IMPEDANCE CARDIOGRAPHY

A Dissertation
submitted in partial fulfillment of the
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Master of Technology

by

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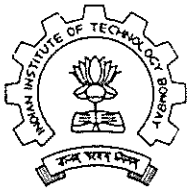

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DISSERTATION APPROVAL SHEET

Dissertation entitled **"Impedance Cardiography"** by **Pravin K. Survase** is approved for the award of the degree of Master of Technology in Electrical Engineering.

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ABSTRACT

In impedance cardiography, the cardiac output is obtained from stroke volume which in turn is obtained by monitoring the changes in the impedance of the thoracic region due to the flow of blood. Various attempts were made since 1932 to develop a system which is reliable, portable and gives satisfactory results while measuring cardiac output. Several methods have been proposed for estimating the stroke volume from impedance waveform, using different models. In one such system under development at IIT Bombay, the instrument generates 100 kHz low intensity sinusoidal current which is injected into thoracic region and the voltage resulting from modulation of the current waveform because of the variation in thoracic impedance is amplified and demodulated to get impedance waveform z(t). The waveforms z(t), its derivative dz/dt, ecg waveform e(t), and its derivative d(ecg)/dt, are fed through signal isolator to the ADC card installed in PC. The cardiac output is calculated by ensemble averaging of dz/dt.

In this project, a thoracic impedance simulator is designed and developed for testing and calibration of instrument. This simulator can be used for introducing a variation of 0.2% to 1% on a base impedance of 22 ohms. To make the existing system portable, instrument is interfaced to 8-channel, 8-bit ADC, data logger. Software supporting data logger interface and to calculate cardiac output is developed. Finally experiments were carried out on six subjects under rest and four subjects exercising on exercise bicycle. The results showed that at rest base impedance and cardiac output are almost constant. A large variation in stroke volume was observed in all subjects both under rest and while exercising.

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LIST OF ABBREVIATIONS

Abbreviation	Description
ADC	Analog to Digital Converter
CO	Cardiac Output in liters/min
DAS	Data Acquisition System
DMA	Data Memory Address
ECG	Electrocardiogram
HR	Heart Rate in beats/min
LVET	Left Ventricular Ejection Time in sec
PC	Personal Computer
SV	Stroke Volume in ml

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LIST OF SYMBOLS

Symbols	Description
ρ	Blood resistivity in ohm-cm.
Z	Base impedance in ohms.
L	Distance between the voltage electrodes in cm.
z(t)	Impedance waveform at the output of impedance cardiograph.
dz/dt	Derivative of z(t) waveform.
e(t)	ECG waveform at the output of impedance cardiograph.
de/dt	Derivative of e(t) waveform.
$(dz/dt)_{min}$	Most negative deflection on dz/dt waveform.

CONTENTS

CHAPTER 1

INTRODUCTION

1.1 Problem Overview

Impedance cardiography measures the blood output of heart using changes in the electrical impedance of the thoracic region. In this method an electrical current is injected in the thoracic region and voltage changes due to modulation of current by variation in the thoracic impedance is measured. This voltage is processed to obtain derivative of impedance, dz/dt, and there after cardiac output [1].

Impedance cardiography is a non-invasive method for continuous monitoring of cardiac output and is simple to work with. However, there are some drawbacks in this method. The thoracic impedance levels tend to change due to breathing. Further the impedance waveform obtained from patients with cardiac malfunction such as heart failure, mitral and aortic valve disorder may get distorted causing inaccurate results. The impedance change in the thorax is of the order of 1% to 2%, therefore the electronic circuit used for measurement must be accurate and sensitive to such small changes. Also the constraints on frequency and current intensity to avoid any irritation or physiological damage to the patient, should be considered.

1.2 Project Objective

In an on going effort to monitor cardiac output at IIT Bombay, an instrument was earlier developed by Joshi [2]. This instrument generates a current of 100 kHz frequency and 5 mA rms magnitude which is injected into body through exciting electrodes. The sensing electrodes sense the voltage due to change in thoracic impedance. On processing the voltage sensed instrument provides the impedance waveform z(t) and its derivative dz/dt, and ECG waveform e(t) and its derivative de/dt. The instrument is interfaced through an analog optoisolator to PC for calculating the stroke volume and cardiac output using an appropriate model and analysis algorithm. The waveforms z(t), dz/dt, e(t). and de/dt are displayed on the PC monitor along with the estimated heart rate, stroke volume, and cardiac output.

The impedance cardiograph developed was studied, tested, and experiments were carried out on subjects. While testing the instrument a need of simulator was felt. Hence simulator has to be developed which simulates the variation in thoracic impedance. The thoracic impedance simulator should also simulate ECG waveform to check the software written for calculating cardiac output and to test the ECG circuit in instrument. Thus thoracic impedance simulator should have ECG simulator as one of its building blocks. The simulator after development has to be calibrated. Presently the

instrument developed to monitor the cardiac output is interfaced to mains operated PC. To make the system portable, the instrument has to be interfaced to battery operated data recording unit. One such data recording unit is a "SITE" data logger which is battery operated. The software for interfacing the instrument to data logger has to be developed. Finally the experiments have to be carried out on subjects under rest and exercising on exercise bicycle.

1.3 Outline of the Dissertation

Chapter 2 presents an overview of the principles of impedance cardiography. Chapter 3 gives the details of the impedance cardiograph developed earlier at IIT Bombay. Chapter 4 focuses on the design and development of thoracic impedance simulator and testing of simulator. Chapter 5 deals with the software developed for data logger interface to instrument. Chapter 6 gives results of calibration of simulator and the results obtained from experiments carried on subjects at rest and exercising on exercise bicycle. Chapter 7 gives the summary and suggestions for further work. Appendix A is provided to give the circuit design of the thoracic impedance simulator. Appendix B gives the information of setting of ADC PCL208 used to capture the waveforms from instrument. Appendix C gives the details of "SITE" data logger used for recording signal waveforms. Appendix D gives the PCB layouts of thoracic impedance simulator circuits.

CHAPTER 2

PRINCIPLES OF IMPEDANCE CARDIOGRAPHY

2.1 Introduction

In early years of this century the scientists were curious to know about heart function and the various events taking place in the heart as it pumps blood. They were in the search of the system or unit by which they can study the function of heart and sort out the reasons for heart diseases. Scientists tried out various methods and performed the experiments on animals to find out electrical signal conveying the information of heart's function. Empirical formulae were obtained, various methods were implemented to support these formulae. This chapter deals with the origin of impedance cardiography, principle of impedance cardiography, various methods used for measuring the cardiac output, and the significance of the electrical signal to describe the events in heart. Before getting started with the details of impedance cardiography let us have a glance at a human heart.

2.2 Human Heart

The Fig. 2.1 illustrates the physical structure of heart and also the course of blood flow through heart. Heart has four chambers or pumps: two primer pumps, the atria and two power pumps, the ventricles. The period from end of one heart beat contraction to the end of next is called cardiac cycle. Each cycle is initiated by spontaneous generation of an action potential in the S-A node. The node is located in the anterior wall of right atrium, near the opening of superior vena cava and the action potential travels rapidly through both atria and then through A-V bundle into ventricles. Because of special arrangement of conducting system from atria into ventricles, the atria contract ahead of ventricles. This pumps blood into ventricles prior to the very strong ventricular contraction. Thus the atria acts as primer pumps for the ventricles and the ventricles then provide major source of power for moving blood through the vascular system [3].

Systole and Diastole

The cardiac cycle consists of period of relaxation called *diastole* followed by a period of contraction called *systole* cycle. The Fig. 2.2 illustrates the different events during cardiac cycle. The top three curves shows the pressure changes in aorta, the left ventricle and left atrium respectively. The fourth curve depicts the changes in the ventricular volume, the fifth curve is the electrocardiogram and sixth a phonocardiogram which is the recording of sounds produced by heart as it pumps. The Fig. 2.3 shows electrocardiogram with P, Q, R, S, and T wave. These are electrical voltages generated by heart and recorded by electrocardiography from the surface of body. The P wave is caused by spread of depolarization through atria

and this is followed by atrial contraction which causes the slight rise in atrial pressure curve immediately after P wave. Approximately 0.16 sec after onset of P wave, the QRS wave appears as a results of depolarization of the ventricles which initiates contraction of the ventricles and causes the ventricular pressure to begin rising. Therefore QRS complex begins slightly before the onset of ventricular systole. The T wave at which the ventricular muscle fibers begin to relax. Therefore T wave occur slightly prior to the end of ventricular contraction [3]. T wave occurs in the ventricle muscle 0.25 sec to 0.30 sec after depolarization and wave is repolarization wave. Thus electrocardiogram is composed of both depolarization and repolarization wave. Fig. 2.3 shows the normal electrocardiogram. The period of occurrence of PQRST wave is 0.41 sec to 0.51 sec.

Work output of heart

The work performed by left ventricles to raise the pressure of blood during each heart beat is called left ventricles stroke work output, and is equal to stroke volume output × (left ventricles mean ejection pressure). Likewise the work performed by the right ventricle to raise pressure of blood is called right ventricles stroke work output. Right ventricles stroke work output is usually about $1/6^{th}$ the work output of the left ventricles stroke.

For a normal person at rest, heart pumps 4-6 liters of blood per minute. However during severe exercise it may be required to pump as much as four times the value at rest.

2.3 Origin of Impedance Cardiography

The biological tissues show variation in the values of their electrical resistivity. For instance, the electrical resistivity of urine, plasma, blood, cardiac muscle, lung, tissue, fat, and bone are 30, 63, 150, 750, 1275, 2500, and 1600 ohm-cm respectively [4]. Though these values are much higher as compared to that of good conductor of electricity like copper (1.724×10^{-6} ohm-cm), the wide variation in the values of resistivity from one biological tissue to another makes the measurement of electrical resistance useful in understanding the function of internal organs of body.

In 1932 Atzler and Lehmann came out with the idea to record the thoracic impedance changes accompanying the cardiac activity in the human. Their technique consists of placing the thorax of the subject between two fixed plates of a capacitor, which was a part of the tuned circuit of an oscillator operating at approximately 50 MHz. With the subject holding his breath to prevent capacitive changes during the respiratory act, variation in the resonant frequency of the tuned circuit produced by the small changes in the effective dielectric of capacitor gave cardiac output. The voltage change accompanying the cardiac activity was sensed which on rectification and amplification was recorded with oscillograph. Atzler and Lehmann called their technique *dielectrographie*.

Kubicek et al. [5] investigated the impedance technique to measure cardiac stroke volume. Kubicek made the use of four electrodes, two current and two voltage as shown in Fig. 2.4. A 100 kHz 6 mA rms current is applied through electrode 1 and 4. The potential changes on the surface of the thorax produced by modulation in the current density distribution associated with cardiac activity are picked by electrode 2, and 3 and recorded as dz/dt. A typical record is shown in Fig. 2.5. The method used by Kubicek to calculate cardiac output and to measure peripheral volume from impedance is based on the familiar equation, $R = \rho L/A$.

Hill et al. [6] showed that the cardiac output measured using dye dilution method supported similar results as impedance method. In dye dilution method the cardiac output is obtained from so called dilution curve. They calculated the cardiac output for the different approximated methods namely,

i. Gamma function method
ii. Forward triangular method

The cardiac output was calculated using following formula

$$Q = \frac{m \times 60 \times \alpha}{C_t}$$

m = mass of injected dye in mg
C_t = maximum value of dilution curve in mg of dye per liter of blood
α = correlation factor

Baker [1] measured the cardiac output using dye and saline dilution technique supporting Kubicek's formula of stroke volume. They also measured stroke volume by implementing the Baker's method of neck and chest for measurement of impedance. It was concluded that mean stroke volume values from impedance method was 39% greater than mean stroke volume calculated by dye dilution method.

In 1989, Qu et al. [7] had proposed the system to measure thoracic impedance change and there after monitor the cardiac output. The system designed was supporting the Kubicek's formula of stroke volume. In this system voltage changes picked up from spot electrodes due to modulation of current by variation in thoracic impedance change is input to the instrument. The instrument processes the input and outputs base impedance Z, change in impedance dz/dt, ECG waveform e(t), and derivative of ECG, de/dt waveforms. Cardiac output is calculated by ensemble averaging of dz/dt. At IIT Bombay this system is developed with a few modifications and experiments are being carried out.

2.4 Fundamentals of Impedance Cardiography

Impedance cardiography is a study of cardiac function determined from measurement of the electrical impedance of the thorax. During systole the impedance decreases by 0.1 to 0.2 ohm. This decrease in impedance implies that the region have measured increase in blood volume [8]. The model used to quantify impedance changes is known as the parallel column model as shown in Fig. 2.6. This model assumes that their exists conducting material with an impedance Zo which is paralleled by a column with a uniform cross sectional area, length L and unknown resistivity ρ. If the cross sectional area of the parallel column varies from zero to a finite value and the impedance change, z(t), measured across the parallel column is small then the volume changes of column can be expressed as

$$\Delta v = \rho \frac{L^2}{Z_o^2} \Delta z$$

ρ = The resistivity of the material in the small parallel column
L = Length of the column
Z_o = The impedance of fixed conduction volume
Δz = Change in impedance across column

The formula for stroke volume assumes that the lungs were the source of impedance change. Just before systole the volume of small parallel column is zero. During systole the blood volume in the lungs increase thereby increasing cross-sectional area of the parallel column. To attempt to account for blood that leaves the lungs during systole, a forward slope extrapolation technique is used. A straight line is drawn parallel to the steepest part of the early impedance change. The early rise in the impedance pulse is due to only arterial inflow to the lungs. Thus drawing the slope by forward extrapolation can be replaced by first derivative of impedance change. The extrapolation of Δz is obtained by using the product of negative peak (minimum) of dZ/dt and the left ventricular ejection time [8].

Thus modified Kubicek's formula or extrapolated Kubicek's formula for stroke volume is

$$SV = \rho \frac{L^2}{Z^2} \left(\frac{dZ}{dt} \right)_{\min} T$$

dZ/dt = The most negative deflection during systole measured from zero.
T = The Left Ventricular Ejection Time
SV = Stroke Volume in ml

The cardiac output is given as

Cardiac Output = SV x HR liters/min ; HR = Heart Rate (beats/min)

The regions most likely responsible for the impedance change are atria, lungs, and aorta. The stroke volume equation assumes that the measured impedance is due to lung blood volume change [8].

2.5 Thoracic Impedance in the Time Domain

The first derivative of impedance recorded in Kubicek neck abdomen configuration gives the information about dynamics of heart. The Fig. 2.7 shows schematic representation of the first time derivative waveform and its relationship with cardiac cycle. The normal sinus rhythm of first time derivative dZ/dt is comprised of three distinct waves A wave, C wave, and O wave. The A wave corresponds to the atrial systole and can occur independently of C and O wave and is therefore absent before premature ventricle contraction. It appears 40-100 ms after the P wave of ECG. The B wave is synchronized with maximal deflection of the first heart sound at the apex of heart. It is synchronized with the opening of aortic valve. The C wave occur in very close in time to peak of ascending aortic blood flow. The C wave gives the required $(dZ/dt)_{min}$ in Kubicek's formula for calculation of cardiac output. X wave is synchronized with closure of aortic valve. The C wave is followed by a second major deflection called O wave. This wave begins at X and end near the base line. The Y wave represents the significant change in slope between O and X wave. The Y wave is synchronized with closure of the pulmonary valve. The O wave occurs during early diastole and is reportedly associated with mitral valve opening. The O wave shows significant increase with heart failure and some cases of coronary artery disease. Z wave shows the down word slope change of O wave. The Z wave is synchronized with the third heart sound representing the end of rapid filling phase [9].

Physiological correlation of impedance waveform

The recorded impedance changes are due to altered electrical resistivity of the blood as the blood cells are not spherically symmetrical and they align themselves in the direction of blood flow. The variation in the blood volume in a given body segment caused by blood flow is responsible to produce the changes in impedance [9].

The various factor that account for thoracic impedance change during cardiac cycle are

i. instantaneous change in shape, size and movement of heart.
ii. blood flow pattern in vena-cava and pulmonary veins.
iii. blood velocity related signals.

2.6 Shapes of Thoracic Impedance waveform

Fig. 2.8 shows a few ICG waveforms of subject under rest. The type-A waveforms are observed with subjects who are normal or healthy and waveform is physiologically normal waveform. Type-B is a distorted waveform of normal subjects above the age of 40 yrs. Type-C waveform is near normal waveform of normal subject. Subjects with the history of smoking, hypertension or diabetes but without any cardiac disorder exhibit type-D waveform [10].

In the type-B waveform, A wave is more negative and B wave is almost in the mid of up stroke of C wave. The estimation of stroke volume in this type of waveform becomes unreliable due to ambiguity in marking the opening of aortic valve [10].

CHAPTER 3

DEVELOPMENT OF IMPEDANCE CARDIOGRAPH

3.1 Introduction

As discussed in previous chapter various attempts were made to develop the unit to monitor cardiac output of subject under test. In order to monitor the cardiac output at IIT Bombay a instrument was developed and designed by Nagvenkar [11] in 1991. With this system one was able to monitor the variation in thoracic impedance. Then in 1993 Joshi [2] developed an other instrument with modifications in circuits. A DC cancellation circuit was designed by Joshi that completely eliminates the base impedance and leaves with varying impedance, dz, which is of more interest for calculating the cardiac output. A software developed by Joshi [2] calculates stroke volume and cardiac output. This software also gives the visual representation of base impedance Z and its derivative dz/dt, ECG waveform e(t) and its derivative de/dt along with stroke volume and cardiac output for the subject under test. Lakdawalla [12] tried to make the system portable. Lakdawalla has provided the isolator circuit with better isolation than that was provided by Joshi [2]. Lakdawalla has provided the visual representation of respiration events of subject undergoing the test. Due to high degree of isolation between instrument and PC, subject under test is protected from getting shocks, since the subject is directly in contact with battery operated instrument (impedance cardiograph) through electrodes.

3.2 Blocks in Impedance Cardiograph

Fig. 3.1 shows the overall block diagram of impedance cardiograph. The excitation circuit produces a 100 kHz low intensity (around 5 mA) sinusoidial current. The current is injected in the subjects thoracic region using current electrodes. The voltage electrode sense the voltage due to modulation of current by the variation in thoracic impedance. This modulated wave is amplified by instrumentation amplifier. The demodulator recovers the modulating impedance wave form z(t).The base impedance component of the impedance waveform is removed by DC cancellation circuit, hence Δz is obtained which represent only the variable component of the thoracic impedance due to the blood volume changes in the thorax. Impedance change due to breathing is removed by DC cancellation circuit. The dz waveform is differentiated to get the dz/dt wave form. ECG waveform e(t) is obtained by another instrumentation amplifier from the same voltage electrode pair the one used for z(t). The 100 kHz impedance measured voltage is removed by low pass filtering at the input of ECG circuit to obtain e(t). This waveform is differentiated to obtain its derivative de/dt. This de/dt is used for R wave detection, thence used for averaging the dz/dt waveform over the number of cycles to obtain the better estimation of the stroke volume. For visual indication of the breathing the dz/dt is

fed to a level detector circuit. During inhalation one led glows while other glows during exhalation.

The system explained above is battery operated. The battery operation of the circuit ensures the patients safety as the patient is directly connected to the circuit, more ever the built in opto-isolator keeps patients away from the mains operated PC. The instruments feeds the waveforms like z(t), dz/dt, e(t), and de/dt to the PC through opto-isolator for the calculation of stroke volume and cardiac output via data acquisition system.

3.3 Brief Explanation of Blocks in Impedance Cardiograph

Impedance Circuit

Excitation circuit

The excitation circuit consists of Wein Bridge oscillator and voltage to current converter. The oscillator is built to provide a stabilized output of 100 kHz with 4.5 V peak to peak voltage.

Sensing electrodes

Current output of V-I converter is fed to current electrodes. The voltage electrodes sense the modulated voltage and this acts as input to instrumentation amplifiers.

Instrumentation amplifier

The incoming signals from voltage electrodes are very small in strength so it is processed through instrumentation amplifier. Prior to that the incoming signal is high pass filtered with cut off frequency 7.8 kHz, this ensures the removal of 50 Hz pick up. The gain of amplifier is set to 16.4.

Demodulator

The demodulator circuit is a precision full wave rectifier, followed by a low pass filter to obtain the desired impedance waveform generated by thorax.

DC cancellation

The impedance waveform obtained after detector circuit has a large component of base impedance Zo and small component of, dz, variable thoracic impedance. It is this small component, is of interest, used to obtain dz/dt. Amplification of impedance signal, Zo, will give rise to saturation and subsequent information loss. This constant part, Zo, is eliminated by automatic DC cancellation circuit. The DC cancellation

circuit involves dual slope ramp generator and is an analog adaptation of the circuit given by Qu et al. [7] using successive approximation register.

Differentiator

The, Δz, small component of variable thoracic impedance is fed to differentiator to get dz/dt signal. The differentiator has band limited operation of 50 Hz in order to remove excessive high frequency noise amplification.

ECG Circuit

In order to get derivative of ECG waveform, e(t), de/dt, the voltage sensed by voltage electrodes acts as input to this circuit, hence an instrumentation amplifier is built at the input with low pass filter prior to it for the removal of high frequency. To enhance ECG waveform before differentiator, the constant dc part is filtered out by high pass filter with cut off 0.5 Hz. For a typical impedance waveform, the fundamental frequency is 1.2 Hz corresponding to 72 beats/min.

Signal opto-isolator

The waveforms like z(t), dz/dt, e(t), and de/dt are usually monitored on mains operated device like PC and oscilloscope. To avoid the possibility of electrical shock to the subject, as subject is well coupled to circuit, and to provide the galvanic isolation between battery driven circuit and mains operated equipment opto-isolators are used.

3.4 Electrodes Used in Measurements

Normally band electrodes are used for the measurement of cardiac output by the impedance technique. The band electrodes provide consistent waveform shape. However there are several problems in using them

i. They are inconvenient in clinical practice especially for long term monitoring. They can produce choking sensation in many patients.

ii. It is very difficult to place these electrode on patients with chest burns, chest tubing or for patient recovering from cardiac surgery as dressing and tube interfere with the placement of electrode.

To overcome these problems, spot electrodes are used as exciting and sensing electrodes in impedance cardiography. The spot electrodes give only local information i.e. near the region of placement. The spot electrodes furnish the signal with better SNR than band electrodes for exercising subjects Qu et al. [7].

3.5 Placement of Electrodes

The spot electrodes give, only local information, hence the placement of spot electrode is an important criteria in order to get the required information correctly. The impedance signal measured between two electrodes is voltage, so any body movement will change the current distribution and thereby the measured voltage. Thus impedance change caused by respiration and body movement is unwanted background signal, which must be suppressed to get the desired one. Qu et al. [7] has proposed the arrangement for the placement of spot electrode to extract the desired signal. In the proposed arrangement exciting and sensing electrodes are placed on the opposite side of body. The sensing electrode are placed far from exciting ones. The electrodes are placed in the central area of body where body movement cause minimal displacement of electrodes. This minimizes the noise introduced by body movement. Fig. 3.3 shows the appropriate arrangement of the spot electrodes.

The spot electrodes used are of suction type. The spot ECG/chest electrode consists of nickel plated steel rim attached to a rubber bulb as shown in Fig. 3.4. Conduction gel is applied on electrodes to make a good contact with the body.

3.6 Software

The opto-isolated analog outputs from the instrument are connected to the ADC channels of PCL-208 card through connector. The data is sampled with specific sampling rate, typically, 200 samples/sec/channel. The program "ICGREAL.C" inputs the data from A/D channel, samples it with adequate rate. Then from this digitized data stroke volume and cardiac output is calculated after successful detection of QRS complex in de/dt signal from which dz/dt is ensemble averaged. Then LVET (Left Ventricle Ejection Time) and $(dz/dt)_{min}$ are found. The base impedance is obtained by averaging dz/dt. The analysis programs "ICGREAL.C", "ICGOFFL.C" [2] are written in C language.

Brief Description of program "ICGREAL.C"

The inputs to the program are blood resistivity ρ, distance between the voltage electrodes L and gain of amplifier. The default values of each parameter are provided. Since PCL-208 card is used for ADC, the function '0' of PCL-208 is initialized to set base address level, interrupt level and DMA level. The function '1' sets multiplexer scan range of A/D channel. In order to carry out ADC the trigger rate is specified using function '17'. Trigger rate is calculated as

Pacer trigger rate = Input clock rate × (c2×c1)
Input clock rate = 1 MHz
c2 × c1 = 10000/SR SR = Sampling Rate
Using function '4' the A/D conversion of the input signal is performed. The resulting digitized data is stored in an array.

Calculation of SV and Cardiac output

The de/dt waveform is used to count the number of cycles in the acquired data. The down transition of de/dt corresponds to QRS complex of ECG waveform. The down transition is detected by setting a threshold, such that it eliminates the spurious down transition. The points corresponding to the transition are stored in an array. Using this time points of threshold crossing of the de/dt waveform, the dz/dt waveform is averaged. Since the $(dz/dt)_{min}$ and LVET occur only in first half of the waveform [7] the average of only $5/8^{th}$ of cycle is taken to save the computation time. To overcome the baseline problem while computing $(dz/dt)_{min}$ peak and valley of (dz/dt) waveform are compared and the difference is multiplied by 0.78 to get $(dz/dt)_{min}$ Qu et al. [7]. LVET is obtained by measuring the time when dz/dt crosses $1/8^{th}$ of the value of $(dz/dt)_{min}$. The heart rate is calculated using the average time between downward threshold crossing of de/dt waveform. Using Kubicek's formula the stroke volume is computed and cardiac output in liters/minute is found. All this data is stored in the *.BIN files.

The graphical representation of the events can be visualized by executing the "ICGOFFL.C".

3.7 Development Required

The system developed and discussed above works satisfactorily. Before resuming the test or experiments one has to test the instrument. Simulator is needed to test the instrument, to avoid dependence of instrument testing on subject. Thus thoracic impedance simulator has to be designed which will simulate the variation in thoracic impedance. Presently instrument is interfaced to PC, hence to make system totally battery operated, as the instrument is battery operated, instrument has to be interfaced to battery operated data logger. A software for interface of data logger to instrument has to be written. After doing all above things experiments have to be carried out on subjects under rest and exercising on exercise bicycle.

CHAPTER 4

IMPEDANCE SIMULATOR

4.1 Introduction

Impedance cardiograph developed at IIT Bombay to monitor the cardiac output was discussed in the previous chapter along with development required to make present system portable. This chapter deals with the block diagram of thoracic impedance simulator. The design of each block is discussed separately. Testing of thoracic impedance simulator is also discussed.

4.2 Impedance Simulator

As was discussed in previous chapter a need of thoracic impedance simulator was felt to test the instrument. The simulator should have two input terminals for injecting current and two output terminals representing the voltage due modulation of current by variation in impedance. Further a simulated ECG waveform should also be available across these terminals. The impedance simulator is analogous to the parallel column model as shown in Fig. 4.1.

In the two cylinder model, thorax impedance consists of a fixed resistance R_o paralleled by time varying resistance R_t due to blood volume change in the thorax region. The simulation of variation in thoracic impedance is similar to two cylinder model and is obtained by fixed resistance R1 in parallel with a some fixed resistance R2, and varying resistance R3, which is switched by analog switch. The switching control to the analog switch simulates opening and closing of the ventricular values. The change in resistance obtained from simulator is a switching type and not a gradual one. The electrode-tissue contact resistances are simulated by appropriate resistances. The ECG waveform in the impedance is introduced by connecting the output of ECG simulator through R7, C1, and R8, C2 as shown in Fig. 4.1.b. The complete simulator block diagram is shown in Fig. 4.2. It consists of blocks like cardiac beat generator, ECG waveform generator, and thoracic impedance simulator

4.3 Cardiac Beat Generator

The cardiac beat generator essentially is a square wave generator as shown in Fig. 4.3. The frequency of square wave is decided by resistor R1 and capacitor C1 along with the feed back factor β that is fed to non-inverting terminal of op-amp. The output swing of square wave is $\pm V_{sat}$ of op amp. The time period of square wave is given by the following expression [13].

$$T = R_1 C_1 \log_e \left(\frac{1+\beta}{1-\beta} \right)$$

where β = R3 / (R3+R2)

The output frequency of square wave can be varied by varying the potentiometer R4 which is in series with R1. If R2 = R3 then the time period of square wave is given by following expression,

$$T = 2.2\ R_1 C_1$$

The frequency of square is given as

$$f = 1/T = 0.76\ RC$$

and frequency can be varied from 0.7 Hz to 2 Hz by varying R4, which gives cardiac beat of 42 beats/min to 120 beats/min.

4.4 ECG Waveform Simulator

To test ECG circuit in instrument (impedance cardiograph) and to check the software written to calculate cardiac output, an ECG waveform simulator is needed. Fig. 4.4 shows the blocks in ECG waveform simulator, and it simulates P, Q, R, S, and T waves representing the different electrical events taking place in heart during cardiac cycle. The input to simulator is output of cardiac beat generator, V_E. The square wave output of cardiac beat generator triggers monostable multivibrator, M1, which in turn triggers M2, and so on till M6. Each of these monostables generate a pulse of time duration equal to P, Q, R, S, and T wave segments respectively. The P, Q, R, S, and T waves are simulated by passing the output pulse of monostable multivibrator through integrator or differentiator as required.

Monostable multivibrator (M) is an important block of ECG simulator. It is this block that decides the time period of different waves in ECG waveform, e.g., P wave occurs only with in time decided by pulse width of M1. All monostable multivibrators use op-amp to generate required pulse width. The essential circuit schematic of monostable is shown in Fig. 4.5. The output pulse width depends on initial voltage across capacitor, value of R1, C1, and feed back factor β. The general formula for pulse width is given as [13]

$$T = R_1 C_1 \log_e \left(\frac{1+\left(V_1 / V_o\right)}{1-\beta} \right)$$

where, $V_o = V_{sat}$ of op amp before triggering mono shot
β = R3/(R3+R2)
V_1 = initial voltage across capacitor

4.5 Generation of ECG waveform

Here we will describe the generation of each wave followed by the adder circuit. The ECG waveform is obtained essentially by generating a pulse corresponding to each of the P, Q, R, S, and T segments, and by appropriate wave shaping of the pulses, and finally adding these as shown in Fig. 4.4. The Fig. 4.6 shows the timing diagram for generation of ECG waveform.

P wave

The elements that constitute P wave are M1 and integrator 1. Fig. 4.7 shows the circuit diagram for generation of P wave. Monostable M1 generates positive pulse. Refer to appendix A for design values. R4 and C2 pair differentiates the trigger pulse (square wave) and the following diode allows only positive pulse to trigger M1. The $\pm V_{sat}$ pulse at the output of M1 is converted to 0 to 5 V by a zener of 5.1 V. R6 and C3 are components responsible for the generation of P wave.

Q wave

M2 along with differentiator 1 generates Q wave. The trigger pulse for M2 is the output pulse of M1. Resistor R3 and C1 decides pulse width of M2. Pulse width of M2 is 10 msec. M2 is negative edge triggered so the output pulse is negative. The negative output pulse of M2 is differentiated by R4, C2 combination and only negative edge of differentiated pulse is allowed by D2. Fig. 4.8 shows circuit diagram of Q wave generator.

R wave

R wave is simulated by M3 and integrator 2. The trigger input for M3 is the output pulse of M2. The diodes in M3 are so arranged that the monostable M3 is positive edge triggered, hence output is positive pulse. The pulse of 0 to 5 V obtained using zener of 5.1 V at the output of M3 is integrated by integrator (R6, C3). The back diode D3 in series with R7 and parallel to R6 allows rapid discharge of C3 . Fig. 4.9 shows circuit diagram of R wave generator.

S wave

S wave is simulated by M4, M5, and differentiator 2. M4 simulates the time between R and S waves. M5 is positive edge triggered by output pulse of M4. The output pulse width of M5 is 300 msec. M5 along with differentiator 2 simulates S wave. Positive output pulse of M5 is differentiated by C5, R10. Fig. 4.10 shows the circuit diagram for the generation of S wave. The output pulse width of M5 decides the time between S wave and T wave.

T wave

T wave is obtained from M6 in conjunction with integrator 3. Monostable M6 is negative edge triggered, hence output pulse of M6 is negative. This negative pulse is integrated by R7, C3. Appendix A gives design values of components used in this circuit. Fig. 4.11 shows circuit diagram for T wave generation.

Adder

Adder, adds all waves so that the ECG waveform, V_{ecg}, is obtained at the output of adder. The attenuation factor required for each wave to obtain ECG waveform is given below. R wave has amplification factor of 1 (attenuation factor is 1) as indicated by values of R2 and R5. All other waves are attenuated by suitable attenuation factor with respect to attenuation provided for R wave.

Wave	Attenuation factor	Resistance and its value
P	3.0	R3 = 68 k ohm
Q	1.5	R1 = 33 k ohm
S	2.1	R6 = 47 k ohm
T	3.0	R7 = 68 k ohm

The simulated ECG waveform at output of simulator is shown in Fig. 4.13. This waveform is recorded using data acquisition card and the program ADC.asm, Scale.exe and fed.exe developed by Prasad [14].

4.6 Thoracic Impedance Simulator

Fig. 4.14 shows the schematic of the impedance simulator. The variation in thoracic impedance is simulated by fixed resistor R1 parallel to another resistor R2 which is switched in and out of circuit by analog switch DG 201. The resistors R4, R5, R6, and R7 represents the resistances due to skin electrode interface. Two values of thoracic impedance are

$$R_{z1} \text{ (switch off)} = R1$$

$$R_{z2} \text{ (switch on)} = R1 \parallel (R2+R3)$$

The analog switch is controlled by square wave output from cardiac beat generator. It is normally set to 1 Hz (60 beats) and can be varied from 42 beats/min to 120 beats/min. The simulated ECG waveform is coupled to point x and y by control signal through series combination of R8 and C1, and R9 and C2. Due to this coupling

the simulated thoracic impedance wave and ECG wave modulate the current waveform of 100 kHz frequency, and 5 mA rms value. The output of simulator is voltage waveform which acts as input to instrument that provides the signals like z(t), dz/dt, e(t), and de/dt. The thoracic impedance simulator so designed has facility to vary the base impedance and also to vary the change in impedance. The base impedance can be decreased to 19 ohms by placing the resistor of 100 ohm across the fixed base impedance i.e. 22 ohms. A variable resistance R3 is placed in series with R2. By varying this resistance the variation in thoracic impedance can changed.

4.7 Testing of simulator

The current of 100 kHz frequency, 5 mA magnitude is injected at the excitation terminal and the output voltage is taken across the sensing terminal, due to the variation in impedance. This voltage serves as input to the instrument, which on processing outputs the waveform like z(t), dz/dt, e(t), and de/dt. This output signals are acquired using DAS (Data Acquisition Card). The acquired dz/dt, e(t), and de/dt waveforms are shown in Fig. 4.15, and Fig. 4.16. The dz/dt waveform shown in Fig. 4.15 is different from the dz/dt of normal subject. The reason for this difference is that switching of analog switch is very sharp, i.e. switching is done by square wave, hence the impedance change is switching type where as the change in impedance in thorax region of normal subject is a gradual change.

CHAPTER 5

INTERFACING TO DATA LOGGER

5.1 Introduction

In Chapter 3, it was discussed that to make system portable the instrument has to be interfaced to "SITE" data logger. The interfacing of instrument to data logger will make the system totally battery operated because data logger works on 9 V battery and instrument works on 18 V battery.

5.2 Software for Data Logger Interface with Instrument

Data-logger, a DMS product, is small and compact device for recording the data. Basically data logger has 8 channel for 8-bit analog to digital conversion. Refer appendix D for details. The opto-isolated analog outputs of the instrument are connected to four channels of data logger. z(t) waveform is connected to ch1. dz/dt to ch2, de/dt to ch3, and finally e(t) to ch4. The program to record the data is written in Site language and is stored as "ECG.LIT". In this program four channel of data logger are initialized to acquire the data at a rate of 200 Hz. The data of ch1, ch2, ch3, and ch4 is stored in variables V1, V2, V3, and V4 respectively. Data logger acquires data and stores in *.dat files. To have the view of recorded data one has to evoke "SITE SHOW" package under windows (ver 3.1) environment.

To calculate the SV and CO the data logged by data logger is fed to DLGICG.C program. Since the data stored in *.dat files is binary, and the format of storing recorded data is not known. Hence XSITE command is used to convert the binary files to text files (XSITE -f [filename.txt] -s 1 -m 1 data filename[*.dat]). The XSITE command is supported by data logger. The *.txt files obtained by using XSITE command contains the information of recorded data and the time at which data is recorded. To retrieve the recorded data the *.txt files are input to another program named DGLCON.C. This program is general purpose program for data retrievation that is recorded by data logger. The inputs to DLGCON.C program are *.txt file, total number of channels used for recording data, and number of channels used for recording data in ascending order. The output of DLGCON.C program is the data recorded by data logger and is stored in *.dat files. The *.dat files created by DLGCON.C program is different from *.dat file created by data logger. This program also stores the data of each channel separately in ch*.dat (* = 1-8) files, e.g. data recorded using ch1 is stored in ch1.dat, if one uses more than one channel. Hence the user can use the data recorded by each channel separately, if required.

Thus *.dat files obtained from program DLGCON.C contains the true data recorded using data logger. These files are input to the program DLGICG.C that

calculates SV and CO. The inputs to program DLGICG.C are resistivity, distance between electrodes, and gain of circuit. The DLGICG.C program accepts the data stored in file and stores it in an array for further processing.

Calculation Stroke Volume and Cardiac Output

The de/dt waveform is used to count the number of cycle in the acquired data. The down transition of de/dt corresponds to QRS complex of ECG waveform. The down transition is detected by setting a threshold, such that it eliminates the spurious down transition. The points corresponding to the transition are stored in array. Using this time points of threshold crossing of the de/dt waveform, the dZ/dt waveform is averaged. Since the $(dZ/dt)_{min}$ and LVET occur only in first half of the waveform [7] the average of only $5/8^{th}$ of cycle is taken to save the computation time. To overcome the baseline problem while computing $(dZ/dt)_{min}$ peak and valley of (dZ/dt) waveform are compared and the difference is multiplied by 0.78 to get $(dZ/dt)_{min}$ Qu et al. [7]. LVET is obtained by measuring the time when dZ/dt crosses $1/8^{th}$ of the value of $(dZ/dt)_{min}$. The heart rate is calculated using the average time between downward threshold crossing of de/dt waveform. Using Kubicek's formula the stroke volume is computed and cardiac output in liters/minute is found. All this data is stored in the *.bin files.

The graphical representation of the events can be visualized by executing the "DLGOFL.C".

CHAPTER 6

RESULTS

6.1 Introduction

As discussed in previous chapter the necessity of simulator was faced for testing instrument. T' :s a simulator was designed, implemented and tested. This chapter deals with calibration of simulator. Experiments were conducted on subjects under rest and exercising on exercise bicycle by placing the spot electrodes on body of subject in an arrangement proposed by Qu et al. [7].

6.2 Calibration of Simulator

The block diagram of thoracic impedance simulator was discussed in chapter 4. The current waveform of frequency 100 kHz, and magnitude 5 mA generated by instrument is input to the simulator. The output of simulator is a voltage waveform due to modulation of current by variation in simulated thoracic impedance. The thoracic base impedance is set to 22 ohms which is placed in parallel to 2.2 k ohms. The switching of 2.2 k ohms in and out of circuit gives the change in thoracic impedance. On the front panel of simulator a thoracic impedance control is provided. Table. 6.1 shows the values of change in thoracic impedance by varying thoracic impedance control. From table 6.1 it is clear that the percentage change in thoracic impedance actually varies from 0.20% to 1.00% and the measured percentage change in impedance is 0.16% to 0.91% for the base impedance of 22 ohms. The thoracic impedance simulator has a facility to change base impedance. Table. 6.2 gives the percentage change in thoracic impedance on base impedance of 20.6 ohms (22 ohms || 330 ohms). The actual percentage change in impedance is 0.18% to 0.97% and measured from instrument is 0.15% to 0.90%. Similarly the actual percentage change in impedance is 0.19% to 0.99% and measured is 0.15% to 0.92% for base impedance of 21.6 ohms (22 || 1200 ohms). Fig 6.1 shows the graph of actual percentage change in impedance verses measured percentage change in impedance. The graph is not linear because,

i. the variation in impedance is caused by switching of R2 in and out of circuit, by analog switch DG 201, paralleled to R1. Since the analog switch has some ON resistance, of the order of 100 ohms, which adds up to R2 while introducing the variation in impedance.

Fig. 6.2 shows the results of testing and calibration of simulator. The Fig. 6.2 shows z(t), dz/dt, e(t), and de/dt waveforms for percentage change in impedance of 1% and 0.2 % on base impedance of 22 ohms. The thoracic impedance simulator has another control, called beats control. The beats can be varied from 42 beats/min to 120 beats/min by varying this control.

6.3 Experimental results

Before going into the details of experimental results let us have a glance at experimental method.

Experimental method

The spot electrodes were used to extract the thoracic impedance change by placing them according to the arrangement proposed by Qu et al. [7]. The instrument excites the current of frequency 100 kHz, and magnitude 5 mA which is injected into the body of subject through current electrodes and voltage electrodes sense the voltage developed due to modulation caused by variation in thoracic impedance. The sensed voltage is input to the instrument which in turns outputs base impedance z(t) and its derivative dz/dt, and ECG waveform e(t) and its derivative de/dt waveforms. These waveforms are interfaced to data logger. Two types of measurement were taken

i. during resting condition
ii. during recovery after exercise on exercise bicycle

Results from subjects under rest

The subjects under rest were asked to stand without making unnecessary movements. Each observation was taken at a duration of 5 sec. All subjects were males in the age group of 23 to 26 years. Six subjects ABC, SVK, SPT, KGS, MSH, and CSK under went this test. Table 6.4 shows the results obtained from subjects under rest. The mean of heart rate over a number of observation shows that subject SPT has heart beat rate 68 beats/min were as other subjects have heart rate ranging from 81 to 85 beats/min. The heart rate of individuals were checked manually and it supported the obtained results. The base impedance was found to be distributed from 18 to 27 ohms among the subjects. The s.d. in base impedance is small for individual subject. The small deviation in base impedance in an individual indicates consistency of measurements. Relatively large variation across the subjects may be attributed to variation in the thoracic physiology across these subjects. A large s.d. in stroke volume is observed for a given subject. The reason for such a large variation in stroke volume is a wrong detection of LVET (Left Ventricular Ejection Time) [15]. The factors that affects the stroke volume magnitude are maximum blood flow velocity, rise and fall times of the aortic inflow [10]. Also a large variation in SV across the subjects is observed. Cardiac output for six subjects ranges from 1.97 to 4.18 liters/min. Only for subject KGS it appears that CO is near normal and for others it is low. The CO for a given subject may be low because of improper positioning of electrodes. Fig. 6.3 to Fig. 6.5 shows that the subjects under rest did not show any signs of heart disorder and all of them resemble a type A waveform of impedance [10].

Results from subjects exercising on bicycle

Subject was asked to exercise on an exercise bicycle vigorously till the heart rate roughly doubled. The subject was then asked to stop exercise and relax. The readings were taken at the interval of 5 sec. Four subjects, ABC, KGS, SPT, and MSH under went this test. Table 6.5 to Table 6.8 gives the measurement values. Fig. 6.8 and Fig. 6.9 shows the plots of HR, SV, CO with respect to time.

For subject ABC, it is observed from the Fig. 6.8 that initially when the heart rate is high the stroke volume and cardiac output has maximum value. After highest HR, both SV and CO start decreasing. There is a decrease in stoke volume at maximum HR, because the time available for pumping per stroke is less and hence the CO is also less. The subject is asked to relax, at a point marked "s/r" in plot. After this point a rise in SV and CO is observed because the oxygen requirement for a body remains high. As the HR decrease, first an increase and than a decrease in SV is observed. The same is observed in CO for certain period, but as heart rate comes to normal CO also decrease and attains the normal value. Even though there is large increase and decrease in SV, the CO tries to attain normal value (value at rest) because the CO also depends on HR.

Similar variation in HR, SV, and CO were observed for subjects SPT, MSH, and KGS. In case of subject SPT the increase in stroke volume is large when subject is relaxing, but for subject MSH the increase in stroke volume and CO is same (by same factor). This variation in increase in stroke volume and cardiac output after the exercise is stopped and subject is relaxing differ from subject to subject because of the physiology and the amount of oxygen consumed by each subject is different.

CHAPTER 7

SUMMARY, CONCLUSIONS AND SUGGESTIONS FOR FURTHER WORK

In impedance cardiography, the cardiac output is obtained from stroke volume which in turn is obtained by monitoring the changes in the thoracic impedance due to the flow of blood. Various attempts were made to develop the system since 1932 to monitor the cardiac output. Various models were designed to calculate the cardiac output. The model suggested by Kubicek was then used by many researchers and depending on this model various systems were developed. The model designed by Kubicek assumes that the impedance variation is due to lungs blood volume change. On the basis of Kubicek's model Qu et al. [7] had proposed the system to calculate the cardiac output. At IIT Bombay a system proposed by Qu et al. [7] is developed with a few modifications to monitor cardiac output.

The instrument developed at IIT Bombay to monitor the cardiac output excites the current of frequency 100 kHz, and magnitude 5 mA to inject into the thoracic region of subject. The voltage sensed due to modulation of current by variation in thoracic impedance is input to instrument, which in turn outputs the signals like z(t), dz/dt, e(t), and de/dt. These signals are interfaced to PC through optoisolator to provide the safety to subject under test as the subject is directly in contact with instrument. The electrodes used for exciting and sensing are spot electrodes. The placement of electrodes on body is as proposed by Qu et al. [7] to minimize the error caused in measuring cardiac output due to motion artifacts. The signals interfaced to PC are acquired by DAS and fed to program "ICGREAL.C" to calculate stroke volume and there after cardiac output.

While testing the instrument, a need of simulator was felt. Thus a thoracic impedance simulator was designed and developed. The thoracic impedance simulator has a building blocks cardiac beat generator, ECG waveform generator, and impedance simulator. The impedance simulator is analogous to the two cylinder model of thorax impedance. The change in impedance obtained from simulator is switching type and not a gradual one. The aim behind introducing ECG simulator in thoracic impedance simulator was to test the ECG circuit present in instrument and also to check the software written for calculating cardiac output, since cardiac output is calculated by ensemble averaging of dz/dt waveform after detection of QRS complex. The variation in thoracic impedance is obtained by switching a resistance of 2.2 k ohms in and out of circuit placed parallel to base resistance of 22 ohm. The switching caused by analog switch is analogous to opening and closing of ventricular valves. To make the system portable, instrument was interfaced to "SITE" data logger. "SITE" data logger is 8-channel ADC used for recording data. The program "ECG.LIT" is written in Site language to acquire the data from instrument. The signals z(t), dz/dt, e(t), and de/dt are connected to channel ch1, ch2, ch3, and ch4 respectively of data logger. The

sampling rate is fixed to 200 samples per sec per channel. Data logger on recording the data stores it in *.dat files in binary mode, to retrieve the recorded data for estimation of cardiac output "XSITE" command is used, which converts binary file to text file. These text files have information of recorded data along with the time at which it is recorded. Thus to retrieve the recorded data *.txt files are input to program DLGCON.C. The program DLGCON.C stores recorded data in *.dat files. Finally the *.dat files obtained from program DLGCON.C are input to program DLGICG.C which calculates stroke volume and cardiac output. Experiments were carried out on healthy subjects under rest and exercising on exercise bicycle.

The simulator so designed was calibrated and it showed that thoracic impedance change ranges from 0.2% to 1%, and 0.18% to 0.97% for base impedance of 22 ohms and 20.6 ohms respectively. The error in measuring the percentage change in impedance is because of addition of ON resistance (100 ohm) of switch while introducing the variation in impedance. Simulator also provides beat control ranging from 42 beats/min to 120 beats/min. Experiments carried on six subjects under rest showed that the HR of individual subject checked manually supported the obtained results. The large variation in stroke volume in all subjects was observed and this is because of wrong detection of LVET [15]. CO for the subjects other than KGS was less because of improper position of electrodes. The impedance waveform obtained from all subjects is similar to type A waveform [10]. Four subjects had under gone test of exercising on exercise bicycle and results showed that as heart rate decreases the stroke volume decreases, because time available for pumping per stroke is less, and hence CO also decreases. When the subject is relaxing a rise in SV and CO is observed because the oxygen requirement for a body remains high.

Suggestions for Further Work

A large variation in stroke volume is observed in all subjects at rest. To sort out the problem experiments have to be carried out on a subject under rest by changing the distance between voltage electrodes. Kubicek's formula for SV, derived from parallel column model, assumes that blood volume changes is due to the impedance change in thorax region. This impedance change neglects the 10%-20% change in the blood resistivity with flow and organ movement [16]. While carrying out the experiment, the blood resistivity of all subjects was assumed to be 145.2 ohm-cm. Hence experimentally measuring of blood resistivity of individual subject undergoing test have to be done because according to Kubicek's formula stroke volume is depended on blood resistivity. Experiments were done on the healthy subjects under rest and exercising on exercise bicycle, and the results obtained from experiments were satisfactory. Experiments have to be performed on the subjects with cardiac disorder under rest and exercising on bicycle.

Table 6.1 Results of calibration of simulator
Base impedance R_o = 22 ohm
R = Variable impedance, R_s = 2.2 Kohm
R_δ = $(R+R_s) \parallel R_o$
Actual % change in impedance = $(R_o - R_\delta)/R_o$

R K ohms	$R_t = R+R_s$ K ohms	Actual % change in impedance	Obtained % change in impedance
8.5	10.7	0.20	0.16
6.5	8.7	0.25	0.20
5.1	7.3	0.30	0.24
4.2	6.4	0.34	0.30
2	4.2	0.52	0.48
1	3.2	0.60	0.54
0	2.2	1.00	0.91

Table 6.2 Results of calibration of simulator
Base impedance R_o = 22 ohm $\parallel$ 330ohm = 20.6 ohms
R = Variable impedance, R_s = 2.2 Kohm
R_δ = $(R+R_s) \parallel R_o$
Actual % change in impedance = $(R_o - R_\delta)/R_o$

R K ohms	$R_t = R+R_s$ K ohms	Actual % change in impedance	Obtained % change in impedance
8.5	10.7	0.18	0.15
6.5	8.7	0.23	0.19
5.1	7.3	0.28	0.22
4.2	6.4	0.31	0.25
2	4.2	0.48	0.43
1	3.2	0.63	0.58
0	2.2	0.97	0.90

Table 6.3 Results of calibration of simulator

Base impedance R_o = 22 ohm || 1200 ohm = 21.6 ohms

R = Variable impedance, R_s = 2.2 K ohm

R_δ = $(R+R_s)$ || R_o

Actual % change in impedance = $(R_o - R_\delta)/R_o$

R Kohms	$R_t = R+R_s$ Kohms	Actual % change in impedance	Obtained % change in impedance
8.5	10.7	0.19	0.15
6.5	8.7	0.22	0.19
5.1	7.3	0.27	0.22
4.2	6.4	0.33	0.29
2	4.2	0.51	0.46
1	3.2	0.66	0.59
0	2.2	0.99	0.92

Table 6. 4 Standard deviation and mean of results obtained from subjects under rest

Subject	Zo	HR	SV	CO
	ohms	beats/min	ml	liters/min
	Mean (s.d)	Mean (s.d)	Mean (s.d)	Mean (s.d)
ABC	25.8 (0.07)	81 (2.72)	24.9 (4.01)	1.97 (0.28)
SVK	25.5 (0.24)	87 (4.09)	23.7 (6.49)	1.99 (0.54)
SPT	26.8 (0.34)	68 (4.36)	25.6 (3.08)	1.76 (0.25)
KGS	17.7 (0.26)	82 (4.60)	50.1 (12.75)	4.18 (1.17)
MSH	19.0 (0.64)	86 (3.34)	38.4 (8.10)	3.30 (0.66)
CSK	23.3 (0.85)	88 (4.34)	26.0 (4.06)	2.40 (0.40)

Table 6.5 Test results from subject under exercise on exercising bicycle. The list of results obtained were taken at interval of 5 sec.
Zo = Impedance, HR = Heart Rate. SV = Stroke volume,
CO = Cardiac Output
ρ = 142.5 ohm-cm. L = 15 cm

Subject	Test no.	Zo	HR	SV	CO
		ohms	beats/min	ml	Liters/min
ABC	1	24.7	98	27.0	2.76
	2	25.7	121	24.4	2.47
	3	25.1	120	18.6	2.23
	4	25.6	117	19.2	2.25
	5	25.4	117	19.7	2.30
	6	25.3	116	7.1	0.83
	7	26.0	114	17.5	1.99
	8	25.5	113	24.6	2.78
	9	25.1	105	20.8	2.18
	10	25.4	102	27.3	2.79
	11	25.1	101	22.3	2.27
	12	26.0	99	21.9	2.17
	13	26.0	90	24.4	1.91
	14	26.2	89	21.2	2.20
	15	25.8	86	25.4	2.42
	16	26.6	85	28.2	1.88
	17	26.4	84	22.3	1.81
	18	26.1	80	22.6	1.52
	19	25.7	78	19.5	1.60
	20	25.8	68	23.3	1.22
	21	25.5	78	15.6	1.35

Table 6.6 Test results from subject under exercise on exercising bicycle. The results were taken at the interval of 5 sec.
Zo = Impedance, HR = Heart Rate, SV = Stroke volume,
CO = Cardiac Ouptut
ρ = 142.5 ohm-cm L = 15 cm

Subject	Test no.	Zo	HR	SV	CO
		Ohms	Beats/min	ml	Liters/min
SPT	1	25.9	127	25.0	3.19
	2	26.0	124	27.2	3.39
	3	25.8	122	31.2	3.79
	4	25.8	119	22.4	2.68
	5	25.4	118	14.0	1.64
	6	25.5	116	12.7	1.47
	7	25.3	101	21.8	2.20
	8	25.3	97	52.4	5.10
	9	25.5	96	43.4	4.18
	10	25.4	91	59.4	5.42
	11	25.3	89	43.7	3.95
	12	26.7	85	51.3	4.39
	13	26.6	85	42.3	3.87
	14	26.2	79	33.8	3.34
	15	26.5	78	26.2	3.31
	16	26.8	78	30.8	2.64
	17	26.3	76	24.9	2.02
	18	27.0	73	39.0	2.25
	19	26.7	70	25.3	1.75
	20	26.4	69	24.8	1.69
	21	26.5	63	25.6	1.61

Table 6.7 Test results from subject under exeercise on exercising bicycle. The results were taken at a interval of 5 sec.
Zo = Impedance, HR = Heart Rate, SV = Stroke Volume
CO = Cardiac Output
ρ = 142.5 ohm-cm. L = 15 cm

Subject	Test no	Zo	HR	SV	CO
		ohms	beats/min	ml	liters/min
MSH	1	19.6	125	38.7	5.21
	2	19.7	123	40.9	5.44
	3	20.0	120	40.8	5.31
	4	20.3	124	36.8	4.55
	5	20.1	117	21.3	2.49
	6	20.1	116	31.1	3.60
	7	20.2	113	44.9	5.08
	8	20.2	112	29.2	3.34
	9	20.1	112	37.2	4.17
	10	20.1	111	29.9	3.34
	11	20.2	109	43.3	4.65
	12	20.1	106	28.5	3.03
	13	20.1	104	31.2	3.25
	14	20.0	103	32.4	3.34
	15	20.0	102	29.4	3.01
	16	19.9	99	29.2	2.91
	17	19.7	99	32.3	3.20
	18	19.6	96	29.1	2.89
	19	19.7	95	29.9	2.87
	20	19.6	94	32.1	2.64
	21	19.7	94	31.4	2.97

Table 6.8 Test results from subject under exercising on bicycle. The results were taken at a inteval of 5 sec.
Zo = Impedance, HR = Heart Rate, SV = Stroke Volume,
CO = Cardiac Output
ρ = 142.5 ohm-cm. L = 15 cm

Subject	Testno.	Zo	HR	SV	CO
		ohms	beats/min	ml	liters/min
KGS	1	16.9	98	59.3	5.86
	2	18.0	121	41.9	5.04
	3	17.6	117	40.8	4.77
	4	17.7	114	40.8	4.67
	5	17.0	113	50.5	5.72
	6	17.3	105	43.6	4.56
	7	17.3	103	74.4	7.83
	8	17.7	102	56.6	5.77
	9	17.3	101	46.7	4.74
	10	18.2	99	44.5	4.41
	11	16.9	98	59.3	5.86
	12	18.1	89	49.7	4.42
	13	18.0	86	51.9	4.49
	14	18.8	85	58.3	4.96
	15	18.3	84	48.7	4.13
	16	18.4	84	59.6	5.01
	17	18.7	83	45.7	3.80
	18	18.4	83	45.6	3.80
	19	18.4	82	47.8	3.93
	20	18.3	80	45.8	3.66
	21	18.5	80	31.0	2.50

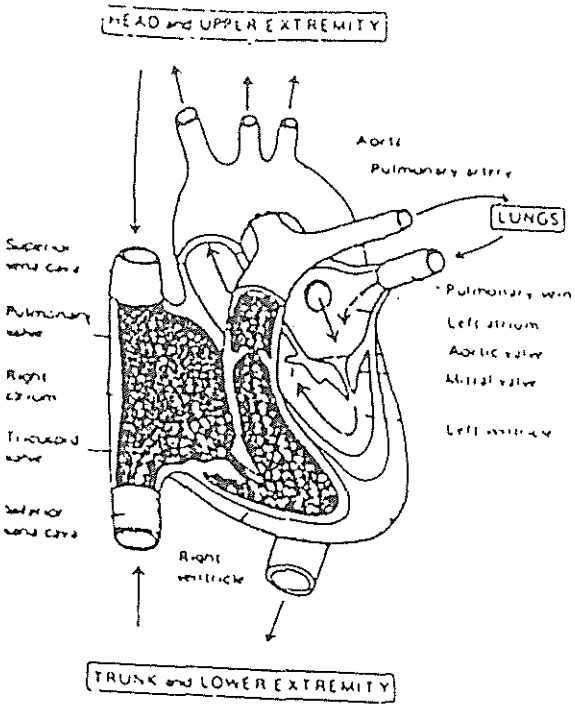


Fig. 2.1 Physical structure of heart. Source : [3]

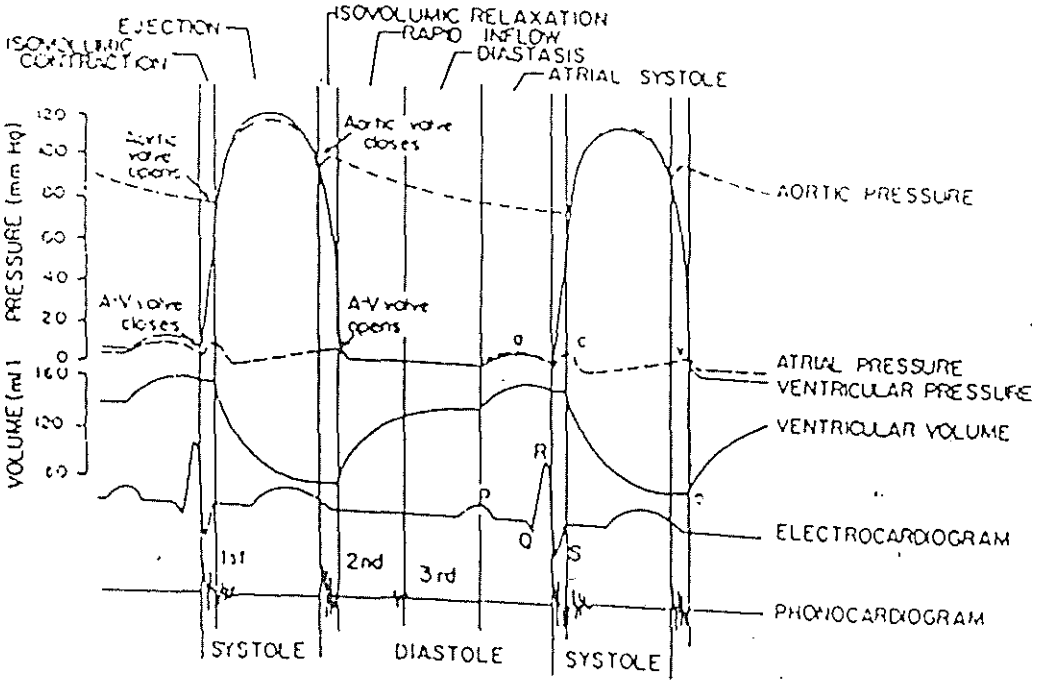


Fig. 2.2 Cardiac output. Source : [3]

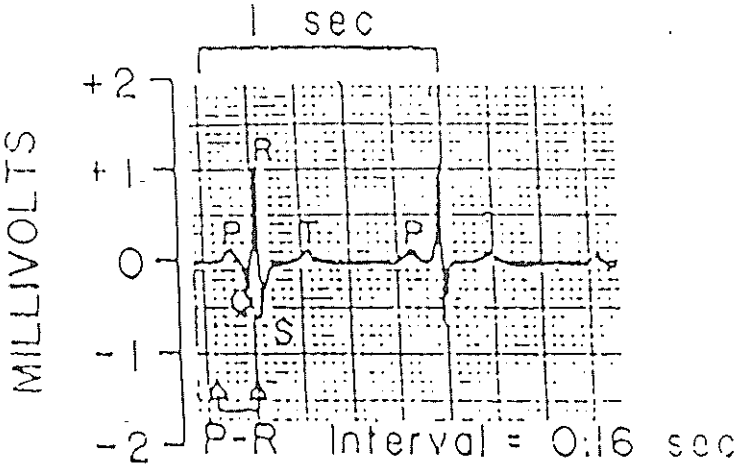


Fig. 2.3 Normal electrocardiogram. Source : [3]

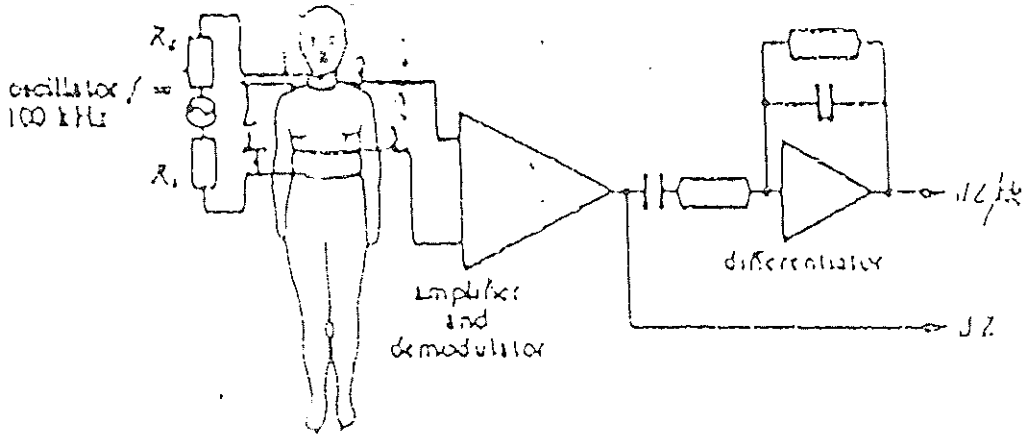


Fig. 2.4 4-terminal system for indirect measurement of cardiac output (Kubicek's method). Source : [5]

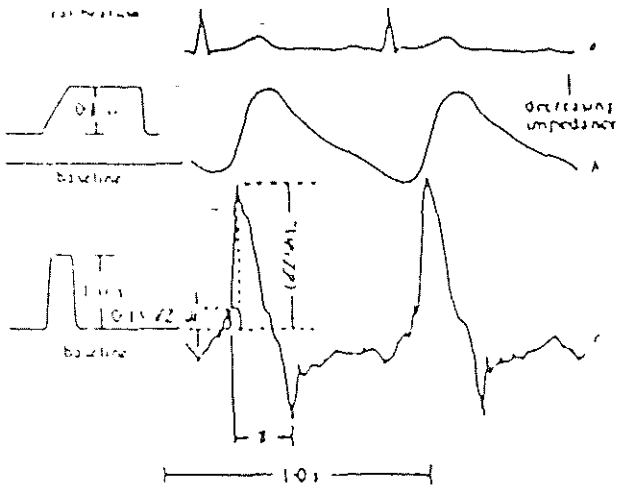


Fig. 2.5 Typical impedance waveform from thorax of human subject (Kubicek's method). Source:[5]

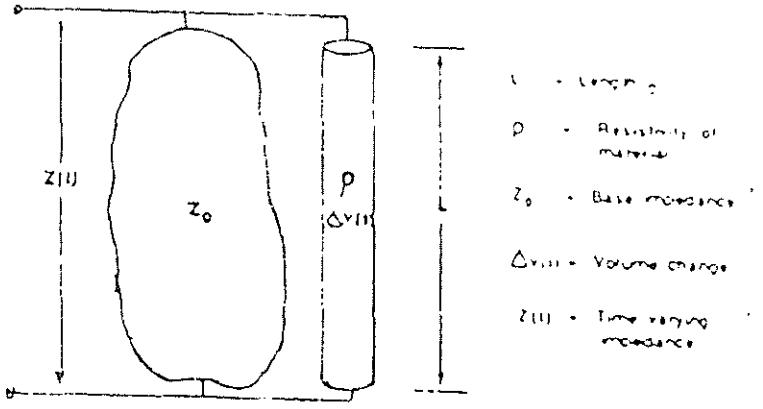


Fig. 2.6 Parallel column model. Source : [8]

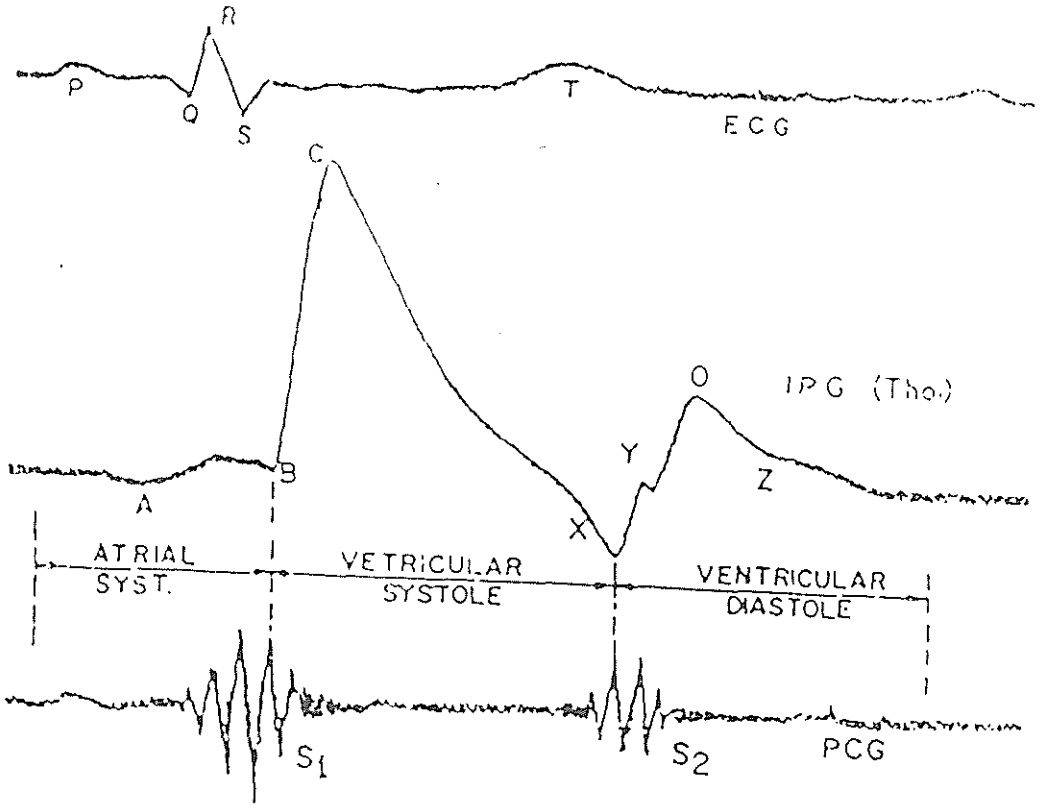


Fig. 2.7 Schematic representation of ECG, dz/dt, and phonocardiogram in time domain. Source : [9]

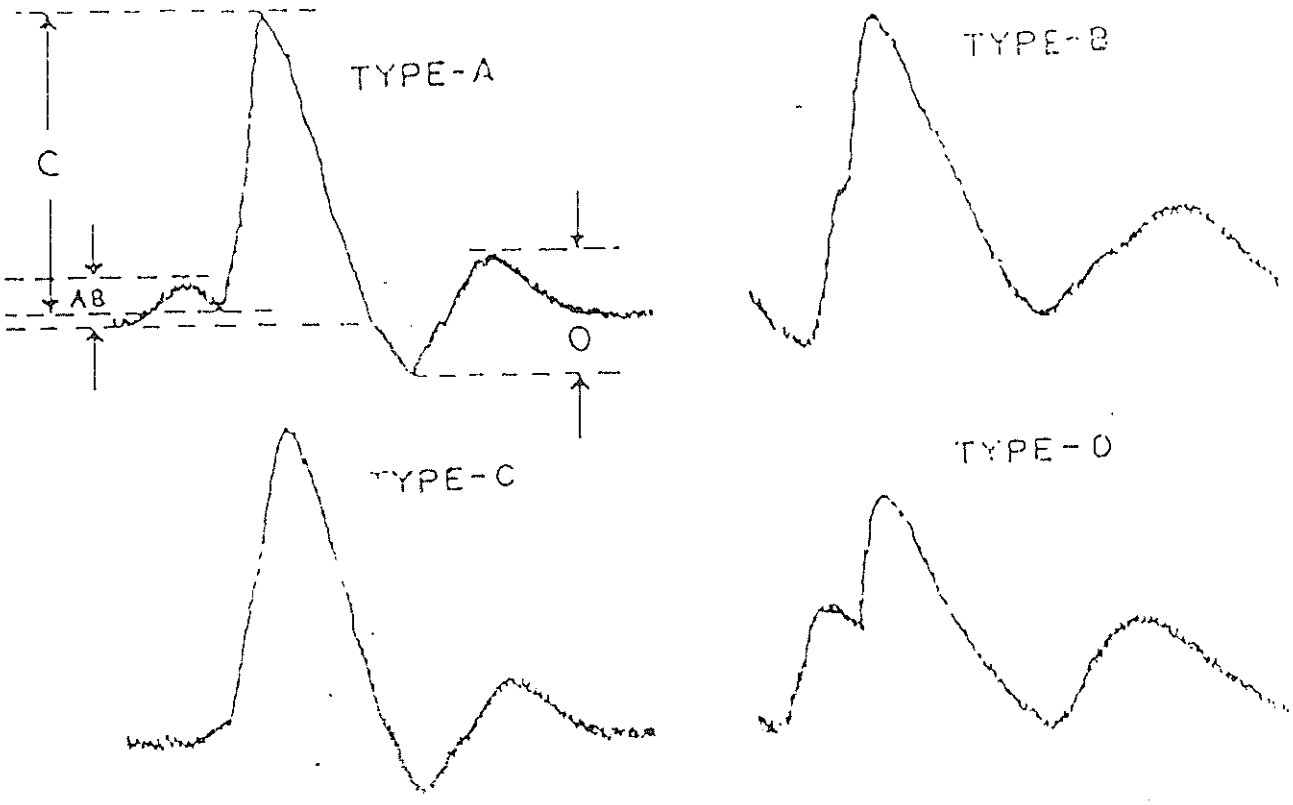
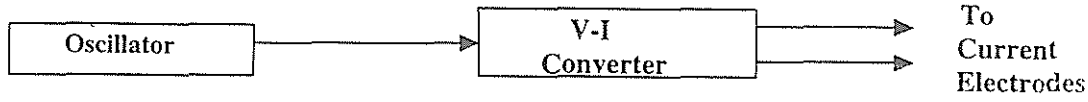
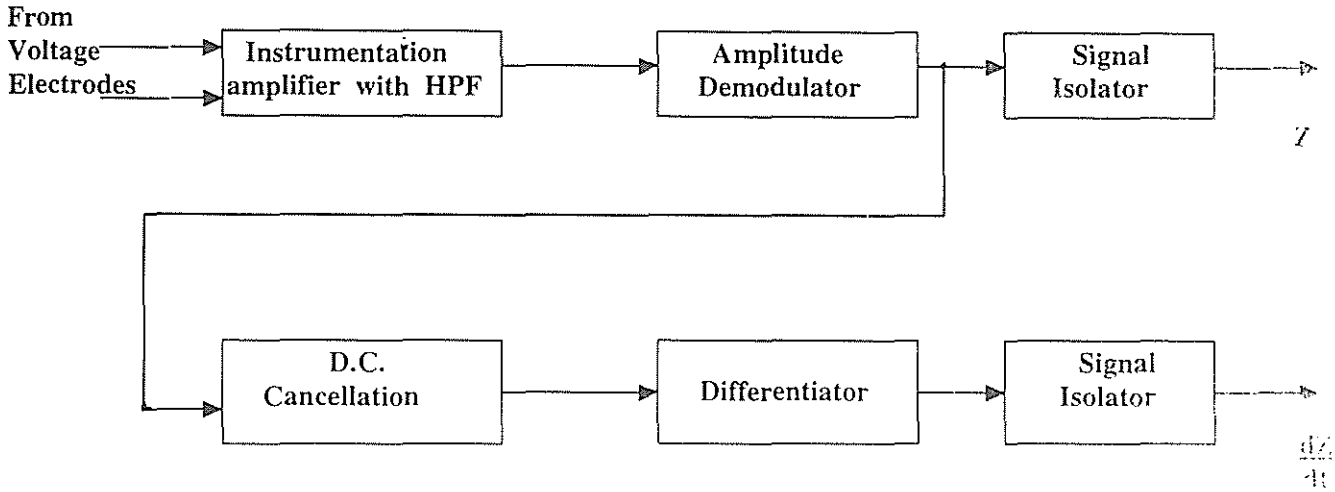


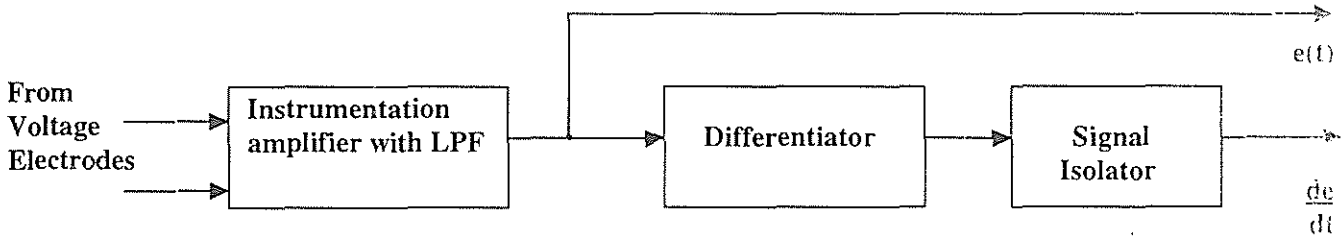
Fig. 2.8 Shape of impedance waveforms. Source : [10]



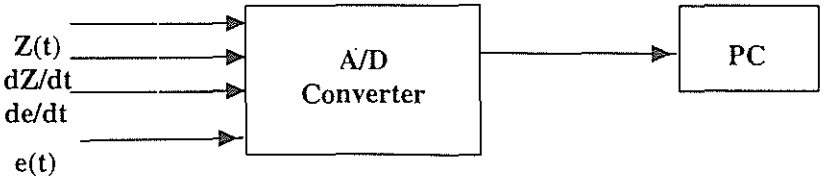
(a) Excitation circuit



(b) Impedance Circuit



(c) ECG circuit



(d) Interface Setup

Fig. 3.1 Block diagram of impedance cardiograph developed by Joshi [2].

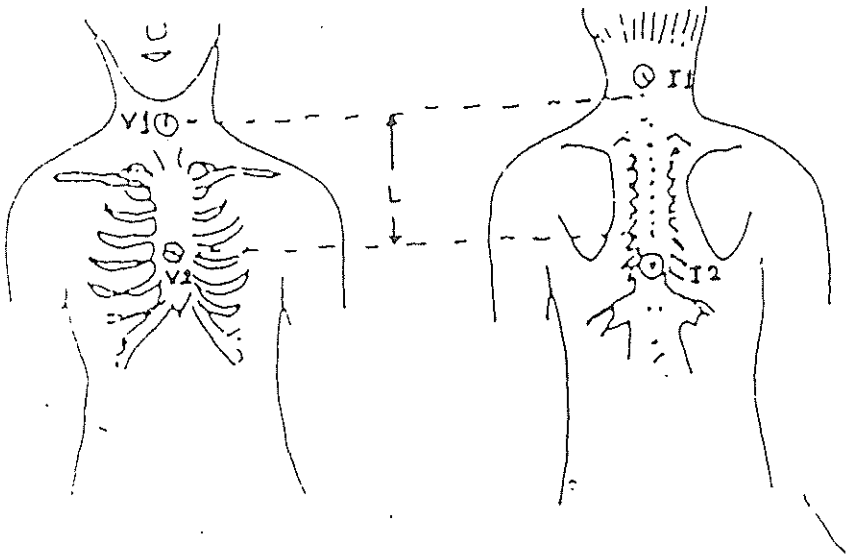


Fig. 3.2 Arrangement for placing spot electrodes proposed by Qu et al. Source : [7]

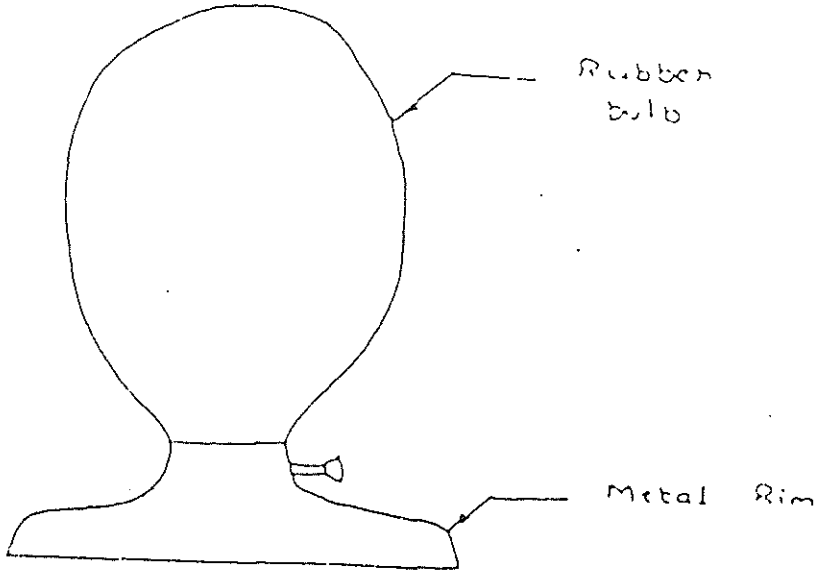
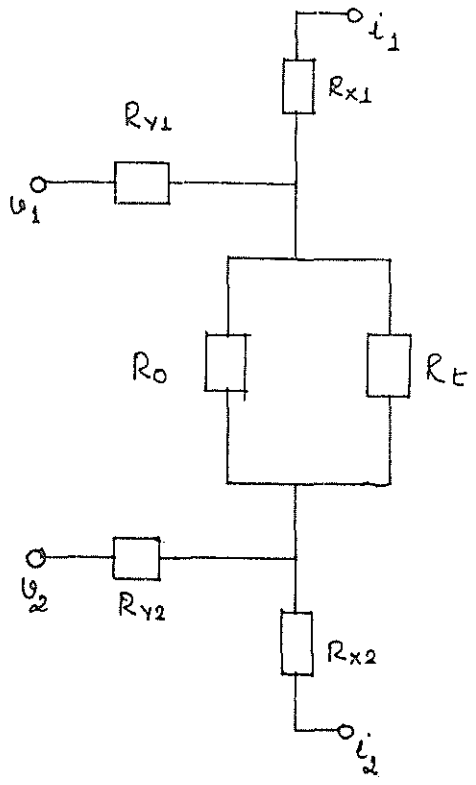
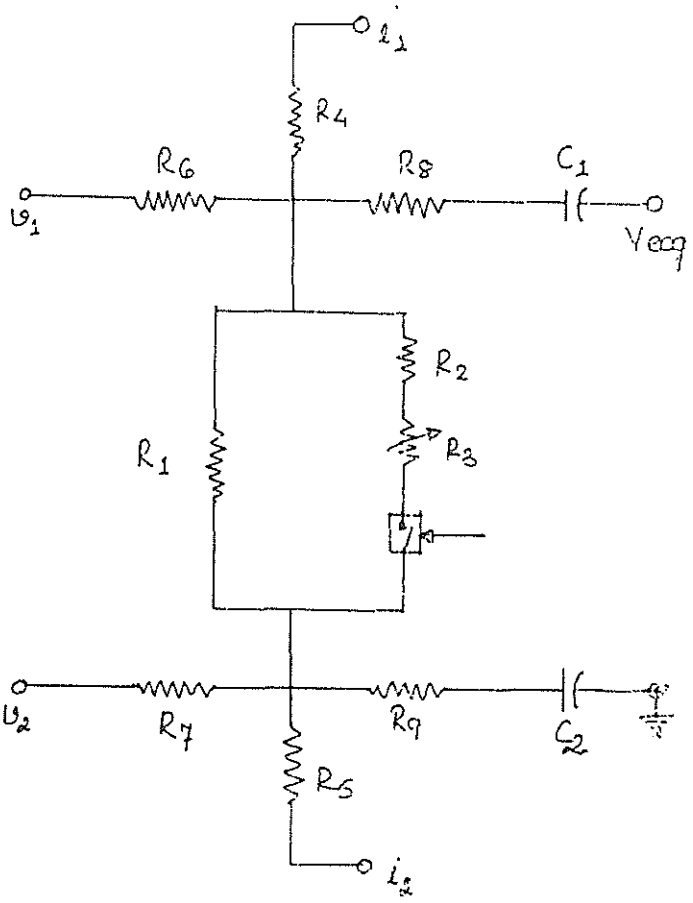


Fig. 3.3 ECG chest electrodes



(a)



(b)

R_0 = Base Impedance

R_t = Time Varying impedance.

R_{x1}, R_{x2} = Contact resistance for excitation Current electrodes

R_{y1}, R_{y2} = Contact resistance For sensing voltage electrodes

$R_0 = R_1$

$R_t = R_2 + R_3, \infty$

$R_{x1} = R_4$

$R_{x2} = R_5$

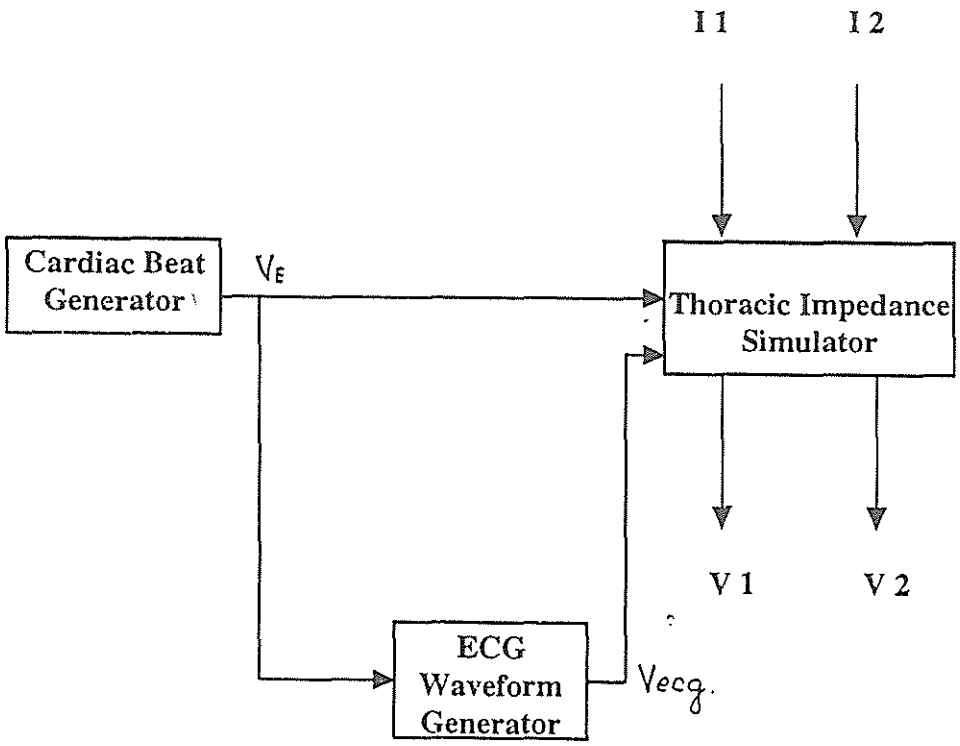
$R_{y1} = R_6$

$R_{y2} = R_7$

Fig. 4.1.a Two-Cylinder model of thorax impedance

Fig 4.2.b. Resistance Simulator.

Fig. 4.1 Impedance simulator



I 1, I 2 = Current Electrodes

V 1, V 2 = Voltage Electrodes

Fig. 4.2 Block Diagram of Thoracic Impedance Simulator

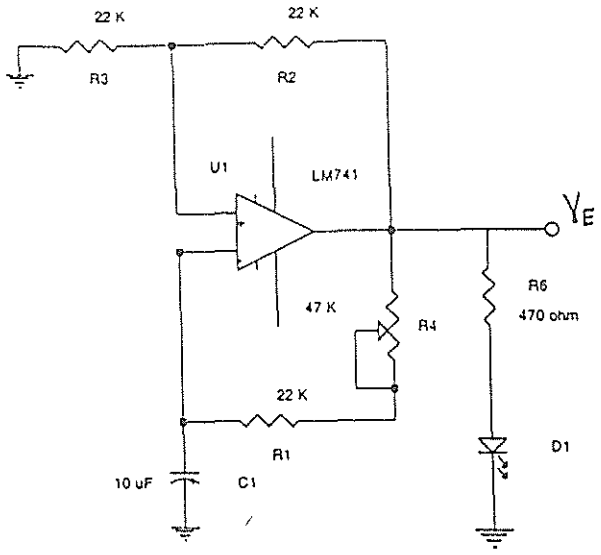


Fig. 4.3 Cardiac beat generator

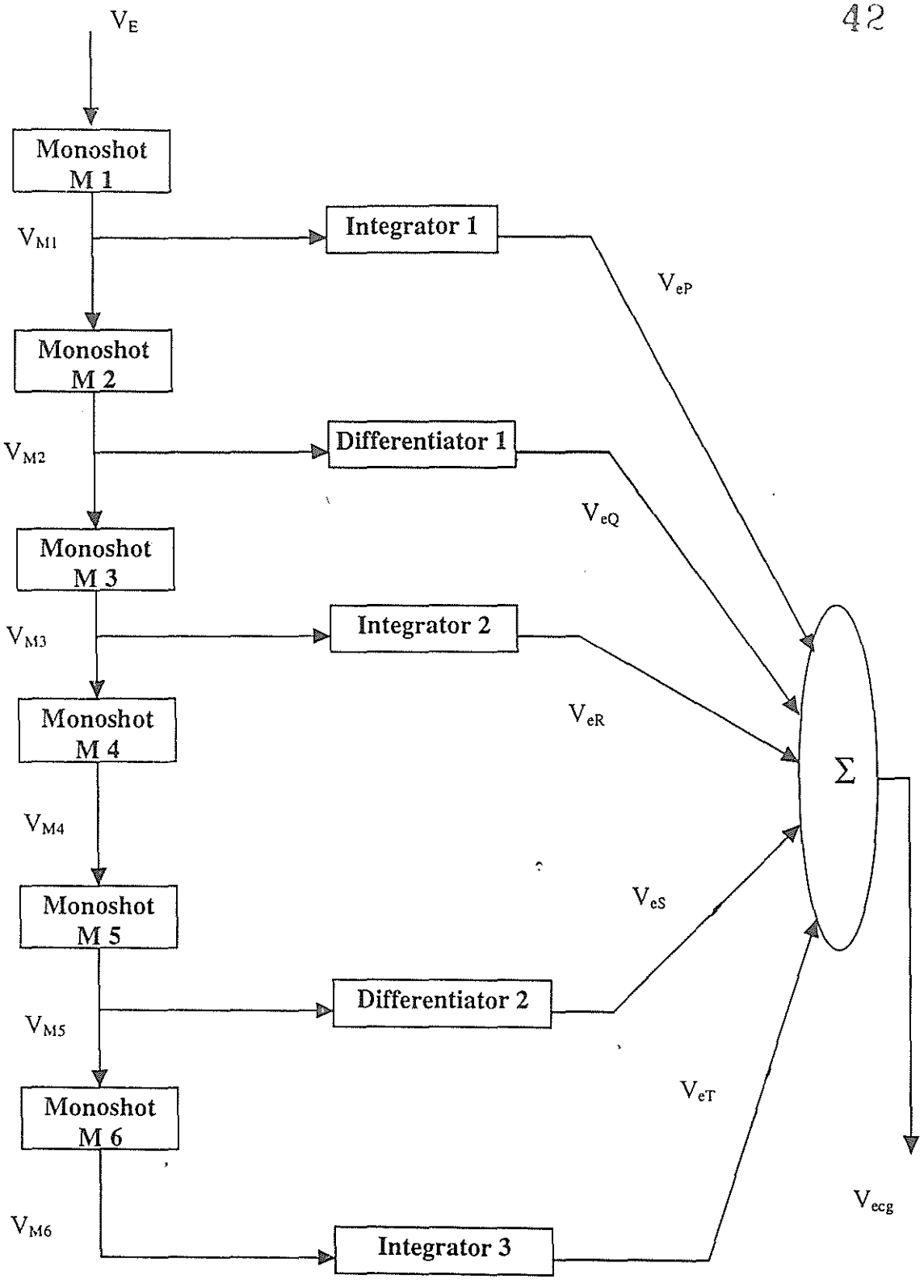


Fig. 4.4 Block diagram of ECG waveform Simulator

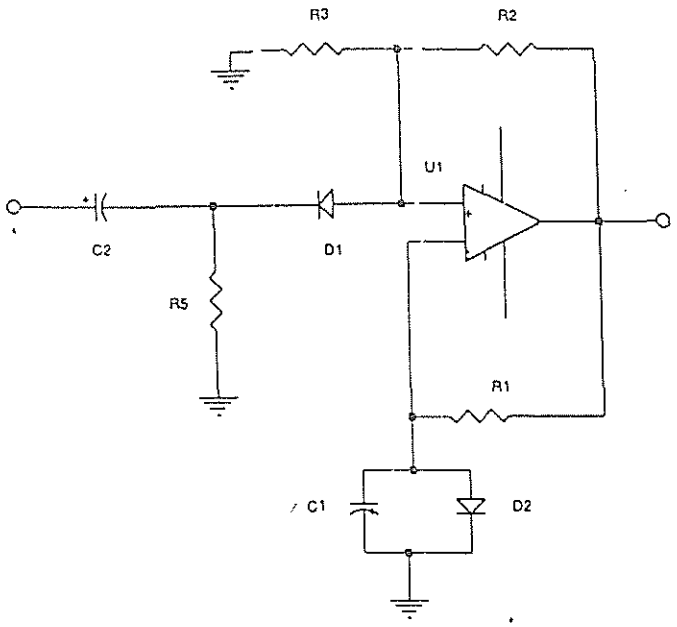


Fig. 4.5 Monostable multivibrator

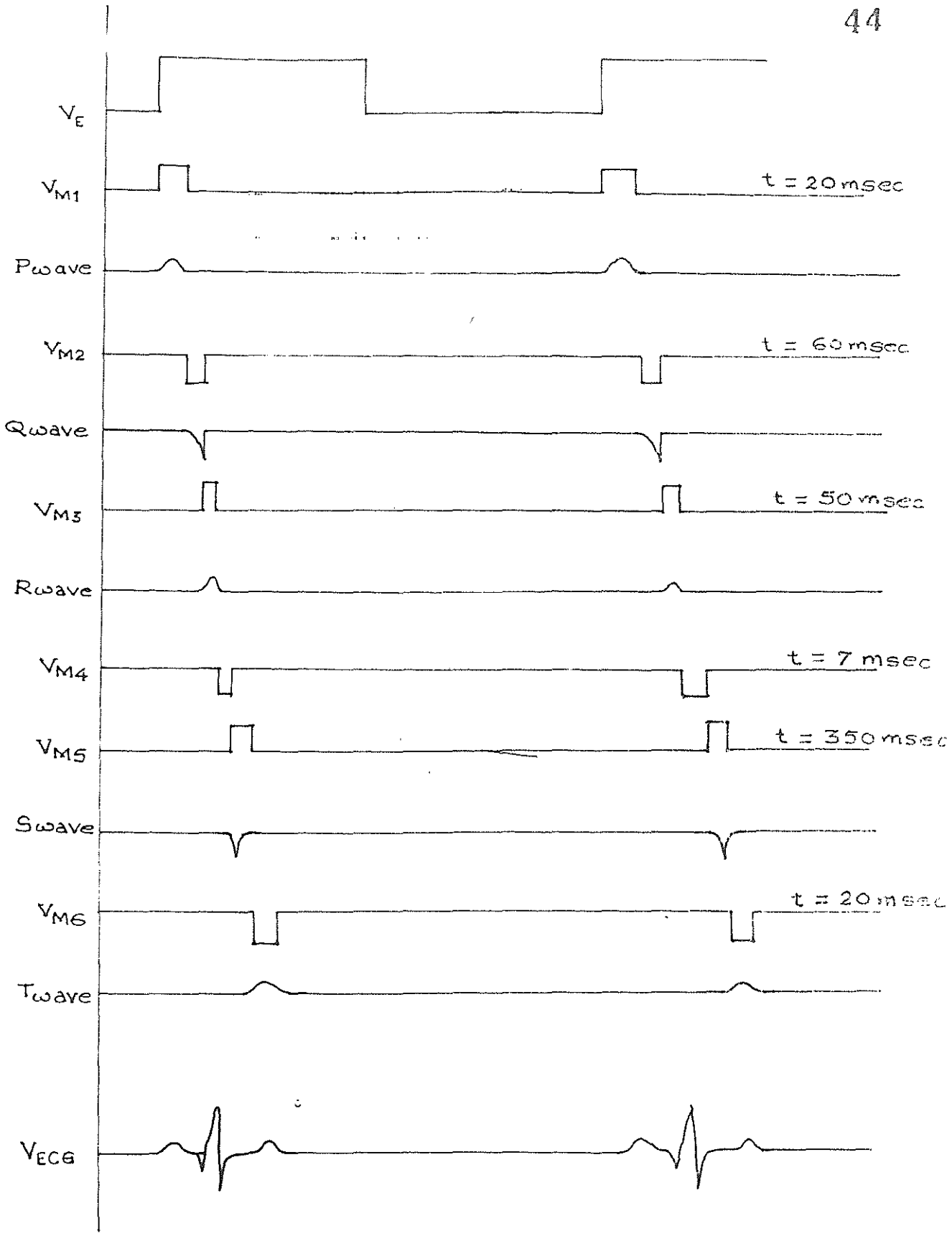


Fig. 4.6 Timing diagram for generation of ECG waveform

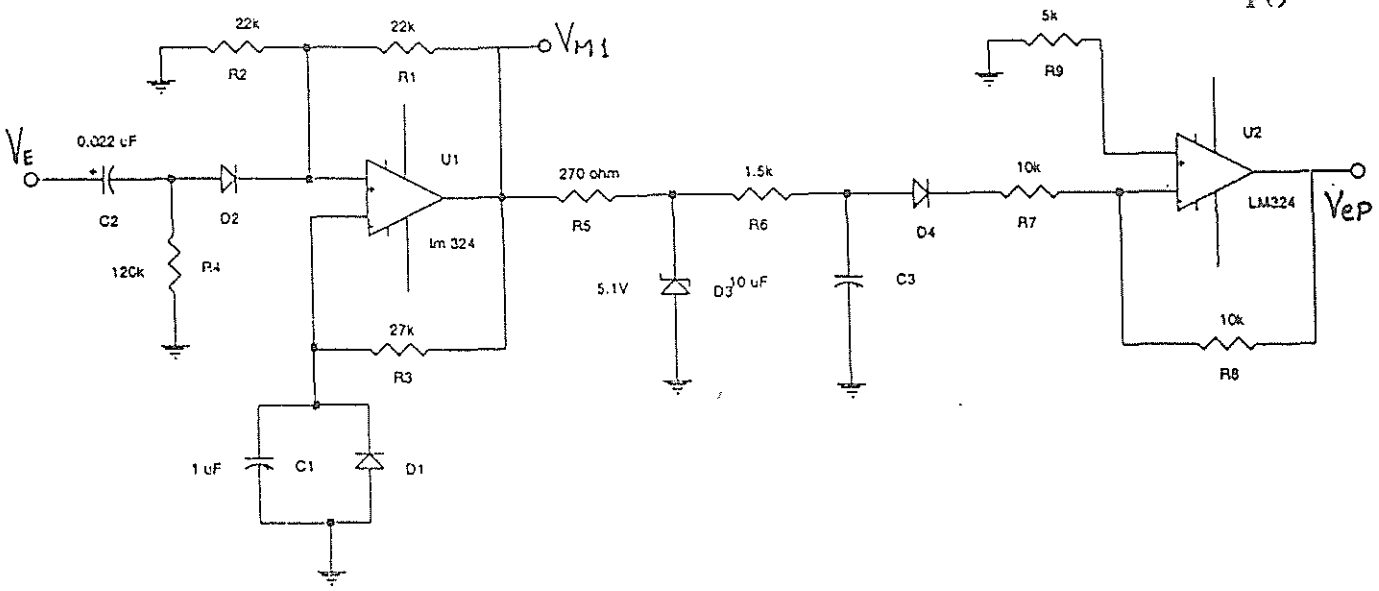


Fig. 4.7 Circuit diagram for generation of P wave

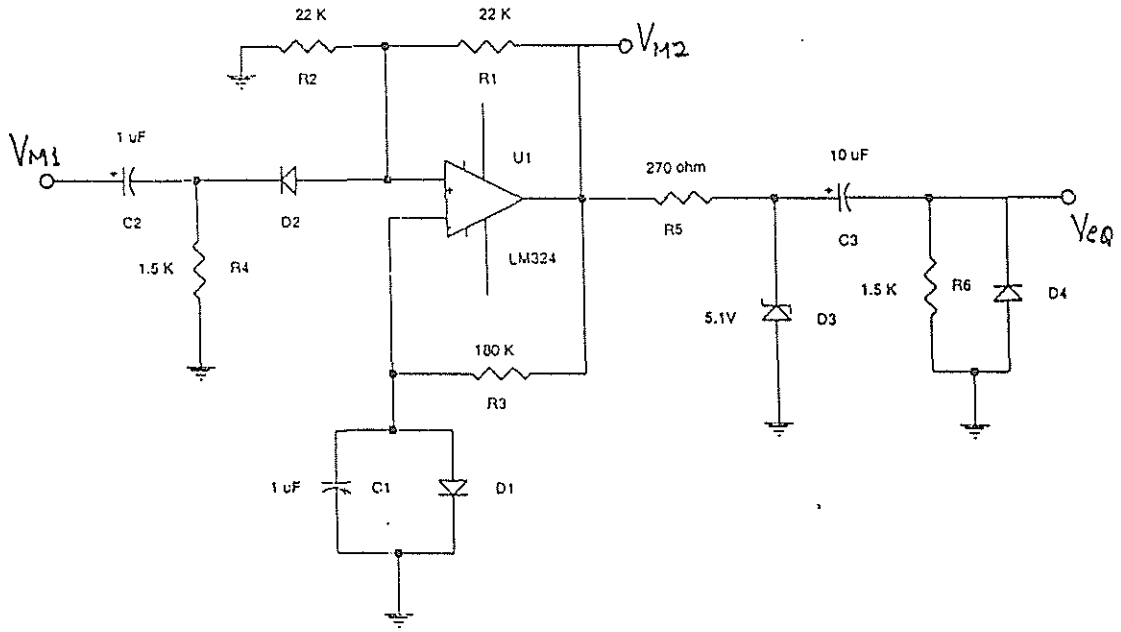


Fig. 4.8 Circuit diagram for generation of Q wave

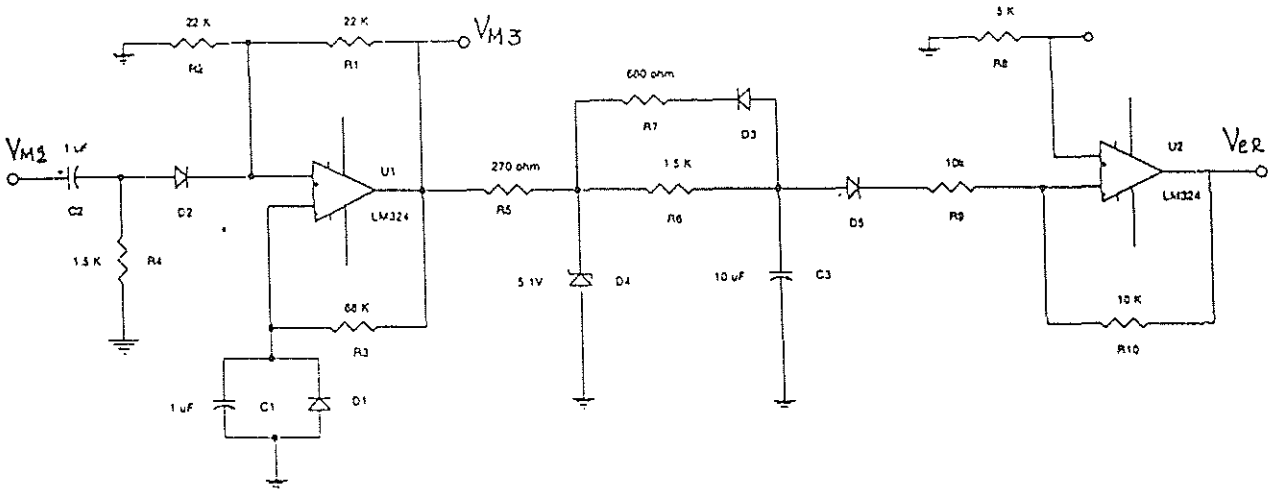


Fig. 4.9 Circuit diagram for generation of R wave

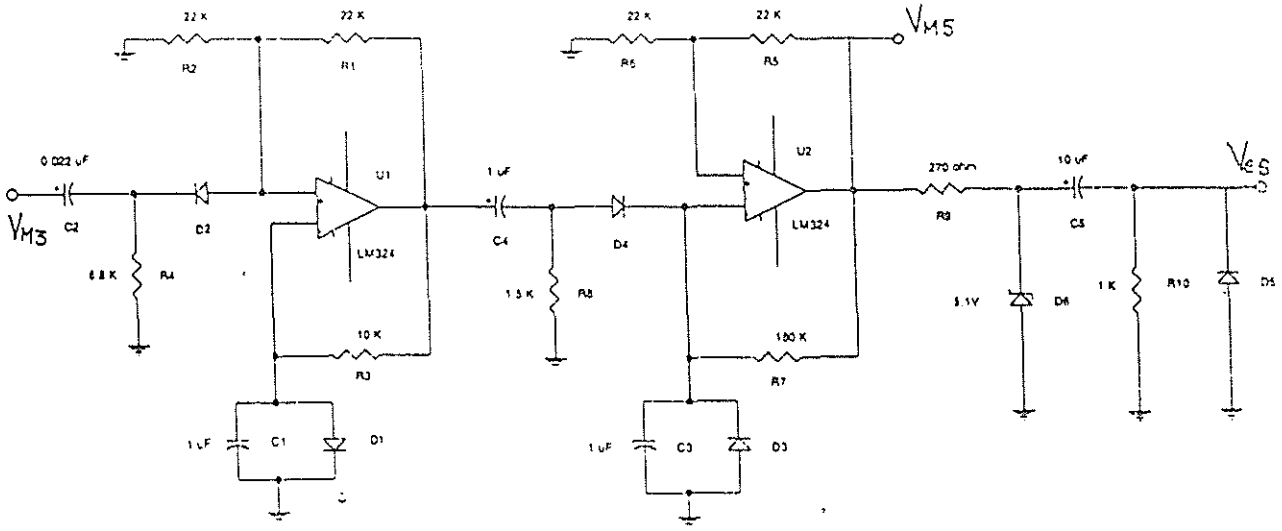


Fig. 4.10 Circuit diagram for generation of S wave

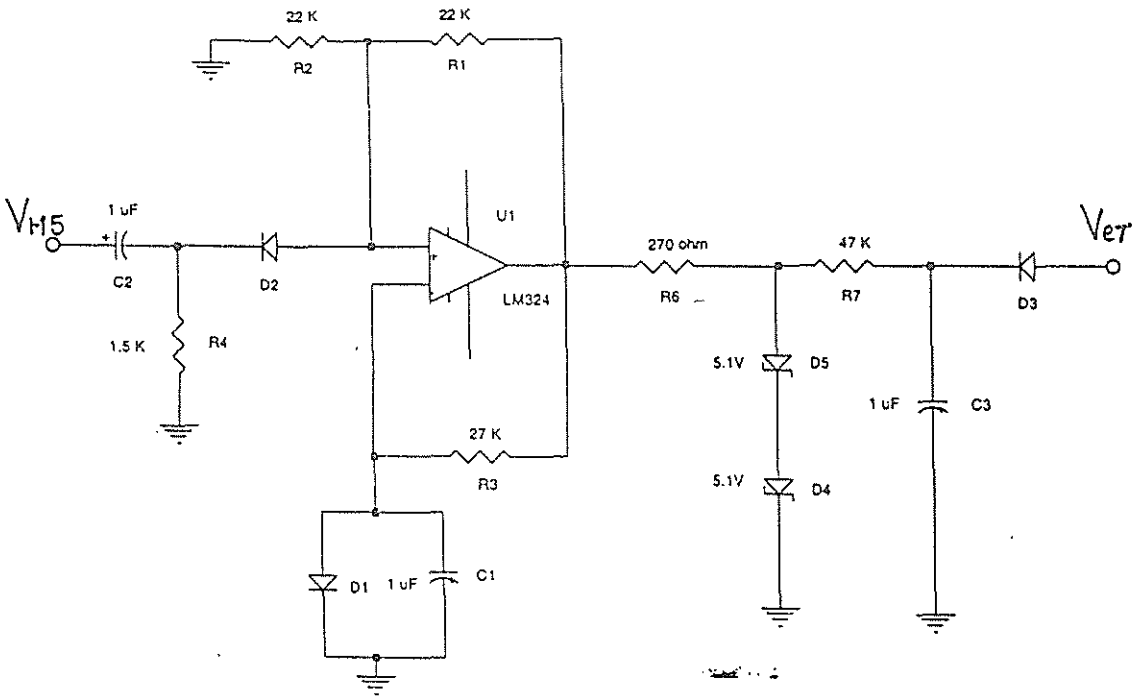


Fig. 4.11 Circuit diagram for generation of T wave

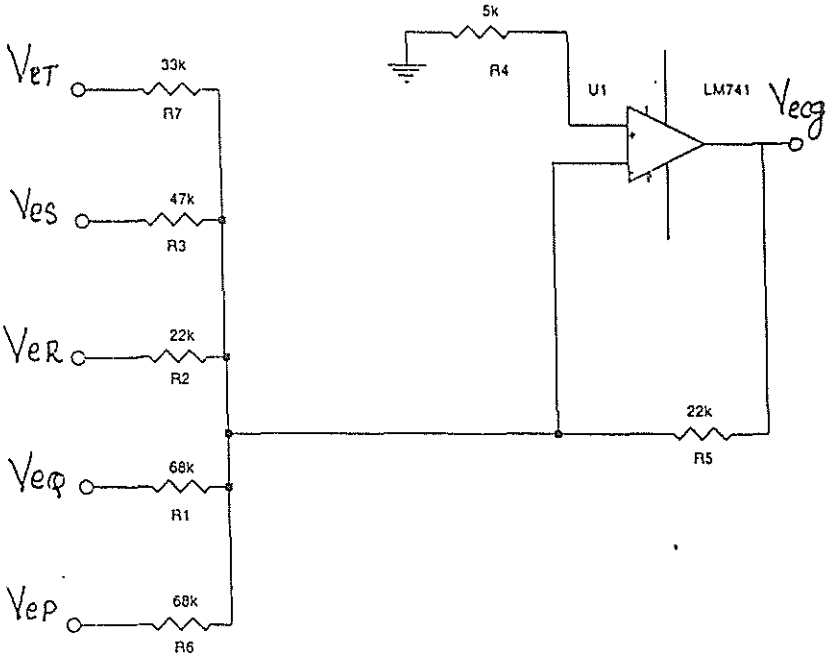


Fig. 4.12 Circuit diagram for generation of adder

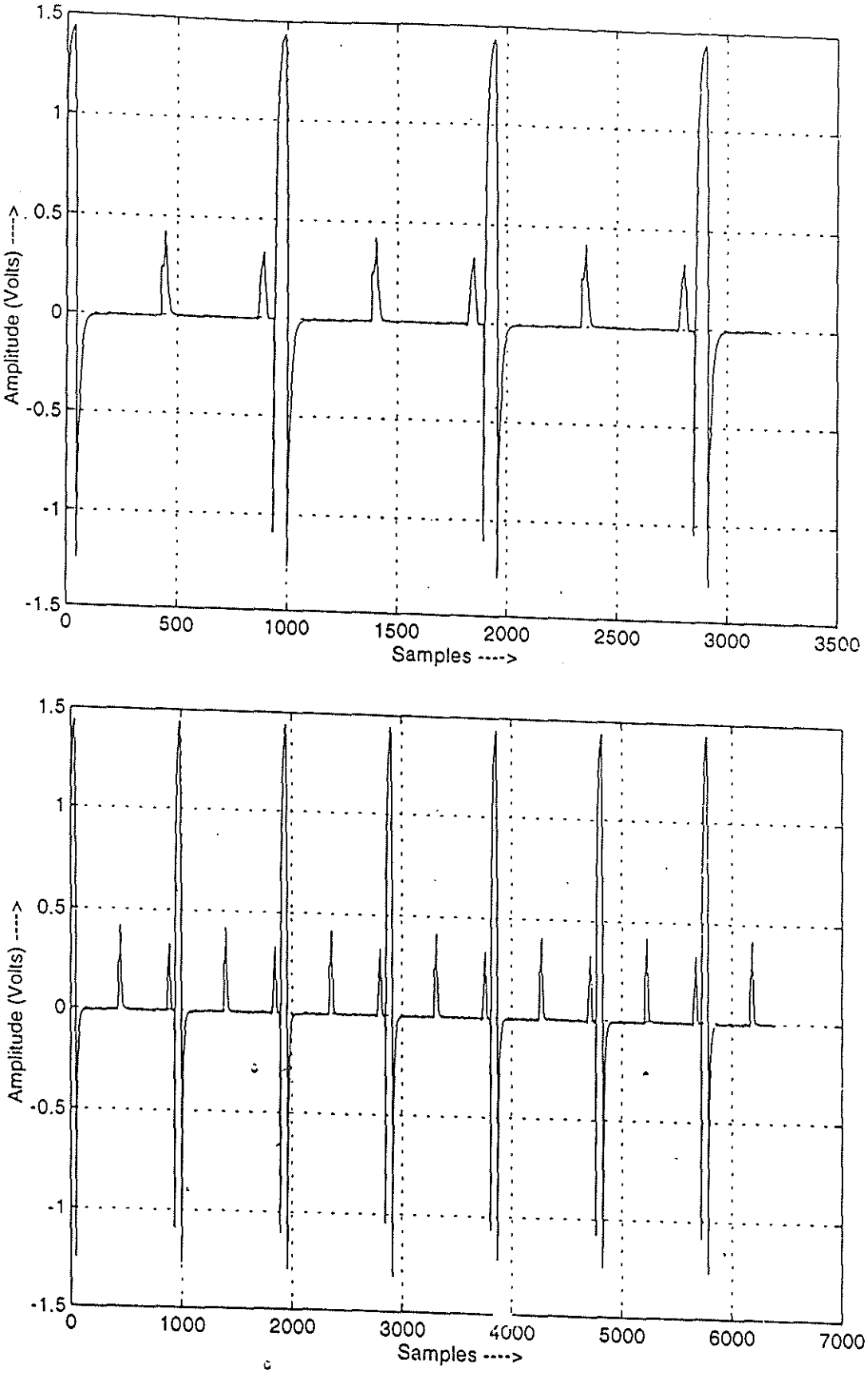


Fig. 4.13 Simulated ECG waveform

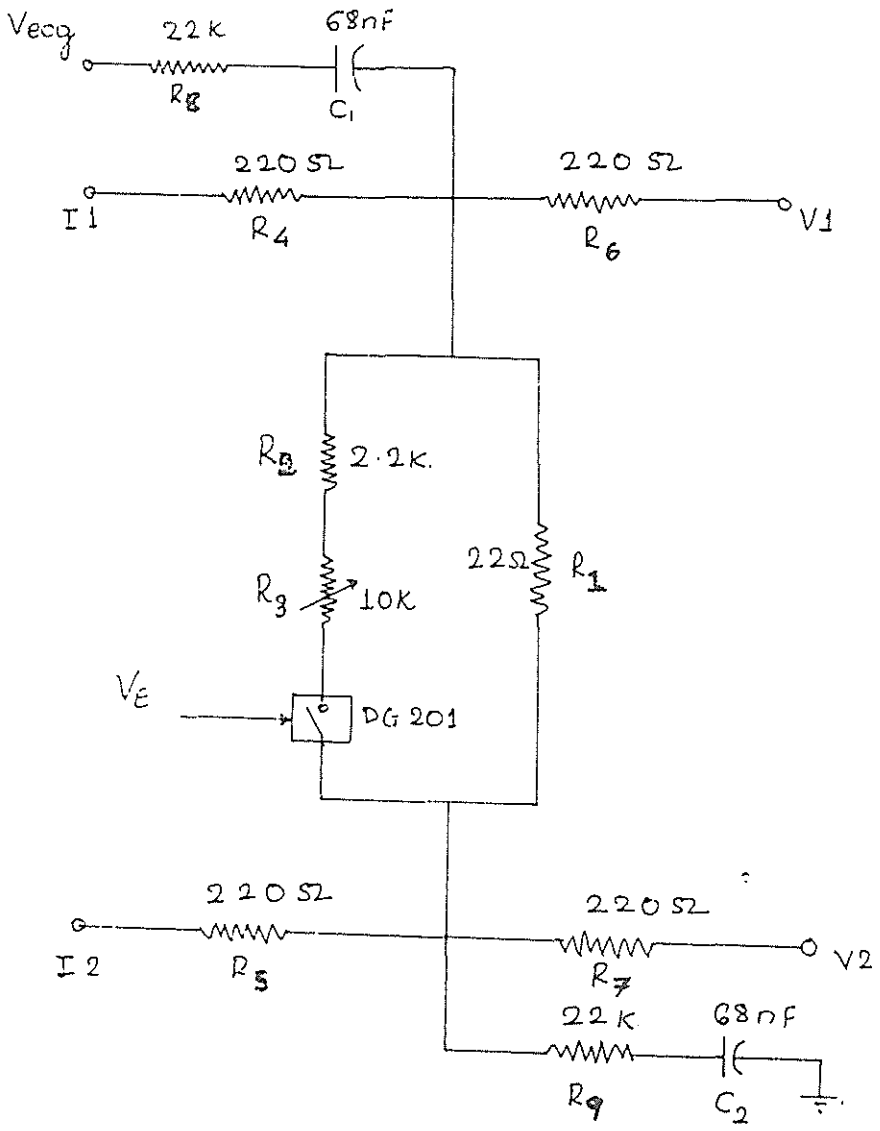


Fig. 4.14 Circuit diagram of impedance simulator

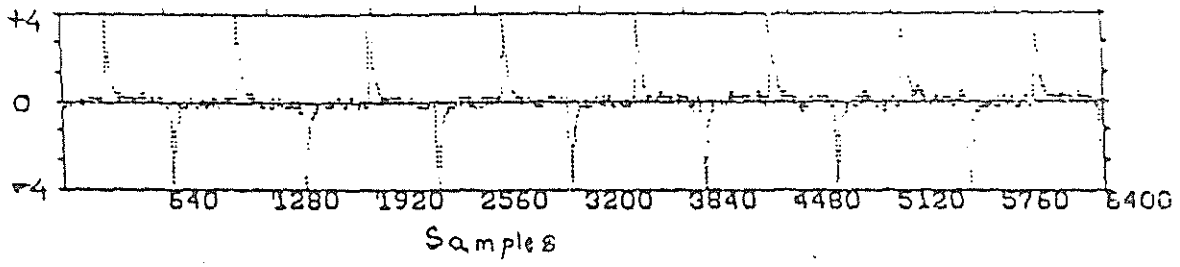


Fig. 4.15 Derivative of impedance, dz/dt, obtained from simulator

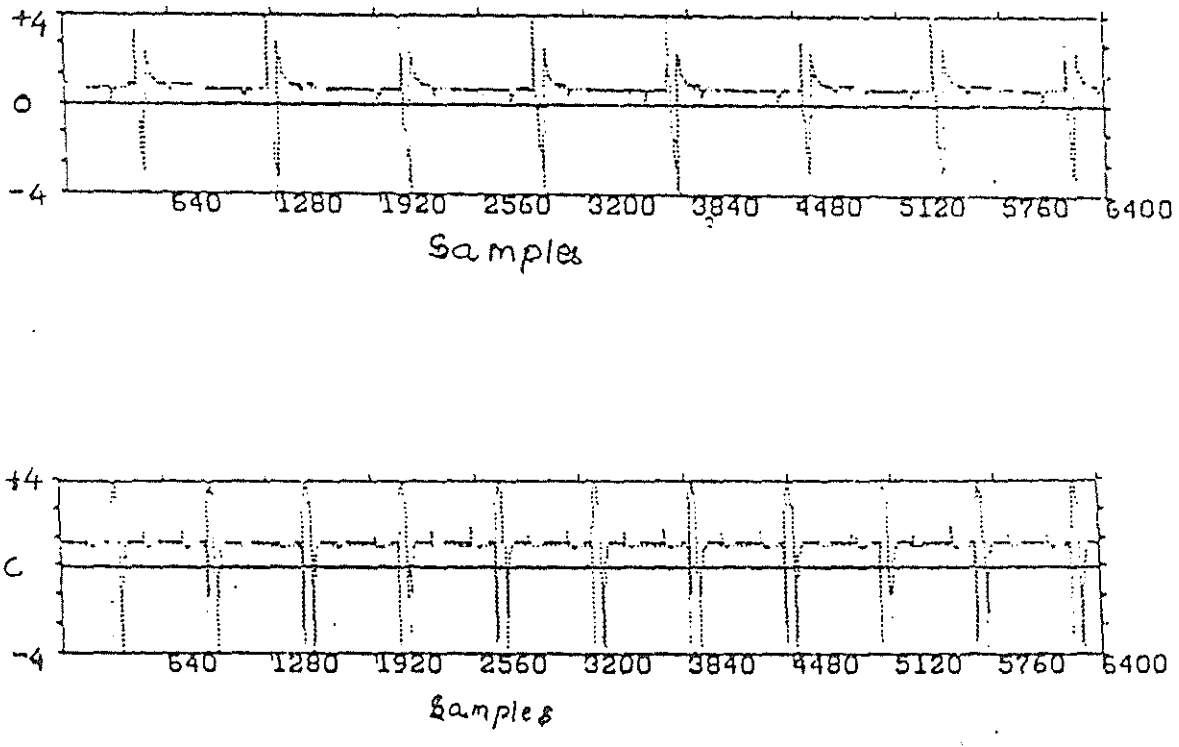
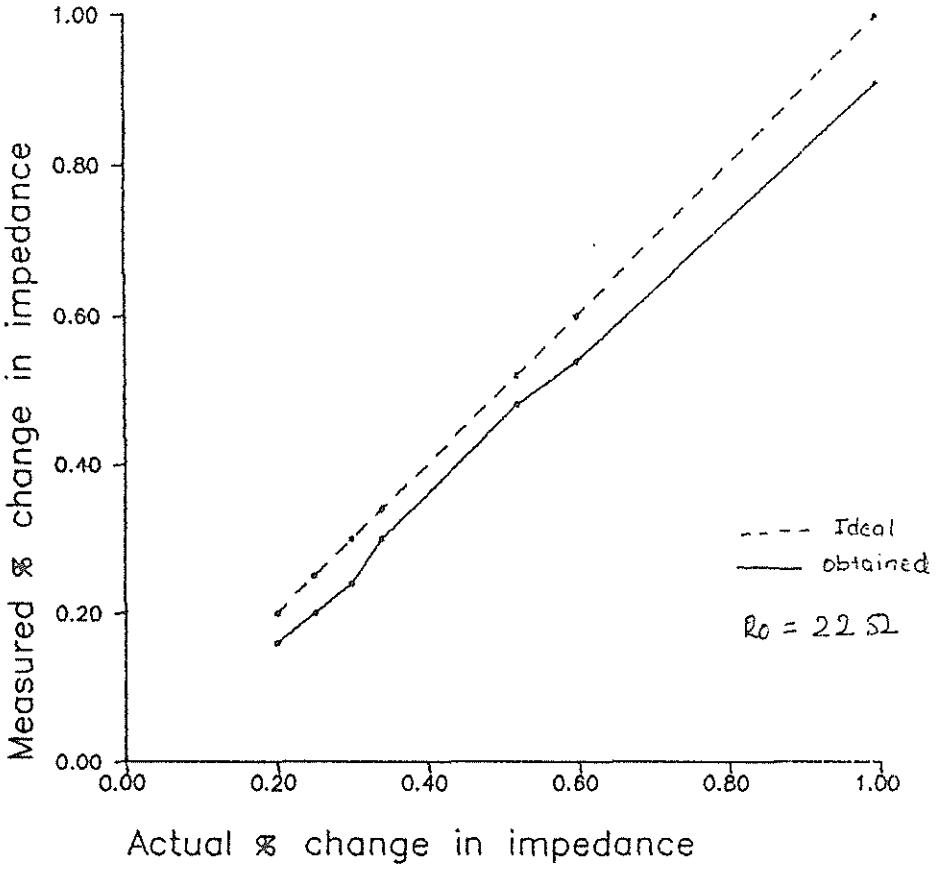


Fig. 4.16 e(t), and its derivative de/dt obtained from simulator



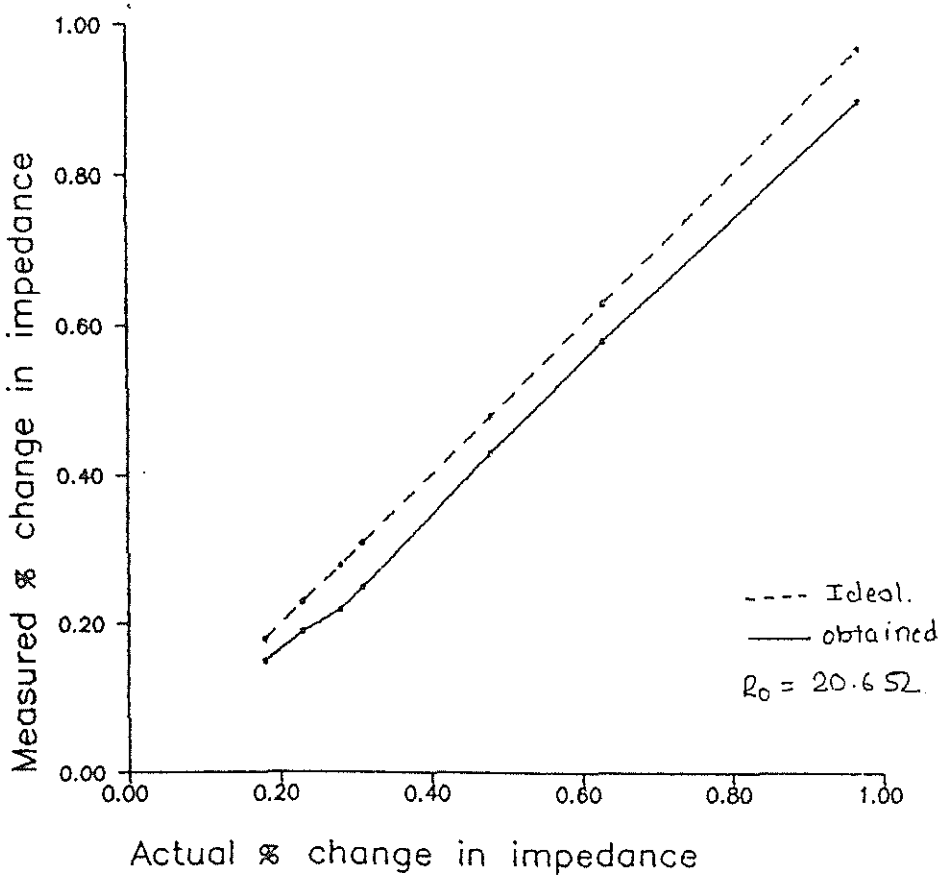
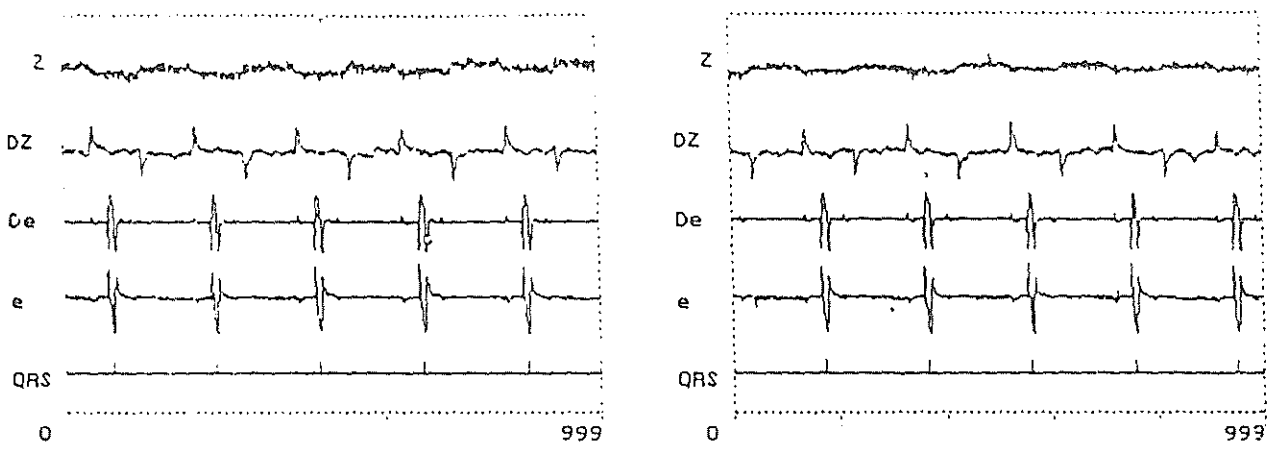
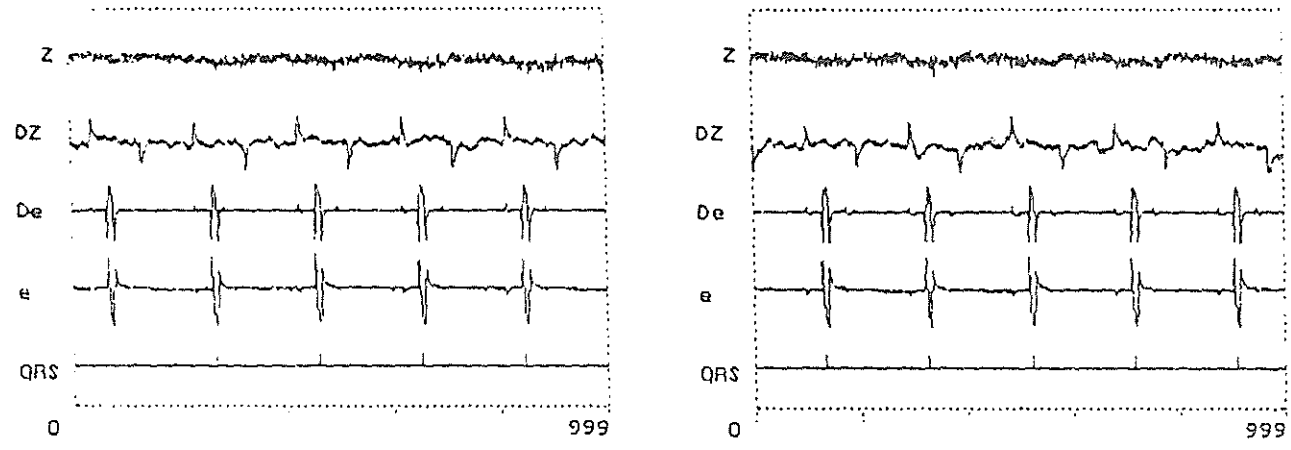


Fig. 6.1 Graph of actual % change in impedance v/s measured % change in impedance



1% Change



0.2% change

Fig. 6.2 Results from simulator

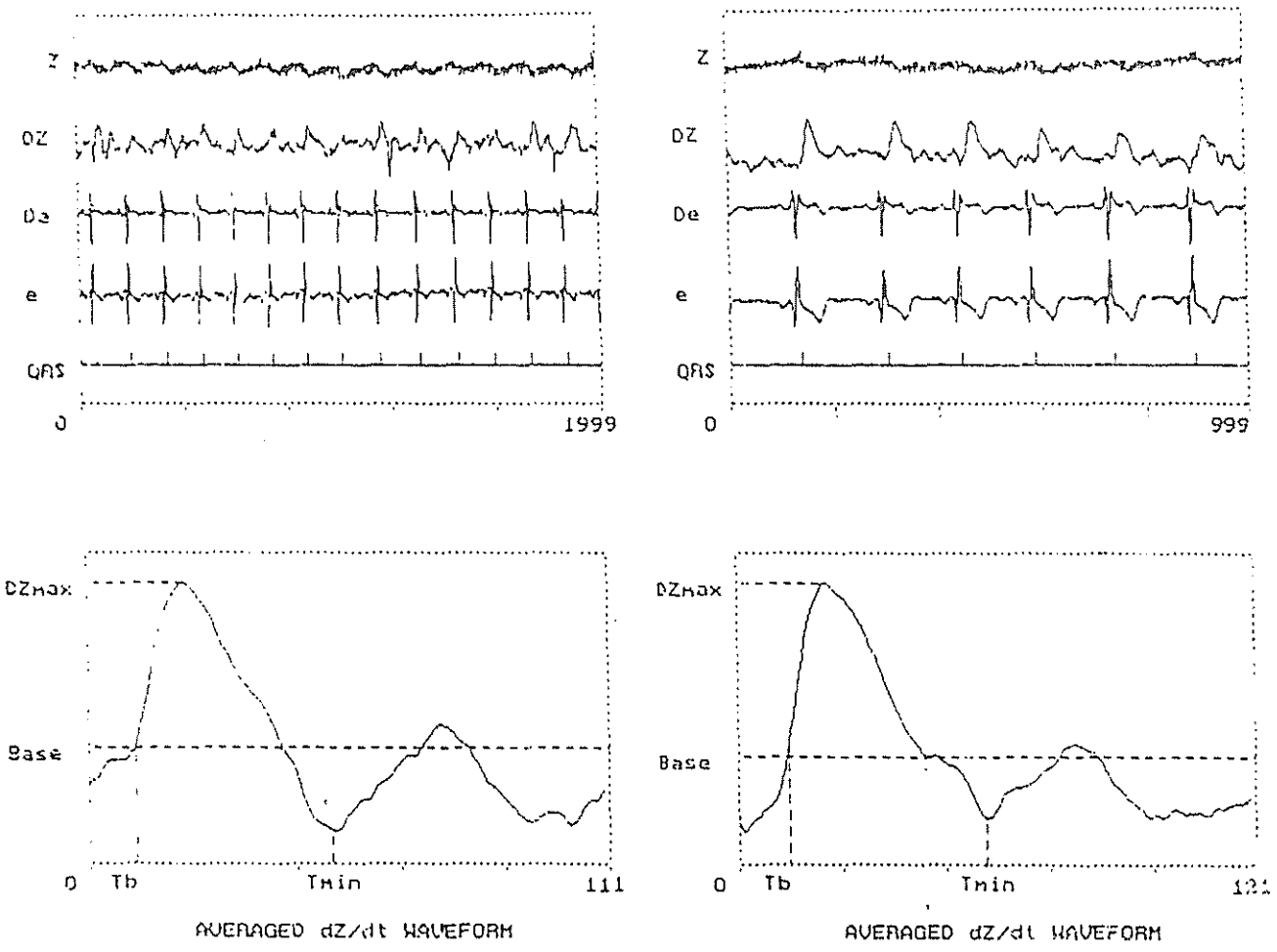


Fig. 6.3 Results from subjects ABC and SPT

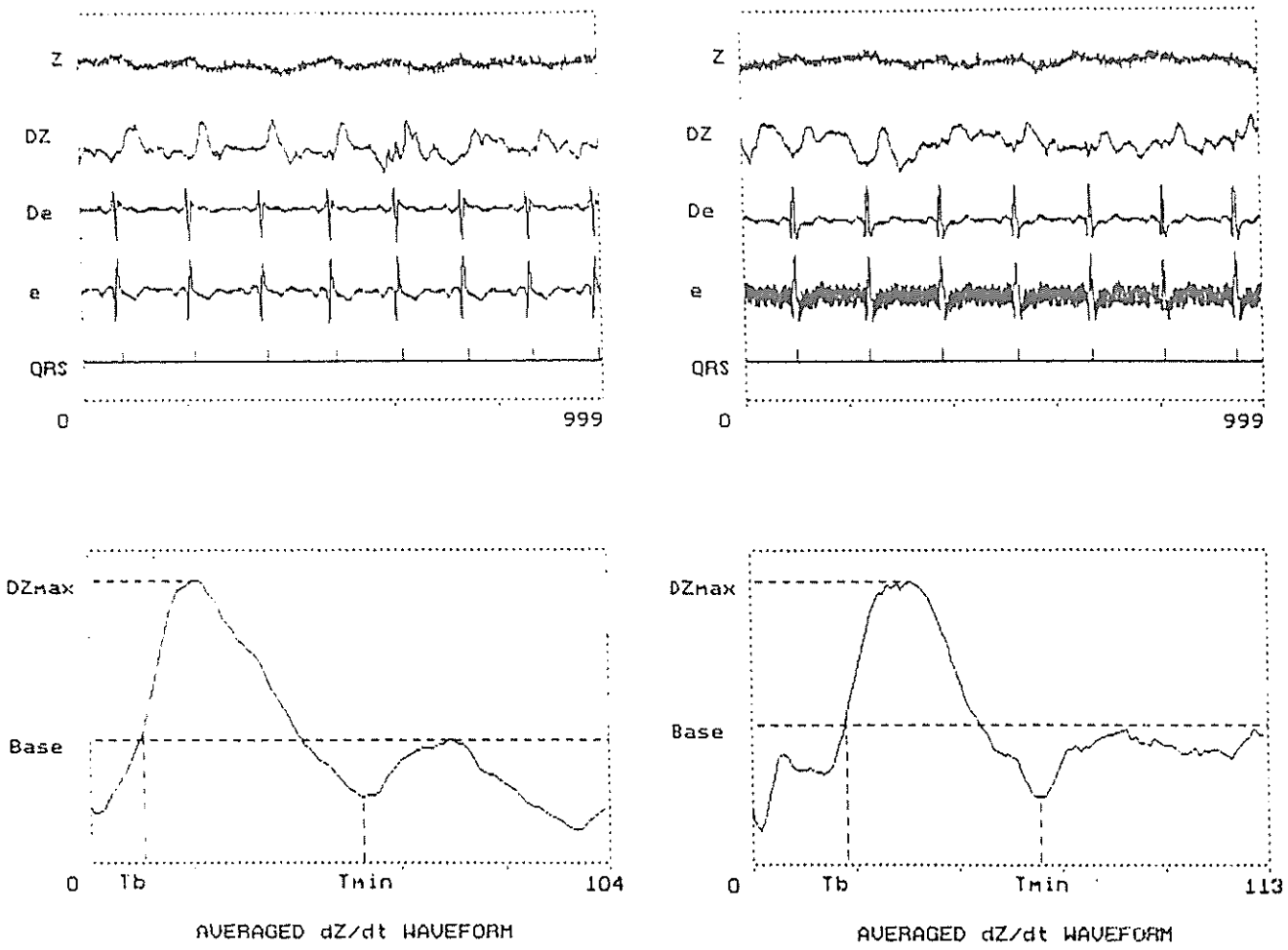


Fig. 6.4 Results from subjects SVK and MSH

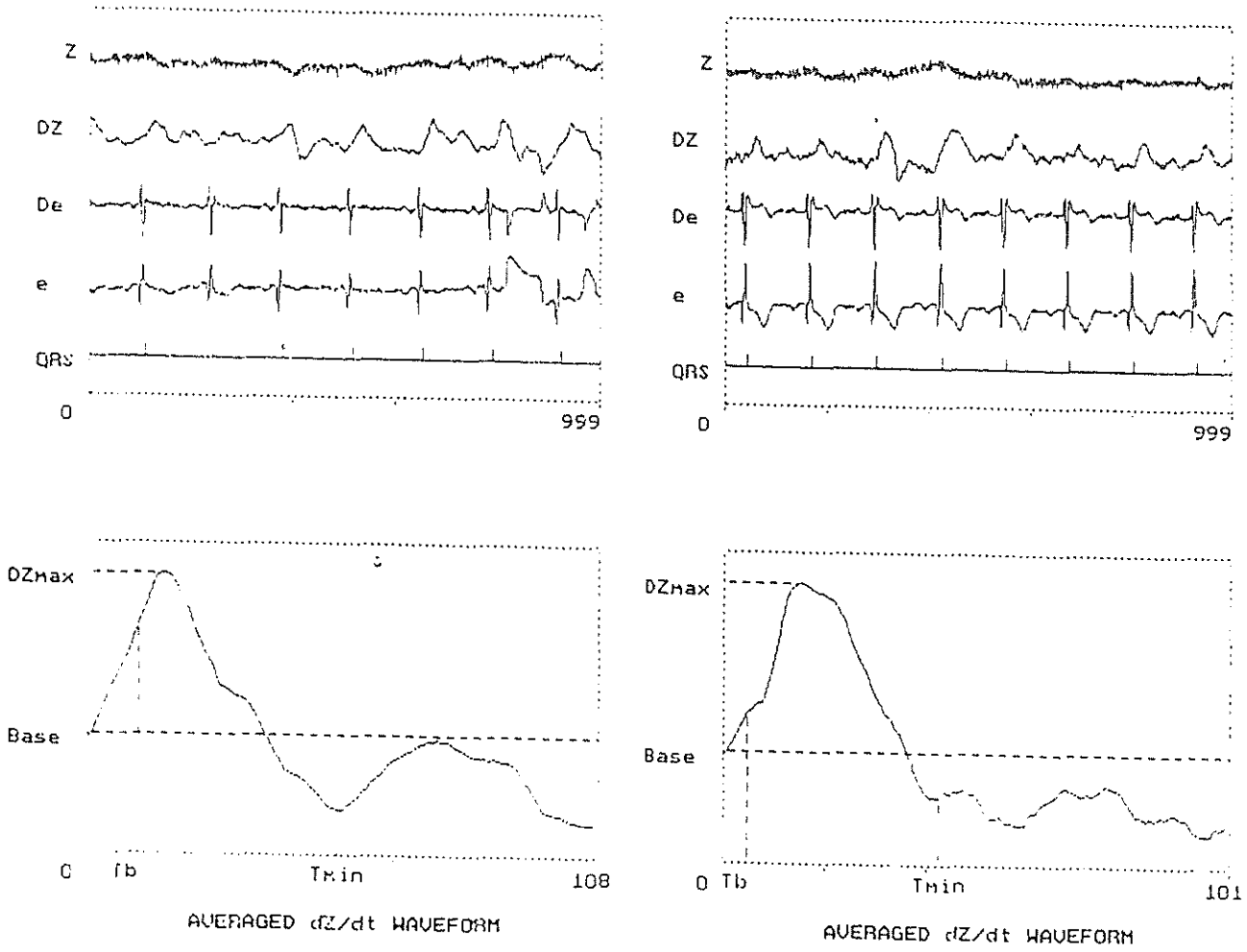


Fig. 6.5 Results from subjects KGS and CSK

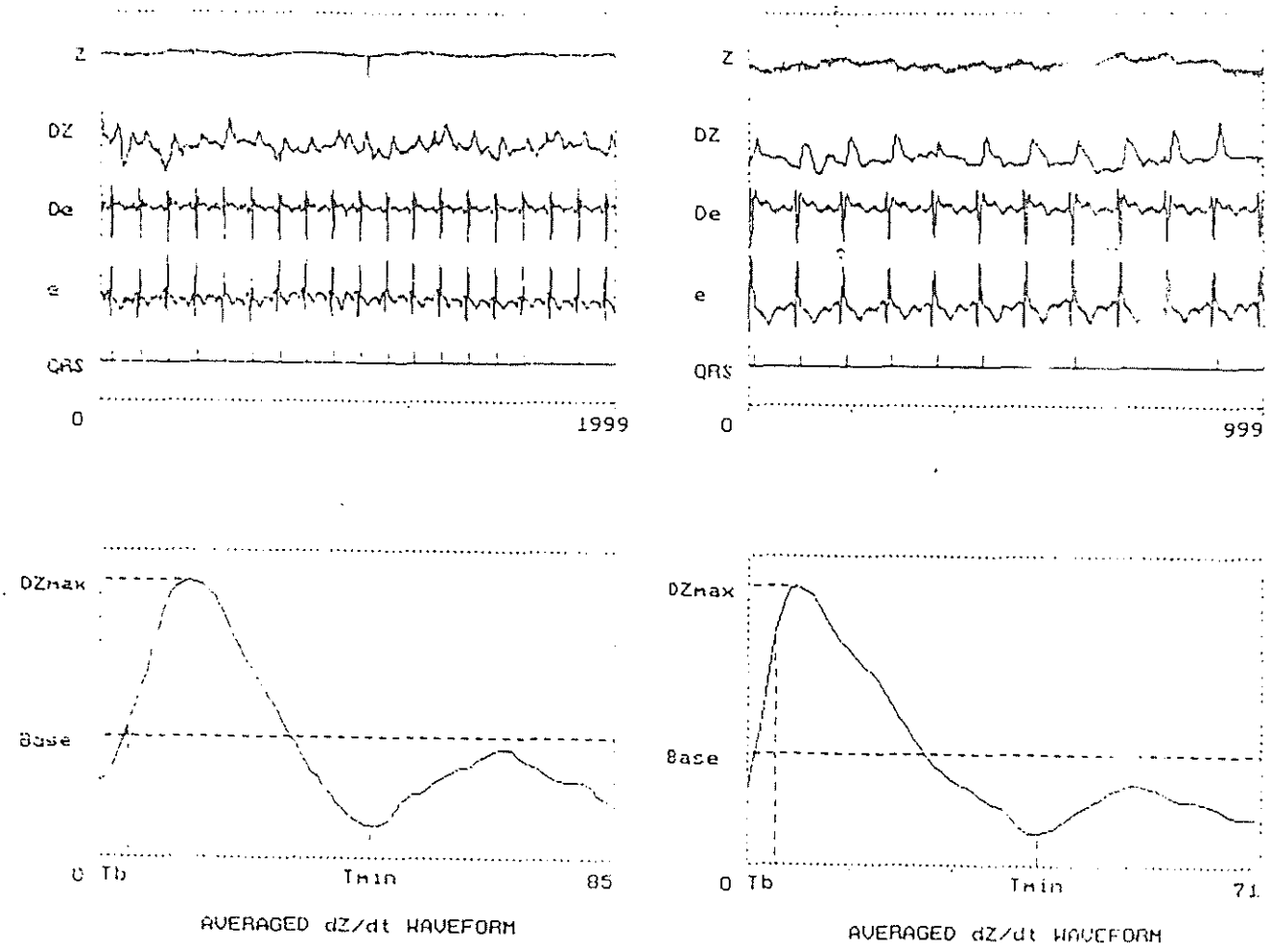


Fig. 6.6 Results from subjects ABC and SPT during exercise.

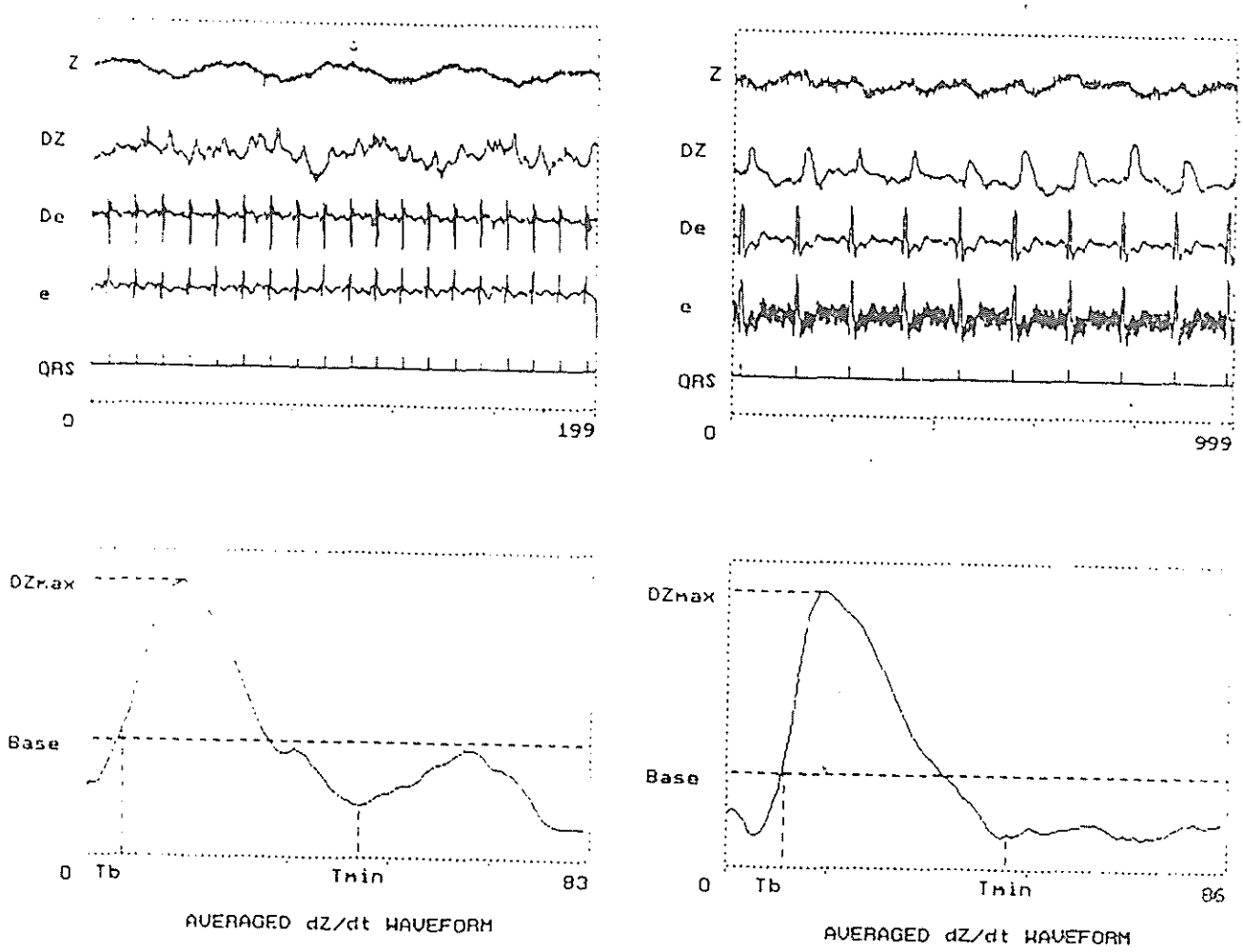
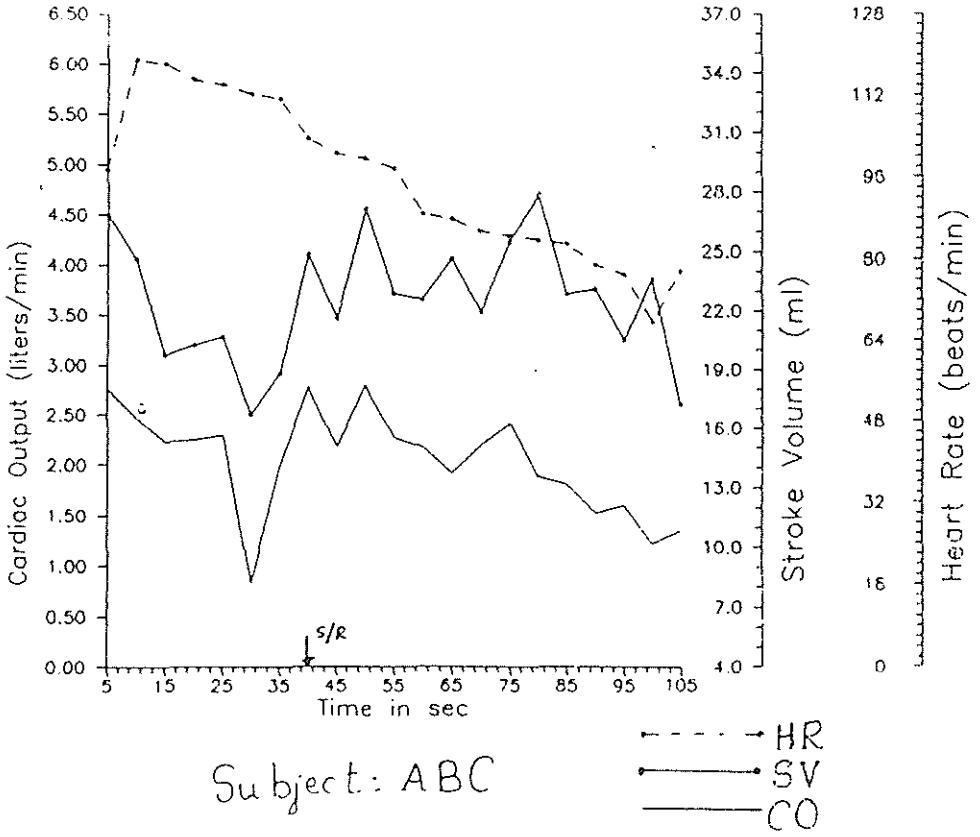


Fig. 6.7 Results from subjects KGS and MSH during exercise.



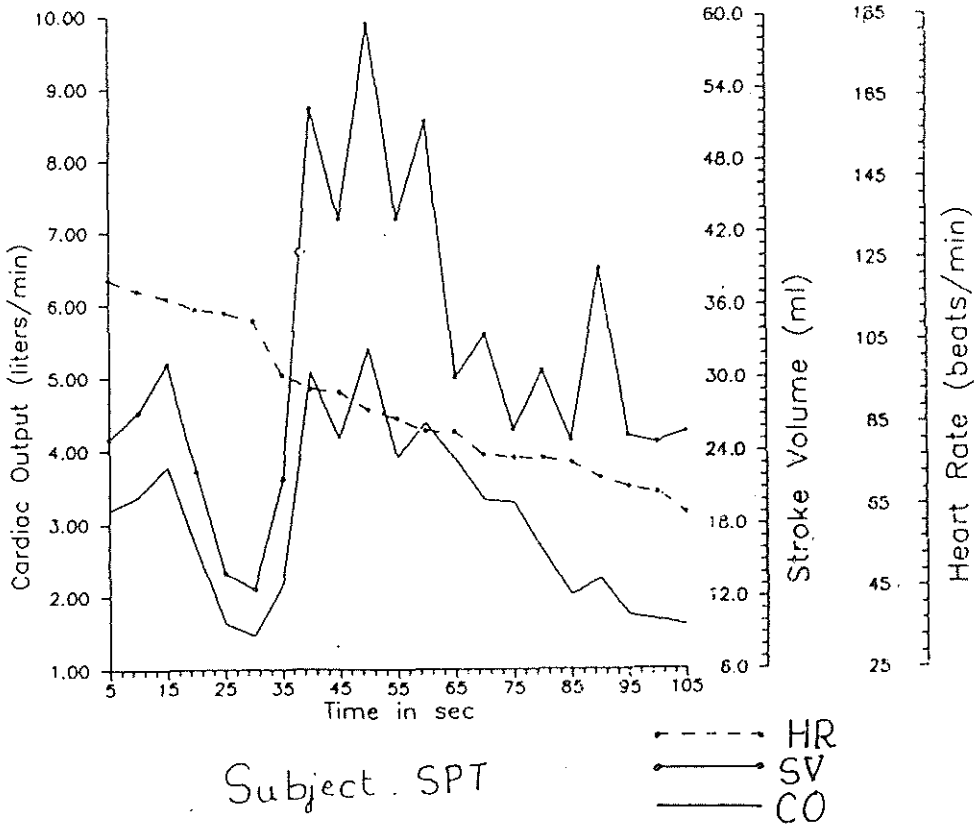


Fig. 6.8 Variation in heart rate, stroke volume and cardiac output of subjects undergoing exercise and then relaxing

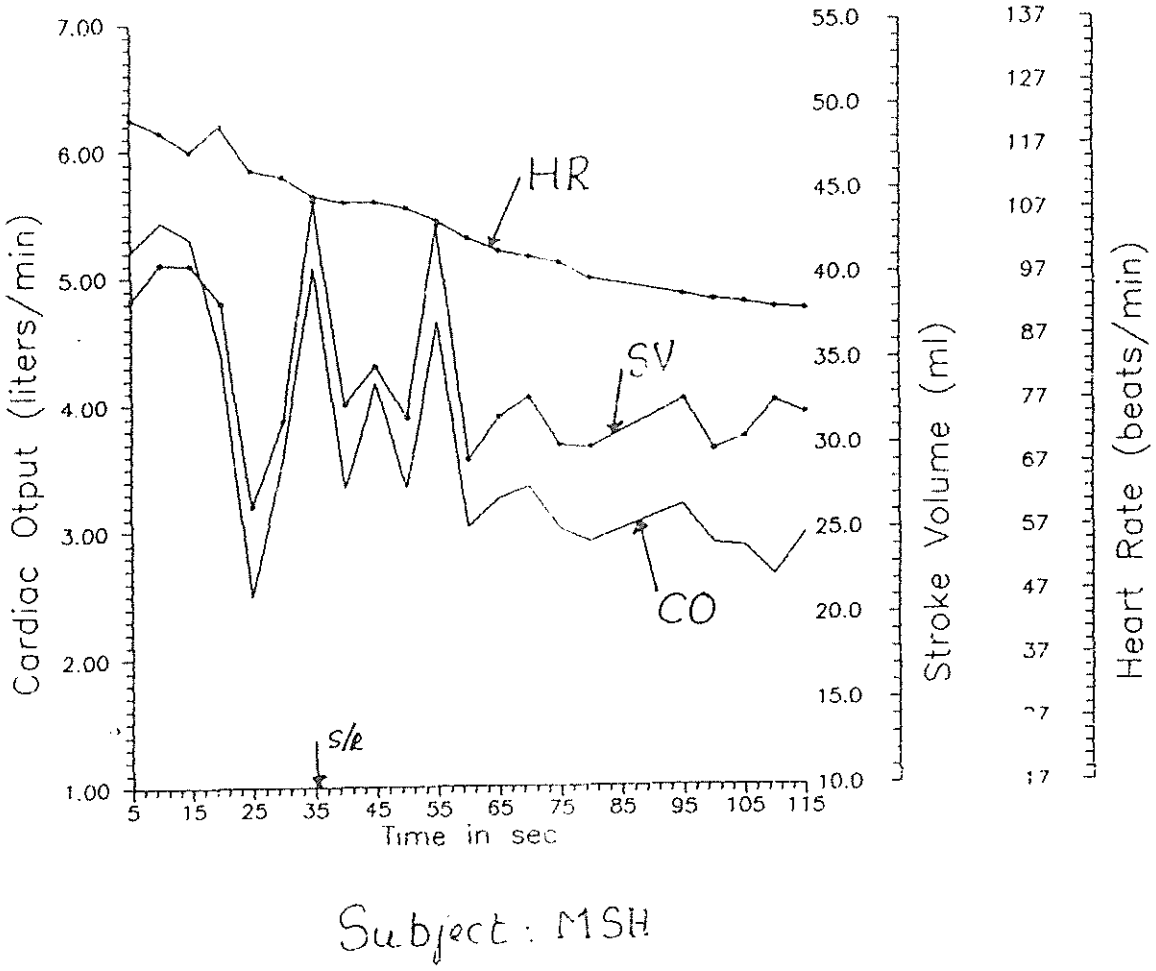


Fig. 6.9 Variation in heart rate, stroke volume and cardiac output of subjects undergoing exercise and then relaxing.

APPENDIX A

Circuit Design

1. P wave circuit [17]

pulse width of monostable M1 is 20 msec

$$T = RC \ln \frac{1+(V1/Vo)}{1-\beta}$$

$\beta = ½$, $V_1 = 0.7V$, $V_o = 9V$

$T = 0.76\ RC$ A.1

$0.02 = 0.76\ R_2 C_1$

$R_2 C_1 = 0.026$

let $C_1 = 1\ \mu F$

$R_2 = 26\ k\Omega \cong 27\ k\Omega$

for C_2 and R_4 $t << T$

$t = 0.015$

$t_p = 2\pi R_4 C_2$

$R_4 C_2 = 0.0025$ for $C_2 = 0.022\ \mu F$, $R_4 = 120\ k\Omega$

Integrator of R_3 C_6

$t_r = 2.2\ R_3 C_6 = 0.03$

$R_3 C_6 = 0.013$ for $C_6 = 10\ \mu F$, $R_3 = 1.3\ k\Omega \cong 1.5\ k\Omega$

2. Q wave circuit

pulse width of M2 is 60 msec

$T = 0.76\ R_3 C_1 = 0.06$

$R_3 C_1 = 0.067$ for $C_1 = 1\ \mu F$, $R_3 = 68\ k\Omega$

$t_p = 2\pi R_4 C_2 = 0.01$

$R_4 C_2 = 0.00159$ for $C_2 = 1\ \mu F$, $R_4 = 1.59\ k\Omega \cong 1.5\ k\Omega$

Differentiator of $R_6 C_3$

$t_p = 2\pi R_6 C_3 = 0.09$

$R_6 C_3 = 0.0143$ for $C_3 = 10\ \mu F$, $R_6 = 1.43\ k\Omega \cong 1.5\ k\Omega$

3. R wave circuit

M3 pulse width is 50 msec

$T = 0.76\ R_3 C_1 = 0.05$

$R_3 C_1 = 0.065$ for $C_1 = 1\ \mu F$, $R_3 = 65\ k\Omega \cong 68\ k\Omega$

$t_p = 2\pi R_4 C_2 = 0.01 << 0.06$
$R_4 C_2 = 0.00159$ for $C_2 = 1\ \mu F$, $R_4 \cong 1.6\ k\Omega$

Integrator of $R_6 C_3$
$t_r = 2.2\ R_6 C_3 = 0.03$
$R_6 C_3 = 0.0136$ for $C_3 = 10\ \mu F$, $R_6 = 1.3\ k\Omega \cong 1.5\ k\Omega$

4. S wave circuit

M4 pulse width is 7 msec
$T = 0.76\ R_3 C_1 = 0.007$
$R_3 C_1 = 0.0092$ for $C_1 = 1\mu F$, $R_3 = 9.2\ k\Omega \cong 10\ k\Omega$

$t_p = 2\pi R_3 C_1 = 0.001 << 0.007$
$R_3 C_1 = 0.000159$ for $C_1 = 0.022\ \mu F$, $R_3 = 7.2\ k\Omega \cong 6.8\ k\Omega$

M5 pulse width is 350 msec
$T = 0.76\ R_7 C_3 = 0.35$
$R_7 C_3 = 0.026$ for $C_3 = 1\mu F$, $R_7 = 2.6\ k\Omega \cong 2.7\ k\Omega$

$t_p = 2\pi R_8 C_4 = 0.0148$
$R_8 C_4 = 0.0149$ for $C_4 = 1\ \mu F$, $R_8 = 1.49\ k\Omega \cong 1.5\ k\Omega$

Differentiator of $R_{10}\ C_5$
$t_p = 2\pi\ R_{10} C_5 = 0.06$
$R_{10} C_5 = 0.01$ for $C_5 = 10\ \mu F$, $R_{10} \cong 1\ k\Omega$

5. T wave circuit

M6 pulse width is 20 msec
$T = 0.76\ R_3 C_1 = 0.02$
$R_3 C_1 = 0.026$ for $C_1 = 1\mu F$, $R_3 = 26\ k\Omega \cong 27\ k\Omega$

$t_p = 2\pi R_4 C_2 = 0.015 < 0.02$
$R_4 C_2 = 0.0025$ for $C_2 = 10\ \mu F$, $R_4 = 2.5\ k\Omega$

Integrator of $R_7\ C_3$
$t_r = 2.2\ R_7 C_3 = 1.03$
$R_7 C_3 = 0.468$ for $C_3 = 10\ \mu F$, $R_7 = 46.8\ k\Omega \cong 47\ k\Omega$

APPENDIX B

Data acquisition card setting

The PCL208 card is an add on card for PC. It has 16 single ended or 8 differential 12-bit A/D input channel. The channel range selection and pacer trigger rate selection is done by means of appropriate software. For the A/D conversion as per required by program ICGREAL.C, following switch position should be set.

The I/O port base address is selectable via an 8 position DIP switch SW6. For HEX address 300, the switch setting are shown in table B.1. The slide switch SW3 controls the selection of analog input configuration. Slide the switch to the left mark 16CH for 16 single-ended inputs. Since the outputs of the cardiac output monitor circuit are bipolar, set the SW2 switch to BIP.

The analog input range selection is made by SW5, a 6 position DIP switch. Since the operating power supply of impedance cardiography is 9V, the maximum signal level can be 9V. Therefore the range chosen is $-^{+}$ 10V. The switch positions are shown in table B.2. The switch SW4 offers two level of DMA transfer capability. Slide SW4 to right side mark "DRO1" for DMA, which is used in the program.

The PCL208 have two clock input frequencies to programmable timer, 10/1 MHz, to generate programmable pulses to trigger the A/D. Slide the switch SW1 to 1 MHz position. Since the output is slowly varying, smaller clock frequency is sufficient. Since no D/A conversion is required, the position of the jumpers JP1 and JP2 are immaterial.

The PCL208 has two 20-pin insulation displacement connector accessible from the rear plate. Of these, connector 1 (CN1) is used for A/D conversion.

Table B.1 Base address selection of PCL 208 card

I/O Address (hex)	switch position					
	1	2	3	4	5	6
	A9	A8	A7	A6	A3	A4
300	1	1	0	0	0	0

Table B.2 Input range selection

switch position					
1	2	3	4	5	6
ON	OFF	OFF	OFF	OFF	OFF

Appendix C

SITE Data logger Information

SITE Data-Logger, is a DMS product with SITE single chip data logger. The data logger basically has 8 A-D channels and 8 programmable digital I/O port. It is 9V battery operated. It has RS-232 serial port communication terminal. All these facilities are available on add on board. The A-D converters are 8-bit successive approximation ADC.

The converter has a sample acquisition time of about 12 microsec. The A-D input range of ground to the positive reference (5V maxi. unipolar). The inputs can be used as pseudo-13 or 15-bit converter inputs using site 'atod 13 and 'atod 15 commands that use a dithering technique which almost eliminates the quantisation noise of the converter. SITE offers trade off between different A-D conversion viz.

Conversion Command	Conversion Rate
atod 15	173 per sec
atod 13	504 per sec
atod 8	1200 per sec

The peak conversion rate is 25,600 conversion per sec.

The communication in data logger is through so called SITE language. Thus SITE language is a new language, borrowing some features from BASIC and others from 'C' and even some from programmable hand calculators. Site language is tailored to data logger task and produces very compact code due to limited program memory of the SITE (255 byte).

The SITE receives inputs from eight analog inputs and two counter inputs. Site language has 16 two byte variable and 1 two byte accumulator. Data output can be made to an LCD, UART or to data storage. Site language breaks data collection into two distinct types of operation : data gathering and presentation. Data gathering commands move data from the SITE's input to variable and presentation commands take what ever is currently resident in a variable and presents it to the UART, LCD or PLOT. For more details about site language refer Dynalog, Might-Micro SITE data logger, User's Manual, Software reference.

The site language is written in the SiteUp package which is installed in Windows. Thus site language is complied in the Windows environment. There is facility to write program in editor in site language and program can be compiled using commands provided by SITE data logger supported by DOS. Some of commands are SC, SITECHK, LAUNCH, RELUNCH, SITEOFF, XSITE, SPLIT, BLAST. All these commands are program supporting site language and are stored as *.exe files.

SC : This program complies a text files produced with an editor and written in the site programming language. The name of file produced with editor must have file extension *.LIT.

SITECHK : It interrogates the selected serial port to see if SITE is on the line. The default port is COM1.

LAUNCH : This program loads the program having *.LIT extension into the SITE.

SITEOFF : Off loads SITE data file to a disk file on PC.

XSITE : This program translates SITE data files into text files.

BLAST : It can be used if SITE hangs up and would not respond to SITECHK

After compiling the site language program data is recorded in the SITE. To see the recorded data one has to off load the data from SITE. The off loaded data is stored in *.dat files in binary mode. For having visual representation of data recorded one has to evoke "Site Show" package which exclusively runs under Windows environment. For more details refer Software reference, User's Manual

APPENDIX D

PCB LAYOUT

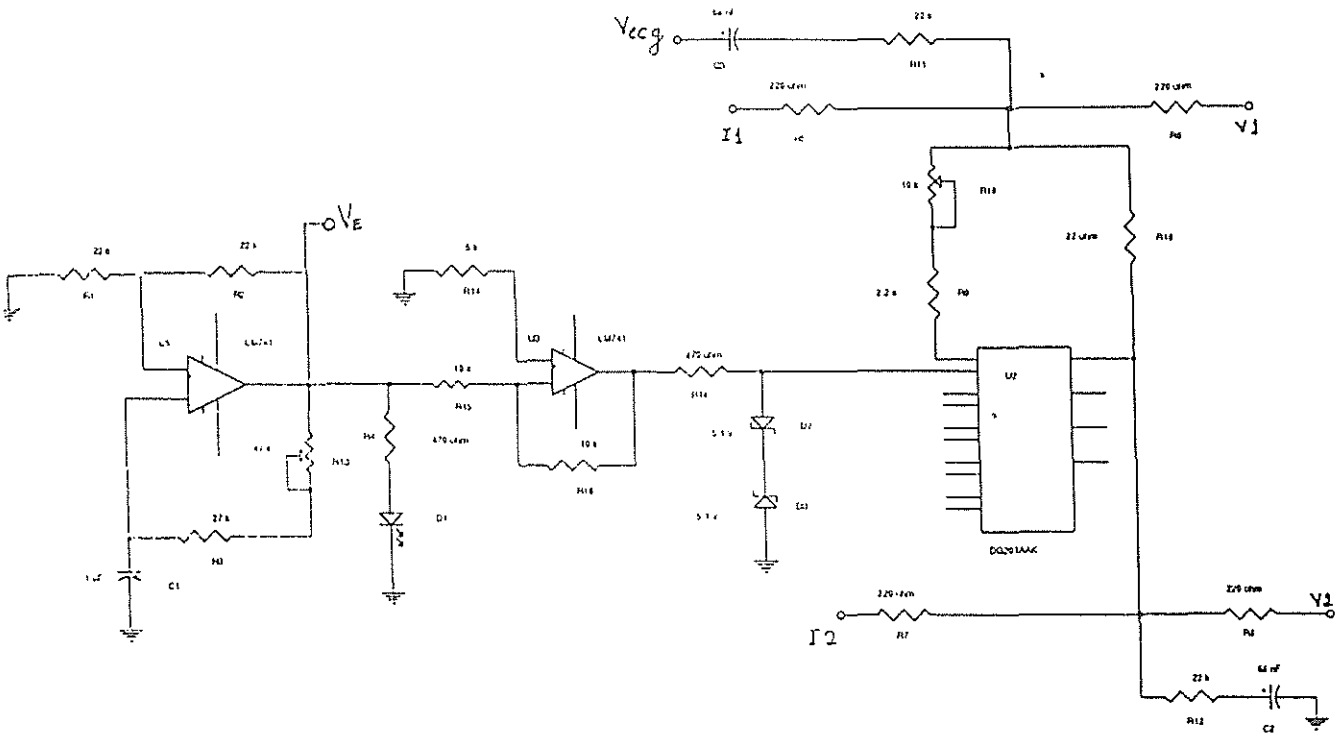


Fig. D.1

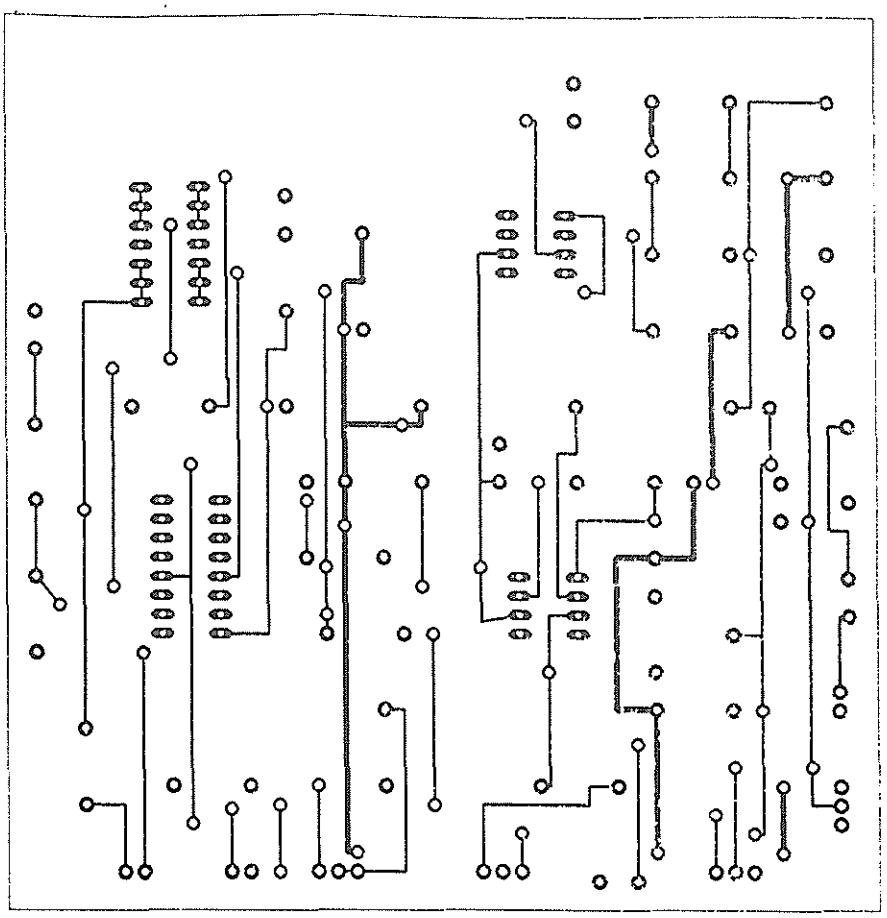


Fig. D. 2.

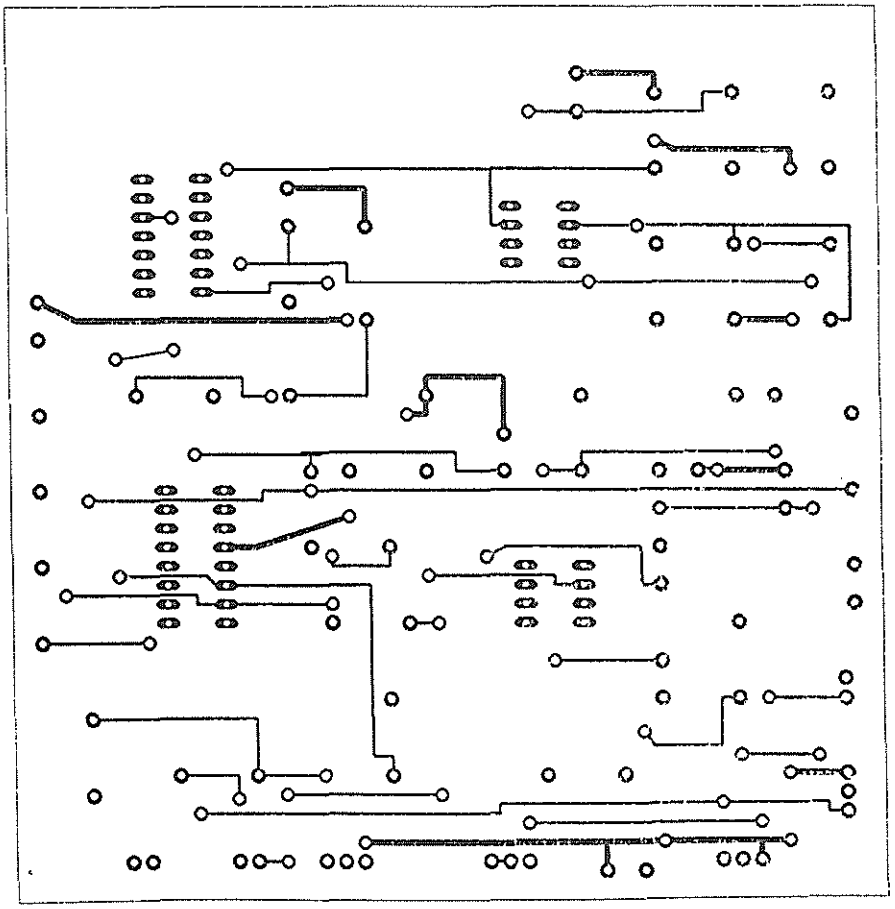


Fig. D. 3

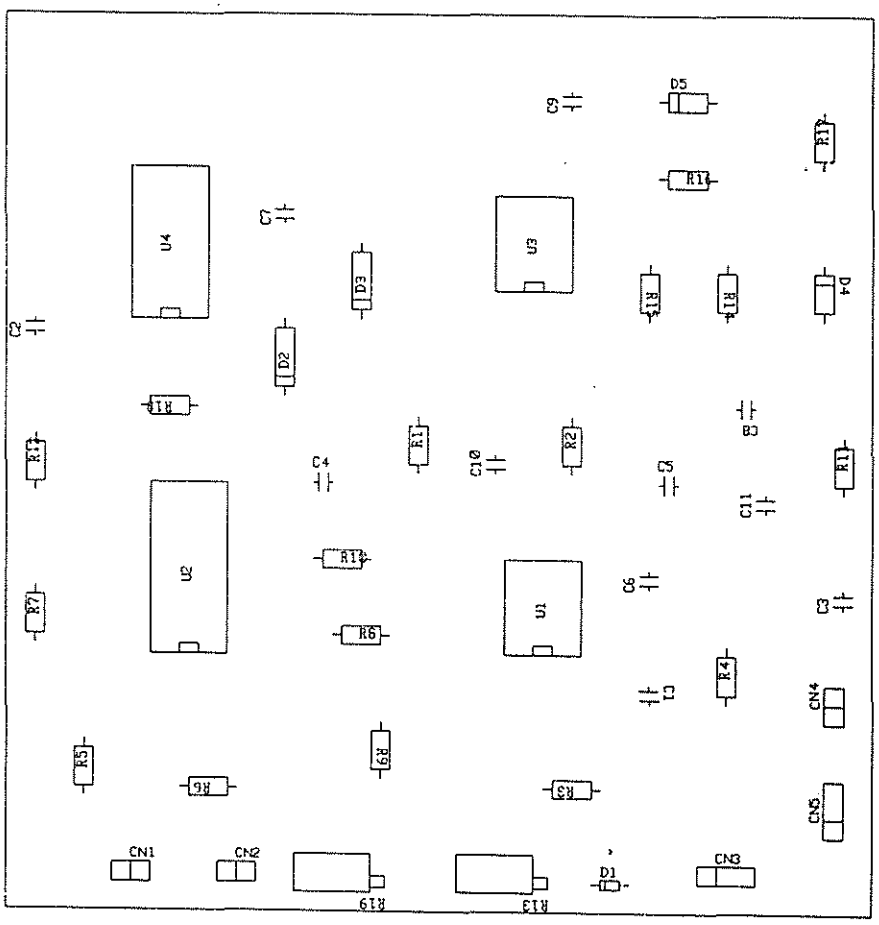
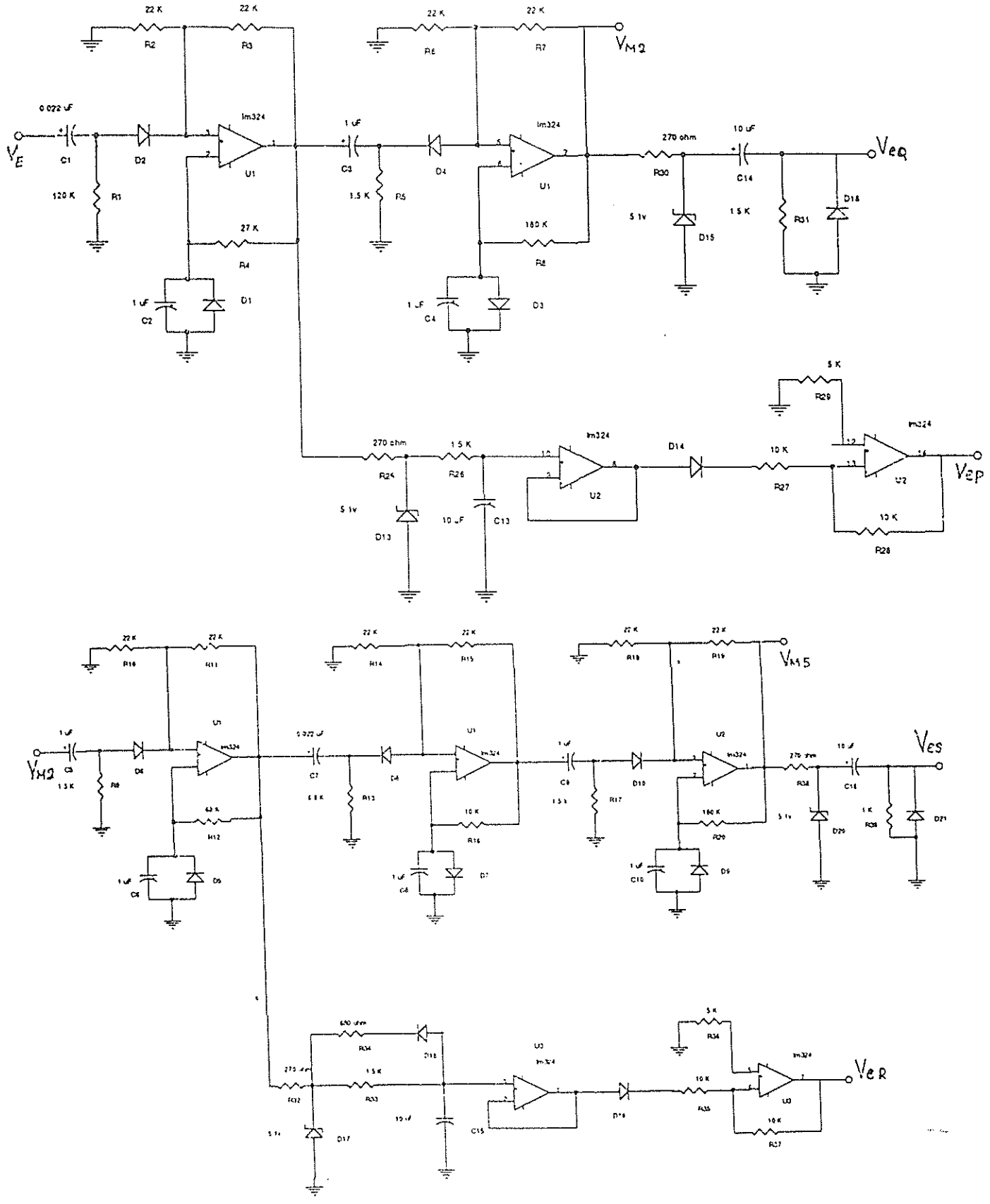
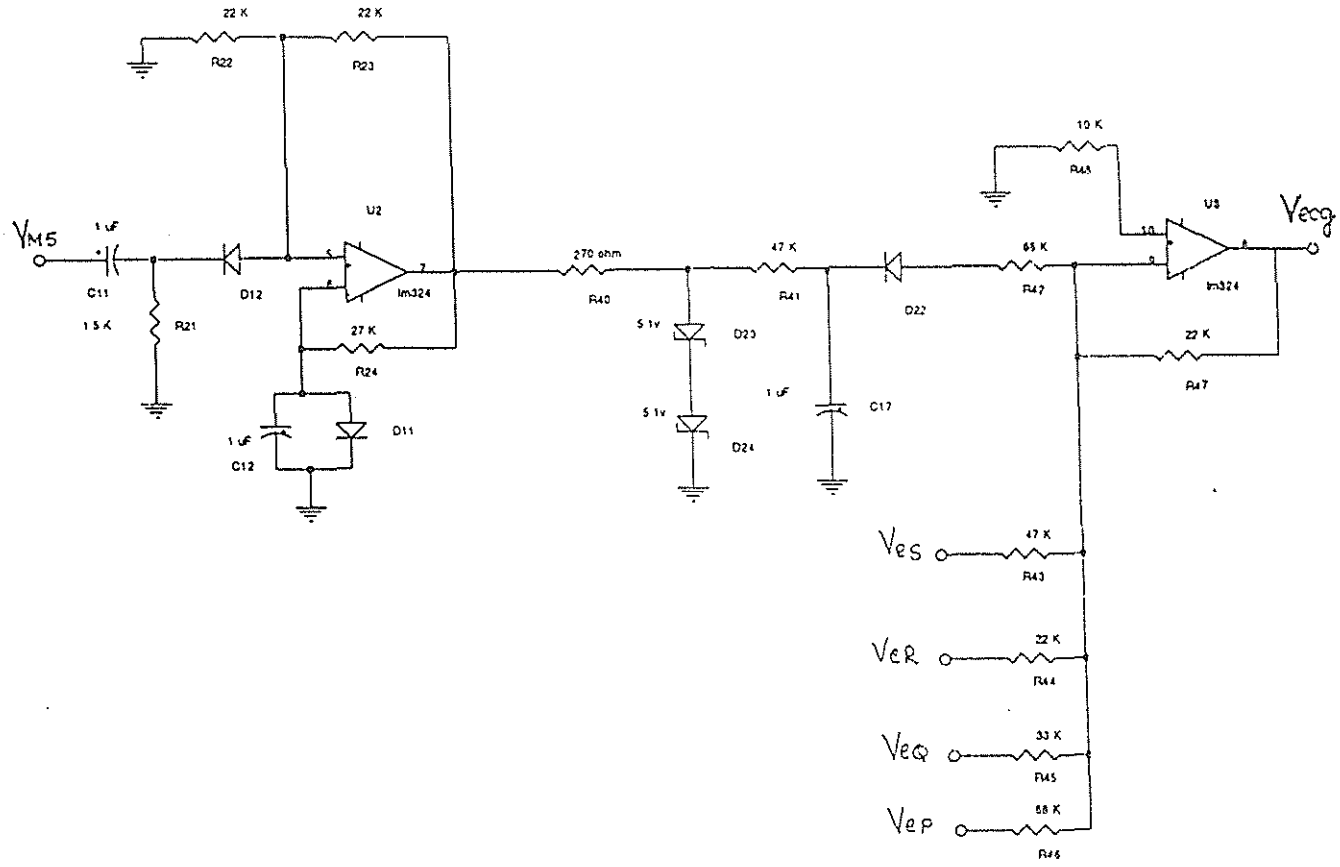

U4
U3
U2
U1
D5
D4
D3
D2
D1
C9
C7
C2
C4
C10
C5
C11
C8
C6
C3
C1
R11
R14
R13
R12
R1
R2
R7
R6
R4
R5
R9
R3
R19
R13
CN1
CN2
CN3
CN4
CN5

Fig. D.4.


22 K
22 K
R2
R3
22 K
22 K
R6
R7
VM2
0.022 uF
C1
VE
D2
lm324
U1
120 K
R1
27 K
R4
1 uF
C2
D1
1 uF
C3
D4
1.5 K
R5
lm324
U1
180 K
R8
1 uF
C4
D3
270 ohm
R30
10 uF
C14
Veq
5.1v
D15
1.5 K
R31
D16
5 K
R29
lm324
U2
270 ohm
R24
1.5 K
R26
lm324
U2
D14
10 K
R27
Vep
5.1v
D13
10 uF
C13
10 K
R28
22 K
22 K
R16
R11
22 K
22 K
R14
R15
22 K
22 K
R18
R19
VM5
VM2
1 uF
C5
1.5 K
R9
D6
U1
lm324
62 K
R12
1 uF
C6
D5
0.022 uF
C7
6.8 K
R13
D4
U1
lm324
10 K
R16
1 uF
C8
D7
1.5 K
R17
D10
U2
lm324
180 K
R20
1 uF
C10
D9
270 ohm
R34
10 uF
C16
Ves
5.1v
D20
1 K
R36
D21
640 ohm
R34
D18
270 ohm
R32
1.5 K
R33
10 uF
C15
5.1v
D17
U3
lm324
D19
5 K
R34
10 K
R35
lm324
U3
10 K
R37
Ver


22 K
22 K
R22
R23
10 K
R48
U3
Vecg
VM5
1 uF
C11
1.5 K
R21
D12
U2
lm324
270 ohm
R40
47 K
R41
D22
65 K
R42
lm324
27 K
R24
5.1v
D23
22 K
R47
1 uF
C17
1 uF
C12
D11
5.1v
D24
Ves
47 K
R43
VeR
22 K
R44
VeQ
33 K
R45
Vep
68 K
R46

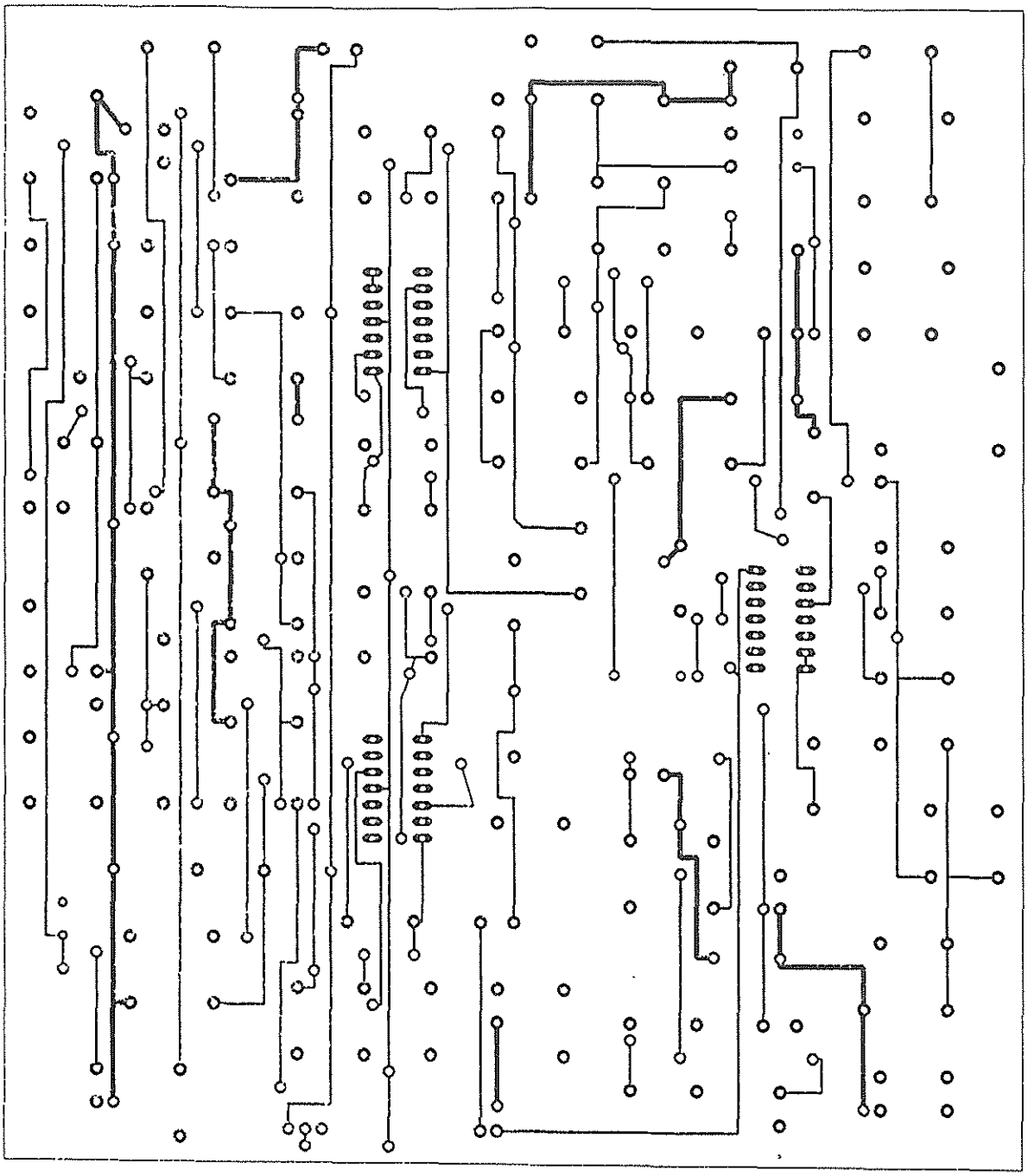


Fig. D.6

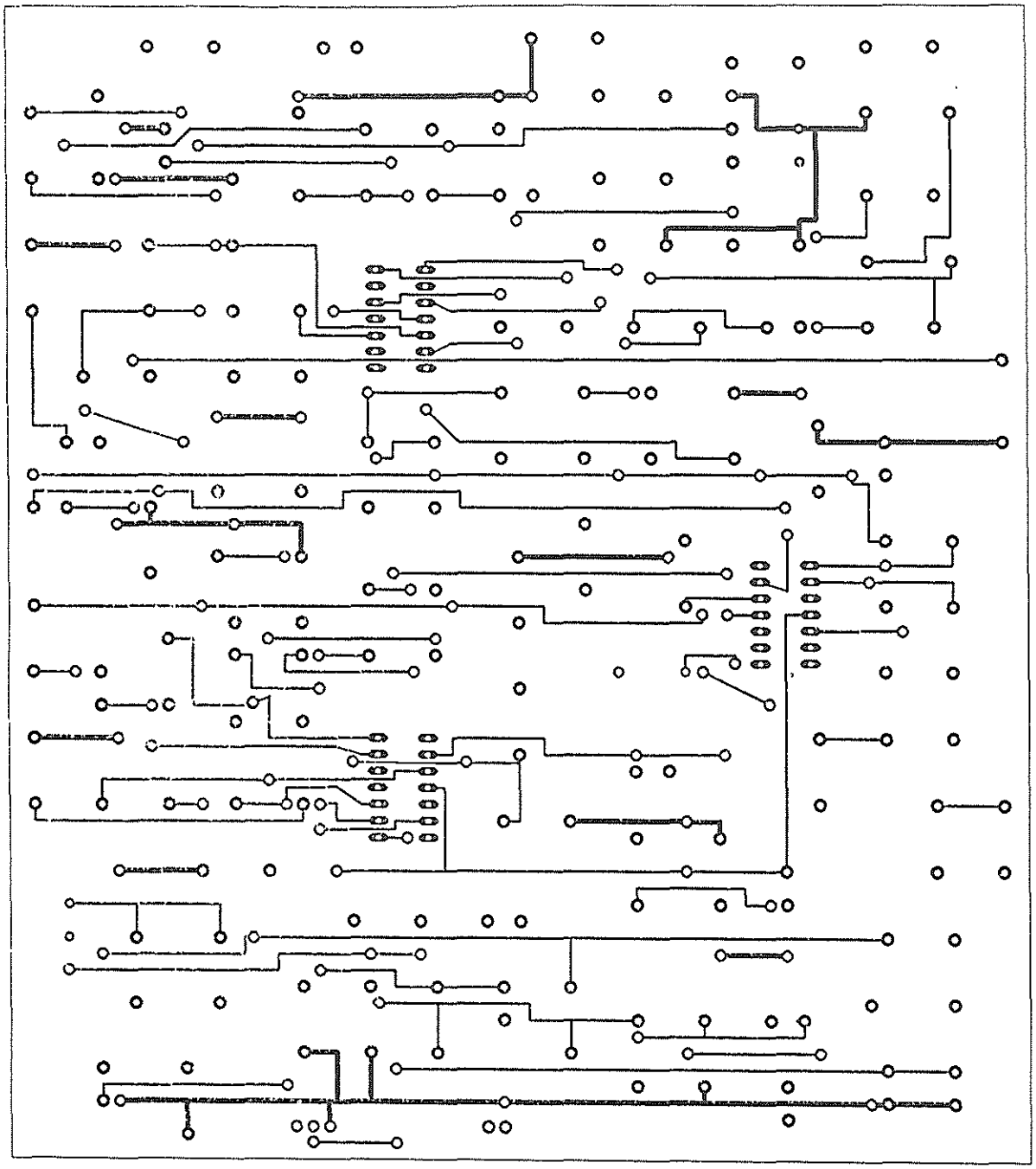
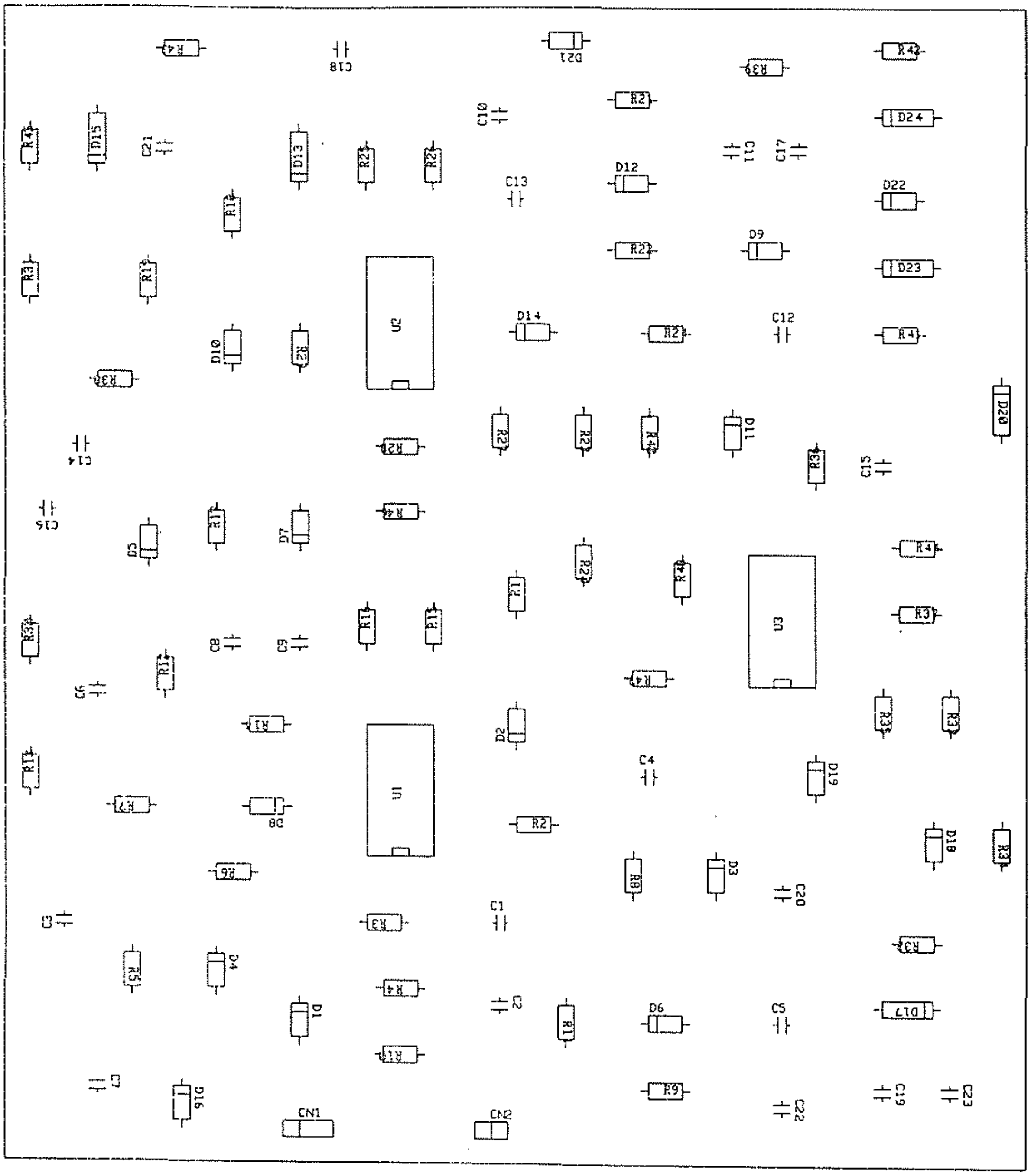


Fig. D.7


U1
U2
U3
CN1
CN2
D1
D2
D3
D4
D5
D6
D7
D8
D9
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D12
D13
D14
D15
D16
D17
D18
D19
D20
D21
D22
D23
D24
C1
C2
C3
C4
C5
C6
C7
C8
C9
C10
C11
C12
C13
C14
C15
C16
C17
C18
C19
C20
C21
C22
C23
R1
R2
R3
R4
R5
R6
R7
R8
R9

Fig. D.8

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