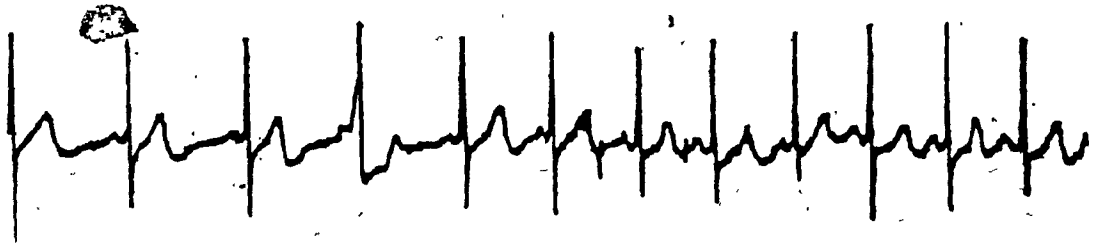


Figure 2.10 : Sinus Arrhythmia.

Ventricular Fibrillation is a completely irregular rhythm caused by random patterns of electrical activation of the Ventricles. The pattern is bizarre with no rhythm of any sort, Figure 2.11. This is a most dangerous rhythm which, if not corrected immediately, will lead to death in minutes. Ambulatory rhythm analysis is of no practical use in this case, as any information derived will be too late for medical attention. It is included in this description because of its clinical importance.



Figure 2.11 : Ventricular Fibrillation.

CHAPTER THREE

SYSTEM DESIGN

INTRODUCTION

The experimental system consists of two subsystems, a development system and a prototype analysis system. The development system was designed as a facility to support off-line testing and refinement of different detection algorithms using a stable data base. The analysis system is a working prototype of a high speed Holter tape analysis system. The development system consists of two modules: a data acquisition system, ACQSYS, which digitizes and stores on a disc file a large segment of a Holter tape recording, and an algorithm testing system, EXSYS, which provides a host for algorithms to be tested on the digitized ECG data file. The prototype analysis system can be subdivided into two sections: an on-line phase and an off-line phase. The on-line phase performs processing functions which extract the ectopic beats from the data, and stores them in an event file in secondary storage. The off-line phase then classifies these beats and generates a report.

The entire experimental system, including the development system and the analysis system, was implemented in software. No special purpose hardware device was employed for the preprocessing of the signal or the detection of the QRS. The partitions of the experimental system are shown in Figure 3.1.

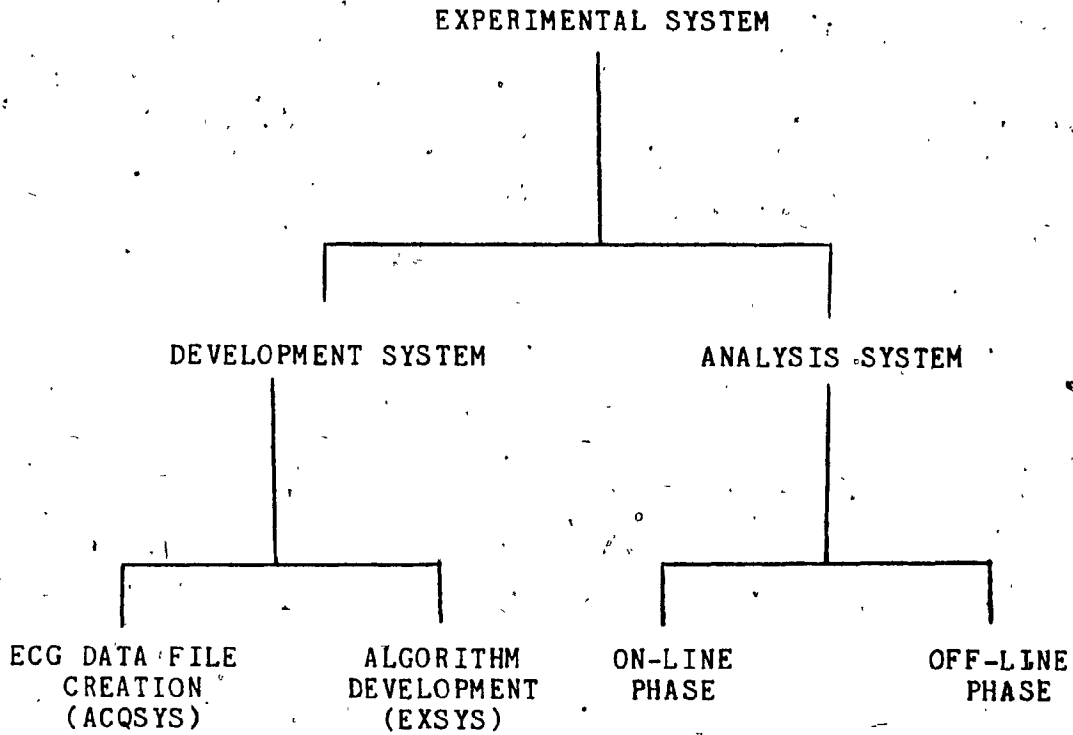


Figure 3.1 : Subdivisions of The Experimental System.

DEVELOPMENT SYSTEM

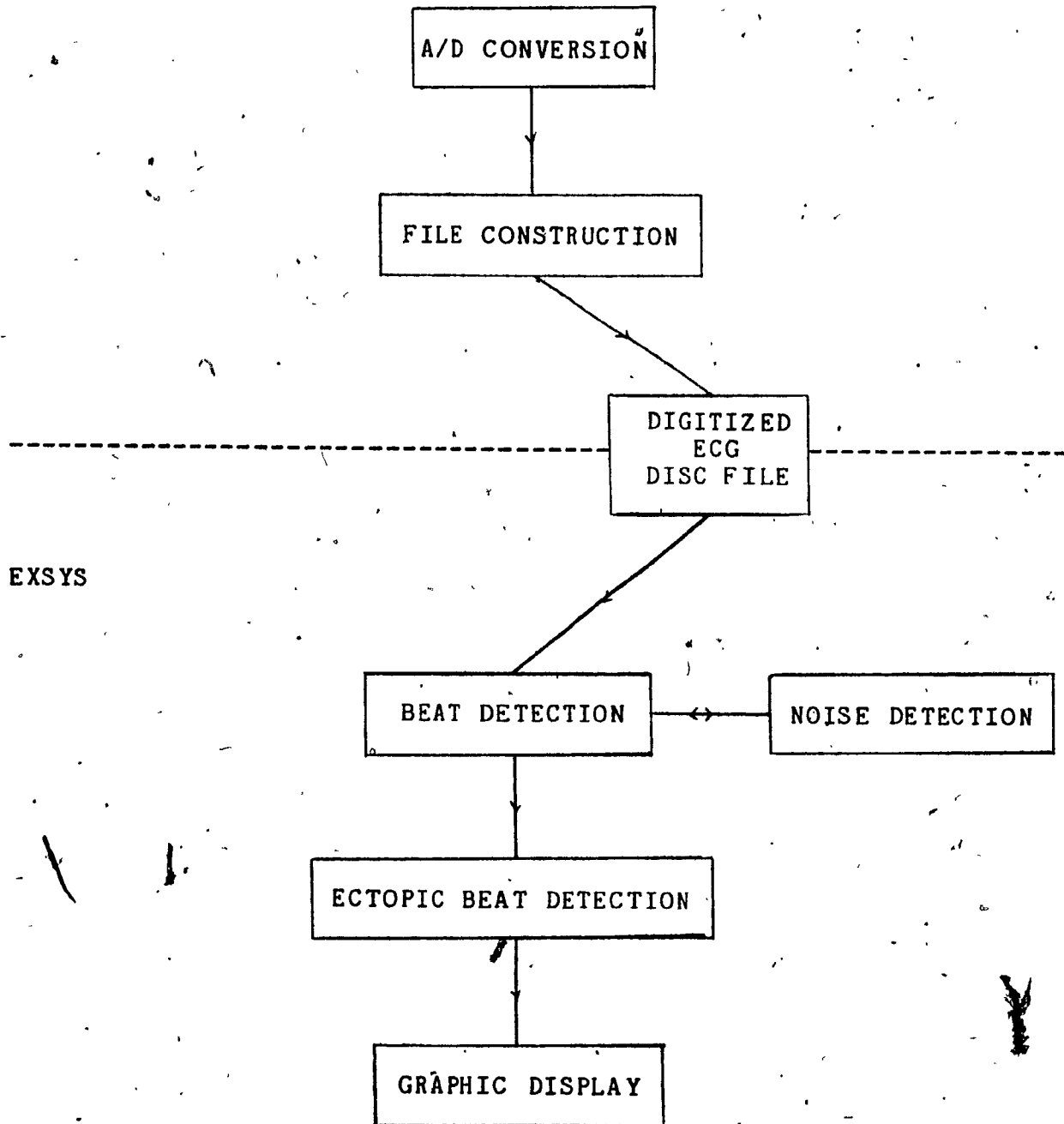
Under ideal conditions, beat detection and classification would not be a complex task. The QRS waves are well defined in the time domain, and have several features that distinguish them clearly from other cardiac activities. Examination of a typical set of ECG traces acquired by Holter recording reveals, however, that the actual detection process will necessarily be quite complex. Under the poorly controlled conditions where ambulatory ECG is recorded, the presence of artifacts and other abnormalities renders the detection of the QRS complexes a tedious process. Factors such as baseline drift, poor electrode attachment and variation in recording quality introduce many additional variables into the detection and classification process. Furthermore, the wide range of normal inter patient variations in the ECG characteristics, the intra patient changes in a 24 hour recording period, and the large variety of pathological conditions, must also be taken into consideration.

In order to develop an effective analysis system, a comprehensive data base containing a variety of the common arrhythmias such as VE, SVE, Bigeminy and Trigeminy is needed to provide a precise comparison of algorithms, as the possible range of variation in waveshapes of both normal and pathological QRS beats is very wide. The use of a small sample in the development of beat detection and classification algorithms, will tend to optimize the system towards the characteristics of the sample QRS's.

For reasons of convenience, it is preferable to have the data in digital, rather than analog form. Redigitization of a tape for every test of a new or modified algorithm is a time consuming task. In addition, the set of beats used in each successive experiment will vary according to the tape positioning at the start of each sample run. There is, however, a significant advantage in reusing the same set of sample points for successive tests. In the final stage of refinement of an algorithm, a deterministic evaluation of the reason for its performance on a given waveshape may provide the adjustment necessary to improve it in marginal cases. A stable data base provides the means for directly evaluating the results of the adjustment, on exactly the same data points which were used as the basis for its development. While the ultimate performance of the system must be independent of sample point positions, this feature provides a "fine tuning" support which can save considerable time and effort in the final phase of algorithm development. Figure 3.2 illustrates the structure of the development system.

Tapes from five different patients were used for the construction of the data base. Thirty minutes were selected from each tape, digitized and stored on disc. An effort was made to include all the common arrhythmias and as many different QRS waveshapes as possible. Figure 3.3 shows a representative subset of the different QRS waveshapes (both for normal and abnormal rhythms) included in the data set.

ACQSYS



EXSYS

Figure 3.2 : Software Organization of the Development System.

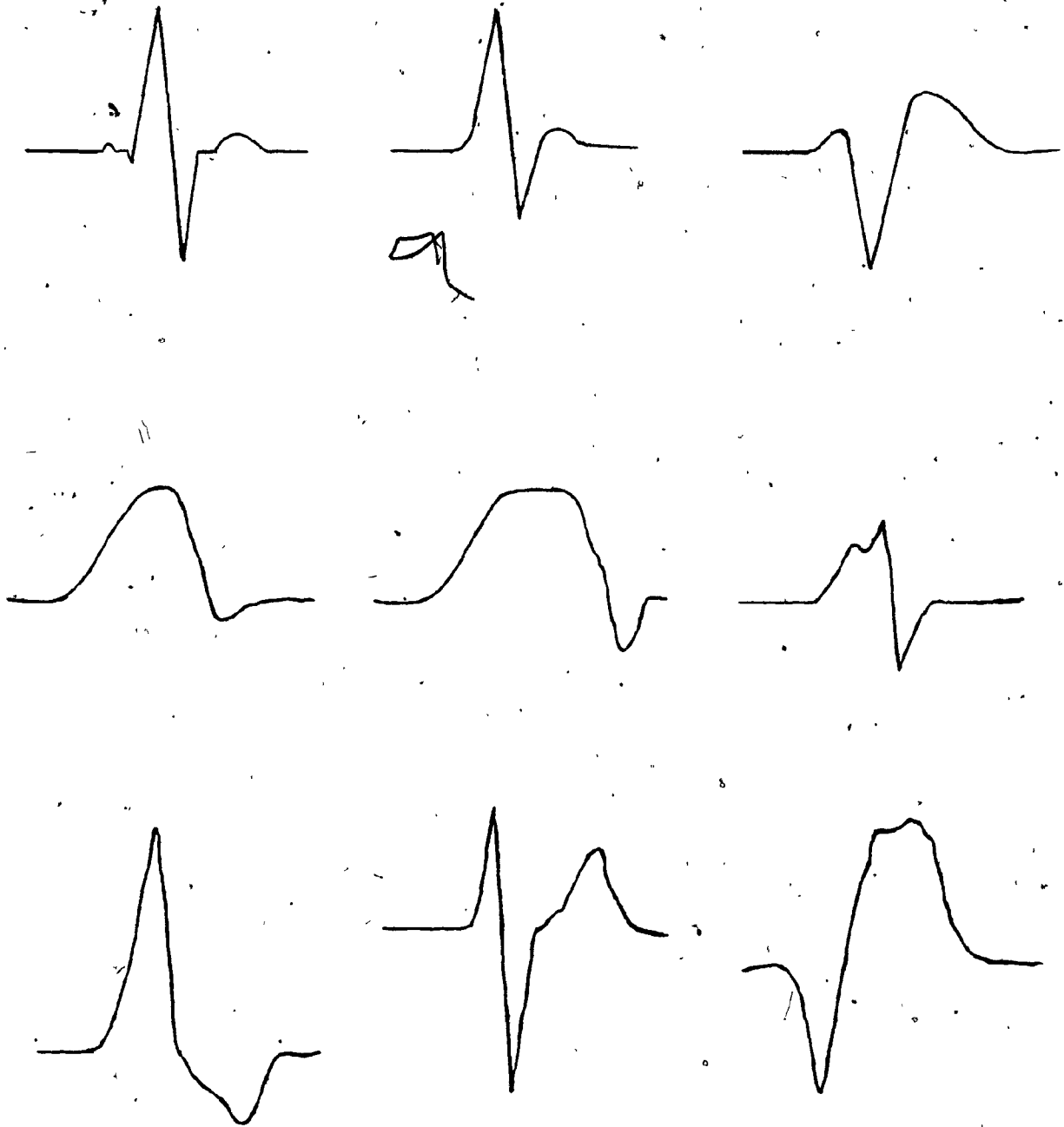


Figure 3.3 : Different QRS Waveshapes (Normal and Abnormal) Included in the Data Set.

ACQSYS - DATA BASE CREATION

A standalone system, ACQSYS, was used to build the disc files for the data base. It accepts data from the A-D converter and creates a binary disc file together with identity information. The flow-chart of ACQSYS is shown in Figure 3.4.. The files created by ACQSYS are compatible with the file structure of TI980B FORTRAN. The capacity of the disc is approximately 4.5 megabytes, i.e. 2.25 megawords (16-bit words), so that a maximum of about 6 hours of digitized ECG can be stored on one disc pack. A standalone system was used because the operating system is too slow for the sampling rate used.

The acquisition rate is 6000 samples per second on the analog recording which is being played back at 60 times real time. The real time sampling frequency is therefore 100 Hz, corresponding to a Nyquist frequency of 50 Hz. This upper frequency limit is more than sufficient for the analysis of Holter recordings [Coulmel, 1975; Bussman, 1975].

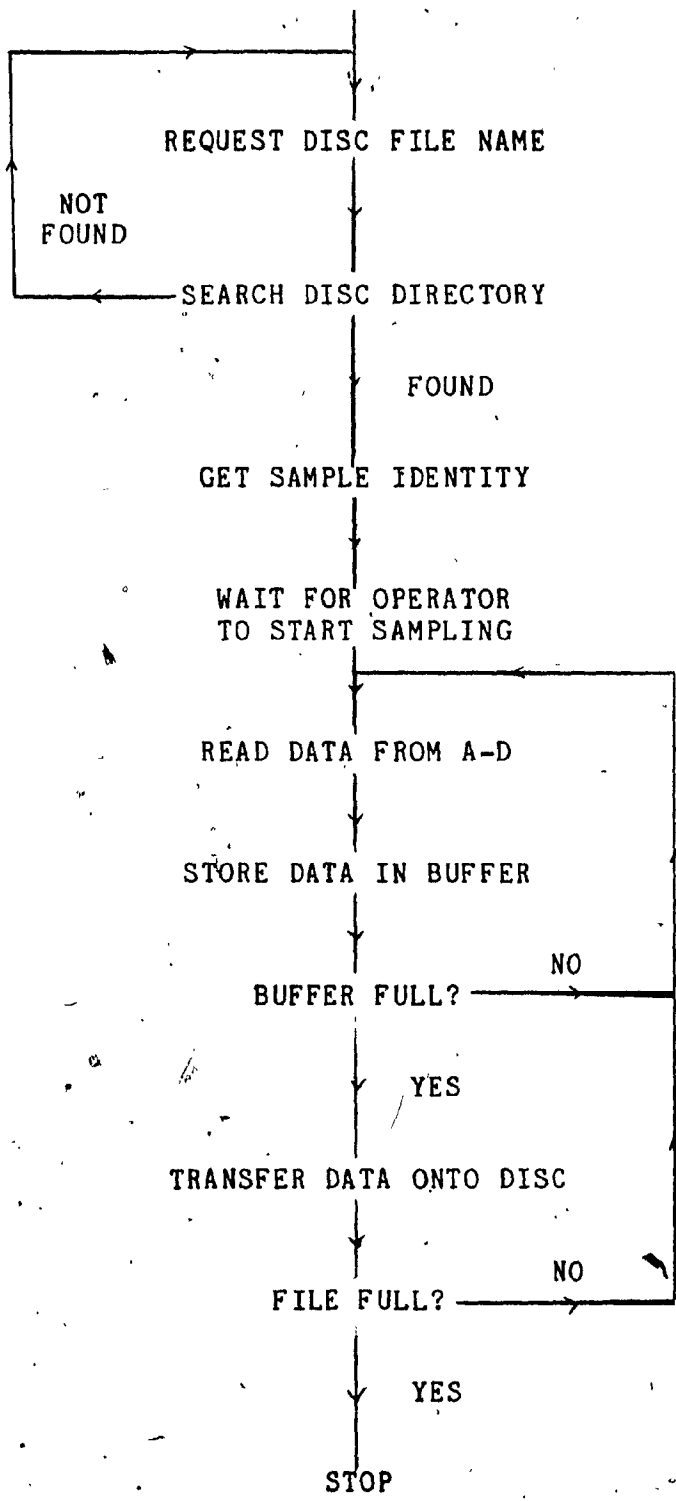


FIGURE 3.4 : Flowchart of ACQSYS.

EXSYS - ALGORITHM TESTING

Medical literature in the past 15 years has been unable to communicate precise measurement data to characterize arrhythmia. This is primarily due to the large variability of the ECG signals and the limited understanding of its origin and effects. Since there is no definitive standard for ECG interpretation the evaluation of algorithms must be done either by cardiologists or according to their criteria. To a large extent, the evaluation process is performed by visual inspection of the data. Although some measurement criteria exist to categorize ECG they depend on the existence of accurately determined fiducial points on which to base the decisions. The accuracy of these points cannot, however, be guaranteed in less than ideal conditions. The ultimate test for a system is therefore always visual inspection by a medical expert. For this reason, a highly visual (graphic) system is needed to facilitate the development, testing and verification of such an analysis system.

The development system, EXSYS, with very versatile graphic capabilities was designed for the selection, testing and refinement of various algorithms. The purpose of this system is twofold: to facilitate development and precise comparison of algorithms and, as will be described later, for the verification of analysis results. The modular design of the system allows algorithms to be inserted, replaced, or modified with minimum of programming effort. Since the development system works with off-line data, processing speed was not the main concern and the

majority of the program could be written in FORTRAN, with the exception of the graphic screen driver which was written in TI980B assembler language. EXSYS performs its analysis using the digitized data base acquired by ACQSYS.

The output of EXSYS is fed directly to the graphics display terminal for visual assessment of the behaviour of the algorithms. The ECG data can be scanned at a maximum speed of approximately 4 times real time or in a "single step" mode under the operator's control. Figure 3.5 shows a simplified flowchart of EXSYS.

OPERATION OF EXSYS

The operation of EXSYS is initialized through a routine which requests operator approval of the initial parameters of slope and R-R interval. This is done by displaying approximately 2.5 seconds of digitized ECG data on the graphic screen, and requesting the operator to signify if the data displayed is satisfactory for initialization by typing a "yes" or "no" on the CRT terminal. The initialization routine selects 3 consecutive QRS complexes, calculates the average slope and R-R interval, and produces the display. The operator is simply requested to recognize whether the beats are normal or not. If he signifies disapproval with an "no", the operation is repeated. A "yes" causes the parameters to be initialized in the detection algorithms, and passes control to the next routine.

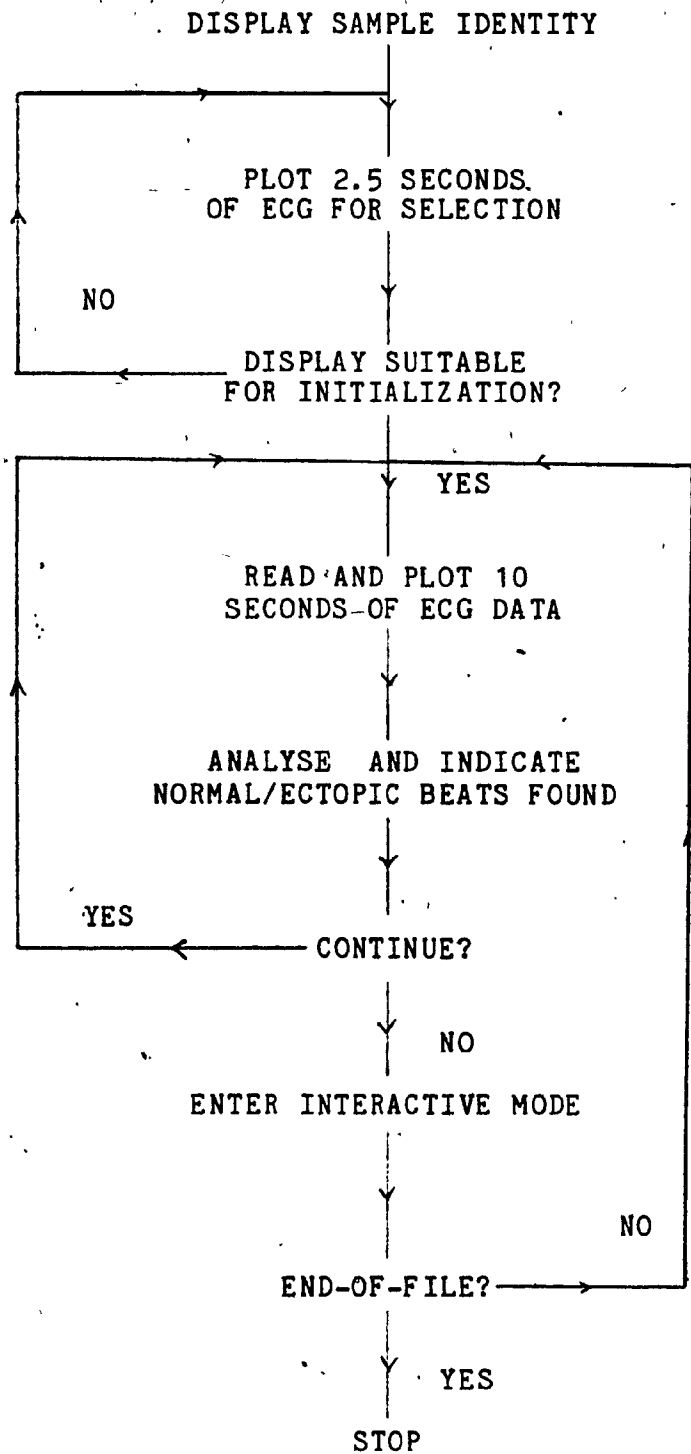


Figure 3.5 : Flowchart of EXSYS.

After initialization, approximately 10 seconds of ECG data is displayed on the graphic screen and the detection points are graphically indicated by two rows of markers, Figure 3.6.

>
HEART RATE = 75 TIME = 0 HRS 1 MIN SLOPE = 685

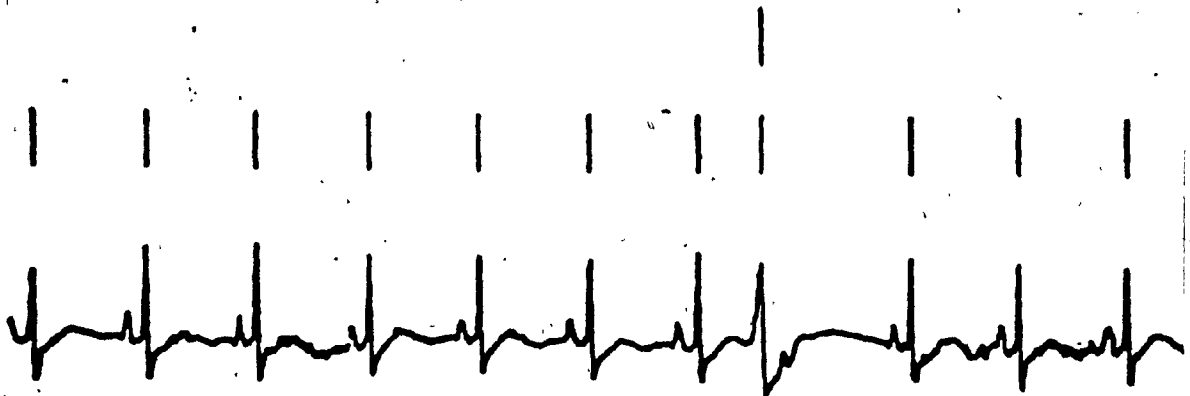


Figure 3.6 : Graphic Display of Digitized ECG Showing Beat Detection (Bottom Row of Markers) and Ectopic Beat Identification (Top Row of Markers).

The lower markers indicate the detection of a beat and the upper one signify that the beat has been identified as premature. At the end of processing the current buffer (the displayed data), the program will either display and process the next section (if the end of the file has not been reached), or pause and enter an interactive mode, depending on the position of a sense switch

() on the front panel of the computer. In the interactive mode the operator can select one of the following options:

- a) magnify any portion of the display (the ECG), Figure 3.7,
- b) dump the numerical value of any portion of the display,
- c) skip ahead a specified amount of time,
- d) terminate the analysis,
- e) advance to the next section.

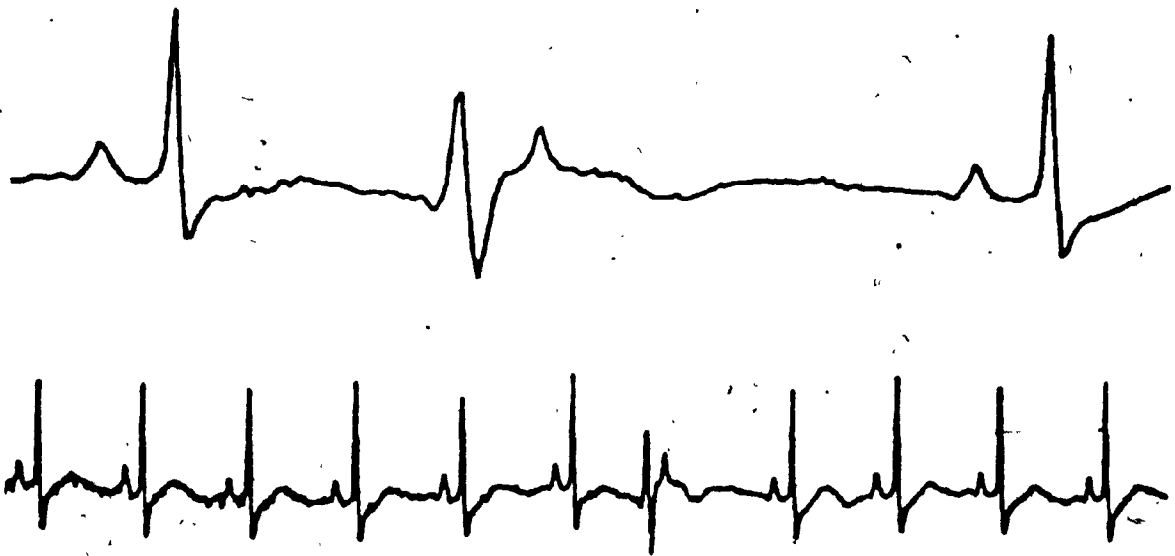


Figure 3.7 : Visual Magnification : The Original Trace is Displayed With the Enlargement of the Segment Selected for Magnification.

THE HARDWARE SYSTEM

The ambulatory ECG analysis system described was implemented on a TI980B mini-computer manufactured by Texas Instruments Incorporated, Houston, Texas, with a 9 megabyte disc unit, (2 drives, 4.5 megabytes each), a console, a printer and a Tektronix 4010 graphic display unit. The ECG data is acquired through a 12-bit analog to digital converter operating at 6000 samples per second. For further discussion on the hardware configuration and software support refer to Appendix D.

THE ANALYSIS SYSTEMOVERVIEW

The system described here was designed to analyse 24-hour ECG recordings at 60 times real time, and produce a report containing quantitative information about the occurrences of arrhythmia events pertinent to the patient's condition. More precisely, exact counts of Premature Ventricular Contractions, Premature Atrial Contractions and related rhythms (Bigeminy, Trigeminy, Ventricular Tachycardia and Sinus Arrhythmia), are presented with a low percentage of false positive and false negative.

Since a large variety of noise has been observed in these recordings, a test is incorporated into the beat detection routine to detect and report the occurrences of significant noise disturbances. Multiple scanning of the analog ECG tape recording is not necessary. The complete analysis requires only a single pass over the original data.

SYSTEM DESIGN CONSIDERATIONS

The implementation of a high speed arrhythmia analysis system is complicated by a number of fundamental constraints. It is not possible to effectively detect arrhythmia by examining individual heart beats as they must be examined in the context of surrounding beats. The ECG signal is recorded on analog medium and must be digitized before it can be processed by the digital computer. Since the analog tape is a continuous and contiguous source of data, it cannot be started and stopped without the loss of synchronization and, therefore, loss of some information. This implies that the entire analog tape must be processed in one continuous run. Since asynchronous processing of the analog tape cannot be used, a very tight time constraint is imposed on the implementation of a high speed analysis system on a small digital computer. There are three ways to solve this problem:

- A) use a hardware preprocessor to detect beats and measure various parameters,
- B) digitize the tape and store the data on an asynchronous digital medium such as a disc,
- C) develop an analysis system which can keep up with the data rate.

The first solution will not be considered for reasons already discussed in chapter 1. The second alternative requires a digital storage device with a capacity of approximately 18 mega bytes ($100 \text{ samples/second} * 60 \text{ seconds/minute} * 60 \text{ minutes/hour} * 24 \text{ hours} * 2 \text{ bytes/sample}$). Since the system was designed to be implemented on a small computer system, this solution is

impractical and will also not be considered. The third solution was therefore chosen as the design goal of the analysis system.

The processing speed is of prime importance when a large number of long electrocardiographic records have to be analysed. Each of the Holter tape recordings contains 24 hours of information and unless it can be processed at high speeds (50 - 100 times real time) the analysis will be intolerably slow. The fundamental design problem is to allow adequate time for processing, while emphasis must also be placed on the overall system efficiency. The processing speed of 60 times real time was chosen for a number of reasons: 1) it is compatible with available equipment since the ECG recordings are made at a speed of 7.5 inches per minute (when played back at 60 times real time the speed becomes 7.5 inches per second, which is a standard playback speed on most audio tape recorders); 2) it was estimated to be within the performance range of a small digital computer; and 3) it allows a 24 hour tape to be processed in a reasonable period of time.

The analysis process can be divided into two major steps. The first is to locate the ectopic beats and the second, to classify them (into VE's or SVE's). These functions can logically be separated into two distinct sequential processes, i.e. the detection process need only to provide the classification process with sufficient information for classification, while no feedback is required from the second process to the first. The complexity of ectopic beat

classification is of an order of magnitude greater than that of ectopic beat detection. Since no assumptions can be made about the frequency of occurrence of ectopic beats, performing the classification at the same time as detection would require allotment of the necessary processing time for every beat. This would result in a drastic reduction of the overall system speed.

In comparison with the total number of beats in a ECG recording the percentage of ectopic beats is usually very low (less than 5% in most cases). This implies that more than 95% of the time allotted for parameter computations would be wasted. A much more efficient approach is to separate the analysis into two phases. The first phase collects information about the occurrences of ectopic beats and records them, along with any information that may be required for subsequent analysis, on secondary storage. The second phase classifies these beats based on the information collected. In most other ambulatory ECG analysis systems, the equivalent of the first phase analysis is performed by a hardware preprocessor which selects beats for further analysis by a software program [Fitzerald, 1975]. The system described here was implemented entirely in software.

ADVANTAGES OF A TWO PHASE SYSTEM

Figure 3.8 shows a diagram relating the different stages of the analysis system.

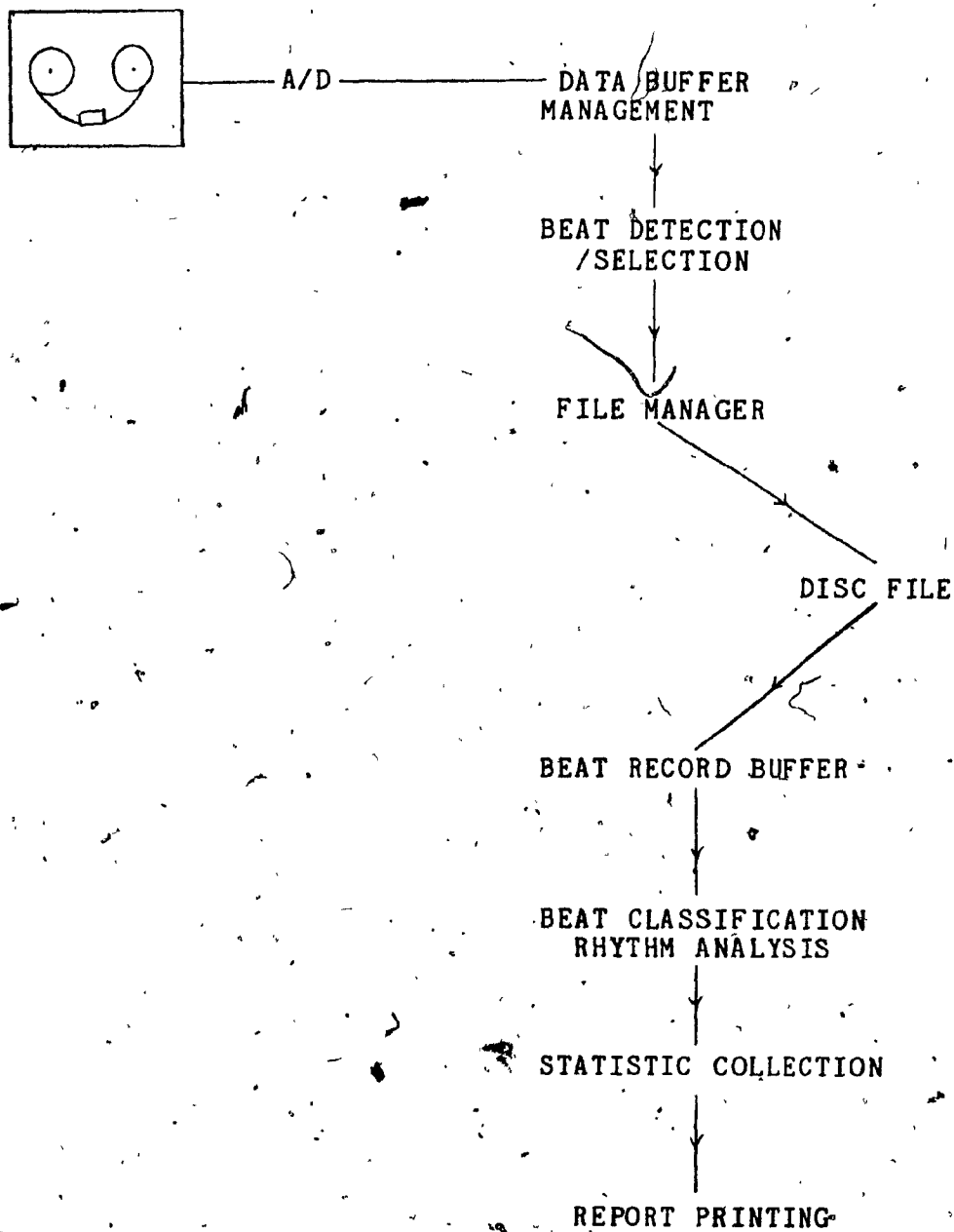


Figure 3.8 : Software Organization of the Analysis System..

During the first phase the tape is scanned at high speed and ectopic events are selected and stored on a disc file. This phase selects beats within the context of the surrounding beats, and performs on-line beat detection and ectopic beat identification. The second phase program performs detailed analysis, ectopic beat classification and rhythm analysis, on the beats selected by the phase 1.

There were a number of advantages in separating the analysis into two phases. The total processing time was greatly reduced because very little computation time is wasted. Each phase was developed and optimized independently. The second phase was also designed to support the development of classification and rhythm detection algorithms, false detections caused by failures of the detection algorithm. Since the ectopic beats detected are stored on a disc file, the performance of the beat classification algorithms can be examined under static conditions.

PHASE 1 - ON-LINE BEAT ANALYSISDESIGN CONSIDERATIONS

This phase of the analysis is basically a data reduction process, where individual heart beats are examined in the context of surrounding beats. The tape is scanned at 60 times real time and ectopic beats are identified and stored on a disc file for further analysis in the second phase.

The critical factor in the design of this phase of the analysis is processing time. At 6000 samples per second (100 samples per second at 60 times real time), the system must be able to process each data point in under 166 msec. This time period is equivalent to approximately 65 typical instructions. In addition to the process of acquiring the digitized sample point, the task of managing the system buffers, beat detection, beat classification, and secondary storage management must also be performed. This time constraint excluded the use of any standard software packages. A special highly efficient system was designed in order to attain the required analysis speed. The operation of the PHASE 1 program is described by the flowchart in Figure 3.9.

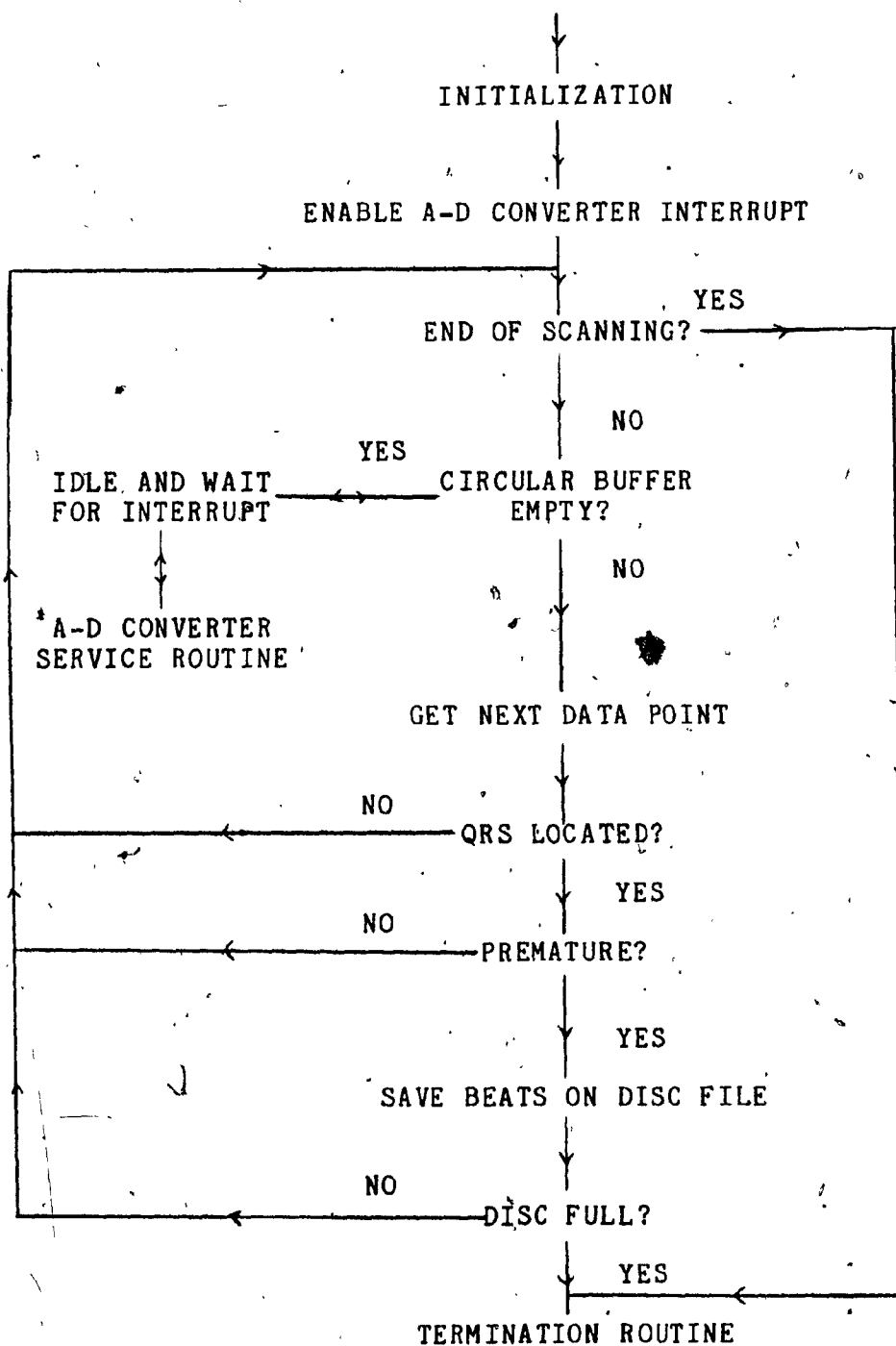


Figure 3.9 : Flowchart of the PHASE 1 On-line Analysis System.

DATA ACQUISITION AND SYNCHRONIZATION

The digitized ECG data is read in via an interrupt driven routine. Upon acknowledgement of the interrupt, control is passed to an A-D converter service routine which reads and stores the data in a 250-word long circular buffer, Figure 3.10. This buffer is shared by the beat detection routine and the two routines are synchronized using a common counter. Each time the beat detection process requires a data point the counter is decremented and when a data point is to be stored in the buffer by the A-D converter service routine the counter is incremented once. If at any time the beat detection routine requires a data point and the counter is equal to zero, then it will idle and wait for the next interrupt (the next data point from the A-D converter).

During the development phase of the analysis system, this counter is also used to detect buffer overflow, i.e. the data points cannot be processed at or above the speed they are acquired. If buffer overflow is detected, the analysis is terminated and an abort message is printed on the CRT. Such a condition would occur only when the system fails to keep up with the data rate and should not happen during normal operations. Its principal purpose is to act as a check during the development period to indicate whether the phase 1 program runs efficiently enough to process the signal at 60 times real time. This test was included in the system as a "fail-safe" feature for algorithm testing and will not be needed in the final system.

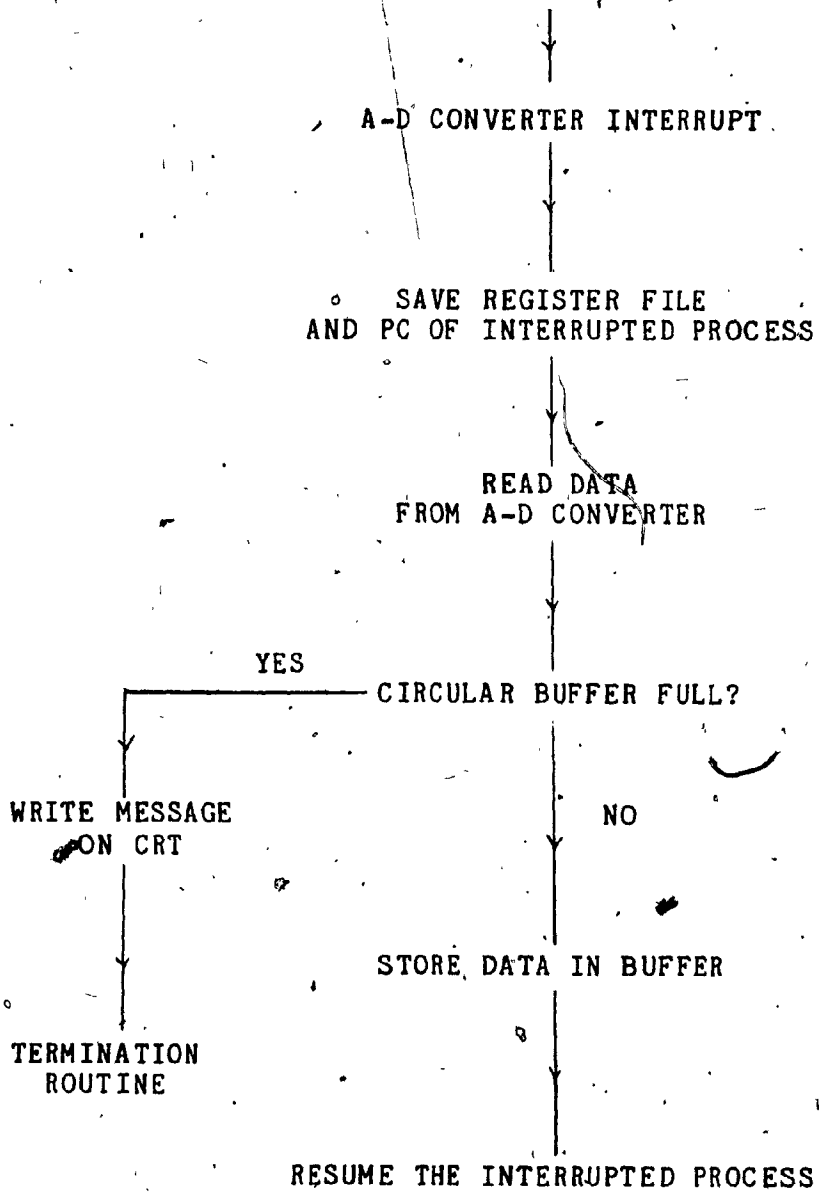


Figure 3.10 : Flowchart of the Data Acquisition and Buffer Management Routine.

FILE CONSTRUCTION

An ectopic event is recorded as a record of 250 data points centered about the identified event. Except in the case of extremely low heart rate, the beat preceding and following the premature beat will be stored. A complete event record consists of the circular buffer, pointer indicating the start of buffer, current R-R interval, and other timing information, associated with the ectopic beat.

When noise is detected, the entry is similar to the premature beat record, with the exception of the heart rate, which is set to a negative value. This will be interpreted by the phase 2 program as an indicator that the record contains noise. The last premature beat record contains a zero heart rate. The only valid information in this record is the time (total elapsed time) and a floating point number indicating the total number of beats detected.

These records are stored in memory in a pair of interface buffers, each of which is 2816 words long (11 records, the number of records in each buffer, was chosen as a trade off between memory space and the frequency of transfer to disc). After a buffer has been completely filled, it is transferred to a disc file via a direct memory access (DMA) channel. While one buffer is being copied to disc, the premature records are stored in the other. The status of the disc is checked before each transfer and at no time does the analysis halt and wait for the disc to

become ready. Direct memory access is used so that as little processor time as possible is consumed in transferring the data. By using twin buffers, the creation of premature records and building of the file can be performed simultaneously.

ECTOPIC BEAT FILE

The general organization of the file is shown in Figure 3.11.

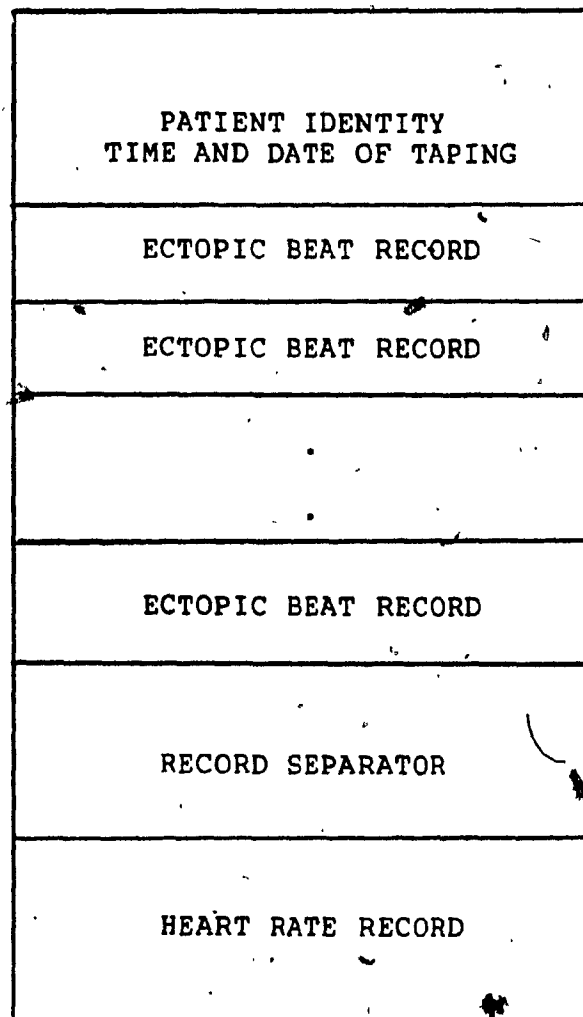


Figure 3.11 : Event File Organization.

The data on the disc is organized in sectors of 32 words. The identity and time information is stored on the first sector of the file. Each ectopic beat record is made up of 8 sectors (256 words). The format is shown in Figure 3.12.

WORD 1	TIME (MINUTES)
2	TIME (MILLI-SECONDS)
3	HEART RATE
4	POSITION OF THE PREMATURE BEAT IN THE BUFFER
5	HEAD POSITION OF THE CIRCULAR BUFFER
6	PREMATURE RATE
7	CIRCULAR BUFFER

256 CIRCULAR BUFFER

Figure 3.12 : Ectopic Beat Record Organization.

HEART RATE RECORD

The average heart rate is sampled every two minutes and stored in a buffer in memory. It is expressed in terms of R-R intervals in 10's of milli-seconds. When the analysis is terminated this record is transferred onto the disc file after the last ectopic beat record.

INITIALIZATION OF THE ANALYSIS SYSTEM

The purpose of initialization is to set the adaptive parameters of tape analysis to initial values appropriate for the tape under analysis. This is done by measuring the R-S slopes and R-R intervals, of a set of three acceptable reference beats. The operator intervention in this phase of analysis is to confirm that the three beats represent a characteristic reference sample for the individual in question.

The initialization procedure is as follows. The operator enters through the console key board the identification data of the patient, including name and date of analysis, Figure 3.13. He then starts the tape, and types the "CR" key to command the system to start sampling the tape. He allows the tape to run for a few seconds and then stops it. The initialization routine digitizes the input signal from the tape and stores it in a temporary file. It then issues the display of Figure 3.14, and requests approval. If the sample is good the operator types Y, and the routine takes the appropriate measurements and sets up initial values for the analysis. If the sample contains an ectopic beat the operator types an N, and the system displays the next three beats from the temporary file. This can be repeated until an acceptable set of beats is displayed. Once the initialization is complete, the operator restarts the tape at the beginning, and the analysis proceeds for twenty four minutes (for a 24 hour tape).

ECG ANALYSIS SYSTEM
ENTER DISK UNIT NUMBER, 0, OR 1 ? 0
? SCR8
ENTER TAPE DATE (YY/MM/DD)
? 78/02/12
ENTER TAPE START TIME (HH/MM/SS)
? 15/35/00
ENTER SAMPLE IDENTIFICATION
? JOHN^DOE

START TAPE RECORDER AND HIT CR
?

Figure 3.13 :: PHASE 1 Initialization : Patient Information.

ARE THESE POINTS SATISFACTORY FOR INITIALIZATION
>Y

HEART RATE = 72 SLOPE = 616
> ARE THESE PEAKS SATISFACTORY ?

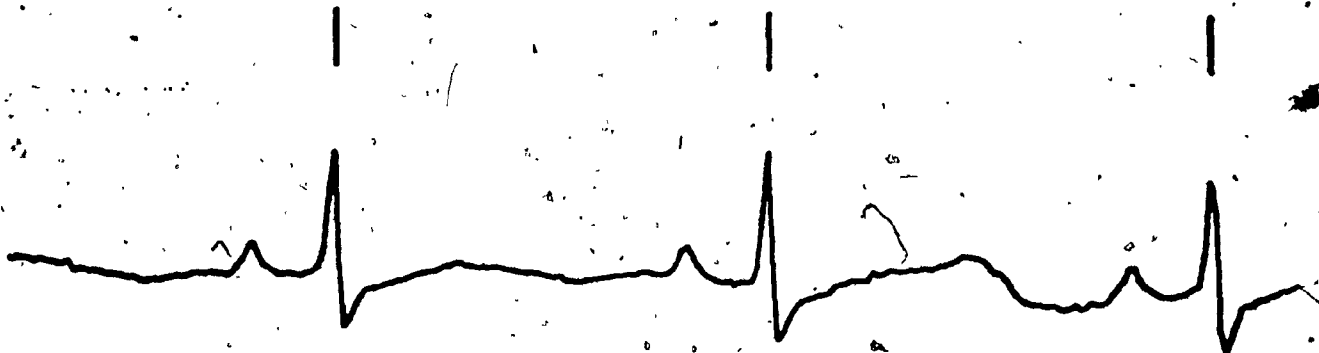


Figure 3.14 : PHASE 1 Initialization : Three Beats Displayed for Operator Approval.

TERMINATION OF THE ANALYSIS

The system can be terminated by one of the following conditions: a) the total time count (tape) has reached 24 hours, b) the disc file is full, c) by the operator via a sense switch on the front panel of the computer, d) hardware error (for example, disc write error). When the analysis is terminated (normally or abnormally) a procedure is executed to close the file, so that all information collected up to the point of the termination will be saved even in the abort case. With the present disc capacity, approximately 8900 records can be stored.

The entire phase 1 program requires approximately 9K words of memory, 5.5K words of which are used as buffers for the disc file of abnormal beats.

PHASE 2 - CLASSIFICATION, RHYTHM ANALYSIS AND REPORT GENERATIONDESIGN CONSIDERATIONS

This phase of the analysis consists of parameter measurement of the ectopic beats; ectopic beat classification, rhythm detection and report generation using the event records in the file constructed by the phase 1 program. The program can be divided into three major sections: a) beat analysis and classification, b) rhythm detection, and c) organization of information and report generation. Figure 3.15 shows the flow-chart of the phase 2 program.

The first section can be subdivided into three parts: a) locating the beats in the record, b) parameter measurements (QRS width, QRS amplitude, baseline and compensatory pause), and c) VE/SVE classification. The second section performs rhythm detection utilizing the timing information on the event record and the ectopic beat classification result to detect various rhythms (Bigeminy, Trigeminy, Sinus Arrhythmia and Ventricular Tachycardia). The third section computes the average heart rate from the heart rate record, organizes the ectopic beat information collected and prints a report.

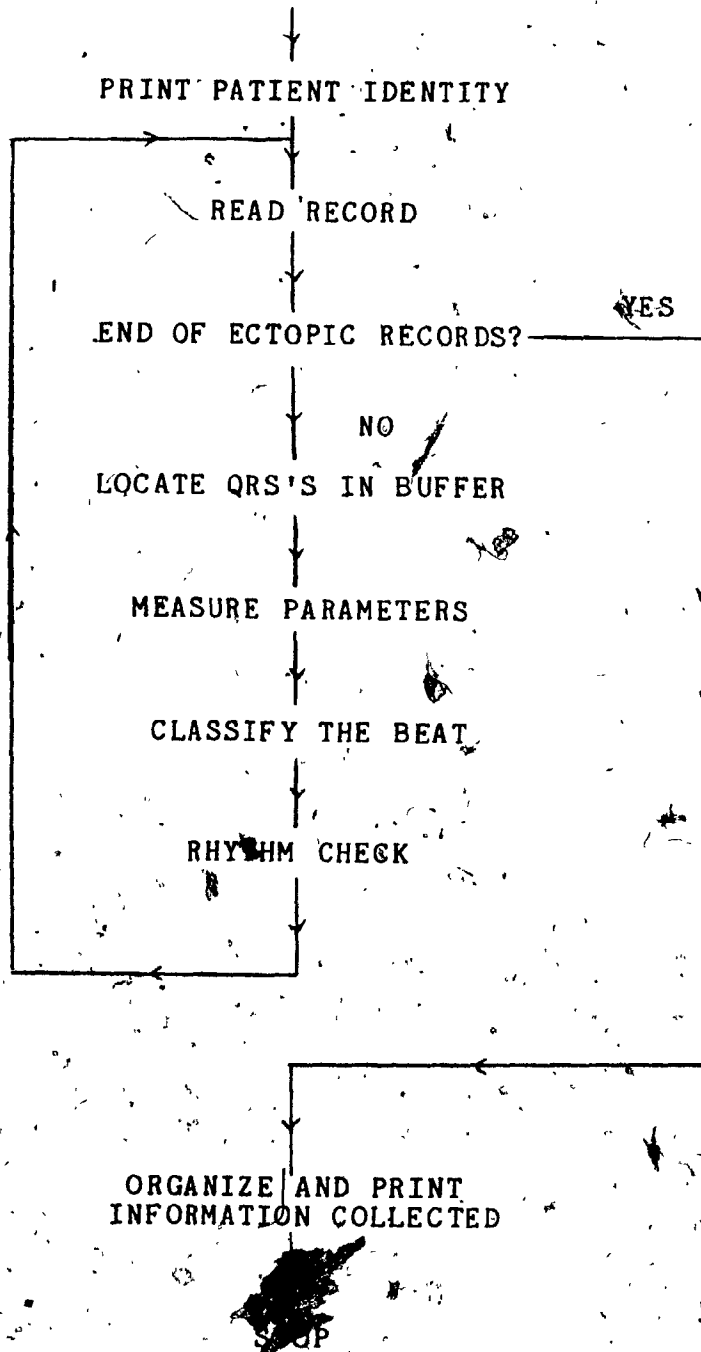


Figure 3.15 : Software Organization of the PHASE 2 Off-line Analysis System.

STATISTIC ORGANIZATION AND REPORT GENERATION

When the analysis is complete, the system produces the set of reports illustrated in the APPENDIX E. The reports were produced on a line printer, and were therefore formatted to accommodate the limitations of that device. Although the analysis is given by half hour segment, it could be easily modified to give greater resolution on a hard copy peripheral capable of providing a more refined output format.

The full report consists of an event summary followed by a set of histograms illustrating the specific results of the analysis. In the event summary, each event is listed in order of occurrence. The Ventricular Ectopic and Supraventricular Ectopic beat counts are listed by half hour summaries. The events listed, apart from the ectopic beat counts, include all occurrences of Bigeminy, Trigeminy, Ventricular Tachycardia, and Sinus Arrhythmia, as well as segments of noise that are rejected from the analysis. The histograms show the VE and SVE counts, the averaged heart rate, the coupling interval and the percent of prematurity.

PHASE 2 IN GRAPHIC MODE

The phase 2 analysis can also be operated in an interactive mode. In this mode, event records can be displayed on the graphic screen, together with the baseline, QRS onset and end points (the two vertical bars bracketing the premature beat), parameters measured and the classification results, Figure 3.16.

0: 4: 35: UE HEART RATE = 83 TOTAL COUNT = 18
QRS WIDTH = 140 MS NORMAL RR = 720 MS COUPLING = 620 MS

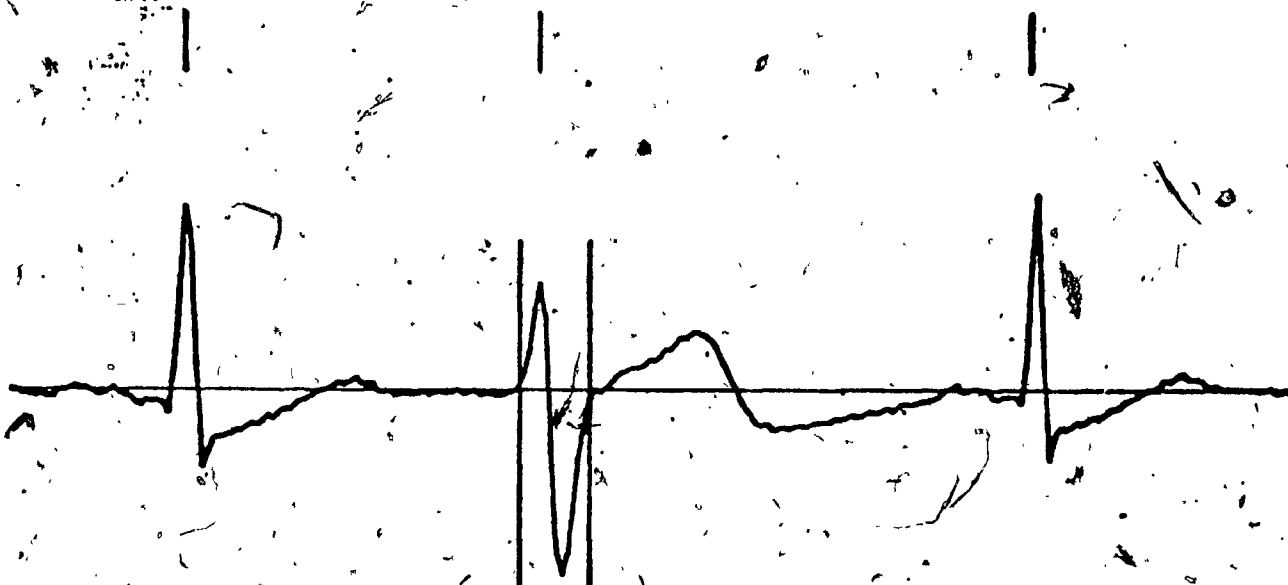


Figure 3.16: Measurements Displayed by PHASE 2 Classification System, a Ventricular Ectopic Beat.

Figure 3.17 shows a false Supraventricular ectopic beat caused by Sinus Arrhythmia, this was detected by the program and the message "***delete***" was displayed.

SUE
DELETE
0: 4: 39: HEART RATE = 84 TOTAL COUNT = 3
QRS WIDTH = 80 MS NORMAL RR = 710 MS COUPLING = 490 MS

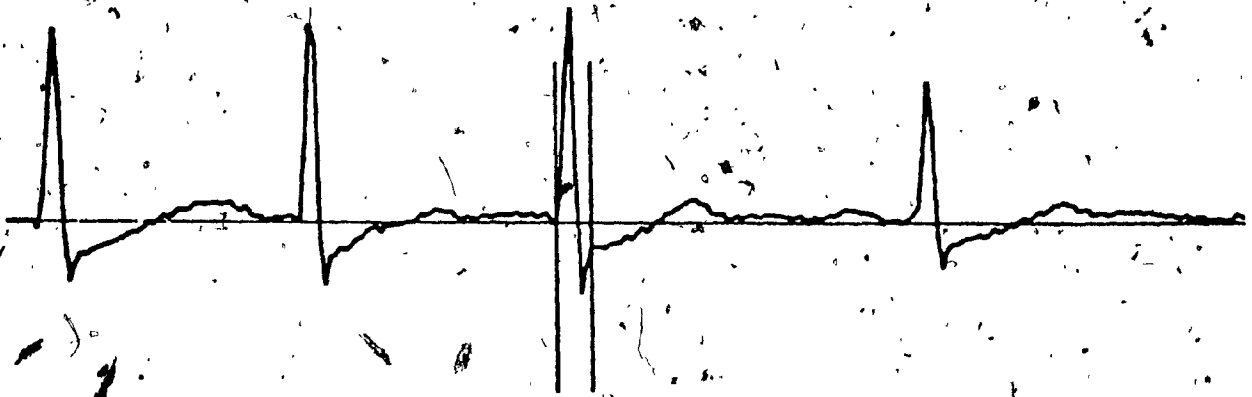


Figure 3.17: False Supraventricular Ectopic Beat Caused by Sinus Arrhythmia, Detected and Deleted by PHASE 2 Analysis system.

The graphic presentation of results permits close observation of the accuracy of the baseline, calculation, QRS width measurement and classification can be observed visually for each beat. In this mode of operation, an interactive function is provided which allows the operator to quickly locate any record according to its time of occurrence. The program can also be operated in a continuous mode, where each record with its classification is displayed on the graphic screen for a period of 2-15 seconds (depending on the positions of four sense switches) before advancing to the next record or in a "single-step" mode under the operator's control. With this graphic and interactive capability, reviewing of the results becomes a relatively simple task. Cardiologists can examine the beats visually, verify the results of the classification, QRS onset/end points and baseline estimation. This feature of the off-line system was designed to facilitate the testing and development of algorithms. It was also part of the verification tool used to determine the accuracy of the various measurement and classification logic.

SUMMARY OF THE SYSTEM

Special care was employed in the design of the user interface of the system. Two criteria guided the design of the user interface of the system: minimum operator intervention and maximum visibility of the results of the analysis. The former is an essential criterion for a practical system. The time and effort required to analyse a tape must necessarily be minimized if this diagnostic tool is to be of practical use to health professionals. The second criterion, visibility, is required to provide a basis for establishing the clinical acceptability of the system. The ability to display sample waveforms classified by the system as ectopic is an essential feature of the system, as illustrated by the typical mode of use of commercial analog tape scanners [Holter, 1961].

Minimal operator training is required as the basic analysis mode of operation requires only a simple "yes" or "no" answer prompted by the system on initialization. The second phase can be performed entirely without operator intervention and is capable of correcting errors made by the phase 1 program. By splitting the analysis into two phases the total amount of memory required is greatly reduced making it suitable for implementation on micro-processors. Very flexible interactive graphic functions are provided for system development and validation.

A rough estimate of the phase 1 program's speed reveals that it could operate up to 120 times real time. The lack of a

suitable tape player, however, prevented the testing of the program's maximum speed.

The phase 1 scanning of the tape requires about 26 minutes, approximately 2 minutes for mounting the tape and initialization and 24 minutes for scanning a 24-hour tape. The analysis time required for the second phase program is directly dependent on the number of ectopic beat records. Approximately 1 minute of processing time is needed for every 1200 records. Experience so far has shown that most tapes contain less than 5000 ectopic beats, so that the total processing time for a 24-hour tape is under 30 minutes. If a suitable tape recorder were available to operate the phase 1 program at 100-120 times real time, the total processing time could be reduced to about 15 minutes. This compares very favourably to the Avionics commercial system; and to all other published experimental systems.

CHAPTER FOUR

DETECTION AND CLASSIFICATION ALGORITHMS

INTRODUCTION

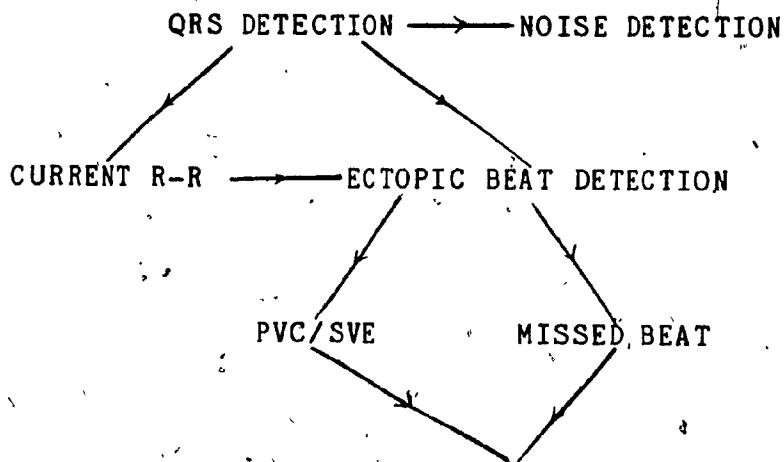
As in the case of most biological signals, the ECG is fairly well defined in the time domain but poorly characterized in the frequency domain. It lends itself to heuristic feature extraction methods rather than mathematical techniques, particularly in the case of Holter tape analysis, where a high volume of data must be processed in a limited period of time. Linear pattern recognition techniques such as cross correlation and Fourier analysis are of limited use because of these factors. The design of the detection and classification algorithms reflect these general constraints, moderated, to a certain extent, by the partition of the analysis process into two phases.

During the on-line phase, the ECG recording is scanned at 60 times real time. The detection algorithm must, therefore, operate effectively in an environment where a new sample is accepted every 166.6 microseconds, and a QRS complex must be processed every 80-100 sample points, while the event records in the worst case must be written onto secondary storage approximately every 10 beats. These factors impose very strict time constraints on the first phase detection and classification algorithms. In the second phase of the analysis, the event records are retrieved from the disc file and classified as VE or

SVE. Ectopic rhythm detection is performed on the basis of the result of the classification and timing information in the event record. In order to classify the ectopic beats, a number of parameters have to be measured and classification logic has to be developed. The parameter measurement algorithms must perform detailed and complicated measurements on the recorded beat. In contrast to the first phase, however, the only time constraint imposed on this analysis is the consideration of practical system design. The on-line, off-line algorithms and their relations are illustrated in Figure 4.1.

In the following sections, the characteristics of the ECG signals and different beat detection methods will be examined, and the algorithms designed for this implementation will be described. The VE/SVE classification parameters, their measurement algorithms and classification logic will be presented.

ON-LINE PHASE



-----EVENT FILE-----

OFF-LINE PHASE

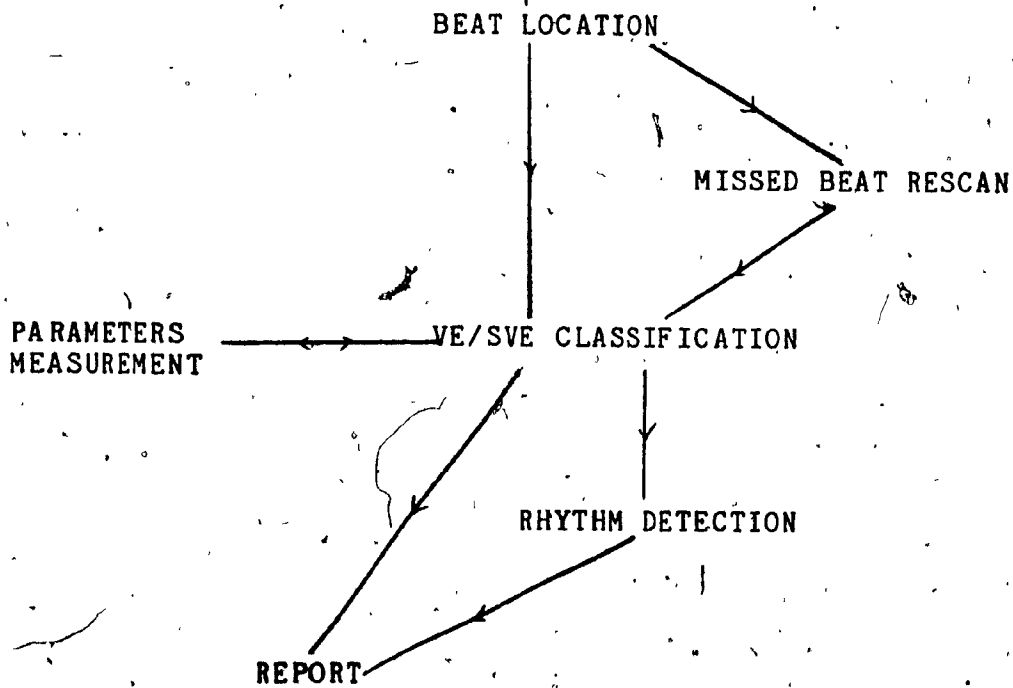


Figure 4.1 : Software Organization of the Analysis System.

BEAT DETECTIONAMBULATORY ECG RECORDING CHARACTERISTICS

Body surface ECG is predominantly contained in the frequency band from 0.5 to 100 Hertz [Pipberger, 1975]. Some small notches occur during Ventricular depolarization (the QRS complex), and slow moving parts of the ST segment may have frequencies outside this band. In general, the information required for arrhythmia analysis on these long term recordings is contained well within this range. Spectral analysis of ECG signal shows that the P wave is about 6 Hertz, the QRS about 15 Hertz and the T wave around 4 Hertz [Scher, 1960]. These frequencies, however, vary and sometimes overlap [Feldman, 1971]. Since the objective of ambulatory ECG analysis is to detect the gross changes in the shape and timing of QRS complexes, a upper frequency of 50 Hertz is sufficient.

The quality of Holter tape recording varies considerably from patient to patient. These variations are primarily the result of slightly different electrode placements, variations in the physical position of the heart, the build of the patient and his activities [Oliver, 1971]. A variety of artifacts have also been observed in these recordings. They can be divided into two basic categories, pseudo-arrhythmia and non-arrhythmia [Karsnow, 1976].

Pseudo-arrhythmias are usually caused by problems related to the functioning of the recorder. Examples are "premature contraction" caused by intermittent slowing of the recorder for

brief periods of time, "Sinus or Atrial Tachycardia", a relatively common artifact caused by battery failure, which usually appears towards the end of the recording and "Atrial Fibrillation" due to loose electrodes. Non-arrhythmia artifacts are usually in the form of high frequency noise contamination which completely obliterates the ECG signal, baseline drift and 60 Hertz interference. The 60 Hertz interference and baseline drift can be eliminated by suitable filters. The high frequency noise usually has components that overlap with that of the ECG signal but it can be detected by virtue of the fact that the signal is erratic. The pseudo-arrhythmias, however, are extremely difficult to recognize.

ON-LINE DETECTION ALGORITHMS

Detection and classification of arrhythmia types through the high speed scan of a Holter tape must satisfy requirements of efficiency and reliability in the recognition of the appropriate waveforms. To satisfy these requirements, it is necessary to determine a minimal complete set of features that characterize the different arrhythmia types. While completeness, which is necessary for reliability, is a quality shared with all types of an automated ECG analysis, minimality is the quality that is particular to Holter tape analysis, and is essential to the design of a practical system.

For the derivation of the required features it is not sufficient to consider standard or typical "heartbeat" waveforms.

The many non-pathological factors that can influence these waveforms must be taken into account, and features that are independent of these variations must be postulated, adapted, and exhaustively tested. Features which are differential, rather than absolute measurements, must also be considered, together with the means of calculation of the adaptive threshold values that determine their effectiveness. Many of the features discussed in the following were derived from previously published work, but their particular application, in the environment of high speed on-line digital processing without the use of a preprocessor, has required refinement and adaptation.

Since QRS detection is an on-line function operating at very high speeds, the overhead of the algorithm must be kept to a minimum. As the presentation on existing ECG recorders is limited to a single lead, the beat detection method used for ambulatory ECG recording analysis must be based on the information available on a single ECG lead. In addition to having a low overhead, the algorithm must be insensitive to the following factors inherent in the data:

- 1) the natural patient to patient variations due to physical differences
- 2) variations due to differences in lead placement in different recordings
- 3) variations in recording quality, such as gain, flutter, and noise pick up.

In order to automatically measure the parameters of an ECG, a stable reference point must be determined for each heart beat

complex. The most logical region to establish a beat detection point is the QRS complex. It is the most consistent component in the ECG signal and it has a number of distinctive features: it has a characteristic shape, a high amplitude, a large area under it and rapid changes in the slope.

Beat detection based on a combination of these criteria has been successfully applied in a number of monitoring systems [Frankel, 1975; Bussmann, 1975]. The required processing speed for ambulatory ECG recording analysis does not, however, leave time for multiple parameter measurement. The detection algorithm should, therefore, be based on a single criterion. The different distinctive characteristics of the QRS peak are as follows.

SHAPE

The QRS can be recognized by its characteristic shape. This can be accomplished by a number of pattern recognition techniques, such as, for example, cross correlation with a model, or piecewise detection of various deflections. These techniques are, however, very time consuming especially in consideration of the fact that the inter and intra patient variation of the QRS shape can be considerable. QRS detection based on its shape alone is therefore not a practical method in 'high speed' ECG analysis.

AMPLITUDE

Beat detection by the amplitude of the QRS is a very simple and fast method [Feldman, 1971], which is very effective for "well-behaved" ECG's. Amplitude measurements of QRS complex are, however, highly subject to lead placement which can affect the shape of the complex, and recording problems which can introduce such as noise contamination and baseline drift. These factors, together with the large patient-to-patient and beat-to-beat variation in the ECG signal render this detection method unacceptable for recordings taken in less than ideal conditions.

AREA

The area under a portion of the curve can be estimated by summation of a number of points on the curve, i.e. integration through a window. This technique is basically equivalent to the enhancement of waveform in a frequency dependent manner. This algorithm requires adaptation to waveshape variation and may result in confusion in cases where large T wave and baseline drifts are present.

SLOPE

The slope of the ECG signal can be positive (up slope) or negative (down slope), either one of these can be used as a detection criterion [Milliken, 1970]. The maximum positive slope is found on the uprise of the QRS complex, however, it is not of

very large magnitude and noise spikes can have values very close to it. The negative slope (down slope), on the other hand is much steeper and remains to be the most dominant feature even in abnormal QRS complexes.

Slope or the first difference of the ECG signal can be calculated by simply subtracting the $X(i)$ th point from the value of the $X(i+1)$ th ECG data point, Figure 4.2.

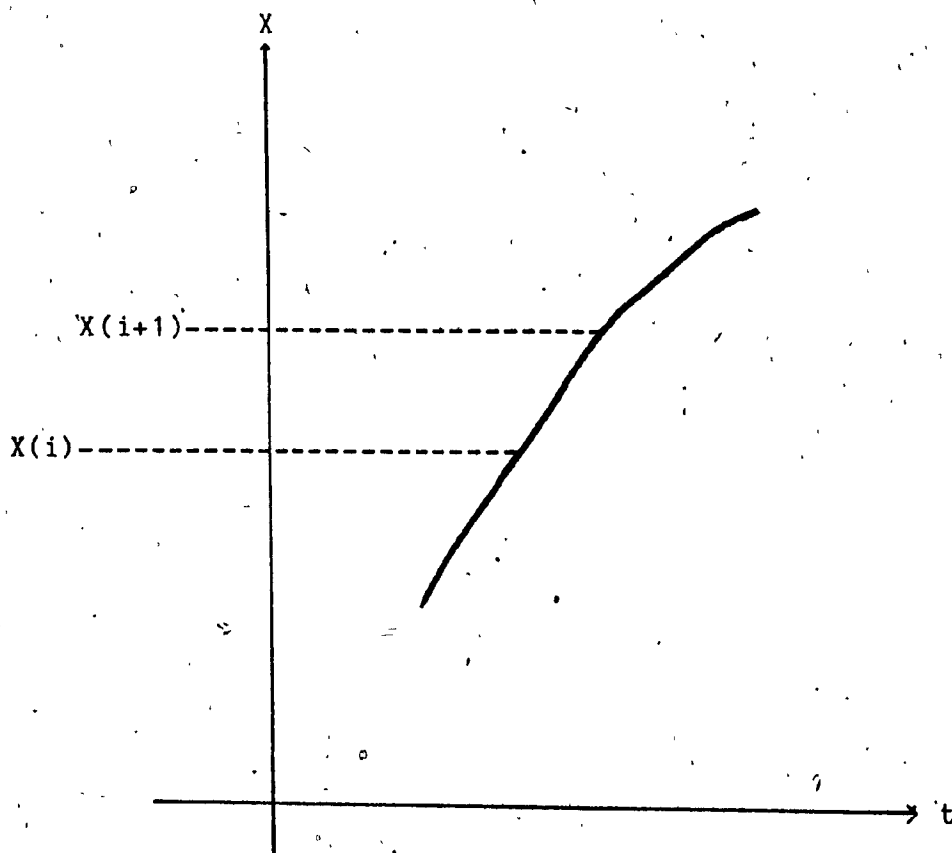


Figure 4.2 : Slope Measurement.

ON-LINE QRS DETECTION ALGORITHM

Of the various detection parameters tested, the negative slope was retained as the basis for QRS detection. Slope measurements are, however, susceptible to noise in the highest frequency range of the digitization (40 to 50 Hz). This noise, which can be attributed to either muscular activity, or pick up due to faulty connections during recording or digitization, can result in spurious high slope indication between adjacent points. It can also create slope reversals (notches) in the QRS complex. Although digital filtering could be used to attenuate this noise, it would require a complex filtering function to avoid excessive distortion of the QRS wave, whose signal components are only an octave lower in frequency. The resulting overhead would be in conflict with the requirement for a very efficient detection method. The use of slope alone as a parameter for detection of QRS complexes requires, therefore, a means of ensuring noise immunity as part of the measurement process, in addition to adaptivity to cope with normal, non-pathological variations.

The algorithms developed for QRS detection fulfill these requirements. The QRS is detected by searching for a negative slope calculation at 20 millisecond intervals which is below an adaptive threshold. The effect of noise in the higher frequency range of digitization is reduced by using this two point slope measurement instead of a point to point first difference:

Two point slope:

$$S_n = x(n) - x(n-2)$$

First difference of signal smoothed with a two point average:

$$\begin{aligned} S_n' &= \frac{x(n) + x(n-1)}{2} - \frac{x(n-1) + x(n-2)}{2} \\ &= \frac{x(n) - x(n-2)}{2} \end{aligned}$$

$X(n)$: value of sample n

S_n : two point slope at n

S_n' : first difference at n of the signal smoothed with two point average

While this requires no more calculation than the first difference it is equivalent, apart from a scale factor of 2, to the first difference of a signal smoothed with a two point average. The resulting attenuation curve, Figure 4.3, effectively reduces frequency components between 35 and 50 Hz, the frequency range where noise most seriously disturbs the QRS detection algorithm. The quality of the threshold value is maintained by updating with the maximum negative slope of a QRS complex rather than simply the first sample greater than the threshold. Since this slope rarely has more than four sample points, the maximum can be determined from the greater of the first two slopes which exceed the threshold:

$$S_n > S_t: \quad S_f = \max \langle S_n, S_{n+1} \rangle$$

S_n - slope at sample point

St - threshold slope

Sf -- slope at detection point

This ensures that the greatest possible margin is maintained between the slope threshold and high frequency noise.

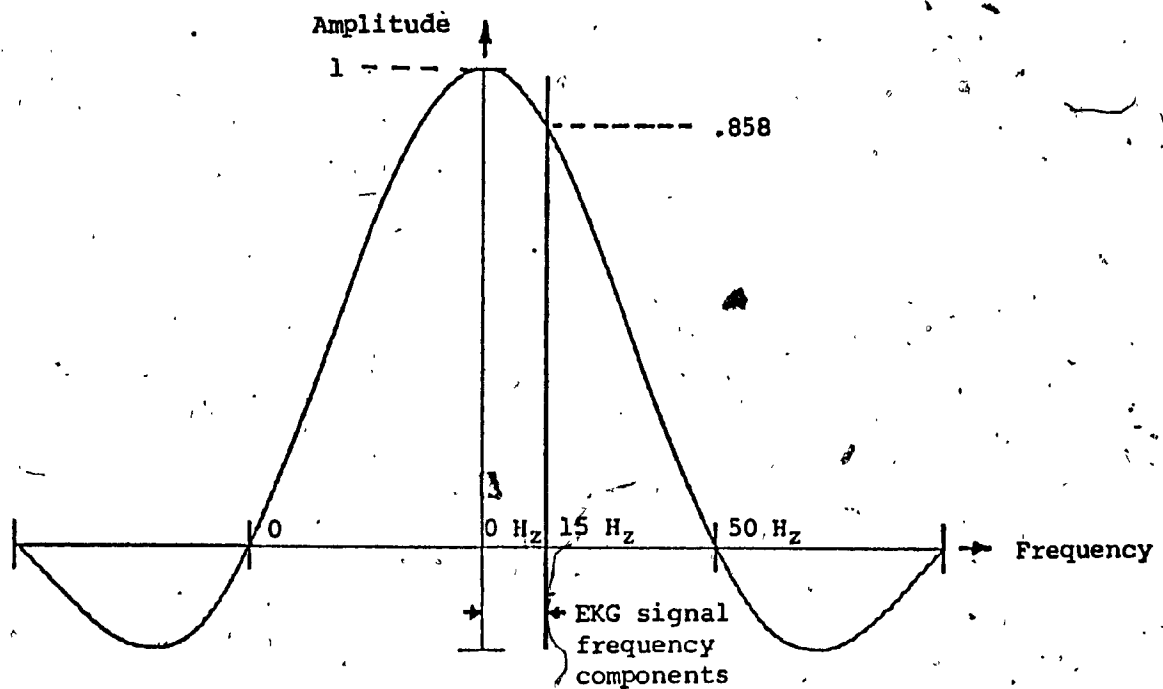


Figure 4.3 : Smoothing Effect of 2 Point Average.

In order to maintain an accurate threshold level, an initialization routine at the start of each tape average measures the maximum negative slope of three normal QRS complexes. The threshold, calculated on the basis of this measurement, is then maintained and updated by a recursive averaging formula which accepts each new value for a normal QRS complex and averages it with the previous samples. The use of two levels of adaptivity, the initial adaptation to the tape under analysis and the continuous adaptation to non pathological variations within the tape, permit a narrow margin between the normal slope and the threshold slope. This in itself improves accuracy of detection, as it eliminates lower frequency noise and artifacts. After a beat has been located the detection algorithm is inhibited for a period of time equal to $1/4$ of the average R-R interval to avoid multiple detection of the same complex.

The normal QRS complex is a relatively invariant characteristic of the ECG waveform. The majority of the classical Premature Ventricular contractions are easy to detect, because they have a large amplitude and a steep slope (Figure 2.4). Experience has shown, however, that there are a large variety of Ventricular premature beats which are markedly different from the classical "well-behaved" PVC's. Some Premature Ventricular Contractions are of very low amplitude, (Figure 2.5), and are difficult to detect. The problems posed by this type of beat have been noted by a number of researchers, [Gerling, 1972; Oliver, 1971; Hinkle, 1967], but no solution has

been offered. These beats represents a significant percentage of the ectopic beats and should be taken into consideration when designing a detection algorithm. Premature Atrial Contractions have the same shape as normal QRS's and can thus be detected only by their asynchronism. The detection method for the various types of ectopic beats will be described below.

ECTOPIC BEAT IDENTIFICATION

PVC AND SVE

An ectopic beat can be recognized by two criteria: synchronism and shape (see chapter 2). In an overwhelming majority of cases, it can be detected by the fact that it is out of synchronization with the normal rhythm. The clinical definition of prematurity is, however, qualitative and insufficient to define a computer identification algorithm. Research experience in arrhythmia analysis has established that a beat prematurity of 10-13% relative to the current R-R interval constitutes an adequate threshold for automated analysis [LeBlanc, 1975; Willems, 1972; Geddes, 1971; Wantanabe, 1969]. The optimal threshold value will, however, necessarily be affected by the method of calculating the current R-R interval, and therefore cannot be simply adopted from previous results. In the system described here, prematurity is determined on a threshold comparison of -12% of the current R-R interval. Once a beat has been detected, it is relatively simple to calculate the R-R interval and determine if it is premature. The average R-R

interval, however, is subjected to fluctuations due to the normal physical or emotional activities of the patient. These fluctuations occur in a fashion which is completely unpredictable for the analysis process, and the speed and extent of variation from the current heart rate may be considerable. The average rate (average R-R) must therefore be an adaptive value which follows the true rate very closely. The degree of success of this premature beat detection method is directly affected by the accuracy of the estimated heart rate.

The most common method for the computation of the average heart rate is a running mean of several beats [Sheffield, 1976; Gerlings, 1972]. This alone, however, is not sufficient, since transient changes (such as premature beats) may result in a sudden decrease in the rate. Some other constraints must be imposed on when and how the updating should be done. The average heart rate computation algorithm must have the ability to track normal heart variations but must not be affected by transient fluctuations. The main criterion of the updating algorithm introduced in this work is to include only the R-R interval of "normal" beats. Any arrhythmia that has an irregular rate is not used in the computation of the average rate.

One major type of arrhythmia that hampers the operation of this algorithm is Sinus Arrhythmia, (Figure 2.10), which is a sudden change (increase or decrease), sometimes transient, in the heart rate. If ectopic intervals are excluded from the averaging algorithm, a sudden increase in rate by more than 12% will result

in a large number of false premature beats, Figure 4.4. Compensating logic must, therefore, be built into this updating method to recognize this case and adapt to it as quickly as possible.

FALSE DETECTIONS

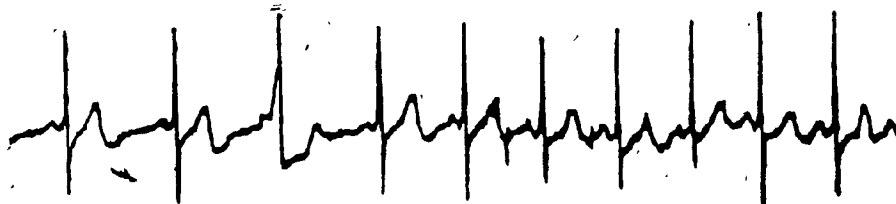


Figure 4.4 : False Supraventricular Ectopic Beat Caused by Sinus Arrhythmia.

When a premature beat occurs, a maximum of 2 beats (the premature beat and the compensatory pause) are excluded from the updating. Spurious detection caused by a sudden increase in rate, is identified by simply counting the number of beats skipped since the last update. If no updating has been done for more than 3 beats, a "forced" update takes place, using a different updating formula.

The computation of the current R-R interval was designed as a compromise between tracking ability and computational efficiency. It is a 1st order recursive averaging formula with a built-in scaling factor:

$$RRavg(n) = A*RR(n)+B*RRavg(n-1)$$

RRavg(n): value of current R-R interval at sample n

RRavg(n-1): value of previous calculation

RR(n): measured value of R-R interval

A = .25 constant determined by experimentation

B = .75 constant determined by experimentation

The parameters A and B are constants whose values were optimized by experimentation, and then subsequently adjusted to correspond to coefficients that could be computed by shifting rather than floating point multiplication. The recursive algorithm will track variations in heart rate more closely than a 5 point average, with a smoother response due to the natural exponential weighting of past inputs. The recursive computation is considerably simpler than the 5 point average used in most

previous works, as it not only requires fewer additions, but does not need any buffer management of recent readings. A higher order averaging function would give marginally better tracking but would increase the overhead of the algorithm considerably.

The values of these constants directly effect the performance of the averaging algorithm. A weight factor that is biased more towards the previous average, for example,

$$RRavg(n) = 0.1 * RR(n) + 0.9 * RRavg(n-1),$$

gives a more sluggish response to changes, while a ratio that favours the new rate, for example,

$$RRavg(n) = 0.9 * RR(n) + 0.1 * RRavg(t-1),$$

results in extreme sensitivity to small fluctuations. The 0.25 and 0.75 weight factors provide a near optimal rate of response to heart rate changes under normal condition.

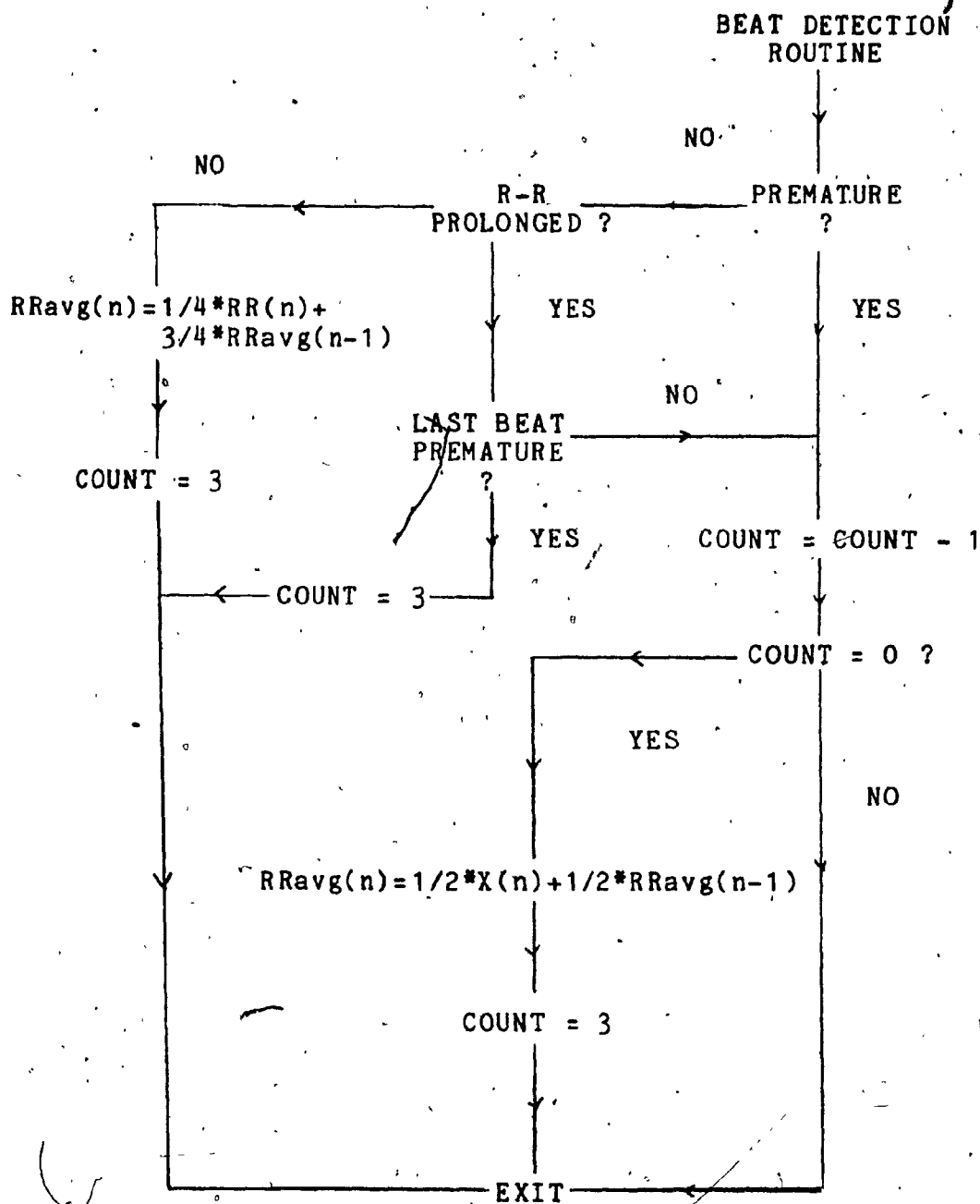
In case of a "forced" update, a different ratio is used:

$$RRavg(n) = 0.5 * RR(n) + 0.5 * RRavg(n-1).$$

this ratio is biased more towards the new rate than the normal updating formula so that the running average will adapt to the current rate in a short period of time. The purpose of this second updating formula is to provide a means for the algorithm to adapt to abrupt heart rate variations.

This method was found to be superior to the simple averaging method used in many ECG monitoring and analysis systems. If no truncation occurs in the computation of this rate, i.e. an infinite accuracy and word size, then the algorithm is in fact a sum of all past rates with exponentially decreasing weight factors. In the context of this analysis program, (word size of the computer and the range of the rate), it is approximately equal to the summing of the past 10 to 15 heart rates with exponentially decreasing weight factors. Such a truncation is advantageous in this case, since errors in the rate will not propagate.

Figure 4.5 shows a flow chart of the heart rate updating condition and algorithm.



COUNT is initialized to 3.

Figure 4.5 : Flowchart of the Average R-R Computation Algorithm.

FUSION BEATS

In most cases of fusion beats, the waveshape is so irregular that the maximum slope is less than the threshold of the QRS detection algorithm, (see chapter 2). Their detection is especially difficult due to the limited choice of detection algorithms dictated by the scanning speed requirement. Their presence can nevertheless be determined by the failure to detect a beat resulting in an apparently prolonged R-R interval. Upon recognition of this, the interval can be rescanned in phase 2 using a different detection strategy which is tailored to detect this type of beats. The rescan method will be described in more detail later.

ECTOPIC BEAT DETECTION ALGORITHM

Ectopic beats are detected by 2 criteria based on the R-R interval. A beat is flagged as premature if the R-R interval is more than 12% shorter than the running average R-R. If the RR interval is more than 150% of the average R-R, then it is almost certain that the QRS detection algorithm has failed to detect a beat, and the "missed" beat is probably a aberrant Ventricular fusion beat described above. Figure 4.6 shows the logical operation of the ectopic detection algorithm.

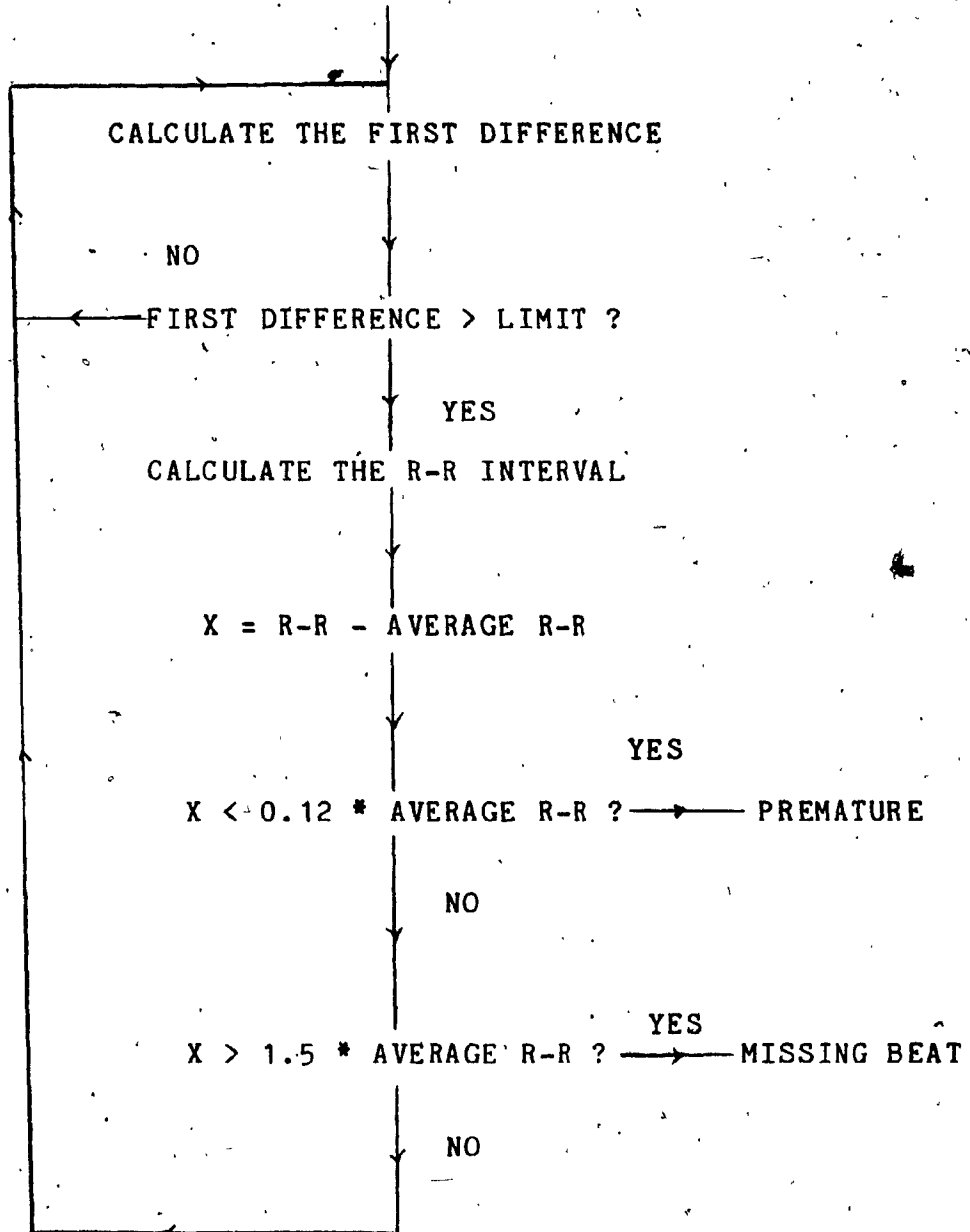


Figure 4.6 : Ectopic Beat Detection Algorithm.

HIGH LEVEL NOISE ELIMINATION

The presence of noise spikes resembling QRS complexes in the ECG recordings is a constant source of problems that cannot be ignored. High frequency noise and baseline drift in an ECG signal can be removed by filtering the signal before processing by the computer. The filtering could be performed by an analog or a digital filter [Lynn, 1971]. The frequency spectrum of the noise, however, often overlaps with that of the ECG and filtering is of limited use [Feldman, 1971]. Furthermore, when the signal is filtered, some distortion is inevitably introduced. In this case the reshaping of noise spikes into QRS like waveforms is a common side effect.

The noise manifestation illustrated in Figure 4.7 is the most common type, where the ECG signal is completely obstructed and no rhythmic waveform can be recognized. This noise interference, if undetected, could result in serious errors in the analysis. Most of these disturbances can be detected by two characteristics: they usually have much steeper slopes and are very close together, resulting in a rate much higher than the normal range of heart rate [Bussmann, 1975; Hochberg, 1969]. Since these noise disturbances are usually transient and obscure only a minute fraction of the information on a tape, a loss of several minutes would not result in any significant changes in the overall analysis result. It is therefore more practical to omit portions of the recording that are contaminated by artifacts, rather than to attempt to filter or analyse them. In the beat

detection algorithm lower amplitude noise is eliminated by the two point averaging scheme and more pronounced disturbances are recognised by a sudden change to a rate outside normal limits (greater than 230 or less than 30 beats per minute), "QRS complexes" that are too close together or too far apart. Once they are recognized the beat detection algorithm is inhibited for a short period of time (two seconds recording time) before analysis is resumed.

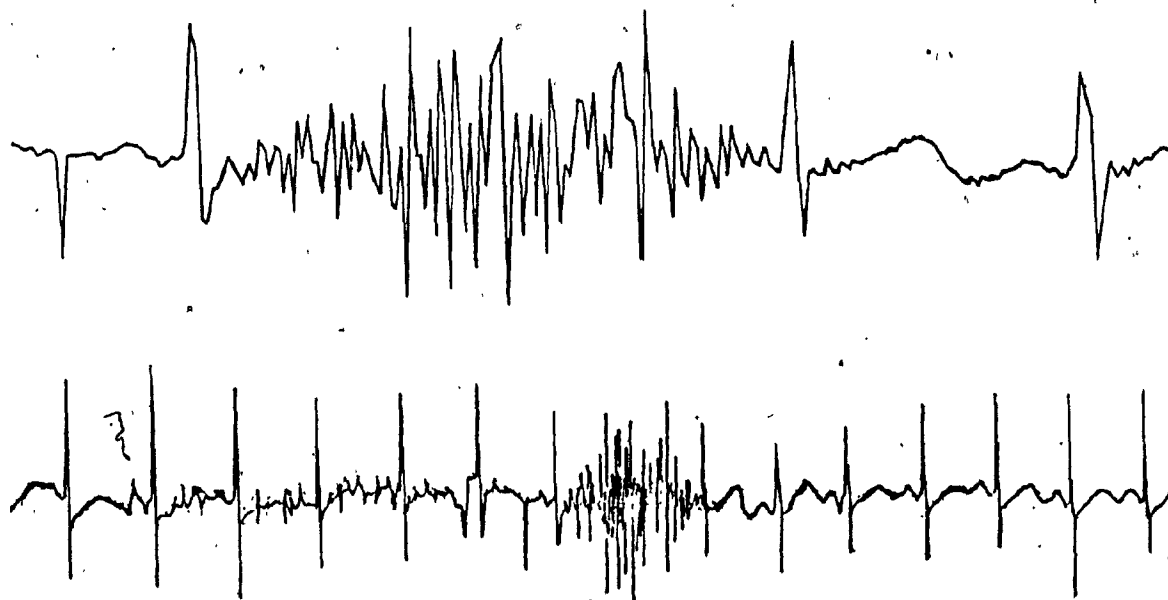


Figure 4.7 : Typical Noise Disturbance (Top Display is the Magnification):

OFF-LINE ECTOPIC BEAT CLASSIFICATION

In the first section of the second phase analysis, a reliable set of ectopic beat classification logic has to be developed. The parameters needed for the classification logic have to be extracted from the event records. This involves the location of beats in the record (including rescan for missed beats), parameter measurement and ectopic beat classification. In the second section, rhythm detection is performed based on the result of the classification and the timing information on the event record. The operations of these two sections are interleaved, with rhythm detection being performed as the beats are classified. Since this phase of the analysis is an off-line operation which does not require real time data acquisition, processing speed is not as critical as in the on-line system.

BEAT LOCATION

The circular buffer is first converted to a straight buffer using the pointer stored in the record. If the ectopic R-R interval is greater than the average R-R, then there probably is a low amplitude aberrant beat which was not detected. In this case the interval is scanned for the maximum negative slope Figure 4.8. If this is greater than 20% of the normal QRS slope, the point at which it is measured is taken to be the detection point for a "missed" beat. If the slope is beneath the threshold, the record is classified as unknown and discarded. If the premature R-R is shorter than the average R-R then the premature beat is located using the information in the record.

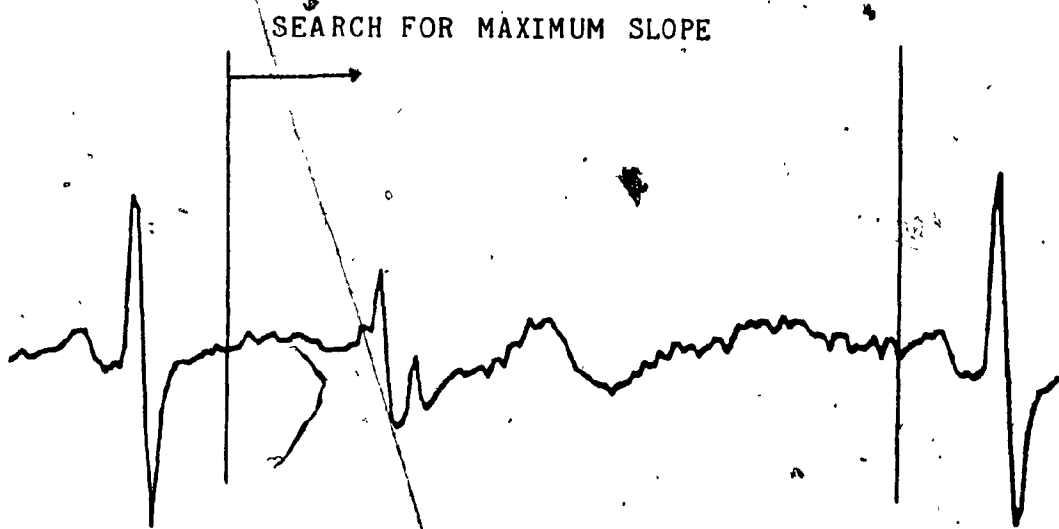


Figure 4.8 : Rescan for "missed" Beat.

ECTOPIC BEAT CLASSIFICATION PARAMETERS

The two basic types of ectopic beat, VE and SVE, can be differentiated by three parameters of the ECG waveform: the presence of a P wave, the shape of the QRS complex and the existence of a compensatory pause. Numerous methods have been developed for the detection of the P wave [Klusmeier, 1976; Breithardt, 1975; Rey, 1971]. These detection methods usually involve laborious computations and correlation of information from several ECG leads. The degree of success has been very limited and even less in the detection of P wave based on the information contained in a single lead. In view of these factors, it is impractical to use the P wave as a classification parameter. The shape of the QRS complex can be divided into two parts: the QRS width and amplitude. In an overwhelming majority of cases, the VE has a much wider QRS complex and an amplitude that is markedly different from that of a normal QRS complex. SVE beats, in contrast, are very similar in shape to normal heart beats. The VE is usually followed by a full compensatory pause (see chapter 2), while the SVE is not.

BASELINE

In order to measure the QRS width and amplitude, a baseline is needed as a reference point. The baseline is estimated by averaging the flat portion before the QRS complex. Such a portion is taken at 160 milli-seconds to 450 milli-seconds (30 points) before the R peak. The region chosen for determining the baseline is justified by the fact that it is not affected by any cardiac activity, Figure 4.9.

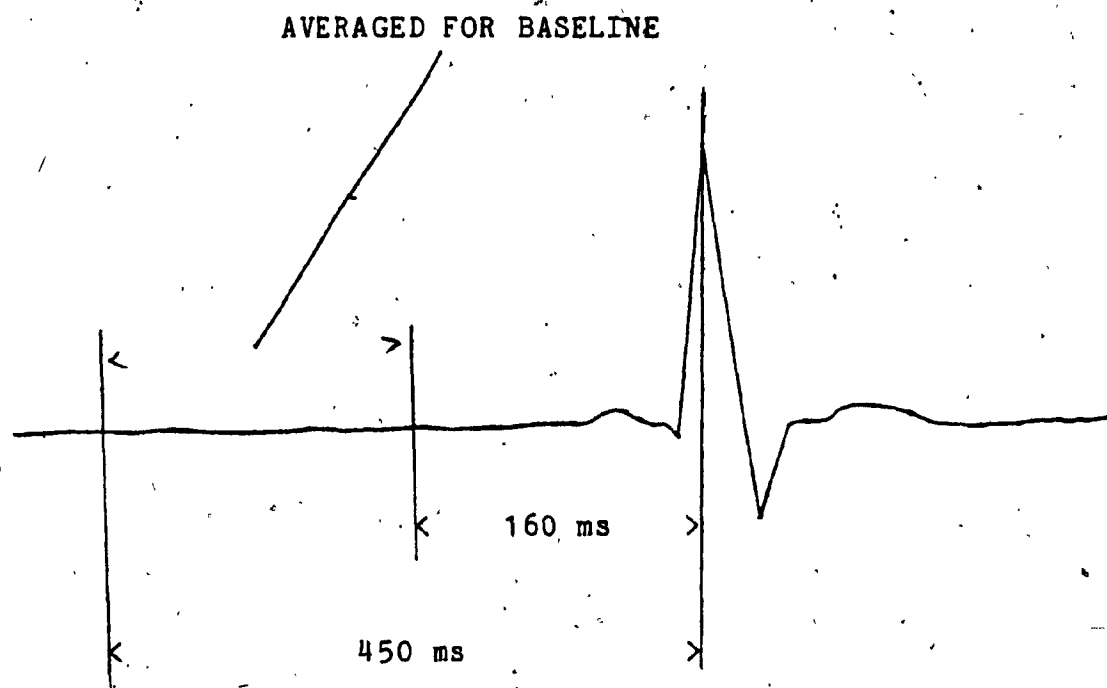


Figure 4.9 : Baseline Estimation.

QRS AMPLITUDE

The QRS amplitude is represented by two values, the height of the R peak above the baseline and that of the S wave below the baseline, Figure 4.10.

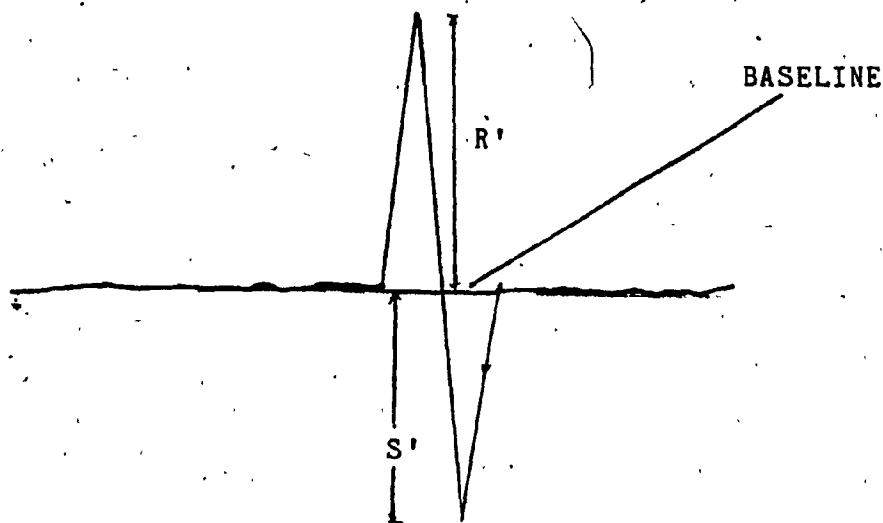


Figure 4.10 : QRS Amplitude Measurement.

QRS WIDTH

The QRS width is the interval between the commencement of Ventricular contraction to the termination of contraction (complete depolarization), Figure 4.11. In a normal ECG, the width is approximately 80 milli-seconds. It is fairly constant and does not vary much in normal ECG's of different patients.

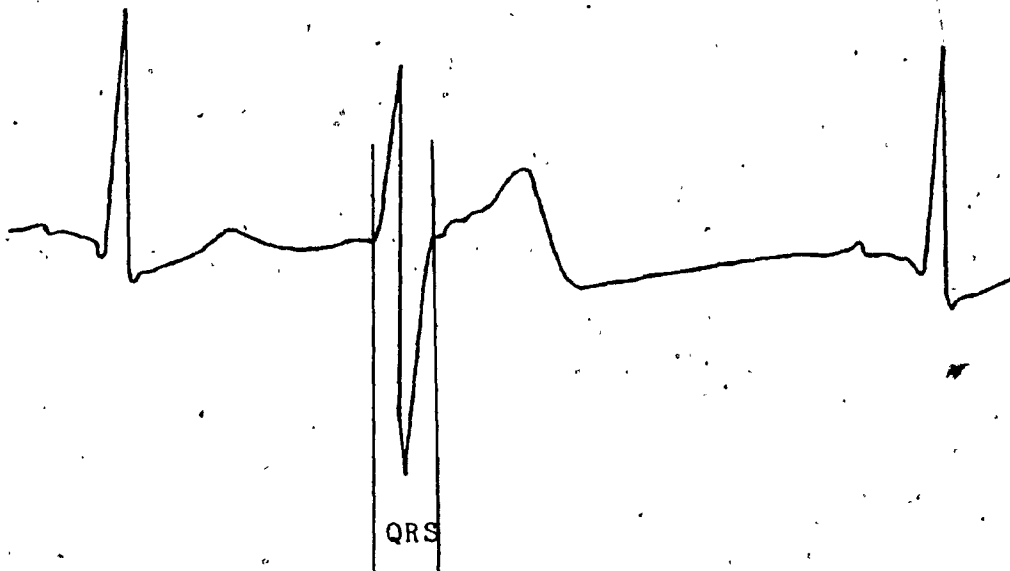


Figure 4.11 : QRS Width of a PVC.

As a concession to the presence of noise in ECG recordings and the large variations in QRS waveshape, no attempt was made to detect the exact QRS onset and end. The assumption was made that these points are detected only accurately enough so that the complexes could be classified.

The onset and end points of the QRS are estimated by a combination of change in slope and proximity to the baseline. The onset point is found by searching backward from the R peak to a point where a significant change in the slope occurs. A significant change in slope is defined as an area where the difference between two consecutive slope values is greater than 80%. The slope is calculated by subtracting two points which are 20 milli-seconds apart. Once such a point has been located its amplitude is compared to that of the R wave (R' of Figure 4.8), if it is above 25% of that of the R wave then the search is resumed, otherwise the point is taken as the QRS onset point. The end of the QRS is estimated in the same manner except the slope is taken as an absolute value (to include the S wave). Figure 4.12 shows the estimated QRS width for a Premature Ventricular Contraction.

0: 4: 35: ^{UE} HEART RATE = 83 TOTAL COUNT = 18
 > QRS WIDTH = 140 MS NORMAL RR = 720 MS COUPLING = 620 MS

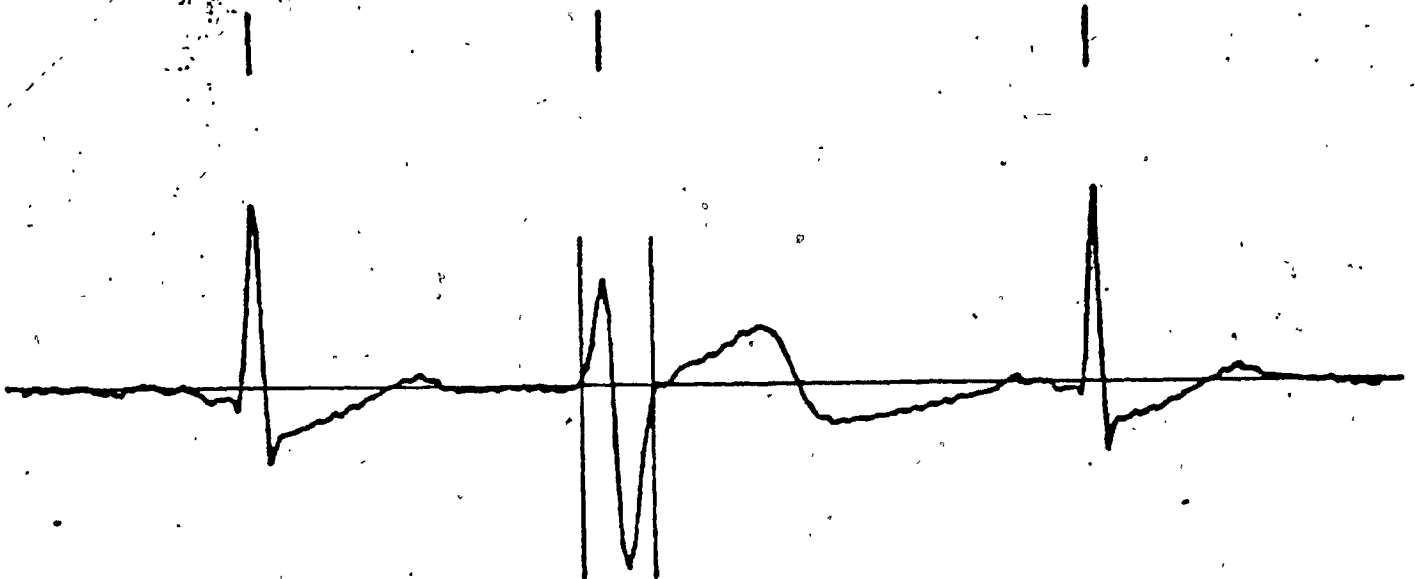


Figure 4.12 : QRS Width of a PVC Estimated by the Analysis system.

COUPLING INTERVAL AND COMPENSATORY PAUSE

The coupling interval is the time interval between the premature beat and the beat preceding it, indicated as interval X1 in Figure 4.13.

A compensatory pause is a cycle following a premature beat which is long enough to compensate for the premature interval. A compensatory pause is said to exist if the combined length of the premature R-R interval, X1, and the length of the cycle following the premature beat, X2, is greater than 188% ($200 \pm 12\%$) of 2 normal R-R intervals.

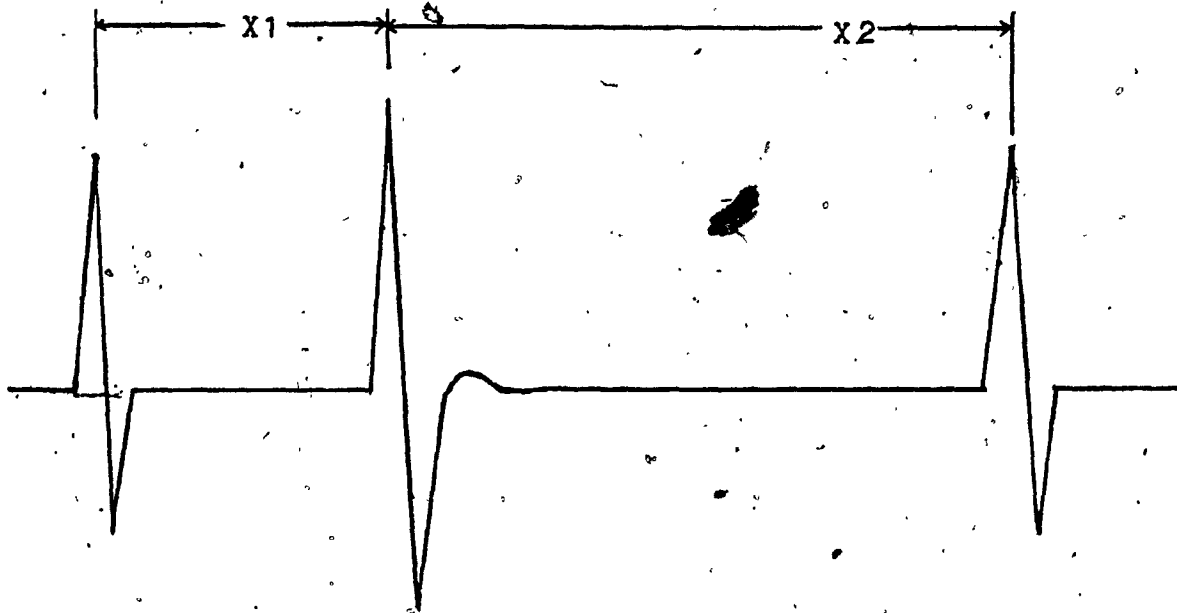


Figure 4.13 : Coupling Interval and Compensatory Pause.

VE/SVE CLASSIFICATION LOGIC

The premature beats are classified according to the results of the parameters measured. A beat is classified as Ventricular if one of the following conditions, A or B, holds:

A) the QRS width is greater than 110 milli-second;

OR

B) two or more of the following :

- 1) the QRS is greater than 90 milli-seconds,
- 2) there is a compensatory pause,
- 3) R' or S' of the premature beat differ by more than 50% from that of the preceding beat.

The classification logic was designed in such a way that it will function with a complete or partial set of parameters. The width was chosen as the primary decision parameter since it is one of the most dominant features which differentiates between VE's and SVE's. In cases where the measured width is between 90 and 110 milliseconds, however, waveshape, R' and S', and the compensatory pause are included in this logic. This compensates for the potential inaccuracy in the width measurement due to recording variation. The amplitude and compensatory pause are only used as secondary parameters. This arrangement ensures the highest reliability possible with the available data. Since the ectopic beat is centered in the event record, its width can always be measured. The shape measurement requires the presence

of the beat preceding the premature beat which is included in the record except in cases where the heart rate is extremely low, below approximately 43 beats per minute. In most cases the percent of prematurity is greater than 12%, so that in practice the preceding beat will be stored. By the same reasoning, the beat following the ectopic beat will also be stored.

The VE and SVE counts are stored in two separate arrays with 48 elements each (summed over 30 minutes). For the Premature Ventricular Contractions, the premature R-R interval (coupling interval) is also recorded.

RHYTHM DETECTION

Cardiac rhythm is associated with the timing and the order of occurrence of ectopic and normal heart beats. The event records contain complete timing information about the ectopic beat stored and its type, which is determined by the classification section. These, together with the current R-R interval stored in the event record constitute sufficient information for the detection of ectopic beat related rhythms. The detection methods for Bigeminy, Trigeminy, Ventricular Tachycardia and Sinus Arrhythmia are described below.

BIGEMINY

Bigeminy is a rhythm disorder where every second beat is a Premature Ventricular Contraction (Figure 2.6). It can be detected by the ratio of normal and premature beats in continuous recording. In this system the event records are short (2.5 seconds), but include accurate timing information locating the premature beat. The beat classifications together with this timing information are used to identify Bigeminy.

The time elapsed between two Premature Ventricular contractions can be calculated by the difference in the time of arrival of the two beats. This interval compared to that of the average R-R rate yields an approximation of the normal and premature beats ratio (at least for short intervals). A message is printed if such a rhythm is detected and it persists for more than 8 beats (8 premature beats). Figure 4.14 shows how this

algorithm works.

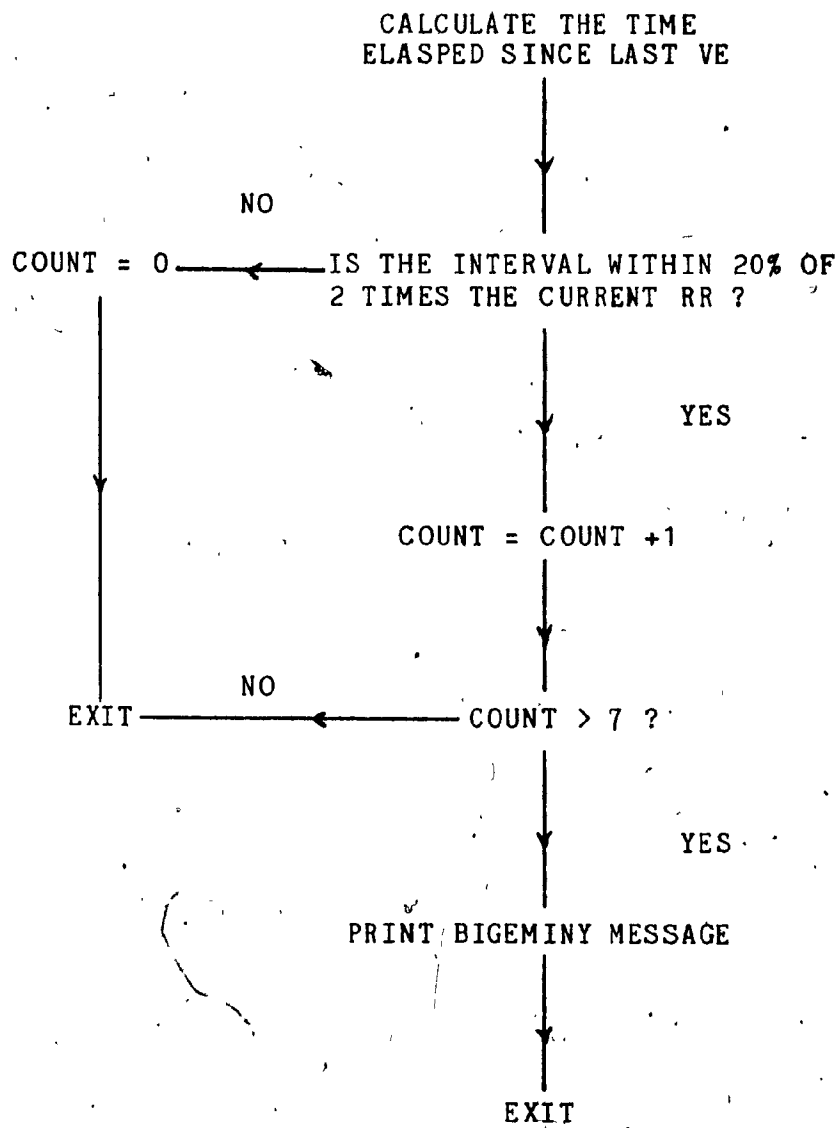


Figure 4.14 : Bigeminy Detection Algorithm.

TRIGEMINY

Trigeminy is a rhythm where every third beat is a Premature Ventricular Contraction, (Figure 2.7). It is detected by a method similar to that used for Bigeminy.

VENTRICULAR TACHYCARDIA

Ventricular Tachycardia is a continuous series of Premature Ventricular Contractions, (Figure 2.8). It is recognized by the fact that the premature beat to premature beat interval (Ventricular beats) is shorter than the average R-R. A message is printed if the series is longer than 3 beats.

SINUS ARRHYTHMIA.

Sinus Arrhythmia is characterized by a sudden change in the heart rate, in a series of normal beats (Figure 2.9). A sudden increase in rate is usually detected by the phase 1 program as a series of Premature Atrial Contractions. This classification is corrected by the second phase. If a continuous series of Premature Atrial Contractions has been detected, the beats are not counted and a Sinus Arrhythmia statement is printed indicating the time of occurrence.

In order to achieve the efficiency required, the on-line beat processing algorithms were implemented in assembly language. For

① simpler implementation the off-line ectopic classification and rhythm detection algorithms were programmed in FORTRAN. A summary of these algorithms is given in Appendix F.

CHAPTER FIVE

SYSTEM EVALUATION

EVALUATION PROCEDURE

The evaluation of the performance of an ECG analysis system is an important part in the development of the system. Since no standard evaluation procedure is available for testing this type of ECG analysis system, designers have to devise their own evaluation methods. The normal evaluation procedure is to compare the cardiologists' interpretation to the computer's result. Most of these evaluations have had one or more of the following faults in common:

- A) very limited amount of test data,
- B) the data on which the system was evaluated was sometimes the data on which it was developed,
- C) limited number of cases of arrhythmias, especially Premature Ventricular Contractions.

The evaluation methods used for the system reported here were designed specifically to avoid these faults.

The cardiologist is the ultimate judge of whether a given heartbeat is normal or abnormal, and his decision is made on the basis of visual inspection of the beat and its relationship to other beats in a recorded trace. In this context, an accurate performance evaluation of an automated arrhythmia analysis system requires a beat by beat comparison of a set of recorded ECG

segments with the results produced by the detection algorithms of the system. This can be done statistically, by a comparison of the results of a set of segments analysed by a cardiologist with the results of machine analysis on the same segments. It can also be done deterministically, through a beat by beat inspection of the ECG trace with the analysis results superimposed on each ectopic beat. This second technique has the advantage of ensuring an exact correspondence between the segments analysed by the cardiologist and those analysed by machine. In addition, it provides a display of the measurements made by the computer which can assist the cardiologist in marginal cases.

The evaluation of the analysis system was performed with the assistance of the support software used in algorithm development. This acquisition and display software was used to acquire one hour segments of ECG tapes for storage on digitized files, to subsequently submit them to the analysis routines of the system, and then to provide the interactive display capability that supports the deterministic evaluation of the system. The visual magnification feature was particularly valuable for the judgement of marginal beats (Figure 3.7).

In the following the evaluation procedures are outlined for each aspect of the performance of the system. The results of the system testing are discussed with reference to each of these procedures.

MATERIAL

The test data set includes 6 tapes selected from 18 which were not used in the development of the system. An attempt was made to include as much variation in the QRS waveshape as possible when the tapes were selected. For the verification of the rhythm detection algorithms, 4 tapes with frequent arrhythmia were used. To test the repeatability of the analysis, 3 randomly selected tapes were used. The ectopic beat content of the tapes selected ranges from very frequent (over 800 per hour) to occasional (20 per hour).

ECTOPIC BEAT DETECTION/CLASSIFICATION

A total of 5 1/2 hours of EKG recording selected from the six tapes of the test data was used in this evaluation. The selected data consisted of contiguous segments of about one hour taken from each test tape. The procedure consisted of digitizing the tape segment, submitting the file to the analysis routine, and then displaying the results as described above on the Tektronix display screen. The operator, trained to recognize Ventricular and Supraventricular beats under the supervision of a cardiologist, recorded all false positive and false negative beats. A total of 23,722 beats were examined, with the results of the analysis classifying 22,573 beats as normal and 1149 as beats premature. The details of these results are shown in TABLE 5.1. The beats are classified according to Ventricular Ectopic (VE) and Supraventricular Ectopic (SVE).

The total false positive count, consisting only of SVE's, is 3, out of a total of 1149 detected ectopic beats. This gives a false positive rate of .013% of the 22,573 normal beats. The false negative count, including both VE's and SVE's is 6, for an overall percentage of 0.5% of the 1149 ectopic beats. Divided according to VE's and SVE's, the false negative rate is .4% and 2.7% respectively.

The higher incidence of SVE errors is due to the similarity between normal beats and SVE's. Because the width of these beats may be the same as that specified for normal beats, all false positive detections will be classified as SVE's. False negatives in this category are beats that fall just beyond the prematurity criteria of -12%. SVE's are more likely to fall in this area than VE's which tend to be clearly out of synchronism. It should be noted that these results were obtained without any "tuning" of the speed of the tape recorder, and without any special adjustments.

PATIENT	DURATION (HOUR)	TOTAL # OF BEATS	DET	VE FP	FN	DET	SVE FP	FN
A	1	4073	78	0	0	3	0	0
B	1	4258	429	0	0	1	1	0/
C	1	4371	50	0	2	16	2	0
D	1/2	2142	444	0	1	54	0	2
E	1	4500	21	0	0	0	0	0
F	1	4378	49	0	1	0	0	0
<hr/>								
TOTAL	5 1/2	23722	1071	-	4	74	3	2
			99.6%	0%	0.37%	-	0.013%	2.7%

*** DET - Number Detected

*** FP - False Positive

*** FN - False Negative

TABLE 5.1 : Evaluation Results of Premature Beat Detection and Classification.

RHYTHM DETECTION

The evaluation of the rhythm detection capability of the system is different in nature from that of the premature beat detection and classification algorithms. In general, rhythm patterns characterized by series of ectopic beats, such as Trigeminy and Ventricular Tachycardia, are much less common than the occurrences of single premature beats, even in the case of seriously ill patients. Consequently, a much larger volume of recording is needed to verify the accuracy of the rhythm detection algorithms.

It is impractical to visually scan long ECG records manually (it would require 7200 feet of electrocardiographic tracing paper for 24 hours of ECG plotted at the speed of 1 inch per second, or approximately 12 hours using EXSYS). Tape segments containing high occurrences of arrhythmia were visually selected using an oscilloscope. The result of the visual scan is compared to the result generated by the analysis system. Twelve hours of recording, from 4 different tapes, were used in the evaluation. False positives were evaluated by verifying the rhythm indicated on the printed reports. For this, a total of 576 hours (24 tapes) were used. No false positives or false negative were observed in this evaluation.

ANALYSIS SPEED

During periods of Ventricular Tachycardia, every beat is detected as ectopic and stored in the event file. Since Ventricular Tachycardia can last from seconds to hours, no assumption can be made about its duration. The system was, therefore, designed to handle the worst case condition of an indefinite string of ectopic beats. As Ventricular Tachycardia is not very common among these tape recordings, it was necessary to simulate it in order to test the system's capacity to meet this design specification. The threshold of ectopic beat detection was reduced from 12% to 0%, causing the ectopic beat detection algorithm to flag every beat as premature and therefore stored on the disc file. This test was performed using a number of different tapes and the program was never terminated due to an overflow of the circular buffer.

REPEATABILITY

One measure of the accuracy of an analysis system is the repeatability of the results. Under ideal circumstances, the results should be identical for the analysis of any given tape repeated at two different times. Under actual conditions, however, these results may be affected by a number of factors. Typical problems associated with the play back and digitization are as follows:

- 1) alignment of the playback head.
- 2) "flutter" (speed variations) in the tape drive. Different sample points resulting in minor differences in fiducial point locations. This may tip measurements beyond thresholds in marginal cases.
- 3) Noise in the playback system.
- 4) Start up conditions: this may affect a small number of QRS complexes. If a tape is started at a different point each time, some beats at the start of the tape may be included or excluded, depending on the starting point of the tape.

Most of these factors have a random effect on the analysis process, differing each time the tape is analysed. Some can, however, be eliminated by taking appropriate precautionary actions. Noise generated by the playback system, for example, can be suppressed with careful shielding and grounding, and the effect of flutter can be reduced by using a high quality tape deck.

The reproducibility of the rhythm classification results was tested by repeating the analysis 3 times on each of 3 tapes

selected at random from a library of 24 tapes. None of these tapes had been used during the system development process. The results are given in TABLE 5.2.

The variations in the total number of ectopic beats detected is very close to the total error determined in the previous tests. The standard deviation in the worst case for VE's is 0.4%, of the mean. The average percentage standard deviation for the three tests was 0.265%. The average percentage standard deviation for Supraventricular beats was 2.36%. These two results match the error observed in the beat detection evaluation, and suggest a system limitation imposed entirely by the data acquisition process.

PATIENT: A

TEST #	TOTAL # OF BEATS	VE	SVE
1	81582	3018	209
2	81601	3038	201
3	81598	3027	207
AVERAGE	81594	3028	206
STANDARD DEVIATION	10	10	4
%	0.012	0.33	2.0

PATIENT: B

TEST #	TOTAL # OF BEATS	VE	SVE
1	100973	1806	46
2	101022	1813	44
3	100989	1798	47
AVERAGE	100995	1806	46
STANDARD DEVIATION	25	8	1.5
%	0.025	0.4	3.3

PATIENT: C

TEST #	TOTAL # OF BEATS	VE	SVE
1	87232	8445	61
2	87237	8455	63
3	87229	8446	61
AVERAGE	87233	8447	62
STANDARD DEVIATION	4	6	1
%	0.0044	0.065	1.8

TABLE 5.2 : Results of Repeatability Tests.

CHAPTER SIX

SUMMARY AND CONCLUSIONS

SUMMARY

The system presented here has demonstrated the feasibility of the implementation of a high speed ambulatory ECG analysis system on a small general purpose digital computer. The entire system was implemented in software with no special purpose hardware preprocessor. The analysis requires only a single pass of the analog tape at 60 times real time, played back on an ordinary inexpensive audio tape player. No special training is required to operate the system and operator interaction with the system is kept to a minimum. Parameters for initialization of the phase 1 system are selected automatically, the operator intervention being acceptance or rejection of the displayed sample with a "yes" or "no" answer. Artifact filtering and detection functions are built into the beat detection routine. The beat detection parameters are adaptive, so that they could track normal changes in patient condition that may occur in the duration of the recording. A noise detection algorithm is built into the beat detection routine to recognize high frequency artifact.

Accurate quantitative information pertinent to the patient's condition is presented in a compact report organized according to the category of cardiac rhythm and time of day. Premature beats detected are identified as VE or SVE and classified where

appropriate into related rhythm disorders, including Bigeminy, Trigeminy, Ventricular Tachycardia and Sinus Arrhythmia. Since a detailed record is kept for each detected abnormal beat, the phase 2 analysis is not limited to the types mentioned above but is extensible to meet any further specific classification requirements that may be desired. All detection and classification parameters, with the exception of the QRS width, are adaptive. The second phase analysis is capable of recognizing and correcting the errors made by the phase 1 program which results in a much lower error rate than in other contemporary analysis systems. The system is capable of detecting low amplitude Premature Ventricular Contraction which had been found to be a problem in ECG analysis (see chapter 4). All premature beats detected are stored in a detailed event file. This feature together with the interactive graphic ability of the phase 2 program provides a very powerful research capability to supplement the practical analysis system.

In order to keep a detailed record of the premature beats detected, a fast mass storage device with approximately 2 mega bytes of storage is needed ($4500 \text{ records} * 256 \text{ words/record} * 2 \text{ bytes/word}$). This could be reduced to about 1 mega byte if the ECG signal is digitized to 8 bits instead of 12 (an accuracy of 8 bits is considered sufficient for ECG analysis [Dower, 1975]). The present disc capacity can hold approximately 8900 records. If a tape contains very frequent arrhythmia, the entire tape cannot be analysed in one single pass. In such cases, however, a

complete 24-hour analysis is probably unnecessary, since a partial analysis (10 to 12 hours) would provide sufficient information to support a complete diagnosis by a cardiologist. In our experience, however, only a few tapes were found to contain over 700 premature beats per hour. Due to the lack of suitable data, the system was not designed to analyse tapes from patients with an electronic pace-maker, however, modification of the system to include this type of tapes would not be difficult.

CONCLUSIONS

An automated, quantitative system for the analysis of ambulatory ECG recordings has been implemented and tested. From the test data available, this system appears to meet, if not surpass, the performance of any other reported system (TABLE 1.1). The results obtained with this system demonstrate a degree of accuracy that is more than adequate to justify its use as a research tool and as a clinical diagnostic aid for cardiologists. More significant, however, is the difference in the fundamental design concept of the system. The implementation of the entire process in software on a small digital digital has demonstrated the feasibility of computer analysis without the use of special hardware. Adaptation of the software to take advantage of standard microprocessor technology is a logical next step. The implications of the cost of such a system are significant, given the downward price trends of contemporary microprocessor hardware. The design features developed here could thus form the

basis of a low cost system that would be attractive as a mass screening diagnostic system. A second unique aspect of the system, its generation of the ectopic beat file in the first analysis phase, is also of interest for future clinical use. In its present implementation, it provides a convenient off-line facility for inspecting ectopic beat types occurring in a patient tape record.

APPENDIX A

COMPUTER ASSISTED ECG DIAGNOSIS

Research work on automated ECG diagnosis started in the late 1950's. The first computer program for automatic ECG wave recognition was developed by Pipberger et al in 1961 [Pipberger, 1962; Zywietz, 1973]. Most ECG analysis programs developed since have employed the same general principles of wave detection. Since his initial attempt, Dr. Pipberger and his group have developed both short record diagnostic [Willems, 1972], and ICU monitoring systems [Haisty, 1972]. At the same time, D. Carceres has created a fully automatic system for processing 12-lead ECG's on a short record basis [Caceres, 1969]. His diagnostic system utilizes a decision tree while Dr. Pipberger has based his system on multivariate statistical techniques to relate ECG parameters to diagnosis. According to the published information, the two different approaches yield approximately the same result. Numerous researchers have since entered the field and developed diagnostic systems with varying degrees of success. TABLE A.1 is a list of some the programs and the evaluations of their performance.

SYSTEM	# OF LEADS	RECORD LENGTH (seconds)	EVALUATION
V.A. Hospital [Willems, 1972; Zyweitz, 1973]	3	6-10	89%
Medical Systems Development Lab. [Caceres, 1970]	12	4	95%
Queen's University [Wartak, 1971]	3	10-12	80%
University of Minnesota [Burchell, 1976]	12	-	95%
University of Montreal [LeBlanc, 1972; 1976]	3	10-15	---

NOTE: The evaluation results quoted are those claimed in published reports.

TABLE A.1 : Short Record ECG Diagnostic Systems.

Published information concerning the evaluation procedure and data is usually sketchy and far from standardized. The lack of a definitive interpretation of a given ECG has prevented rigorous quantification of ECG diagnostic programs' strengths and weaknesses. These uncertainties were pointed out by Dr. Caceres [Caceres, 1976; 1973; 1969]. In a series of tests he discovered that the average agreement in ECG diagnosis between cardiologists ranges from 60 to 80%. In view of this, one might expect to find the accuracy of an automated ECG diagnosis system to peak at about the same percentage points. The diagnostic methods of a program are, however, inevitably tailored to suit the cardiologist(s) who participated in the development and evaluation of the system. Therefore, it is not surprising to find that a much higher accuracy is usually claimed.

APPENDIX B

AUTOMATED ECG MONITORING.

There are many significant differences between arrhythmia monitoring and short record ECG diagnosis by computer. In the former case, the objective is to detect transient cardiac abnormalities, while the latter attempts to determine the specific condition of the heart from the shape of the ECG waveforms. In consequence, quite different approaches are needed for optimum solutions in the two cases. In off line ECG diagnosis there are virtually no time constraints except for those imposed from an economic point of view. As much computing time as required can be spent in examining the data collected. In ECG acquisition procedure, recordings can be repeated by the technician until a record which is free of muscle tremor, baseline drift and other artifacts, has been obtained. This is, however, not true in ECG monitoring. A substantial part of the effort must go into the detection and elimination of noise disturbances. Since arrhythmia could occur at any time and lasts from seconds to hours, all beats should be analysed.

In order to optimize the cost effectiveness of a monitoring system, it is desirable to handle several patients simultaneously. The analysis, therefore, must be performed at speeds in excess of real-time. Another important consideration in real-time monitoring is that the patient is expected to go through a series of states and the monitoring system must be able

to track, and report these dynamic changes. Provision must be made so that alarm criteria can be modified to suit individual conditions. As a tremendous amount of data is examined in monitoring, verification of the analysis result is much more difficult than for ECG diagnosis, and even a very low error rate can lead to an intolerable number of false alarms.

Almost all computer arrhythmia monitoring systems contain five distinct functions: data acquisition-digitization, preprocessing (filtering, waveform coding), pattern recognition (beat detection), parameter measurement, and rhythm analysis (presentation of result or activation of alarm). While each individual system may employ special techniques in one or more areas, the general configuration remains much the same. The testing procedures are, however, usually not very well documented and the test data quantity and variety differ considerably from one system to another. An attempt to tabulate some of these systems is found in TABLE B.1. It should be noted in the assessment of these results, that the testing of a monitoring system is inherently more difficult than that of a diagnostic system due to the vast amount of data.

SYSTEM	# OF LEADS	ALGORITHM	EVALUATION
Washington University [Cox, 1972]	1	feature extraction	78%
Harvard University [Feldman, 1971]	1	stored normal cross correlation	80%
Medical Systems Development Laboratory [Hochberg, 1969]	1	QRS shape	---
V.A. Hospital [Haisty, 1972]	1	RR interval	85%
University of Virginia [Gerlings, 1972]	1	cross correlation	97%
University of Chicago [Fozzard, 1976]	1	feature extraction	90%
IBM [Bonner, 1969]	1	feature extraction	---
HEWLETT-PACKARD (COMMERCIAL)	1	"AZTEC" coding [Cox, 1968]	88%

NOTE: The evaluation results quoted are those claimed in the published reports.

TABLE B.1 : Arrhythmia Monitoring Systems.

APPENDIX C

THE ELECTROCARDIOGRAM LEAD SYSTEM

Traditionally, short record electrocardiograms were taken using a 12 electrodes system, commonly called "12-lead" system [Shaub, 1965]. Later work has shown that the same information can be captured by 3 leads which assume a certain degree of orthogonality, Figure C.1. Since the human body and the heart are 3 dimensional objects, the electrical activities of the heart cannot be completely represented by the information contained in a single plane (single lead). The 3-lead system, the "vectorcardiogram", attempts to capture a component of the cardiac vector in each of the 3 planes.

This 3-lead system measures the potential difference between two limbs. Lead I measures the potential difference between the right arm and the left arm; Lead II measures the potential difference between the left leg and the right arm and Lead III, the potential difference between the left leg and the left arm. This type of lead, which measures the potential difference between two points is called a bipolar lead [Shaub, 1965]. The different electrode placements of the various leads record a different projection for each lead. Figure C,2 illustrates the typical differences between the three leads in a normal ECG.

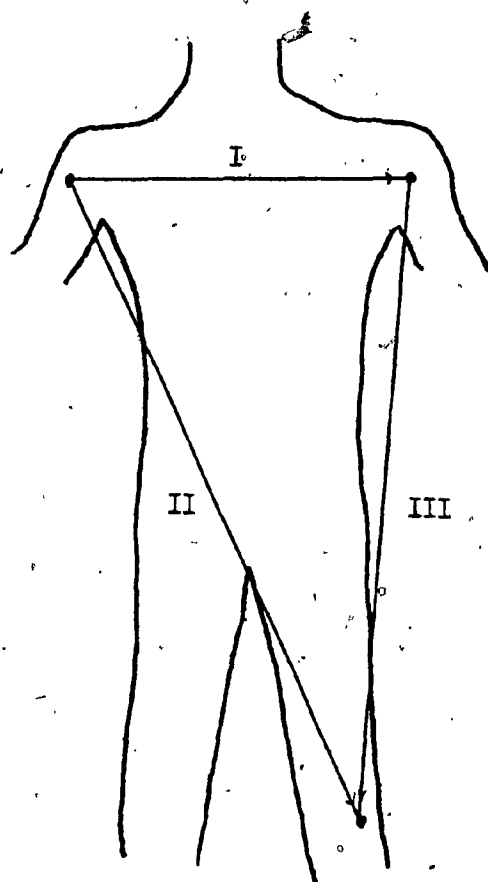
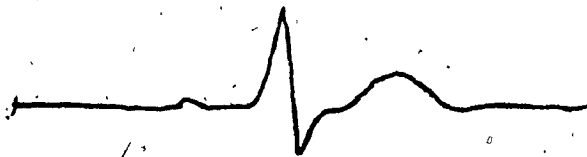


Figure C.1 : Electrodes Placement of the 3-lead System.



LEAD I



LEAD II



LEAD III

Figure C.2 : Typical ECG Waveshapes Obtained by the 3 Leads.

APPENDIX D

THE HARDWARE SYSTEM

The hardware configuration of the system is illustrated by Figure D.1. Figure D.2 is a photograph of the system.

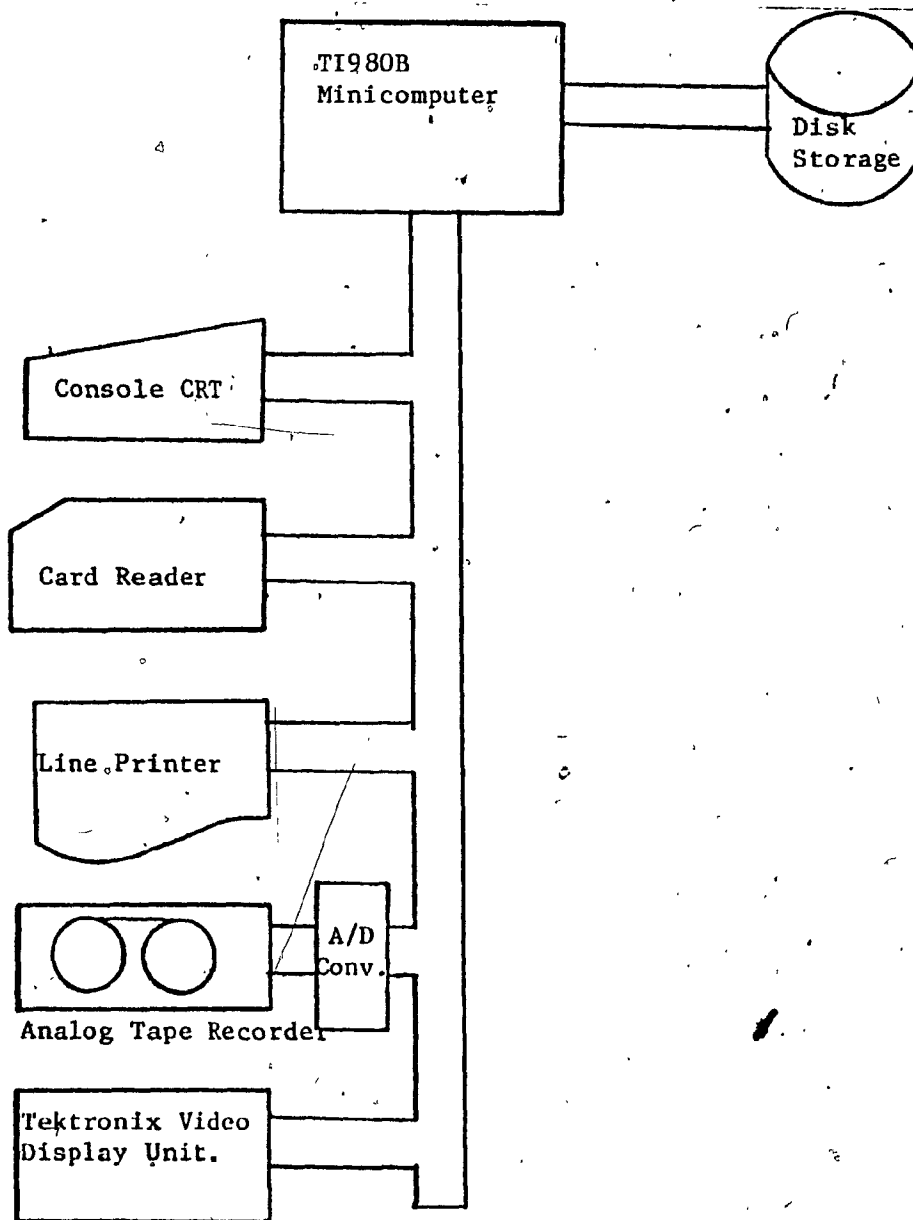


Figure D.1 : Hardware Configuration of the Experimental System.



Figure D.2 : Experimental set up of Holter Tape Analysis System.

ANALOG

The tape playback unit is an ordinary two-track monaural audio tape recorder-player, SONY model TC-106. The analog-to-digital (A-D) converter [Berson, 1973], has a resolution of ± 2048 (12 bits) and a maximum conversion time of 35 micro-seconds. It is connected to the input/output (I/O) bus of the computer and can issue interrupts to the central processor via the bus.

DIGITAL

The computer is a TI980B mini-computer manufactured by Texas Instruments Incorporated, Houston, Texas, with 32K words (16 bits per word) of main memory. Approximately 12K words of memory are actually required to run all the programs developed. The peripherals include a Tektronix 4010 graphics unit, a CRT terminal, a line printer and a magnetic disc unit. The disc unit has two drives, one fixed and one removable with a total storage capacity of approximately 9 megabytes.

The peripherals used were not specially chosen for the implementation, but were simply the available ones. They are not necessarily the optimal equipment, nor were they part of the design specification, as any other compatible equipment could also have been used. The use of a mini-computer with a 16-bit word size is not essential.

SOFTWARE SUPPORT

The software support includes a single user disc operating system supplied by the manufacturer (Basic System Manual #961961-9710), a FORTRAN IV compiler (FORTRAN Manual #961961-9740, 9742), and a symbolic assembler (Assembler Language Manual #943013-9701), which reside on the fixed disc unit.

APPENDIX E

ANALYSIS REPORT

Figures E.1 to E.6 illustrate a typical report generated by the PHASE 2 analysis system.

24-HOUR ECG ANALYSIS

PAGE 1

RECORD OF PREMATURE BEATS

IDENTIFICATION: B - C

TAPING DATE: 76/01/05

TAPING TIME: 00/00/00

TIME = 0 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 130	SVE = 0
SINUS ARRHYTHMIA	0 HRS 52 MIN 5 SEC		
TIME = 1 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 164	SVE = 0
TIME = 1 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 280	SVE = 0
TRIGEMINY START	1 HRS 47 MIN 11 SEC		
TRIGEMINY END	1 HRS 47 MIN 16 SEC	COUNT = 10	
TIME = 2 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 319	SVE = 0
TIME = 2 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 48	SVE = 1
TIME = 3 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 89	SVE = 2
TIME = 3 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 50	SVE = 0
TIME = 4 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 82	SVE = 0
TIME = 4 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 122	SVE = 1
TIME = 5 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 83	SVE = 1
TIME = 5 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 13	SVE = 1
TIME = 7 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 10	SVE = 0
TIME = 7 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 1	SVE = 0
TIME = 8 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 2	SVE = 1
TIME = 9 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 6	SVE = 0
SINUS ARRHYTHMIA	9 HRS 32 MIN 27 SEC		
TIME = 10 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 20	SVE = 0
TIME = 10 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 10	SVE = 1
SINUS ARRHYTHMIA	10 HRS 51 MIN 27 SEC		
TIME = 11 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 156	SVE = 3
TIME = 11 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 270	SVE = 23
SINUS ARRHYTHMIA	11 HRS 32 MIN 15 SEC		
SINUS ARRHYTHMIA	11 HRS 39 MIN 54 SEC		
SINUS ARRHYTHMIA	11 HRS 41 MIN 26 SEC		
SINUS ARRHYTHMIA	11 HRS 59 MIN 13 SEC		
TIME = 12 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 224	SVE = 16
TIME = 12 HRS 30 MIN	COUNT FOR THIS 1/2 HOUR	VE = 180	SVE = 3
TIME = 13 HRS 0 MIN	COUNT FOR THIS 1/2 HOUR	VE = 23	SVE = 0

Figure E.1 : Event Summary.

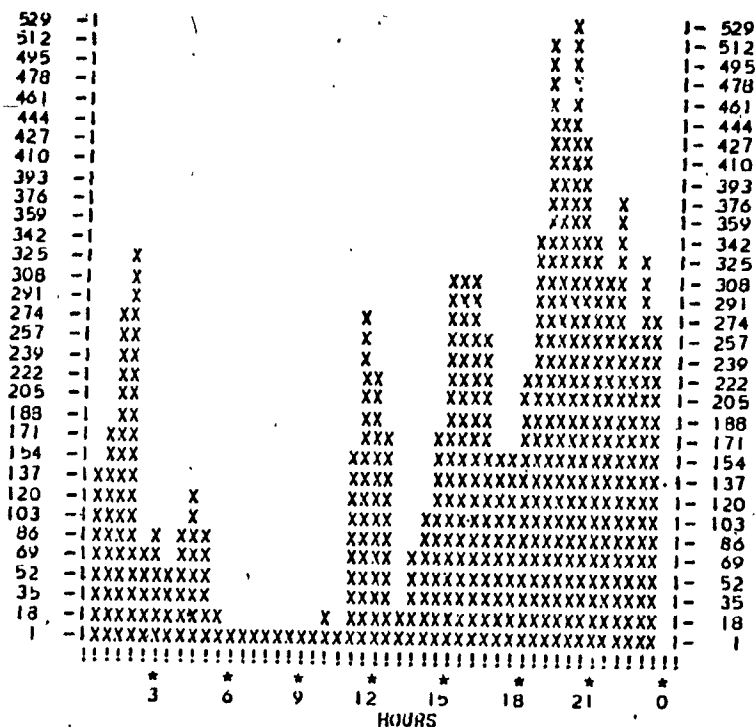
TIME = 13 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 72 SVE = 2
 SINUS ARRHYTHMIA 13 HRS 44 MIN 22 SEC
 SINUS ARRHYTHMIA 13 HRS 56 MIN 60 SEC
 TIME = 14 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 111 SVE = 0
 SINUS ARRHYTHMIA 14 HRS 6 MIN 21 SEC
 TRIGEMINY START 14 HRS 20 MIN 59 SEC
 TRIGEMINY END 14 HRS 20 MIN 59 SEC COUNT = 8
 TIME = 14 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 175 SVE = 2
 SINUS ARRHYTHMIA 14 HRS 31 MIN 12 SEC
 TIME = 15 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 310 SVE = 0
 TIME = 15 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 306 SVE = 0
 TIME = 16 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 308 SVE = 0
 TIME = 16 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 256 SVE = 0
 SINUS ARRHYTHMIA 16 HRS 52 MIN 28 SEC
 TIME = 17 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 148 SVE = 0
 TRIGEMINY START 17 HRS 11 MIN 21 SEC
 TRIGEMINY END 17 HRS 11 MIN 24 SEC COUNT = 9
 TIME = 17 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 162 SVE = 0
 TIME = 18 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 223 SVE = 0
 TRIGEMINY START 18 HRS 21 MIN 33 SEC
 TRIGEMINY END 18 HRS 21 MIN 33 SEC COUNT = 8
 TIME = 18 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 339 SVE = 0
 TRIGEMINY START 18 HRS 30 MIN 36 SEC
 TRIGEMINY END 18 HRS 30 MIN 39 SEC COUNT = 9
 TRIGEMINY START 18 HRS 34 MIN 35 SEC
 TRIGEMINY END 18 HRS 34 MIN 38 SEC COUNT = 9
 TRIGEMINY START 18 HRS 38 MIN 19 SEC
 TRIGEMINY END 18 HRS 38 MIN 23 SEC COUNT = 10
 TRIGEMINY START 18 HRS 41 MIN 11 SEC
 TRIGEMINY END 18 HRS 41 MIN 18 SEC COUNT = 11
 TIME = 19 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 515 SVE = 1
 TIME = 19 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 441 SVE = 2
 TIME = 20 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 529 SVE = 0
 TIME = 20 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 426 SVE = 0
 TIME = 21 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 342 SVE = 0
 TIME = 21 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 307 SVE = 0
 TIME = 22 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 371 SVE = 0
 TIME = 22 HRS 30 MIN COUNT FOR THIS 1/2 HOUR VE = 341 SVE = 0
 TIME = 23 HRS 0 MIN COUNT FOR THIS 1/2 HOUR VE = 331 SVE = 0

TIME = 23 HRS 27 MIN COUNT FOR THIS 27 MIN VE = 268 SVE = 2

TOTAL TIME = 23 HRS 27 MIN 16 SEC
 TOTAL VE COUNT = 9503
 TOTAL SVE COUNT = 62
 TOTAL TIME LOST DUE TO NOISE = 0 HRS 0 MIN 0 SEC
 TOTAL NUMBER OF BEATS = 87301.

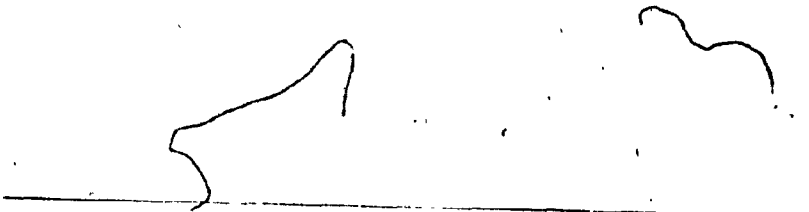
Figure E.2 : Event Summary.

PREMATURE VENTRICULAR CONTRACTION

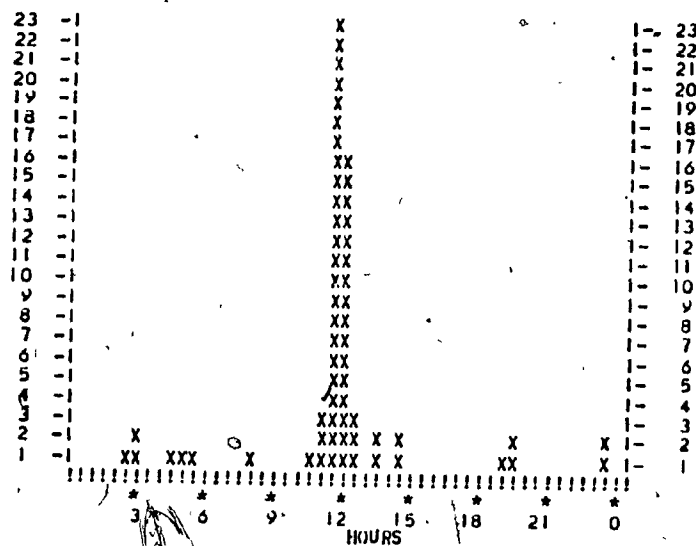


TOTAL TIME = 23 HRS 27 MIN 16 SEC

Figure E.3 : Ventricular Ectopic Beat Histogram:



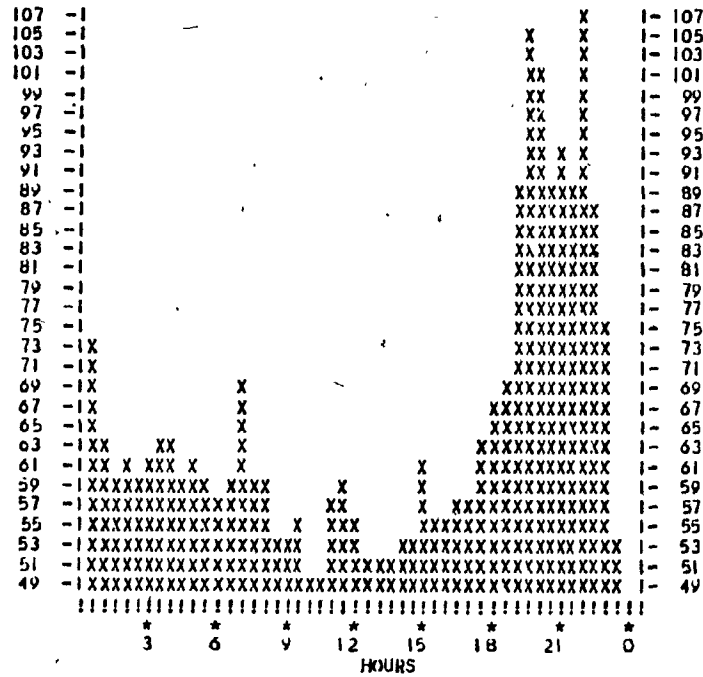
PRÉMATURE SUPRA-VENTRICULAIRE CONTRACTION



TOTAL TIME = 23 HRS 27 MIN 16 SEC

Figure E.4 : Supraventricular Ectopic Beat Histogram.

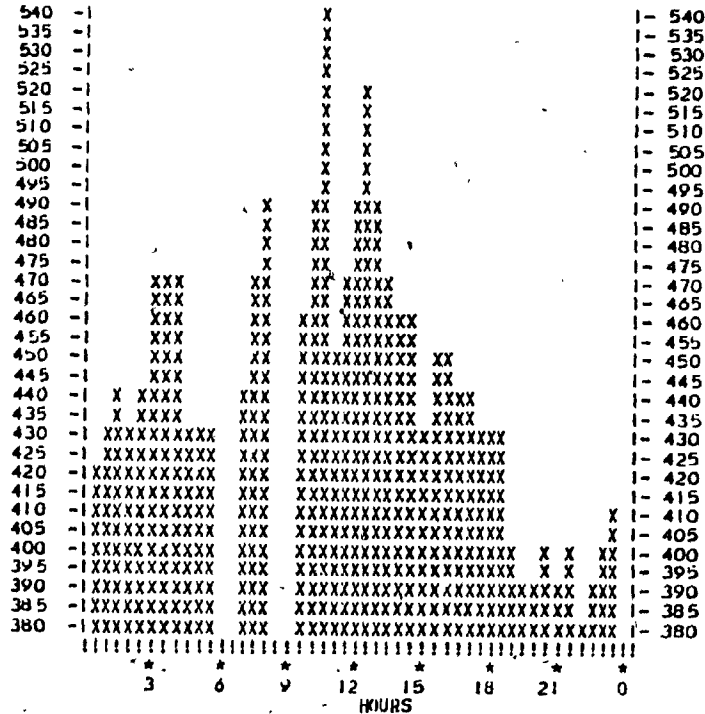
HEART RATE



TOTAL TIME = 23 HRS 27 MIN 16 SEC

Figure E.5 : Averaged Heart Rate.

COUPLING INTERVAL
ASEC



TOTAL TIME = 23 HRS 27 MIN 16 SEC

Figure E.6 : Averaged Coupling Interval.

PERCENT OF PREMATUREITY

	3	6	9	12	15	18	21	0		
59	-			X					1-	59
58	-			XX		X			1-	58
57	-			XX	A	XXXX			1-	57
56	-	X		XX	X	XXX			1-	56
55	-	X X	X	XX	XXX	XXXX			1-	55
54	-	XXXX	XXX	X XX X X	XXX	XXXX			1-	54
53	-	XXXX	XXX	X	XXXX	X-XXXXXXX			1-	53
52	-	XXXX	XXX	X	XXXXXXXXXXXXXXXXXX				1-	52
51	-	XXXX	XXX	X	XXXXXXXXXXXXXXXXXX				1-	51
50	-	XXXXXX	XXX	X	XXXXXXXXXXXXXXXXXX				1-	50
49	-	XXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX				1-	49
48	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX				1-	48
47	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX				1-	47
46	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX				1-	46
45	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX			X	1-	45
44	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX			X	1-	44
43	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX			X	1-	43
42	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		X XXX		1-	42
41	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		X XXX		1-	41
40	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		X X XXX		1-	40
39	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	39
38	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	38
37	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	37
36	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	36
35	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	35
34	-	XXXXXXXXXX	XXX	XX	XXXXXXXXXXXXXXXXXX		XXX XXX		1-	34
33	-	XXXXXXXXXX	XXX	XXX	XXXXXXXXXXXXXXXXXX		XXXX XXXX		1-	33
32	-	XXXXXXXXXX	XXX	XXX	XXXXXXXXXXXXXXXXXX		XXXX XXXX		1-	32
31	-	XXXXXXXXXX	XXX	XXX	XXXXXXXXXXXXXXXXXX		XXXX XXXX		1-	31
30	-	XXXXXXXXXX	XXX	XXX	XXXXXXXXXXXXXXXXXX		XXXX XXXX		1-	30

TOTAL TIME = 23 HRS 27 MIN 16 SEC

-END OF REPORT-

Figure E.7 : Averaged Percent of Prematurity.

APPENDIX F

SUMMARY OF ALGORITHMS

The detection methods and thresholds of the various waves are listed in TABLE F.1.

WAVE TYPE	DETECTION PARAMETER(S)	DETECTION THRESHOLD
QRS	First difference taken at 20 milli-second intervals.	> 50% of the average maximum slope
Premature beat	RR interval	< -12% of average RR
Late premature, "missing" beat	RR interval	> 150% of average RR
Noise spike	Negative slope, "RR"	negative slope < 150% of average maximum slope rate > 250 beats per minute

Table F.1 : Detection Methods and Thresholds.

TABLE F.2 shows the updating methods and conditions for the two adaptive parameters.

PARAMETER	UPDATING METHOD AND CONDITION
Average RR	$RR_{avg}(n) = 0.25 * RR(n) + 0.75 * RR_{avg}(n-1)$, using "normal" beats only.
"forced" update	$RR_{avg}(n) = 0.5 * RR(n) + 0.5 * RR_{avg}(n-1)$
Maximum negative slope	same as average RR

TABLE F.2 : Adaptive Threshold Updating Methods and Conditions.

VE/SVE CLASSIFICATION LOGIC

The premature beats are classified according to the results of the parameters measured. A beat is classified as Ventricular oriented if one of the following conditions, A or B, holds:

A) the QRS width is greater than 110 milli-second;

OR

B) two or more of the following :

- 1) the QRS is greater than 90 milli-seconds,
- 2) there is a compensatory pause,
- 3) R' or S' of the premature beat differ by more than 50% from that of the preceding beat.

RESCAN FOR MISSED BEAT

The interval where the suspected missed beat occurs is searched for the maximum (magnitude) negative slope. If the slope found is less than 20% of that of a normal QRS then the point is taken as a detection for the missed beat.

BASELINE

The baseline is estimated by averaging the points which are between -160 milliseconds to -450 milliseconds from the R peak.

QRS AMPLITUDE

The QRS amplitude is measured in two parts: the R amplitude and the S amplitude. The R amplitude is calculated from the R peak to the baseline and the S amplitude is calculated from the baseline to the bottom of the S wave.

QRS WIDTH

The QRS width is the difference between the QRS onset and end points. The QRS onset point is estimated by back tracking from the R peak to a point where the difference between two consecutive slopes is greater than 80% and the amplitude (from the baseline) is less than 25% of the R amplitude. The QRS end point is estimated from the R peak forward in the same manner except that the slopes are taken as absolute values so that the rising edge of the S wave will be included.

COUPLING INTERVAL

Coupling interval is the time interval between the premature beat and the preceding beat.

COMPENSATORY PAUSE

A compensatory pause is said to exist if the combined length of the coupling interval and the cycle following the ectopic beat

is greater than 188% (200-12%) of 2 normal R-R intervals.

BIGEMINY

Bigeminy is detected by comparing the ratio of the time interval between ectopic beats and the current R-R interval. If they occur in a 2 to 1 ratio, i.e. a premature beat occurs every 2 intervals, and this rhythm persists for more than 8 (ectopic) beats a Bigeminy message is printed.

TRIGEMINY

Trigeminy is detected in the same manner as Bigeminy except that the intervals are in a 3 to 1 ratio.

VENTRICULAR TACHYCARDIA

Ventricular Tachycardia is detected by a continuous sequence of Ventricular Ectopic beats. If more than 4 beats occur in a sequence a message is printed.

SINUS ARRHYTHMIA

Sinus Arrhythmia is detected by a continuous sequence of 3 or more Supraventricular Ectopic beats. When such a rhythm is detected a message is printed and the preceding 2 SVE's are removed from the array of SVE events.

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