AN IMPEDANCE CARDIOGRAPH FOR STRESS TESTING

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ABSTRACT

The aim of this project is to design an impedance cardiograph system for non-invasively monitoring the stroke volume and cardiac output of a subject during stress testing. The system hardware extracts the required physiological signals, namely, the electrocardiogram, impedance cardiogram and its derivative, and the phonocardiogram (PCG). These signals are then digitized and processed to obtain certain parameters which give information about heart condition during stress testing. The basic system consists of the signal conditioner hardware for extracting the physiological signals and a Personal Computer PC, with PC bus based A/D card for signal acquisition, processing, and off-line display of the recorded signals. In order to make the system portable a notebook PC with PCMCIA bus based A/D card is also provided and can be used in lieu of the mains operated PC. For field use of the instrument, interface with a hand held data logger for digitizing and recording the signals has been provided. The data logger eliminates the need of PC at the data recording site. The signals recorded using the data-logger can be down-loaded into a PC and processed off-line. In order to facilitate easy calibration of the signal conditioning circuitry for extracting the physiological signals, a calibrator has been provided.

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M. Tech Dissertation Approval

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LIST OF ABBREVIATIONS

A/D	Analog-to-digital
CO	Cardiac output
ECG	Electrocardiogram
HR	Heart beat rate
ICG	Impedance cardiogram
PC	Personal Computer
PCG	Phonocardiogram
PCMCIA	Personal Computer Memory Card Interface Association
QRS	The QRS complex in the ECG signal
R	The R point of QRS in ECG signal
SV	Stroke volume
SPA-1	Signal Processing Algorithm 1
SPA-2	Signal Processing Algorithm 2

LIST OF SYMBOLS

ρ	Blood resistivity
$(dz/dt)_{max}$	Maximum value of ensemble averaged dz/dt
L	Distance between the voltage electrodes
T_{base}	The base line crossing point of dz/dt signal
T_{lvet}	Left ventricular ejection time
T_{min}	The first valley point after peak in dz/dt signal
T_{peak}	The peak point in dz/dt signal
Zo	Base impedance of thorax
z(t)	ICG signal

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CHAPTER 1

INTRODUCTION

1.1 PROBLEM OVERVIEW

Many persons have heart disease without being aware of it because their activities are usually well within the normal limits of their cardiac capacity. In these cases physiological parameters recorded during stress testing, which increases metabolic activity and oxygen demand of the cardiovascular system, can serve as valuable tools in assessing the overall cardiovascular fitness of the person. Stress testing unmasks the cardiovascular abnormalities that remain undetected in the resting state [1].

Monitoring the amount of blood pumped by the heart per minute, the cardiac output (CO), during exercise is a very important diagnostic tool [2, 3, 4]. Several instruments meant for this purpose have already been developed and their use is quite common in hospitals outside India. However these instruments are not indigenously manufactured and their use is also not common.

1.2 PROJECT OBJECTIVE

The aim of this project is to develop an instrument for monitoring the amount of blood pumped by the heart per minute (CO) during stress testing, by using the non-invasive technique of impedance cardiography. This technique involves processing the impedance cardiogram (ICG) and its derivative (dz/dt). The impedance cardiogram (ICG) is the recording of the impedance of the thorax during a heart beat. Appropriate signal processing techniques based on a model of the thorax are used for estimating the cardiac stroke volume and this along with the heart beat rate yields the estimate for cardiac output.

The main challenges in the development of the instrument include development of portable hardware for signal acquisition

and implementing the appropriate signal processing techniques. Generally we need electrocardiogram (ECG) and phonocardiogram (PCG) recordings made simultaneously, for cross-checking some of the estimates obtained from ICG waveform. The PCG would provide additional information regarding status of heart valves. The ICG, ECG, and PCG signals are also displayed graphically off-line, since their origins are directly related to the functioning of heart and hence they have a vast diagnostic value [2, 5, 6, 7, 8, 9, 10, 11].

Earlier efforts at IIT Bombay [12, 13, 14, 15, 16, 17, 18, 19] had resulted in a system which could record and process the ECG and ICG of a subject at rest. The processing was done in order to monitor the heart rate, stroke volume, and cardiac output of the subject. A simulator for simulating the ECG and ICG was also developed. This project is a continuation of those earlier efforts and consists of making modifications or redesigning the existing hardware and software, as necessary, in order to improve its consistency and accuracy when used with exercising subjects. Since the recorded physiological signals have a diagnostic value, it is necessary to display the recorded signals for detailed examination by cardiologists. So software has now been developed for off-line display of all the recorded physiological signals.

The basic system consists of the signal conditioner hardware for extracting the physiological signals and a PC with PC bus based A/D card for signal acquisition, processing, and off-line display of the recorded signals. A linear opto-isolator circuit interfaces the PC bus based A/D card with signal conditioner hardware and isolates the PC bus ground from signal conditioner ground. In order to make the system portable a notebook PC with PCMCIA bus based A/D card is also provided and can be used in lieu of the mains operated PC. The battery operated notebook PC does not require the opto-isolator circuit interface. For field use of the instrument, interface with a hand held data logger for digitizing and storing the signals has been developed. The

calibration of the signal conditioner for extracting the physiological signals a thorax simulator circuit has also been developed.

1.3 REPORT OUTLINE

The second chapter describes the technique of impedance cardiography for measuring the cardiac output. The importance and need of recording ECG and PCG are discussed. This is followed by a brief discussion of various cardiac output monitoring techniques including impedance cardiography technique. This chapter concludes with a review of some of the literature on impedance cardiography and instrumentation including the work done earlier on it at IIT Bombay.

Chapter 3 describes the impedance cardiography system developed. The block diagram of the system is explained and the necessity and requirements of the hardware and software developed are discussed. The fourth chapter explains the design details of the hardware and software. The results obtained from using the system developed on exercising subjects are presented in Chapter 5. The last chapter provides summary of the work done and some suggestions for future work.

Appendix A explains the circuit design of the thorax simulator circuit. Appendix B lists the circuit diagram and PCB layouts of the signal conditioner and thorax simulator hardware. Appendix C presents the performance specifications of various parts of the signal conditioner hardware. Appendix D lists the data acquisition card details for PCL-208, a PC-bus based A/D card. Appendix E gives the data acquisition card details for DAQ700, a PCMCIA based A/D card. Appendix F gives the data logger details. The tables of the various physiological parameters monitored with exercising subjects are listed in Appendix G. Appendix H lists the operating instructions for the impedance cardiograph system. The system specifications are provided in Appendix I.

CHAPTER 2

IMPEDANCE CARDIOGRAPHY

2.1 INTRODUCTION

Impedance cardiography deals with measuring the impedance of the thorax with the objective of assessing the functioning of heart. This can also be used to calculate the blood pumped by heart in each stroke and hence cardiac output [2, 10, 20]. Cardiac output is the stroke volume multiplied by the heart rate. Since blood carries oxygen to the body tissues, cardiac output is a major determinant of the oxygen delivery to tissues. There are many ways to establish cardiac output such as, thermodilution and dye-dilution methods, aortic pulse pressure contour method, CO_2 rebreathing method, electromagnetic blood flow measurement method, *etc.* [9, 10]. However these methods are complicated and difficult to repeat. Hence none of the methods are suited for cardiac output measurement during heavy exercise. However impedance cardiography technique is non-invasive and simple.

In this chapter we will first review the working of the heart, and then the ECG and PCG. Then some methods for measuring the cardiac output are discussed. This is followed by detailed discussion of the impedance cardiography technique. Then a review of the literature on development of impedance cardiography technique and instrumentation is presented. This chapter concludes with the summary of the work done earlier at IIT Bombay on impedance cardiography.

2.2 WORKING OF HUMAN HEART

The structure and the course of blood flow in human heart [9, 10, 21] is shown in Fig.2.1a. The heart consists of four chambers, two atria and two ventricles. The cardiac muscles of the atria are separated from those of the ventricles by fibrous

tissues at the atrio-ventricular groove.

The right atrium receives blood returning from the various body organs brought by the inferior and the superior vena-cavae. This blood is pumped by the right atrium into the right ventricle. The right ventricle pumps blood into the lungs via the pulmonary artery for oxygenation. Triscupid valve which is present at the junction of the right atrium and the right ventricle prevents back flow of blood into the right atrium during ventricular systole (compression). After oxygenation, the blood comes via the pulmonary vein into the left atrium from where it gets pumped into the left ventricle. The left ventricle pumps blood into the aorta. This blood gets circulated through all tissues of the body. The mitral valve prevents back flow of blood into the left atrium from the left ventricle during ventricular systole (compression). The inlets of the pulmonary artery and the aorta are guarded by the pulmonary and aortic valves respectively.

2.3 ECG AND PCG SIGNALS

The heart rhythmically beats due to certain impulses (electrical) which are generated in the S-A node of the heart shown in Fig.2.1b. These impulses pass through the atria first, making them contract. This is called the atrial systole. The contracting atria pump blood into the ventricles. After an impulse leaves the atria, the atria relax. This is called atrial diastole. The impulse then enters the ventricles through a special conduction system formed by the bundle of His and the Purkinje fibers. Since it takes time for the impulse to reach the ventricles, they contract after the atria contract [21]. This impulse formation and conduction through the heart muscle give rise to weak electrical currents that spread through the body and can be recorded by proper application of voltage electrodes [11]. Electrocardiogram (ECG) is the graphical recording of these electrical potentials produced in

association with each heart beat. These recordings are immensely useful in the diagnosis of disorders in the electrical activity of the heart [11, 21].

When the heart pumps blood, vibrations are produced [22], due to, disturbances in the blood velocity caused by sudden opening and closing of valves, flow of blood into the atria and ventricles, and turbulence in the valves during the rapid flow of blood. All these vibrations travel to the chest where they can be heard as sound and recorded by a microphone. Phonocardiogram (PCG) is the recording of the sounds made by the pumping action of the heart. The PCG can be used to diagnose the disorders in the activity of heart valves. The PCG also contains information regarding the fitness of the heart muscle [22] and hence serves as an important diagnostic tool. The ECG and PCG signals do not give sufficient information regarding the overall cardiovascular fitness of a person. However this information can be obtained from the variations observed in the HR, SV, and CO, recorded during stress testing [1].

The temporal distribution of ECG and PCG signals as generated in one heart beat is shown in Fig.2.2, [10, 11, 21, 22] The 'P' wave in ECG occurs prior to the atrial systole. This corresponds to the contraction (depolarization) of the atria. The atria contract prior to the ventricles. The rapid contraction of the atrial walls creates a weak sound shown in Fig.2.2, as the fourth heart sound in PCG. After the atria contract and fill the ventricles with blood, the ventricles contract. The QRS complex in the ECG signal occurs at the same time. The first heart sound in PCG is generated by the mitral valve closure and occurs at the onset of ventricular systole. This first heart sound occurs synchronously with the QRS complex in the ECG.

The end of ventricular systole, marked by the second heart sound, corresponds to aortic valve and pulmonary valve closure.

The time difference between first heart sound and second heart sound corresponds to the time for which blood was pumped out of the heart by the left ventricle (T_{LVET}). The 'T' wave in the ECG occurs during ventricular repolarization. During the next heart beat the atria start contracting once again, filling blood in the ventricles. The sound produced due to this rapid filling is recorded as the third heart sound.

2.4 METHODS FOR MONITORING CARDIAC OUTPUT

We will first review some general methods other than the impedance cardiography method for monitoring the blood flow and hence cardiac output. These methods include thermodilution method, dye dilution method, pulse pressure contour technique, CO_2 rebreathing method, and electromagnetic blood flow measurement method [9, 10]. This will be followed by a detailed discussion of impedance cardiography technique.

2.4.1 Thermodilution Method

In this method a catheter is inserted through a vein in the arm and advanced to the right ventricle of the heart. This catheter is used to insert a bolus of cold physiological saline into the pulmonary artery. This cold saline mixes readily with blood thus dropping its temperature. The same catheter also carries a thermistor at its tip which is used to record the temperature of the blood. The initial temperature of the cold saline prior to injecting it into the blood and the profile of the temperature drop of blood after the injection of the saline can be used to calculate the blood flow rate, [9, 10].

The basic source of inaccuracies in this technique arise due to imperfect mixing of the cold saline and blood. Also temperature of the cold saline has to recorded before being injected into the subject. This temperature of the cold saline increases during its flow through the the catheter due to the warm blood surrounding the catheter. This has been reported to

cause errors in the calculated blood flow rate [10]. Also this technique requires catheterization of the subject and hence is not suited for use during stress testing.

2.4.2 Dye Dilution Method

This method is based on the 'Fick Technique' wherein a fixed amount of substance (dye) is injected into the blood flow at some constant rate dm/dt, and the difference between its initial (C1) and final (C2) concentrations can be used to calculate the blood flow rate (F) [9, 10] as given by,

$$F = \frac{dm/dt}{C1 - C2}$$

In this method two catheters are inserted, one through a vein in the arm into the right atrium and other through an artery into the aorta near the left ventricle. The catheter in the right atrium injects the dye at rate dm/dt. The other catheter in the aorta measures the initial (C1) and final (C2) concentrations of the dye by taking a sample of blood for this purpose. The process of determining the dye concentration from blood samples is very complex and involves examining optical density of blood. Also this method is very difficult to repeat continuously over a period of time due to the slow decrease in concentration of the injected dye in blood.

2.4.3 Pulse Pressure Contour Technique

In this method the pressure waveform observed over aorta during ventricular systole is analyzed. During ventricular systole blood is pumped into the aorta increasing the pressure inside it as shown in Fig.2.3. The total blood flowing into the aorta and hence stroke volume of left ventricle is related to the pressure increase inside aorta, the duration of ejection, and the resistance faced by blood flow into the aorta [10] as

given by the relation,

$$SV = \frac{1}{Z_\alpha} \int_{T_1}^{T_2} (P_2 - P_1)\,dt$$

where P_2 and P_1 denote the final and initial pressures inside the aorta, and T_2 and T_1 denote the start and end of ventricular systole. Z_α is the resistance to blood flow presented by the aorta. Changes in the value of the integration shown as area A in Fig.2.3 correspond to changes in blood volume. The aortic pressure waveform is analyzed using catheters inserted into the aorta.

This technique has been reported to be sensitive regarding the place of recording the pressure inside the aorta and small deviations from the correct place of recording can introduce inaccuracies [10]. This technique is invasive as catheters are to be inserted into the aorta to record pressure.

2.4.4 Co_2 Rebreathing Method

This method is a indirect 'Fick Technique' wherein the subject is made to rebreathe in an air bag which contains an appropriate concentration of Co_2 in oxygen. The concentration of Co_2 in the bag is varied by varying the supply of oxygen to the air bag till there occurs a point at which there is no net change in the Co_2 concentration in the air bag. This point denotes that the oxygen consumed by the lungs is equal to the oxygen supplied to the air bag. The concentration of Co_2 in the air bag is now equal to the concentration of Co_2 in blood entering the lungs. By using this Co_2 concentration and a Beckman metabolic chart the cardiac output can be calculated [9].

This procedure though non-invasive, is time consuming and cannot be used to calculate the instantaneous stroke volume.

2.4.5 Electromagnetic Blood Flow Measurement

This is a direct method to measure blood flow in intact blood vessels. This method is based upon Faraday's law which states that when a conductor is moved at right angles through a magnetic field in a direction at right angles to both the magnetic field and the conductor length, an emf gets induced in the conductor.

In this technique an electromagnetic assembly provides a magnetic field at right angles to a blood vessel through which the flow is to be measured. The stroke volume of heart, can be calculated by measuring the blood flow in aorta. This blood vessel has to be exposed so that the electromagnetic assembly can be put over it. The blood in this vessel represents a moving conductor and an emf gets induced in the blood stream. The magnitude of the emf is recorded by placing electrodes over the blood vessel and gives the blood flow rate [10]. Although this method is very accurate, it is invasive as it requires exposing the blood flow vessel.

The methods discussed until now for blood flow measurement were invasive except for the CO_2 rebreathing method. Also they are very difficult to repeat and hence not suited for use during stress testing.

2.5 IMPEDANCE CARDIOGRAPHY TECHNIQUE

This method is based upon impedance plethysmography which measures the impedance of a body segment. The body impedance is measured by passing a current through it and measuring the voltage developed. In particular the thoracic impedance is required in this technique, and is measured by passing a small ($\leq$ 5 mA to avoid tissue damage) high frequency (20 - 200 kHz) current through the thorax and measuring the voltage developed [2, 20, 23]. At frequencies greater than 20 kHz the nerves and muscles cannot be stimulated however at frequencies greater

than 200 kHz most of the excitation current passes through the body surface. A general scheme as used by many researchers for measuring the thoracic impedance is shown in Fig.2.4. The electrodes I1 and I2 represent current electrodes and pass a constant current through the thorax. The resultant voltage developed is sensed by voltage electrodes V1 and V2. The electrodes used are band electrodes which encircle the body. The position of the electrodes is important as it decides which part of the thorax contributes significantly to the recorded impedance [23, 24, 25].

The waveform recording the varying impedance of thorax is known as the impedance cardiogram (ICG). It has been reported that ICG can be used for estimation of stroke volume in subjects not suffering from certain cardiovascular diseases like left to right shunts, valvular insufficiency, isolated valvular disorder, septal disorder, *etc.* [2, 5].

2.5.1 Basis for Impedance Cardiography

During systole, there is a decrease in thoracic impedance. It was thought [26, 27] that the increase in blood volume of thorax was the reason of this impedance decrease. During ventricular systole there is an increase in the blood entering the lungs. This blood pumped by the left ventricle represents the stroke volume and was taken as the the dominant factor contributing to the systolic impedance decrease [26, 27]. This was investigated by Kubicek and based on his parallel column model of thorax he derived a formula for stroke volume [26].

2.5.2 Kubicek's Method

Kubicek proposed a parallel column model [26] of the thorax shown in Fig.2.5 which consists of a conducting material with impedance Z_0 which is paralleled by a column of uniform cross sectional area A, with length L and effective resistivity ρ. The impedance Z_0 represents the base impedance of the thorax

and the parallel column models the lungs. Now if the cross sectional area of the parallel column varies from zero to a finite value producing an impedance change ΔZ in $Z(t)$, then the the corresponding change in volume of the column is given by

$$\Delta V = \rho \left(L^2 / Z_0^2 \right) \Delta Z$$

where ΔV is the change in volume.

This model assumes that blood pumped into the lungs, which causes increase in its volume during ventricular systole is the the primary source of thoracic impedance decrease. However since blood also leaves the lungs starting from mid ventricular systole the recorded impedance decrease at any time arises only due to a part of the blood pumped by the right ventricle. To account for this outflow of blood from lungs, a forward extrapolation technique originally developed by Patterson [27] is used.

2.5.3 Modified Kubicek's Method

In this technique [27], it is assumed that during the initial systole blood only enters the lungs and causes decrease in impedance and that during the entire systole the rate of blood flow into the lungs is approximately constant. So the initial rate of decrease of impedance during ventricular systole is obtained which corresponds to the peak $dz/dt_{(max)}$, in dz/dt waveform shown in Fig.2.6. The ΔZ is extrapolated from this peak over the entire duration of ventricular systole, the left ventricular ejection time, T_{lvet} as,

$$\Delta Z = \left(dz/dt_{(max)} \right) T_{lvet}$$

where, ΔZ is total impedance decrease that would have arised in absence of out-flow of blood from lungs during ventricular systole.

Thus the modified Kubicek's equation for computing stroke volume (SV) is given by,

$$\Delta V = \rho \left(L^2 / Z_0^2 \right) \left(dz/dt_{(max)} \right) T_{lvet}$$

The cardiac output is calculated as the product of heart rate and stroke volume.

2.6 DEVELOPMENT OF IMPEDANCE CARDIOGRAPHY

In this section a review of some of the work done on impedance cardiography is presented. This section will briefly cover the impedance cardiogram and its applicability for diagnostic purposes. Some of the impedance cardiograph systems developed and reported in literature are also reviewed. We will first discuss the impedance cardiogram.

When the heart beats there is a redistribution of blood in thorax, e.g. during ventricular systole volume of blood in left ventricle decreases and that in aorta increases. This changes the impedance of the thorax. Anderson et al. [5] concluded from their experiments that impedance cardiography recorded only those changes in resistivity of thorax that occur between the voltage electrodes V1 and V2 shown in Fig.2.4. In 1988 Kim et al. [2] confirmed that that impedance change as noted between the voltage electrodes chiefly occurs during ventricle systole and contribution to it from change in blood volume of heart is negligible. They observed that the insulating effect of the heart muscle due to its higher resistivity prevents changes in heart volume from changing the thoracic impedance appreciably. In 1985 Patterson [28] has shown that during systole there is a decrease in impedance and also that for a given blood volume change in ventricles, decrease in impedance produced by the aortic segment is much greater than the increase in impedance shown by the ventricles.

In the following subsections the applicability of impedance

cardiogram for determination of cardiac output and for clinical purposes is presented.

2.6.1 The Minnesota Impedance Cardiograph

In 1974 Patterson et al. [2], developed an instrument called *The Minnesota Impedance Cardiograph.* In this system ECG, ICG, dz/dt and PCG were recorded during exercise. The electrode placement was done as shown in Fig.2.4. A constant 4 mA rms current at 100 kHz was passed through the current electrodes. The thoracic impedance of the subject between the voltage electrodes V1 and V2 (refer Fig.2.4) was recorded and was processed to calculate the change in thoracic impedance in each beat ~~cycle~~. The ICG was differentiated to get the dz/dt. The PCG was recorded using a microphone connected to the patient's chest. The stroke volume and cardiac output were calculated from dz/dt using the modified Kubicek's formula for stroke volume.

They used the system on normal and diseased subjects in order to provide the physician with stroke volume and cardiac output. They also investigated the effect of various diseases on ICG waveforms particularly in cases which are very difficult to detect using just ECG. They observed that certain diseases like acute myocardial infarct and left ventricular enlargement show an easily detectable abnormality in the z(t) and dz/dt during stress testing but fail to produce any change in the ECG.

2.6.2 Kim's Thorax Model

In 1988, Kim et al. [31] modeled the thorax by using a three dimensional finite element model. Their model consisted of small cubes which together form the thorax and neck. The resistivity of each cube depends on the resistivity of particular tissue or muscle that it represents. When the heart beats, the aorta, ventricles *etc.*, change in volume. To simulate this the number of cubes that make up the aorta,

ventricles, pulmonary circuit etc. also change. Those cubes that are inside the aorta, ventricles, pulmonary circuit etc. have the resistivity of blood. Thus by varying the size of different organs as they vary during each heart beat, the impedance change as measured by the voltage electrodes is simulated. The results of their simulation indicated that

1) The impedance change as measured by the voltage electrodes was linearly related only to the blood volume change in aorta among all other origins of impedance change.
2) Although the blood volume change in the ventricles is not linearly related to the thoracic impedance change its contribution to the thoracic impedance change is among the largest.
3) The contribution to total impedance change due to blood perfusing into the lungs is much smaller compared to the first two effects.
4) The contribution of blood resistivity change to total change in thoracic impedance is approximately 40% of that caused due the aortic expansion.

However it should be noted that all the above mentioned sources of impedance change occur at various intervals during cardiac cycle and effect (2) does not interfere in Kubicek's method of calculation of blood volume [31].

2.6.3 Monitoring CO during Treadmill Exercise6

Another instrument for monitoring cardiac output during treadmill exercise was developed by Zhang et al. [3, 23]. Their system obtained the ECG, ICG and dz/dt of the subject during the treadmill test. The ICG of the subject was obtained by rectifying and filtering the carrier signal between voltage electrodes. The changing thoracic impedance component was then extracted from the ICG and differentiated to get the dz/dt. These signals were then digitized and processed using a Z80 based microprocessor system. The ECG was processed for

detection of QRS complexes and heart beat rate. The ensemble average of dz/dt in each heart beat for 64 beats was then calculated to minimize noise. Stroke volume was then computed using Kubicek's modified formula. The cardiac output was also calculated from heart beat rate and stroke volume.

They tested the system on normal subjects with the aim of comparing the cardiac output as obtained from impedance cardiograph system with that obtained from CO_2 rebreathing method and found that both the techniques show good correlation. They also suggested and tested a new four-spot electrode array method shown in Fig.2.9. In this arrangement electrode 1 is placed on the back of the neck behind cervical vertebra C4 and electrode 4 over vertebra T9. Voltage electrode 2 is placed 4 cm. above the clavicle on the front of the neck and electrode 3 is placed over the sternum at the fourth rib. This method was found to introduce less noise compared with the conventional band electrode array method, shown in Fig.2.4, especially in recordings taken during exercise.

2.6.4 Monitoring Air Volume of Lungs Using ICG

The investigation of the physical state of thorax was carried out by Vachogiannis et al. [25] in 1988. They developed a instrument for recording the variation in thoracic impedance and based on a model of human thorax explained the impedance change as a function of the air that enters or gets exhaled from the lungs. The principle for detection of thoracic impedance was the same as discussed until now. Their instrument directly obtained the impedance change of thorax by multiplying the impedance modulated carrier between the voltage electrodes with the excitation oscillator voltage and low pass filtering the multiplier output. They also extracted the ICG by rectifying and low-pass filtering the impedance modulated excitation carrier between voltage electrodes.

They showed that the air volume of lungs obtained from

processing the thoracic impedance show a good agreement with that obtained from a spirometer (an instrument to record air volume of lungs) and that their instrument had a diagnostic value as it could be to detect the non-uniformities of thoracic impedance arising due to plasma in the lungs.

2.6.5 Use of ICG for Medical Diagnosis

In India a system was developed by Jindal et al. [5, 6, 7, 8] at the Bhaba Atomic Research Centre, Trombay (BARC) and used extensively on normal and diseased subjects to explore the advantages and limitations of using the ICG as a diagnostic tool. They found that with subjects suffering from cardiovascular diseases, impedance cardiography provides an effective and inexpensive screening procedure to other diagnostic tools like, 2D-echo, cardiac-catheterization, and angiography. They also showed that the impedance cardiograph system cannot be reliably used in subjects suffering from certain heart ailments like septal disorder which introduce a deformity in the impedance cardiogram signal.

2.7 CALIBRATOR FOR THE IMPEDANCE CARDIOGRAPH.

The impedance cardiography technique for monitoring stroke volume and cardiac output requires extraction of the subject's ECG and ICG. The ICG is differentiated to get dz/dt. The hardware required for all this must be calibrated to guarantee accuracy of results. A calibrator is a device which simulates the working environment of the circuit to be calibrated and can be used for its calibration as well as for fault finding purposes. We will first discuss the general working of the hardware for extracting the ECG and ICG with reference to electrode placement scheme shown in Fig.2.4 and then two different calibrator schemes for calibrating it.

The ECG extraction hardware extracts chest ECG of the subject between electrodes V1 and V2. The voltage between

electrodes V1 and V2 consists of the chest ECG riding over the thorax impedance modulated excitation signal from the current electrodes I1 and I2. The chest ECG which is a low frequency signal compared to the excitation carrier, can be extracted by low pass filtering the voltage between V1 and V2 and then amplifying it. The ICG between electrodes V1 and V2 can be extracted by removing the ECG first by high pass filtering the voltage between V1 and V2 and then demodulating the excitation carrier modulated by the thoracic impedance.

One way of designing a calibrator is to artificially generate the voltages waveforms between electrodes V1 and V2 and present it to the ECG, and ICG extraction hardware. In the other method the thorax of the subject is simulated by a model. This model is used in place of the actual subject.

A scheme for calibrating the ICG extraction hardware and the differentiator following it to get the dz/dt was published by Jindal [29]. In this scheme the voltage waveform between electrodes V1 and V2 is artificially generated. This is done by modulating the excitation signal from electrodes I1 and I2 by a triangular wave and presenting this modulated signal to the inputs V1 and V2 of ICG extraction circuit. With this input the ICG extraction circuit produces a triangular wave riding over a dc as ICG which is fed to the differentiator to get the dz/dt. The dz/dt is a square pulse. The amplitude of the triangular wave and that of the square pulse can be used to calibrate the ICG extraction circuit and the differentiator respectively.

Another scheme for calibrating the ICG extraction hardware was tried out and used at IIT Bombay and will now be discussed. In this scheme the impedance of the thorax is simulated and this thorax simulator replaces the subject. The circuit for the thorax simulator is shown in Fig.2.7. This thorax simulator is partly based on the Kubicek's parallel column model of thorax shown in Fig.2.5. In the subsequent subsections this existing thorax simulator circuit and some modifications as required in

it are explained.

2.7.1 Thorax Simulator

In Kubicek's thorax model [26] the basal thorax impedance is represented by Z_0 and the parallel column of varying volume represents the lungs whose volume and hence impedance change during each heart beat. The thorax simulator circuit shown in Fig.2.7 is partly based on the Kubicek's parallel column model of thorax. In this circuit resistance Rb represents the basal thoracic impedance and the changing thoracic impedance is generated by closing switch S1. Closing of switch S1 simulates the decrease in thoracic impedance found during ventricular systole. The impedance decrease so obtained is of a switching type and not a gradual one. The resistors R1, R2, R3, and R4 represent the electrode-tissue contact resistances. This circuit was built on a PCB by Survase [19] in 1995. He also designed a circuit for simulating the ECG signal. The simulated ECG was added to the excitation carrier. The setup for interconnecting the thorax simulator and signal conditioner is shown in Fig.2.8. The thorax simulator simulates the thoracic impedance between electrodes V1 and V2 (refer Fig.2.4) which modulates the 100 kHz excitation carrier supplied from terminals I1 and I2. It also simulates and adds an ECG signal to the carrier. This signal is now given at the voltage electrode inputs V1 and V2, of the ECG and ICG extraction hardware.

2.7.2 Modifications Required In The Thorax Simulator

The existing thorax simulator circuit had a few drawbacks. During ventricular systole there is a decrease in thoracic impedance ΔZ, which was simulated by a step change (decrease) in the impedance value. However this does not truly approximate the actual impedance decrease waveform found in the ICG. The ECG simulator lacked proper timing relationship between various

segments in the ECG. There was also a lack of proper synchronization between R point in ECG and the initiation of the ΔZ waveform. So the thorax simulator had to be redesigned. In the redesigned thorax simulator, the thoracic impedance decrease during systole is a triangular waveform. This impedance change is generated approximately 30 ms after the R point in the simulated ECG. The ECG simulator circuit was suggested by Pandey [30]. The block diagram of the redesigned thorax simulator is explained in the next chapter.

2.8 Development of Impedance cardiograph At IIT Bombay

Since 1990 efforts have also been initiated at IIT Bombay under the guidance of Dr. Pandey to develop an impedance cardiograph system for measuring the CO and SV of resting as well as exercising subjects. The aim was to build a PC based instrument for carrying out signal acquisition and processing of the required physiological parameters. Bendre [12], Nagvenkar [13] and Patil [14] worked on a circuit for monitoring the thoracic impedance of a subject based on the technique developed by Qu et al. [23]. This circuit was further refined and successfully tested by Joshi [15] in 1993. He modified the DC cancellation circuit and built the whole circuit on a PCB. He implemented a linearised opto-isolator interface for interfacing the impedance cardiograph hardware with a PC and developed software for calculating the SV and CO [15, 17].

The system at this stage consisted of hardware for obtaining the ECG and its derivative de/dt, Zo , and dz/dt of the subject interfaced to a PC with data acquisition interface for digitization and software processing of the digitized signals. In 1994 the opto-isolator interface was critically studied and tested by Lakdawalla [16]. Ingle [18] in 1994 completed many trials of the system developed till then, on subjects and proved the workability of system. In 1995 Survase [19]

developed a thorax simulator by incorporating an ECG simulator in the earlier developed circuits.

CHAPTER 3

IMPEDANCE CARDIOGRAPH SYSTEM

3.1 INTRODUCTION

An impedance cardiography system has been designed to monitor the cardiac functioning of a subject during stress testing. The system developed extracts the ECG and ICG signals and processes them to calculate the heart rate (HR), left ventricular ejection time (T_{lvet}), base impedance of thorax (Z_0), stroke volume (SV) and cardiac output (CO). All these parameters are calculated and displayed on-line and the recorded signals are stored for off-line display and processing. The system also provides an option for recording the PCG of the subject. This PCG is displayed off-line for critical diagnostic examination.

The aim of this project was to redesign the existing impedance cardiograph system for use with exercising subjects. The software processing of the ECG, ICG and dz/dt has been critically studied and modified to work on exercise recordings. The system essentially consists of the signal conditioner hardware, opto-isolator interface, and a PC with A/D card for signal conditioning and processing. The system designed lacks portability due to the mains operated PC and a hence a notebook PC has now been been provided as a substitute. Since most of the notebook PCs have PCMCIA bus slots for connecting external cards, an interface with a PCMCIA bus based A/D card has been developed.

In order to allow the recording of ECG, ICG, dz/dt, and PCG signals during outdoor exercise, an instrument is required that can be easily carried along. In this case it is not important for the instrument to give an on-line reading of cardiac output as the cardiac output results can be interpreted only after completion of the exercise. Since a PC for recording the data becomes very inconvenient because of its bulkiness a data logger interface has been provided which eliminates the presence of a PC

at the data recording site. The data logger is a small handy battery operated unit and can digitize and store data over an interval of time. The recorded data can then be down-loaded into any PC, which could be far from the site of recording, and processed off-line as per convenience. When the battery operated notebook PC or data logger is used the opto-isolator interface is not required.

A block diagram of the entire system is shown in Fig.3.1. The entire system consists of the signal conditioner unit, the linear opto-isolator unit, the signal acquisition and processing unit and the thorax simulator unit.

Each sub-block of Fig.3.1 is briefly explained in the following sections. The design aspect of each sub-block is explained in the next chapter.

3.2 SIGNAL CONDITIONER CIRCUIT

This unit consists of hardware for extracting the ECG, ICG, and PCG signals. The ICG is also differentiated to get the dz/dt signal. The block diagram of the signal conditioner unit is shown in Fig.3.2. The four patient leads interface the subject with the hardware as shown in Fig.2.9. Electrodes 1 and 4 pass a 100 kHz, low amplitude ($\leq$ 5 mA rms) current through the subject. The voltage between electrodes 2 and 3 consists of the 100 kHz sine wave modulated by the changing thoracic impedance and riding over the subject's ECG.

The ECG of the subject is extracted by removing the 100 kHz carrier between electrodes 2 and 3. This ECG is amplified, high pass filtered to remove the baseline fluctuations arising due to improper contact between electrodes and skin [10], and then low pass filtered to suppress the 50 Hz and other high frequency noise. The low pass filtering also serves as an anti-aliasing filter prior to sampling. The circuit design details are explained in next chapter.

The thoracic impedance is extracted from the 100 kHz component

present between electrodes 2 and 3. The 100 kHz component is amplified and then demodulated (rectified and low pass filtered) to get the Z(t). This signal consists of ΔZ, a low frequency changing thoracic impedance component, riding over the basal thoracic impedance Z_0, and a respiration component. This respiration component arises because thoracic impedance increases during inhalation and decreases during exhalation [23]. The Z(t) is severely low pass filtered to obtain the base impedance Z_0 and then digitized. The changing thoracic impedance component ΔZ, is extracted using a special DC cancellation circuit. The respiration component cannot be removed by a simple RC high pass filter as the lower band limit of the z(t) signal ($\leq$ 0.5 Hz) overlaps the spectrum of the respiration component (2 Hz-0.2 Hz). This ΔZ signal is then differentiated to get dz/dt. This dz/dt is now digitized. The Z_0 and dz/dt are digitally processed and used for calculation of stroke volume. The circuit details of signal conditioner hardware are presented in next chapter.

To record the phonocardiogram an electronic stethoscope "STETHMATE" from Electronic Engineering Corporation has been used. It has a chest piece which encloses a microphone. The signals picked up by the microphone are amplified and filtered. The instrument has both audio as well as electrical output. The electrical output is amplified and then digitized. Though a microphone is not the best transducer for recording PCG it is used as other transducers like light-weight accelerometer, which are better suited for this purpose are not readily available.

3.3 OPTO-ISOLATOR CIRCUIT

Until now the signal conditioner hardware for extracting the physiological signals was discussed. This hardware is battery operated. This is done because there always exists a small leakage current flowing between the primary and secondary of the step down transformer which is used to power circuits from mains supply. With the hardware mains operated, this small leakage

current (ground fault current) flowing from the hardware circuit ground to the subject, who is electrically grounded can be fatal [10]. The output of the signal conditioner is digitized by a mains operated PC. In order to electrically isolate the signal conditioner and PC grounds an opto-isolator circuit has been used. The opto-isolator circuit developed by Lakdawalla [16] has been used unmodified in this project.

3.4 SIGNAL ACQUISITION AND PROCESSING UNIT

The ECG, Z_0, dz/dt, and PCG are digitized by a computer interfaced with a A/D card. The mains operated PC uses a PCL208 A/D card from Dynalog Microsystems. For the notebook PC a PCMCIA bus based DAQ700 A/D card from National Instruments has been used. The software processing of the signals is the same for both types of PCs and is independent of the type of A/D card used. The ECG, ICG, and dz/dt signals are digitized at 200 samples per second and then processed to obtain some additional parameters about heart condition. This sampling rate is adequate as the the power spectrum of each of these signals has negligible component above 70 Hz [10, 11]. The PCG has to be digitized at 1000 samples per second [22] and is not processed by the computer. It is meant only for off-line graphical display.

The digitized ECG signal is processed to remove noise and then based upon detection of R points in it, the heart beat rate (HR) is calculated. The ICG waveform z(t), is low pass filtered and its average value is taken as the base impedance Z_0 of thorax. The dz/dt signal in each heart beat is ensemble averaged to remove noise. The R point location in ECG serves as a fiducial mark for ensemble averaging the dz/dt. From this filtered dz/dt the peak value of dz/dt signal $(dz/dt)_{max}$ and the positions of the baseline crossing point T_{base} and the first global minima T_{min} are detected by software. Now the left ventricular ejection time T_{lvet}, is calculated as $T_{min} - T_{base}$ [3]. The stroke volume (SV), is estimated using the modified Kubicek's formula followed by

calculation of cardiac output (CO) as the product of SV and HR. These calculated results are then displayed. The digitized signals are stored in binary files and can also be processed and displayed off-line. The PCG is displayed separately along with the ECG signal for critical diagnosis. The algorithm used for processing the ECG, ICG, and dz/dt signals is explained in the next chapter.

3.5 THORAX SIMULATOR

A thorax simulator has been designed for calibration of the signal conditioner hardware. The thorax simulator is used in place of a subject for calibrating and testing the signal conditioner hardware. The thorax simulator simulates the changing thoracic impedance and also generates and adds a ECG signal at the voltage electrodes. The thorax simulator consists of the ECG simulation circuit and the thoracic impedance simulator circuit. The block diagram of both the circuits is shown in Figs.3.3a and 3.3b and is discussed in the following subsections.

3.5.1 ECG Simulator

The ECG simulator circuit generates only the QRS complex in ECG. The P and T waves in ECG are not simulated as they would add complexity to the circuit, and also the QRS complex can adequately serve the purpose of calibration of the ECG extraction circuit. The QRS is generated by integrating the output pulses of the four monoshots, M1, M2, M3, and M4 shown in Fig.3.3a. The monoshot chain gets triggered by the low to high going pulse of astable multivibrator AM1. The frequency of AM1 determines the period between successive QRS and hence heart beat rate. This ECG is added at the voltage terminals V1 and V2 of signal conditioner hardware.

3.5.2 Thoracic Impedance Simulator

The block diagram of this circuit is shown in Fig.3.3b. The

impedance simulator simulates the time varying thoracic impedance for the four electrode configuration. The thorax impedance is simulated as a base impedance Z_0 with the changing thoracic impedance ΔZ, riding over it. The ΔZ waveform is simulated as a negative going triangular wave and is generated 20 ms after the R point in simulated ECG. This circuit requires the ECG from the ECG simulator as an input. The R point in the ECG is first detected by an R point detector circuit. After the R point has been detected, monoshot M5 provides a delay of 20 ms before generation of the changing thoracic impedance component ΔZ. This value of this delay is so kept to simulate the interval between an actual QRS in ECG and onset of ΔZ in ICG. The high to low going transition of monoshot M5 triggers the M6, M7 monoshot chain. The output pulses of these monoshots are integrated to get the required triangular wave, (ΔZ in ICG). The base impedance along with the changing thoracic impedance ΔZ modulate the 100 kHz excitation signal supplied from the current electrodes I1 and I2. The ECG is added to the modulated excitation carrier and applied to the voltage electrodes (V1, V2) of the signal conditioner hardware.

The component selection criteria for all components used in the thorax simulator is given in Appendix A.

3.6 DATA LOGGER INTERFACE

The data logger consists of a multi-channel A/D converter chip, a RAM for storing the digitized data, and control circuitry to control sampling rate and number of channels to be sampled. The whole instrument is battery operated and quite handy. In this system, "Micro SITE" a data logger from Dynalog Microsystems has been used. This data logger is first programmed to sample the required number of channels at the required sampling rate. After this programming, the data logger can be connected to the signal conditioner hardware for recording the signals. The data logger starts its operation (digitizing and storing), after a key on its

key pad is pressed. After the recording is complete, the data logger is connected to a PC for down-loading and processing (off-line) the recorded data. The actual data logger interface details are presented in the next chapter.

CHAPTER 4

HARDWARE & SOFTWARE DEVELOPMENTS

4.1 INTRODUCTION

As mentioned in the second chapter there has been an ongoing activity at IIT Bombay towards the development of an impedance cardiograph system. In order to achieve the aims of this project, the existing software and hardware was modified or redeveloped as necessary, in order to improve the system for use on exercising subjects. Software was also additionally developed for interface of a notebook PC with a PCMCIA based A/D card and for recording and display of the signals, digitized using a data logger.

In this chapter the design aspect of each block of the system and the algorithms of the software developed are explained.

4.2 SIGNAL CONDITIONER UNIT

This circuit extracts the ECG and the ICG of the subject. The overall working of the circuit and the setup for monitoring physiological parameters using the developed circuit is now explained. The detailed explanation of each block of the signal conditioner hardware is given in the following subsections.

The excitation circuit consists a 100 kHz, constant current source. The amplitude of the current has to be less than 5 mA, rms and for the designed excitation circuit, it is set at 3.3 mA, rms. This current is applied to the current electrodes. Electrode placement is done in such a way as to reduce motion artifacts during exercise. For this, an arrangement as suggested by Zhang et al. [23] shown in Fig.2.9 has been used. In this arrangement, electrode 1 is placed on the back of the neck behind cervical vertebra C4 and electrode 4 over vertebra T9 (Refer Fig.2.9). Voltage electrode 2 is placed 4 cm above

the clavicle on the front of the neck and electrode 3 is placed over the sternum at the fourth rib. Electrodes 1 and 4 pass a constant current through the thorax. Voltage electrodes 2 and 3 measure the voltage across the thorax because of this excitation. The changing thoracic impedance amplitude modulates the 100 kHz carrier. This modulated carrier also has the subject's chest ECG riding over it.

4.2.1 Excitation Circuit

It consists of a Wein-bridge oscillator and a voltage to current converter as shown in Fig.4.1. The oscillator is built around opamp U1 and produces a 100 kHz, 3.3 V rms sine wave. At this frequency and level of excitation neither the nerves nor the heart muscle can be stimulated [2, 10, 20, 23]. The amplitude of oscillator is stabilized by the circuit consisting of FET T2 and zener D1. The output of the oscillator is fed to a V/I converter built around opamp U2. It excites the subject with a constant current of 3.3 mA rms. The capacitors C4 and C5 block the dc offset from the current electrodes. The 100 nF capacitors have a low impedance (15 ohms) at the frequency of excitation and block any dc current that might flow otherwise. The resistor R10 has a value of 2.2 K and limits the amplitude of oscillations when the current electrodes are not connected to a subject. The value of this resistance is ten times greater than the combined impedance of capacitors C4, C5, and the subject (less than 200 ohms) and hence does not introduce any considerable error in the V/I converter action [23].

4.2.2 Electrodes

Four suction type ECG electrodes are used, as shown in Fig.4.2. The electrodes are dipped in ECG jelly before applying on the subject's body to ensure good skin contact. These electrodes are connected as per description given earlier with electrodes 1 and 4 as current electrodes and electrodes 2 and 3

as voltage electrodes.

4.2.3 Instrumentation Amplifier for ICG

The voltage electrodes (electrodes 2 and 3) sense the 100 kHz voltage across thorax which has been modulated by the changing thoracic impedance. The electrodes also pick up a large 50 Hz signal (common mode), the ECG signal and other low frequency artifacts. In order to extract the ICG, these low frequency components have to be filtered out. Fig.4.3 shows the instrumentation amplifier with high pass filtering. The high pass filter comprising of R9, C1, and R10, C2, with 3 dB cutoff frequency of 16 kHz, has almost unity gain at 100 kHz. The instrumentation amplifier is built around opamps U1, U2, U3 and has a differential gain of 14. The output Vo1 of this stage represents the modulated impedance signal and has to be extracted by the demodulator circuit.

4.2.4 Demodulator Circuit

The two-opamp full wave rectifier based demodulator shown in Fig.4.4 is built around opamps U4 and U5. It rectifies and then lowpass filters (C4, R17) the incoming 100 kHz signal. The low pass filter has upper 3 dB cutoff of 30 Hz and provides smoothing of the rectified signal and minimizes the 50 Hz interference. The output of the demodulator Vo2, represents the thoracic impedance Z(t). The waveform Z(t) consists of the base impedance Z_0, impedance variation z(t) due to blood flow, and other factors, caused mainly by respiration. For impedance cardiograph analysis we need Z_0 and derivative of z(t). For obtaining Z_0 we severely low pass filter the Z(t), and discretize it, and then obtain Z_0 by averaging it in the digital domain. Since the artifacts due to respiration cannot be removed by filtering, we here use a special DC cancellation circuit, to obtain from z(t) from Z(t) and then differentiate z(t) before digitizing it.

The signal Vo2, representing Z(t), is lowpass filtered by the lowpass filter comprising of C5 and R22 built around opamp U6, to give Vo3 representing Z_0. This low pass filter has a 3 dB cutoff frequency of 0.7 Hz and attenuates the variations in the Z(t) signal that arise due to respiration and body movements. The DC cancellation circuit and differentiator circuit to obtain a representation of dz/dt from Vo2 are discussed in the next two subsections.

4.2.5 DC Cancellation Circuit

In order to extract the changing thoracic impedance component (ΔZ), the DC cancellation circuit shown in Fig.4.5 is used. The complete circuit is built around opamps U7, U8, U9, U10, and U11. Opamps U9 and U10 are used to design a window comparator circuit. Opamp U11 is a part of an integrator. When z(t) exceeds the threshold (±Vth) of the comparators the integrator gets activated. The threshold voltage is determined by potentiometer R31. The output of the integrator gets subtracted from the incoming signal. This ensures that the output of the DC cancellation circuit remains effectively clamped between ±Vth. The respiration component is a slow varying signal with amplitude much greater than the ΔZ component. The value of ±Vth is kept slightly greater than the normal strength of the ΔZ component but less than that of the respiration component. So now only the fast changing low amplitude ΔZ component escapes unaffected as rest of the z(t) signal gets canceled out due to integrator action. The RC time constant of the integrator charging circuit formed by R32 and C5 or R33 and C5 is around 0.22 seconds and is sufficiently fast to remove the respiration component. This circuit has been earlier developed [32] as an analog adaptation of a circuit developed by Qu et al. [3] using successive approximation register.

4.2.6 Differentiator Circuit

The output of the DC cancellation circuit Vo4, represents z(t) waveform. It is further amplified by the noninverting amplifier shown in Fig.4.6, built around opamp U12. This amplified signal is then differentiated by the circuit built around opamp U13 to get the dz/dt signal. The differentiator shown in Fig.4.6 differentiates frequencies less than 50 Hz (determined by C8 and R3), which is sufficient to differentiate the z(t) signal (with maximum frequency content of around 30 Hz). C7 and R40 form a low pass filter with a 3 dB cutoff frequency of 50 Hz to limit the gain of the differentiator at high frequencies.

4.2.7 ECG Extraction Circuit

The ECG of the subject is also obtained from the same voltage sensing electrodes 2 and 3. This ECG is coupled to the three opamp instrumentation amplifier built around U1, U2, and U3 as shown in Fig.4.7. The instrumentation amplifier has a gain of 16 and amplifies only the low frequency components because of the low pass filters formed by R2, C1 and R5, C2. The low pass filters have a 3 dB cutoff frequency of 40 Hz and effectively remove all the 100 kHz carrier component from the ECG signal. The output of the instrumentation amplifier is high pass filtered by C4 and R10 which has a 3 dB cutoff frequency of around 0.2 Hz. This high pass filter attenuates the base line drift and presents a clean ECG signal. It is to be noted over here that our main interest is to get the chest ECG waveform for detection of the QRS complex and hence relatively higher value of cutoff frequency has been chosen. This ECG is further amplified by the noninverting amplifier built around opamp U4. The noninverting amplifier has a low pass filter (C3, R11) with a 3 dB cutoff of 12 Hz to remove 50 Hz noise and also serves as anti-alias filter prior to digitization. This low pass filtered signal is the chest ECG and is digitized.

4.2.8 Amplifier for PCG

The phonocardiogram of the subject is obtained using an electronic stethoscope. The stethoscope used, "STETHMATE" from Electronic Engineering Corporation has a chest piece which is placed over the chest of the subject. The instrument gives an electrical output which corresponds to the phonocardiogram of the subject. This output has a strength of around 300 mV and is amplified by the circuit shown in Fig.4.8 prior to digitization. The amplifier is built around opamp U1. The amplifier is a non-inverting amplifier with a gain of 10. The input high pass filter comprising of R1 and C1 has a lower 3 dB cutoff frequency of 9 Hz and decouples any dc offsets from the stethoscope output. The output of the amplifier is digitized.

4.2.9 PCB Design of the Signal Conditioner

In order to provide a modular approach, two PCBs have been designed. One PCB contains the excitation circuit and the ECG extraction circuit and the other has the hardware for extraction of Z_0 and dz/dt. The PCBs were designed using 'Circuit Maker', a PCB design software. The layout of the designed PCBs and the actual circuit diagrams are provided in Appendix B, refer figures B.1 to B.8.

4.3 OPTO-ISOLATOR CIRCUIT

The opto-isolator circuit is required to protect the subject from ground fault currents arising from any mains operated equipment like the PC. This circuit is built around the opto-isolator IC 4N25. The circuit used in this project was built on a PCB by Lakdawalla [16]. This circuit should be used when the signal conditioner outputs are interfaced to the mains operated PC through a data acquisition card. This circuit is not required when data acquisition is done by the battery operated notebook PC or the battery operated data logger.

4.4 THORAX SIMULATOR

For reasons discussed earlier in Chapter 2, the thorax simulator had to be redesigned. The redesigned circuit consists of an ECG simulator and an impedance simulator circuit. The design of the thorax simulator developed is explained in the subsequent sections. The component selection criteria for all the components used is presented in Appendix A.

4.4.1 ECG Simulator

This circuit simulates the QRS complex in ECG signal and is shown in Fig.4.9a. The QRS complex simulation is initiated at the positive going transition of astable multivibrator AM1 built around opamp U1. The QRS complex is generated by integrating the pulse outputs of monoshots M1, M2, M3, and M4. These monoshots are built around opamps U2, U3, U4, U5. The periods of the various monoshots are determined so as to simulate an actual QRS complex in ECG. The monoshots M1 and M4 determine the period of down-going transition of the Q wave and up-going transition of the S wave in ECG respectively. So their period is set at 4 ms as determined by R8, C2 and R20, C7. The monoshots M2 and M3 determine the period of the up-going and down-going transitions of the R wave in ECG respectively. Their period as determined by R12, C5, and R19, C6 is set at 40 ms.

The output amplitude of the monoshots vary due to the uncertain opamp saturation voltages. So the circuit around zeners D13, D14, D15, and D16 (V_z = 3.3 V) is built to get fixed voltage pulses of appropriate polarity. The amplitude stabilized output pulses of the monoshots are integrated by the circuit built around opamp U6 shown in Fig.4.9b to simulate the QRS in ECG. The value of integrating resistor C10 is selected as 0.1 uF to get a reasonable value (5 V) for the R wave peak in QRS.

4.4.2 Impedance Simulator

In this circuit the changing thoracic impedance is simulated as a triangular wave. The impedance signal is generated approximately 20 ms. after the R point in the simulated ECG has been detected. This R point detection is done by the circuit built around opamps U1 and U2, shown in Fig.4.10a. Opamp U1 forms a half wave rectifier whose output is differentiated by R2 and C1. The time constant of R2C1 is not very critical, and has been kept at 47 ms to get a square pulse from the R pulse in ECG, after differentiation. This differentiated output is fed to a Schmitt trigger built around opamp U2. The low to high transition of Schmitt trigger, triggers monoshot M5 which provides 20 ms. delay between R point in QRS complex of ECG and generation of the impedance signal.

The high to low transition of monoshot M5 triggers M6. The pulse outputs of monoshots M6 and M7 are amplitude stabilized by circuit built around zeners D11 and D13 (Vz = 3.3 V) and diodes D10 and D12. The pulse duration of these monoshots M6 and M7, are determined by R13, C5 and R17, C7 respectively. The pulse duration of each monoshot is kept at 150 ms so that the duration of triangular wave corresponds to the normal value of T_{lvet} in an actual dz/dt. The integrator built around opamp U6 generates the required triangular wave by integrating the amplitude stabilized outputs of monoshots M6 and M7. This triangular wave represents the changing part of the thorax impedance. The impedance simulator simulates the time varying thorax impedance for the four electrode configuration and is shown in Fig.4.10b. Resistors R32 and R34 represent the contact resistances of the current stimulation electrodes. Similarly resistors R31 and R34 represent the contact resistances of the voltage sensing electrodes. The time varying thorax resistance is simulated by the R_{ds} of JFET T2. The triangular waveform Vo5 representing z(t) is added to a negative dc bias by the circuit around U7 and its output Vo6 drives the gate of T2. Resistors

R24 and R25 bias the JFET T2 at -1.3 V. The magnitude of z(t) can be altered by R36.

In order to simulate the presence of chest ECG at the voltage sensing electrodes, the simulated ECG waveform is coupled by R30 to the impedance simulator circuit. Resistor R35 helps in establishing the gate bias. This voltage between voltage electrodes is fed to the the signal conditioner hardware. The thorax simulator is built on a PCB layout with one PCB for the ECG simulator and another for the impedance simulator. The PCB layouts of the thorax simulator are provided as Figs. B.9 to B.16 in Appendix B.

Until now the system hardware was explained. Now the programs for data acquisition and software processing of ECG, ICG, dz/dt and PCG will be discussed.

4.5 SIGNAL ACQUISITION AND PROCESSING

The estimation of cardiac output (CO), requires the knowledge of the heart beat rate (HR), and stroke volume (SV) which are obtained by processing of the digitized ECG, Z_0, and dz/dt. As mentioned in the previous chapter the signal acquisition and processing can be done by using one of the three setups, a) signal conditioner interfaced to a PC through PC-bus based data acquisition card PCL-208 (Dynalog Microsystems), b) signal conditioner interfaced to a notebook PC through a PCMCIA-bus based data acquisition card, DAQ700 (National Instruments), or c) signal acquisition by a data logger for signal recording and down-loading the data to a PC for off-line processing. A signal acquisition and processing program has been developed for either of the first two setups. The use of the third setup requires a different program for signal acquisition although processing remains the same. The card settings and details for both the data acquisition cards are given in Appendix D and E respectively.

Joshi [15] had earlier developed software for monitoring the

cardiac output. His program digitized the ECG waveform e(t), de/dt, Z_0, and dz/dt and stored them in binary files. de/dt is the first derivative of ECG and was required to be generated by the signal conditioning hardware. The de/dt, Z_0, and dz/dt signals were processed to calculate the HR, LVET, SV, and CO. The algorithm used by Joshi for processing the signals (referred to as SPA-1 in this report) is now briefly explained. The shortcomings of this algorithm when used with noisy exercise recordings and the solutions for these shortcomings are also briefly mentioned. A modified signal processing algorithm named SPA-2 has been developed and implemented.

4.6 ALGORITHM SPA-1

In Joshi's [15] algorithm the ECG derivative, de/dt signal was first low pass filtered and the R-point locations in it were determined using a peak-detection scheme. Once all the R-points locations were determined, the heart rate was calculated from the average R-R interval. These R point locations also served as a fiducial mark for ensemble averaging the dz/dt signal, in order to remove the dominating muscle and motion noise. Then the base line crossing point T_{base}, the $(dz/dt)_{max}$ point Tpeak, and the 1st global minima after the $(dz/dt)_{max}$ point Tmin, as shown in Fig.2.6 were detected. The time difference between the T_{base} and T_{min} points give the left ventricular ejection time T_{lvet}. The base impedance of the chest was calculated by averaging the Z_0 signal. Kubicek's modified formula was used to calculate stroke volume.

This algorithm was found to give erroneous values for SV especially when used on noisy recordings common during exercise. So the working of the existing algorithm was critically examined and the following problems were noted. The solution for each problem has also been mentioned.

1) The R-point detection was carried by a simple peak-detection

scheme and it gave false detections on noisy de/dt recordings, especially those taken during exercise. This problem has now been solved by implementing Hamilton's algorithm [33] for detecting R points in ECG. This algorithm is much better suited to work on noisy ECG recordings.

2) The ensemble averaged dz/dt signal some times lacked the base line crossing point T_{base} (Refer Fig.2.6), and hence the value of LVET calculated as

$$T_{lvet} = (T_{min} - T_{base})$$

was in error. This problem has now been solved by ensemble averaging dz/dt signal from 60 points prior to the R point in ECG signal.

3) The algorithm for detection of the T_{min} point in the dz/dt signal sometimes gave incorrect results. This happened because usually there exists more than one valley around the T_{min} point in the ensemble averaged dz/dt signal and the algorithm takes the first valley occurring after T_{peak} signal as the T_{min} point.
To solve this problem some *a priori* information is required about the position of the T_{min} point. This information is available through the heart rate as suggested by Zhang et al. [3]. They have also suggested an algorithm for deciding the T_{min} point which has now been implemented.

4) Even after ensemble averaging, the dz/dt signal, it looked too noisy. A noisy dz/dt signal can give wrong estimates for the T_{base}, T_{peak}, and T_{min} points. This happened because only 10 second data samples were recorded and they contained hardly 8 to 12 heart beats. Since ensemble averaging principle works best on 40 to 60 heart beat recordings [3], which corresponds to a sampling record time of 30 to 80 seconds the sampling record time has to be increased.

In SPA-1 'arrays' were used to store the recorded data in computer memory. However the array size could not be made to exceed 64 k bytes, thus limiting the recording time to 10 seconds. This was a limitation of the compiler used (Turbo C).

To cure this problem, software was developed which used ('C' structures), a different memory allocating and handling mechanism to record and store 40 second data in computer memory. But this program worked too slowly due to the lengthy mechanism for addressing 'structures' in memory and also required lots of free RAM in the computer to work properly. So a new approach which solves both these problems was implemented.

In this approach, data are continuously recorded and stored. After every 10 seconds, data recorded is semi-processed and these intermediate results of processing are temporarily stored. The semi-processed results require only small amount of memory for storing. The memory that had the unprocessed ten second data is now free and can be used to store newly recorded data. In this way all the data for the programmed amount of recording time is gathered and then the intermediate results obtained are processed to calculate the HR, Z_{base}, T_{lvet}, SV, and CO.

A new signal acquisition and processing algorithm SPA-2 has been developed, incorporating all the modifications in SPA-1. This algorithm uses the waveforms e(t), Z_0, and dz/dt.

4.7 ALGORITHM SPA-2

The ECG is first processed to enhance the QRS complexes in the noisy ECG, so that the R-point detection becomes reliable. This processing is done according to Hamilton's algorithm [33] and is carried out as follows. First ECG is low pass filtered to remove the 50 Hz and other high frequency noise. The equation for low pass filtering is given as,

$$y(n) = 2y(n - 1) - y(n - 2) + x(n) - 2x(n - 2)$$

$$+ x(n - 12)$$

This is followed by high pass filtering to enhance the QRS complexes in the ECG signal.

$$y(n) = y(n - 1) - x(n)/32 + x(n - 16) - x(n - 17) + x(n - 32)/32$$

The low pass filtering and high pass filtering attenuate the out-of-band noise which gets introduced in the ECG during exercise. The filtered ECG is now differentiated to enhance the QRS complex.

$$y(n) = (2x(n) + x(n - 1) - x(n - 3) - 2x(n - 4))/8$$

The differentiated signal is squared to further enhance the QRS complexes and then a 32 point moving averaging filter is used to suppress all peaks of width less than that of QRS complex.

$$y(n) = \sum_{m=0}^{31} [x(n - m)]^2 / 32$$

The peaks in the filtered signal are now detected by a simple peak detection scheme. Once a peak is detected, the point half way down the negative slope of the peak is located and marked as a possible QRS complex. If the maximum value of signal in this peak exceeds some threshold called 'Peak-Threshold' and if the width of the peak exceeds 120 ms, then the peak is taken as a valid QRS complex. The value of 'Peak-Threshold' is now updated as 0.28 times the peak value of the last detected QRS complex.

Once all the R point locations in the ECG have been determined, ensemble averaging of the dz/dt signal is carried out, from 60 sample points prior to the location of the R-point. The ensemble averaging of dz/dt in each heart beat is done over 0.8 times the average R-R interval as this segment length contains all the features of interest shown in Fig.2.6. Now the ensemble averaged dz/dt signal is processed.

The point T_{peak} in the ensemble averaged dz/dt is detected as the point having the largest value in the entire ensemble

averaged dz/dt. The baseline crossing point, T_{base} is detected as the first point prior to the T_{peak} point that has a negative value. Calculation of T_{lvet} requires the location of T_{min} point,

$$T_{lvet} = T_{min} - T_{base}.$$

It has been observed that there is a linear relationship between heart rate and Tlvet [3] given in ms as,

$$Tlvet = (391 - 0.91\ HR)$$

This relationship is used to get some *a priori* information about the location of Tmin point. This is achieved by using the formula for T_{lvet} to get value for T_{min}. Now a window of ± 0.125(R-R) interval is placed around this estimated value of T_{min} and the point at which minimum value of signal in this window occurs is taken as the actual T_{min} point. Now T_{lvet} is recalculated as

$$T_{lvet} = T_{min} - T_{base}$$

Kubicek's modified formula is now used to calculate the SV. The results of calculation are then displayed.

The program "SPA2A" implements the above mentioned algorithm for on-line cardiac output monitoring. This program also gives an option of digitizing the PCG along with the other signals. This PCG is not processed but stored in PC in the form of binary files. Another program "SPA2B" uses the same above mentioned algorithm for off-line processing and display of the recorded physiological signals. This program also displays the PCG along with the other signals.

4.8 DATA LOGGER INTERFACE

The data logger interface was developed to allow digitization of ECG, ICG, dz/dt and PCG during outdoor exercises. Data logger allows easy portability of the instrument but the processing of the sampled signals has to be done off-line.

To provide this data logger interface, a program "SPA2C.LIT"

for digitizing and storing the ECG, Z_0, and dz/dt signals has been developed. Another program "SPA2D.LIT" has been developed and allows the PCG to be sampled along with. These programs control number of channels sampled and the sampling rate of the data logger and have to be down-loaded into the data loggers memory and executed. This can be done by using the software 'LAUNCH' from Dynalog Microsystems (Refer Appendix C for details). This software down-loads a specified "*.LIT" program from the PC into the data logger and starts its execution. The down-loaded program after execution waits for a key press on data logger's key pad, to start its digitizing and storing operation. The digitized data is stored in data logger's RAM.

After recording is complete, the logged data is off-loaded into a PC. This is done using the software "OFFLOAD" from Dynalog Microsystems. This software uploads data from the data logger into the PC and stores them in "*.DAT" files. These files are not in a ASCII format. The conversion to ASCII format is done by using the software 'Xsite' supplied by Dynalog Microsystems. This software converts the "*.DAT" files into text files, "*.TXT". The text file so generated has to be converted into binary file which can be understood by the off-line monitoring program. Software for doing this conversion has been developed in 'C' language. This conversion is done using a program called "CONVT.C", when only the ECG, ICG, and dz/dt are sampled. When PCG has also been sampled "CONVT1.C" has to be used. Now the data recorded using data logger is in a format that can be processed and displayed using the off-line processing program "SPA2B.C".

CHAPTER 5

TEST RESULTS

5.1 INTRODUCTION

This chapter deals with testing the impedance cardiograph system, on the thorax simulator and exercising subjects. The signal acquisition and processing was done by the program "SPA2A.C" for on-line cardiac output monitoring. This program stores the recorded ECG, z(t), dz/dt, and PCG signals in binary files for off-line processing. These stored signals were reprocessed off-line using "SPA2B.C" which displays the results of processing along with a graphical display of the recorded signals. The PCL-208 A/D card from Dynalog Microsystems interfaced with a mains operated PC, was used for signal acquisition purposes.

In the following sections the results obtained in terms of the various physiological parameters (HR, Z_0, T_{lvet}, SV, CO), from testing the impedance cardiograph system first with thorax simulator and then on five exercising subjects are discussed. These physiological parameters are tabulated in Appendix G. These results were obtained by processing the ECG, z(t), and dz/dt signals of each subject recorded during the exercise. The PCG of each subject was monitored only for ten seconds before the start of exercise as during exercise it contained severe noise originating from the motion and heavy breathing of the subject.

5.2 TEST RESULTS FROM THORAX SIMULATOR

The impedance cardiograph system was tested using the thorax simulator in place of the subject. The outputs of the signal conditioner hardware were digitized and processed. The waveforms of the recorded signals are shown in Fig.5.1. The amplitude of the ECG, z(t), and dz/dt waveforms can be used to

calibrate the signal conditioner hardware. The heart beat rate set in the thorax simulator circuit was 70 beats/min. The heart beat rate calculated by software processing of the recorded ECG had a mean value of 68 beats/min with a variation of $\pm$ 1 beat/min over a recording interval of 5 minutes. The impedance simulator presents a 20 ohm base impedance between its V1 and V2 terminals. The triangular wave in thorax simulator, which generates the change in base impedance is set to produce a 2.3 ohms/sec change in base impedance and corresponds to a $\pm$ 1.5 V bipolar pulse as dz/dt, from the signal conditioner hardware. The recorded z(t) signal from the signal conditioner had a strength of 5.8 V which, with the calibration factor of 3.2 ohms/V corresponds to a base impedance of 19 ohms. The dz/dt output had a strength of $\pm$ 1.6 V which with the calibration factor of 1.51 ohms/sec/V corresponds to the base impedance change at the rate of 2.4 ohms/sec. Both these observations (z(t) and dz/dt) were constant over the recording interval of 5 min.

5.3 TEST RESULTS WITH EXERCISING SUBJECTS

Before going into the details of the results, the method used for obtaining the same is discussed. Four, ECG spot electrodes were used to extract the ECG and thoracic impedance of the subject, and were placed using the scheme proposed by Qu et. al. [23] shown in Fig.2.9. A constant current of 3.3 mA rms, with frequency 100 kHz was passed through the current electrodes. The signal conditioner inputs were connected to the voltage electrodes V1 and V2. The subjects were initially at rest and were asked to exercise on an exercise bicycle until they felt tired or their heart rate roughly doubled. The recording and processing of the signals was done throughout the exercise and was continued after exercise till the heart rate came down to normal. The value of blood resistivity, required to calculate SV by Kubicek's method was taken as 142 ohm/cm for

all subjects. Since the impedance cardiograph system used, is not calibrated, only the relative variations in the monitored values of SV and CO should be considered.

The values of HR, Z_0, T_{lvet}, SV and CO were monitored every ten seconds during the entire exercise period and are listed in Appendix G for all the subjects. A sample of the waveforms of ECG, Z_0, dz/dt, and PCG as recorded at start of exercise for subject NMH is shown in Figs.5.2. Figure 5.3 shows the variation in the monitored value of Z_0 and T_{lvet} with respect to time for subject NMH during the exercise. In this figure the X axis is time and starts with 0 min, which corresponds to the start of exercise. The variation of HR, SV, and CO with respect to time are shown in Figs.5.4 to 5.8. for all the subjects. In these figures the point s/r marks the time at which the exercise was stopped. Fig.5.9 and Fig.5.10 shows the variation in Z_0 and T_{lvet} and HR, SV, and CO respectively for subject SVP during rest.

For subject NMH it can be observed from Fig.5.3 that the value of Z_0 is stable at 13 ohms during the exercise period. Also the variation in T_{lvet} is gradual without any drastic changes which suggests proper working of the algorithm for determining it. The Z_0 and T_{lvet} of all the other subjects were also found to exhibit a similar behavior.

Fig.5.4 shows the variation in the HR, SV, and CO as recorded during exercise for subject NMH. The exercise was continued till the point s/r marked in the figure. In all the subjects the exercise had to be stopped due to fatigue. The HR shows a definite increase till the point s/r marked the figure. The SV and CO show an initial decrease at the start of the exercise but recover soon and show an increasing trend till the point s/r marked in figure. The SV and hence CO show a constant fluctuation but still a definite trend in their variation is observed. The initial decreasing trend in SV occurs because with increased HR less time is available for pumping the blood.

Once the HR starts stabilizing the SV shows an increasing trend thus increasing the CO. After the termination of exercise the heart rate starts decreasing. This happens because during the resting phase the oxygen requirement of muscles decreases producing the slow decrease in HR. The body now recovers slowly bringing the HR, SV, and CO back to normal.

The variation trend in HR, SV, and CO with exercise is similar for subjects KSP, MRV, and MRP. However, in the case of subject SAP (refer Fig.5.5) a decrease in the SV and CO with increasing levels of exercise is noted. This could indicate a poor exercise capability in the subject. This subject is also slightly obese.

5.4 DISCUSSION

The values of Z_0, HR, T_{lvet}, SV, and CO were monitored every 10 seconds in order to examine the fine variations in these parameters during exercise which may otherwise be lost due to averaging effect over a longer monitoring interval. This value of 10 seconds is actually a compromise as in 10 seconds there can be at the most 25 heart beats and the digital processing of ECG, z(t), and dz/dt works best when the recorded signals contain 60 to 64 heart beats. It can be observed from Figs. 5.4 to 5.8 that in all subjects, SV and hence CO show wide fluctuations during exercise. These fluctuations are absent during rest as can be noted from Table 5.1 which lists the mean and variance of all the recorded parameters (Z_0, T_{lvet}, HR, SV, and CO) for all subjects during rest. Fig.5.9 and Fig.5.10 show the variation in Z_0 and T_{lvet} and HR, SV, and CO respectively for subject SVP during rest over an interval of 5 min with the parameters monitored at 10 sec intervals. In Fig.5.10 the SV and CO show some fluctuations.

Hence it can be concluded that the fluctuating value of SV and hence CO, during exercise, is either because of an actual variation in SV of heart during exercise or because the short

time interval of 10 sec, over which the SV was monitored, was insufficient to remove the muscle and motion artifacts from the exercise ECG and dz/dt recordings, with the employed digital processing methods. In a noisy ECG signal the R-points in QRS may not get properly detected and also, the algorithm for detection of T_{max}, T_{base}, and T_{min} points in dz/dt may fail on noisy dz/dt records. The actual source of fluctuations in SV can be pin-pointed only after more extensive testing on exercising subjects. This would include monitoring the SV and CO at longer time intervals and observing the effect of the same on the variance of SV and CO.

CHAPTER 6

SUMMARY AND CONCLUSIONS

In impedance cardiography the cardiac output is obtained from stroke volume which in turn is obtained by monitoring the changes in thoracic impedance due to blood flow. The model of thorax as suggested by Kubicek has been used by many researchers to develop impedance cardiograph systems. This model assumes that the primary source of impedance change in thorax during systole is due to the blood entering the lungs.

A system based on Kubicek's thorax model was proposed by Qu et al. [3, 23] for monitoring cardiac output. This system had been implemented earlier at IIT Bombay with modifications. In this system a constant current of 3.3 mA, at frequency of 100 kHz is injected into the thorax of the subject. The spot electrode placement for injecting the current and recording the resulting voltage across thorax is as suggested by Qu et al. [23]. The instrument developed recorded the ECG, de/dt, z(t), and dz/dt of the subject. This recording was done by a PC with an A/D card interfaced to the signal conditioner hardware through an opto-isolator interface. The signals recorded by the PC were processed to monitor the stroke volume and cardiac output. This system was based on a PC and was tested by Ingle [18] on subjects at rest.

In this project, a new system, incorporating certain modifications in the earlier systems, has been built and tested with exercising subjects. Provision for additionally recording the phonocardiogram of the subject has been implemented. The system modification has mainly been carried out in the software processing algorithm for monitoring the stroke volume and cardiac output. The algorithm for detection of R-point in ECG as suggested by Hamilton [33] has been implemented. The signal acquisition scheme has also been completely modified to

remove the time limit on the recording interval. The redesigned system can now be used on exercising subjects.

A need was also felt to make the system portable. The existing system was built around a PC which lacks portability. So a notebook PC was provided for use, in lieu of the mains operated PC. An A/D converter interface for the notebook PC based on the PCMCIA bus was also developed. For field use of the system, a data logger interface for recording the required physiological signals was also developed. This data logger replaces the need of a PC for digitizing and storing data. The data logged into the data logger is down-loaded into a PC and processed off-line.

The existing calibrator (thorax simulator) for impedance cardiography hardware needed modifications. So the thorax simulator circuits were redeveloped, implemented and tested. Finally experiments were carried out using the impedance cardiograph system developed with the redesigned thorax simulator and five exercising subjects. From these experiments it was noted that the estimate for SV obtained for a subject undergoing exercise showed wide fluctuations which could either be because of their actual presence in the subject or because of some characteristic of the digital processing algorithm which was used to estimate the SV.

Suggestions for future work

The hardware for extraction of the z(t) and dz/dt will have to be carefully studied for further modifications. The capability of the DC cancellation circuit in removing the heavy respiration component from Z(t) signal during exercise should be thoroughly examined. It has been observed in all the subjects that the estimate for CO obtained, is quite low for a normal, male subject, in the age group of 22 to 25 years. In order to sort this problem out, first of all, the actual value of blood resistivity for the subject will have to be determined

and then used in the Kubicek's formula for SV, as against the current procedure of using a fixed value of 140 ohms/cm. Secondly experiments will have to be carried out by varying the distance between voltage electrodes. As this distance is varied the strength of the carrier voltage, and also the impedance variation between the voltage electrodes varies, thus giving a scaled value for the estimate of SV and hence CO. This CO will then have to be compared with another estimate obtained from some other calibrated instrument meant for monitoring CO. The experiments should be carried out till both the instruments show a good correlation in their estimates of CO.

Table 5.1 Test results with subject under resting condition. The values are the mean with standard deviation s.d., in paranthesis. Observations are over six, 10 sec intervals.

Subject	H.R. beats/min	Z_0 ohms	T_{lvet} sec	SV ml	CO lit/min
NMH	85 (1)	14 (0)	0.28 (0.02)	35.0 (2.1)	2.9 (0.25)
KSP	86 (0)	18 (0)	0.20 (0.01)	22.8 (2.5)	1.9 (0.21)
SAP	61 (2)	13 (0)	0.34 (0.01)	89.8 (5.4)	5.4 (0.23)
MRP	87 (4)	14 (0)	0.19 (0.03)	12.0 (3.0)	1.0 (0.28)
MVN	83 (1)	17 (0)	0.30 (0.02)	41.6 (2.5)	3.4 (0.24)

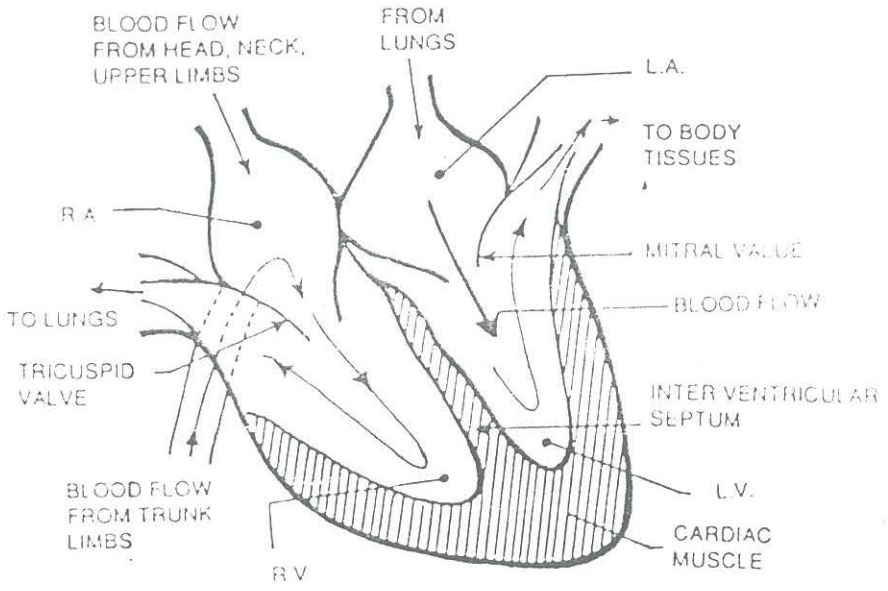


Fig. 2.1a Schematic representation of blood flow through heart, [20].

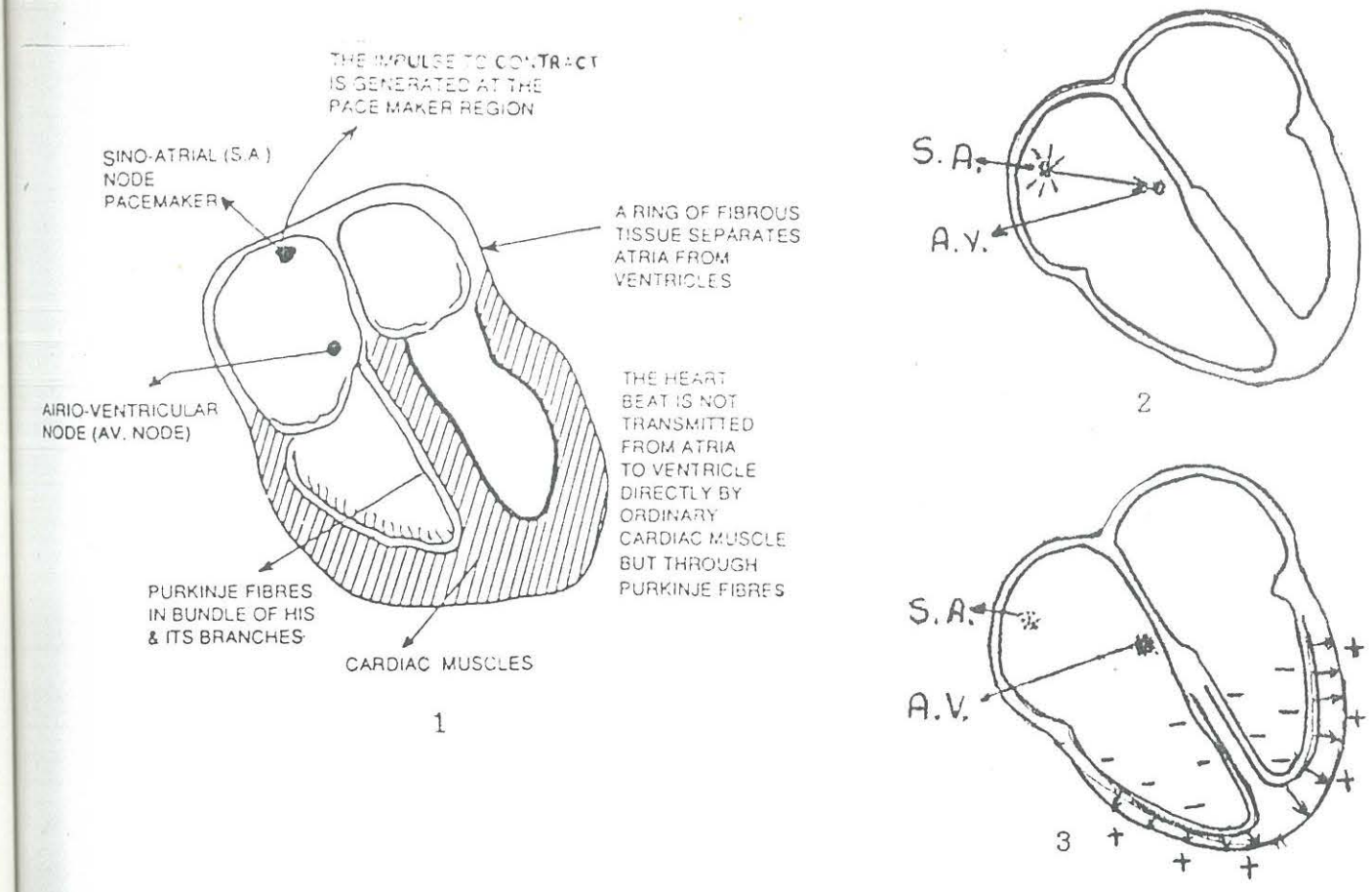


Fig. 2.1b Origin and conduction of heart beat. 1) Conduction mechanism of heart. 2) Spread of excitation through the atria from S.A. node to A.V. node 3) Spread of impulse through the ventricles. Adapted from [11, 21]

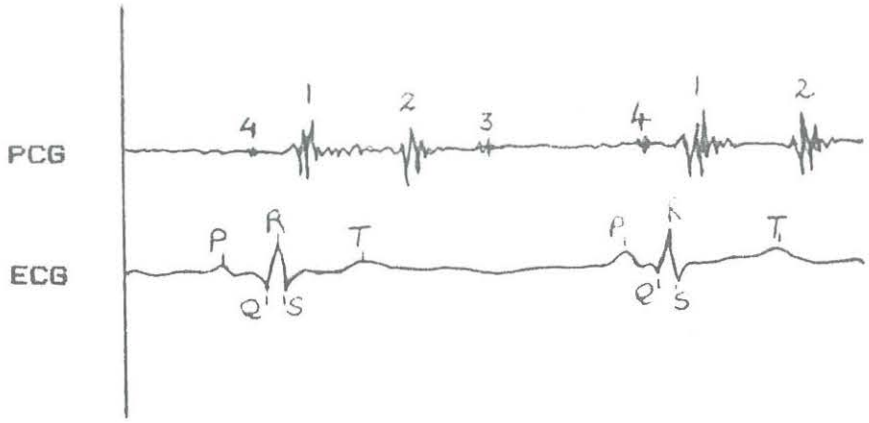


Fig.2.2 Typical recordings of ECG and PCG during a heart beat, [21]

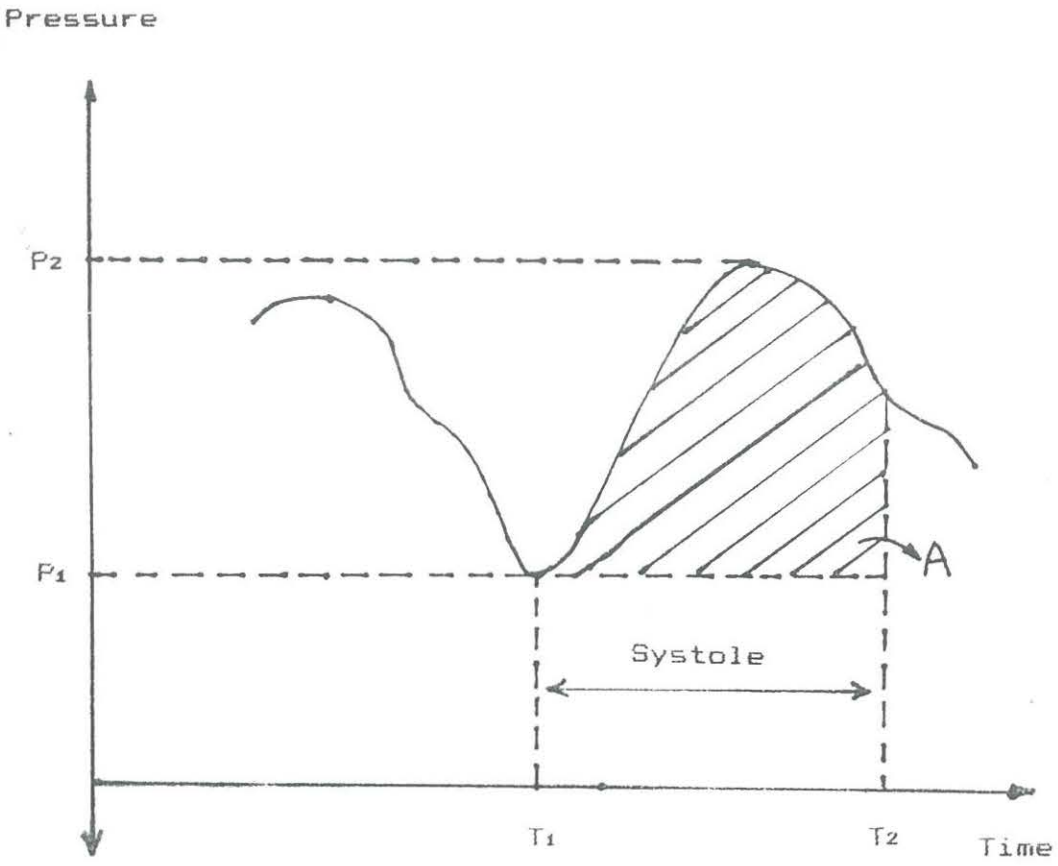
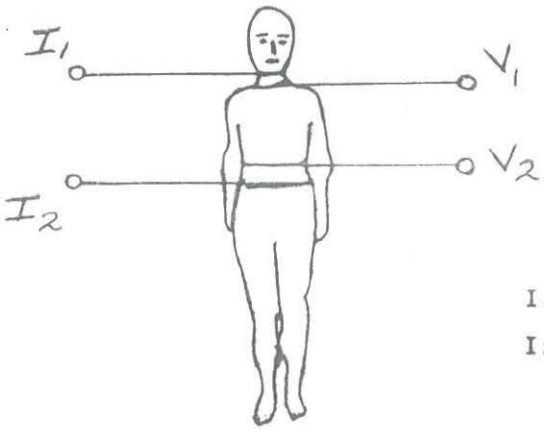
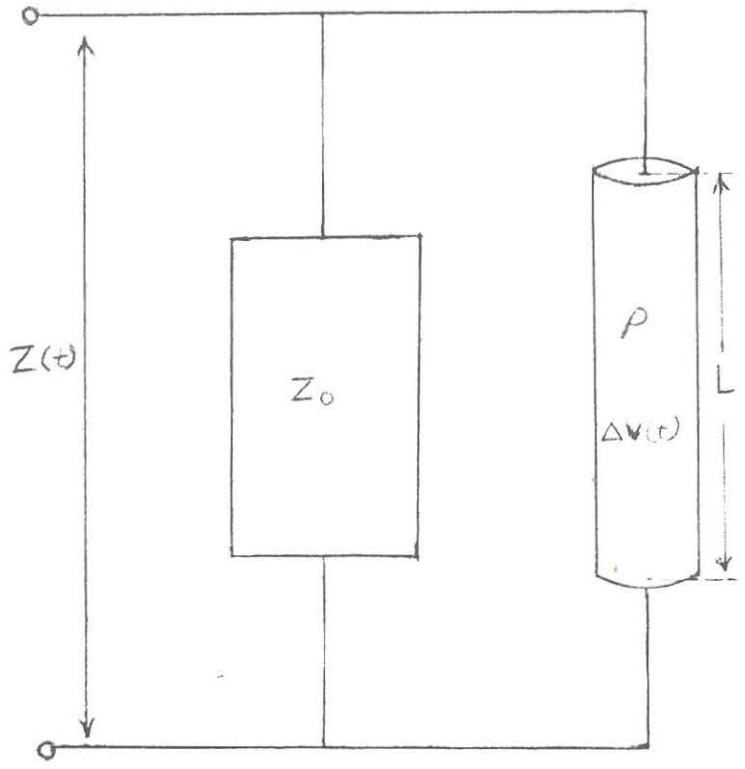


Fig.2.3 Typical aortic pressure waveform observed during a heart beat, [10]



I1 Electrode 1 V1 Electrode 2
I2 Electrode 4 V2 Electrode 3

Fig.2.4 A general electrode placement scheme for impedance cardiography using band electrodes, [2]



L = Length
ρ = Resistivity
Z0 = Base impedance of thorax
ΔV(t) = Volume change
Z(t) = Time varying impedance across thorax

Fig.2.5 Parallel column model of thorax as suggested by Kubicek, Adapted from [2].

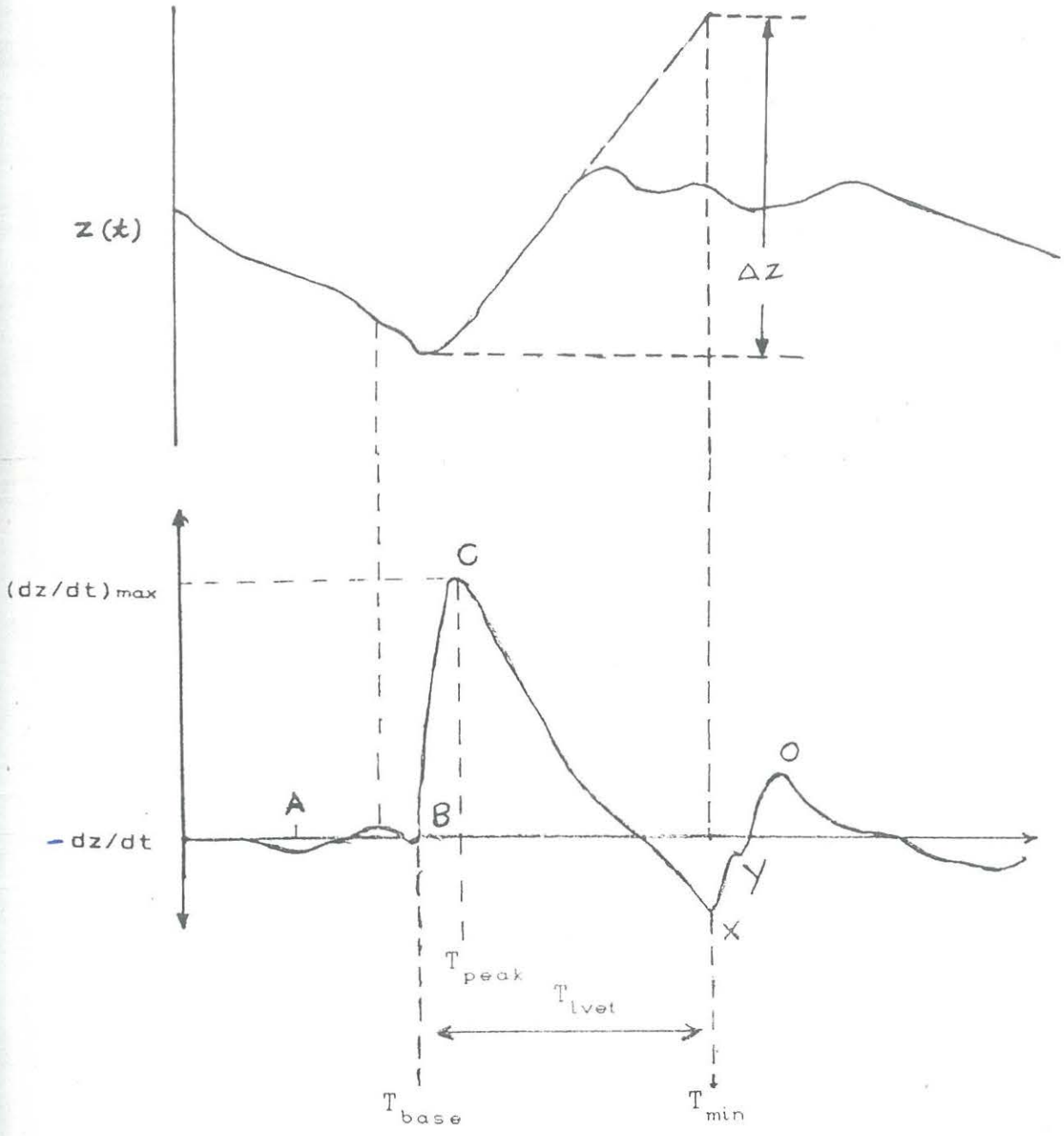
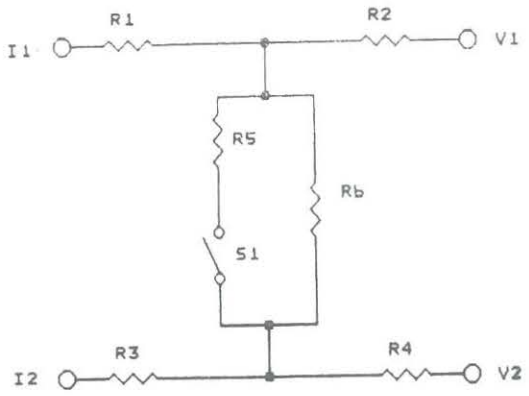


Fig.2.6 Typical ΔZ and dz/dt waveforms from the thorax of a human subject, Adapted from [2]



Rb = Base impedance of thorax

R5 = Switched resistance

R1, R2, R3, and R4 = Skin-electrode contact resistances

Fig. 2.7 Electrical model of thorax built using resistors.

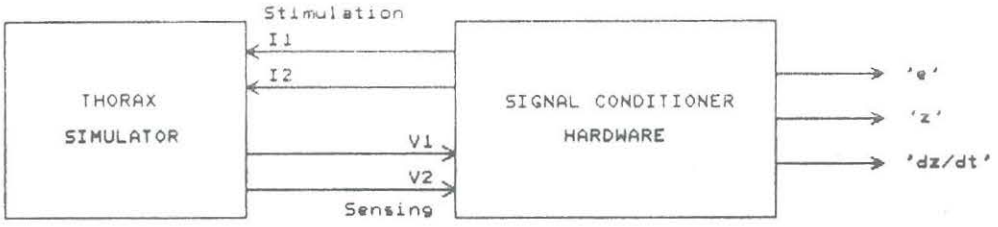
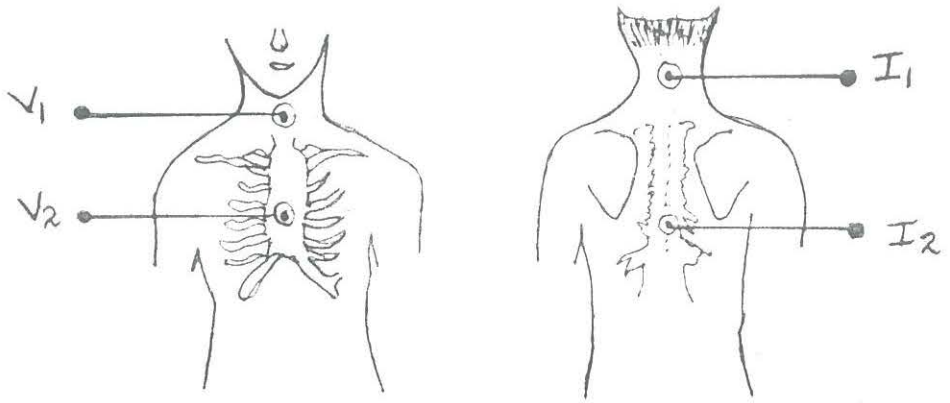


Fig. 2.8 Setup for calibration of signal conditioner hardware.



I_1	Electrode 1	V_1	Electrode 2
I_2	Electrode 4	V_2	Electrode 3

Fig.2.9 Electrode placement scheme for impedance cardiography as proposed by Zhang et al., [23]

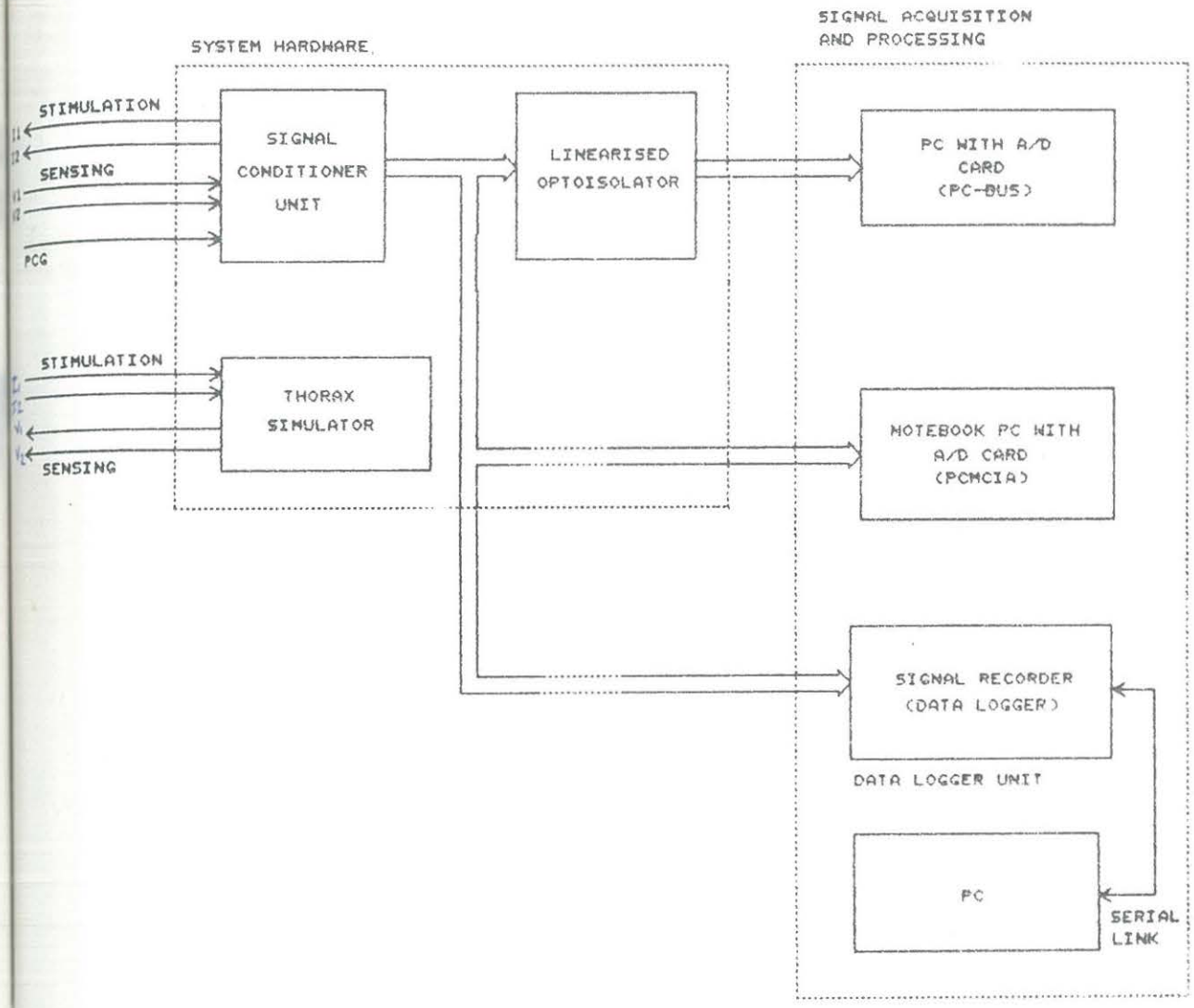
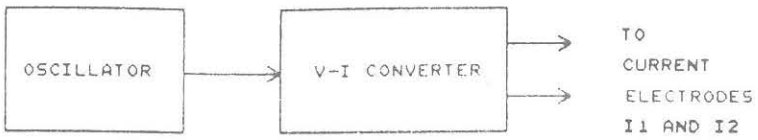
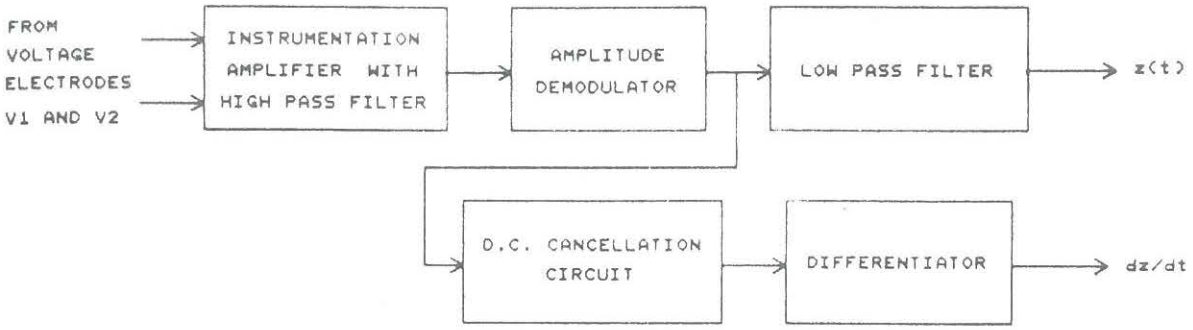


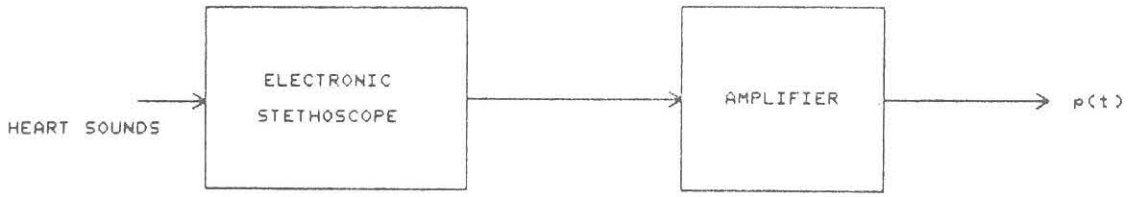
Fig. 3.1 Block diagram of impedance cardiograph system.



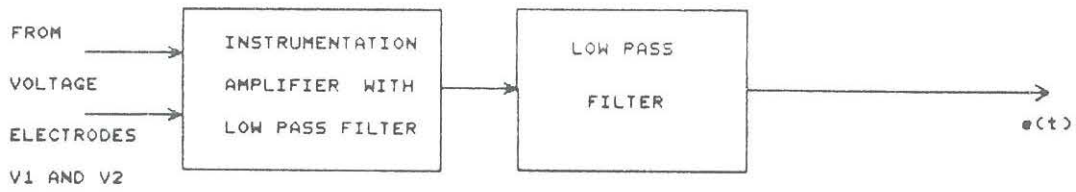
SUBJECT EXCITATION CIRCUIT



ICG EXTRACTION AND dz/dt CIRCUIT



PCG EXTRACTION CIRCUIT



ECG EXTRACTION CIRCUIT

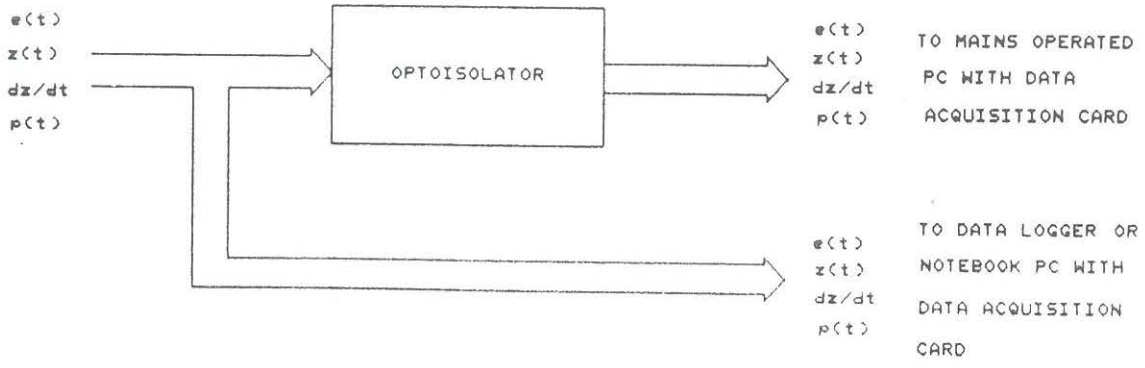


Fig. 3.2 Block diagram of signal conditioner unit.

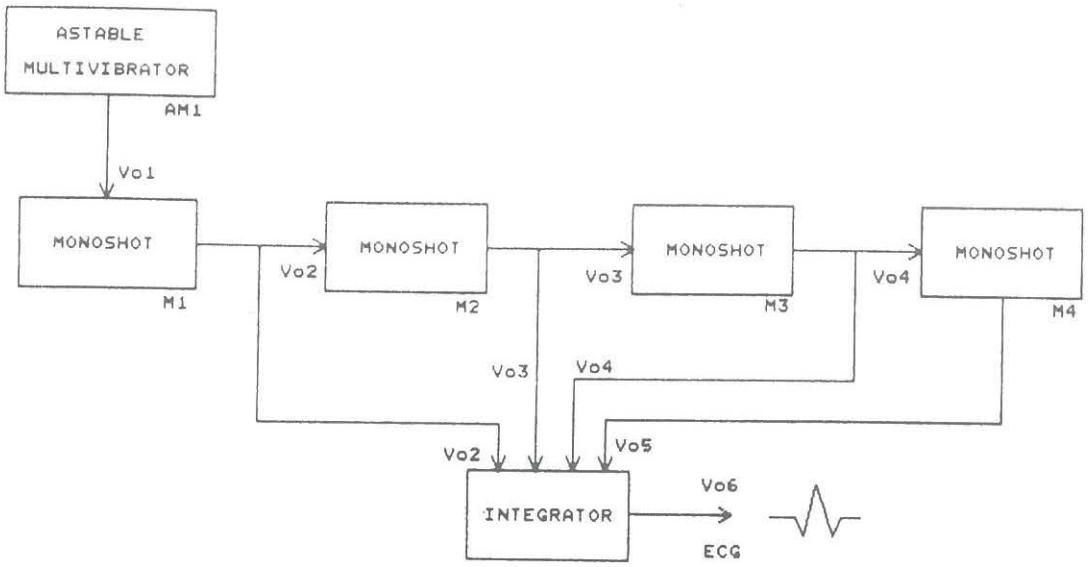


Fig. 3. 3a Block diagram of ECG simulator circuit.

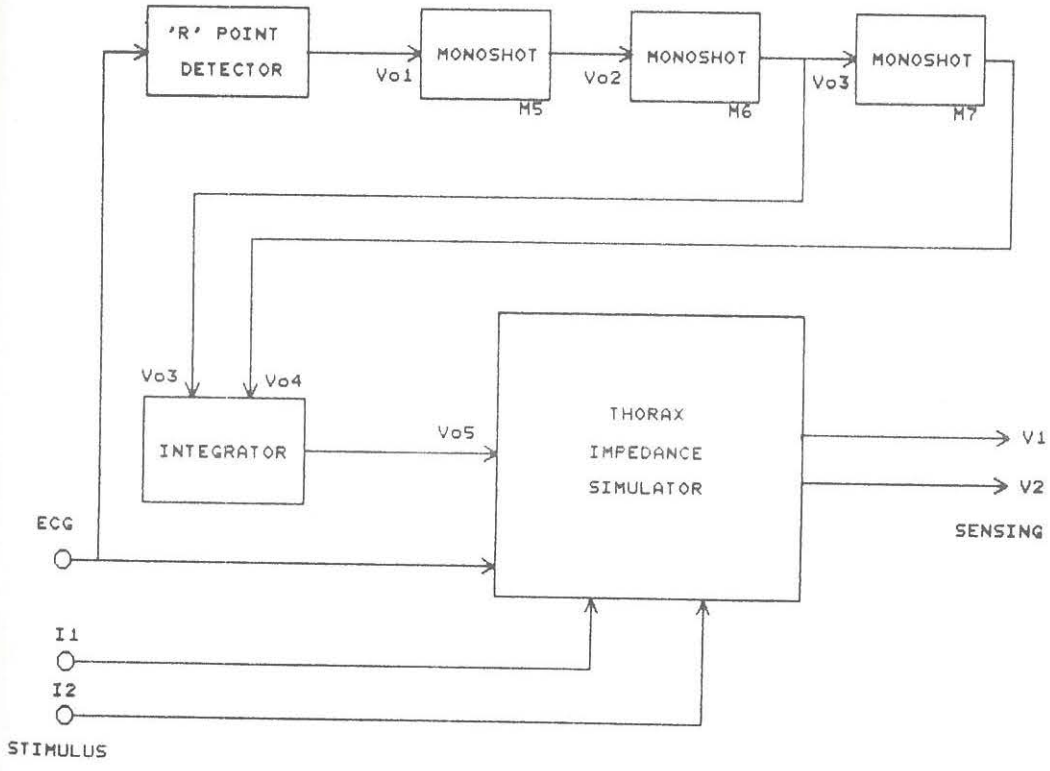
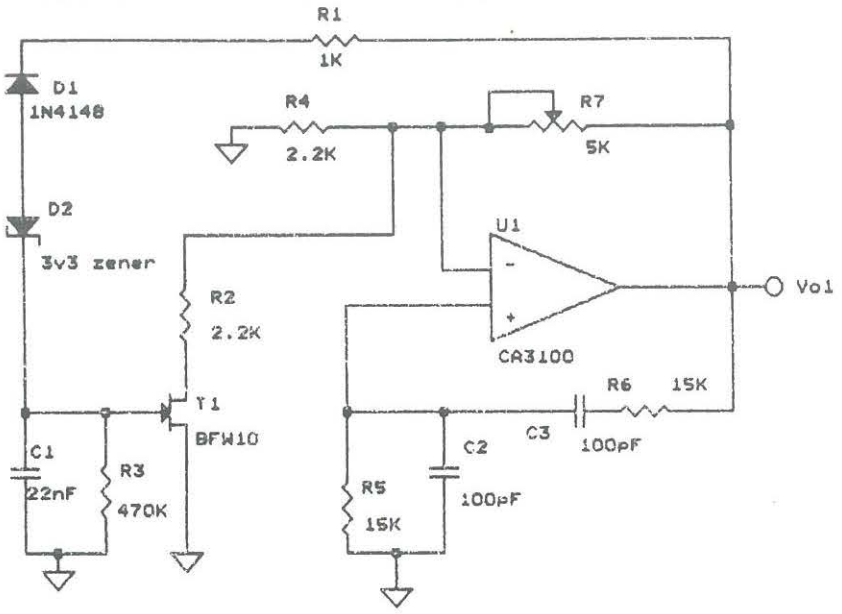
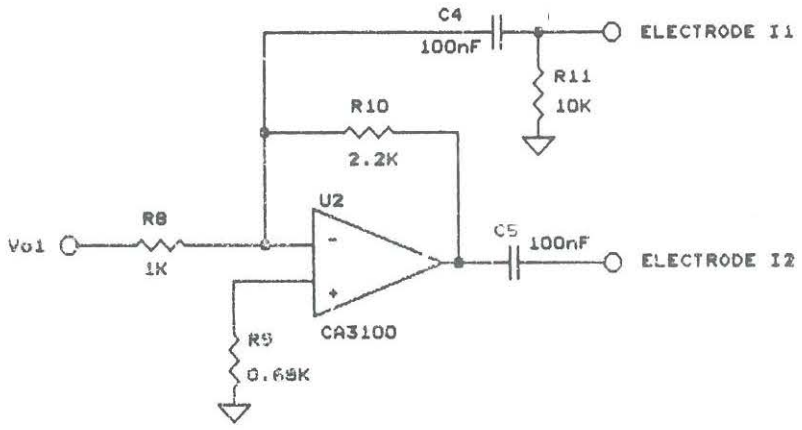


Fig. 3. 3b Block diagram of thoracic impedance simulator circuit.



OSCILLATOR CIRCUIT



V-I CONVERTER

Fig. 4.1 Subject excitation circuit.

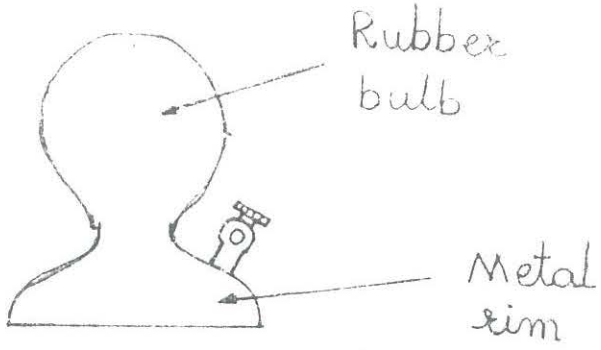


Fig. 4.2 ECG chest electrodes.

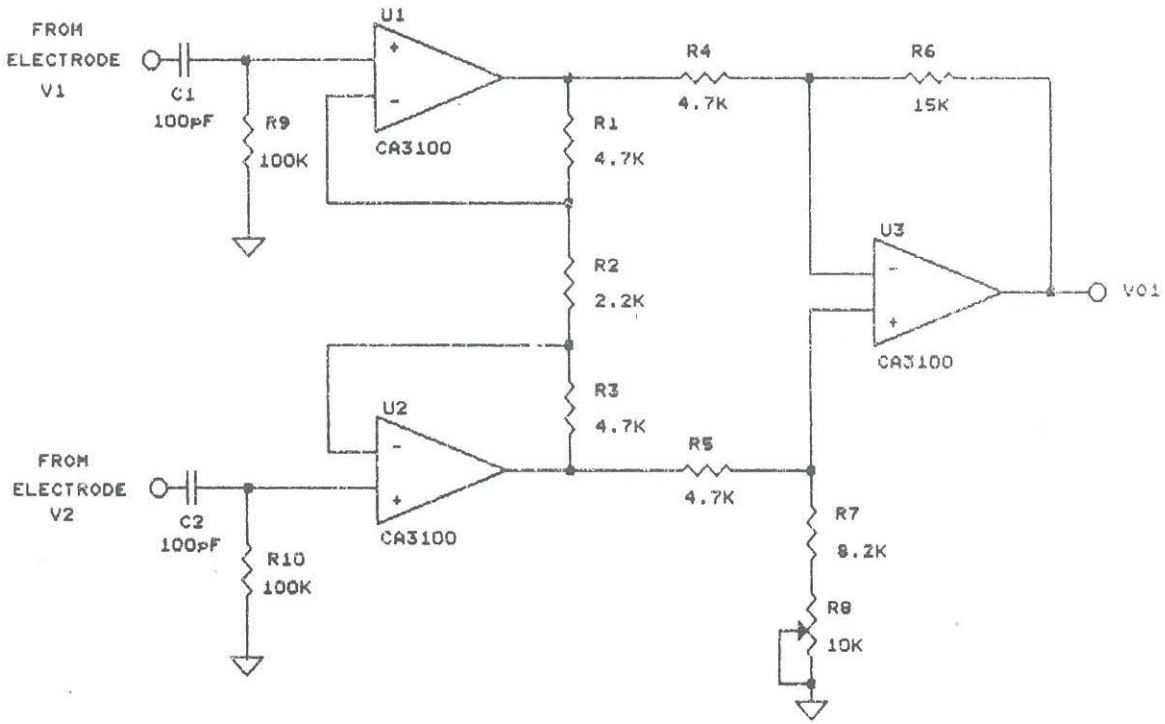
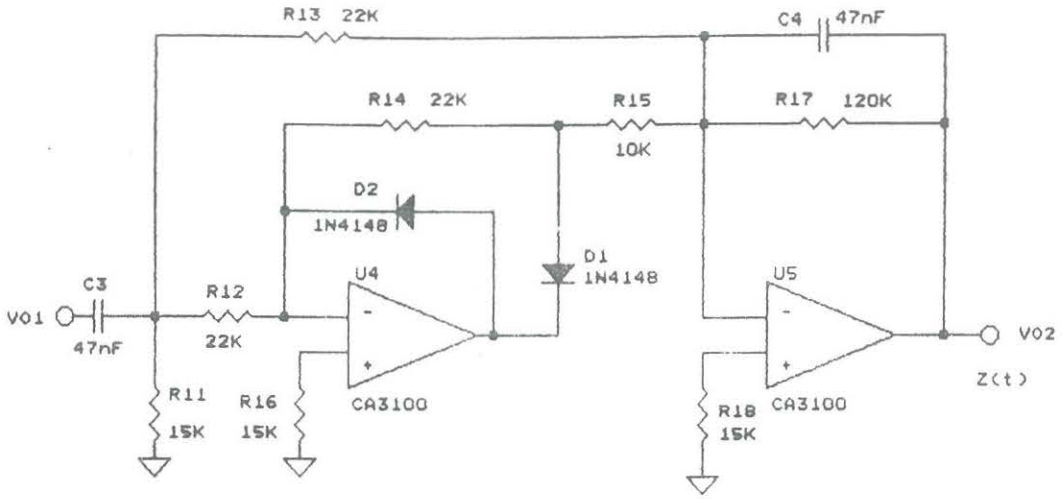
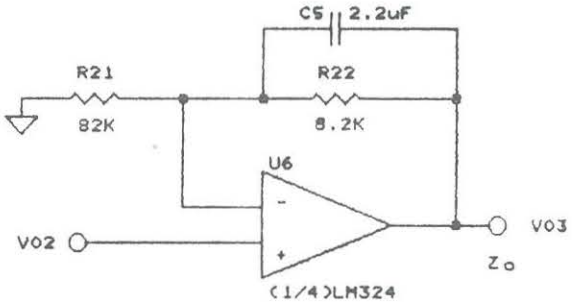


Fig. 4.3 Instrumentation amplifier for ICG.

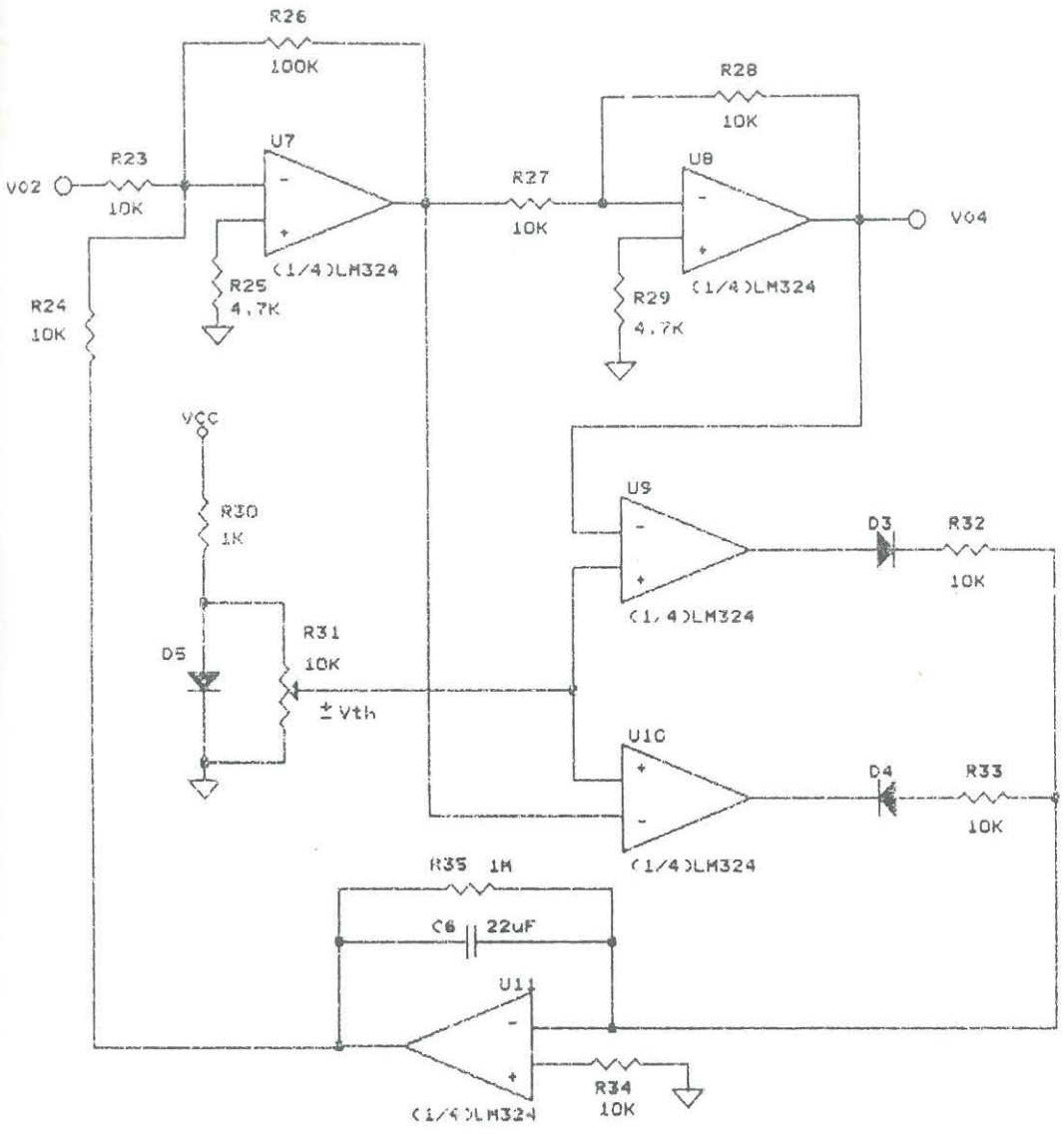


DEMODULATOR (PRECISION RECTIFIER)



LOW PASS FILTER

Fig. 4.4 Demodulator and low pass filter for ICG.



DC cancellation circuit.

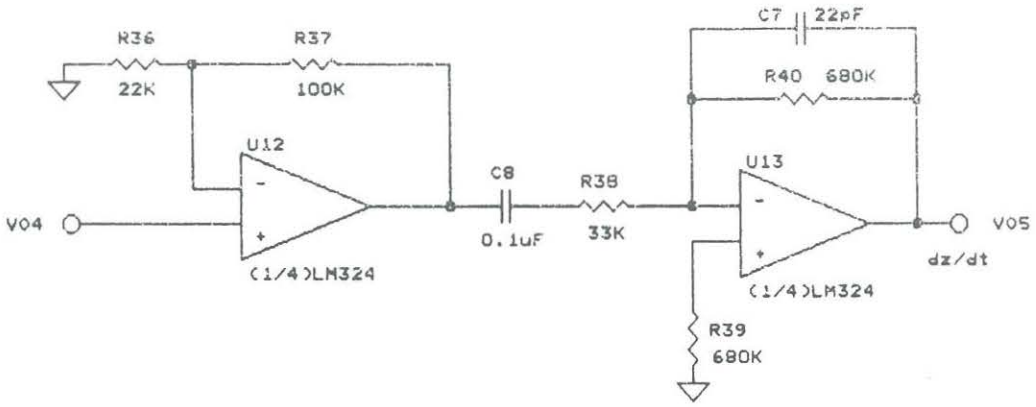


Fig. 4.6 Amplifier and differentiator to get dz/dt.

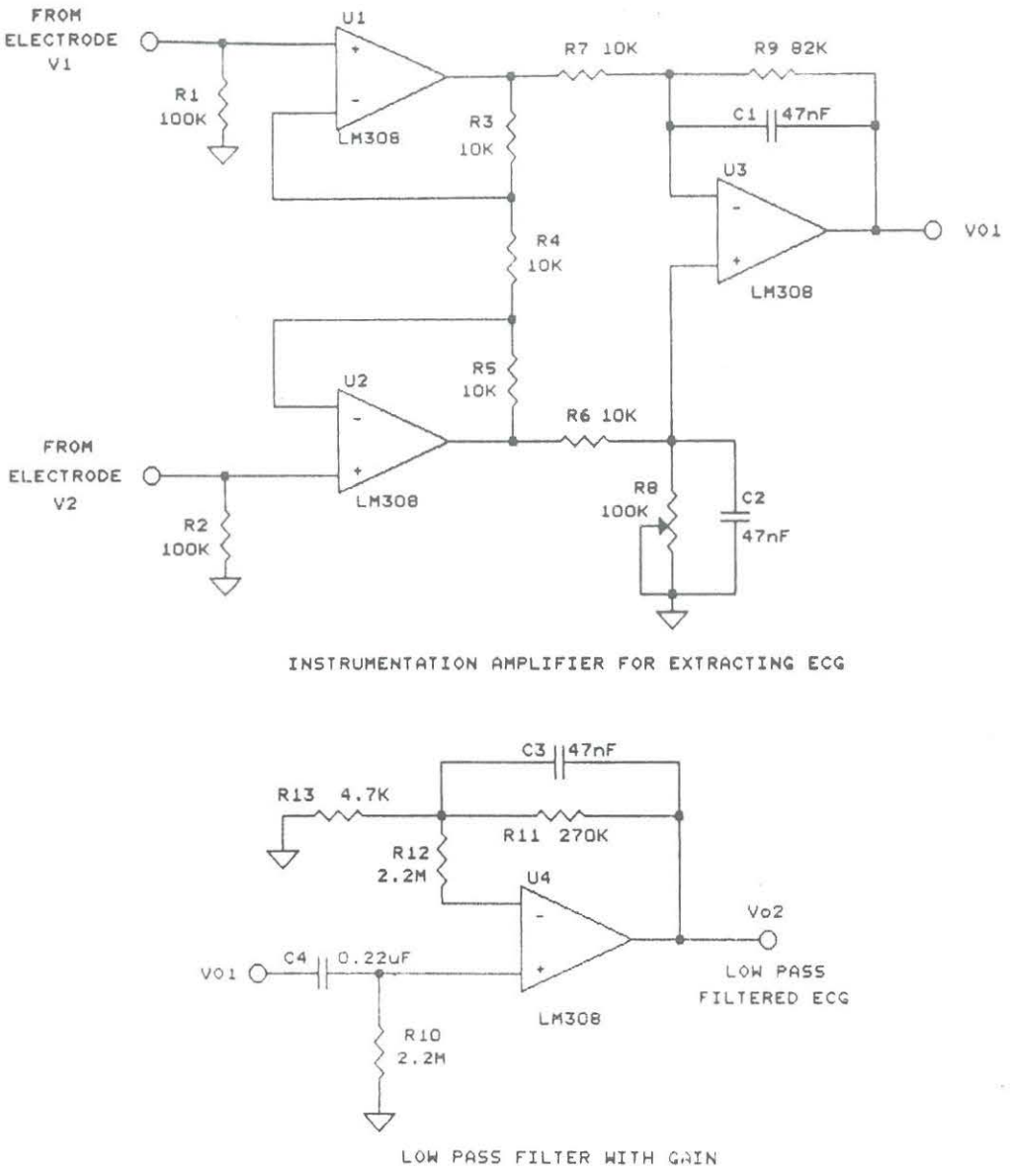


Fig. 4.7 ECG extraction circuit.

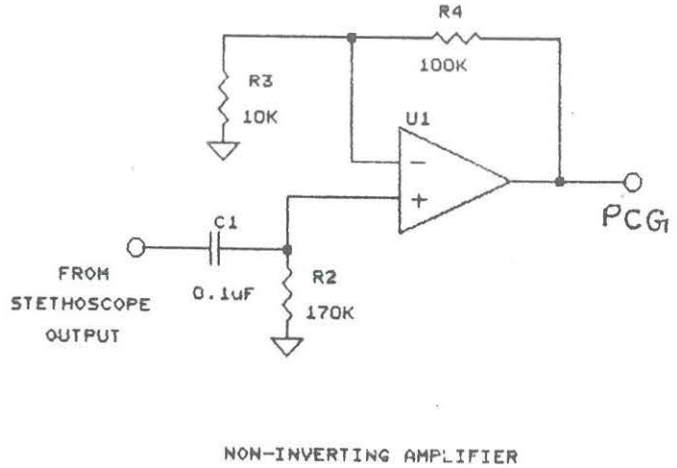


Fig. 4.8 Amplifier for PCG.

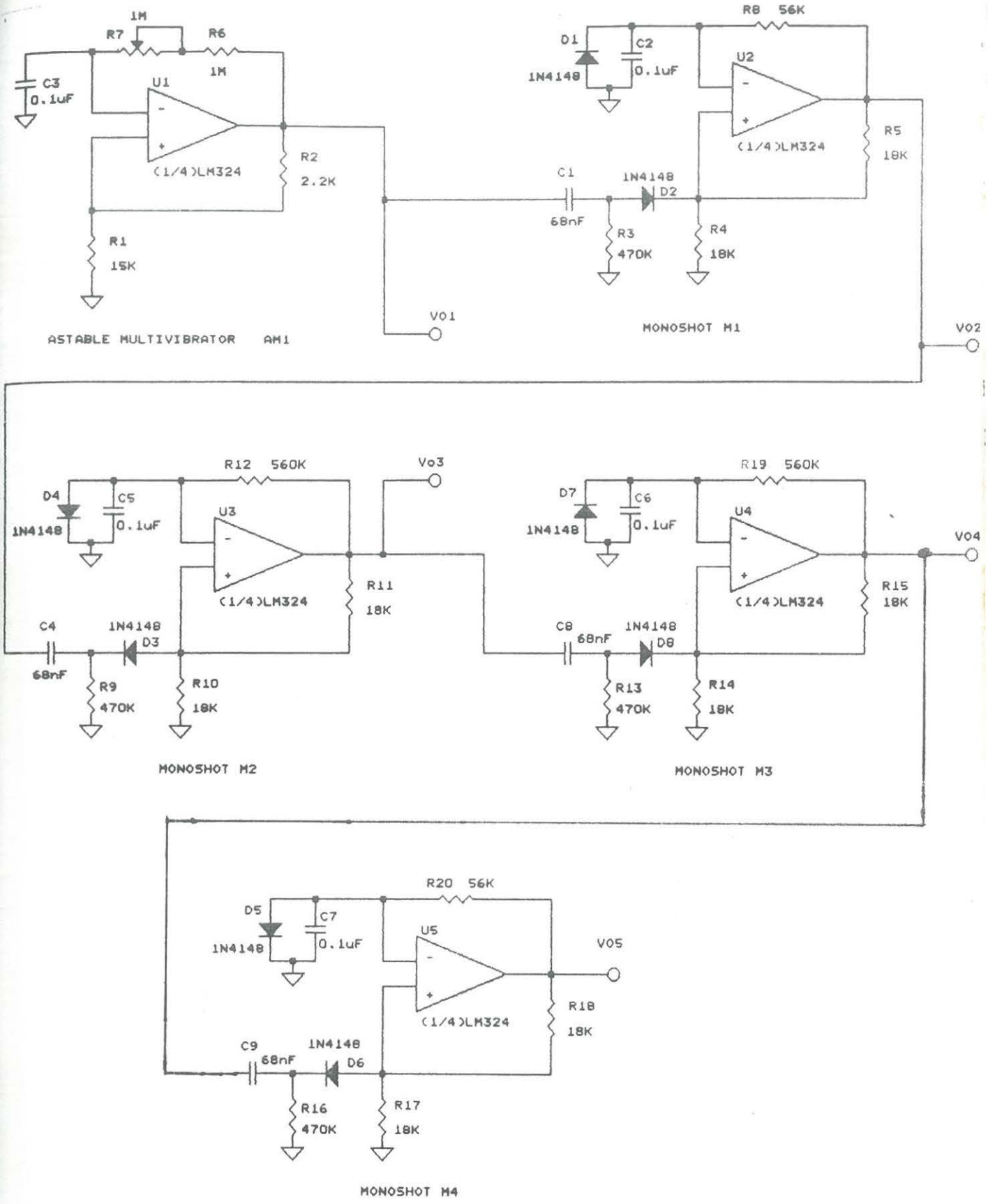


Fig. 4.9a Circuits of the monoshots for ECG simulator (refer Fig. 3.3a).

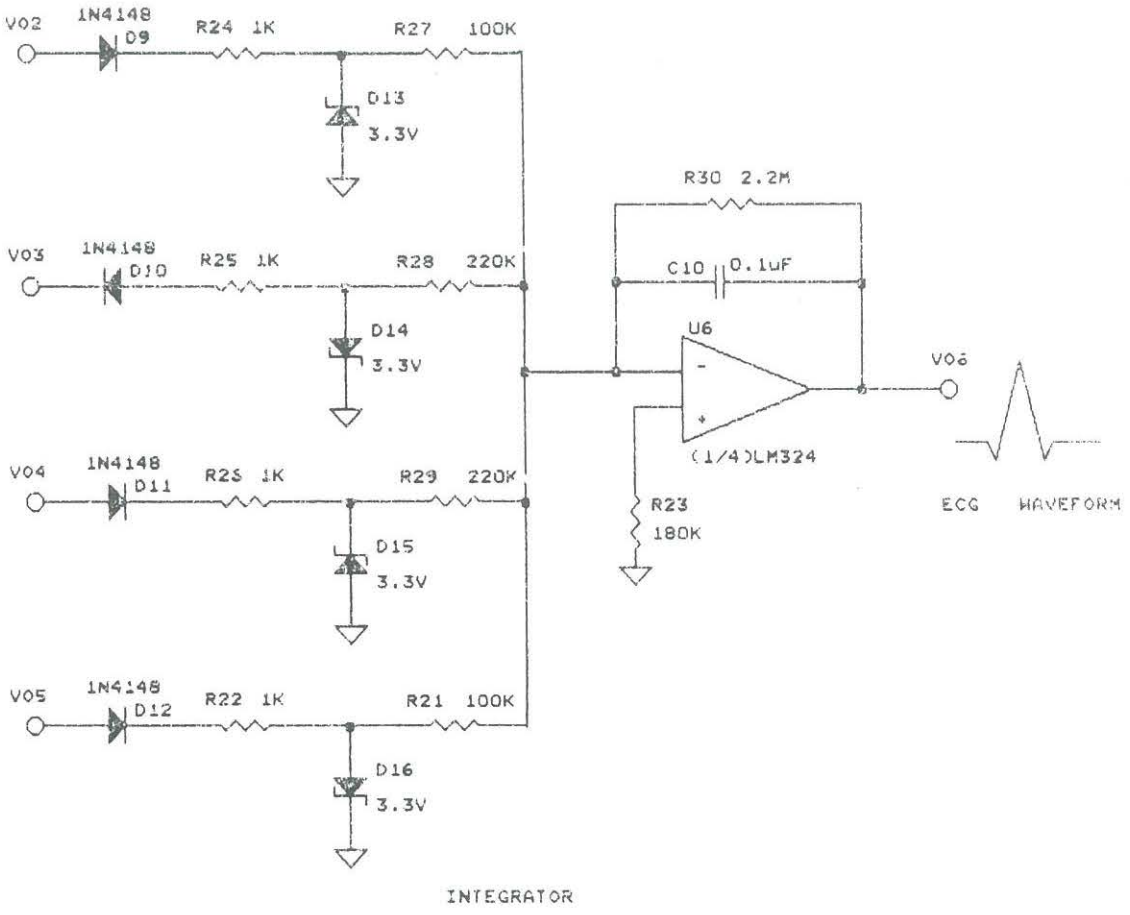


Fig. 4.9b Circuit of the integrator for ECG simulator (refer Fig. 3.3a).

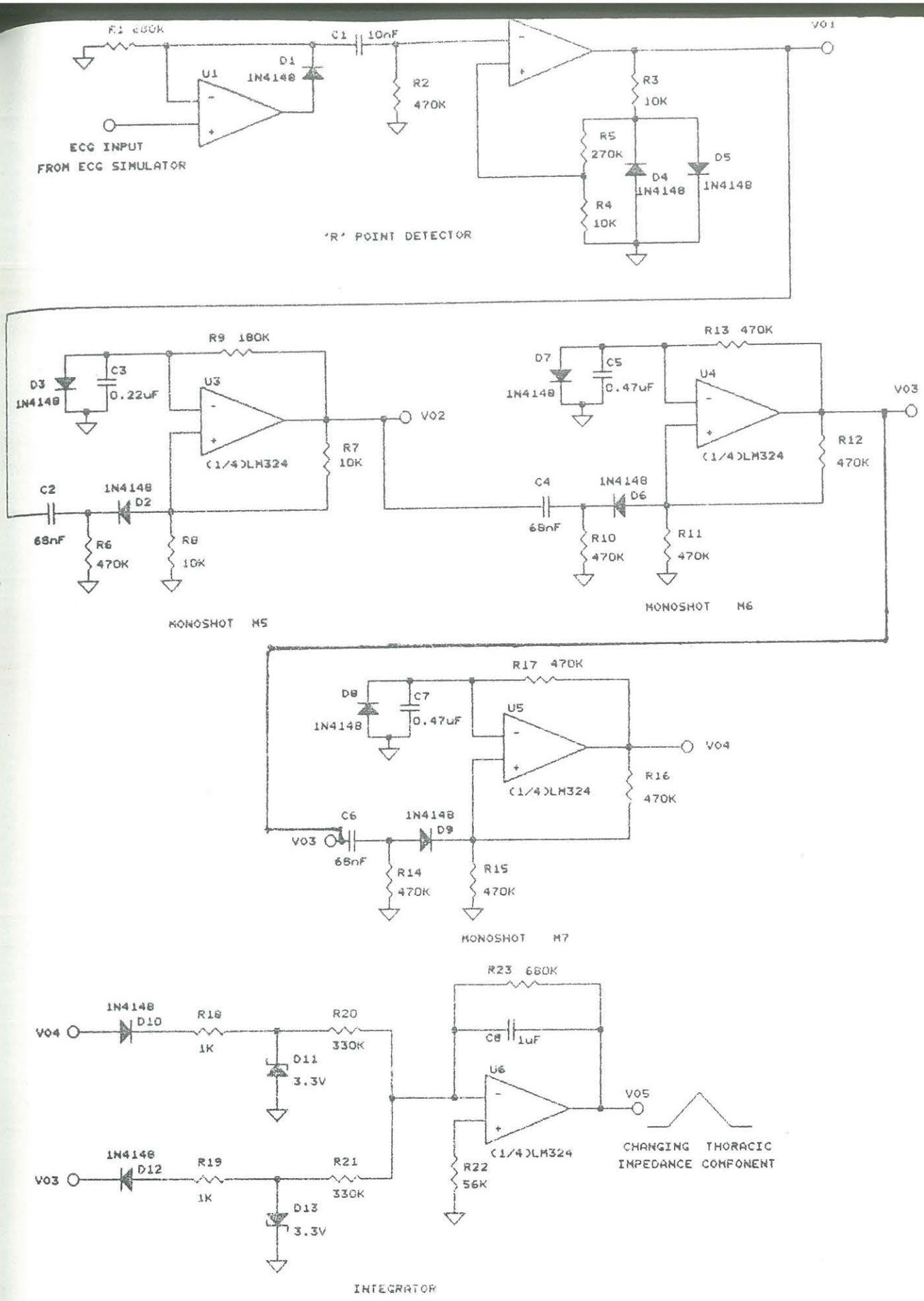
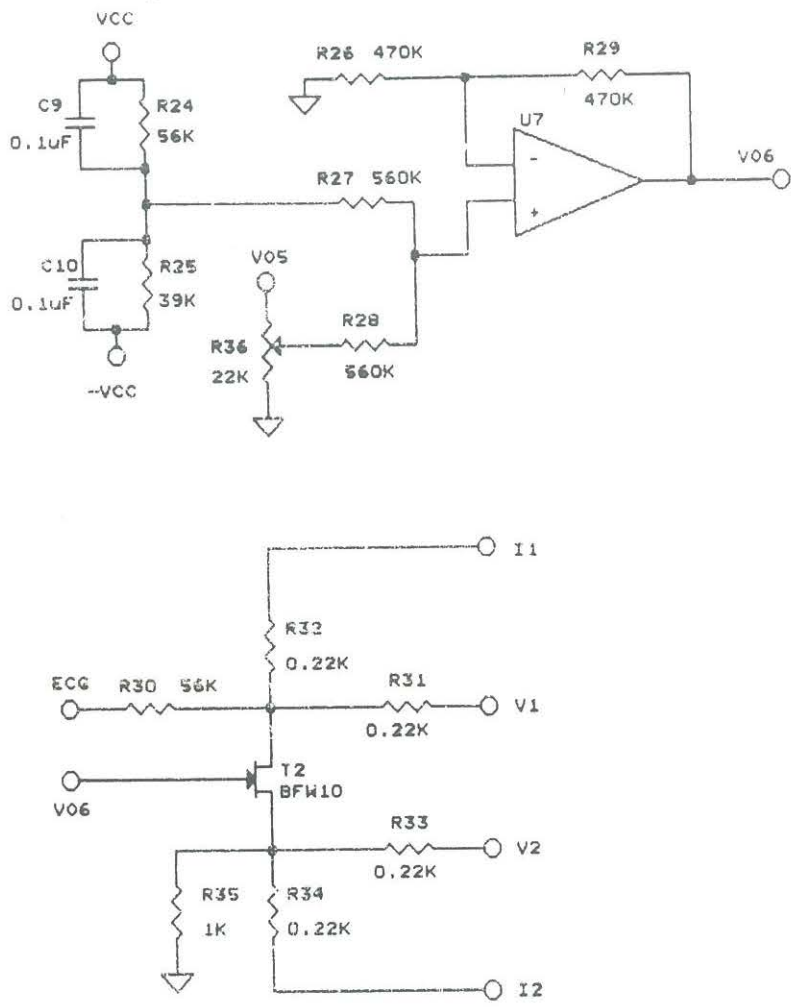
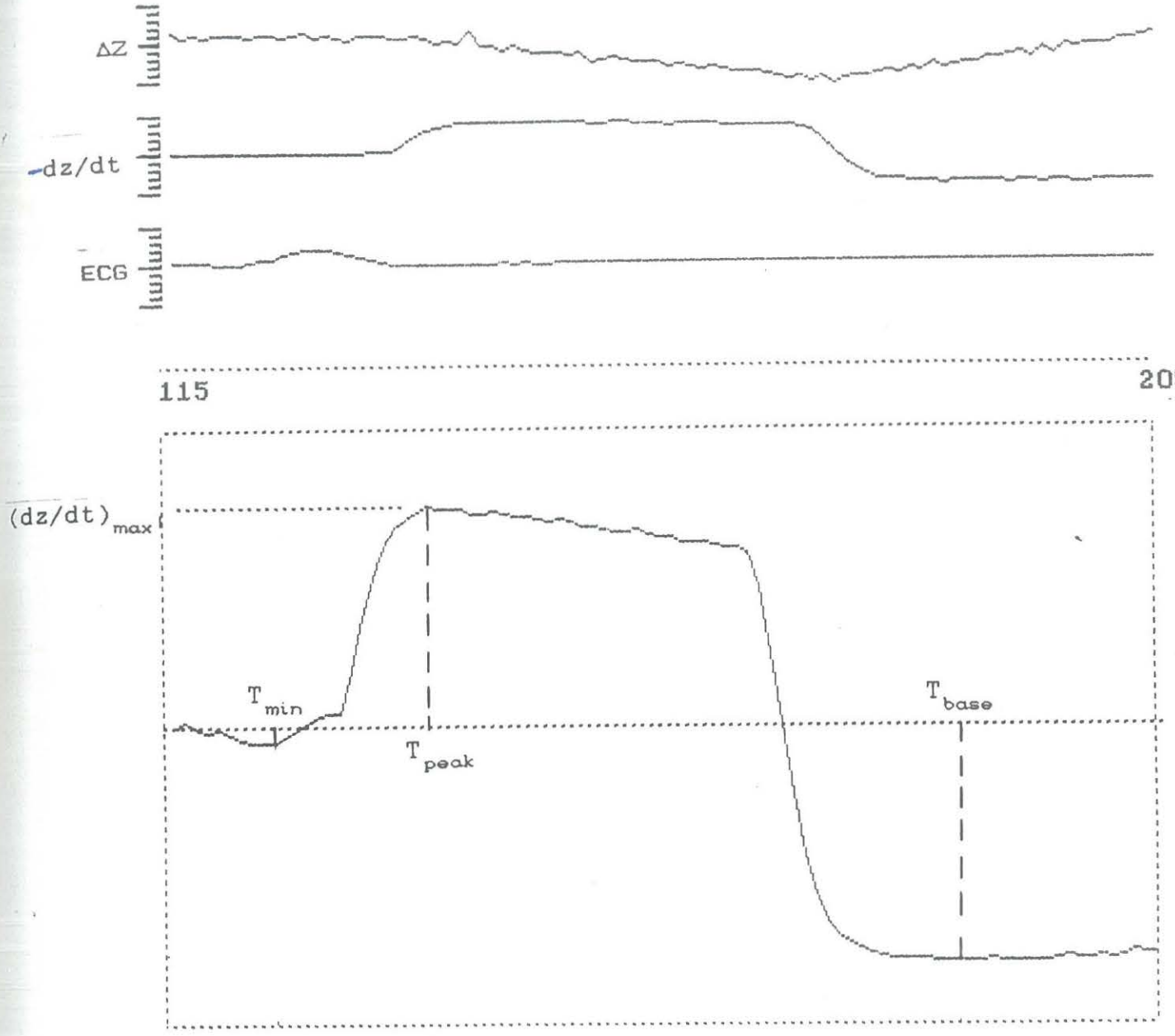


Fig. 4.10a Circuit generating the triangular impedance waveform (from the input ECG) for impedance simulator.



THORAX IMPEDANCE SIMULATOR

Fig. 4.10b Circuit diagram of impedance simulator.



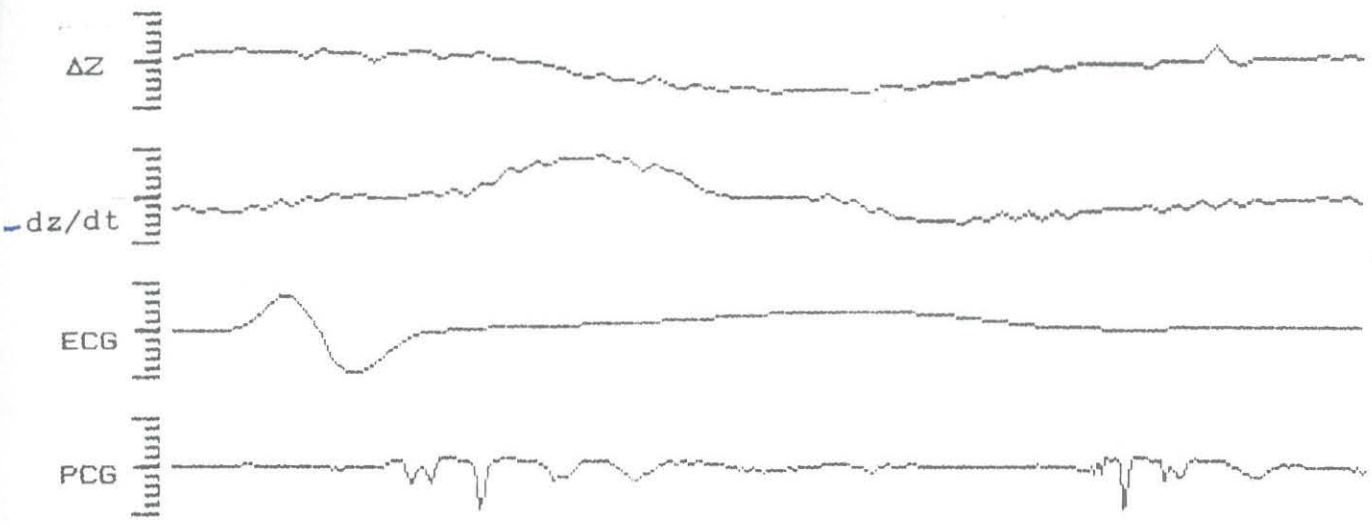
Averaged dz/dt waveform

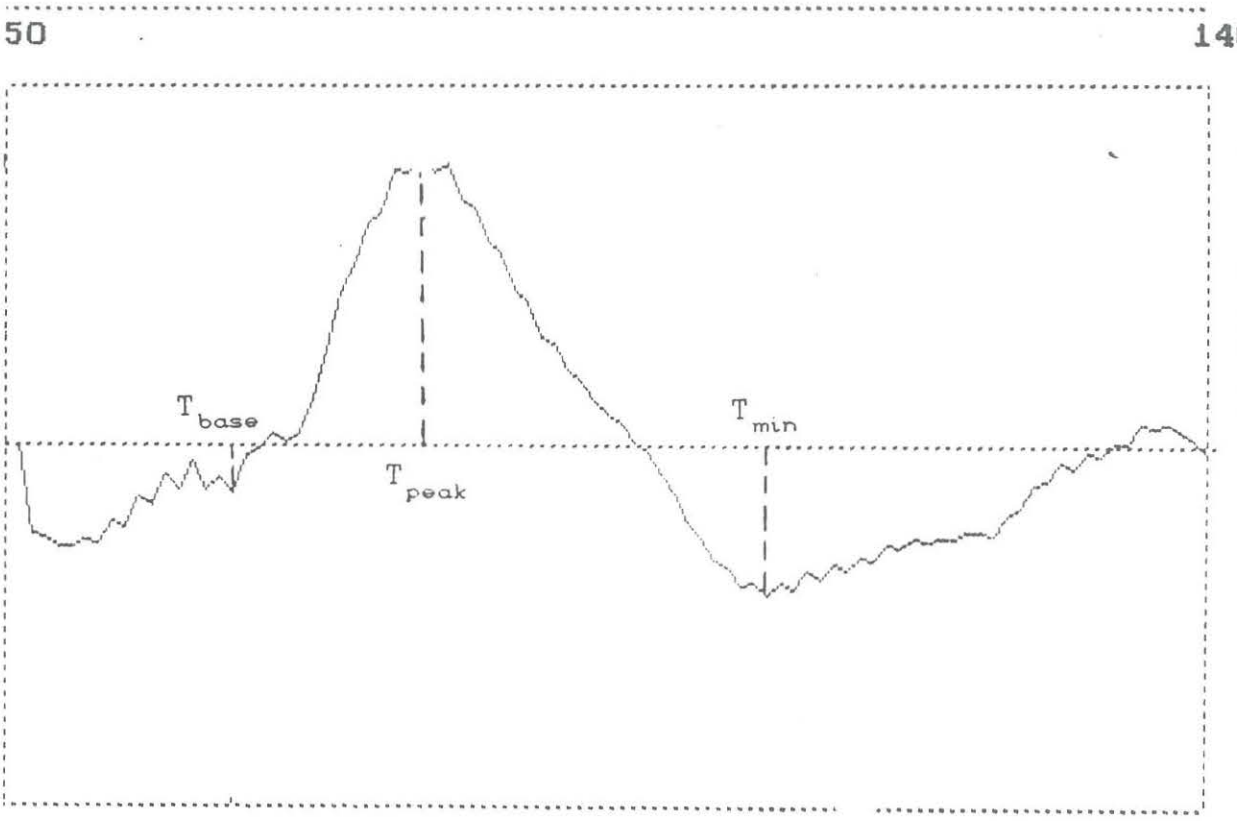
Screen dump of a section of the recorded parameters from sample numbers 115 to 205.

$(dz/dt)_{max}$ = 1.6 V
Heart beat rate = 68 beats/min
Zo = 20 ohms

Ampitude spans
ECG : 5.5 V
dz/dt : 1.54 ohms/sec
ΔZ : 16e-03 ohms

Fig. 5.1 Waveforms recorded using the thorax simulator.





Averaged dz/dt waveform

Screen dump of a section of the recorded parameters from sample numbers 50 to 140.

Heart beat rate = 90 beats/min
Zo = 14.8 ohms
T_{lvet} = 0.28 Sec
Stroke volume = 39.0 ml
Cardiac output = 3.5 lit/min

Amplitude spans
ECG : 5.5 V
dz/dt : 2.54 ohms/sec/V
ΔZ : 16e-03 ohms

Fig.5.2 Typical recordings of ECG, ΔZ, dz/dt, and PCG obtained during rest from subject NMH

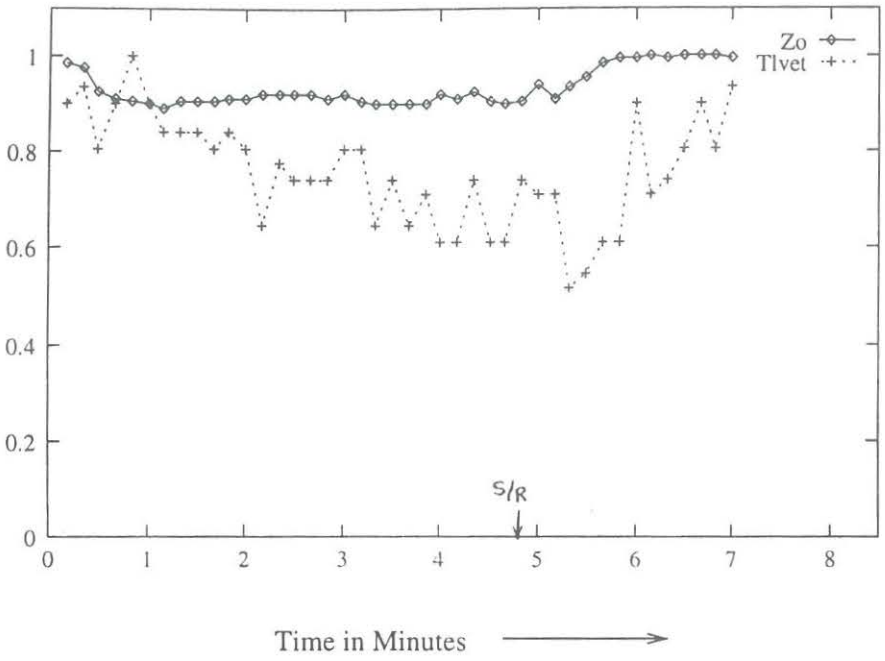


Figure 5.3 : Graph of variation in Z_o and T_{lvet} for subject NMH (Normalised values of Z_o and T_{lvet} are plotted. Z_o max. = 15 ohms, and T_{lvet} max. = ~~1.06~~ 0.31 sec.)

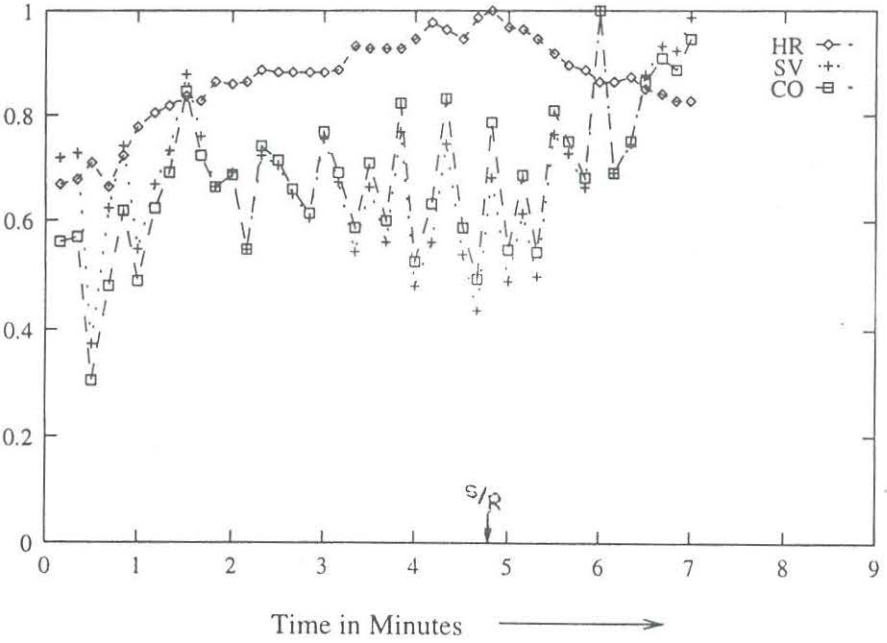


Figure 5.4 : Graph of variation in HR, SV, and CO for subject NMH (Normalised values of HR, SV and CO are plotted. HR max. = 134 beats/mins., , SV max. = 54 ml., and CO max. = 6.28 lit./min.)

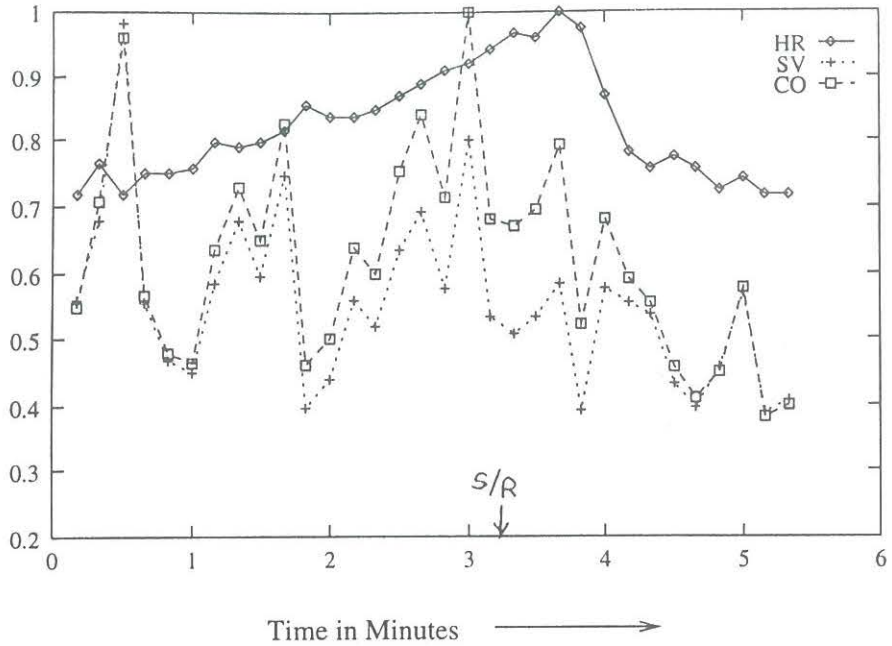


Figure 5.5 : Graph of variation in HR, SV, and CO for subject KSP (Normalised values of HR, SV, and CO are plotted. HR max. = 124 beats/mins. , SV max. = 26.16 ml., and CO max. = 2.24 lit./min.)

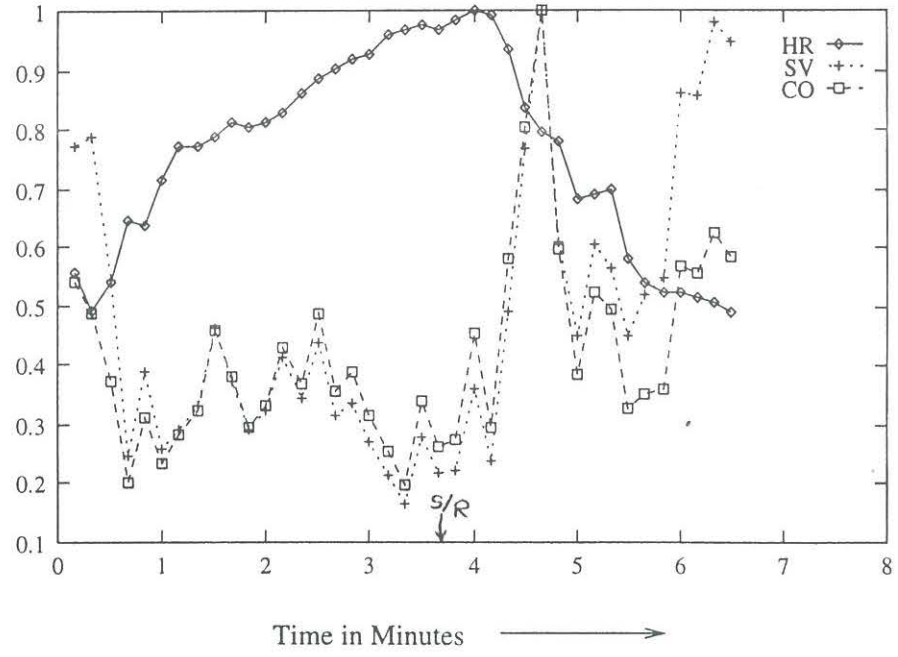


Figure 5.6 : Graph of variation in HR, SV, and CO for subject SAP (Normalised values of HR, SV and CO are plotted. HR max. = 122 beats/mins. , SV max. = 117.2 ml., and CO max. = 11.3 lit./min.)

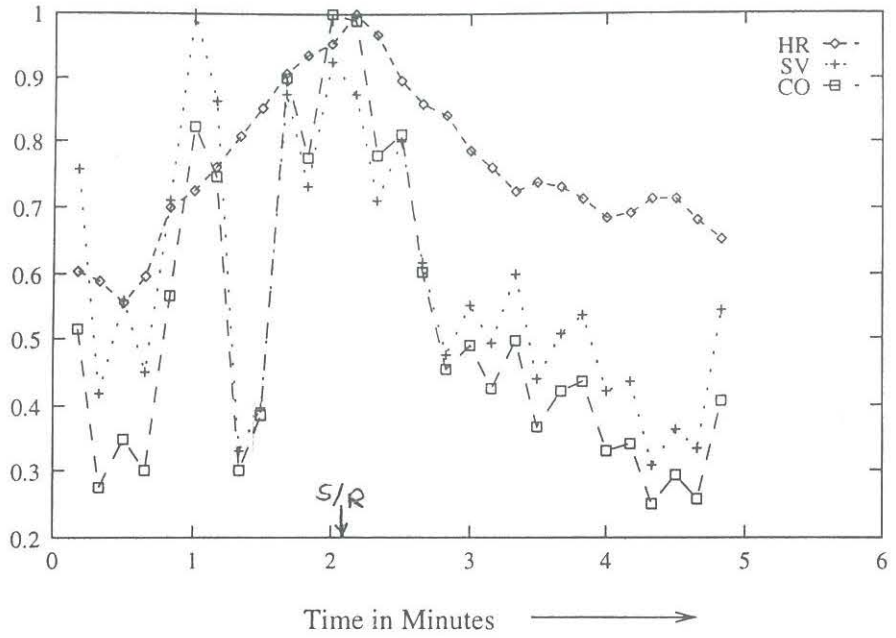


Figure 5.7 :Graph of variation in HR, SV, and CO for subject MRP (Normalised values of HR, SV and CO are plotted. HR max. = 150 beats/min. , SV max. = 31.7 ml., and CO max. = 4.2 lit./min.)

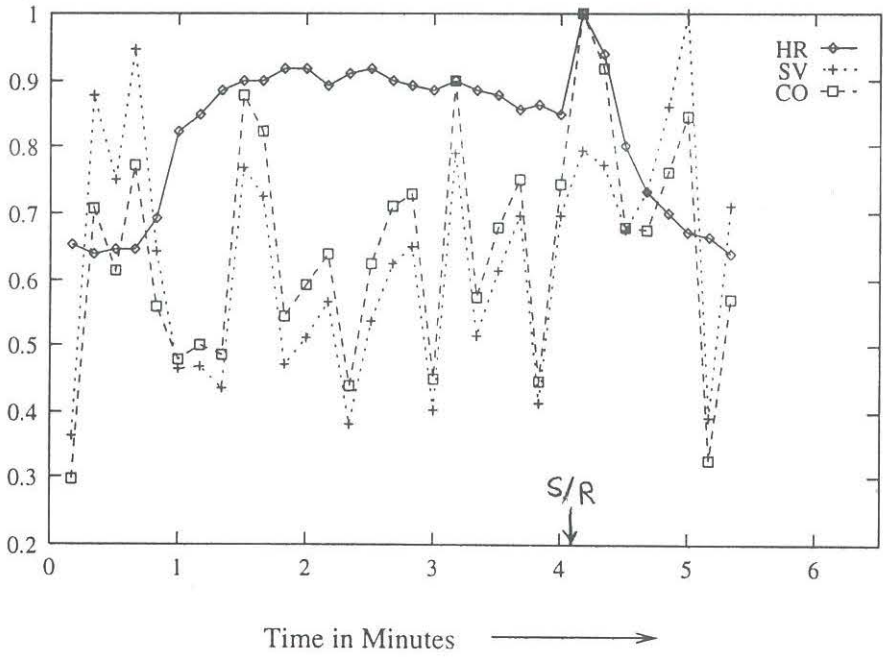


Figure 5.8 :Graph of variation in HR, SV, and CO for subject MRV (Normalised values of HR, SV and CO are plotted. HR max. = 130 beats/min. , SV max. = 39 ml., and CO max. = 4.0 lit/min.)

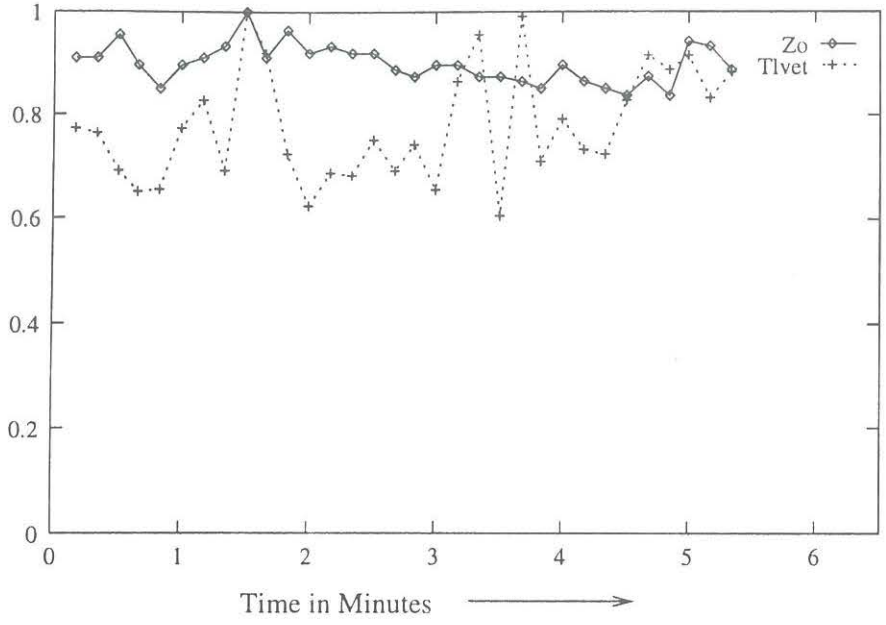


Figure 5.9 : Graph of variation in Z_o and T_{lvet} for subject SVP (Normalised values of Z_o and T_{lvet} are plotted. Z_o max. = 14.5 ohms and T_{lvet} max. = 0.28 sec.)

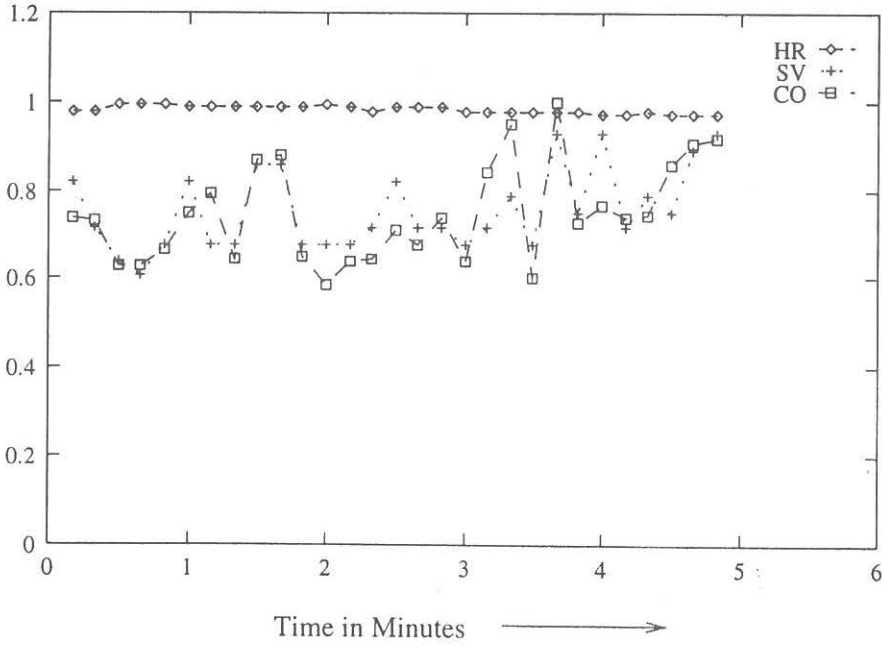


Figure 5.10 : Graph of variation in HR, SV, and CO for subject SVP (Normalised values of HR, SV, and CO are plotted. HR max. = 87 beats/mins., SV max. = 24.6 ml., and CO max. = 1.8 lit./min.)

APPENDIX A

Component Selection Criteria for Thorax Simulator Circuit

1 ECG generator circuit (refer Fig.4.9)

1.1 Astable multivibrator AM1 (refer Fig.4.9a)

Period of square wave of AM1 = T s.

where, (0.5 s > T < 0.8 s).

$T = 2(R6 + R7)C3 \ln((1+\beta)/(1-\beta))$,

and $\beta = R1 / (R1 + R2)$.

With R2 = 2.2K and R1 = 18K,

$\beta = 0.89$.

Let C3 = 0.1uF.

Condition 1

T = 0.5, giving (R6 + R7) = 1M,

and Condition 2

T = 0.8, giving (R6 + R7) = 1.9M.

Select R6 = 1M, giving R7 = 1M (potentiometer)

1.2 Monoshots M1, M2, M3, and M4 (refer Fig.4.9a)

The period of monoshots M1 and M4 is 4 msec. and that of M2 and M3 is 40 msec. Monoshots M1 and M4 are similar. So design of only monoshot M1 is explained. Period of monoshot M1, (T) is given by,

$T = RC \ln(1 + Rx/Ry)$

where Rx = R5, Ry = R4, R = R8 and C = C2

Let C2 = 0.1 uF, and R4 = R5 = 18K

For T = 4 ms,

R3 = 57K = 56K

Monoshots M2 and M3 are similar. So design of only monoshot M2 is explained. Period of monoshot M2, (T) is given by,

$T = RC \ln(1 + Rx/Ry)$

where Rx = R11, Ry = R10, R = R12, and C = C5.

Let C5 = 0.1uF, and R11 = R10 = 10K.

For T = 40 msec,

R12 = 577K = 560K

1.3 Integrator (refer Fig.4.9b)

This integrator has a input charging current of magnitude fixed at 15 uA. To get a peak voltage of 5V at 'R' point of ECG (V_r = 5V), we have,

V_r = (15 uA)(30 msec)(1/C10),

giving C10 = 0.9 uF = 0.1 uF.

2 Thoracic impedance simulator (refer Fig.4.10a and Fig.4.10b)

2.1 'R' point detector circuit

Differentiator C1,R2 has an active frequency (F_1) of 30 Hz.

$F_1 = 1 / (2\pi R2C1)$

Taking C1 = 0.1 uF, we get R2 = 530K = 470K.

2.2 Monoshots M5, M6, and M7

The period of M5 is 20 msec. and that of M6 and M7 is 150 msec.

2.2.1 Design of Monoshot M5

T = 20 msec.

$T = RC \ln(1 + Rx/Ry)$

where Rx = Ry = R7 = R9 = 10K, R = R9 and C = C3.

This gives R9C3 = (20/ln(2))msec, for T = 20 msec.

Choose C3 = 0.22 uF, giving R9 = 196 K = 180 K.

2.2.2 DESIGN OF MONOSHOTS M6 AND M7

Since both monoshots M6 and M7 are similar only the design of M6 is explained.

T = 150 msec, where $T = RC \ln(1 + Rx/Ry)$.

With Rx = Ry = R11 = R12 = 470K, R = R13, and C = C5 we get,

R13C5 = (150/ln(2))msec.

Choose C5 = 0.47 uF, giving R13 = 520 K = 470 K.

2.2.3 Integrator (Fig.4.10b)

The triangular wave output of this integrator should have a peak amplitude of 1.5 V to present a proper base drive for FET T2. The peak voltage of the integrator (V_p = 1.5 V) is given by,

$$V_p = (V_{D11} / R20)\ (150\ msec)\ (1/C8)$$

where V_{d11} is zener break-down voltage = 3.3 V.

Let C8 = 1 uF, giving R20 = 330K.

APPENDIX B

PCB Layouts for Signal Conditioner and Thorax Simulator Hardware

Fig.B.1 Circuit diagram of excitation and ECG extraction circuit

Fig.B.2 Component side layout of excitation and ECG extraction circuit

Fig.B.3 Solder side layout of excitation and ECG extraction circuit

Fig.B.4 Component silk of excitation and ECG extraction circuit

Fig.B.5 Circuit diagram of Z_0 and dz/dt extraction circuit

Fig.B.6 Component side layout of Z_0 and dz/dt extraction circuit

Fig.B.7 Solder side layout of Z_0 and dz/dt extraction circuit

Fig.B.8 Component silk of Z_0 and dz/dt extraction circuit

Fig.B.9 Circuit diagram of ECG simulator

Fig.B.10 Component side layout of ECG simulator

Fig.B.11 Solder side layout of ECG simulator

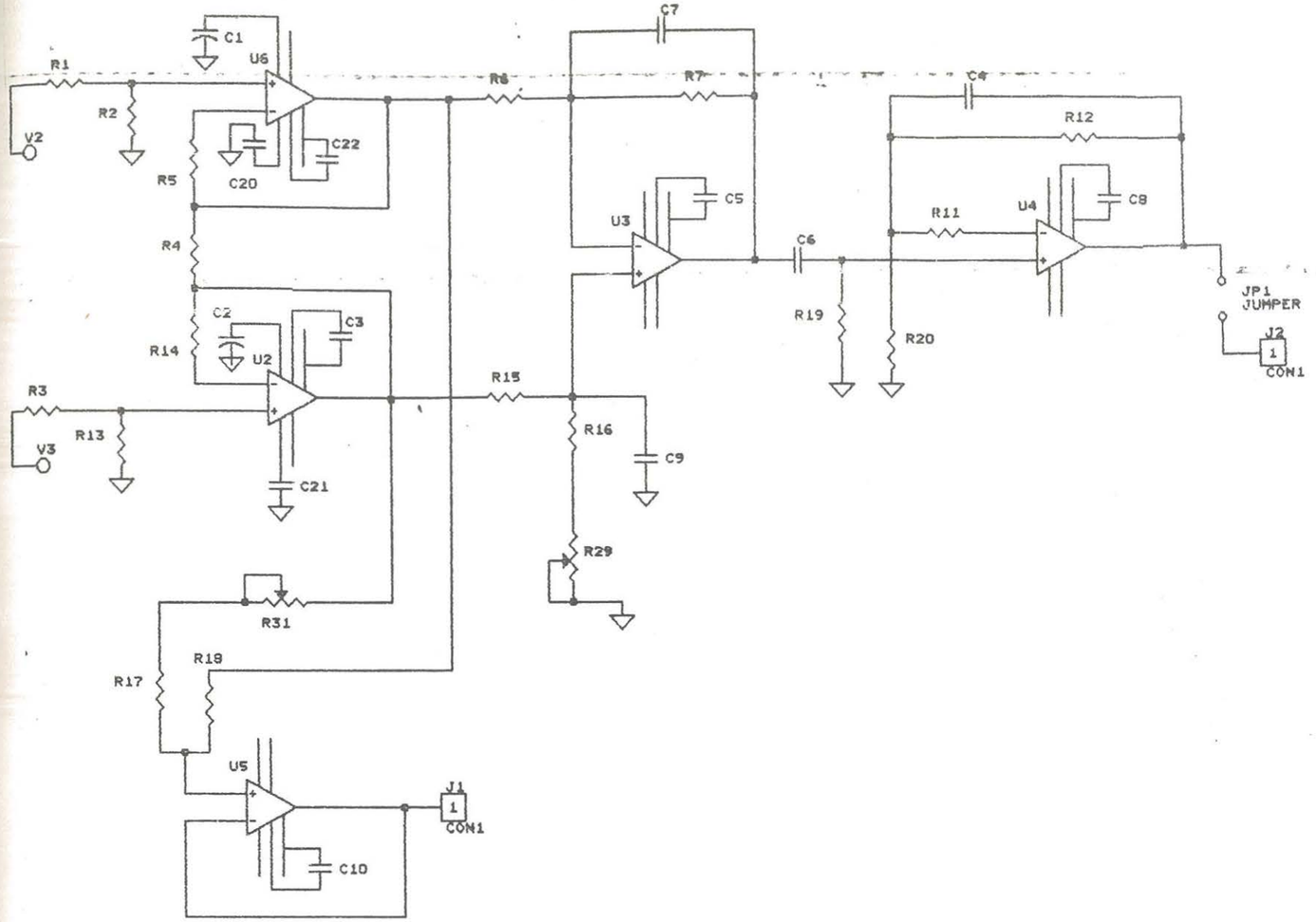
Fig.B.12 Component silk of ECG simulator

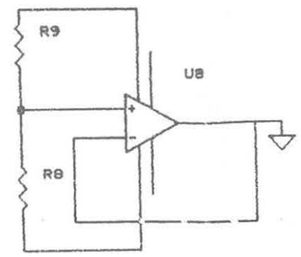
Fig.B.13 Circuit diagram of impedance simulator

Fig.B.14 Component side layout of impedance simulator

Fig.B.15 Solder side layout of impedance simulator

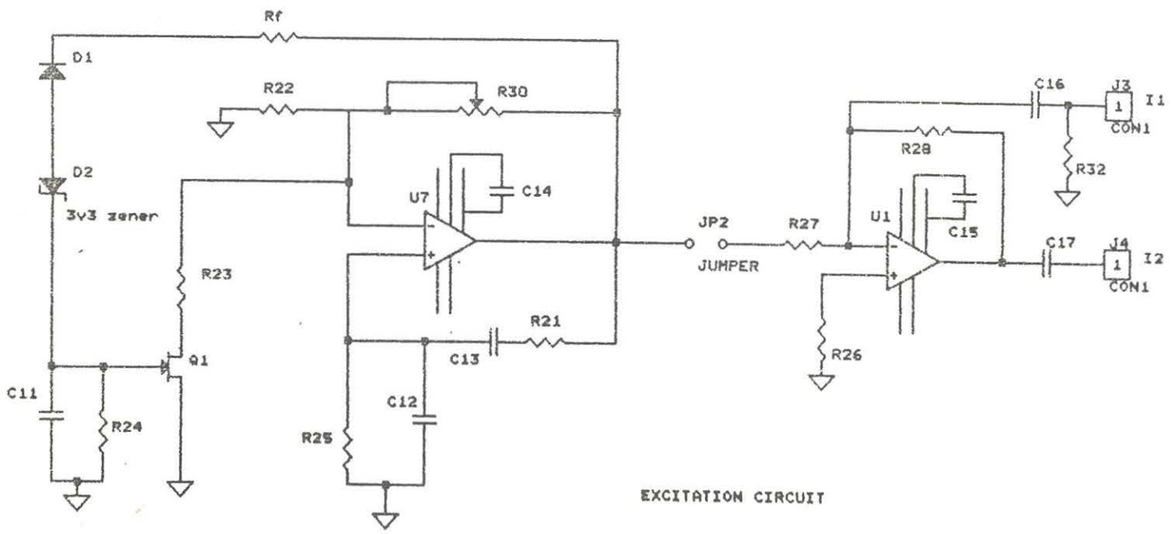
Fig.B.16 Component silk of impedance simulator





GROUND GENERATION CIRCUIT

ECG EXTRACTION CIRCUIT



EXCITATION CIRCUIT

Fig. B.1

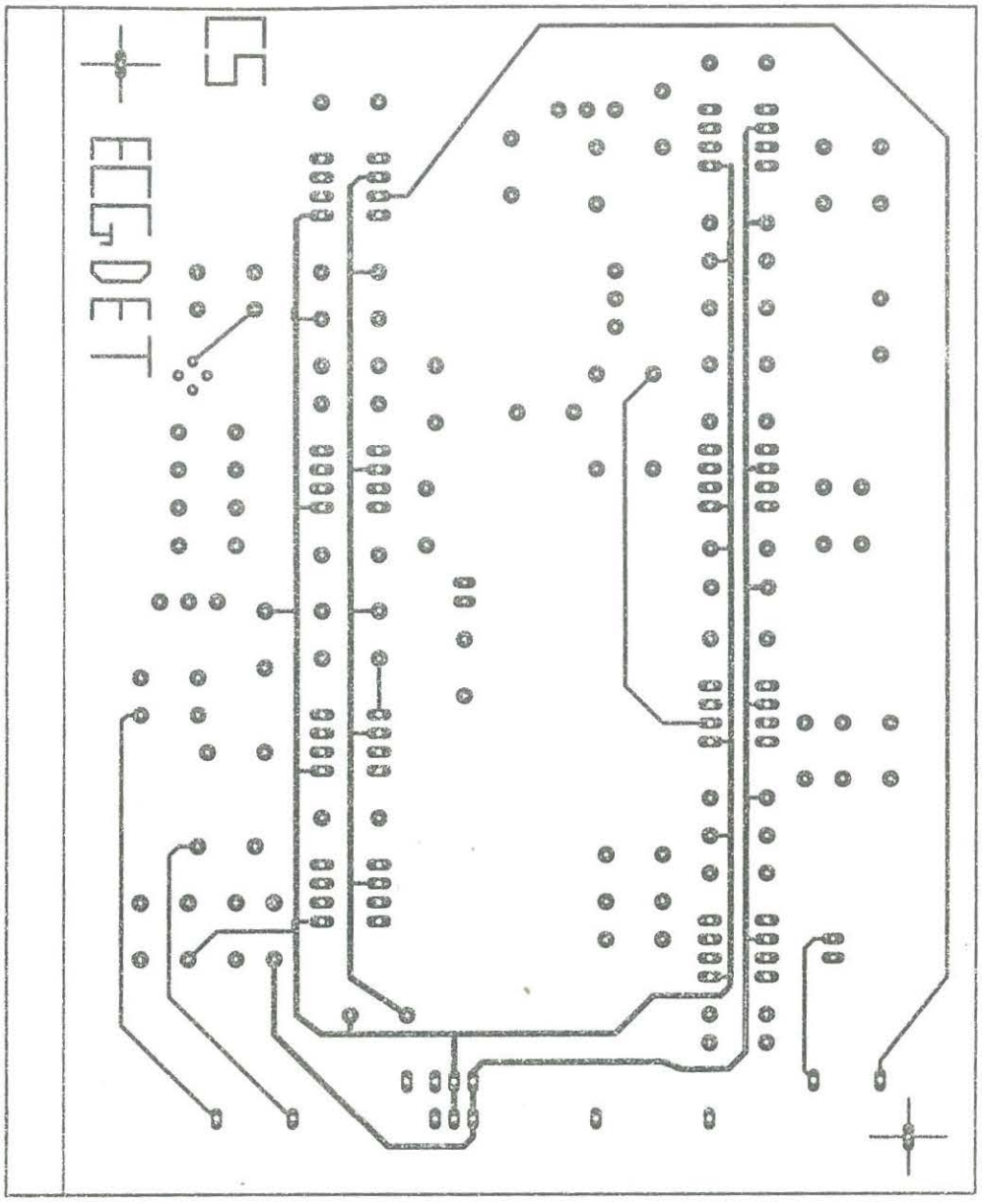


Fig. B.2

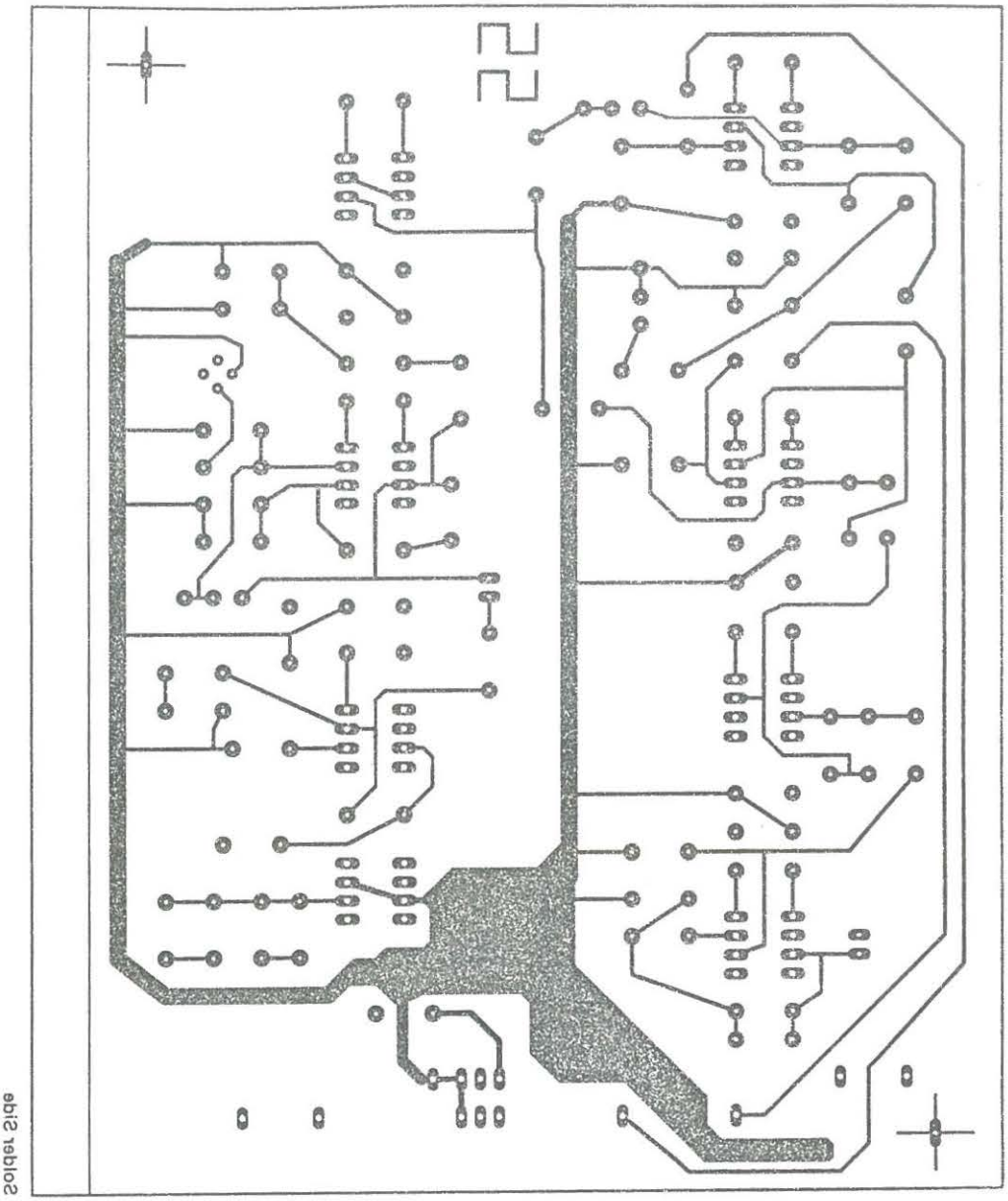


Fig. B. 3

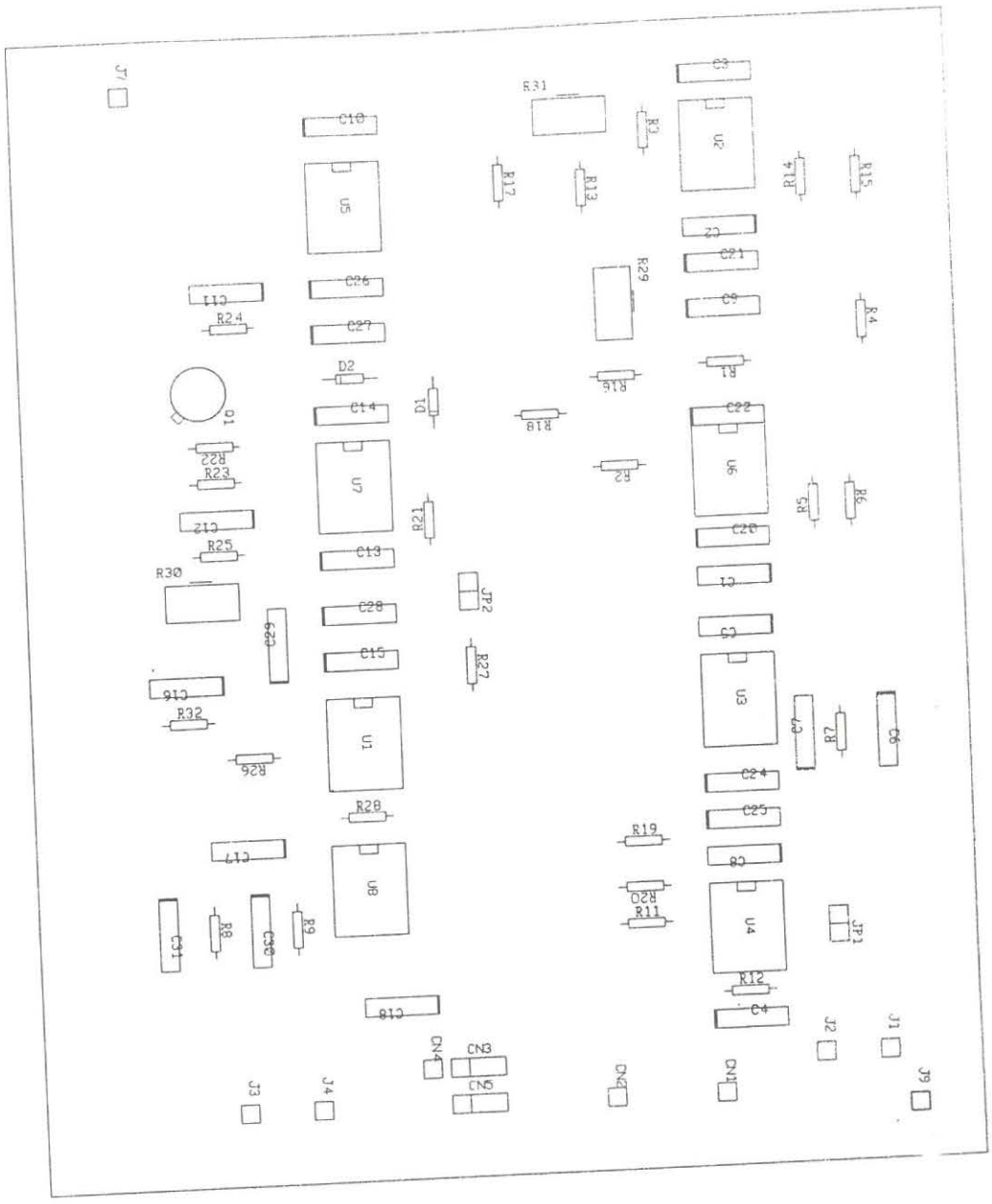

J7
C10
U5
C26
C27
C11
R24
D2
Q1
C14
R22
R23
U7
C12
R25
C13
R30
C28
C29
C15
C16
R32
U1
R26
R20
C17
U8
C31
R8
C30
R9
C18
CN4
CN3
CN5
J3
J4
R31
R3
U2
R17
R13
R14
R15
C3
C21
R29
C9
R4
R16
R1
D1
R18
C22
U6
R5
R6
C20
C1
C5
JP2
R27
U3
C7
R7
C6
C24
C25
C8
R19
R20
R11
U4
JP1
R12
C4
J2
J1
J9
CN2
CN1

Fig. B. 4

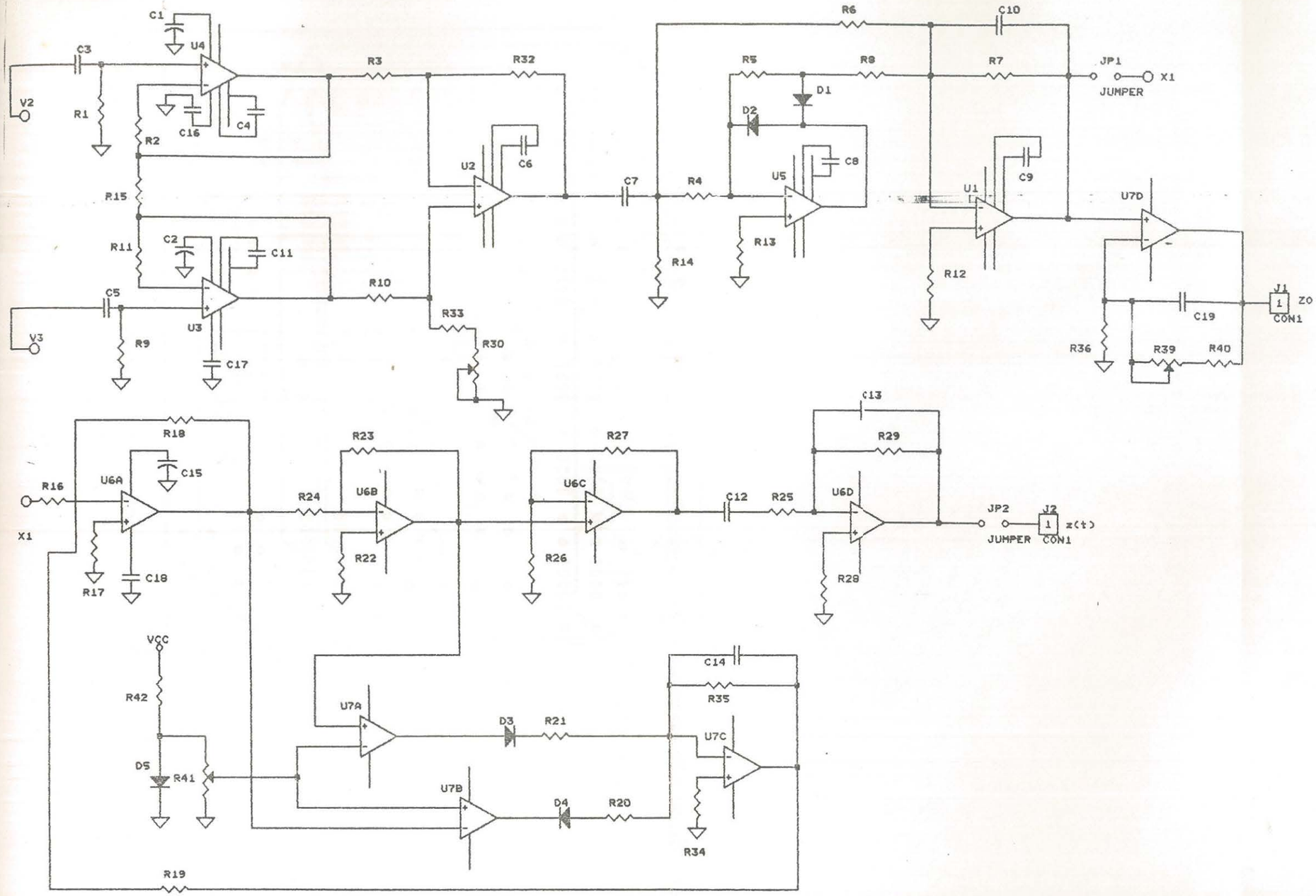

C1
C3
U4
R3
R32
R6
C10
R5
R8
R7
JP1
X1
JUMPER
V2
R1
C16
C4
R2
D1
D2
U2
C6
C7
R4
U5
C8
U1
C9
U7D
R15
R11
C2
C11
C5
R10
R33
R30
V3
R9
U3
C17
R14
R13
R12
R36
R39
R40
C19
J1
CON1
Z0
R18
C13
R29
R23
R27
U6A
C15
R16
U6B
U6C
C12
R25
U6D
JP2
J2
z(t)
X1
R24
R22
R26
R28
C18
R17
VCC
C14
R35
R42
U7A
D3
R21
U7C
D5
R41
U7B
D4
R20
R34
R19

Fig. B. 5

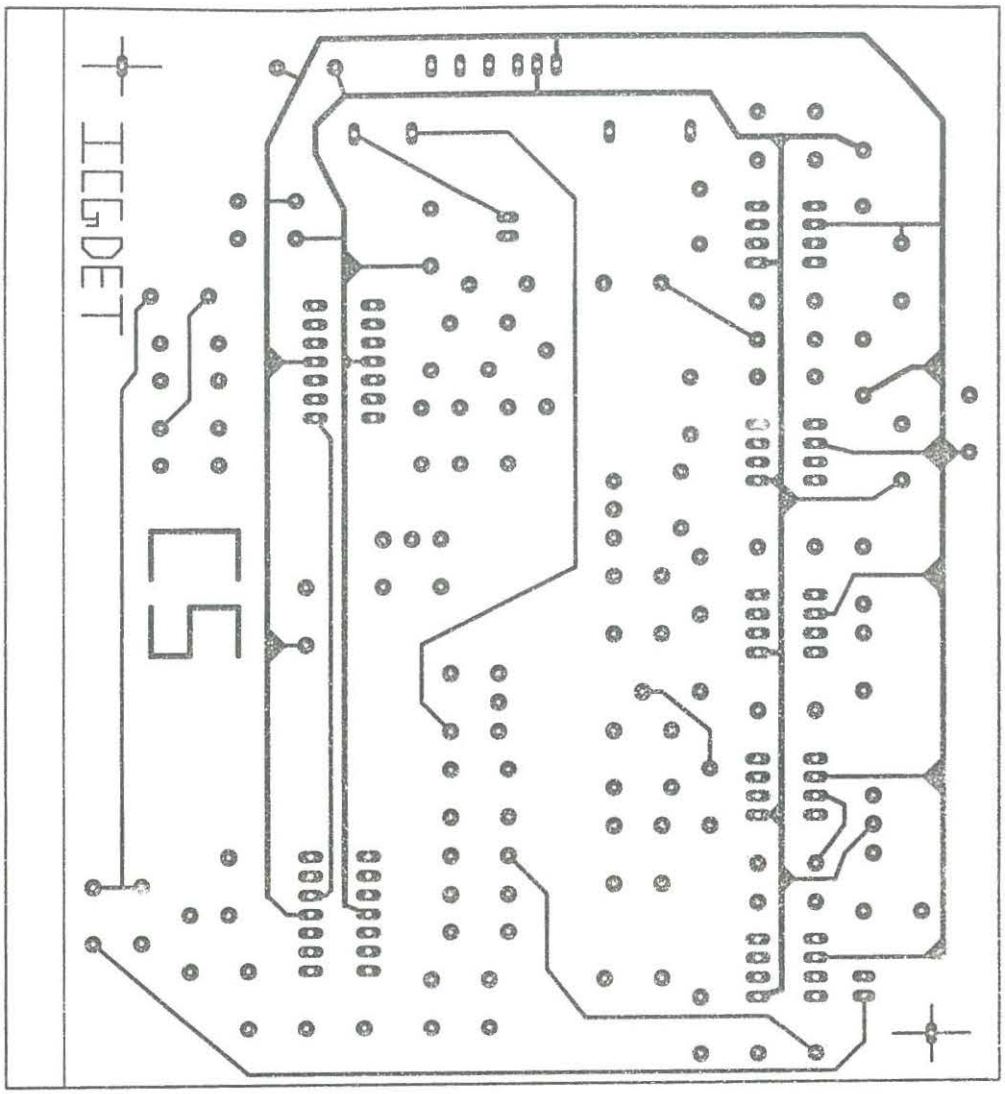


Fig. B. 6

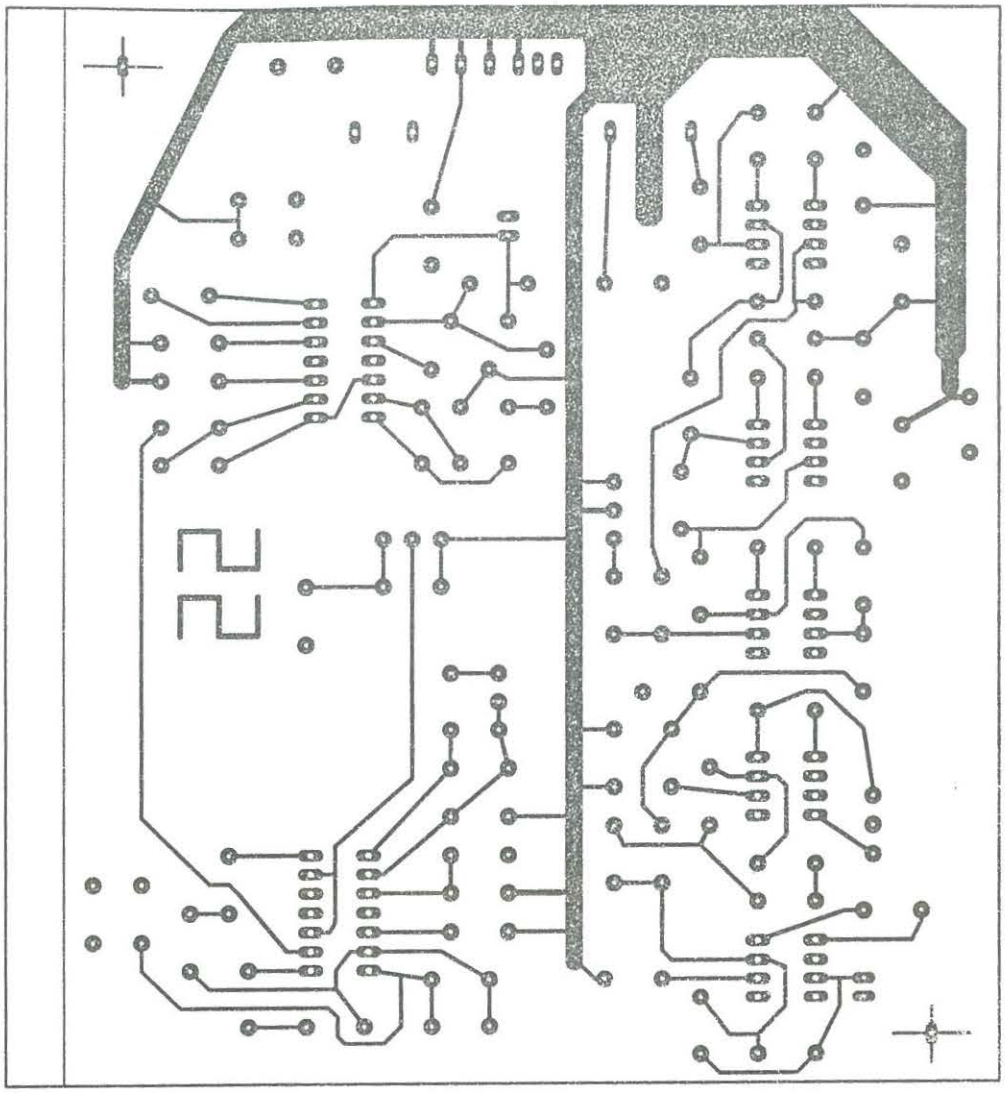


Fig. B. 7

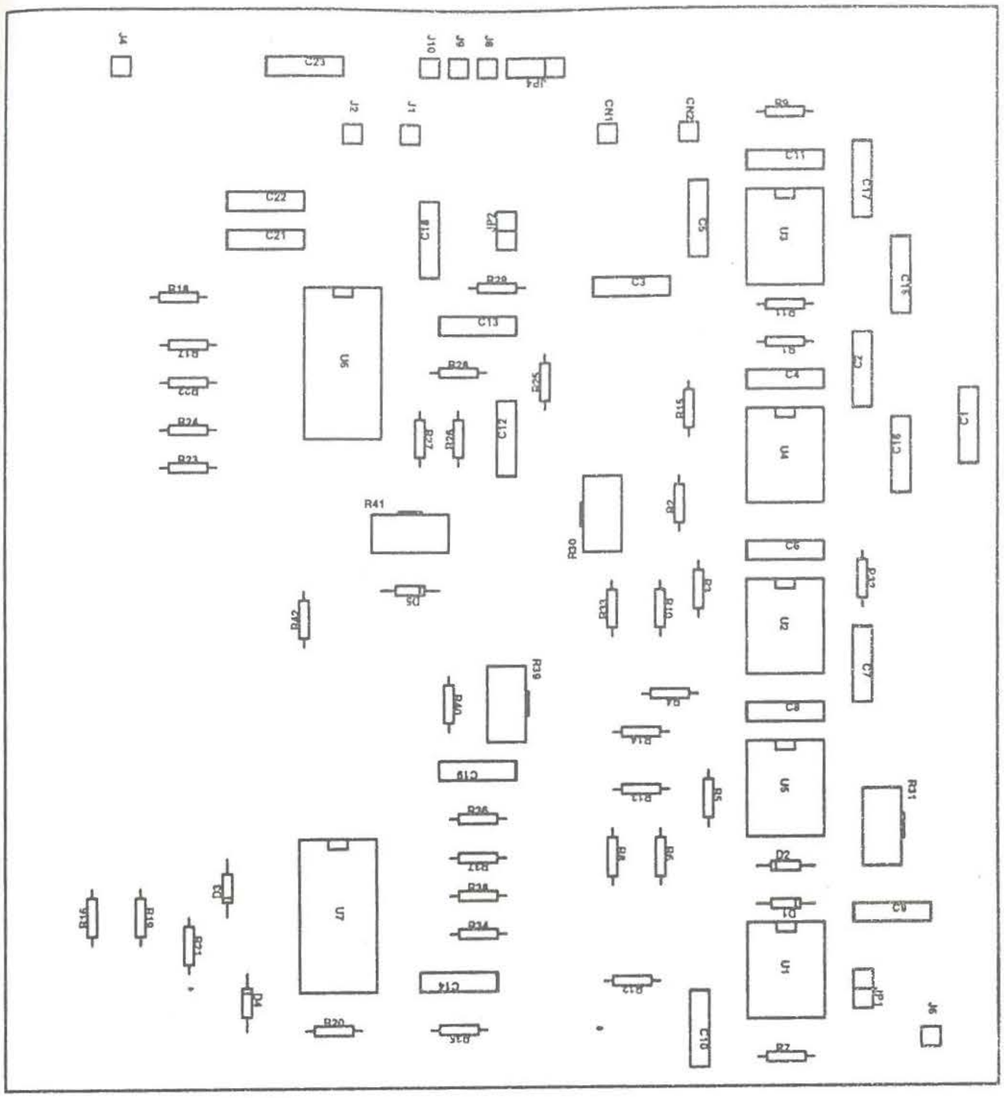


Fig. B. 8

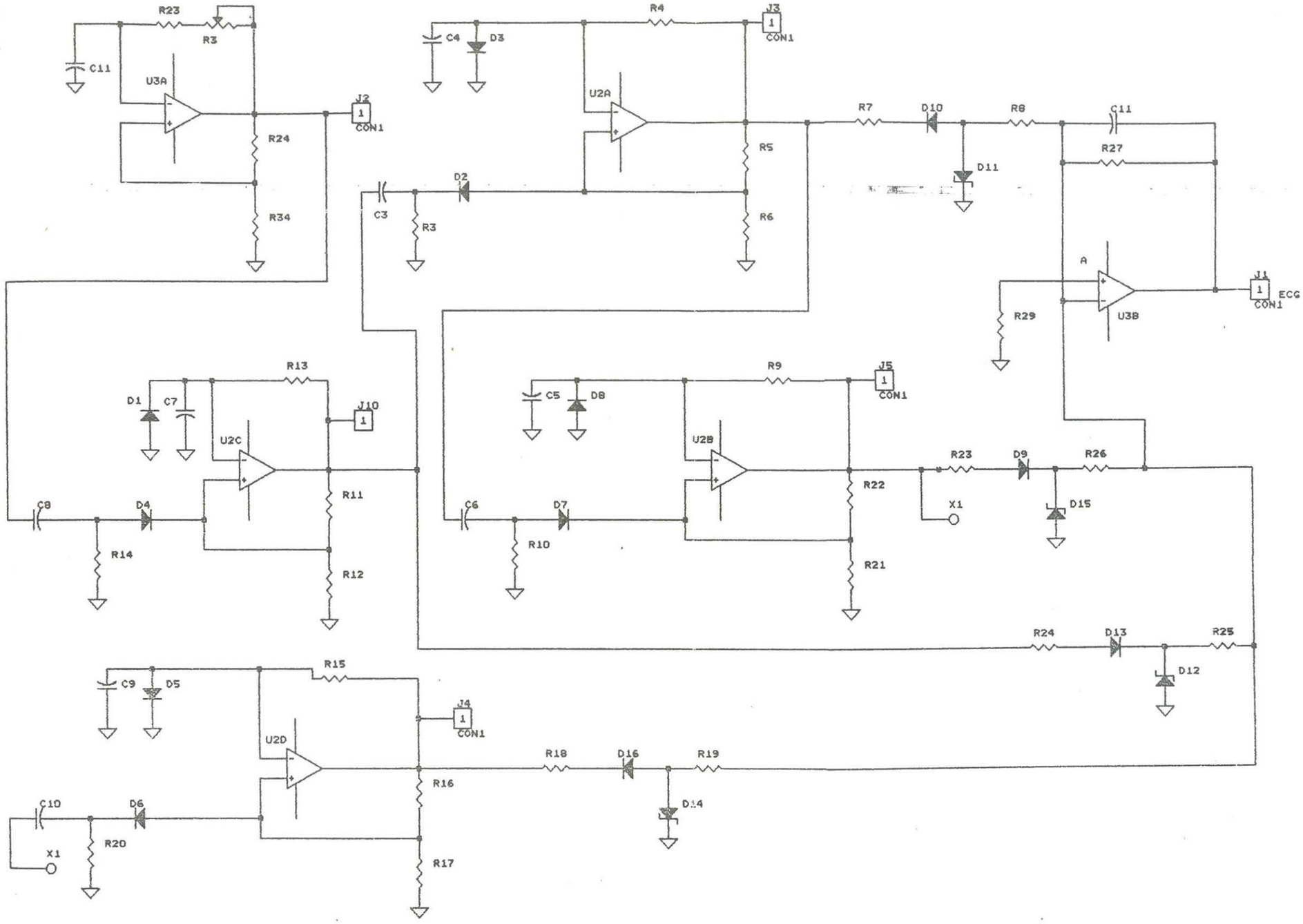

R23
R3
C11
U3A
J2
CON1
R24
R34
C4
D3
R4
J3
CON1
U2A
R7
D10
R8
C11
R5
R27
D2
C3
R3
R6
D11
A
J1
ECG
CON1
U3B
R29
R13
D1
C7
J10
U2C
R11
C8
D4
R14
R12
C5
D8
R9
J5
CON1
U2B
R23
D9
R26
R22
X1
D15
C6
D7
R10
R21
R24
D13
R25
D12
R15
C9
D5
J4
CON1
U2D
R18
D16
R19
R16
D14
C10
D6
X1
R20
R17

Fig. B. 9

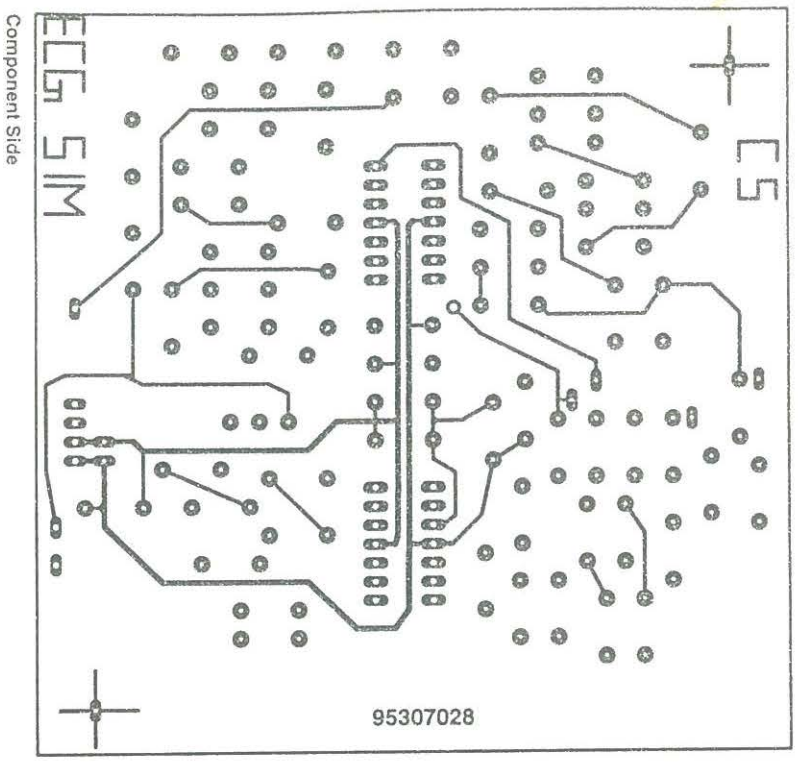


Fig. B.10

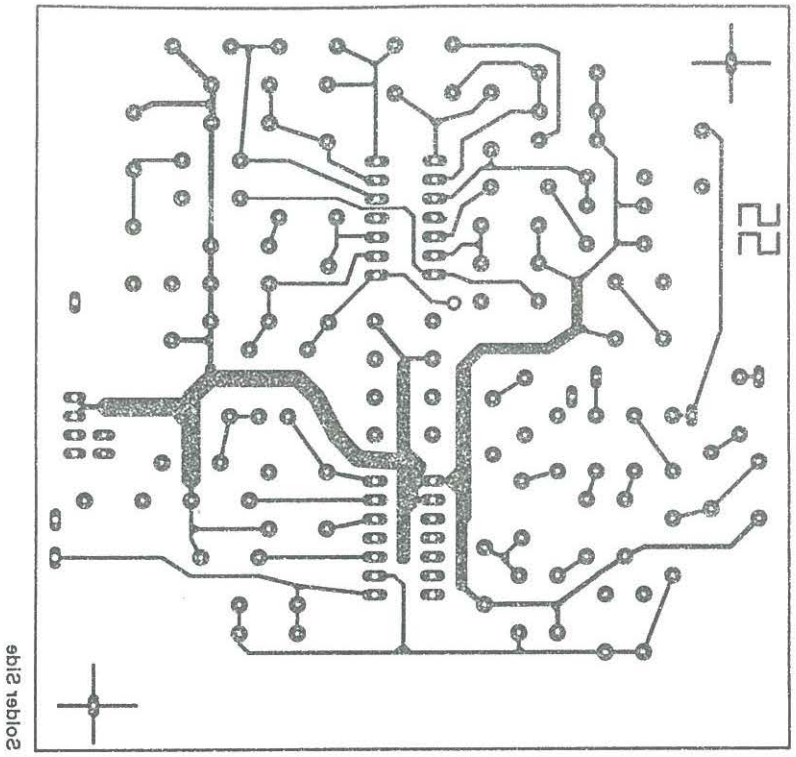


Fig. B. 11

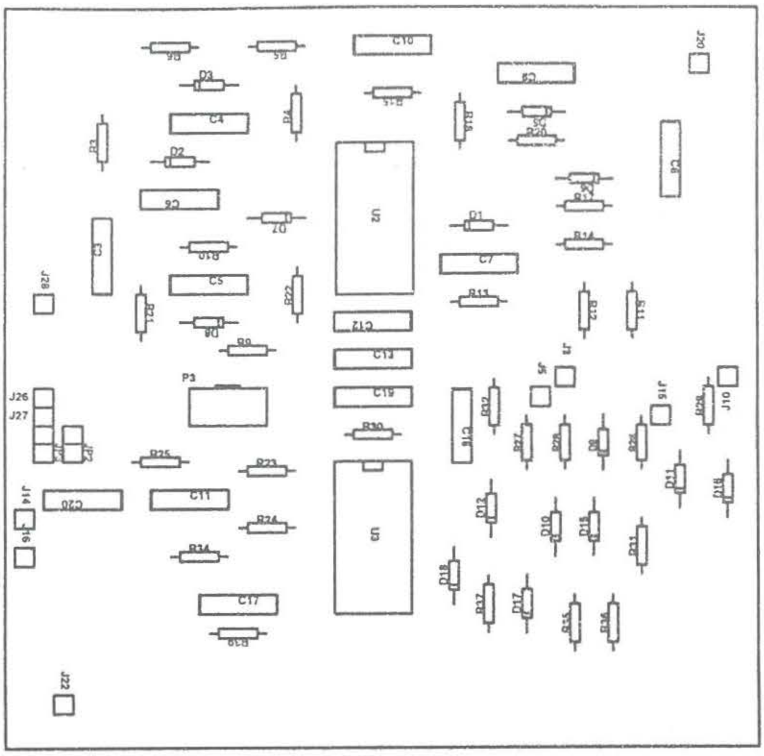
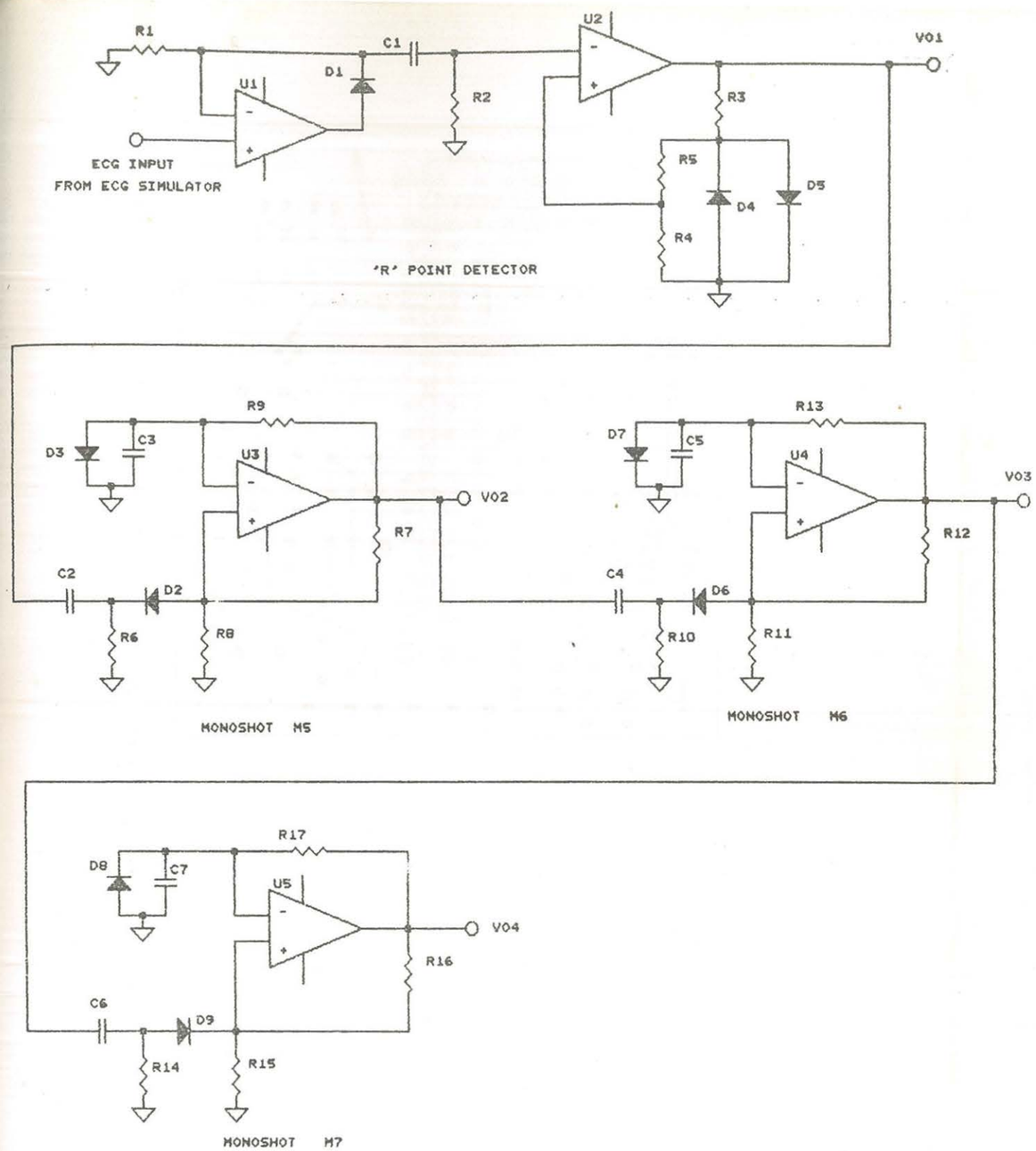
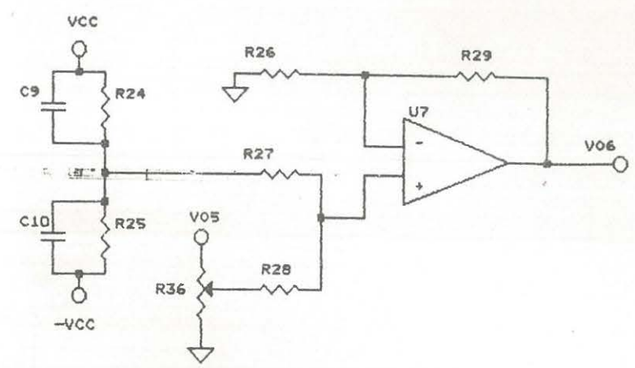
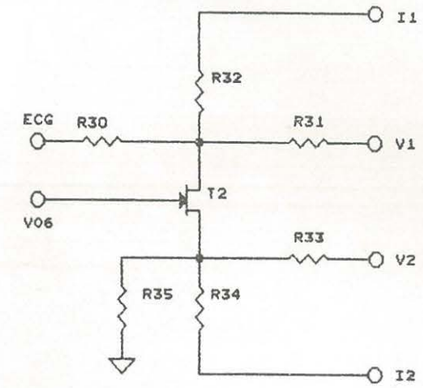


Fig. B.12







THORAX IMPEDANCE SIMULATOR

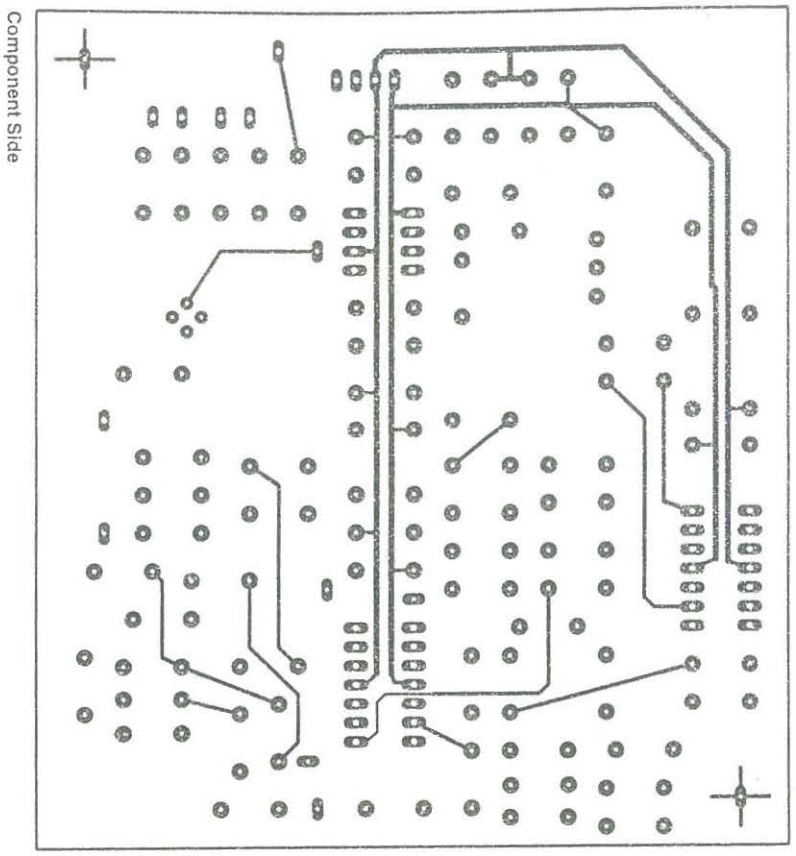


Fig. B.14

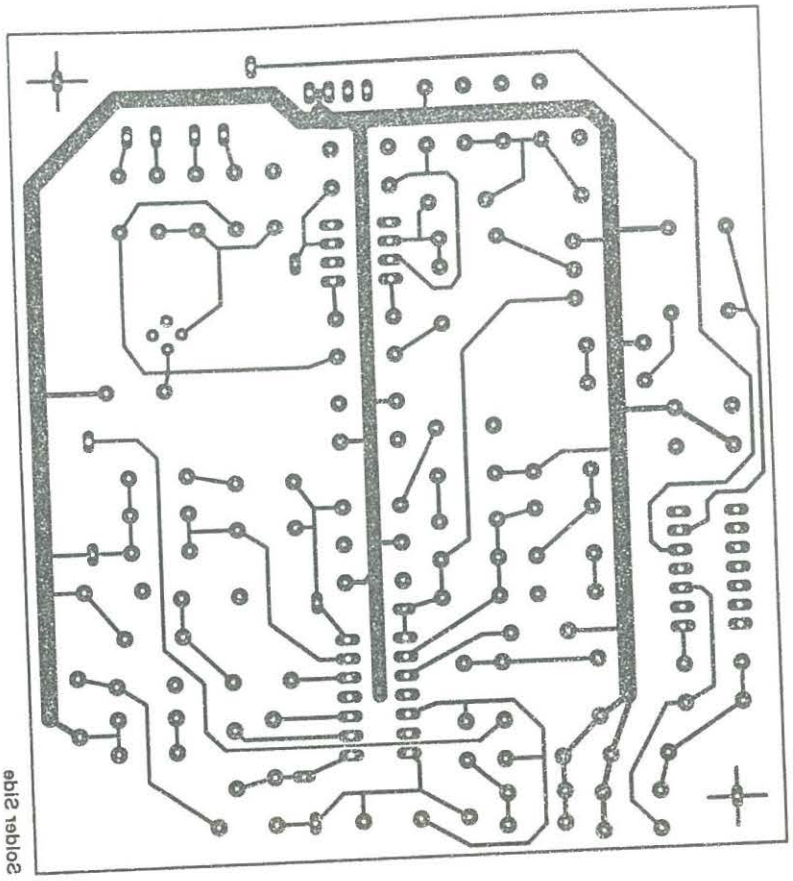


Fig. B.15

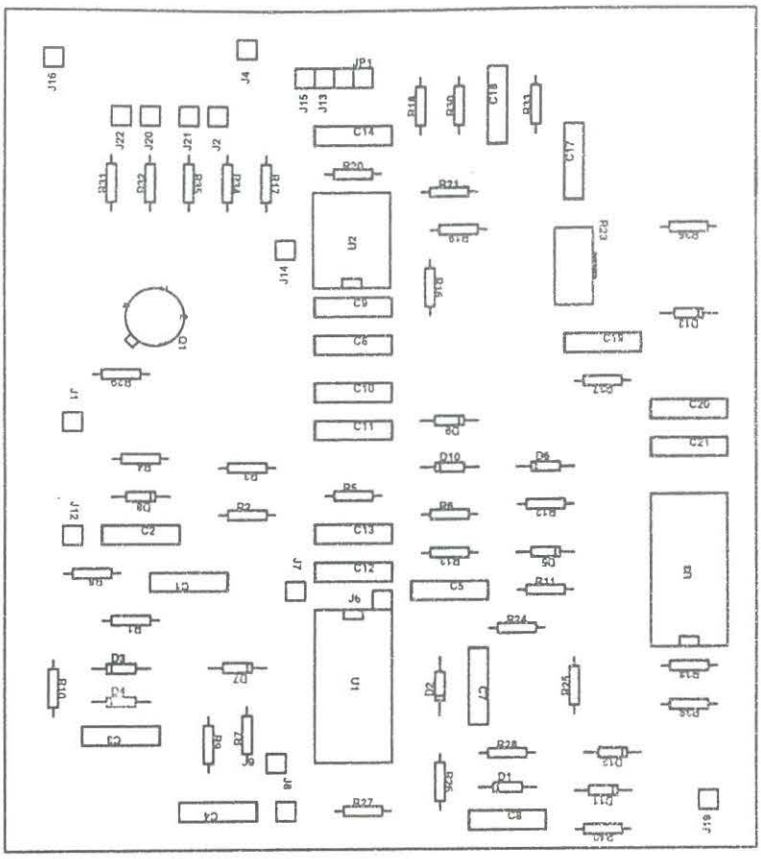


Fig. B.16

APPENDIX C

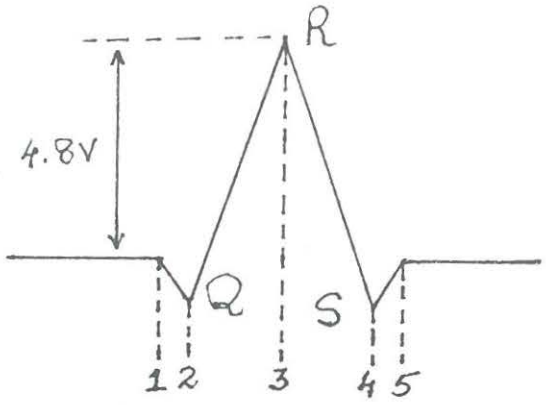
PERFORMANCE SPECIFICATIONS OF THORAX SIMULATOR AND SIGNAL CONDITIONER HARDWARE

Various operational parameters of the thorax simulator and signal conditioner hardware have been listed in this appendix. This will serve in easy calibration or repair of these circuits. The signal conditioner was applied the ECG and thoracic impedance simulated in the thorax simulator circuit and the outputs of the signal conditioner ware observed on CRO. The amplitude and shape of various waveforms present inside the thorax simulator and signal conditioner hardware are listed here.

1) ECG simulator (refer Fig.4.8)

1.1) QRS complex duration = 88 msecs.

Voltage of 'R' point in QRS complex = 4.8 V.
The generated ECG waveform is shown below.



Simulated ECG waveform

T1 = 0 Sec	T2 = 3.4 ms	T3 = 43 ms	T4 = 84 ms	T5 = 88 ms
V1 = 0 V	V2 = -1.2 V	V3 = 4.8 V	V4 = -1.2 V	V5 = 0

1.2) The variation in frequency of astable multivibrator AM1 shown in Fig.4.8 (heart beat rate) with different values of potentiometer resistence R7 is given in the following table.

Value of (R6 + R7) in M X	R-R interval in seconds (period of AM1)
1.2	0.74
1.3	0.80
1.4	0.86
1.53	0.94
1.7	1.00
1.88	1.10

2) Changing thoracic impedance simulator (refer Fig.4.9)

2.1) Minimum amplitude of QRS complex for reliable working of 'R' point detection circuit = 130 mV.

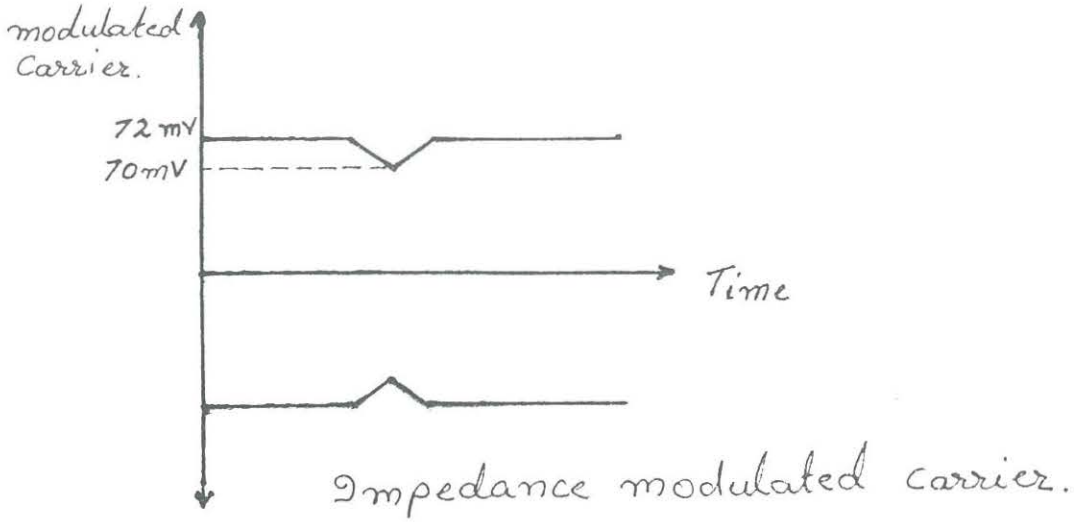
2.2) Delay introduced by various circuits from the detection of 'R' point in ECG to the final generation of triangular wave is given below. (refer Fig.3.3b)

a) 'R' point to the negative transition of schmitt trigger built around opamp U2 = 2 msecs.
b) 'R' point to down going transition of monoshot M5 = 40 msecs.
c) 'R' point to up going transition of monoshot M6 = 190 msecs.
d) 'R' point to down going transition of monoshot M7 = 360 msecs.

2.3) The gata bias of FET T2 is varied by adding a triangular wave to the negative bias of -1.3 V. This changes its channel resistance. The effect of this changing channel resistance on the amplitude of the ECG signal added at the V1 terminal, was determined by recording the amplitude of R-point of detected ECG with gate bias of -1.3 V and -0.3 V and it was observed that

there is no appreciable change in ECG amplitude.

2.4) The changing channel resistane of FET modulates the 100 kHz carrier. The strength of the unmodulated carrier (*i.e.* with gate voltage of FET T2 = -1.3 V) is 72 mVolts. The strength of the carrier reduces to = 70 mVolts when the peak of the triangular wave is present. Following figure shows the modulated envelope.



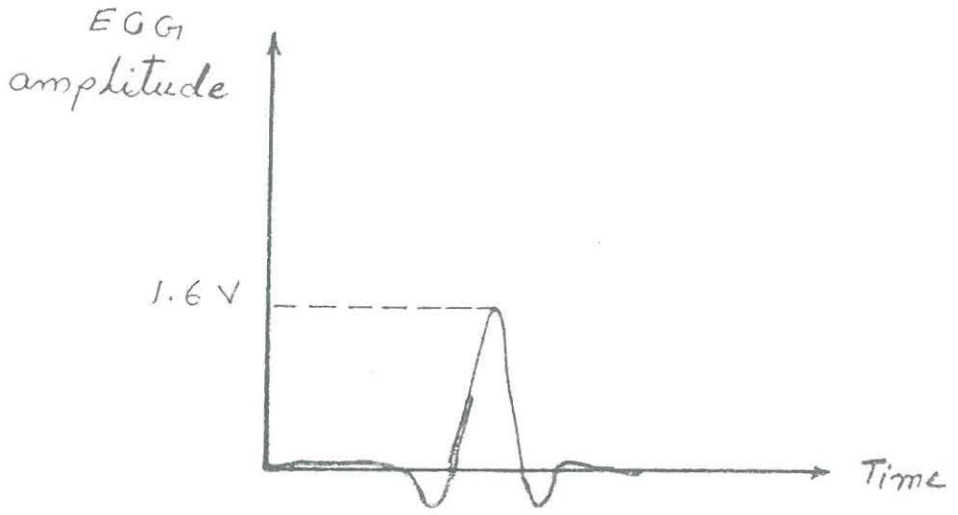
3) ECG extraction circuit (refer Fig.4.7)

The ECG extraction circuit was fed the ECG signal from the ECG simulator as present between outputs V1 and V2 of the thorax simulator. The amplitude of the resultant ECG signal at various locations in the ECG extraction circuit is noted below.

3.1) QRS amplitude at the voltage electrode inputs V1 and V2 of ECG extraction circuit = 1.8 mV.

3.2) The output of the instrumentation amplifier Vo1 has a QRS amplitude strength of 40 mV.

3.3) The final filtered output of ECG detector shown in the figure below has a strength of 1.6 V. The peaks of QRS get rounded off due to the low pass filtering action present in the last stage of ECG extraction circuit.

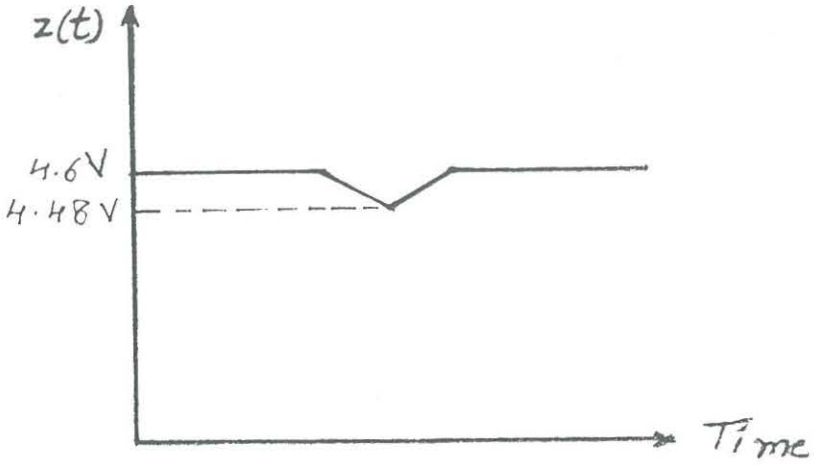


Output of ECG extraction circuit.

4) z(t) extraction and dz/dt circuit (refer Fig. 4.3)

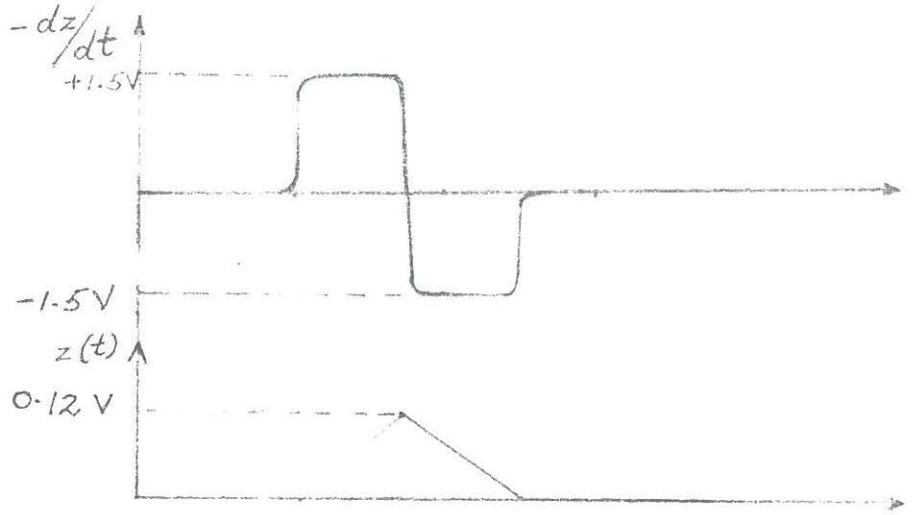
The excitation carrier modulated by the changing channel resistance of FET T2 was fed as input to the circuit. The strength of carrier with triangular wave absent was 72 mVolts. The minimum value of the carrier during the peak of the triangular wave and had a strength of 70 mVolts (refer the carier modulation waveform shown above). The output voltages at various stages of the z(t) extraction and dz/dt circuit are recorded below.

4.1) The output of the demodulator circuit, Vo is shown below. It has a strength of 4.6 Volts with a 0.12 V triangular wave riding on it.



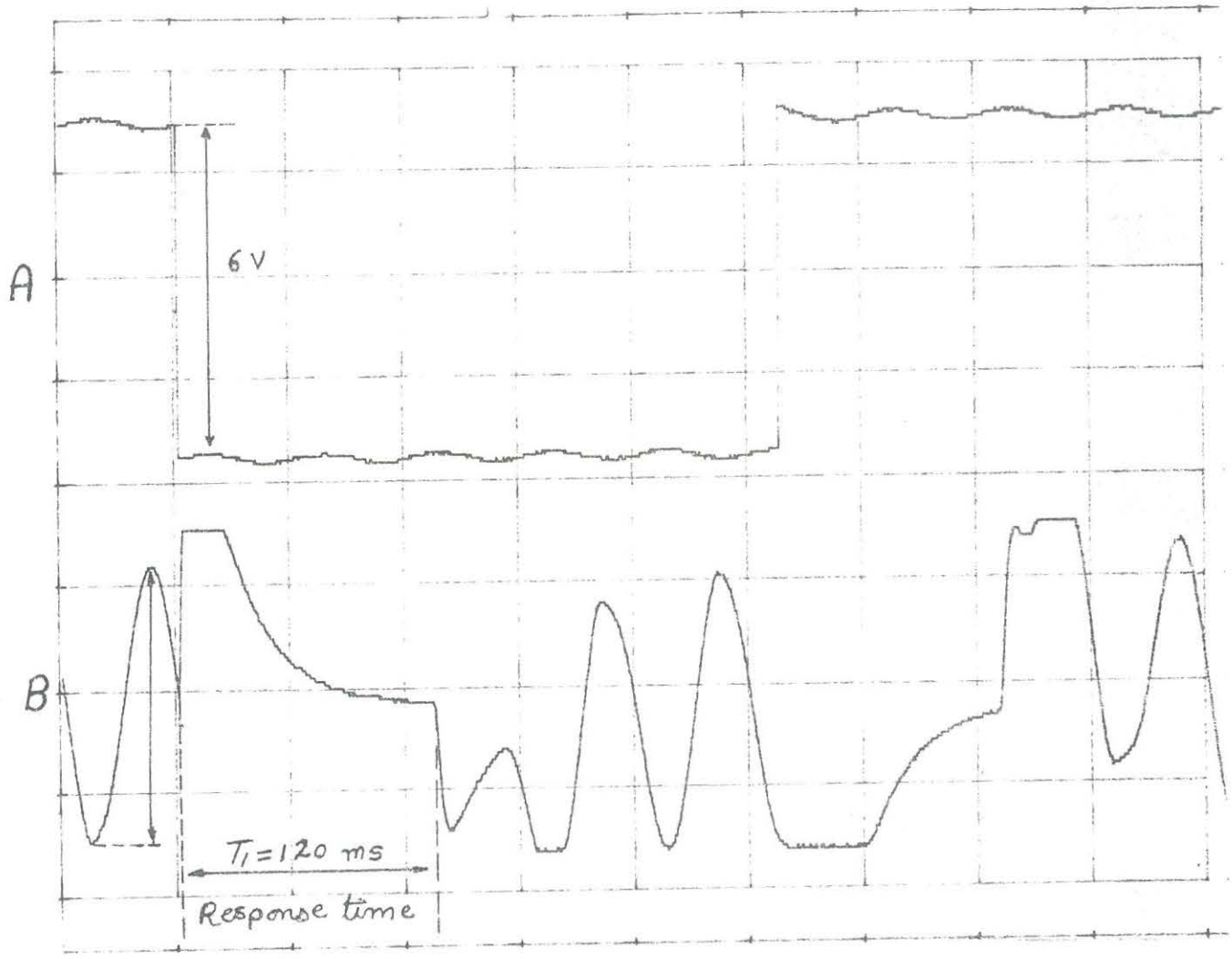
Demodulator output

4.2) The output of the demodulator is differentiated to get the dz/dt. With the above mentioned setup the resulting 'dz/dt' signal was a bipolar pulse with a positive peak of 1.5 V as shown in the following figure.



4.3) The DC cancellation circuit was individually tested by applying a sinusoidal wave riding over a large square wave at its input. The square wave represents a sudden jump in signal level and the response of the DC cancellation circuit in eliminating the square wave was checked. The response is as shown in accompanying figure AF1. The DC cancellation circuit takes a finite time to cancel off the transition of the square wave. This is due to the slow charging of the integrator capacitor C6 shown in Fig.4.5. The square wave had an amplitude of 3.4 Volts and the response time taken for eliminating the square wave transition is marked in the diagram as T1 which is around 120 msecs.

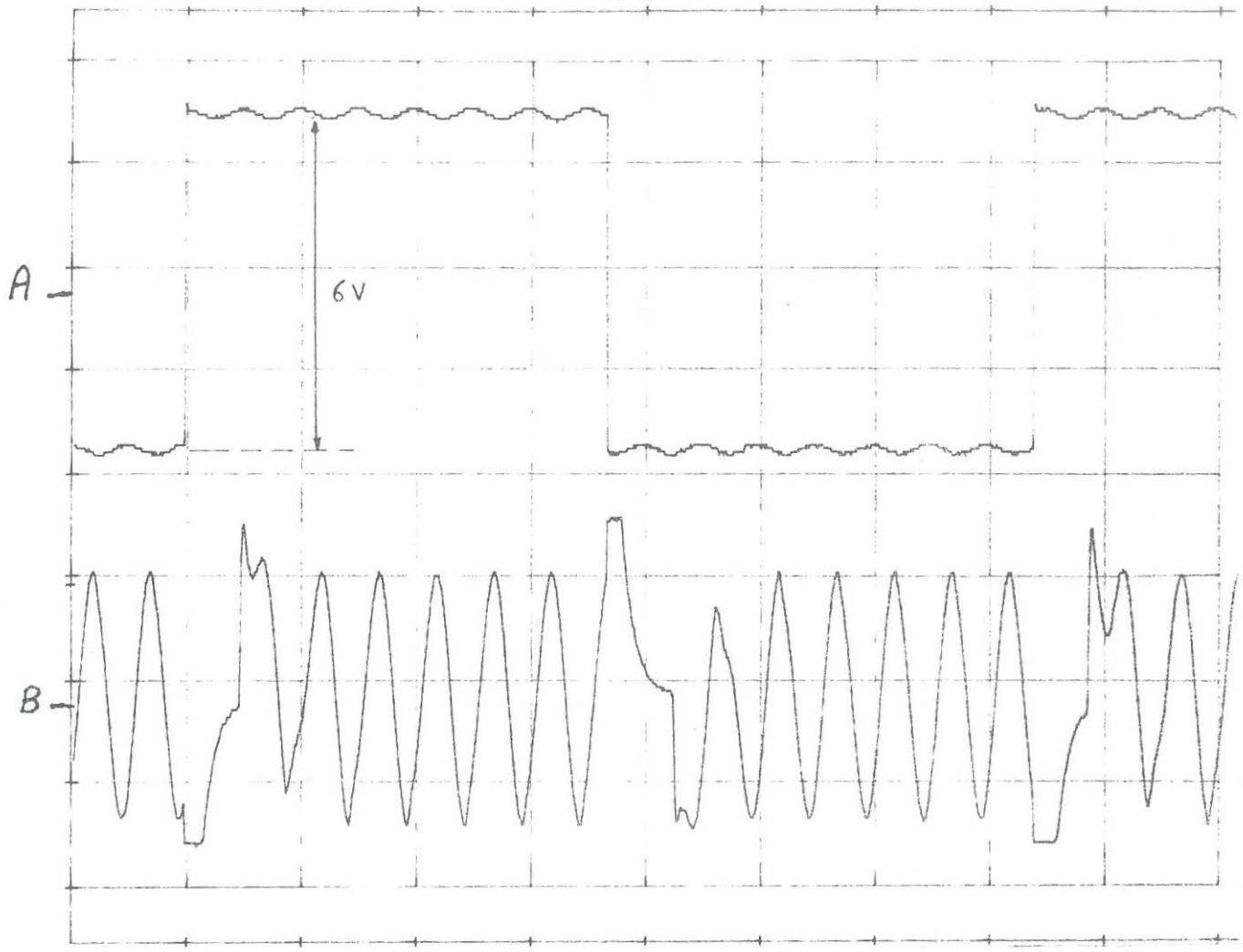
This response time is fast enough as the ICG is a low frequency signal (frequency content < 10 Hz) and has a maximum repeation rate of approximately 2 Hz. To give a pictorial view of the speed of the DC cancellation circuit when used with actual ICG signals a 10 Hz sine wave was added to a 3.4 Volts, 0.71 Hz square wave and the response of the DC cancellation circuit to this input was recorded. This response is shown in Fig.AF2.



Waveform A: Input to DC cancellation circuit.

Waveform B: Output of dz/dt circuit, (V_{05})
(refer Fig. 4.6).

Fig. AF1



Waveform A : Input to DC cancellation circuit.

Waveform B : Output of dz/dt circuit, (Vo5).
(refer Fig. H.6)

Fig. AF2

APPENDIX D

PCL 208 DATA ACQUISITION CARD SETTING

PCL 208 is a data acquisition card form "Dynalag Instruments". It interfaces with the PC bus and has a programmable base address and interrupt. It consists of 16 single ended, or 8 differential inputs. The card has some switches for programming it's operational capabilities. These switch settings as required by the program for online monitoring of physiological signals are now stated.

The base address is selected by switch SW6. The switch settings for selecting the required 300 HEX address are as shown below.

Switch SW6 Base address selection

Base Address 300 HEX

switch position	1	2	3	4	5	6
name	A9	A8	A7	A6	A5	A4
setting	OFF	OFF	ON	ON	ON	ON

The analog signals to be digitized are all single ended and bipolar in nature. The card can be configured for sinlge ended bipolar inputs by switches SW2 and SW3. Slide switch SW2 has to be kept at BIP position to enable bipolar operation. The slide switch SW3 should be kept at 16CH position to enable 16 single ended input channels mode. The analog input range is selected by switch SW5 and should be kept at ~10 Volts setting as shown below.

Switch SW5 Input range selection

Input range ±10 volts

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

The card has two clock frequencies (10 MHz and 1MHz) for it's internal timer operation. The 1MHz clock should be selected by sliding switch SW1 to 1MHz position.

APPENDIX E

DAQ700 A/D CARD DETAILS

DAQ700 is a low-power, digital, and timing I/O card for computers equipped with a PCMCIA type II slot. .The card conatins a 12 bit successive approximation A/D converter with 8 differential or 16 single ended inputs, 8 lines of digital inputs, and 8 lines for digital output. The card also has a clock generator and a programmable counter/timer for timing I/O. The card interfaces with the inputs and outputs through a 50 pin connector.

The card is fully software configurable. The card has to be configured for single ended bipolar configuration. The selection of the base address, the interrupt level, the number of channels to be sampled, the sampling rate, etc is done by software before using the card.

This software configuration is done by writing certain control words at specific addresses. The control words to be written and the order in which they are to written is given below. Please refer "DAQCard-700 Register Level Programmer Manual" for additional details.

1 Write 8*x* hex to command register 1 (base address +00)
2 Write 0*x* hex to command register 1 (base address +00)
x = 2 when PCG is not sampled and *x* = 3 when PCG is also sampled.
3 Write 00 hex to command register 2 (base address +07)
4 Write 00 hex to command register 3 (base address +05)
5 Write 34 hex to counter mode register (base address +0B)
6 Write 00 hex to timer interrupt clear register (base address +06)
7 Write 00 hex to the A/D clear register (base address +01)
8 Read data from the A/D FIFO register (base address +02)

APPENDIX F

"SITE" data logger information

Micro SITE Data Logger, is a product from "Dynalog Microsystems". It has 8 A-D channels and an 8 bit digital I/O port. It also has a key pad of 3 keys. The data logger has to be programmed to control when it starts sampling, its sampling rate, number of channels to be sampled, *etc.* The programming language is the SITE language developed specifically for the data logger. In the data logger interface developed, the data logger digitizises and stores three or four channels as programmed. The data is stored as 8 bits per sample. The interface with the PC for up-loading into the data logger its control program, or for down-loading into the PC the recorded data is via a serial port.

The company "Dynalog Microsystems" has also supplied a few programs for communicating with the data logger. These programs have been used in the interface developed. A brief description of the programs is given below.

SC.EXE This program complies text files written in SITE language. The text files can be typed using any text editor. The name of the text file should have the extension *.lit.

SITECHK.EXE This program interrogates the selected serial port to see if the SITE is online. The default port is COM1.

LAUNCH.EXE This program loads the program compiled by SC.EXE into the data loggers memory.

SITEOFF.EXE This program up-loades the data digitized and stored by the data logger, into the PC. The data is stored in the PC in the form of *.dat files. These files are not normal ASCII files.

XSITE.EXE This program translates the *.dat file up-loaded from the data logger into the PC, into a ASCII format.

BLAST.EXE This program is used to restore normal operation of the SITE instrument in case it should ever hang up.

For more details about the data logger please refer "Software Reference, User's Manual" and "Hardware Reference, User's Manual" from the same company.

APPENDIX G

Tables of the test results on subjects

Table	Description
G.1	Experimental results from subject NMH during exercise
G.2	Experimental results from subject KSP during exercise
G.3	Experimental results from subject SAP during exercise
G.4	Experimental results from subject MRP during exercise
G.5	Experimental results from subject MRV during exercise
G.6	Experimental results from subject SVP during rest

Table G.1 Test results from subject under exercise on an exercising bicycle. The list of results obtained were taken at every 10 sec. intervals.
Z_0 = Thoracic impedance, H.R. = Heart beat rate,
SV. = Stroke volume, CO. = Cardiac output,
ρ = 142 ohm-cm, L = 15 cm.

Subject NMH

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
10	90	14	0.28	39.0	3.5
20	91	14	0.29	39.4	3.5
30	95	13	0.25	20.1	1.9
40	89	13	0.28	33.8	3.0
50	97	13	0.31	40.0	3.8
60	104	13	0.28	29.5	3.0
70	108	13	0.26	36.3	3.9
80	110	13	0.26	39.5	4.3
90	112	13	0.26	47.5	5.3
100	111	13	0.25	41.0	4.5
110	116	13	0.26	36.0	4.1
120	115	13	0.25	37.5	4.3
130	116	13	0.20	29.6	3.4
140	119	13	0.24	39.1	4.6
150	118	13	0.23	38.0	4.4
160	118	13	0.23	35.2	4.1
170	118	13	0.23	32.7	3.8
180	118	13	0.25	40.9	4.8
190	119	13	0.25	36.4	4.3
200	125	13	0.20	29.4	3.6
210	124	13	0.23	35.9	4.4
220	124	13	0.20	30.4	3.7
230	124	13	0.22	41.7	5.1

Time s.	H.R. beats/min	Z_0 ohms	Tivet s.	SV. ml.	CO. litres/min
240	127	13	0.19	26.0	3.3
250	131	13	0.19	30.2	3.9
260	129	13	0.23	40.4	5.2
270	127	13	0.19	29.1	3.7
280	132	13	0.19	23.4	3.0
290	134	13	0.23	36.8	4.9
300	130	14	0.22	26.4	3.4
310	129	13	0.22	33.4	4.3
320	127	14	0.16	26.8	3.4
330	123	14	0.17	41.4	5.1
340	120	14	0.19	39.3	4.7
350	119	14	0.19	36.1	4.3
360	116	14	0.28	54.1	6.2
370	116	15	0.22	37.4	4.3
380	117	14	0.23	40.3	4.7
390	114	15	0.25	47.5	5.4
400	113	15	0.28	50.4	5.7
410	111	15	0.25	50.0	5.5
420	111	14	0.29	53.3	5.9

Table G.2 Test results from subject under exercise on an exercising bicycle. The list of results obtained were taken at every 10 sec. intervals.
Z_0 = Thoracic impedance, H.R. = Heart beat rate
SV. = Stroke volume, CO. = Cardiac output
ρ = 142 ohm-cm, L = 15cm.

Subject KSP

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
10	89	18	0.19	13.7	1.3
20	95	18	0.18	16.6	1.8
30	89	18	0.19	24.1	2.5
40	93	18	0.16	13.6	1.7
50	93	18	0.17	11.4	1.7
60	94	17	0.17	11.0	1.4
70	99	18	0.16	14.3	1.2
80	98	18	0.17	16.6	1.3
90	99	18	0.16	14.6	1.5
100	101	18	0.17	18.3	1.5
110	106	18	0.16	9.70	1.3
120	104	18	0.16	10.7	1.2
130	104	18	0.16	13.7	1.3
140	105	18	0.16	12.7	1.4
150	108	18	0.17	15.6	1.9
160	110	18	0.16	17.0	1.8
170	113	19	0.16	14.1	1.0
180	114	18	0.25	19.6	2.4
190	117	19	0.17	13.0	1.3
200	120	19	0.16	12.4	1.0
210	119	19	0.16	13.0	1.6
220	124	18	0.16	14.3	1.8
230	121	18	0.16	9.63	1.7

Time s.	H.R. beats/min	Z_0 ohms	Tlvet s.	SV. ml.	CO. litres/min
240	108	18	0.17	14.1	1.3
250	97	18	0.17	13.6	1.2
260	94	18	0.17	13.2	1.4
270	96	18	0.17	10.6	1.2
280	94	18	0.17	9.76	0.2
290	90	17	0.19	11.2	1.1
300	92	18	0.18	13.9	1.9
310	89	18	0.19	9.52	0.5
320	89	18	0.19	10.0	0.9

Table G.3 Test results from subject under exercise on an exercising bicycle. The list of results obtained were taken at every 10 sec. intervals.

Z_0 = Thoracic impedance, H.R. = Heart beat rate

SV. = Stroke volume, CO. = Cardiac output

ρ = 142 ohm-cm, L = 15cm.

Subject SAP

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
10	68	13	0.37	90.6	6.5
20	60	13	0.35	92.0	5.5
30	66	13	0.35	63.7	4.2
40	79	12	0.20	29.6	2.0
50	78	12	0.20	45.0	3.5
60	87	12	0.19	30.0	2.5
70	94	12	0.22	34.6	3.1
80	94	12	0.22	39.6	3.8
90	96	12	0.22	54.7	5.9
100	99	12	0.19	43.7	4.1
110	98	12	0.19	34.4	3.7
120	99	12	0.18	37.5	3.6
130	101	13	0.25	48.8	4.1
140	105	12	0.20	40.4	4.1
150	108	12	0.26	51.3	5.4
160	110	12	0.16	36.5	4.6
170	112	13	0.20	39.6	4.3
180	113	13	0.19	31.9	3.0
190	117	13	0.19	24.7	2.1
200	118	13	0.17	19.4	2.7
210	119	13	0.20	32.9	3.9
220	118	13	0.18	25.5	3.1
230	120	13	0.19	26.6	3.5

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
240	122	13	0.25	42.6	5.7
250	121	13	0.16	27.9	3.6
260	114	13	0.26	57.4	6.8
270	102	13	0.28	89.2	9.6
280	97	14	0.31	117.2	11.3
290	95	13	0.27	71.2	6.0
300	83	13	0.20	52.8	4.9
310	84	13	0.25	70.3	5.6
320	85	13	0.29	65.8	5.1
330	71	13	0.22	52.6	3.3
340	66	13	0.22	60.0	4.1
350	64	13	0.24	64.6	4.1
360	64	13	0.32	101.2	6.8
370	63	13	0.34	100.4	6.3
380	62	13	0.35	114.7	7.2
390	60	13	0.35	111.2	6.7

Table G.4 Test results from subject under exercise on an exercising bicycle. The list of results obtained were taken at every 10 sec. intervals.

Z_0 = Thoracic impedance, H.R. = Heart beat rate

SV. = Stroke volume, CO. = Cardiac output

ρ = 142 ohm-cm, L = 15cm.

Subject MRP

Time s.	H.R. beats/min	Z_0 ohms	T_{ivet} s.	SV. ml.	CO. litres/min
10	90	15	0.32	24.2	2.6
20	88	15	0.21	13.3	1.6
30	83	15	0.26	17.0	1.7
40	89	15	0.28	14.0	1.7
50	105	15	0.28	22.1	2.7
60	109	15	0.25	31.5	3.6
70	114	15	0.25	27.3	3.3
80	121	15	0.16	10.2	1.7
90	128	15	0.16	12.9	1.1
100	136	15	0.25	27.4	3.7
110	140	15	0.25	23.6	3.6
120	143	15	0.24	29.7	4.0
130	150	15	0.24	27.5	4.6
140	145	15	0.23	22.0	3.8
150	134	16	0.25	25.8	3.1
160	129	15	0.18	19.5	2.2
170	126	15	0.16	15.5	1.0
180	118	15	0.25	17.5	2.6
190	114	15	0.29	15.6	1.9
200	109	15	0.29	19.4	2.8
210	111	15	0.16	13.9	1.4
220	110	15	0.25	16.0	1.7
230	107	15	0.25	17.7	1.3

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
240	103	15	0.17	13.9	1.8
250	104	15	0.16	13.0	1.4
260	107	15	0.16	9.8	1.5
270	107	15	0.16	11.0	1.3
280	102	15	0.17	10.6	1.8
290	98	15	0.30	17.1	1.0

Table G.5 Test results from subject under exercise on an exercising bicycle. The list of results obtained were taken at every 10 sec. intervals.

Z_0 = Thoracic impedance, H.R. = Heart beat rate

SV. = Stroke volume, CO. = Cardiac output

ρ = 142 ohm-cm, L = 15cm.

Subject MVN

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
10	85	17	0.23	14.2	1.0
20	83	16	0.31	34.9	2.4
30	84	17	0.32	29.8	2.6
40	84	16	0.29	36.3	3.0
50	90	16	0.32	25.3	2.5
60	107	15	0.16	18.6	1.3
70	110	15	0.17	18.5	2.1
80	115	15	0.17	16.6	1.5
90	117	16	0.23	30.5	3.2
100	117	16	0.23	28.3	3.1
110	119	16	0.16	18.8	2.9
120	119	16	0.16	20.0	2.8
130	116	16	0.19	22.3	2.7
140	118	16	0.16	14.0	1.6
150	119	16	0.16	21.1	2.0
160	117	16	0.19	24.3	2.5
170	116	16	0.22	25.5	2.3
180	115	16	0.17	15.4	1.1
190	117	16	0.25	30.9	3.1
200	115	16	0.18	20.0	2.1
210	114	16	0.25	23.4	2.3
220	111	16	0.22	27.0	3.1
230	112	16	0.18	16.0	1.0

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
240	110	16	0.25	27.9	2.8
250	130	15	0.20	30.4	4.2
260	122	17	0.23	30.7	3.8
270	104	17	0.19	26.7	2.3
280	95	16	0.20	28.1	2.1
290	91	16	0.24	33.7	3.6
300	87	16	0.25	39.2	3.9
310	86	17	0.20	15.5	1.1
320	83	16	0.26	27.5	2.9

Table G.6 Test results from subject under rest. The list of results obtained were taken at every 10 s intervals.

Z_0 = Thoracic impedance, H.R. = Heart beat rate

SV. = Stroke volume, CO. = Cardiac output

ρ = 142 ohm-cm, L = 15cm.

Subject SVP

Time s.	H.R. beats/min	Z_0 ohms	T_{lvet} s.	SV. ml.	CO. litres/min
10	79	12	0.23	18.1	1.4
20	79	12	0.20	17.9	1.4
30	83	14	0.18	15.5	1.2
40	78	14	0.17	15.4	1.2
50	74	14	0.19	16.4	1.2
60	78	13	0.23	18.4	1.4
70	79	13	0.19	19.5	1.5
80	81	13	0.19	15.9	1.2
90	87	13	0.24	21.3	1.8
100	79	13	0.24	21.6	1.7
110	84	13	0.19	16.0	1.3
120	80	14	0.19	14.4	1.1
130	81	13	0.19	15.7	1.2
140	80	12	0.20	15.8	1.2
150	80	13	0.23	17.4	1.4
160	77	13	0.20	16.7	1.2
170	76	13	0.20	18.1	1.3
180	78	12	0.19	15.7	1.2
190	78	12	0.20	20.6	1.6
200	76	12	0.22	23.4	1.7
210	76	12	0.19	14.8	1.1
220	75	12	0.26	24.6	1.8

Time s.	H.R. beats/min	Z_0 ohms	T_{1vet} s.	SV. ml.	CO. litres/min
230	74	12	0.21	17.8	1.3
240	78	11	0.26	18.8	1.4
250	75	11	0.20	18.1	1.3
260	74	12	0.22	18.3	1.3
270	73	11	0.21	21.0	1.5
280	76	11	0.25	22.3	1.7
290	73	11	0.26	22.6	1.6
300	82	15	0.28	20.7	1.7

APPENDIX H

OPERATING INSTRUCTIONS

In this appendix the instructions for using the impedance cardiograph system are listed. The impedance cardiograph system consists of the signal conditioner hardware and signal acquisition and processing unit. The signal acquisition can be done using a PC with an appropriate A/D card or by using a data logger. In the following sections, the calibration procedure and test setup of the signal conditioner hardware is explained. This is followed by a discussion of the PC based signal acquisition and processing unit. Finally the data logger interface for recording the signals is explained.

The signal conditioner hardware and thorax simulator circuits are housed in a single box with all the signal leads connected to audio jacks fitted on a common control panel. The view of the control panel is shown in Fig.H.1. The control panel can be roughly split into two sections, one containing the terminals of thorax simulator circuit and the other containing terminals for signal conditioner hardware. The signal conditioner has four outputs namely the z(t), dz/dt, ECG, and PCG. The PCG IN terminal has to be connected to the output of the electronic stethoscope. The PCG out terminal represents the PCG amplifier output. Both, thorax simulator and signal conditioner hardware have the excitation current terminals and sensing voltage terminals brought out on the control panel. The I1/I2 terminal is a stereo jack which contains both I1 and I2 signals.

The impedance simulator requires the ECG simulated by the ECG simulator as an input. This is done by connecting the ECG OUT terminal of ECG simulator to the ECG IN terminal of impedance simulator on control panel. The DZ output on the control panel is the triangular wave generated by the impedance

simulator. The potentiometer P1 can be varied to change the R-R interval in simulated ECG. The LED D1 glows for half the R-R interval in simulated ECG.

SIGNAL CONDITIONER HARDWARE CALIBRATION

The signal conditioner hardware has to be checked for calibration before use. This can be done by interfacing the signal conditioner hardware with the thorax simulator as shown in Fig.2.8. For calibrating the signal conditioner hardware, the I1/I2, V1, and V2 leads of the signal conditioner hardware and the thorax simulator are to be shorted. The ECG simulated is given to the impedance simulator by shorting the terminals ECG OUT and ECG IN. The ECG, z(t), and dz/dt outputs of the signal conditioner hardware are now digitized and used to check its calibration. The digitization and recording of the signals is done by the program SPA2A.EXE and the recorded signals can be observed on screen by using the program SPA2B.EXE. The waveforms of the outputs of a fully calibrated signal conditioner hardware, shown in Fig.5.1 and discussed in Appendix C, can be used for comparison.

TEST SETUP OF SIGNAL CONDITIONER HARDWARE

The test setup for connecting the signal conditioner hardware to the exercising subject is now explained. The electrodes I1, I2, V1, and V2 are connected to the subject according to the Zhang's electrode placement scheme shown in Fig.2.9. The electrodes used, four suction type ECG electrodes, are dipped in ECG jelly and then applied onto the subject's body. The PCG is obtained by connecting the chest piece of the electronic stethoscope onto the chest of the subject. A chest belt for holding the chest piece has also been provided. The outputs of the signal conditioner hardware are digitized and recorded without opto-isolation if the battery operated notebook PC or the data logger is used for the same purpose.

SIGNAL ACQUISITION AND PROCESSING

The outputs of the signal conditioner hardware can be digitized and recorded using the a) mains operated PC with PC-bus based PCL208 A/D card from Dynalog Microsystems, or b) notebook PC with PCMCIA-bus based DAQ700 A/D card from National Instruments or c) data logger. The instructions for signal acquisition and processing using the first two options are now explained. The channels 0 to 3 of the A/D card used, are connected to the z(t), dz/dt, ECG, and PCG outputs of the signal conditioner hardware respectively. Then the program SPA2A.EXE is executed on the PC. This program gives certain options to the user which are listed below. The values shown in brackets for each option are the default values.

* Sample PCG [Y/ret] :

This option decides if PCG is required to be sampled along with the ECG, z(t), and dz/dt signals and can be selected by pressing the 'Y' or 'N' keys as necessary.

* Subject code [SPT] :

The three character subject code, to be entered is used as part of the file names that store the recorded data.

* Use PCL208 A/D card [ret/N] :

This option decides which A/D is being used. PCL208 A/D card is considered connected to the PC if just the 'RETURN' key is pressed. Any other key means the DAQ700 A/D card.

* Base address in decimal [768] :

The base address of the A/D card can be changed from the default value of 768 by entering the appropriate value here and is applicable to both PCL208 and DAQ700 cards. The default value need not be changed unless there is a clash with some

other add-on card.

* Free IRQ no. [7] :
The interrupt number assigned to the A/D card for transferring the digitized data to the PC can be entered here. Normally interrupt 7 of the PC is free and is hence used as default. This choice is applicable to both the PCL208 and DAQ700 cards.

* Sampling record length 10*[1]sec :
This is the time over which filtering of the recorded signals is done prior to processing them for monitoring CO. The recording interval is in multiples of 10 s. This is different from the total recording time, which decides the total time duration over which signals are to be recorded. The total recording time can be anything, say 400 s with recording interval of 10 s, meaning after every 10 s the recorded signals are processed for monitoring CO.

* Blood resistivity [142] :
The blood resistivity of the subject is taken as 142 ohm/cm by default but can be changed by entering the required value.

* Distance between voltage electrodes in cm [15] :
The distance between voltage electrodes V1 and V2, on the chest of the subject, should be measured with a tape and entered here, the default value being 15 cm.

Once all these options have been selected the program starts recording and processing the signals. The values of basal thoracic impedance Z_{base}, HR, T_{lvet}, SV, and CO over the programmed recording interval are now continuously calculated and displayed. The program can be stopped by pressing the 'ESC' key on the keyboard.

Once all recording is complete, the recorded signals can be

processed off-line and displayed on screen using the program SPA2B.EXE. This program asks the name of the subject whose recorded signals are to be analyzed. The recorded signals are then displayed cycle-wise over each recording interval.

The data logger can be used for recording the signals. The channels 2 to 5 of the data logger are connected to the ECG, z(t), dz/dt, and PCG outputs of the signal conditioner hardware respectively. Then the program SPA2C.LIT or SPA2D.LIT is down-loaded into the data logger. Program SPA2C.LIT is to be used when PCG is not to be recorded and SPA2D.LIT is used when otherwise. The procedure for down-loading a "*.LIT" file into the data logger and executing it, consists of executing the program "LOAD.EXE" from Dynalog Microsystems on the PC, with the "*.LIT" file name as a command line option. After this has been done the down-loaded program starts its operation by displaying the sign "SURE" on its display and polling its key-pad till the top-most key on it, is pressed.

A key-press is acknowledged by displaying the sign "WAIT" and then the recording is initiated. After the recording is complete the data logger displays the sign "FULL" on its display. Now the logged data has to be up-loaded into the PC over the serial link. This up-loading is done by connecting the data logger to the PC with the serial link and executing the program "SITEOFF" from Dynalog Microsystems on the PC. This results in up-loading of the recorded data from the data logger into the PC in the form of a "*.DAT" file. This "*.DAT" file is to be converted into an ASCII file ("*.TXT") by executing the program "XSITE" from Dynalog Microsystems on the PC. This ASCII file is transformed by the program CONVT.EXE or CONVT1.EXE into a binary file with a format recognized by the off-line processing program SPA2C.EXE. The program CONVT.EXE should be used when PCG has not been recorded and CONVT1.EXE when otherwise.

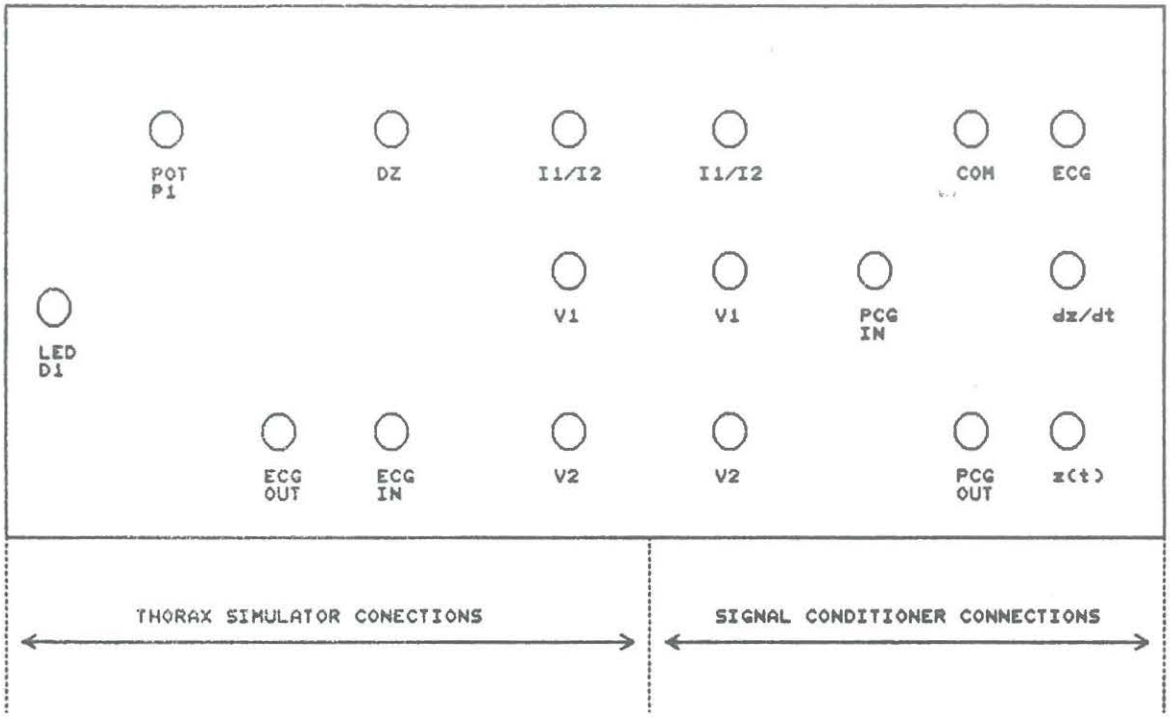


Fig.4.1 View of the control panel of impedance cardiograph system

APPENDIX I

SYSTEM SPECIFICATIONS

Hardware

1. Signal conditioner

Supply voltage	18 VDC
Operating current	60 mA

1.1 Excitation

This circuit passes a constant current through the thorax of the subject.

Subject excitation frequency	100 kHz
Subject excitation current	3.3 mA, rms

1.2 Sensing

The signal conditioner unit has four outputs namely the ECG, z(t), dz/dt, and PCG. The normal strength of each of the ECG, z(t), dz/dt, and PCG signals is around ±4V, 5V, ±3V and ±5V respectively. The input impedance of both the ECG extraction circuit and the z(t) extraction circuit is 100 kΩ. The calibration factors for the various outputs are as follows.

a) ECG 1.04e-3

b) z(t) 0.016 ohms/V

c) dz/dt 2.11 ohms/sec/V

1.3 Electrodes

Four spot type ECG electrodes are used. ECG jelly is applied to the electrodes before placing them on subjects.

1.4 Calibrator

This circuit simulates the ECG and the thoracic impedance. The strength of the R point in the stimulated ECG is 4V. The thoracic impedance simulated has a base value of 20 ohms. The changing thoracic impedance component is 2% of the base impedance. The excitation current of 3.3 mA rms, from the current electrode terminals of the subject excitation circuit passes through this simulated thoracic impedance producing the required voltage proportional to thoracic impedance.

Signal acquisition

The ECG, z(t), dz/dt, and PCG outputs are digitized and stored. This can be done using a) mains operated PC with PCL208 data acquisition card from Dynalog Microsystems b) notebook PC with DAQ700 data acquisition card from National Instruments c) data logger from Dynalog Microsystems. For both the data acquisition cards interrupt based data transfer mode is selected with the input range at ±10 V. The ECG, z(t), and dz/dt outputs are sampled at 200 samples per second and the PCG at 1000 samples per second.

Processing

The processing of the ECG, z(t), and dz/dt is carried out on PC. The PC should have a 80386 processor or better, with minimum 1 MB RAM and a VGA display. During the processing of the digitized signals the R points in ECG are detected and then ensemble averaging of dz/dt with these R points serving as a fiducial mark is carried out. The ensemble averaging is carried out over 0.8 times the average R-R interval. The digitized signals are stored in the PC for off-line processing and display. Hard-disk of the computer should have sufficient space to store the files of the recorded signals during the monitoring. Two

files are made during the recording. One contains the digitized ECG, z(t), and dz/dt signals in binary format (*.BIN), with the file size approximately 12016 bytes for data recorded over an interval of ten seconds. If PCG is also being recorded, another binary file is created (*.PCG) with a size of approximately 20010 bytes for data recorded over an interval of ten seconds.

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