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Electronic measurement of blood pressure

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ELECTRONIC MEASUREMENT OF BLOOD PRESSURE

by

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1. INTRODUCTION

The measurement of blood pressure is of prime importance to both the medical practitioner and researcher. The determination of blood pressure during periods of dynamic physiological testing is of chief concern in the study of circulatory and respiratory functions. The auscultatory method, the present clinically accepted method (15), is suitable for obtaining measurements while the patient is not moving. Movement of the arm prohibits the satisfactory use of the auscultatory method due to movement artifacts. It is desirable to obtain measurements of blood pressure during treadmill exercise tests performed as a part of circulatory research; but these measurements are not feasible using the auscultatory method. The purpose of this thesis is to investigate electronic methods of measuring blood pressure, suitable for use under dynamic test conditions such as a treadmill exercise test.

The historical and theoretical hemodynamics of the blood system were discussed previously (10). Specifications for electronically measuring blood pressure will be presented. An instrument designed to determine the blood pressure of a subject during treadmill exercise, using an extra-arterial measurement technique, has been constructed. This device will be considered in terms of three major sub-systems; the pulse sensor units, the electronic decoding and control circuits,

and the pneumatic system.

The electronic decoding and control circuits determine the occurrence of pressures corresponding to both systolic and diastolic pressure. The pneumatic system provides air at controlled flow rates and pressures to occlude the flow of blood in the digital artery. The pulse sensor unit contains a miniature photoconductive sensor to monitor pulse amplitudes in the distal finger during a gradual programmed occlusion by a more centrally located finger cuff. The design considerations and operation of each assembly will be discussed.

An analysis of the operation of this system has been performed. Measurements of blood pressures by this electronic blood pressure measuring instrument have been performed and the results of these measurements will be discussed.

II. GENERAL SYSTEM CONSIDERATIONS

A. Specifications for the Automatic Determination of Blood Pressure

The methods of obtaining the human blood pressure fall into two general categories. First, the blood pressure may be determined directly from the arterial system through intra-arterial cannulation. This method requires a minor surgical procedure and in general is not suitable for test conditions where patient movement is required. Second, the blood pressure may be obtained by extra-arterial means. The usual procedure is to vary the pressure in an occluding cuff placed around a peripheral artery and to observe the concurrent changes in the pulsatile blood flow.

Several methods of sphygmomanometry have been proposed (3,10,15,28). The auscultatory method consists of first occluding the flow of blood to the distal arm by inflating an arm cuff around the middle upper arm. As the pressure in the cuff is gradually decreased, a stethoscope is placed over the brachial artery, proximal to the radial-ulnar bifurcation. The Korotkow sounds are observed through the stethoscope. Four distinct phases of sound are heard as the pressure in the occlusion cuff is decreased from occlusion. The sudden appearance of a clear tapping sound is termed phase I; during phases II and III the sound grows louder and clearer; in phase IV the sound suddenly becomes muffled and at the end of

phase IV the sound has disappeared. The best index of the occurrence of systolic pressure and diastolic pressure is the start of phase I and the end of phase IV respectively (15). The selection of the proper arm cuff size is of particular interest. The original Riva-Rocci arm cuff was 5 cm wide and produced pressure readings that were about 10 percent higher than expected. A cuff of 12 cm in width was introduced, which is still employed today, and gives acceptable readings. In clinical practice a 5 cm cuff is used for small children.

Instead of using a stethoscope to observe the Korotkow sounds, a sensor may be used to determine the variation in amplitude of the blood pulse waveform with changes in the occlusion cuff pressure. This pulse amplitude monitoring method may be used with either a brachial or digital occlusion cuff (10). The sensor device used to sense the pulse amplitude variations at a distal location to the occlusion cuff may be of several types; a tambour, a pressure bag in communication with the distal arterial bed, an impedance plethysmograph, or a combination of a light source and light sensor (10,11).

The auscultatory method is the commonly accepted method for clinical blood pressure measurements. Since this method requires the use of a sound detector to observe the Korotkow sounds, any spurious sound will distort or obscure the determination of the proper Korotkow phase. Spurious sounds

may arise from background room noise or from patient muscle movement.

The effects of muscle movement have generally prohibited the determination of blood pressure by the auscultatory method under dynamic test conditions. Figure 1 is a curve showing the effect of treadmill exercise on systolic and diastolic pressures. For the first 2.5 seconds the subject is standing at rest. During this period blood pressure may be determined by the auscultatory technique. While the treadmill is in operation blood pressure information is not available. For the last 2.5 seconds the subject is permitted to recover in a standing rest position, again the blood pressure may be determined. The purpose of the treadmill exercise test is to evaluate circulatory and respiratory functions of the human body with controlled work. It will be seen from Figure 1 that during the period of treadmill work, direct information on blood pressure is not available.

A conference to consider the possibility of obtaining blood pressure information during the period of treadmill operation by electronic methods was held between Dr. Harold Margulies, M. D., and Dr. John Gustafson, M. D., of the United Heart Station of the Iowa Methodist Hospital in Des Moines, Iowa, Dr. Victor W. Bolie, Chairman of Biomedical Electronics at Iowa State University, and the author.

The following general design specifications for a system

to measure blood pressure during the treadmill exercise test were established:

1. The system should require a minimum of operating procedures, preferably operating without continual adjustment.
2. The system should provide outputs compatible with recording equipment in use for research purposes at the United Heart Station.
3. Blood pressure information should be available on a cyclical basis.
4. The system should obtain blood pressure measurements during the treadmill exercise test.

A review of the pertinent considerations for blood pressure determination with a sphygmomanometer was conducted (3,5,10,15,23,29). The results of this review and the measurement operating requirements observed in Figures 1 and 2 are summarized in Table 1.

A pulse amplitude monitoring technique was adopted as the primary measuring method. It was decided to investigate the development of a system utilizing a digital occlusion cuff. The use of a digital occlusion cuff will permit the occlusion of arterial flow on a cyclic basis with a minimum of discomfort to the test subject. The choice of a digital cuff scheme should reduce muscle movement effects resulting from arm motion while the subject is walking on the

treadmill.

B. The Synthesized System

A blood pressure measuring system satisfying the specifications of the previous section was designed. This system is composed of three major subsystems; the electronic decoding and control circuits, the pneumatic system, and the pulse sensor. Figure 3 is a block diagram of this system. The pulse sensor determines variations in the pulse waveform with variations in the occlusion cuff pressure. The occlusion cuff is contained as a part of the pulse sensor unit. The electronic decoding and control circuits decode the pulse amplitude variations obtained from the pulse sensor. The decoded pulse amplitude variation activates the pneumatic system through the control circuits. The pneumatic system provides air pressure to the occlusion cuff of the pulse sensor. The pressure of the air supplied to the occlusion cuff is varied in a programmed pressure cycle. The operation of each of the major subsystems will be further discussed in subsequent chapters.

The instrument has two modes of operation; automatic and manual. Figure 4 is a representation of the pulse waveform and occlusion cuff pressure variation for both modes of operation. The manual mode provides occlusion cuff pressures similar to the arm cuff pressure variation in the standard

auscultatory method. One of four values of maximum occlusion pressure for the manual mode may be selected. The rate of pressure decay may also be varied. The automatic pressure mode provides an occlusion pressure which increases at programmed rates thus enabling rapid measurement of the blood pressure. In the automatic mode the occlusion cycle is repeated at a preselected interval.

In both modes of operation the pulse waveform is obtained from the pulse sensor. This pulse waveform is recorded on one channel of a four channel Sanborn, "150" series recorder, and the occlusion pressure waveform is recorded on a second channel (25). This Sanborn recorder is designed to accept several plug-in type preamplifiers for each recording channel. It was decided to utilize this capability and to build as much of this system as possible in the form of a plug-in unit. A storage space below the recording section of the "150" series cabinet is used to house the subsystems not contained on the plug-in unit. In addition, a pulse sensor selector unit is attached to the treadmill hand bar support. The selection of a plug-in type unit and internal mounting within the recorder cabinet imposed overall space limitations on the units to be constructed. A plug-in volume of 500 cubic inches and a lower supplementary chassis volume of 1920 cubic inches represented the available useable equipment space. The decision to limit the use of external equipment

cabinets, other than the pulse sensor, was prompted by a lack of suitable mounting area in the immediate proximity of the treadmill.

The "150" series recorder utilizes vacuum tube circuitry in both the plug-in preamplifier units and the corresponding installed driver-amplifier unit. The plug-in unit must supply the input bias voltage to the direct coupled driver-amplifier unit. The driver-amplifier unit is designed to supply operating voltages to the plug-in units. Vacuum tube circuitry utilizing, for the most part, the same tubes used in the "150" series ECG preamplifier was chosen.

Figure 5 is a block diagram showing the interconnections of the major units of the electronic blood pressure measuring instrument. There are six major units; the pulse sensor unit, the decoder unit, the control relay unit, the negative power supply unit, the programmed pressure valve unit, and the air pump.

Figures 6 and 7 show the instrument that has been constructed. The device has been mounted on the equipment rack shown in these pictures for system test and evaluation purposes. The control panel of the decoder unit is shown in the top center of Figure 6. The pulse sensor units are shown attached to the hand bar in the center of Figure 6. The pulse sensor connects to the decoder unit through a 15 foot cable which contains both shielded electrical cables and an

air hose. The pneumatic system and the control circuits are located on the supplementary chassis rack. This rack is shown in the lower portion of Figures 6 and 7. The inter-unit connecting cables and hoses are shown in the lower portion of Figure 7.

Figures 8 and 9 show the blood pressure measuring instrument installed in the "150" series recorder at the United Heart Station. The supplementary chassis assembly and the interconnecting air hoses are installed on a semi-permanent basis. The units may be easily removed from the recorder for maintenance.

The blood pulse waveform is obtained by a photosensitive detector unit employing transillumination of the arterial bed of the distal phalange. This detector is combined with an occlusion cuff to form the pulse sensor unit. Three pulse sensor units are available for use with this instrument. The pulse waveform output of the pulse detector is passed through a pulse sensor selector switch and the pulse sensor cable, to the differential preamplifier of the decoder unit. As seen in Figure 5, the occlusion cuff pressure waveform is similarly passed to the pressure transducer circuit of the decoder unit.

The decoder unit functions both as a pulse waveform decoder and a system control unit. The differential preamplifier unit has outputs to both the decoder circuits and the

Sanborn driver-amplifier. The output to the driver-amplifier is subsequently recorded along with the occlusion pressure waveform output of the pressure transducer circuit. The blood pulse waveform input to the systolic and diastolic decoders is decoded, and the corresponding output signals are passed to the control relays. The decoder operating controls are connected with both the control relays and the power relays. An automatic timer circuit is included in the control relay unit. This timer circuit permits the automatic occlusion cycle to be repeated after a preselected period of unoccluded blood flow.

Three other units are located on the supplementary chassis. The pump develops air pressure which it supplies to the programmed pressure valve unit. This valve unit is designed to permit air to be passed to the occlusion cuff at pre-set rates and pressures. The programmed valve unit is activated by the control relay unit. The last unit on the supplementary chassis is the negative power supply unit. While positive plate and heater voltages for the decoder circuits are supplied from the driver-amplifier chassis, current limitations in the negative power supply of the driver-amplifier resulted in a requirement for a separate negative 100 volt power supply. Line power and 6.3 volt AC control power are also furnished from this negative power supply unit.

The blood pressure measuring instrument has been constructed in a manner which will introduce a minimum of new signal interference to the operation of the "150" series recorder. First, all units except the pump have been enclosed for electrical shielding purposes. Second, all 60 cycle line voltages have been restricted to the supplementary chassis.

III. PULSE SENSOR UNITS

A. Design Considerations

The pulse sensor unit is the transducer which converts the arterial pressure information of the human subject into an input signal to the decoder unit. Several digital pulse sensor units were constructed and experimentally evaluated. Three pulse sensor units employing digital transillumination are used as transducers in this blood pressure measuring instrument. Each pulse sensor unit contains a miniature photoconductive sensor to monitor pulse amplitudes in the distal finger during a programmed occlusion by a more centrally located finger cuff. The electrical output signal from the distal pulse pickup serves as an input signal to the decoder unit as shown in Figure 5. The decoder unit functions to relate pulse amplitude changes to both the occlusion pressure and the digital blood pressure.

The electronic decoding of the blood pulse utilized in this instrument is based upon a pulse amplitude monitoring method of blood pressure measurements. This method consists of the application of an occlusion pressure to a centrally located occlusion cuff while the blood pulse is observed by a distally located pulse pickup. It has been shown previously that the variation in the amplitude of the blood pressure pulse is related to the variation in the pressure of the oc-

clusion cuff (10). This pulse monitoring method of extra-arterial blood pressure measurement is based upon the consideration that the application of an external pressure to the walls of a distensible vessel will cause the cross sectional area of the vessel to decrease as the external pressure is increased above the pressure in the vessel. The arteries of the body are not entirely plastic as they possess muscle and elastic tissue which introduces considerable elasticity to the arterial walls. It has been found that the pressure volume curve of the human aorta approximates a parabolic form for pressures below 170 mm Hg (8,22). The diameter of the aorta remains almost constant for pressures above 170 mm Hg. Therefore, the vessels of the arterial system appear to be elastic, with the elasticity decreasing for pressures above 170 mm Hg. Extra-arterial blood pressure measurements made external to the body surface must also account for the presence of muscle and other tissue between the skin and the artery. An external pressure in excess of the actual vessel pressure should be required to effect a change in the diameter of the vessel. The amount of pressure required in excess of the vessel pressure, will be a function of the type and physiological condition of the tissue between the point of pressure application and the vessel walls.

The determination of the distal pulse waveform during occlusion is of prime importance in correlating the value of

the externally applied pressure with the actual vessel pressure.

The distal pulse waveform detected by a pulse sensor placed on the body surface will differ from the intra-arterial pulse waveform in two main respects. The pulse amplitude available to the extra-arterial detector will be greatly attenuated due to the presence of various types of tissues between the artery and the body surface. The waveform may also be distorted by movement artifacts. The movement artifacts arise chiefly from compression of the vascular system as a result of muscle and tendon movement, and from motion between the pulse detector and the body surface. The effects of tissue attenuation may be counteracted by amplification of the detected pulse signal. However, if the pulse detector is to transduce the pulse waveform during exercise the movement artifact distortion in the pulse waveform must be minimized.

Several different devices were evaluated for possible use as distal pulse sensors. The pulse detector devices considered may be classified in two categories: Those dependent upon finger volume changes only, and those dependent upon the intrinsic properties of blood flow in the arterial system. Transducers using a plethysmograph principle such as a fluid sensor cuff, and transducers using a linear displacement principle such as a crystal sensor, will be considered in the former category. Transducers employing principles based upon

either the electrical impedance change of the vascular volume or the optical density variation with blood flow will be considered in the latter category. While all transducers will be affected by vascular volume changes, this effect will be of secondary consideration in the optical density transducers. As a preliminary design, a pulse sensor composed of a pressure transducer in communication with a water filled rubber cuff was constructed and evaluated. This pulse detector was combined with a centrally located occlusion cuff to form a torroidal like structure three inches long with a 0.75 inch center opening. The finger of a subject was placed in the center aperture and pressure was gradually applied to the proximal occlusion cuff when a blood pressure measurement was desired. A pulse waveform was obtained from the transducer output and subsequently recorded. This device functioned satisfactorily while the subject was restrained from moving the hand or finger during occlusion, otherwise the pulse waveform was severely distorted by movement artifact. Investigation revealed that the primary cause of the movement artifact was motion of the finger with respect to the fixed pickup housing. A contributing factor to the movement artifact was the low signal to noise ratio. The pulse signal output was approximately 5 mv and the movement artifact was characteristically random with an output amplitude in excess of 100 mv.

Consideration was given to the development of methods to eliminate the movement artifact produced by motion between the skin and the sensor. A sensor employing a barium titanite crystal was constructed. This transducer was designed to slip over the end of the finger and was spring loaded to hold the crystal against the middle phalange and the digital artery. Pulsations of the arterial wall were transferred to a pressure bar mounted along a diagonal of one face of the square crystal wafer. The crystal wafer was supported by two hard rubber posts mounted on the corners of the diagonal; opposite to the pressure bar. Thus, motion on the pressure bar was transduced to an electrical output signal by the flexure of the crystal. Movement artifact was also present in the output of this device. However, the ratio of the artifact signal to the pulse signal was reduced by a factor of ten. An analysis of this device revealed that a considerable reduction in the effect of the skin-sensor artifact would result if the mass of the pulse detector was made smaller and the detector attached to the skin surface.

A capacitive transducer having a small mass and capable of being attached to the finger with adhesive was then constructed. Variations in the spacing between the capacitor plates and in the dielectric constant of the medium separating the plates effected a change in the capacitance of this transducer. The constructed capacitive transducer uti-

used 1 cm square plates, formed of 0.002 inch thick aluminum sheet imbedded in a covering of approximately 0.002 inch latex rubber and connected to a length of low loss coaxial cable. Coaxial cable with a characteristic impedance of 50 ohms and an electrical length of 1/8 of a wavelength was used. This line length was selected to produce an impedance reflected at the input to the transducer, which would vary close to a short circuit value. The transducer was excited by a 100 megacycle signal source. The magnitude and phase of the impedance change resulting as the transducer was attached to the middle phalanx over the digital artery was investigated. The results of this experimental evaluation indicated that the application of this transducer as a digital pulse sensor would require considerable effort to reduce the transducer motion artifacts. The capacitance of the transducer was only 7$\mu\mu$f at 100 megacycle cycles when placed on the finger. Body motion in the proximity of the transducer created changes in excess of 10 percent of the resting value. Development of this type of transducer for application as a digital pulse sensor was not pursued further.

The application of an electronic impedance plethysmograph, reported by Nyboer (20), as a pulse pickup was also considered. This device operates on the principle that blood flow through the finger produces variations in both the dielectric constant and the conductance. These variations may

be studied by applying a high frequency signal to the finger and sensing the output potential intermediate to the point of application. The use of this system in a pulse sensor unit would require the development of the necessary excitation and detection circuits. The development of such a pulse sensor was not pursued, instead primary consideration was given to the design of a photoelectric pulse sensor.

B. Photoelectric Pulse Sensors

The development of a photoelectric pulse sensor, containing a photoelectric pulse detector and an occlusion cuff, was considered.

The photoelectric detector functions to convert variations in the transmitted light into corresponding electrical signals. Human tissue is transparent to radiation of wave lengths from 6000 to 11000 Angstrom units (9). Kramer et al. (13) has conducted studies for the determination of the oxygen saturation of whole blood utilizing the spectral band to which human tissue is transparent. However, the extinction coefficient of human blood, particularly in the nonhemolyzed state, is much higher than that of the tissue (13). Therefore, the pulsations of blood through the illuminated human tissue will result in variations in the amount of light transmitted through the tissue. Vasomotor activity within the human tissue will also affect the quantity of transmitted

Light.

Photoelectric methods have been used previously in the studies of peripheral circulation. Hertzman (11) applied this technique to the study of the blood volume pulse and the blood volume in the fingers. Millikan (17) used a similar principle in the measurement of blood oxygen in the ear.

The pulse detector may utilize either transillumination or parallel-illumination of the tissue volume. Transillumination, or illumination from one side of a body surface to the opposite side such as across a finger or through the ear lobe, is applicable to the study of the circulation in peripheral systems. Parallel-illumination utilizing a light transmission path parallel to the body surface can be employed in circulatory studies over the entire body surface. Weinman and Manoach (27) describe the use of a photoelectric assembly employing a parallel-illumination technique in the study of peripheral circulation.

Several photoelectric schemes, employing both types of illumination were experimentally evaluated in this study. Incandescent lamps were utilized as light sources. The emission spectrum of the tungsten filament lamp covers the spectral transmission region of human tissue (21). Both photodiodes and photoconductive cells were evaluated as photodetectors.

Type 1N77B photodiodes were combined with type T-1 1/2

lamps to form pulse detector assemblies. These pulse detectors employed both transillumination and parallel-illumination principles. The impedance of the photodiodes, a function of the light intensity directed on the photodiode, was approximately 100 kilohms at the maximum useable light intensity for digital light transmission. The maximum useable light intensity was determined by the intolerance of the subject to the local heating produced by the lamp. The 1N77B photodiode has an active light surface in the shape of a square with a side length of approximately 0.05 inches. The impinging light is directed to the photodiode element through a lens mounted on the end of the one inch long by 0.1 inch diameter glass envelope. The maximum sensitivity of this type pulse detector occurred when the lens of the photodiode was directed towards the light source. The overall size of the pulse detector assembly was approximately 2.5 x 0.5 x 0.75 inches for the transillumination device and 1.2 x 0.8 x 0.8 for the parallel-illumination device. Attempts to remove the photodiodes from the glass envelopes were unsuccessful. In addition to exhibiting a high impedance, these pulse detectors were subject to serious movement artifact effects. The movement artifacts were largely due to displacements between the skin surface and the detector housing.

The investigations of the photodiode type pulse detector revealed three considerations to be used in the design of a

photosensitive pulse detector. First a device possessing both high sensitivity and low inherent impedance for low levels of impingent light intensity is required. Secondly, a light source configuration which will produce sufficient illumination without introducing intolerable heating effects of the surrounding tissue is required. Third, in order to reduce the motion artifact, the pulse detector assembly must be constructed in a configuration that will reduce the effective motion between the body surface and the photoelectric transducer. Ideally, the only variable in the light path should be the variation in the transmitted light resulting from blood flow in the vascular volume. The pulse detector assembly must function to reduce or eliminate variations in the transmitted light resulting both from tissue volume changes due to muscle movements and from displacements between the illuminated tissue volume and the detector due to inertial effects.

A review of several commercially available photoelectric devices and incandescent lamp units was conducted. A Clairex (7) photoconductive cell, CL504SL, was selected for use as a pulse detector. The photosensitive element of this cell is cadmium selenide which has a peak spectral response at 6900 angstrom units. The cell is contained in a glass housing which is 0.260 inches in diameter and one inch long. The cell itself is of cylindrical shape, 0.14 inches in diameter

and 0.10 inches thick. The cell was removed from the glass envelope and placed in a plexiglas housing. The housing is only 0.40 x 0.80 x 0.12 inches. The plexiglas unit was painted black except for a lens area directly over the photoconductive cell.

The T-1 1/2 lamps used with the photodiodes were replaced by a subminiature lamp purchased from the Chicago Miniature Lamp Company (6). The lamp, a type CN8-666, has a five volt filament and is only 0.19 inches long and 0.10 inches in diameter.

The subminiature lamp and plexiglas mounted photoconductive cell were combined to form a pulse detector unit. The photoconductive cell and lamp were arranged in the pulse detector to utilize transillumination instead of parallel-illumination. Experimental comparison between the movement artifact resulting from each type of illumination demonstrated the transillumination technique to be superior for reduction of movement artifact. The pulse detector configuration shown in Figure 10 is the result of numerous experimental design evaluations. The use of a rubberized cloth base for the pulse sensor unit permits the detector to conform to the body surface. The four wire shielded cable connects the lamp and photoconductive cell to the pulse sensor selector switch. This cable was shielded to reduce power line interference on the transducer output signal. The cable

was mounted along the dorsal surface to strengthen the rubberized cloth during flection of the finger. The opaque rubber cover has two functions. First the cover serves to reduce the effects of background light. The CL504SL photoconductive cell possesses the requirements of high sensitivity and low inherent impedance; however, the sensitivity is such that considerable 60 cycle interference was introduced into the transducer output by the background artificial lighting. Also, movement artifact was introduced by the variation in background light intensity resulting from body motion. The dark rubber cover reduces these effects. The second function of the rubber cover is to serve as a clamping device to reduce the artifacts caused by changes in the volume of illuminated tissue and changes in the contact between the body surface and the detector. The rubber cover compresses the distal portion of the detector; however, the compression is not sufficient to occlude surface arterial blood flow. The rubberized cloth strip is rough textured around the lamp and photoconductor. The texturizing and compression of the strip serve to reduce movement between the detector and the body surface. The small mass of both the lamp and photoconductor, and the large ratio of contact surface area to the depth of the cell housing function to effectively reduce the inertial effects of this pulse detector.

The subminiature lamp and photoconductive cell were com-

bined to form another type of pulse detector as shown in Figure 12. This unit is constructed of 0.004 inch thick sheet brass shaped to fit the distal phalanx in a manner similar to the pulse detector shown in Figure 10. The proximal end of this transducer is compressed by two springs which are mounted on the side of the unit. The compression springs function in a manner similar to the opaque rubber cover of the pulse detector of Figure 10. The same type of lamp, photoconductive cell and plexiglas housing as used in the variable pulse detector was utilized. The pulse detector shown in Figure 12 is utilized in the cylindrical pulse sensor unit.

Satisfactory pulse waveforms have been obtained with both pulse detector configurations. The pulse detector exhibits an output impedance of approximately five kilohms when placed on the finger of a caucasian subject with a light intensity corresponding to a lamp excitation of two volts DC. Blood pulse waveforms with amplitudes of approximately 30 mv and 15 mv were obtained from the pulse detectors of Figure 10 and Figure 12 respectively with a two volt lamp excitation. The higher output signal level of the variable pulse detector appears to be a result of the smoother contact between the flexible cloth and the body surface of this unit in comparison to the brass sheet holder of the detector of Figure 12. Both detector configurations are essentially free of

nonviolent movement artifacts. However, movement artifact will be produced with sharp, violent motion of the finger. This artifact apparently is introduced by changes in the vascular blood volume of the distal finger introduced by the additional blood flow resulting from the centrifugal force acting on the peripheral blood mass. The vascular volume will be varied with vasomotor activity and venous occlusion due to muscle movement, thereby introducing variations in the optical density of the finger.

Three pulse sensor units were designed. One variable and two fixed type pulse sensor units are provided with the blood pressure measuring instrument. The variable pulse sensor is shown in Figure 11 and is designed to facilitate the investigation of the effects of the location and size of the occlusion cuff in the measurement of digital blood pressure. The ability to vary the location and size of the occlusion cuff used with this pulse sensor is the reason for denoting this unit as the variable pulse sensor. The pulse detector design used in the variable pulse sensor has been combined with a fixed occlusion cuff to form one of the fixed pulse sensors. The other fixed pulse sensor utilizes the pulse detector shown in Figure 12. A pulse sensor selector switch has been provided to connect the desired pulse detector to the decoder circuits and to connect the occlusion cuff to the pneumatic system.

The pulse sensor units were designed to function on any finger of a normal hand. A maximum finger diameter of 1.2 inches was used in the design of the finger aperture of the fixed occlusion cuff units. The pulse sensor selector switch is located for use with cuffs applied primarily to the left hand. The use of the left hand will permit the auscultatory measurement of blood pressure on the right arm without interference to the electronic blood pressure measurement using the pulse amplitude monitoring technique. The measurements of blood pressure as reflected in Figure 1 were taken using the auscultatory method on the right arm as a standard method. The only requirement in the application of the variable pulse sensor to the finger of a test subject is that the detector unit be slipped over the distal phalanx with the lead wires placed along the dorsal side of the finger and hand. An occlusion cuff, selected from the cuffs shown in Figure 13, should be placed on the finger with the air hose on the dorsal side and directed centrally. The choice of finger occlusion location and cuff width should be made by the investigator. The digital occlusion cuffs shown in Figure 13 are color coded to identify the cuff width. Table 2 lists the color code and cuff width corresponding to the cuff order shown in Figure 13. The occlusion cuff should be wrapped around the finger and secured by the Velcro (1) fastener. This arrangement enables the cuffs to be easily

applied to fingers of various diameters.

The variable pulse sensor is connected to the subject switch input circuit by a nine pin subminiature connector and a luer type hose connector. The variable pulse sensor may be used away from the treadmill exercise bar by connecting suitable extension cable and air hose between the pulse sensor selector switch and the pulse sensor. The shielded cable to the pulse sensor should be attached against the subject with adhesive plaster in order to obtain an electrical ground for the pulse detector. The treadmill hand bar functions to provide an electrical ground terminal during normal treadmill operation.

Figure 14 is a cross section drawing of one of the fixed pulse sensor units. In this unit, a fixed occlusion cuff has been combined with a pulse detector identical to the unit used in the variable pulse sensor. The occlusion cuff, A, has a deflated width of two inches and is attached to a brass pressure ring, B. The pressure ring functions to restrain outward inflation of the continuous occlusion cuff A. Both the front retainer ring, C, and the rear retainer ring, D, are fabricated of rubberized nylon. The retainer rings serve to contain the outward distension of the occlusion bag. The occlusion bag is connected to the air connector of the pulse sensor selector unit by the 0.125 inch outside diameter air hose, E. Both the top and bottom strips of the rubberized

cloth of the pulse sensor detector, F, are attached to the rubber aperture ring, G. The subject normally will insert a finger through the aperture ring, G, and along the pulse detector strips F until the finger reaches the blind end of the pulse detector, H. The air hose E, and the pulse detector shielded cable I, should be located on the dorsal surface of the finger and the hand. The lamp J is positioned over the cuticle and the photoconductive cell K is located below the distal arterial arch. The pulse detector will be compressed against the finger by the rubber ring L. The outer covering M, serves to support the cuff and detector unit. This covering also functions as a light shield against undesirable background illumination. Figure 15 shows the portable fixed-cuff pulse sensor applied to the index finger. The pulse sensor selector switch is shown in the lower right of Figure 15. The cylindrical fixed pulse sensor was removed from the pulse sensor switch bracket in this figure.

The second type of fixed pulse sensor unit is shown in Figure 16. The index finger of the subject has been inserted into the sensor aperture. The air hose and electrical cable connecting this pulse sensor to the pulse sensor switch unit are shown in the lower right of Figure 16. An occlusion cuff is contained in this unit along with a device to properly position the pulse detector. Figure 17 is a drawing of the pulse sensor showing the relative location of the occlusion

cuff and pulse detector unit. The occlusion cuff is actually composed of two latex prophylactics, A and B in Figure 17. These prophylactics are extremely thin and form to the contour of the finger inserted in the 1.2 inch sensor opening, C. The rubber ring D functions to contain the occlusion cuff as the pressure is increased. Ring D is sealed against the finger by the occlusion cuff. As the finger advances in the sensor opening, the palmar side of the distal portion rides along the 0.5 inch rubberized cloth, strip E. One end of this strip is fixed to the retainer ring F and the other end is spring loaded to insure the return of the strip to the proper position as the finger is removed. The dorsal portion of the finger is forced against the guide G by the spring loaded finger ramp H. This ramp rotates about a hinge connected to the retainer ring F. The retainer ring F is connected to the automatic positioner housing J by the guide G.

The occlusion cuff is restrained from motion towards the distal portion of the finger by the rubber retainer ring K. This retainer ring, K, is located between prophylactics A and B. Prophylactic A functions to retain the proper position of ring K as the finger advances towards the detector L. The distal portion of the finger enters the detector through the opening M. This detector is spring loaded by the spring N and is held in position by a Velcro (1) fastener at O and a plunger P. The finger is advanced with the detector against

the plunger P. The detector is released from the velcro (1) fastener at O and with the proper interval of advancement the spring loaded plunger P is removed from the detector. The detector is then properly positioned on the finger with the photoconductive cell on the palmar distal surface and lamp on the dorsal distal surface. Connections to the detector are made through four lead wires made of number 38 enamelled copper wire wound into springs of 0.8 inch outside diameter.

During operation, the hand is placed on the treadmill hand bar with one finger extended. The pulse sensor is advanced on the finger by turning the knob located on the rear of the pulse sensor holder. The turning of the knob, rotates a threaded drive which moves the pulse sensor unit. The release of plunger P, which indicates the proper application of the pulse detector, is shown by the lighting of a position lamp located adjacent to the pulse sensor selector switch. When the pulse sensor selector switch is rotated to the fixed pulse sensor position the operation of the position lamp is transferred to a leaf switch contact located in the positioner housing J. This switch is constructed to complete the lamp circuit only when the finger is extended. The subject may be directed to keep the lamp illuminated, thereby requiring him to keep his finger straight and relatively immobile. This procedure greatly reduces the effects of sharp,

violent motion induced artifacts in the pulse waveform output. However, the pulse sensor will function with non-violent motion even though the position lamp feature is not used. Once the pulse output is connected through the pulse sensor selector switch to the decoder unit, the occlusion cuff is ready for the application of pressure. The container Q is constructed of plexiglas and functions as a pressure capsule for the occluding cuffs A and B. Container Q encloses the automatic position housing. The sensor input and output wires are removed from the unit by running them between prophylactics A and B. The wires are then passed around the top of the retainer ring F adjacent to guide bars G and along the outside of the container Q where they are connected to a pulse sensor input cable which enters the rear of the pulse sensor outer housing R. Air pressure enters in the rear of the housing R and is passed through a tube in the rear of the pressure container Q to the inside of the container Q.

The pulse sensor is shown in Figure 18 with the center section of the outer cover removed. The outer cover is composed of three parts. The pressure container held against the sensor opening ring by retaining bolts mounted on the rear plate and shown in Figure 18. The pressure container may be seen inside the end plate supports of the outer container. Removal of the pressure container reveals the oc-

clusion cuff assembly and the automatic positioner housing as seen in Figure 19. The pulse detector cable is shown extending along the occlusion cuff. The detail in the automatic positioner may be observed in Figure 20.

The three pulse sensor units were designed and constructed to minimize the movement artifacts introduced by body motion. Each of the pulse sensor units will transduce the distal blood pulse amplitude variation with changes in the pressure of the occlusion cuff during periods of treadmill exercise. However, distal pulse amplitude variations, will also result from changes in the vasomotor activity of the peripheral circulation. The acral circulation, particularly in the fingers and toes, is subject to extreme fluctuations with temperature changes, pain, and emotions, according to Mendlowitz (16). The vasomotor activity may be stabilized by inhibiting sympathetic nerve discharge to the hand. The digital circulation should then reflect the non-neurogenic or intrinsic changes of the entire systemic circulation (16). The inhibition of the sympathetic nervous system through the use of drugs does not appear feasible during the treadmill exercise evaluation.

Experimental evaluation of the effects of digital vasomotor activity on the amplitude of the pulse sensor output has shown that a pulse waveform, which is satisfactory for obtaining pressure measurements, can be obtained after a

brief stabilization and subject familiarization period. The lamp of the detector cuff produces local heating which aids in the vasodilation of the arterial bed in the distal phalanx. This vasodilation is reflected as an increase in the output pulse signal amplitude. The apprehension, to placing a finger within the pulse sensor unit, experienced by some subjects appears to be allayed after a few measurement cycles. The inflation of the occlusion cuff affects a sensation similar to a mild gripping of the finger. The tingling sensations which are sometimes experienced with prolonged application of a brachial occlusion cuff are not present, even for digital occlusion periods in excess of five minutes.

C. Pulse Sensor Selector Switch

The pulse sensor units are connected to the decoder unit through the pulse sensor selector switch. This switch permits the investigator to connect the pulse sensor in use to the decoder. The switch unit also serves to terminate the input to the decoder unit when the pulse sensor is not in use. Figure 21 is a schematic diagram of the pulse sensor selector switch showing the connection to the pulse sensor units. A component list for this circuit is presented in Table 3. This pulse sensor selector switch may be switched to one of three positions. In the fixed position the cy-

lindrical fixed pulse sensor is connected to the decoder unit. Lamp voltage is supplied to the pulse detector lamp LP_1 from the pulse sensor light intensity control R_2, which is located on the decoder unit. Voltage is applied to the photoconductive cell PC_1, through the 15 foot pulse sensor cable from the load resistors R_3 and R_4 of the decoder unit. The photoconductor is operated with a bias current of approximately 3 ma. The output signal from the photoconductive cell is capacitively coupled to the decoder unit. The fixed position of the pulse selector switch activates the pulse detector of the position circuit which functions to signify that the finger is extended. The proper application of the pulse detector to the distal phalanx is indicated by the position circuit when the sensor switch S_1 is in the off position. When the sensor selector switch is in the off position a one kilohm resistor is connected in parallel with the photoconductor output terminals, thus insuring that the removal of the lamp voltage from the pulse detector unit will not introduce interference signal into the decoder unit as a result of the high dark impedance of the photoconductor.

Both the portable fixed-cuff pulse sensor and the variable pulse sensor units are connected to the pulse sensor switch unit by a nine pin subminiature connector. This connector and the luer air connector are shown in Figures 15 and 16. When the pulse sensor selector switch S_1 is placed in

the variable position, the nine pin connector is connected to the decoder unit. The pulse sensor selector switch also functions to connect the occlusion cuff of the pulse sensor selected to the air hose contained in the pulse sensor input cable.

IV. ELECTRONIC DECODING AND CONTROL CIRCUITS

A. Amplifiers and Decoders

The electronic decoding and control circuits translate changes in the blood pulse waveform into control signals to the pneumatic system. The decoding and control circuits will be presented in separate sections of this chapter. The decoder unit will be discussed first. Figure 22 is a block diagram of the decoder unit. Figures 23 and 24 show the top and bottom of the decoder unit. The front of the unit is located in the bottom of Figure 23 and the top of Figure 24.

A short discussion of the decoder operating controls should be of assistance in presenting the theory and operation of the decoder circuits. Figure 25 shows the front of the decoder unit. All operating controls of the instrument are located on the front panel of the decoder unit except for the pulse sensor selector switch which is located on the pulse sensor assembly attached to the treadmill hand bar. Table 4 lists the function of each of the operating controls. The pulse sensor selector switch serves to connect the pulse sensor to the decoder unit. The amplitude of the recorded pulse waveform may be changed by varying the pulse sensor light intensity control and the sensitivity control. Once a proper waveform amplitude has been obtained, the pump switch is turned to the on position. The maximum manual pressure

control should then be set to the desired initial pressure if the manual mode of operation is to be utilized. The minimum diastolic pressure control should be set for use in the automatic mode of operation. The mode of operation is determined by the position of the mode selector switch. However, the pump switch should be turned to the on position before the mode selector is spring loaded to return to the center position from the manual mode. Therefore, the switch must be depressed and held for manual operation. The occlusion cuff is vented when the mode selector is returned to the center position. However, the pulse waveform will be recorded as long as the pulse sensor selector switch is turned to the proper pulse sensor position.

In addition to the operating controls on the decoder and pulse sensor units, calibration and adjustment controls are located on several of the subsystem units. Adjustment of these controls is required during the maintenance alignment procedure and will be discussed in subsequent sections.

The theory of operation of the decoder circuits is based on the monitoring of the characteristic amplitude changes in the pulse waveform with variations in the occlusion cuff pressure. During the manual mode of operation the pulse waveform amplitude diminishes rapidly to zero as the occlusion pressure is increased above the systolic value. When the pressure diminishes to just below systolic, the blood

pressure pulse is passed to the distal pickup and a pulse output is presented from the pulse sensor. It has been observed that the occurrence of the first maximum in the amplitude of the envelope of the pulse waveform is an index of diastolic pressure (10). The values of systolic and diastolic pressure thus obtained are shown by the S and D respectively in Figure 4a. The pattern of the pulse waveform amplitude changes found in the automatic mode is shown in Figure 4b. During the interval AB, pressure is increased rapidly to a pre-diastolic value and the pulse waveform shows little amplitude change. During the interval from B to C the pulse waveform remains constant until diastolic pressure is reached at point D, after which the pulse waveform diminishes to zero at above systolic pressure. The value of pressure corresponding to zero pulse waveform amplitude again corresponds to systolic pressure. During the period E to F the pressure is returned to zero and the pulse waveform again returns to the normal amplitude.

The decoder unit was designed to translate the pattern of pulse waveform amplitude changes in the automatic mode of operation to indices of systolic and diastolic pressure. The manual mode has been included in this instrument chiefly as a system calibration mode. Therefore, the pulse waveform in manual mode will not be decoded in this instrument. Decoding of this pulse waveform will be performed visually from

the recording.

The decoding of the pulse waveform in the automatic mode is accomplished in the circuits shown in the block diagram of Figure 22. The output of the pulse sensor is passed through a balanced differential preamplifier. The decoder preamplifier is located on top of the decoder chassis and is shown in the left of Figure 11. The preamplifier is constructed on a separate subchassis with twisted-pair shielded cable used for input and output connections.

Three outputs are obtained from the preamplifier. A balanced output signal is fed to the Sanborn driver-amplifier. Both positive and inverted waveforms are available as inputs to the diastolic and systolic decoding circuits respectively. The input to the systolic decoder is amplified and fed to a monostable multivibrator. The output of this multivibrator is a positive pulse of fixed duration for each input pulse. This output is processed in a grid charge circuit and functions to control a relay driver stage. The absence of an input to the systolic circuit for a preset interval of time is passed to the control circuits as an indication that the occlusion pressure has exceeded the systolic pressure of the vessel. The decoder amplifier tubes are located adjacent to the preamplifier chassis as shown in Figure 23. All low level pulse circuits are contained on the left side of the chassis. The monostable multivibrator

circuits are located on the right rear section of the unit. The relay drivers are located on the right front corner of the decoder chassis.

In the diastolic decoding circuit, a positive going output waveform from the preamplifier is amplified and then passed to a cathode follower. The output of the cathode follower is fed to a diastolic detector circuit. This circuit determines changes in the amplitude of the pulse waveform (2). The output of the detector circuit is passed to an amplifier and subsequently to a monostable multivibrator. The output of the multivibrator is fed to the charge circuit and relay driver in a manner similar to the systolic decoder. The diastolic detector permits pulses to pass to the multivibrator until a decrease in pulse amplitude occurs. A decrease in pulse amplitude results in an output change in the relay driver which is an indication that the occlusion pressure has exceeded the diastolic pressure of the vessel.

The large cylindrical component located in the center of the decoder chassis in Figure 23 is the pressure transducer. The thyratron, V_9, is located adjacent to the transducer. Both the transducer and the thyratron are part of the pressure transducer circuit. A test socket is located to the rear of the thyratron. Control switches, potentiometers, and the pressure decay valve are mounted on the front panel of the decoder unit.

The detailed operation of the decoder unit will be presented with reference to the schematic diagrams and voltage waveforms of the decoder circuits. Figure 26 is a schematic diagram of the differential preamplifier. Table 5 is a component list for this circuit. The input to the preamplifier is taken from the shielded housing of the pulse sensor connector shown in the upper left of Figure 24. The pulse sensor output is passed to the grids of V_1 through input networks consisting of 0.05 mfd coupling capacitors and five megohm grid resistors. This coupling network functions to reduce the low frequency variation in the pulse waveform resulting from changes in the blood volume of the finger. Capacitors C_3, C_4, C_5, and C_6 serve as high frequency by-pass networks. There are units of the "150" series recorder which utilize a radio frequency heater power supply of approximately 2.5 megacycles. The cathode circuit of V_1 consists of R_7, R_8, and R_9. The common cathode resistor, R_9, provides in phase rejection, chiefly to reduce power line interference. The plate circuit consists of plate resistors R_{10}, R_{11}, the sensitivity potentiometer R_{12}, and the resistor R_{13}. The sensitivity control is located on the front of the decoder unit and is shown in Figure 25. The networks composed of R_{14}, R_{16}, R_{15}, and R_{17} are bias networks available for possible future use as biased output networks to the direct coupled driver amplifier of the Sanborn recorder. The

normal output of V_1, is through the networks C_7, R_{18} and C_8, R_{19} to the second amplifier stage.

The second amplifier stage, V_2, is located on the center of the preamplifier chassis shown on the left in Figure 23. The cathode circuit of V_2 consists of R_{20}, R_{21}, R_{22}, and R_{23}. The balance control R_{22}, is located on the side of the preamplifier chassis and permits adjustment of V_2, a 5814 medium μ triode, to obtain balanced amplifier operation. The plate circuit of V_2 is similar to the plate circuit of V_1, with the exception that the sensitivity resistor R_{26} is not adjustable. The second stage of the preamplifier, also functions as the preamplifier to the Sanborn recorder which is direct coupled. Therefore, the output of this stage must supply the required grid bias voltage to the first stage of the driver-amplifier. The resistance network composed of R_{27}, R_{29}, R_{28}, and R_{30}, is connected to the negative voltage supply and provides a voltage of 45 $\pm$1 volt for bias purposes. The preamplifier stage signal output to the driver-amplifier is passed through this network and is connected to terminals 13 and 14 of the amphenol connector, J_3, located on the rear of the decoder unit. The positive going signal is connected to terminal 13 in accordance with the input circuit requirements of the driver-amplifier. Terminal 15 is the common system ground and completes the balanced output to the driver-amplifier. Figure 27a shows a typical pulse waveform input to the preamplifier

from the pulse sensor. The sweep time of the photographs of Figure 27 is 1 cm per second. The pulse detector waveform of Figure 27a has an amplitude of approximately 30 mv. The preamplifier output waveform to the Sanborn driver-amplifier is shown in Figure 27b. The amplitude of this waveform is approximately 1.5 volts and will result in a 2 cm recording when passed to the driver-amplifier.

The preamplifier output waveform is also amplified by V_3 and the associated circuits. The grid input circuit consists of C_9, R_{31}, C_{10}, and R_{32}. This amplifier stage is also of differential design. The cathode circuit consists of R_{33}, R_{34}, and R_{35}. The plate circuit consists of R_{36} and R_{37}. The output from the amplifier stage to the systolic and diastolic decoders is shown in Figures 15 and 15d. The output waveform from the pulse sensor, corresponding to the digital pulse waveform, is effectively differentiated by the RC coupling networks of each amplifier section. The waveforms of Figures 15c and 15d are comparable to waveforms of the second derivative of the digital pulse computed with graphical techniques by Burch (4) from data obtained with a digital plethysmograph. The differential of the pulse waveform is demonstrated by comparing the waveforms of Figure 27. The section of the pulse waveform in the neighborhood of the dicrotic notch exhibits the most pronounced change in waveform shape.

The use of the balanced differential amplifiers in this

instrument was necessitated by the requirement to separate the desired pressure pulse waveform from an input signal containing considerable 60 cycle interference (26). The digital pulse sensor signal is composed of approximately 150 mv of interference imposed on 30 mv of desired signal information. The first preamplifier stage has a common mode rejection of approximately 3000. The succeeding stages of the amplifier provide additional common mode signal rejection; however, these stages function chiefly as amplifiers and signal inverters for the output signal of V_1.

The inverted pulse pressure waveform from V_3 of the preamplifier is passed to the systolic decoder circuit. The schematic diagram of the systolic decoder circuit is shown in Figure 28. Table 6 is a component list for this circuit. The inverted pulse waveform, as shown in Figure 27c, is passed through C_{11} to the gain control R_{38}. This potentiometer is seen between tubes V_4 and V_7, adjacent to the preamplifier chassis in Figure 23. The signal is amplified and inverted in tube V_{4A} and passed to the monostable multivibrator through the coupling capacitor C_{12} and the waveshaping circuit consisting of CR_1, CR_2, R_{41}, and R_{42}. This waveshaping circuit clamps the pulse waveforms to a negative potential of approximately 52 volts. The diode CR_1, then clips the positive going pulse. Only pulses with signal amplitudes in excess of the negative clamp-off voltage es-

established on the grid of V_{5A} by the voltage divider network composed of R_{41}, R_{42}, R_{44}, and R_{45} in conjunction with V_{5B} are passed to the grid of V_{5A}. Proper adjustment of the stability potentiometer, R_{41} will permit V_{5A} to remain in the nonconductive state until a positive going signal from V_{4A} causes V_{5A} to conduct. This diode waveshaping circuit is required to limit the effects of spurious signals resulting from the high amplification of the digital pulse waveform. The decoding of a systolic pressure index depends on the determination of the last pressure pulse waveform prior to systolic occlusion. The sensitivity of the decoder circuit therefore depends on the ability of the electronic circuitry to distinguish between the last blood pulse waveform and the inherent system noise. Ideally, the stability potentiometer R_{41} should be adjusted to enable only the positive pulse waveform to pass to the grid of V_{5A}. Thereby, all undesired signal components, or noise, will be clipped by the diode waveshaping circuit.

The positive output pulse from the diode circuit causes V_{5A} of the monostable multivibrator circuit to pass to the conducting state, which effects an output signal from the multivibrator stage. The operation of this multivibrator is fully described in Millman and Taub (18). The multivibrator stability adjustment, R_{41}, and systolic multivibrator tube V_5 is located on the right rear of the decoder unit chassis. For

each positive pressure pulse the monostable multivibrator output is a 50 millisecond square wave pulse with an amplitude of 40 volts. This multivibrator output pulse is coupled to the relay driver stage by the charge circuit.

Capacitor C_{15} in combination with semiconductor diodes CR_3, CR_4, resistors R_{48}, R_{49}, and capacitor C_{16} form the storage circuit for the relay driver stage V_{6A}. The primary purpose of this storage circuit is to insure the occlusion pressure has increased beyond systolic pressure before the relay control circuit effects the release of occlusion pressure. The output voltage pulse of V_5 is coupled through C_{14} and clamped to ground by CR_3. The diode CR_4 permits the positive pulse to pass to the grid of V_{6A} and to the shunt capacitor and resistor. Capacitor C_{16} charges towards the positive multivibrator pulse voltage and discharges through R_{48} and R_{49}. The voltage on C_{16} functions to drive the relay driver stage into conduction. The plate circuit of the relay driver tube, which contains the relay K_1, is connected in series through the mode selector switch, S_{4B}, and contact pole 3 of relay K_5 to the B+ power supply. Therefore, the relay driver stage will only operate when this series path is properly completed. The functioning of the relay circuits will be discussed in connection with the control relay unit. A positive voltage in excess of seven volts at the grid of V_{6A} is sufficient to insure that relay K_1 will be energized

thereby permitting the occlusion pressure to continue to increase. Capacitor C_{16} is charged by the output pulse of the multivibrator. The series resistors, R_{48} and R_{49}, are connected in parallel with C_{16} and form a discharge circuit for the capacitor. The potentiometer, R_{48}, permits the discharge time constant for C_{16} to be varied. The period of time for which the relay K_1 will remain energized after the last digital pressure pulse can be varied from 0.3 to 1.5 seconds. However, the decay time of the storage circuit must be sufficient to insure that K_1 remains energized between succeeding digital pulses. The range of decay time corresponds to heart rates of 200 to 40 beats per minute respectively. A decay time setting of 1.5 seconds will insure that the circuit will operate within the specifications for subject pulse rates as shown in Table 1.

The primary purpose of the diastolic decoder is to obtain an index of diastolic pressure from the digital pulse waveform. The input to the diastolic decoder is a pulse waveform from the preamplifier chassis. The output of this circuit is in the form of a relay position change in the control relay unit. The diastolic decoder circuit monitors the pulse amplitude and senses a decreasing of the pulse amplitude. The occurrence of a diminishing pulse amplitude corresponds to an increase in occlusion pressure over diastolic pressure and is therefore an index of diastolic pressure.

The diastolic decoder is similar to the systolic decoder in theory and operation. The major difference is the addition of a diastolic detector circuit (2). Figure 29 is a schematic diagram of the diastolic decoder. Table 7 is a component list for this circuit. The positive going output of the preamplifier is passed through the RC ccupling network composed of C_{17} and R_{52} to an amplifier stage. The output of the amplifier, V_{7A}, is passed through a capacitor, C_{18}, to the diastolic gain adjustment, R_{55}. The gain adjustment, R_{55}, is located on the left side of the decoder chassis and serves to establish the input voltage level to the cathode follower.

The cathode follower serves to isolate the amplifier stages from the charge circuits of the diastolic detector. The output of the cathode follower is a pulse waveform which has been differentiated by the interstage coupling networks of the amplifier stages. The output of the cathode follower is coupled by the capacitor C_{19} to the diastolic detector circuit.

The operation of the diastolic detector circuit will be presented with reference to Figure 30. An oscilloscope sweep of 1 cm per second was used for this photograph. Both Figures 30a and 30b represent an amplitude sensitivity of 10 volts per cm of oscilloscope deflection. A typical voltage waveshape appearing at the junction of C_{19}, CR_5, and CR_6 is

shown in Figure 30a. The first two pulses of Figure 30a are slowly increasing in amplitude. The base line of Figure 30a represents zero voltage. The positive portion of the pulse waveform causes conduction through CR_5 with a resultant increase in the voltage of capacitor C_{20}. This increase in the voltage of C_{20} is shown by the two steps in the waveform of Figure 30b. The voltage on C_{20} will increase, with each pulse, until a charge balance between capacitors C_{19} and C_{20} has been attained. During the portion of the pulse waveform when the pulse amplitude is less than the voltage on C_{20} the capacitor discharges through R_{57} due to the blocking effect of CR_5. This discharge is shown by the decrease in amplitude of the voltage waveform of Figure 30b. The discharge time of C_{20} and R_{57} is large, as the RC time constant of this circuit is approximately one minute. The third pulse of Figure 30a serves to demonstrate the operation of the diastolic detector as the pulse amplitude decreases. Once the gradually increasing occlusion cuff pressure has exceeded the diastolic pressure of the artery, the amplitude of the distal pulse waveform will gradually decrease to zero. The rate of decrease of the pulse amplitude is dependent upon the rate of increase in occlusion pressure.

The latter pulses shown in Figure 30a represent pulses of decreasing pulse amplitudes occurring with the increase in the occlusion pressure. When the amplitude of the positive

going pulse of Figure 30a has become less than the voltage of capacitor C_{20}, conduction through CR_5 ceases. If the rate of decrease in the pulse amplitude is greater than the discharge rate of capacitor C_{20}, the voltage on the capacitor will continue to discharge as shown in Figure 30b. The negative portion of the waveform of Figure 30a is conducted through CR_6 and appears as an output signal across R_{58}. The voltage in capacitor C_{20} functions to cause the average value of the diminished pulse waveform to remain above zero as the capacitor discharges, thereby permitting the pulse waveform to become completely positive. Once the pulse waveform becomes positive the output of the diastolic detector through CR_6 ceases. If the amplitude of the pulse waveform were to remain at the new, but lower, voltage level the voltage of C_{20} would eventually decrease until the pulse voltage exceeds the capacitor voltage. Charging of C_{20} would then be effected and a waveform similar to that shown on the left of Figure 30b would result. The diastolic detector is not dependent upon the amplitude of the pulse voltage for proper operation. The diode and capacitor charge circuits will function to stabilize the diastolic detector for operation at voltage levels corresponding to the input pulse. The output of the amplifier V_{4B} is passed through C_{22} to the diastolic multivibrator sensitivity adjustment R_{62}. The operation of the remaining stages of the diastolic decoder circuit is identical with that described for the systolic decoder circuits.

D. Decoder Unit Calibration

The calibration of the decoder unit consists of the adjustment of the gain and stability of the decoder circuits to insure proper interpretation of the variations of the pulse waveform. The decoder unit calibration has been divided into two parts, a maintenance alignment and an operating alignment. The maintenance alignment is designed to calibrate the decoder circuit, chiefly through the chassis adjustments. The operating alignment should be performed at the start of the treadmill exercise evaluation. This alignment establishes the proper input waveform to the decoder unit.

The maintenance alignment procedure must be performed on the decoder unit while the unit is operational. A six foot patch cord has been constructed which will permit the operation of the decoder unit external to the Sanborn recorder. The patch cable has connectors which mate with the decoder unit and the driver-amplifier unit. The connector which mates with the decoder is fitted with test points for all the circuits connecting with the driver-amplifier. A balanced phone type jack is also provided to permit the monitoring of the preamplifier output to the recorder. The connecting cable is shown attached to the decoder unit in Figure 31.

The maintenance alignment of the decoder unit is contained in Table 8. Table 8 is the maintenance alignment procedure for all units of the blood pressure measuring instru-

ment. While a separate alignment of only the decoder circuits is possible, a composite alignment for all circuits is more feasible in consideration of the interconnections between the units. The maintenance alignment of Table 8 is designed to calibrate the pressure control circuit, the pneumatic system, the recycle circuit, in addition to the decoder circuits. Figure 32 is a schematic diagram showing the location of the decoder unit chassis maintenance adjustments. The decoder circuits are calibrated using a pulse waveform of 2 cm deflection on the recorder tracing. This amplitude of recorded waveform corresponds to an output voltage of 1.2 volts from the preamplifier chassis of the decoder unit. The decoder unit will function with pulses of greater amplitude. However, the use of this pulse amplitude for calibration procedures, both in the maintenance and operating alignments, will provide sufficient sensitivity in the decoder circuits to insure their proper functioning even with pulses of 0.8 cm amplitude. The maintenance alignment requires the measurement of the voltages required for the proper operation of the instrument. The value of voltage and the corresponding measurement location are presented in Table 9.

The operating alignment procedure is presented in Table 10. While the maintenance alignment involves other units of the instrument, the operating alignment involves only the front panel controls of the decoder unit and the pulse sensor

selector switch. The operating alignment should be completed each time the pulse sensor is connected to the subject. Item eight of this alignment is concerned with obtaining a recorded pulse waveform of 2 cm amplitude which exhibits a minimum of power line interference. Variation in the amplitude of the pulse waveform may result from changes in the vasomotor activity of the finger. The pulse sensor lamp intensity control may be adjusted to restore the amplitude of the pulse waveform.

C. Pressure Control Circuits

The pressure control circuits function to provide the programmed pressure operation of the pneumatic system. Also, these circuits afford remote switching of AC line voltages, thereby permitting line voltages to be excluded from the proximity of the preamplifier and amplifier circuits of the decoder unit. The interconnections of the control circuits with the other units of the instrument are shown in Figure 33. There are four major control circuits; the pressure transducer circuit, the power relay and negative power supply circuit, the programmed control relay circuit, and the automatic timer circuit.

First, consider the operation of the pressure transducer circuit. This circuit converts air pressure changes into electrical signal outputs to a second recorder channel and to

the programmed relays. The output to the programmed relays is used to control the flow of air in a manner which will produce an occlusion pressure waveform as shown in Figure 4.

The pressure transducer circuit is composed of a Bourns pressure transducer, a 2D21 thyratron, pressure selection switches, potentiometer adjustments, an output relay, and the necessary interconnecting networks. Figure 34 is a block diagram of the pressure transducer circuit. Air pressure from the pneumatic system is transduced to a voltage signal by the pressure transducer. The output of this transducer is connected to a balanced output jack for interconnection to the DC input jack of a separate recording channel preamplifier unit. A Sanborn ECG preamplifier is used for this purpose and is located below the decoder unit in the recorder cabinet. The interconnection between the units may be seen in Figure 8. The output of the transducer is also passed through a network of pressure selection switches. These switches permit the selection of the maximum manual pressure and the minimum diastolic pressure as presented in Table 4. The transducer output, as obtained from the switch network is connected to a thyratron relay driver stage.

The schematic diagram for the transducer circuit is shown in Figure 35. A list of components is presented in Table 11. Several transducers were investigated for use in this circuit. These included differential transformer types,

resistive strain gage types, commercially available bourdon tube-photoconductive types, and experimental types with both continuous and digital outputs from a bellows-photo sensitive semiconductor element. The transducer used in this instrument is a Bourns type 509, and is constructed with a bellows linked to a 5000 ohm, 0.5 watt potentiometer. The pressure range of this transducer is 0.10 psi. The availability of a comparatively high voltage DC output, in the order of 50 volts for full scale pressure, permitted the use of this transducer in a thyratron control circuit without further amplification. The frequency response characteristics of this transducer were evaluated experimentally and determined to be adequate for this application.

Air pressure in the pneumatic system is transmitted to the pressure transducer by a network of polyvinyl hoses as shown in Figure 24. The transducer is shown in Figure 23 as the large cylinder in the center of the decoder chassis. Voltage is applied to the transducer through the series resistors R73 and R74. The arrow shown alongside the transducer in Figure 35 corresponds to the direction of transducer resistance change with an increasing pressure. The voltage applied across the transducer is adjusted by R_{74} which is connected to the negative 100 volt power supply. The output voltage of the transducer is connected to both an output attenuator network and a pressure selection network.

The output attenuator network functions to establish an output voltage to the Sanborn ECG preamplifier which is both proportional to the transducer output voltage and has a zero pressure value at approximately ground potential. As pressure is increased in the transducer the voltage across the series resistor R_{77} and R_{78} increases to approximately 13.8 volts for an air pressure of 300 mm Hg in the occlusion cuff system. The voltage across R_{78} is connected to the air pressure DC output voltage jack, J_4. The voltage divider network R_{78} and R_{79} reduces the transducer output voltage J_4 to 5.5 volts for an air pressure of 300 mm Hg. This value of voltage was selected to enable the use of the sensitivity and attenuator controls of a Sanborn ECG preamplifier in establishing the calibration for the occlusion cuff pressure recording. The air pressure DC output voltage jack is connected by a jumper cable to the DC input jack of the ECG preamplifier. The calibration procedure is described in Table 8 and consists of setting the ECG sensitivity to provide a full scale recording which will correspond to a pressure of 150 mm Hg if the ECG preamplifier attenuator is set on 10, and 300 mm Hg if the attenuator is set on 20. At the request of Dr. Gustafson of the United Heart Station, a base line for the occlusion cuff pressure recording of 5 mm from the right margin will be utilized. The capacitor, C_{27}, serves to integrate rapid pressure variations, thereby producing a smoother recorded pressure curve.

The transducer output voltage is also connected to the pressure selection switch network and subsequently to the thyratron, V_9. The function of the pressure selection network is to provide an input voltage to the grid of the thyratron which will cause the tube to conduct when a selected air pressure has been attained in the pneumatic system. Switch S_2 is the minimum diastolic pressure control. Switch S_3 is the maximum manual pressure control. Switch S_{4A} is a section of the mode selector switch. All these switches are located in the front of the decoder unit. Switch S_2 is connected through switch S_{4A} to the grid of the thyratron when the mode selector is in the automatic position. Switch S_3 is connected to the thyratron for the manual mode of operation only. As the pressure in the transducer is increased the voltages on the contact points of switches S_2 and S_3 become less negative. The voltage on a particular switch position will pass through the thyratron firing potential when the pressure in the transducer reaches the value noted in Figure 35. Switches S_2 and S_3 are normally preset to the desired thyratron firing pressure. The firing potential of the thyratron is -96.5 volts and is established by the network composed of the Zener diode, Z_1 and the resistor, R88. When the grid voltage reaches the firing potential the tube conducts and K_3 remains energized until the plate circuit of V_9 is interrupted.

The plate circuit of the thyratron contains a potenti-

ometer, R_{07}, relay K_3, switch S_{4C}, and contact 2 of relay K_4. The potentiometer enables the adjustment of the thyratron conduction current to insure the proper functioning of K_3. Switch S_{4C}, which is a segment of the mode selector switch, deactivates the thyratron circuit whenever the mode selector switch is in the center position. Contact 2 of the relay K_4 is normally open, therefore the thyratron circuit will not function until K_4 is energized. The operation of K_4 will be discussed as part of the relay control unit.

The relay control output of the pressure transducer circuit is combined with outputs from the decoder operating controls, and the systolic and diastolic decoders to form the inputs to the programmed control relays. These relays are located in the control relay unit which is mounted on the supplementary chassis. The units mounted on the supplementary chassis are shown in Figure 36. The power supply unit is located on the left. The control relay unit is next to the power supply. The programmed pressure valve unit is mounted next to the pump which is on the right. The knob on the top of the control relay unit is the automatic timer adjustment. The connections to the relay unit are shown in Figure 8. Figure 33 is a diagram of the cable connections to this unit. The 6.3 volt AC control and thyratron heater, the 115 volt AC relay and pump power, and the negative 100 volt supply voltage are obtained from the power supply unit. Output jacks

to the decoder unit, programmed pressure valve unit, and the pump are located on the control relay unit. Connecting plugs and cables are attached to the unit concerned. Another output from the relay unit is a cable to the relay spark suppression capacitors which are mounted below the control relay unit.

Figure 37 shows the inside of the control relay unit. Relay K_1, the systolic decoder relay, is located on the left, or the rear of the relay unit. Relay K_3, the pressure transducer relay, is located between relay K_1 and relay K_2, the diastolic decoder relay. The next relay, K_4, functions as part of the automatic timer circuit as does relay K_5 which is located in the center foreground. The second relay from the right, operates in the mode selector circuit. The last relay K_7, functions in the line power control circuit. The small tube chassis located in the lower left of Figure 37 contains the time delay relay, K8, and potentiometer, R87, of the pressure transducer circuit.

Figure 38 is a schematic diagram of the control relay unit. A component list for this unit is presented in Table 12. When the main power switch of the Sanborn recorder is turned to the on position, power is supplied to the negative power supply unit. This unit then supplies power to the relay control circuit. Heater voltage is applied to the thyratron, V_9, whenever the main power switch is in the on position.

This procedure insures that the pressure transducer circuit will function when the pump is turned on without loss of control during a warm-up period. The pump switch, S_5, also serves to control the application of line power to the control relay circuits. Closure of S_5 applies 6.3 volts AC to relay K_7 and lamp PL_3. Relay K_7 then connects power to the pump and the contacts of K_6. Relay K_6 is energized by 6.3 volts AC from a segment of the mode selector switch S_{4D}. This relay is shown in Figure 38 in the unenergized state, corresponding to both the stand-by and automatic mode position.

Let us consider the relay functions for the automatic mode of operation. In the automatic mode, AC line voltage from K_7 is applied to the contacts of K_1, the systolic decoder relay. With the mode switch in automatic S_{4B} applies plate voltage to both the relay driver stages of the decoder unit through the normally closed contact 3 of relay K_5. Therefore, relays K_1 or K_2 will be energized when the corresponding relay driver stage is in the conducting state, the mode switch is in the automatic position, and relay K_5 is not in the energized position. The occurrence of arterial pulses effects a switching of K_1 to the closed position which applies line power to relay K_4 through contact 1 of relay K_3. When relay K_4 is energized the thyratron pressure control circuit is completed to relay K_3 by the closing of contact 2 of K_4. The switching of contact 3 of relay K_4 from the normally closed to the energized position connects AC line voltage from K_7 to

terminal 12 of terminal board TB-1. Terminals 12, 13, 14, and 15 of terminal board TB-1 are connected to the programmed pressure valve as input signal terminals. This valve has four stable positions, and changes position when line voltage is applied to the proper input terminal. Application of line voltage to terminal 12 corresponds to setting the valve to permit the occlusion pressure to make an initial pressure jump as shown in Figure 4b. As the air pressure is increased the pressure transducer circuit functions to energize K_3 to correspond to the pressure selected by the minimum diastolic pressure control, S_2. When K_3 is energized, 115 volts AC is applied to terminal 13 to switch the valve to obtain the diastolic rate of increase in the occlusion cuff air pressure. Relay K_3 will remain energized until the thyratron plate circuit is interrupted by the opening of either K_4 or S_{4C} of the mode selector switch.

During the period in which the occlusion pressure is increasing at the diastolic rate, corresponding to the interval BC of Figure 4b, the diastolic decoder relay is energized. When diastolic pressure is attained, corresponding to point D of Figure 4b, the input to the diastolic monostable multivibrator ceases and after a delay of approximately 1.5 seconds the diastolic relay K_4 is un-energized, which then applies line voltage to the valve unit through terminal 14. The valve unit is now switched to the systolic air flow rate, corre-

sponding to the interval CE of Figure 4b. When systolic pressure is attained, at point S, the input to the systolic monostable multivibrator ceases and after a delay of approximately 1.5 seconds the diastolic relay K is un-energized. As relay K_1 is returned to the off position, line power is applied through contact 2 of the energized thyratron relay, K_3, to relay K_5. When relay K_5 is energized, the recycle timing circuit is activated. The automatic system recycle consists of four steps. First, relay K_5 is locked in the closed position by the voltage supplied to contact 1 from the normally closed contacts of the thermal time delay relay, K_8. Secondly, heater voltage is applied to the time delay relay. Third, the plate circuits of both the systolic and diastolic relay driver stages are deactivated by the opening of contact 3 of K_5. The closing of relay K_5 also removes voltage from the coil of relay K_4 by the opening of contact 4 of K_5. When the coil circuit of relay K_4 is opened, the relay returns to the open position, thereby interrupting the thyratron control circuit to relay K_3. The opening of relay K_4 also connects terminal 15 of TB-1 to the AC line power which will position the programmed valve unit to permit the occlusion cuff to vent all air. The last step in the automatic system recycle is the functioning of the time delay relay, K_8. The opening of the contacts of K_8 removes the lock voltage to relay K_5, which will cause the relay to open. The opening of relay K_5 will

complete both the relay driver plate circuit and the coil circuit of relay K_4. Thus, the control system has completed one cycle in the automatic mode. If arterial pulses are available at the pulse sensor, then K_1 energizes and the automatic relay cycle is repeated. The recycle rest interval, the time required for the thermal time delay relay to function, may be varied by either adjusting R89 or by interchanging time delay relays. Two Amperite time delay relays have been provided with this unit, one relay has a nominal delay time of five seconds and the other has a nominal delay of 15 seconds. Recycle rest time intervals of from two seconds to 1.5 minutes have been obtained by varying both the thermal relays and the setting of R_{89}.

The movement of the mode selector to the manual position closes relay K_6, which causes K_4 to become energized. The closing of K_4 connects line power to terminal 12 of TB-1 which causes the value assembly to rotate to the position which will permit the rapid build-up of system pressure. The closing of K_4 also completes the thyratron control circuit. When the pressure in the system reaches the pressure selected on the maximum manual pressure control, S-3, the thyratron is fired and relay K_3 is energized. The closing of relay K_3 connects the programmed value unit to the AC line power so that the value unit may rotate to a closed position. Thus, the occlusion cuff has been filled with air at the pressure level

selected on the maximum pressure control. The bleed out of pressure in the manual mode of operation is effected by varying the manual pressure decay control. The pressure in the occlusion cuff will also be vented by the return of the mode selector to the center position. When the mode selector switch is in the center position, line power is applied to terminals 13, 14, and 15 of terminal strip TB-1 which switches the programmed pressure valve to the stand-by position, thereby venting the occlusion cuff to the atmosphere. The center position of S_4 also removes plate voltage to the thyratron control and decoder relay driver circuits.

The negative power supply unit, while not an integral part of the control relay unit, is directly concerned with the functioning of the control circuits. The inclusion of a power supply in addition to the supply contained in the driver-amplifier system of the Sanborn recorder was due to the inability of the recorder supply to furnish the required 30 ma of current at a negative 100 volts potential. The output characteristics of the negative power supply are presented in Table 13. Figure 39 is a schematic diagram of the power supply unit shown on the left in Figure 36. A component list for this circuit is presented in Table 12. Full-wave rectification is achieved by V_{10}, a 6X4 full wave rectifier diode and regulation are accomplished by V_{11}, a voltage regulating tube.

V. PNEUMATIC SYSTEM

The purpose of the pneumatic system is to supply air pressure to the occlusion cuff. The major components of the system are shown in Figure 40. The pump supplies air to the programmed pressure valve. The programmed pressure valve directs the air flow through the air pressure rate valves to the mode selector switch. Air is passed through the mode selector switch to the occlusion cuff to obtain the occlusion pressure waveforms of Figure 4.

A continuous duty pump, capable of delivering 400 cubic inches of air per minute at 18 psi, was selected as the system air pressure source. The pump is mounted on four springs which in turn are mounted on a chassis which "floats" in a holder composed of one inch foam rubber. This assembly is seen on the right of Figure 36. The "floating" mounting of the pump essentially eliminates all pump vibrations to the recorder unit.

Air input to the pump is through an input muffler located below the pump unit. The pump output is connected to the valve assembly on the top of the programmed pressure valve unit as shown in Figure 36. The system pressure is restricted to less than 10 psi by an overpressure bleed-out valve located on the top of the pressure valve unit. This bleed-out valve is designed to protect the transducer, which has a full scale pressure of 10 psi, from damage due to over pressurization.

Several systems were evaluated in the design and construction of the pressure valve units. A programmed pressure valve system using two stepping relay valves and two variable air orifices was chosen for development. The chief advantage of this system is the sequential program built into the relay valves and the automatic control relay circuit. The choice of the sequential system, a system using the satisfactory completion of one event to initiate a second event, instead of a system using a timed cycle of operation was based upon the shorter occlusion cycle time of the former type.

Since the determination of both the systolic index and the diastolic index is based upon the observation of digital pulse amplitude changes with variations in the occlusion cuff pressure, the accuracy with which the systolic and diastolic pressure measurements are made is dependent upon both the rate of change of the occlusion pressure and the heart rate of the subject. If a change in the successive pulse amplitude is taken as an index of diastolic and systolic pressure, the pulse changes can only be observed with each heart beat. Let us consider a normal heart rate of 72 beats per minute having a pulse period of 0.83 seconds. If the occlusion pressure is increased at a linear rate of 10 mm Hg per second the pressure will change 8.3 mm Hg in one pulse period. Let us now consider that the index of either diastolic or systolic pressure occurs at a time corresponding to the mid-point of a pulse

period. Then there exists the possibility of an error of 4.15 mm Hg between the value of the occlusion cuff pressure at the time of the index and the actual occlusion cuff pressure, which caused the change in the arterial flow characteristics. Therefore, the error between the measured and the actual values of systolic and diastolic pressure may be decreased by reducing the increase in occlusion pressure per heart beat.

It may be seen from Figure 1 that the anticipated change in diastolic pressure with exercise will be considerably less than the change in systolic pressure. Since the chief use of this instrument will be to investigate pressure changes during treadmill exercise greater accuracy in measuring the diastolic pressure than in measuring the systolic pressure is desirable due to the smaller changes involved. Therefore, an increase in occlusion pressure of approximately 5 mm Hg per second for the diastolic pressure measurement and approximately 10 mm Hg per second for systolic pressure measurement is desirable. The accuracy in determining the occlusion pressure corresponding to the blood pressure indices is increased with exercise due to the resultant rise in the heart rate. A heart rate of 120 beats per minute with an occlusion pressure increase of 10 mm Hg per second corresponds to a possible error of 2.5 mm Hg between the value of the occlusion cuff pressure at the time of the blood pressure index and the actual occlusion cuff pressure, which effected the change in the blood

flow in the artery.

The pressure control system used in the programmed relay control unit consists of two valve units which are switched in the proper sequence by control voltages from the control relay unit. The switching of the valve units functions to direct the flow of air through a system of variable orifices which control the rate of pressure increase in the occlusion cuff system.

A schematic diagram of the pneumatic system is shown in Figure 41. Air from the pump passes through a system of pipes and hoses to the two position valve unit. The position of the valve unit is controlled by the position of the rotor of the stepping relay connected to the valve. The output of the valve units is connected to the air orifices or to the pressure transducer and occlusion cuff in order to produce the occlusion pressure waveform as shown in Figure 4b.

An investigation and evaluation of commercially available solenoid valves suitable for use in this system was made. Satisfactory commercial valves were not available, and a valve unit was constructed for use in the pneumatic system. Figure 42 is a photograph of the programmed pressure valve assembly with the top cover removed. A component list for this unit is presented in Table 12. Two relay-valve units were constructed by modifying a standard 24 position, spring return stepping relay. The stepping relays were adapted for continuous ro-

tation. The relay stepper rotates 15 degrees with each application of line voltage to the relay coil. The relay coil was connected in series with a 24 position switch which is mounted between the valves and the relay. A normally closed contact was added to the stepping relay to insure that the coil circuit would be interrupted after each 15 degree rotation. Each relay unit has three stable positions as shown in Figure 41. The programmed relay valve, PRV_1, shown on the top in Figure 41 corresponds to the relay valve shown on the top in Figure 42. Standard metal laboratory stopcocks are used for the valve units. The stopcocks are mounted to the stepper relays as shown in Figure 42. Relay valve, PRV_1, functions to provide the rapid increase in pressure corresponding to the interval AB of the occlusion waveform of Figure 4b and also to vent the occlusion system to the atmosphere. Relay valve, PRV_2, functions both to vent the pump during the rest intervals and to provide the proper rate of increase in the occlusion pressure.

The electrical connections to the programmed relay unit are shown in Figure 41. While each valve unit has three stable positions, the programmed relay unit has four stable positions. These positions are shown in Figure 41, which has been drawn to correspond to the rest or stand-by position. All segments of the 24 position switch on the stepper relays not used as control positions are connected in common. When

the pump switch, S-5 is placed in the on position, line voltage is connected to these common contacts, which causes the stepper relay to rotate to a control contract which is not energized. The control contacts are energized by the control relay unit to insure the programmed relay unit is in the proper stable position. The electrical connection to the control relay unit is made through a seven wire cable and plug, shown in Figures 41 and 42.

The interconnections between the sections of the pneumatic system contained on the supplementary chassis and the decoder unit are shown in Figure 9. Two polyvinyl air hoses are connected to the copper decoder air input pipes which extend through the driver-amplifier chassis of the Sanborn recorder.

In position 1, the occlusion cuff system and the pump are both vented to the atmosphere. When relay K_4 of the control relay unit is energized, in either the automatic or manual mode of operation, the programmed relay unit rotates to position 2. The movement to position 2 is accomplished by rotating PRV_1, through 180 degrees and PRV_2 through 135 degrees. Air now flows directly from the pump to the occlusion cuff through PRV_1. If the automatic mode of operation is being used air will also flow through PRV_2, the variable orifice, VO_1, and the mode selector switch to the occlusion cuff. Position 2 effects the rapid increase in air pressure required

in both modes of operation. Once the air pressure has increased to the value preselected to operate the pressure transducer circuit, the programmed valve unit is advanced to position 3. The change of operation from position 2 to position 3 is accomplished by rotating PRV_1, by 45 degrees. In this position, air flow to the occlusion cuff is directed through the variable orifice, VO_1. The orifice, VO_1, controls the rate of pressure increase during the diastolic decoding period. When the diastolic decoder relay K_2 is released the programmed relay valve is switched to the fourth operating position. The valve unit PRV_1 is now in position 4 and the rotation of PRV_2 by 45 degrees places both units on the fourth position. The fourth stable position permits air to flow through both VO_1 and VO_2, thus obtaining the systolic rate of increase in pressure. The application of voltage to the programmed relay unit from the control relay unit signifying the completion of the occlusion pressure cycle returns both relay valves to the first position. The programmed relay valve unit is dependent upon the control relay unit for all position control voltages. However, the valve units will only operate in the sequence described above, as the satisfactory completion of the pressure operation at each position is required to produce the proper voltage from the control relay unit to enable the stepping relay to advance to the next position.

In the manual mode of operation the occlusion cuff system is connected to VO_3 instead of VO_1 and VO_2. Therefore, when the maximum manual pressure is attained the programmed relay valve will be switched to position 3 without a further increase in the cuff pressure. Counter clockwise rotation of the manual pressure decay control results in the opening of VO_3 with a subsequent decrease in the occlusion cuff pressure. The return of the mode selector switch to the center position returns the programmed valve assembly to the first position which will vent the occlusion cuff.

At the time of initial design and construction of the pneumatic system, an investigation of commercially available air flow control valves failed to produce a satisfactory valve. The design and construction of the variable orifice valves was undertaken. A converted automotive valve assembly was used as a housing for the adjustable valve. The valve is held against a rubber valve seat by a valve spring. A set screw adjustment has been positioned against the piston to facilitate changing the valve aperture. Clockwise rotation of the set screw increases the valve orifice. In addition to a variable orifice, the rate of increase of occlusion pressure during the systolic detection to interval is controlled by the selection of one of three fixed orifice units. The fixed orifice units are located on the top of the valve unit shown in Figure 36. These fixed orifices utilize lengths of 0.58 mm

inside diameter tubing which have been attached to luer connectors and then molded in the form of a cylinder. The fixed orifices facilitate the rapid selection of a calibrated occlusion pressure rate. The set screw assembly of the variable valves was replaced by a knob and gear assembly in the construction of the pressure decay control, VO_3. This variable orifice is shown mounted on the rear of the front panel of the decoder unit in Figure 23. Recently, a commercial valve has been produced which has the same type of flow characteristics as the locally produced orifice valves (19). The commercially available valves are recommended as replacement or substitution components for the orifice valves of the pneumatic system.

The location of the variable orifices VO_1 and VO_2 and the unit interconnecting hoses are shown in Figure 36. The rate of pressure increase during the diastolic and systolic sensing intervals is adjusted by varying VO_1 and VO_2 respectively while observing the recorded occlusion pressure waveform. A uniform rate of increase in the occlusion pressure during the jump, diastolic sensing, and systolic sensing intervals of the occlusion cycle is desirable. In general, a uniform rate of increase in pressure represents the most satisfactory compromise between the overall system accuracy and the total cycle time required. Instead of a uniform rate of increase in pressure a non-linear rate of increase in pressure is attained

(24). This non-linear rate of increase in the occlusion cuff pressure is a result of non-linear air flow. A uniform rate of increase in the occlusion pressure may be produced by varying the orifice diameter. The development of such an orifice diameter change device was not pursued.

VI. MEASUREMENTS

The electronic blood pressure measuring instrument was used to obtain measurements from subjects both at rest in the laboratory and on the treadmill ergometer. A laboratory simulator of the Sanborn "150" recorder was assembled at Iowa State University in Ames, Iowa. A system operational evaluation and calibration was performed using this simulator. The variations in digital pressure measurements caused by changes in the digital occlusion cuff size and location were demonstrated. The blood pressure measuring instrument was installed in the Sanborn "150" series recorder at the United Heart Station of the Iowa Methodist Hospital in Des Moines, Iowa. Blood pressure measurements were made during the standard treadmill test.

A simulator unit, which exhibited the electrical operating characteristics of the Sanborn "150" recorder was assembled. The decoder unit and supplementary chassis assembly were attached to an equipment rack as shown in Figure 6. The handbar shown in the center of Figure 6 serves to simulate the function of the treadmill handbar. The pulse sensor selector unit was mounted to this handbar. The electrical power normally supplied by the Sanborn power supply through the driver-amplifier unit was supplied from two Hewlett-Packard power units. A type 710A power supply was used to supply both 6.3 volt AC heater voltage and the regulated B+ voltage of 250

volts DC. A type 721A semiconductor power supply provided the regulated 4.2 volts DC for the pulse sensor lamp circuit. A dummy load stage was constructed to simulate the loading effect of the Sanborn driver-amplifier. The recorded outputs from the blood pressure measuring instrument were obtained on a Sanborn "Twin-Viso" recorder. One channel of this recorder was connected in parallel to the dummy load and served to record the blood pulse waveform. The other channel of the recorder was used to record the occlusion cuff pressure waveform from the DC air pressure output voltage jack. A Tektronix Type 502 oscilloscope was used for calibration and monitoring of the decoder circuits.

The laboratory simulator was used in performing a system operating evaluation of the blood pressure measuring instrument. All units of the instrument were interconnected and properly aligned and calibrated as outlined in Tables 8 and 10 respectively. Figure 43 is a recording of a typical blood pressure measurement obtained in the manual mode of operation. The variable pulse sensor unit was placed on the middle finger of the left hand of a 21 year old caucasian male. The pulse waveform was recorded on the top channel of the recorder. A 2.5 cm wide occlusion cuff was located centrally over the proximal interphalangeal joint. The digital occlusion cuff pressure was recorded on the lower channel. The marks along the bottom of Figure 43 denote time intervals of one second

duration.

When the mode selector switch was depressed the pressure in the occlusion cuff increased rapidly to the preselected maximum manual pressure value of 150 mm Hg. The pressure in the digital cuff may be increased to 150 mm Hg in 0.8 seconds from the time of depression of the mode selector. Once the air pressure reaches the maximum pressure selected the pressure control circuit halts the further increase of the pressure. The air in the occlusion cuff is then permitted to vent from the cuff at a rate determined by the manual pressure decay control. The instrument will continue to permit the flow of air through the decay control until the mode selector is released and permitted to return to the center position.

The amplitude of the blood pulse exhibits a rapid decrease to zero as the occlusion cuff pressure is increased to above systolic pressure. The return of the blood pulse signifies that the occlusion pressure on the arterial wall has decreased below the actual systolic pressure in the digital artery during the preceding pulse interval. The index of the occurrence of systolic pressure may be taken as a point located half a pulse period prior to the first blood pulse following occlusion. The occlusion pressure applied to the finger can be determined from the value of the occlusion cuff pressure recording at the time of the systolic index. Figure 43 demonstrates a systolic pressure of 90 mm Hg.

The amplitude of the pulse waveform continues to increase until the occlusion cuff pressure becomes less than the diastolic pressure of the digital artery. The index of the occurrence of diastolic pressure may be taken as the mid-point between the first blocd pulse after occlusion which exhibits a maximum amplitude and the pulse immediately preceding. A diastolic pressure of 64 mm Hg is displayed in Figure 43. The changes in the shape of the blood pulse as the occlusion pressure is decreased are typical and are exhibited on all recordings.

The return of the mode selector switch to the center position causes the programmed pressure valve unit to rotate to the pressure relief position. The increase in the occlusion pressure waveform occurring as the occlusion cuff is vented, results from the rotation of the programmed valve unit through the systolic and diastolic air flow positions.

The operation of the electronic blood pressure measuring instrument in the automatic mode of operation is shown in Figure 44. This figure was recorded immediately following the manual operation, but for clarity in presentation has been reproduced below Figure 43. Figure 44 demonstrates two automatic measurement cycles. The first cycle demonstrates automatic operation utilizing only the systolic rate of pressure increase. The second cycle demonstrates operation with both the diastolic and systolic rate of increase in occlusion

pressure. In the first cycle the pressure increases rapidly to the preselected minimum diastolic pressure when the mode selector is placed in the automatic position. Once the initial jump pressure has been reached, the programmed valve unit is switched to direct the air flow through the systolic orifice to obtain the systolic rate of pressure increase. A pressure increase rate of approximately 10 mm Hg per second is shown in Figure 44.

The amplitude of the pulse waveform diminishes as the occlusion cuff pressure is increased above diastolic pressure. The index of diastolic pressure may be taken as the mid-point of the pulse interval between the last blood pulse before the pulse amplitude decreases and the first blood pulse with a decreased amplitude. The occlusion cuff pressure corresponding to this index can be determined from the value of the occlusion cuff pressure waveform occurring at the time of the index. A diastolic pressure of 60 mm Hg is demonstrated in the first cycle of Figure 44. The systolic index may be taken as the mid-point between the last blood pulse before occlusion and the estimated location of the next pulse if occlusion had not been attained. A systolic pressure of 90 mm Hg is shown in Figure 44. The occlusion cuff pressure is permitted to increase for a period of 1.5 seconds after the last blood pulse, then the pressure is vented to zero. The occlusion cuff pressure will decrease from 150 mm Hg to 0 mm Hg in approxi-

mately 1.9 seconds. The pressure was applied to the occlusion cuff for 9.5 seconds in the first cycle of Figure 44.

The automatic recycle circuit determines the rest interval between cycles during the automatic mode of operation. The pneumatic system is vented to the atmosphere during the rest period. A rest interval of 8.5 seconds is demonstrated in Figure 44. At the end of the rest interval the occlusion pressure is increased to the preselected minimum diastolic pressure value. The diastolic rate of pressure increase shown in Figure 44 is approximately 5 mm Hg per second.

A diastolic pressure of 62 mm Hg is demonstrated on the second cycle of Figure 44. The switch from the diastolic to systolic rate of pressure increase should occur approximately 1.5 seconds after the diastolic index has been reached. However, the change of pressure rate has been delayed until seven seconds after the diastolic index in order to demonstrate the changes in the pulse waveform following the occurrence of diastolic pressure. The systolic rate of pressure increase is 10 mm Hg per second. A systolic pressure of 90 mm Hg is demonstrated. The remainder of this cycle is identical to the first cycle shown in Figure 44. The total time required for the second cycle is 17 seconds. The use of the diastolic rate of increase in the occlusion pressure has the effect of lengthening the measuring period. Measurements of the blood pressure may be made without using the diastolic detector

circuit. However, the diastolic detector circuit functions to increase the accuracy with which the diastolic pressure may be determined from the recorded data.

The values of systolic and diastolic pressures measured in the automatic mode as shown in Figure 44 are in close agreement with the measurements of the manual mode of Figure 43. However, measurement of the blood pressure by the auscultatory method using a brachial cuff gives a systolic pressure of 118 mm Hg and a diastolic pressure of 72 mm Hg for this subject. The auscultatory measurement was taken immediately following the digital measurements.

The correlation between the occlusion cuff pressure and the actual pressure in the digital artery was investigated. The actual instantaneous blood pressure in the arterial system can only be obtained by a catherization procedure. Facilities were not available at Iowa State University to perform this procedure on human subjects. Therefore the investigation was made using extra-arterial techniques for blood pressure measurement. A reference blood pressure measurement using the left arm was made with the normal auscultatory method. The digital pulse detector of the variable pulse sensor was positioned over the distal phalanx of the middle finger of the left hand. Both the pulse waveform and the brachial arm cuff pressure variation used in the auscultatory measurements were recorded. The occlusion cuffs shown in Figure 13 were used to

obtain the digital blood pressure. The cuffs were applied to three sites on the middle finger; around the proximal phalanx; around the proximal interphalangeal joint; and around the middle phalanx. The 6 cm wide cuff was applied over the middle and proximal phalanges. The pressures measured by the pulse amplitude monitoring method were expressed as a percentage of the pressure measurements obtained by the auscultatory technique. The pressure measurement tests were performed with six male subjects and one female subject. The median subject age was 25 years.

The mean value of these measurements for each cuff size and site of application is shown in Figure 45. Each measurement obtained from the recorded output of the blood pressure measuring instrument has been expressed as a percentage of auscultatory measurement. The measurement at the brachial location was obtained simultaneously with the auscultatory measurement.

The most salient feature of Figure 45 is the consistently low relative measurements obtained in the area of the proximal interphalangeal joint. Measurements made with the digital occlusion cuff around the middle phalanx produced characteristically higher readings than found at the joint. Both Burch (4) and Mendlowitz (16) report blood pressure measurements comparable to the values of systolic and diastolic pressure measured around the interphalangeal joint. However, neither

investigator reported the occlusion cuff size or location of measurement. The investigations of the correlation between the occlusion cuff size and location were made using other fingers of the hand with the same type of results. There appeared to be little variation in the value of blood pressure determined at the proximal interphalangeal joint for all eight medial digits of a subject. The consistency of measurement between fingers appears to eliminate the possibility that this effect is the result of variations in either the vascular system or the skeletal structure of the hand (12).

The higher relative pressures measured at the proximal and middle phalanges appear to result from the difference in tissue type and structure present in these areas compared to the interphalangeal joints (12). The investigation of the relationship between the pressures measured at the interphalangeal joint and the pressures measured by the auscultatory method may be continued using the electronic blood pressure measuring instrument. An attempt to make a conclusive interpretation of the results of Figure 45 is not justified at this time due to the small number of subjects studied.

The blood pulse waveforms shown in Figures 43 and 44 are typical of the pulse recordings obtained when the pulse sensor light intensity control has been properly adjusted. The pulse sensor light intensity should be adjusted to obtain a pulse waveform which exhibits a minimum of signal interference. A

low level of pulse detector illumination results in an output which normally contains considerable interference. However, if the pulse sensor light intensity is increased to the maximum illumination, a pulse waveform signal which is essentially free of both power line and movement artifact interference is obtained from the pulse detector. Operation of the pulse detector at the maximum light intensity produces a signal which is greatly reduced in amplitude and the effects of local tissue heating, particularly with vascular occlusion, become pronounced and pain may result. The pulse sensor light intensity should be adjusted for an intermediate level which will produce an acceptable pulse waveform without painful tissue heating.

The pulse sensor units have been used to make measurements on both light and dark pigmented subjects. A slightly higher level of pulse sensor lamp intensity is required for measurements on dark pigmented subjects than on caucasians. However, this level of lamp illumination is still below the level required to introduce thermal pain sensations.

Figure 47 is a recording obtained during a treadmill exercise test performed at the United Heart Station of the Iowa Methodist Hospital. The electrocardiograph of the subject is recorded on the second channel of this recording. The top channel of the recorder is not presently utilized. The pulse waveform and the occlusion cuff pressure waveform

are recorded on the third and fourth channels respectively. The portable pulse sensor unit was applied to the index finger of the subject's left hand. The subject was connected to a spirometer during the exercise test.

The recording shown in Figure 46 corresponds to the initiation of treadmill exercise. The notations on the recording were made during the test. The subject began to walk on the treadmill at the exercise notation. The ECG recording reflects body movement of the subject. The blood pressure measurement cycle is not disturbed by the motion of the subject. The recorder paper speed for this recording is 0.5 cm per second. The occlusion cycle is approximately 11 seconds and the rest cycle is approximately five seconds. The exercise evaluation was conducted using the standard test procedures for obtaining the electrocardiogram and the spirogram; however, blood pressure was measured during the entire evaluation by the electronic blood pressure measuring instrument. The measurement of blood pressure during periods of treadmill exercise will aid in the investigation of the cardiovascular system.

VII. SUMMARY AND CONCLUSIONS

An investigation of electronic methods of measuring the blood pressure of a human subject during periods of treadmill exercise has been made. Classical extra-arterial methods of sphygmomanometry were reviewed. Specifications for the automatic measurement of blood pressure during treadmill exercise have been presented, and the causes and effects of movement artifact have been discussed. An instrument has been designed and constructed which determines the blood pressure using a pulse amplitude monitoring technique. The design and construction of the measuring instrument have been presented in terms of three major subsystems: The pulse sensor units, the electronic decoding and control circuits, and the pneumatic system.

Several pulse sensor units were discussed. Photoelectric pulse sensor units containing miniature photoconductive cells were developed and evaluated. The design and operation of the decoder and control circuits have been presented, and the development of a programmed occlusion cycle has been discussed. The functioning of the pneumatic system during both the manual and automatic modes of operation has been described.

Blood pressure measurements have been conducted using the electronic measuring instrument. Systolic and diastolic blood pressures were obtained during treadmill exercise evaluations. The instrument was employed to make a brief study of the re-

lationship of the size and location of the digital occlusion cuff to the blood pressure measurement. The results of this investigation were presented.

The electronic blood pressure measuring instrument constructed and evaluated in this investigation serves to demonstrate the engineering feasibility of measuring blood pressure with electronic techniques.

VIII. BIBLIOGRAPHY

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X. FIGURES

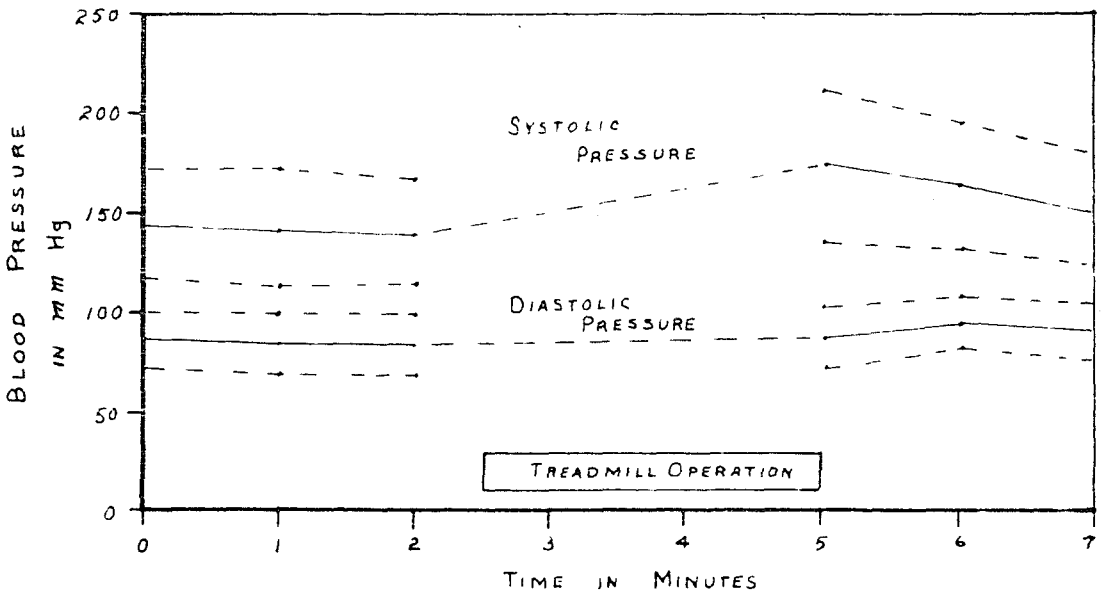


Figure 1. Variation in systolic and diastolic pressure with treadmill exercise

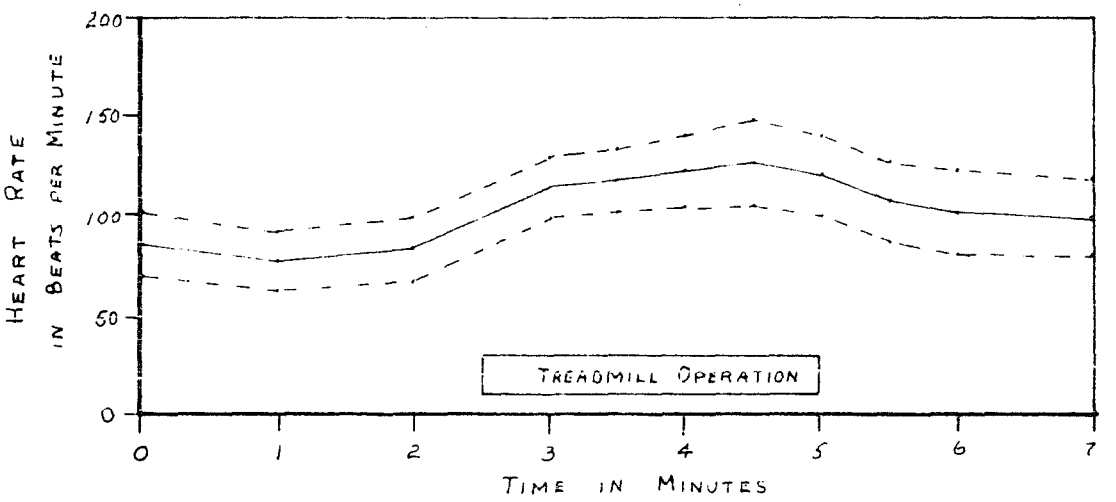


Figure 2. Variation in heart rate with treadmill exercise

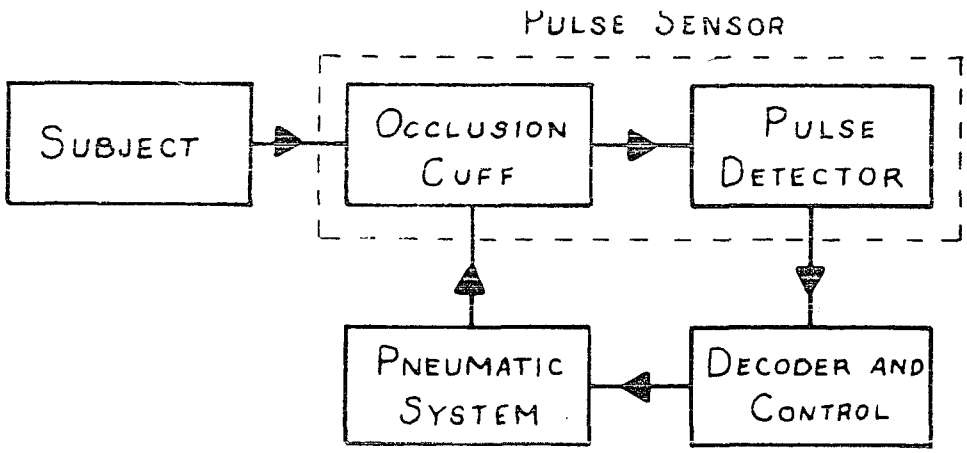


Figure 3. System block diagram for the electronic blood pressure measuring instrument

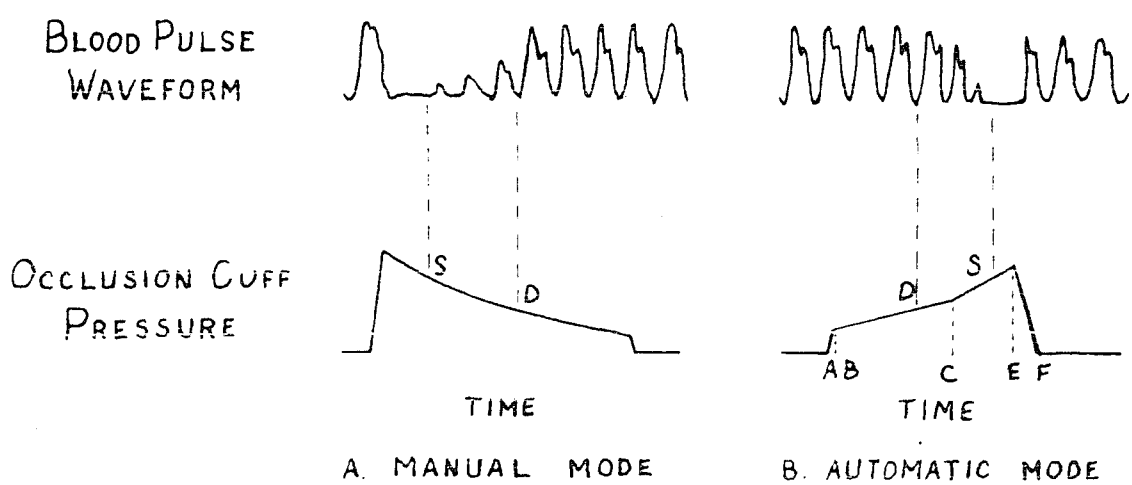


Figure 4. Blood pulse waveform

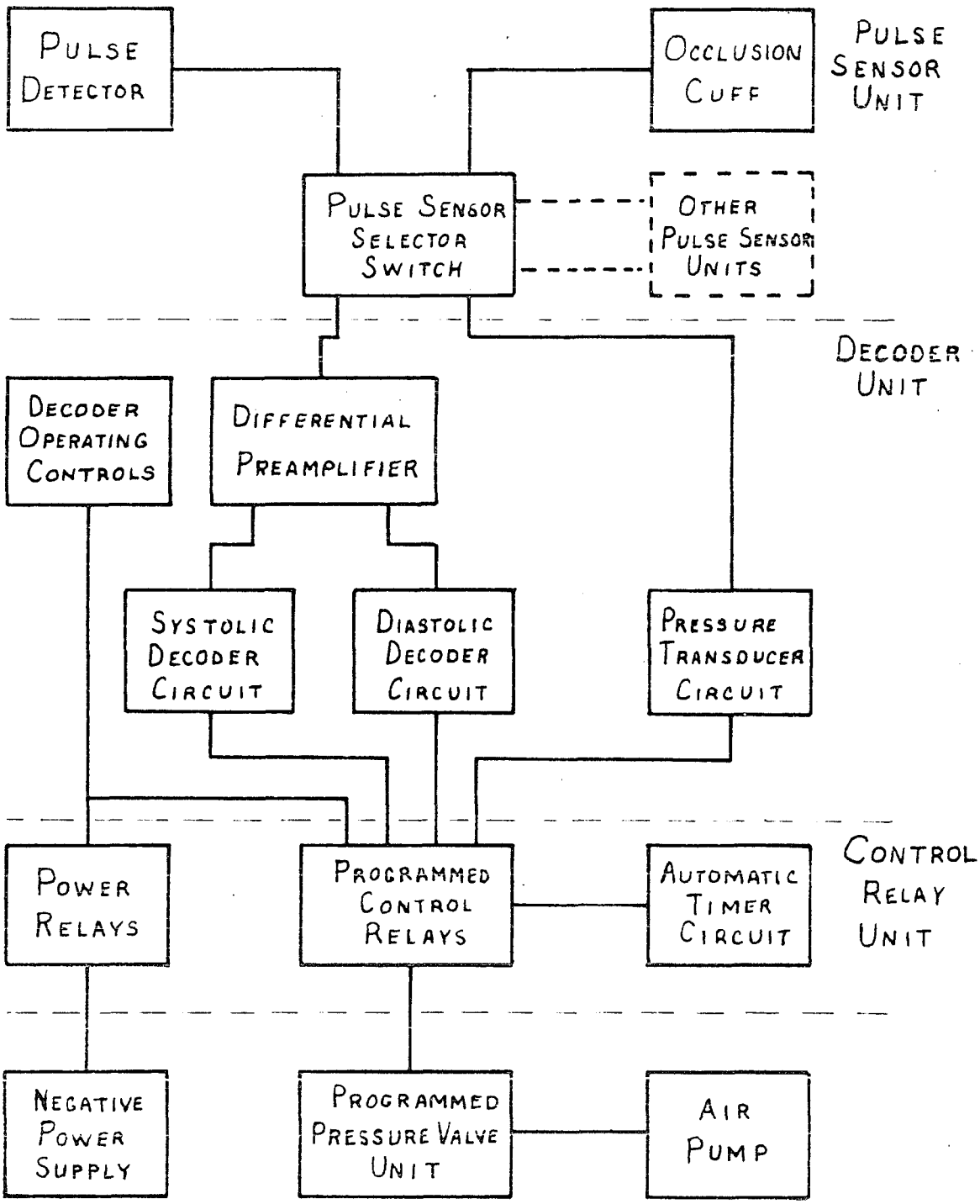


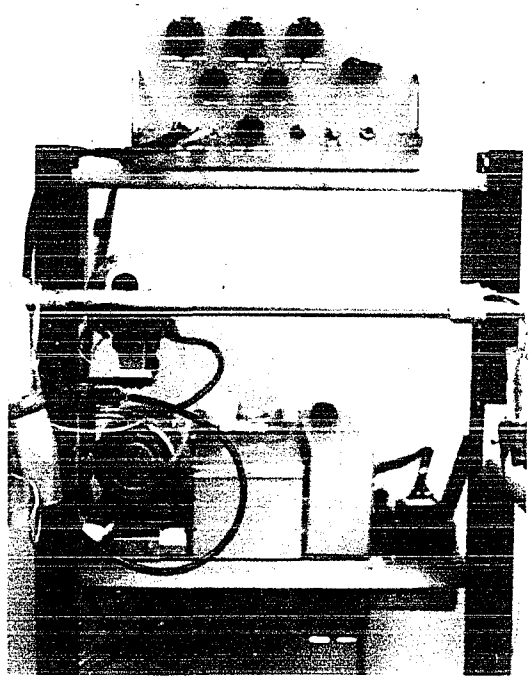
Figure 5. Block diagram of the major units of the electronic blood pressure measuring instrument

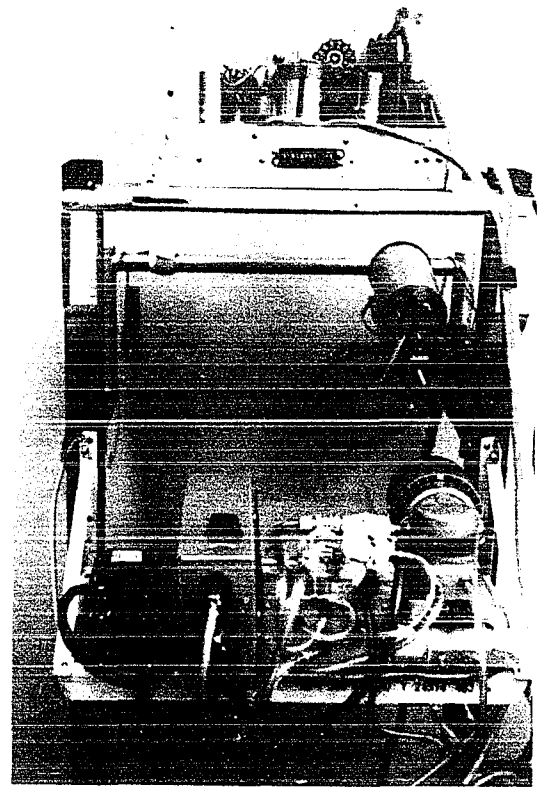
Figure 6. Front view of decoder, pulse sensor, and supplementary chassis units

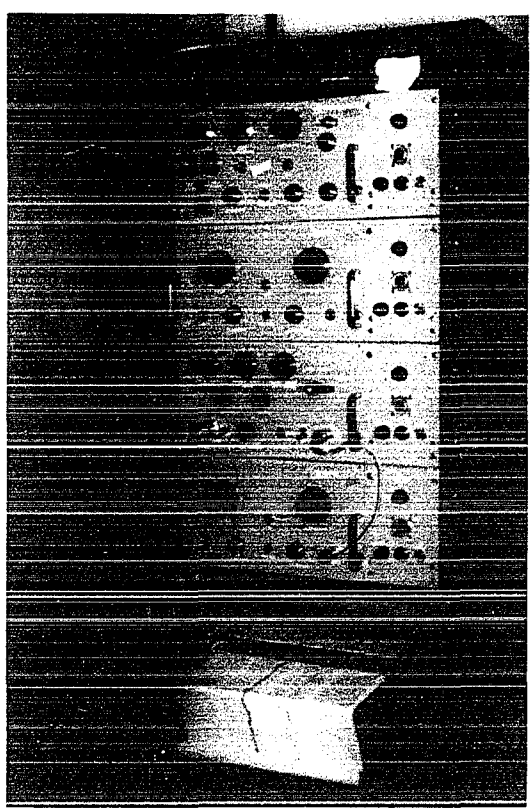
Figure 7. Rear view of decoder, pulse sensor, and supplementary chassis units

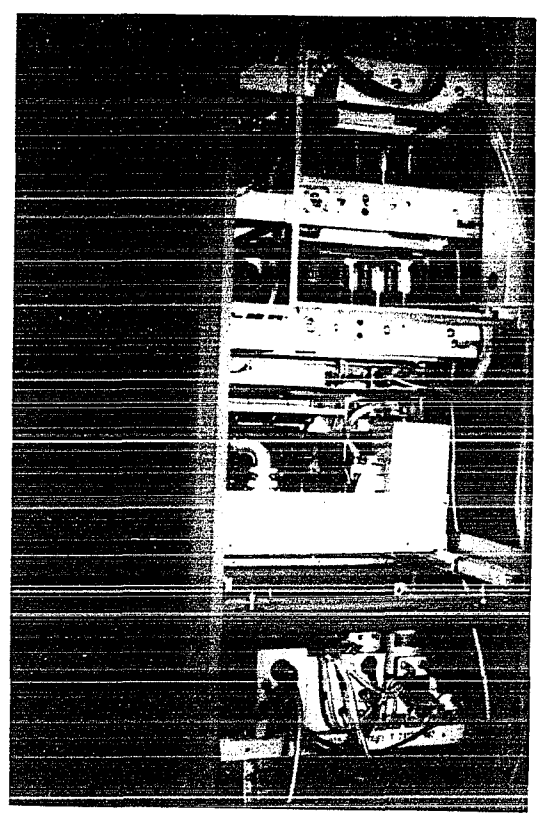
Figure 8. Front view of blood pressure measuring instrument installed in the Sanborn recorder

Figure 9. Rear view of blood pressure measuring instrument installed in the Sanborn recorder









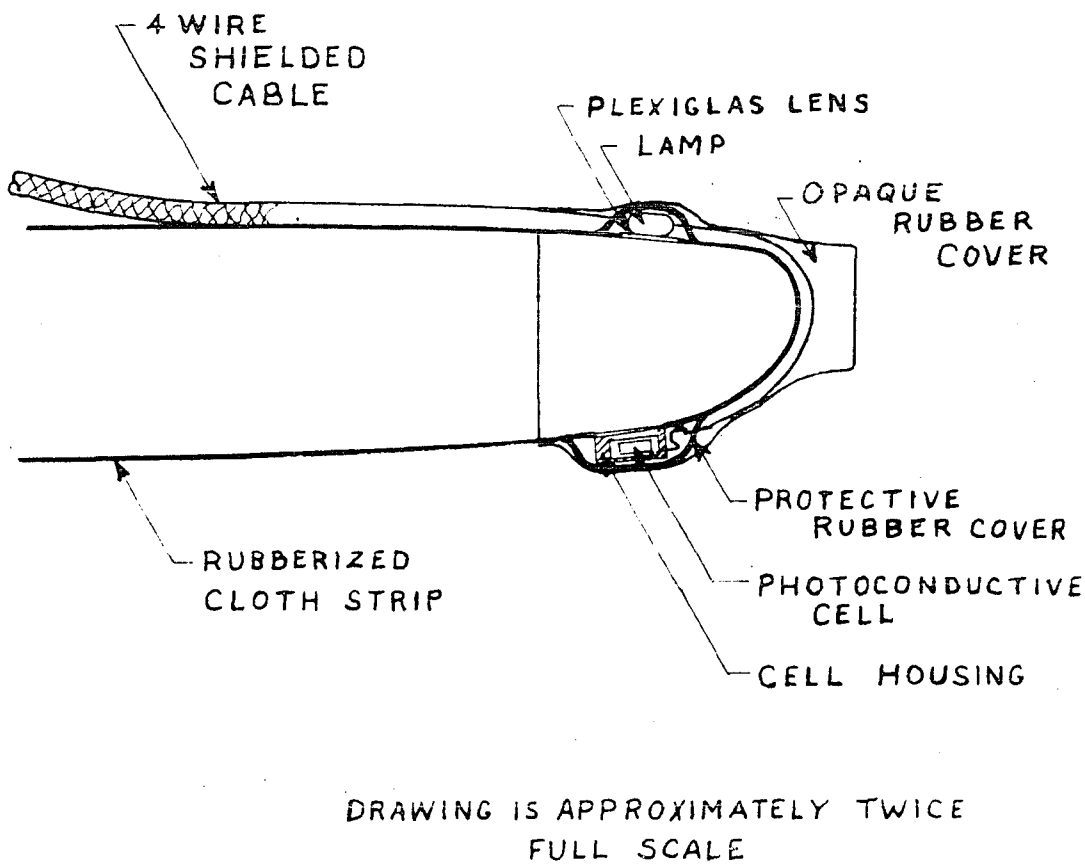
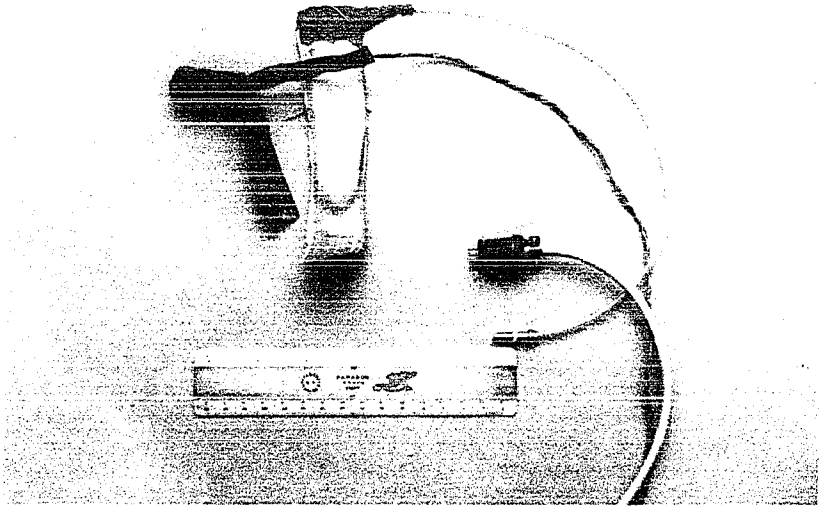


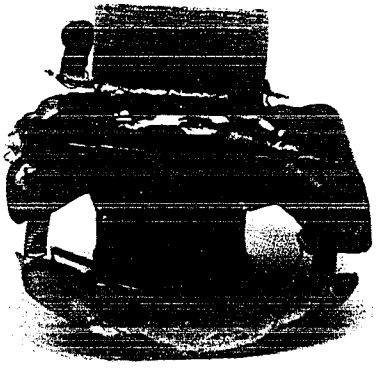
Figure 10. Sectional drawing of the photoconductive pulse detector unit of the variable pulse sensor

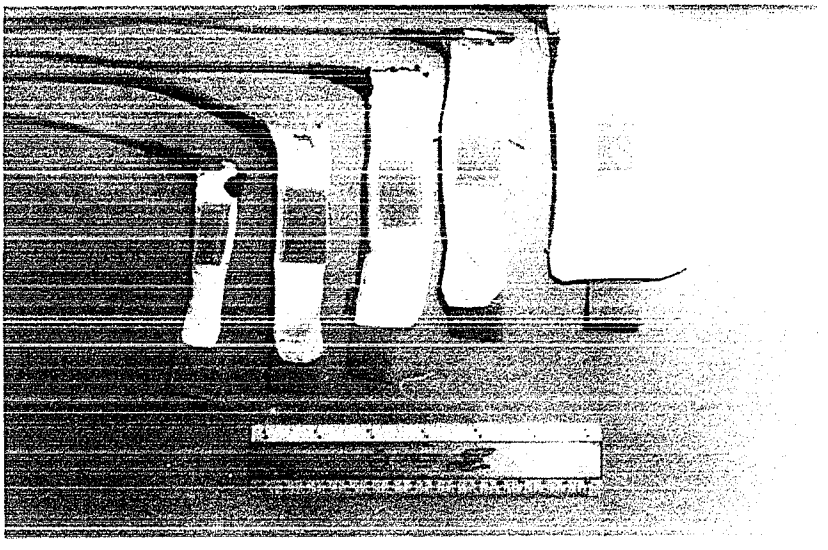
Figure 11. Variable pulse sensor unit

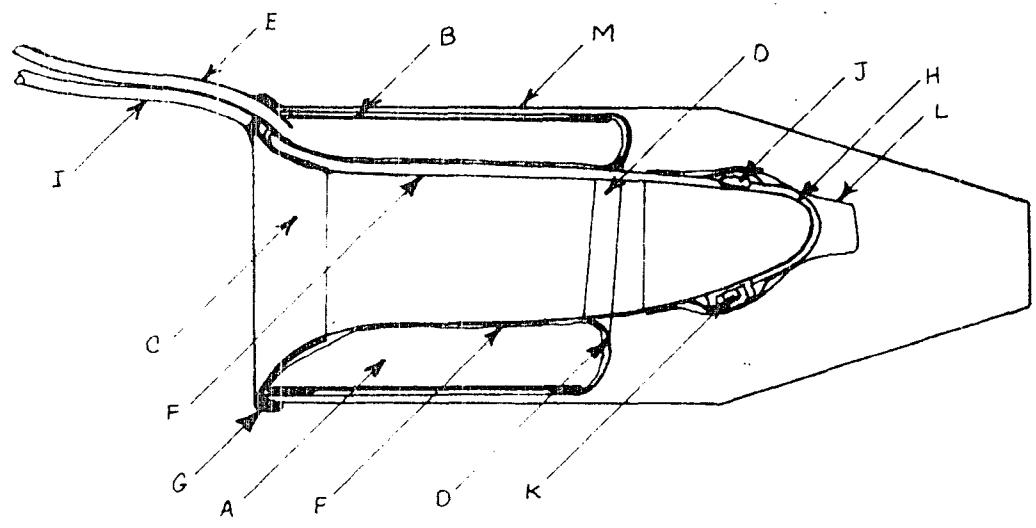
Figure 12. Photoconductive pulse detector unit of the cylindrical fixed pulse sensor

Figure 13. Digital occlusion cuffs for use with the variable pulse sensor







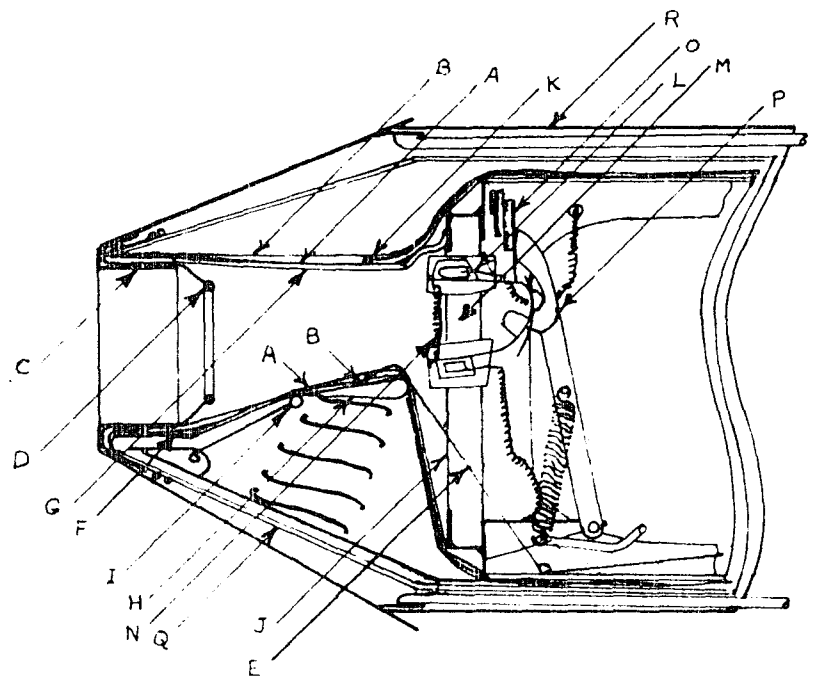


A. Occlusion Cuff
B. Pressure Ring
C. Front Retainer Ring
D. Rear Retainer Ring
E. Air Hose
F. Pulse Detector Strips
G. Rubber Aperature Ring
H. Blind End
I. Shielded 4 wire Cable
J. Lamp
K. Photoconductive Cell
L. Rubber Ring
M. Outer Mounting

Figure 14. Sectional drawing of the portable fixed-cuff pulse sensor unit

Figure 15. Portable pulse sensor unit

Figure 16. Cylindrical pulse sensor unit



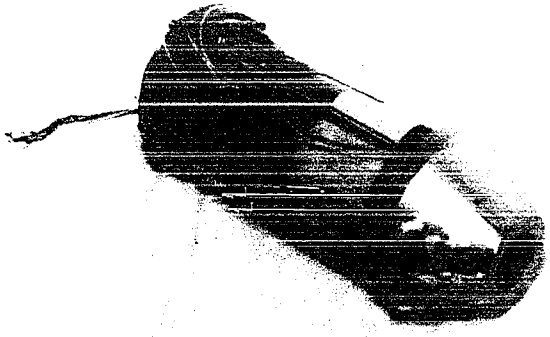
A PROPHYLACTIC
B PROPHYLACTIC
C SENSOR APERATURE
D RUBBER RING
E RUBBERIZED CLOTH
F RETAINER RING
G GUIDE
H FINGER RAMP
I HINGE
J POSITIONER HOUSING
K RUBBER RETAINER RING
L PULSE DETECTOR
M DETECTOR OPENING
N DETECTOR SPRING
O VELCRO FASTENER
P PLUNGER
Q PRESSURE CONTAINER
R OUTER COVERING

Figure 17. Sectional drawing of the cylindrical pulse sensor unit

Figure 18. Cylindrical pulse sensor with the center section of the cover removed

Figure 19. Cylindrical pulse sensor with the pressure container removed

Figure 20. Cylindrical pulse sensor with the position housing removed



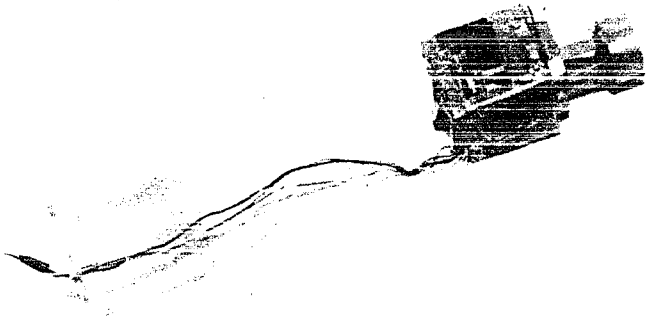
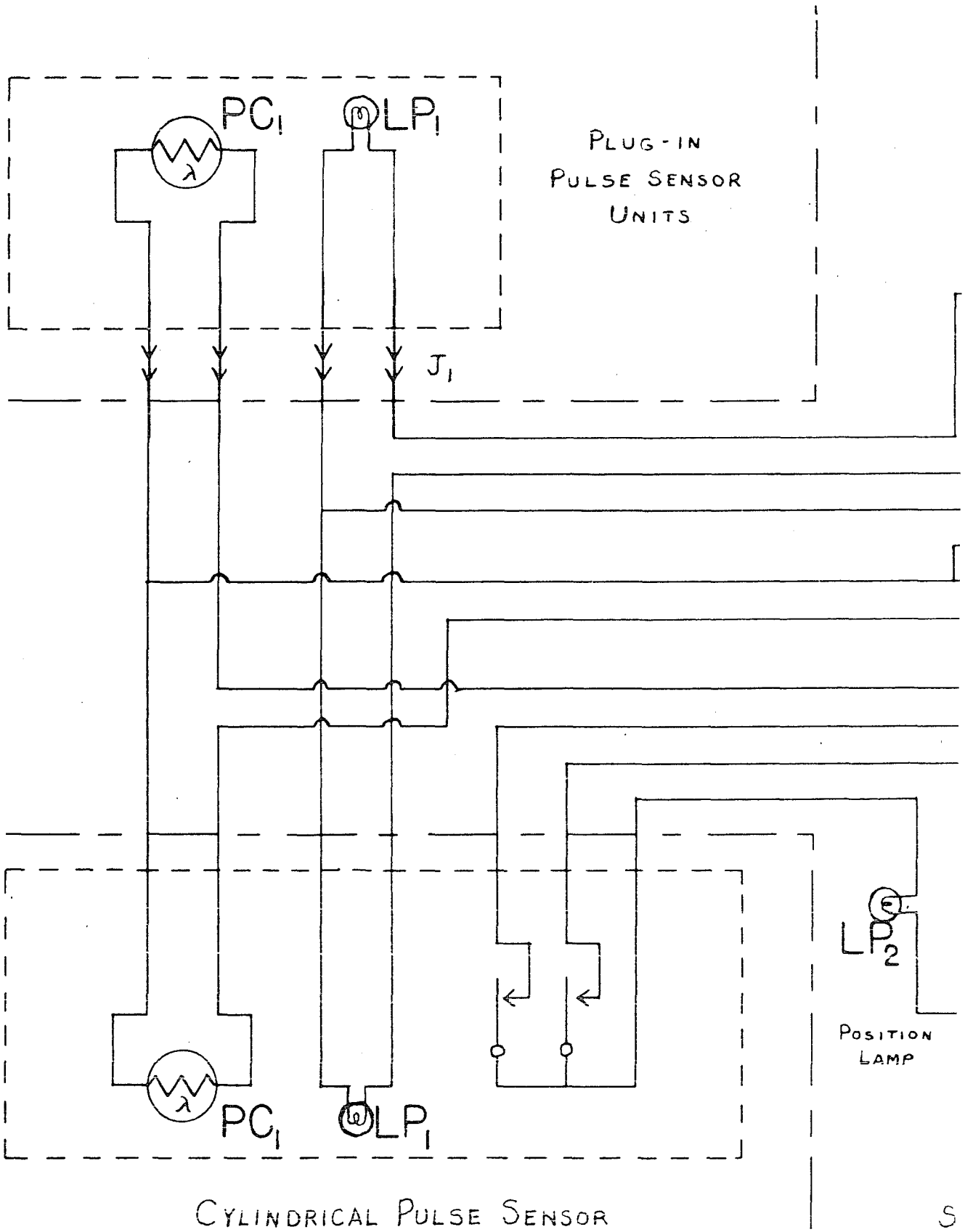
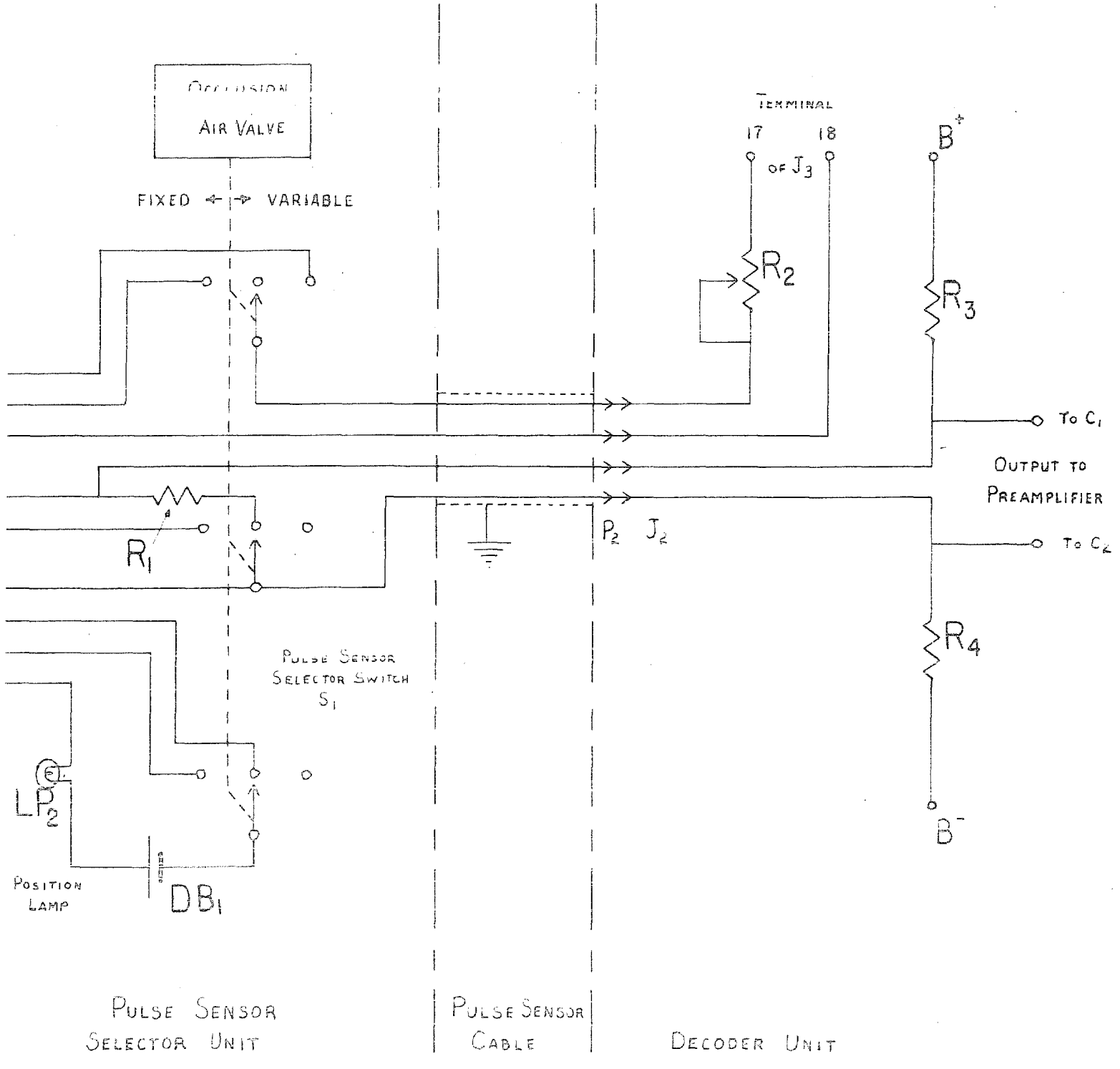


Figure 21. Schematic diagram of pulse sensor selector unit and associated circuitry



CYLINDRICAL PULSE SENSOR


Occlusion
Air Valve
FIXED
VARIABLE
Terminal
17
18
of J3
B+
R2
R3
To C1
Output to
Preamplifier
To C2
R1
P2
J2
R4
Pulse Sensor
Selector Switch
S1
LP2
Position
Lamp
DB1
B-
Pulse Sensor
Selector Unit
Pulse Sensor
Cable
Decoder Unit

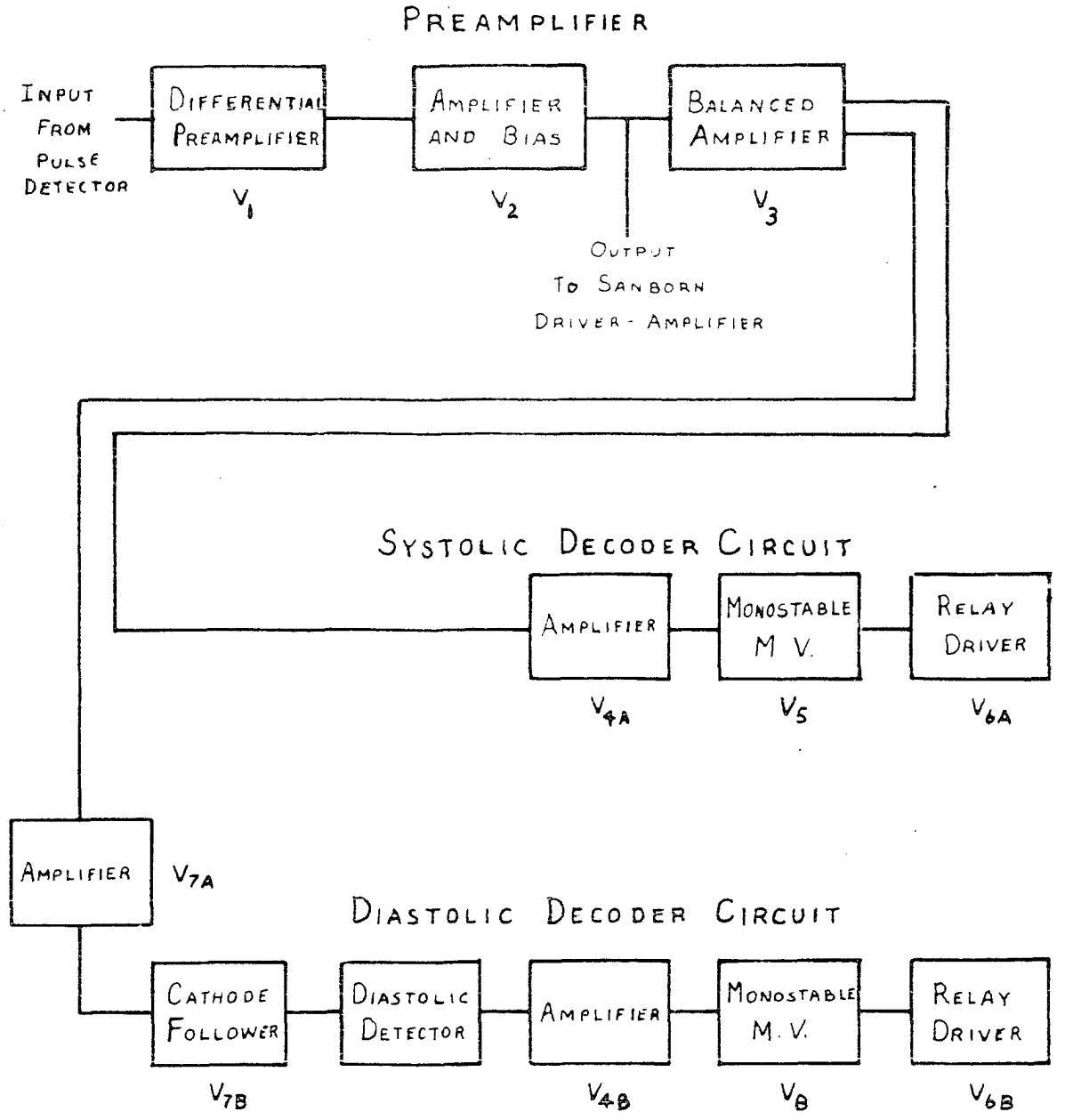
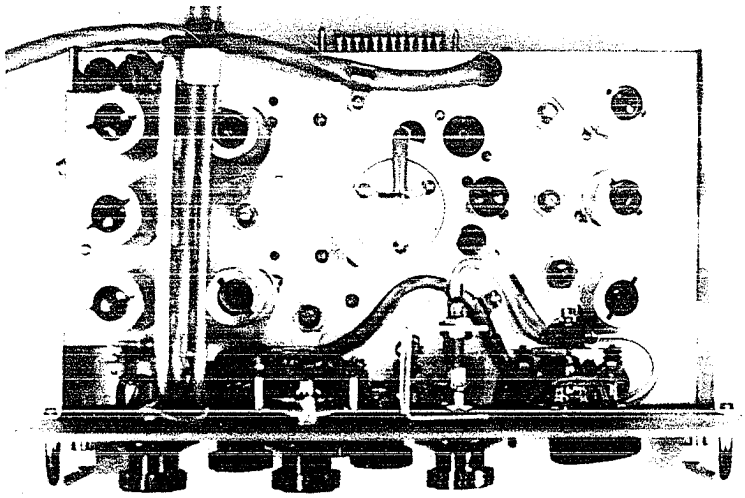


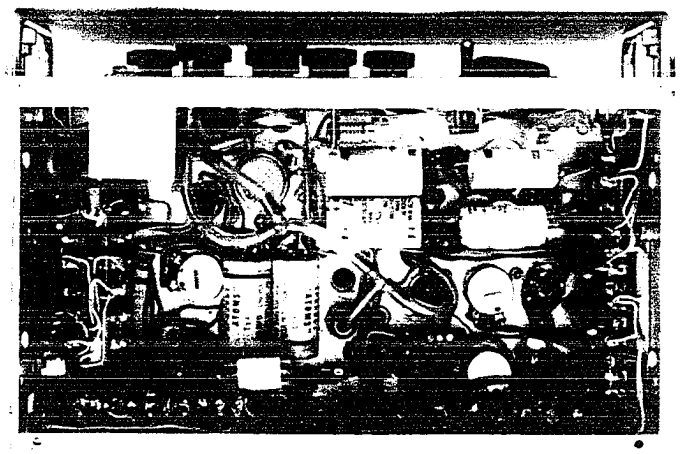
Figure 22. Block diagram of the decoder unit

Figure 23. Top view of decoder unit

Figure 24. Bottom view of decoder unit (with bottom shield removed)

Figure 25. Operating controls of the blood pressure measuring instrument





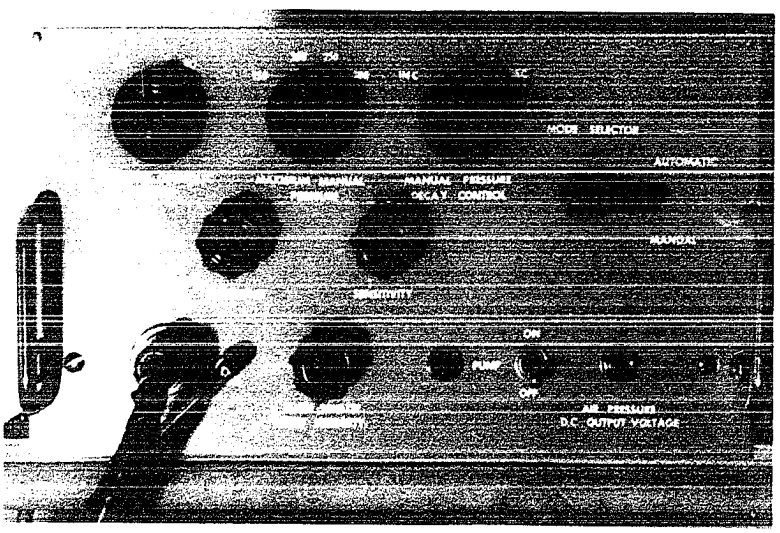
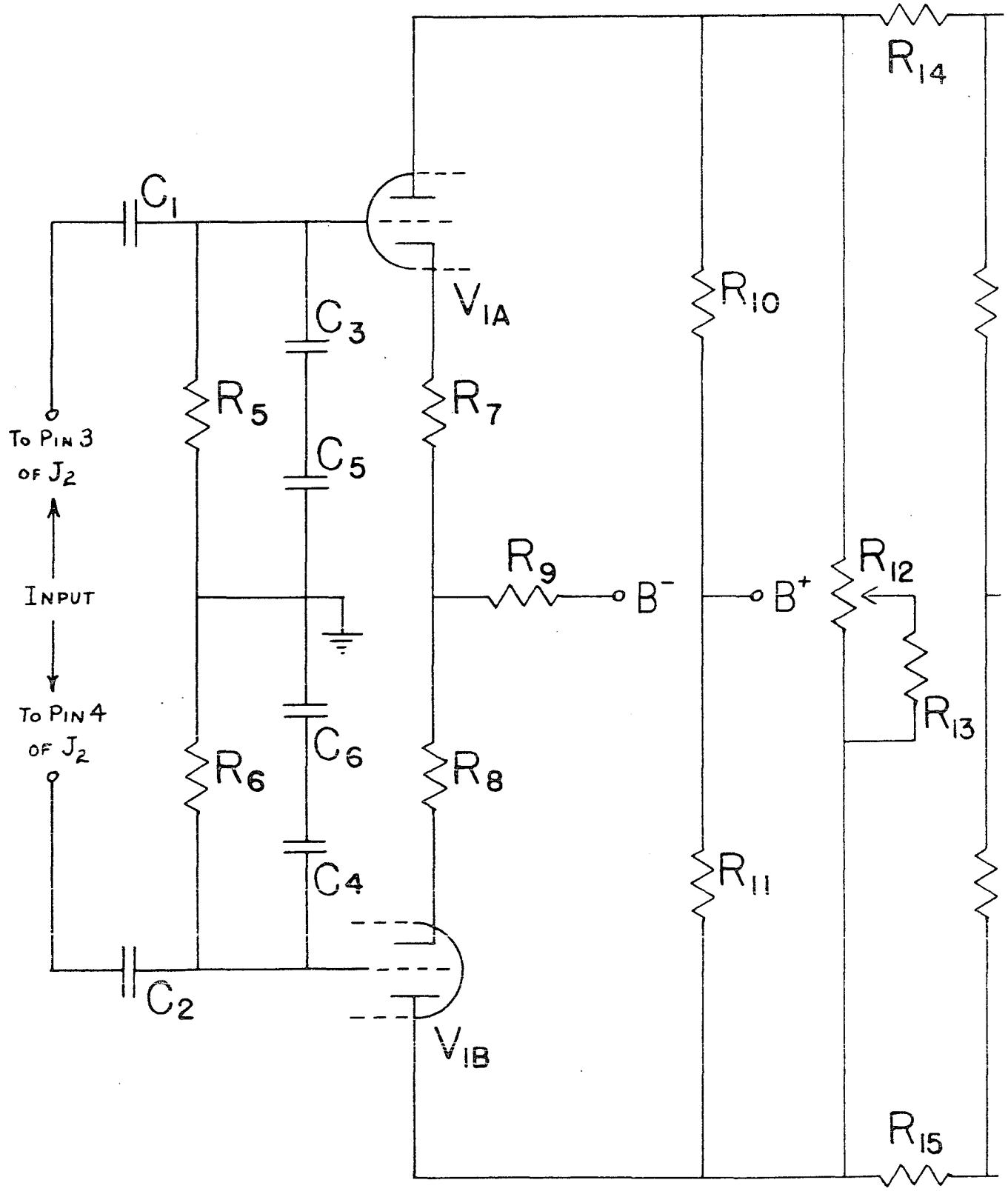
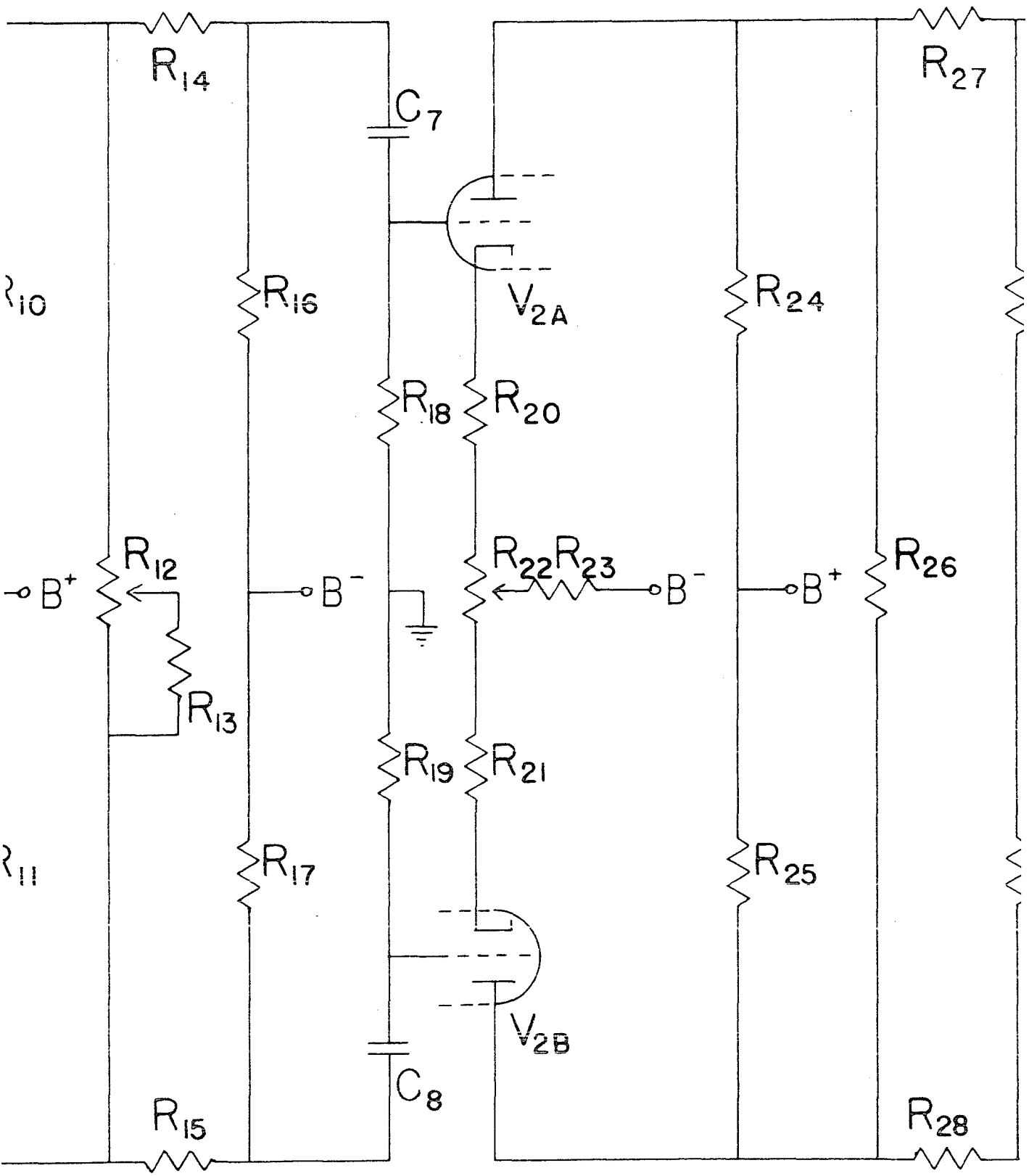

MODE SELECTOR
AUTOMATIC
MANUAL
ON
OFF
AIR PRESSURE
D.C. OUTPUT VOLTAGE

Figure 26. Schematic diagram of the differential preamplifier


C1
C2
C3
C4
C5
C6
R5
R6
R7
R8
R9
R10
R11
R12
R13
R14
R15
V1A
V1B
B-
B+
INPUT
TO PIN 3 OF J2
TO PIN 4 OF J2


R14
C7
R27
R10
R16
V2A
R24
R18
R20
B+
R12
B-
R22
R23
B-
B+
R26
R13
R19
R21
R11
R17
R25
V2B
C8
R15
R28

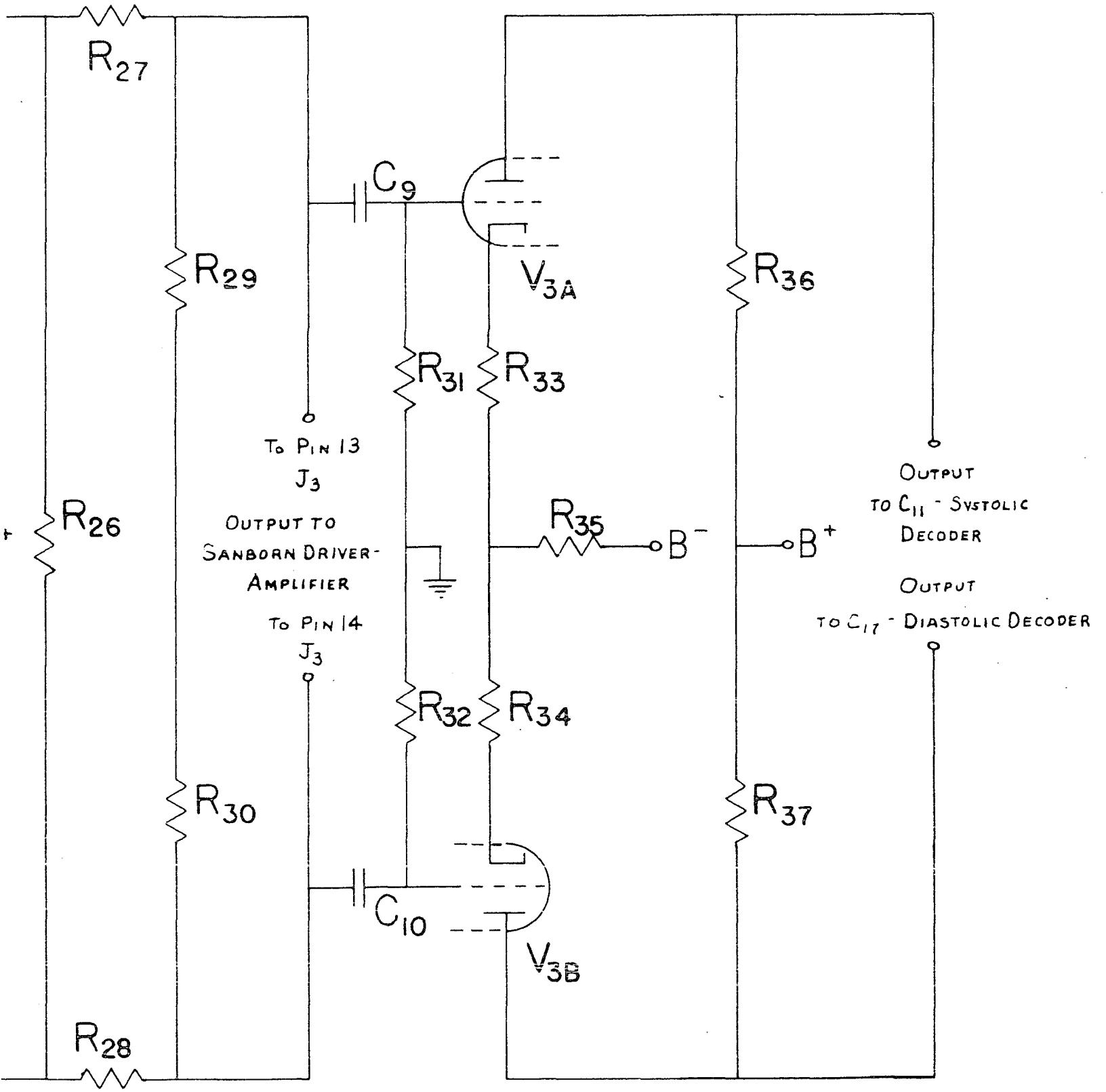

R27
R29
R26
R30
R28
C9
C10
V3A
V3B
R31
R33
R32
R34
R35
R36
R37
B-
B+
To Pin 13
J3
Output to
Sanborn Driver-
Amplifier
To Pin 14
J3
Output
to C11 - Systolic
Decoder
Output
to C17 - Diastolic Decoder

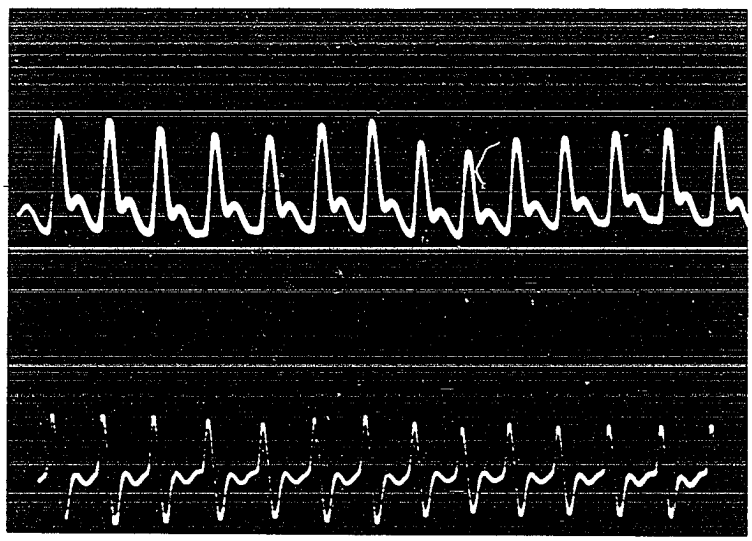
Figure 27a. Blood pulse waveform input to the preamplifier as obtained from the variable pulse detector

Figure 27b. Preamplifier pulse waveform output to the Sanborn driver-amplifier

Figure 27c. Preamplifier pulse waveform output to the systolic decoder circuit

Figure 27d. Preamplifier pulse waveform output to the diastolic decoder circuit

Figure 27. Preamplifier pulse waveforms



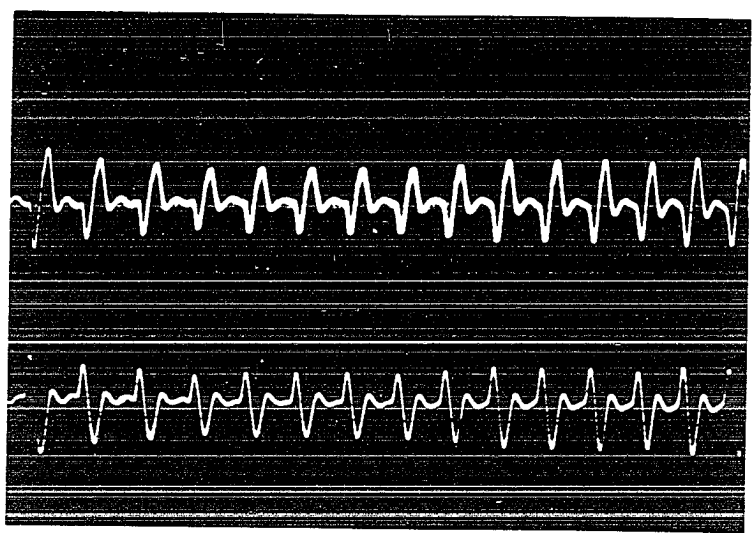
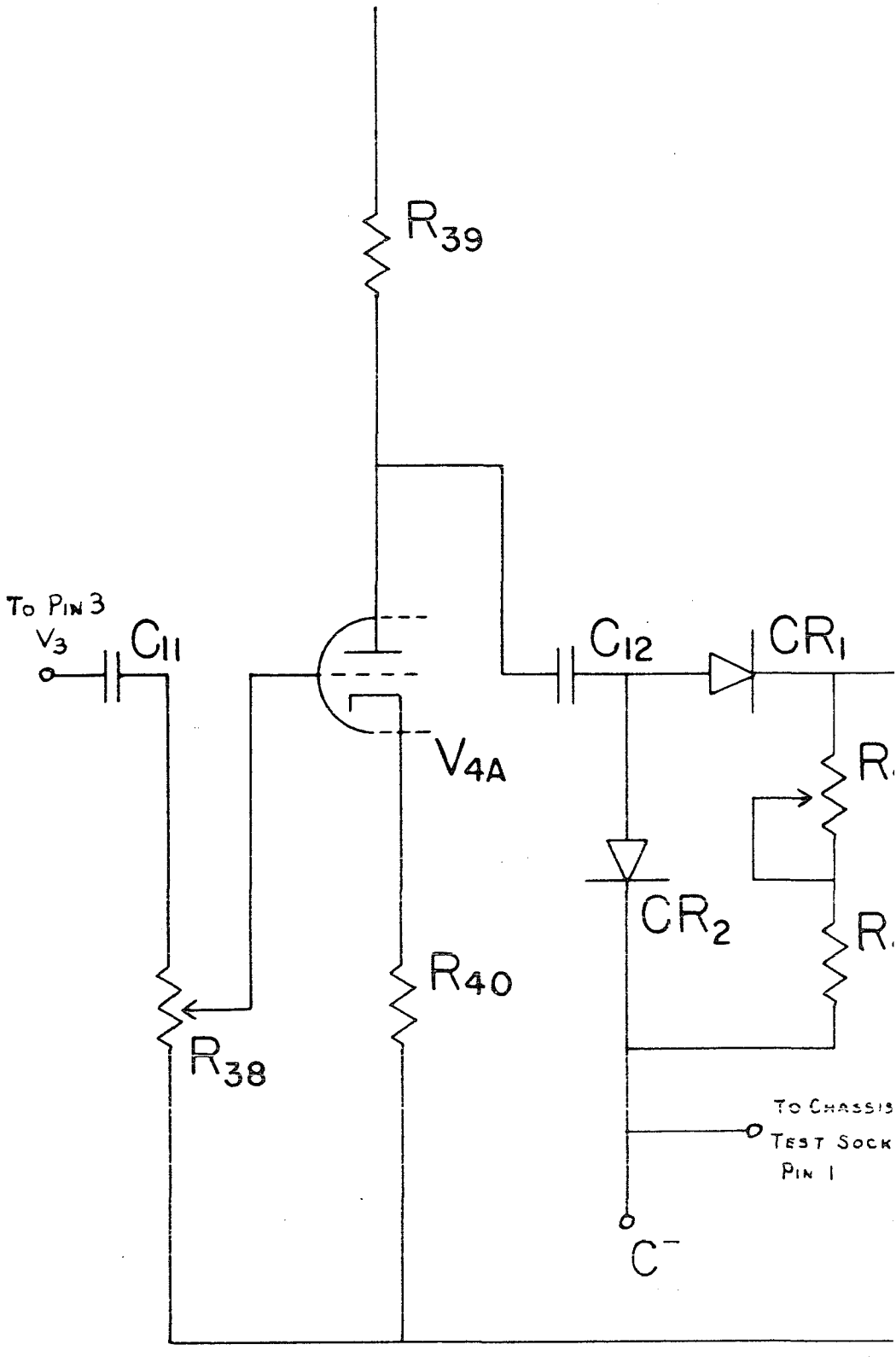
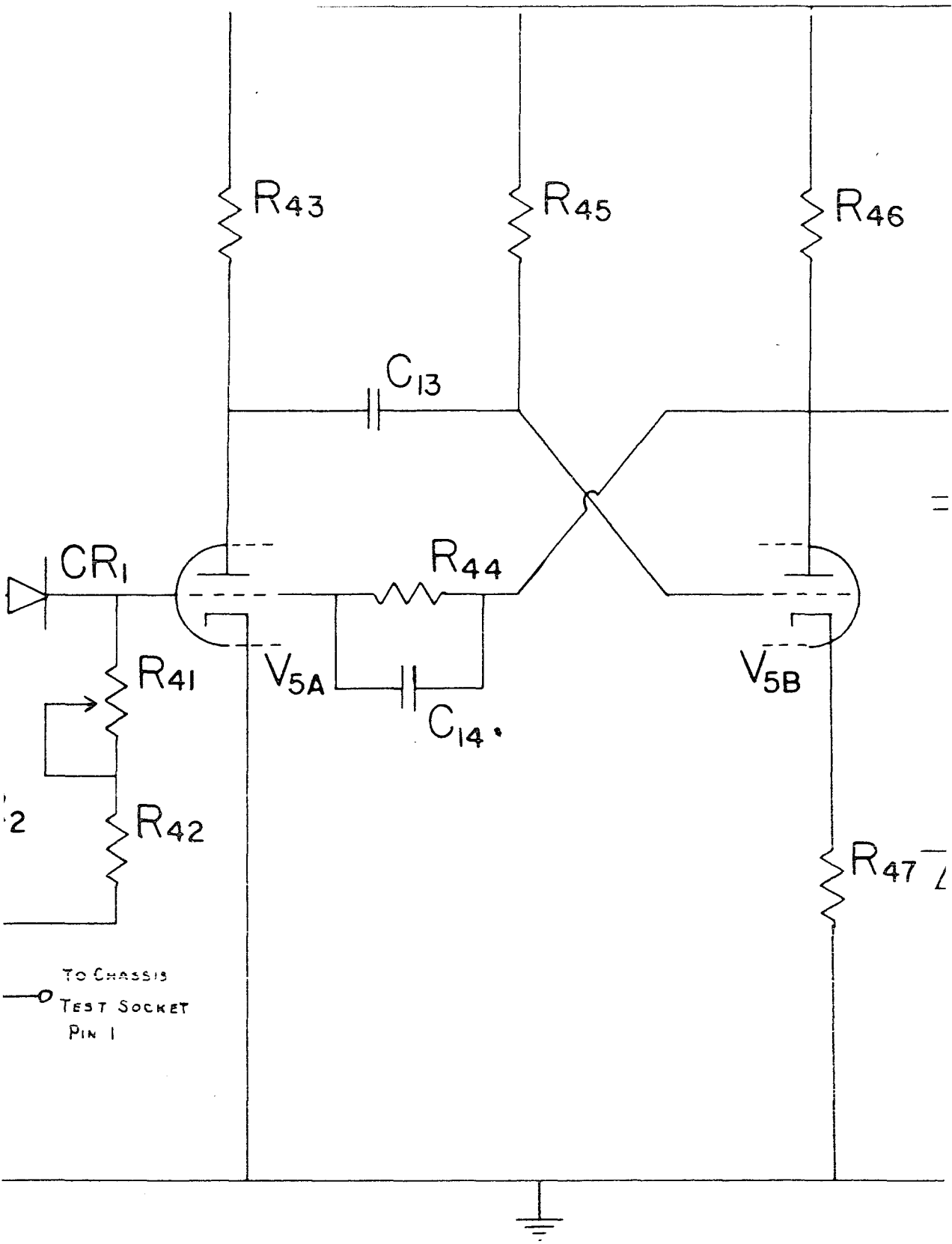


Figure 28. Schematic diagram of the systolic decoder circuit


R39
To Pin 3
V3
C11
V4A
C12
CR1
R
CR2
R
R40
R38
To Chassis
Test Sock
Pin 1
C⁻


R43
R45
R46
C13
CR1
R44
R41
V5A
V5B
C14
R42
R47
TO CHASSIS
TEST SOCKET
PIN 1

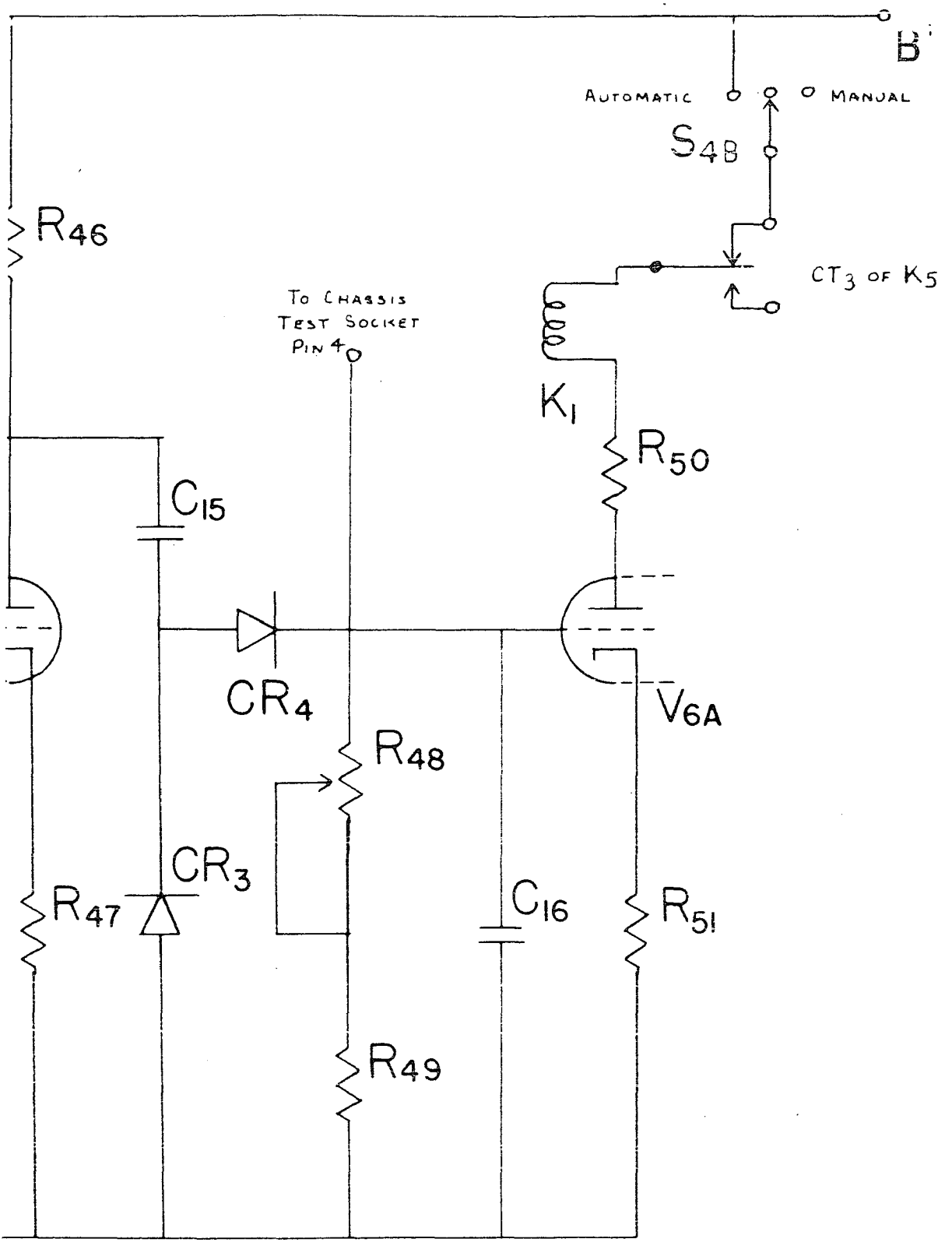
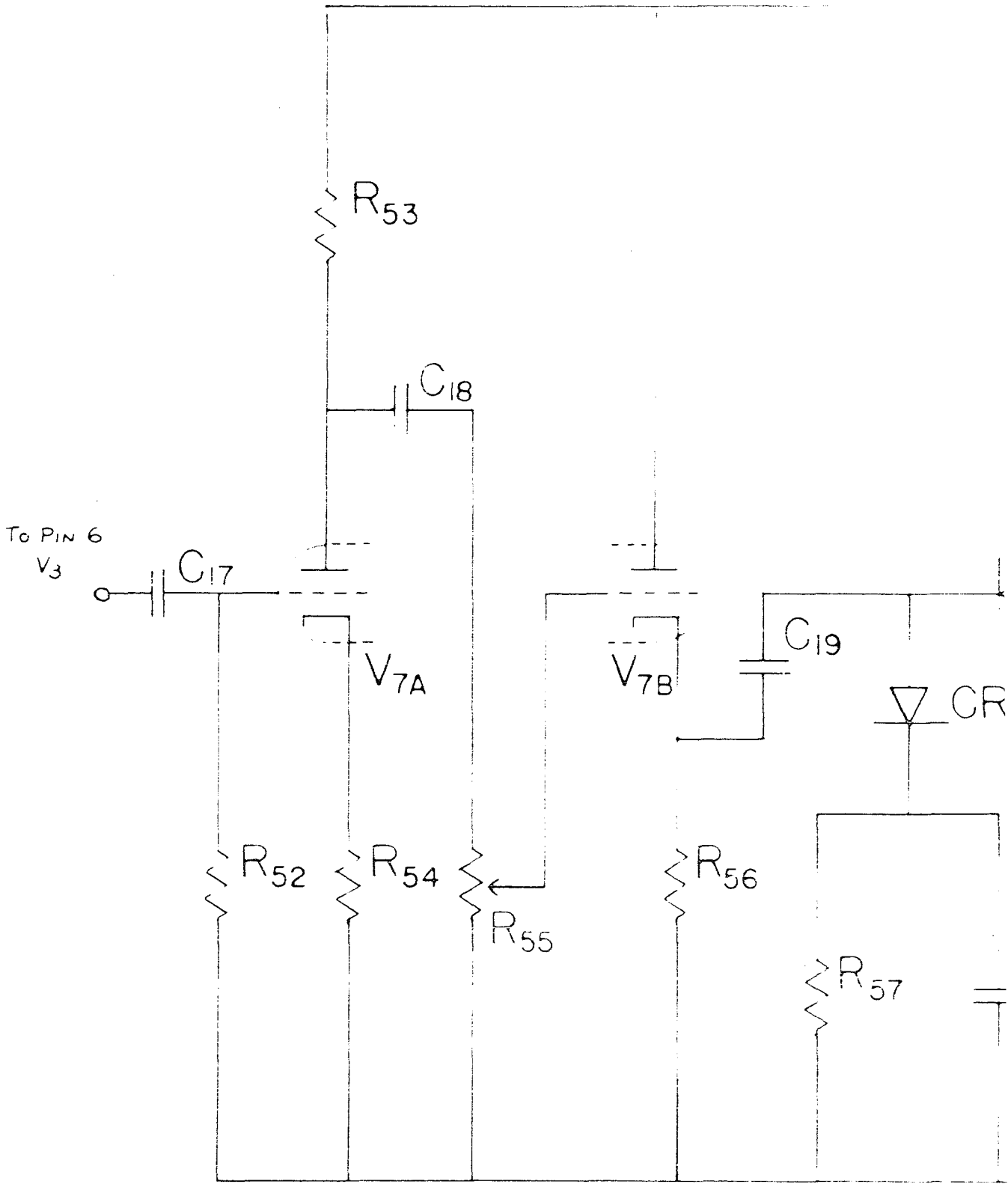
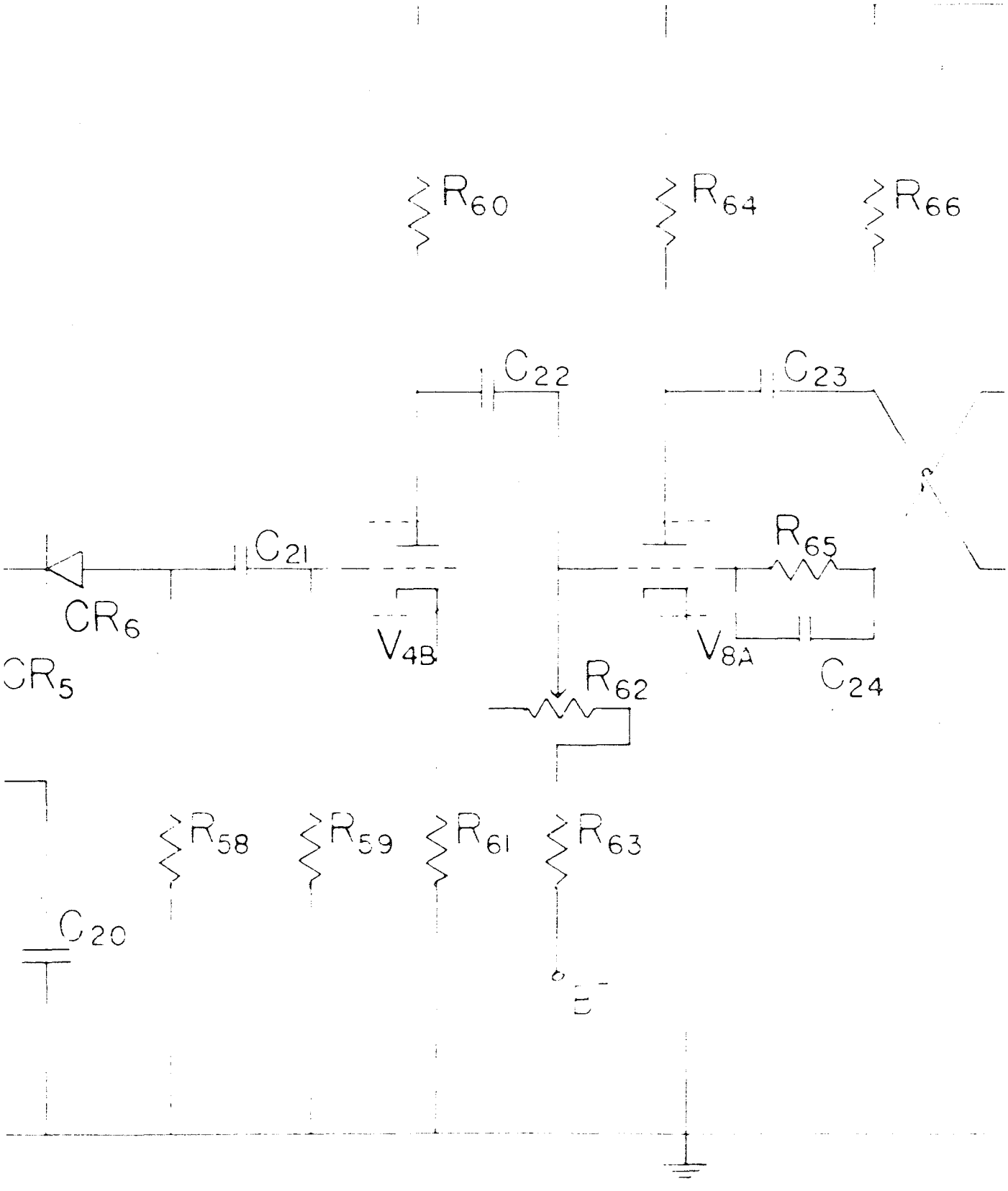

B+
AUTOMATIC
MANUAL
S4B
R46
CT3 OF K5
TO CHASSIS
TEST SOCKET
PIN 4
K1
R50
C15
CR4
V6A
R48
CR3
R47
C16
R51
R49

Figure 29. Schematic diagram of the diastolic decoder circuit


R53
C18
To Pin 6
V3
C17
V7A
V7B
C19
CR
R52
R54
R55
R56
R57


R60
R64
R66
C22
C23
C21
R65
CR6
V4B
V8A
C24
CR5
R62
R58
R59
R61
R63
C20

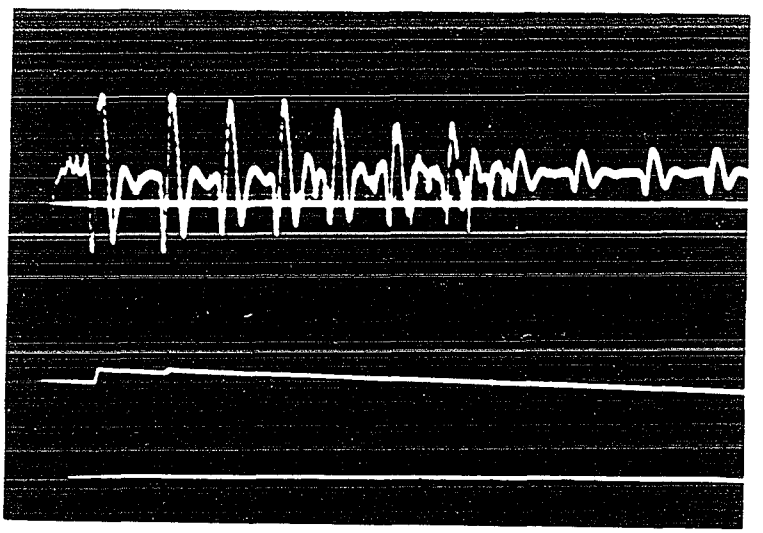
DC +
AUTOMATIC
MANUAL
S4B
CT3 OF K5
K2
R71
R67
C25
V6B
V8B
CR8
R69
CR7
R68
C26
R72
R70

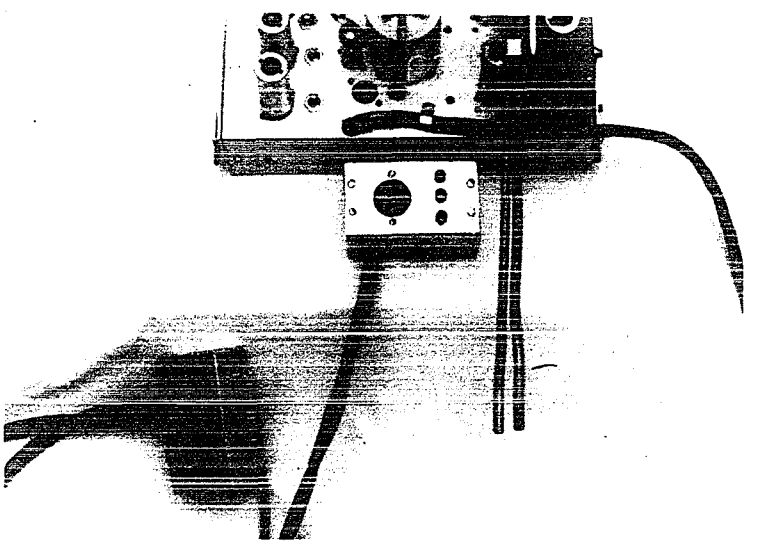
Figure 30a. Input waveform to the diastolic detector circuit

Figure 30b. Storage waveform of the diastolic detector circuit

Figure 30. Pulse waveforms of the diastolic detector circuit

Figure 31. Maintenance connecting cable





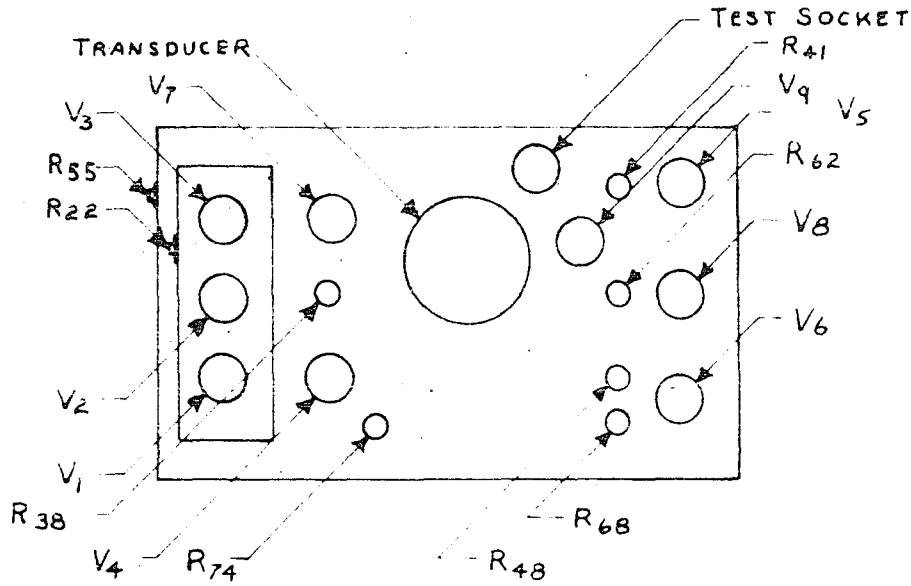


Figure 32. Schematic diagram of the decoder unit maintenance adjustments

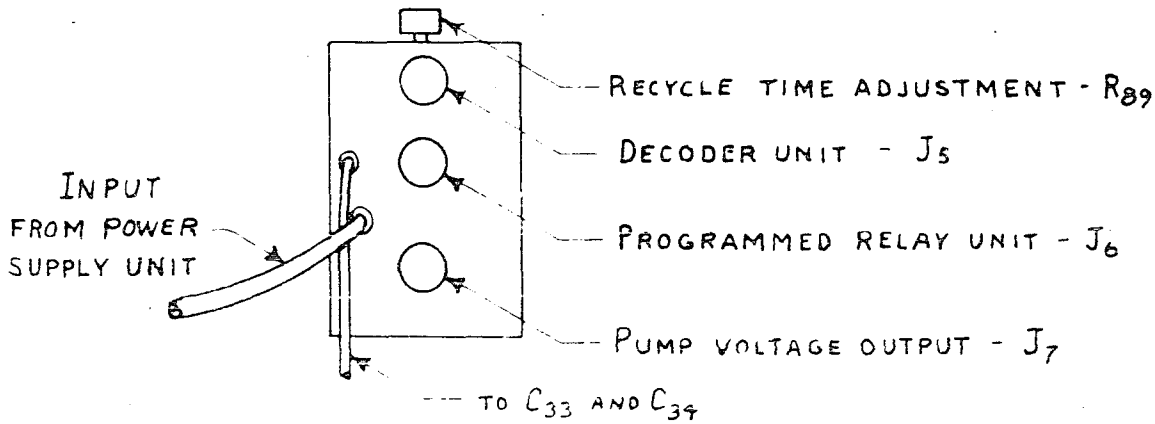


Figure 33. Schematic diagram of the interconnections to the relay control unit

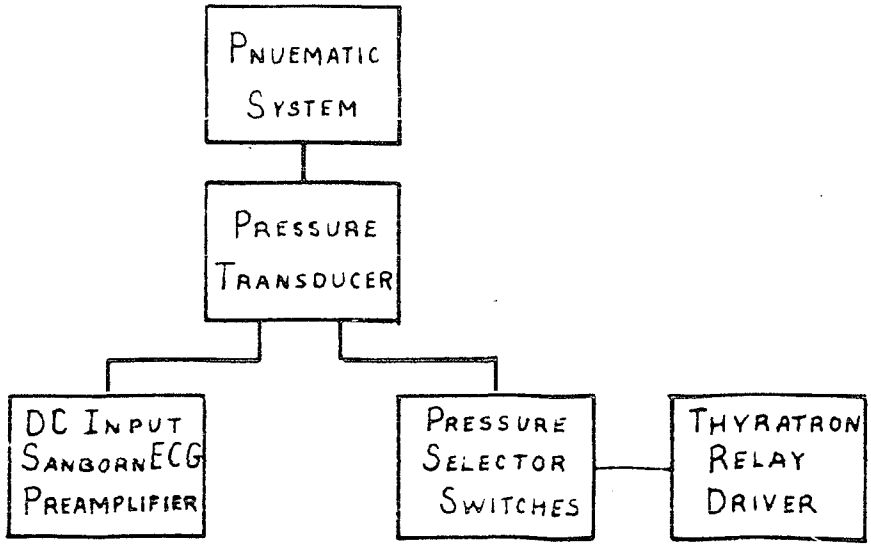


Figure 34. Block diagram of the pressure transducer circuit

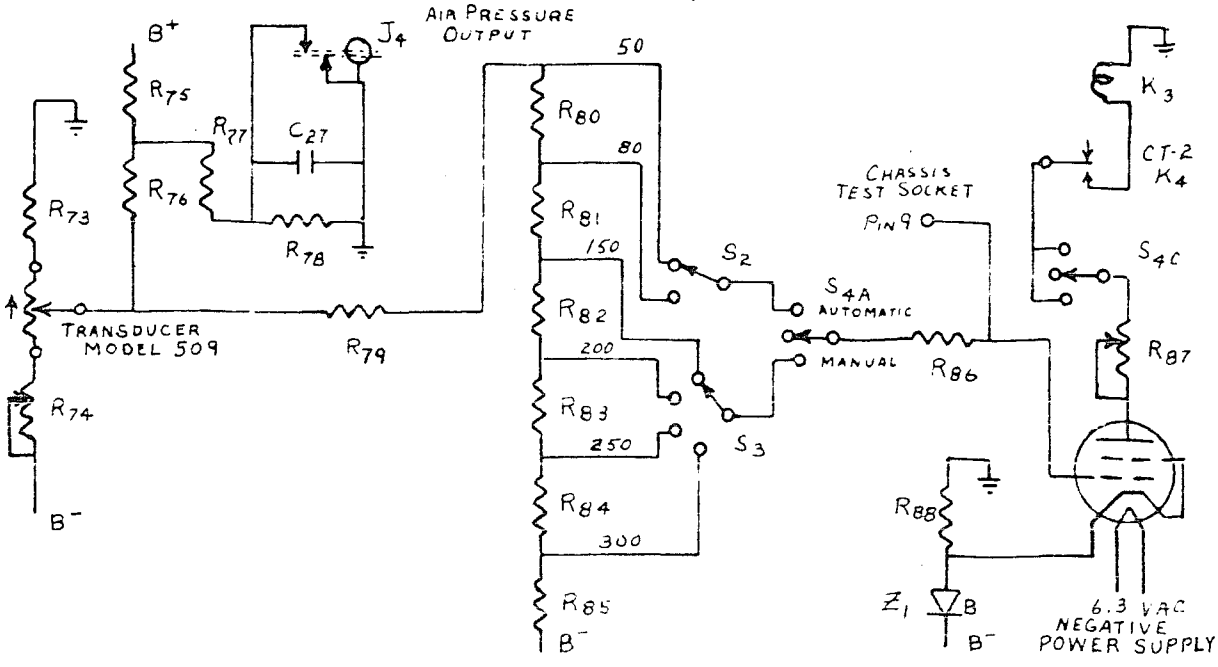
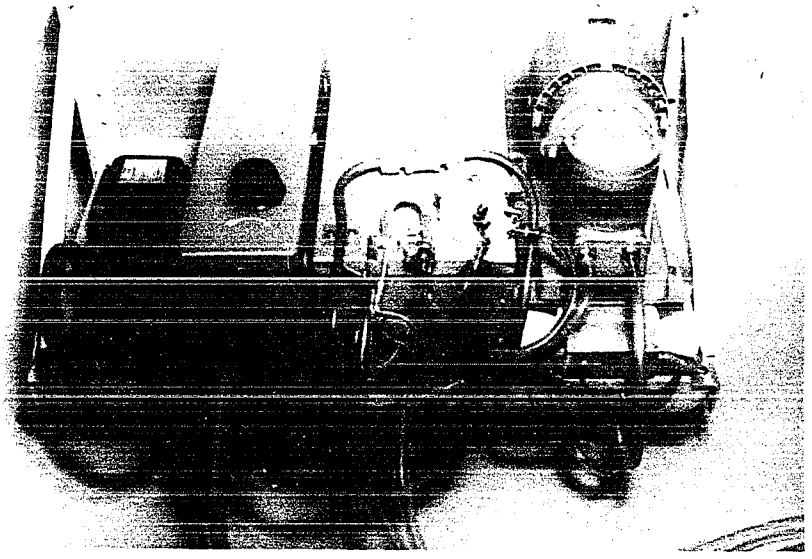


Figure 35. Schematic diagram of the pressure transducer circuit

Figure 36. Supplementary chassis assembly

Figure 37. Control relay unit (with side cover removed)



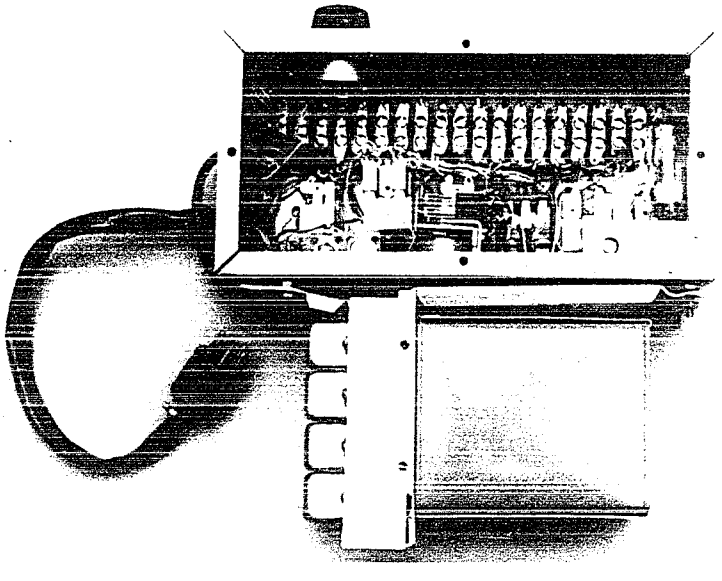
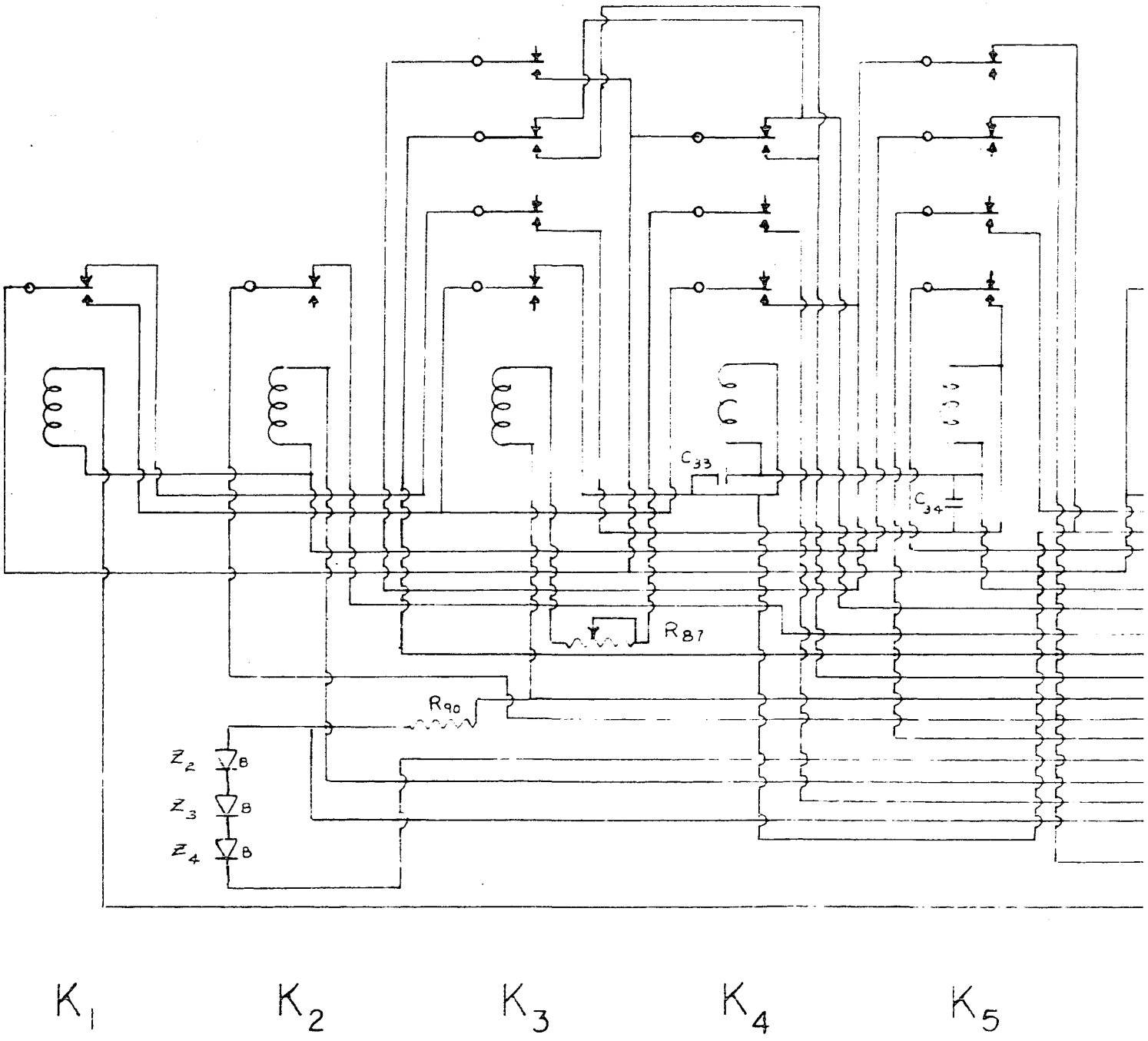
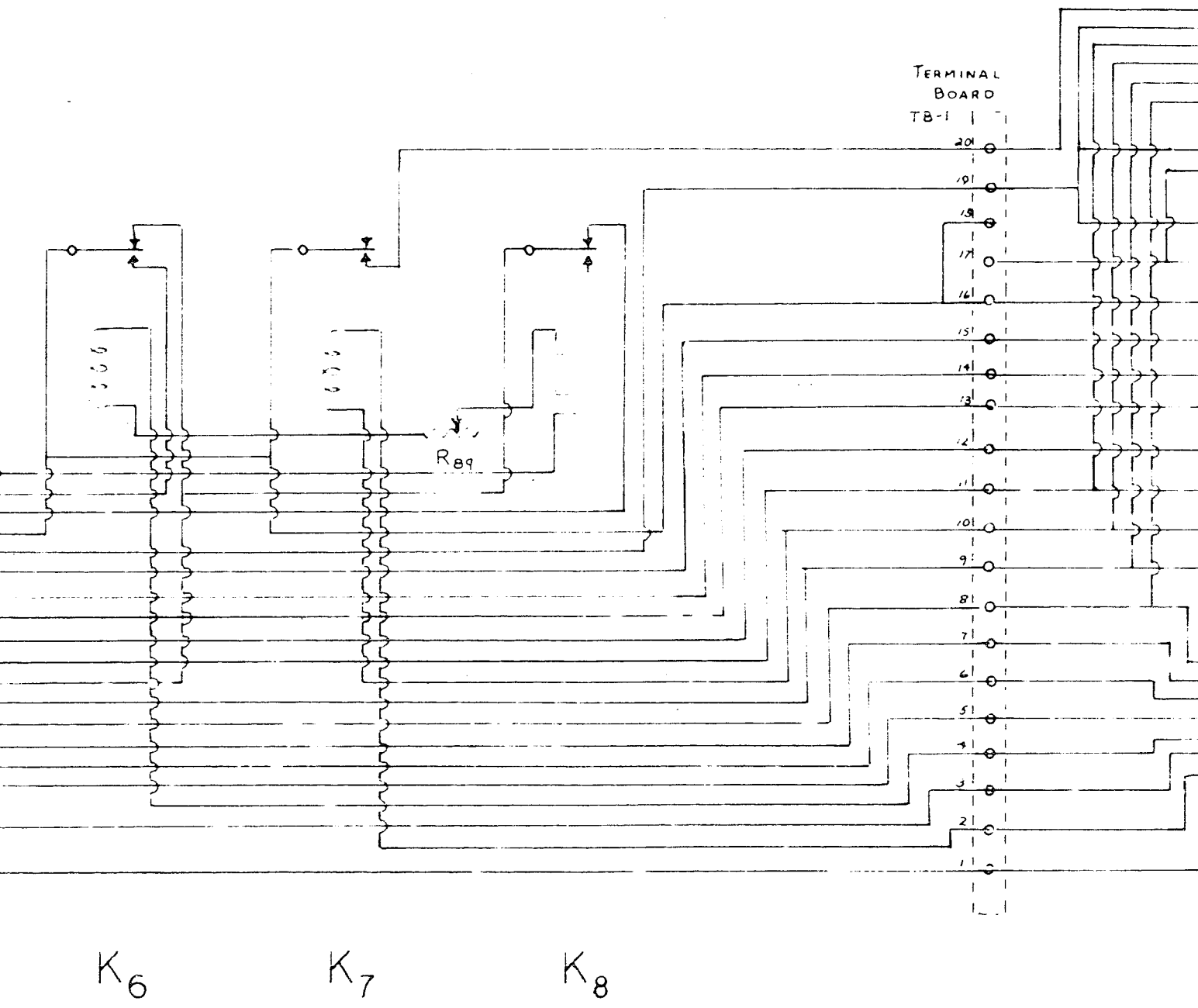
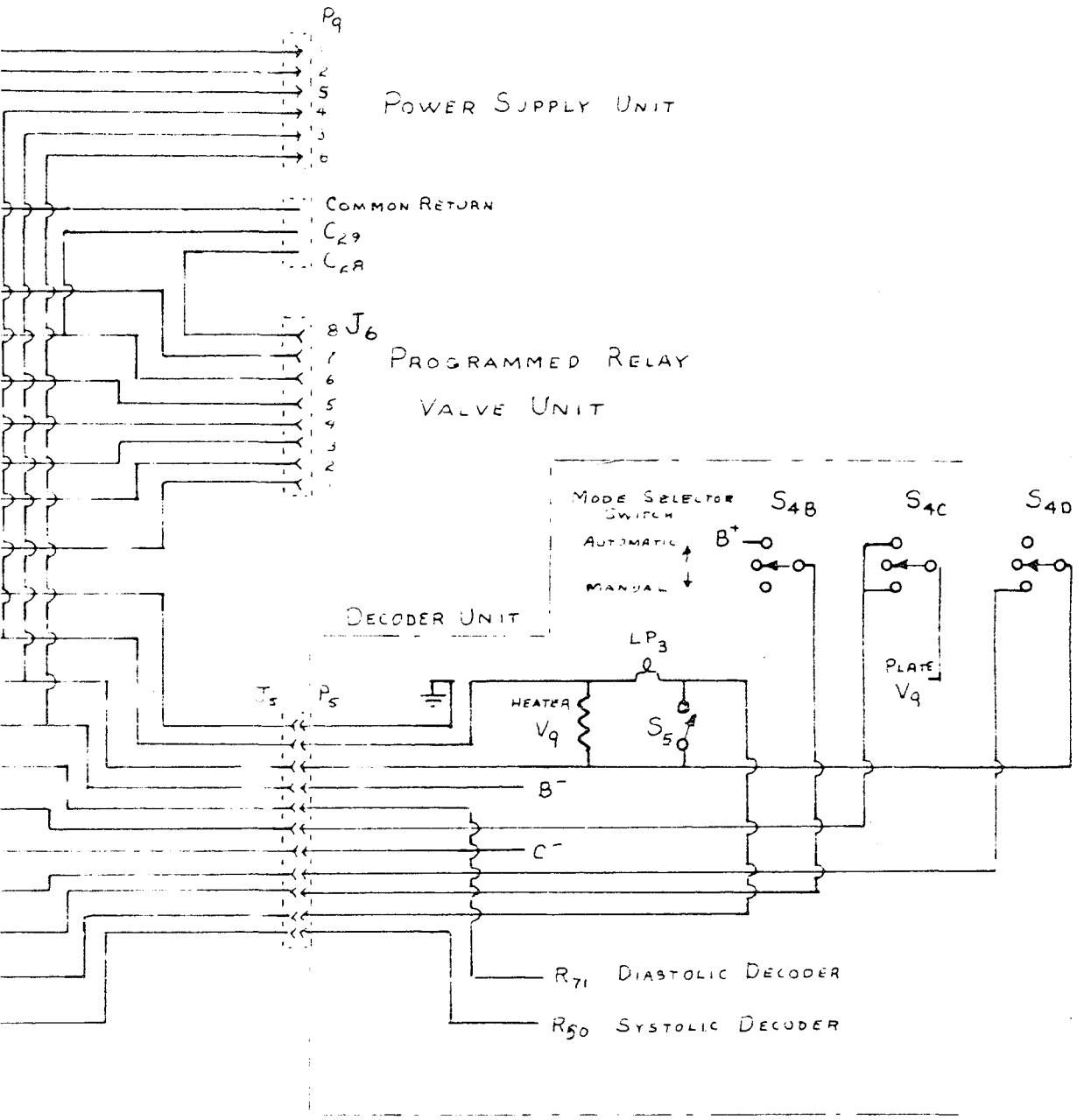


Figure 38. Schematic diagram of control relay unit


C33
C34
R87
R90
Z2 8
Z3 8
Z4 8
K1
K2
K3
K4
K5


TERMINAL
BOARD
TB-1
20
19
18
17
16
15
14
13
12
11
10
9
8
7
6
5
4
3
2
1
R89
K6
K7
K8


P9
2
5
4
3
6
POWER SUPPLY UNIT
COMMON RETURN
C29
C28
8 J6
1
6
5
4
3
2
PROGRAMMED RELAY
VALVE UNIT
MODE SELECTOR SWITCH
AUTOMATIC
MANUAL
S4B
S4C
S4D
B+
DECODER UNIT
LP3
PLATE
V9
J5
P5
HEATER
V9
S5
B-
C-
R71 DIASTOLIC DECODER
R50 SYSTOLIC DECODER

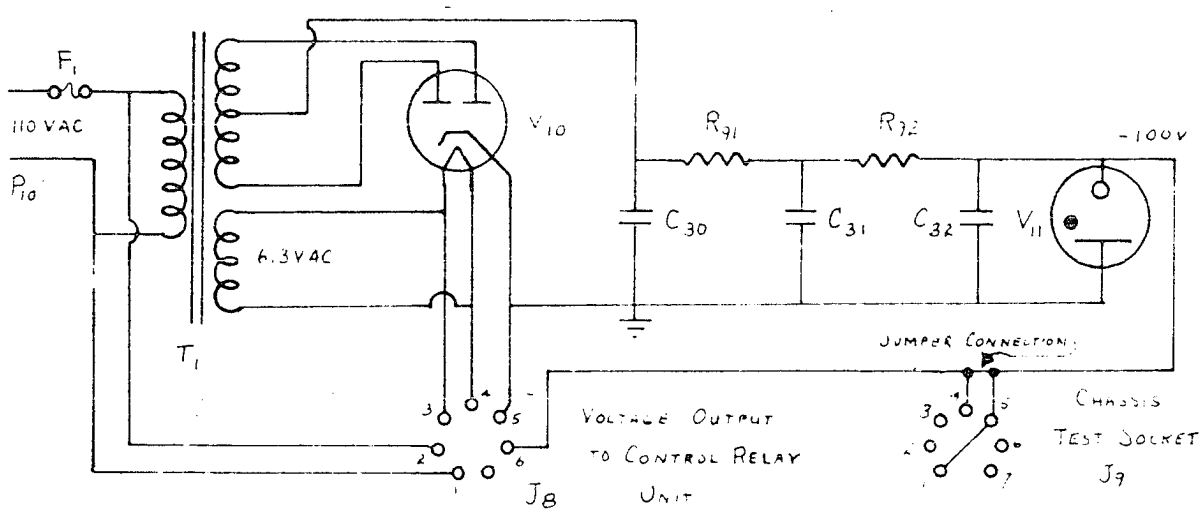


Figure 39. Schematic diagram of the power supply unit

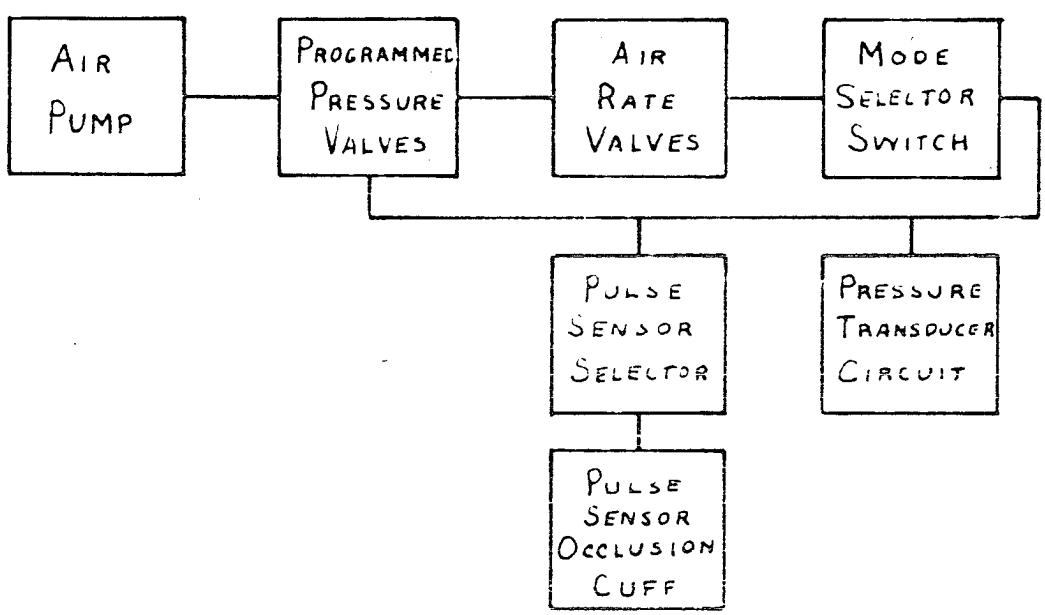
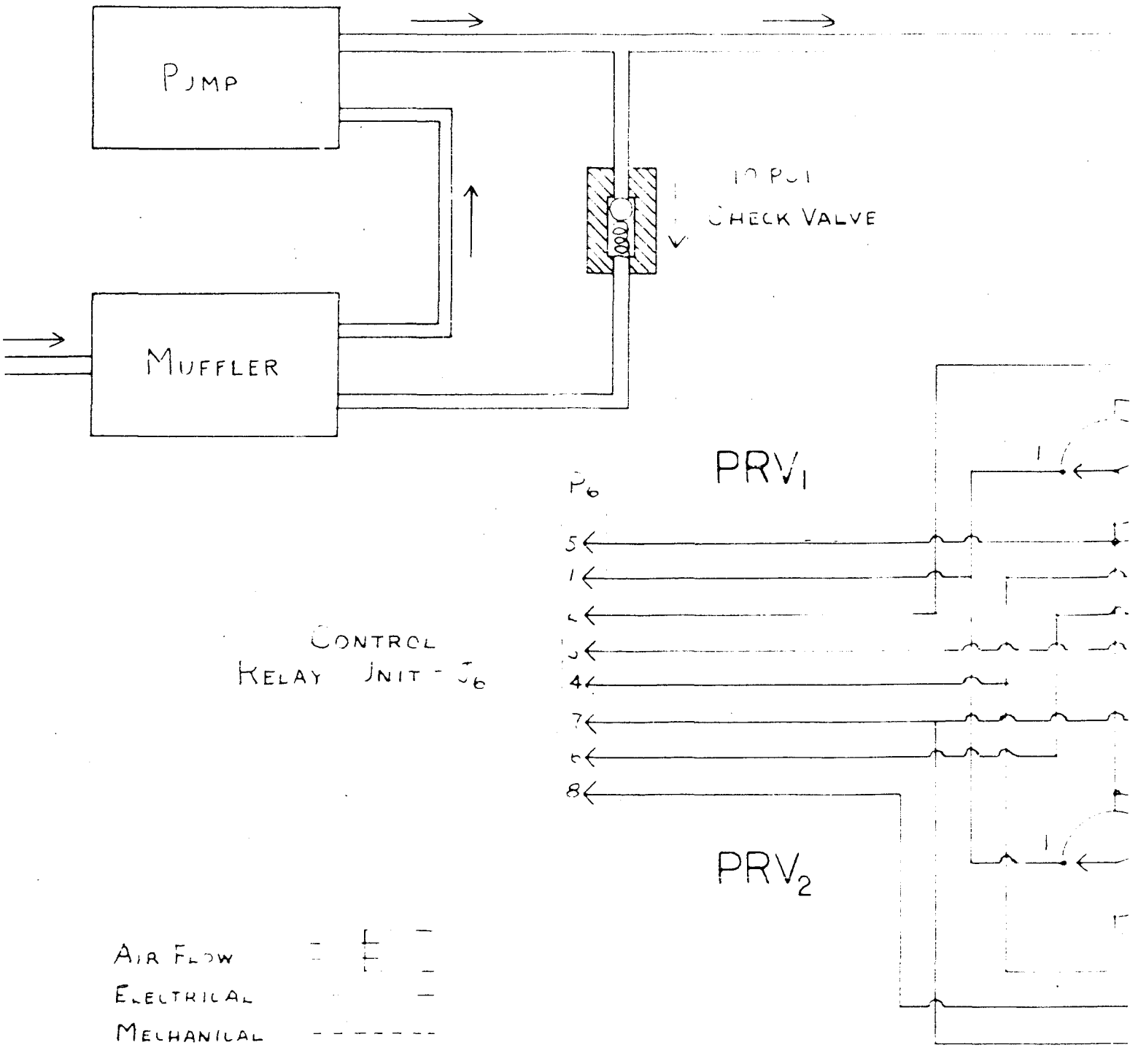
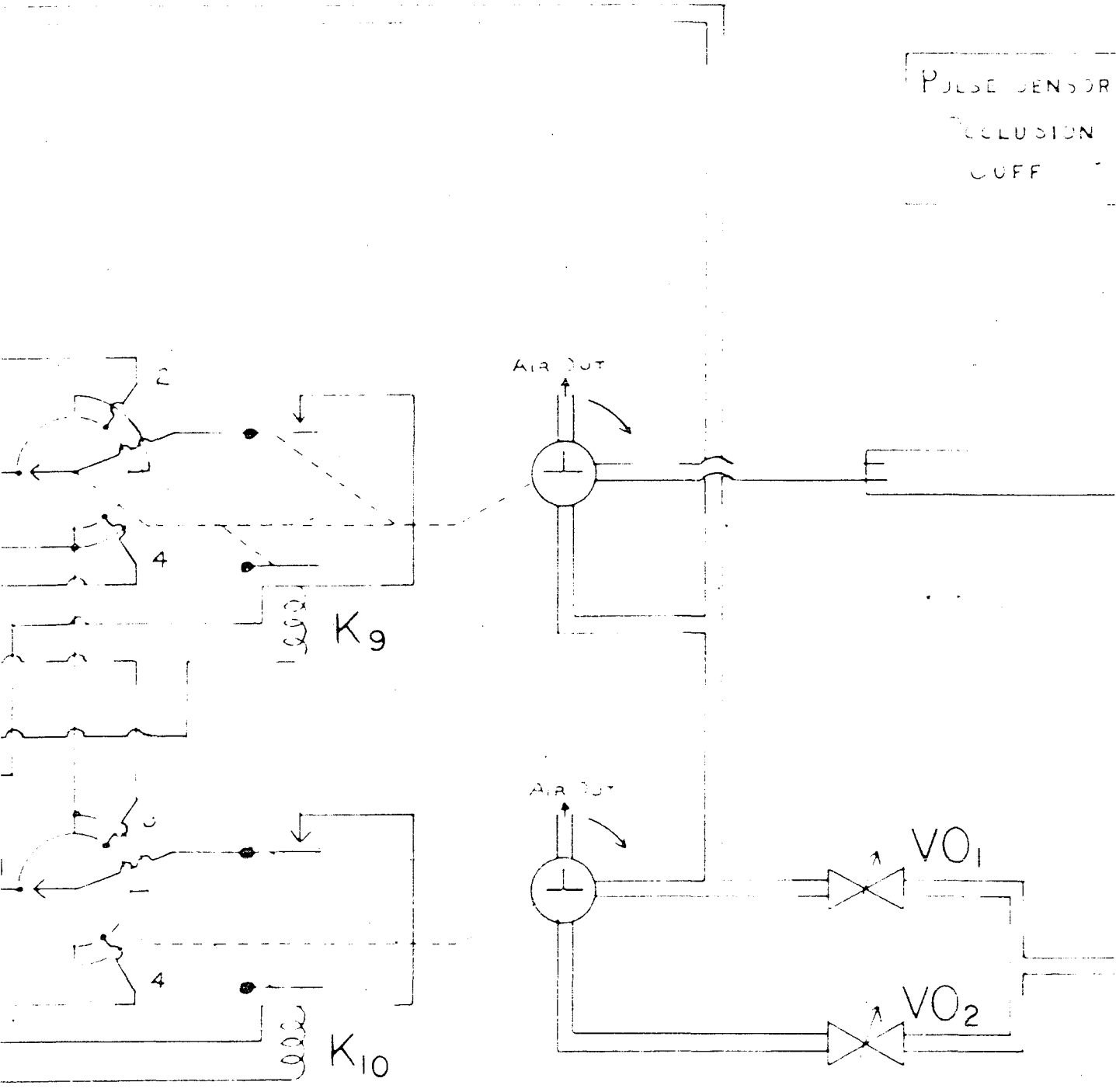


Figure 40. Block diagram of the pneumatic system

Figure 41. Schematic diagram of the pneumatic system


PUMP
CHECK VALVE
MUFFLER
PRV1
P6
5
1
4
7
8
PRV2
CONTROL
RELAY UNIT - J6
AIR FLOW
ELECTRICAL
MECHANICAL


PULSE SENSOR
OCCLUSION
CUFF
AIR OUT
AIR OUT
4
4
K9
K10
VO1
VO2

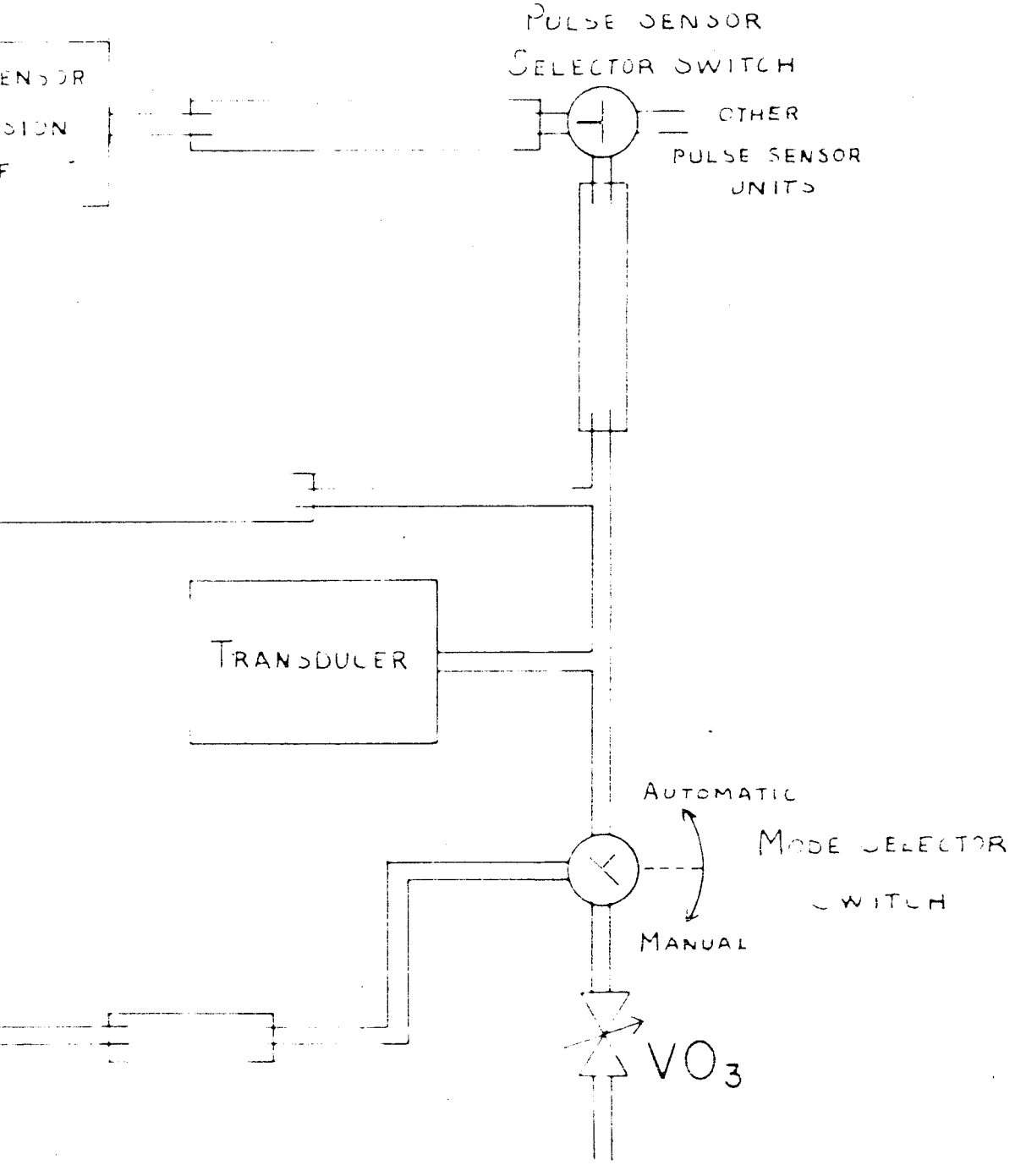

PULSE SENSOR
SELECTOR SWITCH
OTHER
PULSE SENSOR
UNITS
TRANSDUCER
AUTOMATIC
MODE SELECTOR
SWITCH
MANUAL
VO3

Figure 42. Programmed relay valve unit

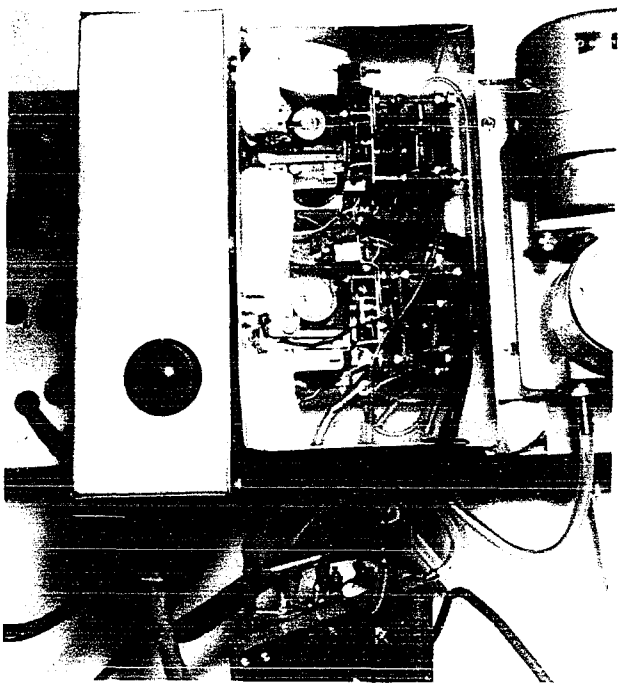
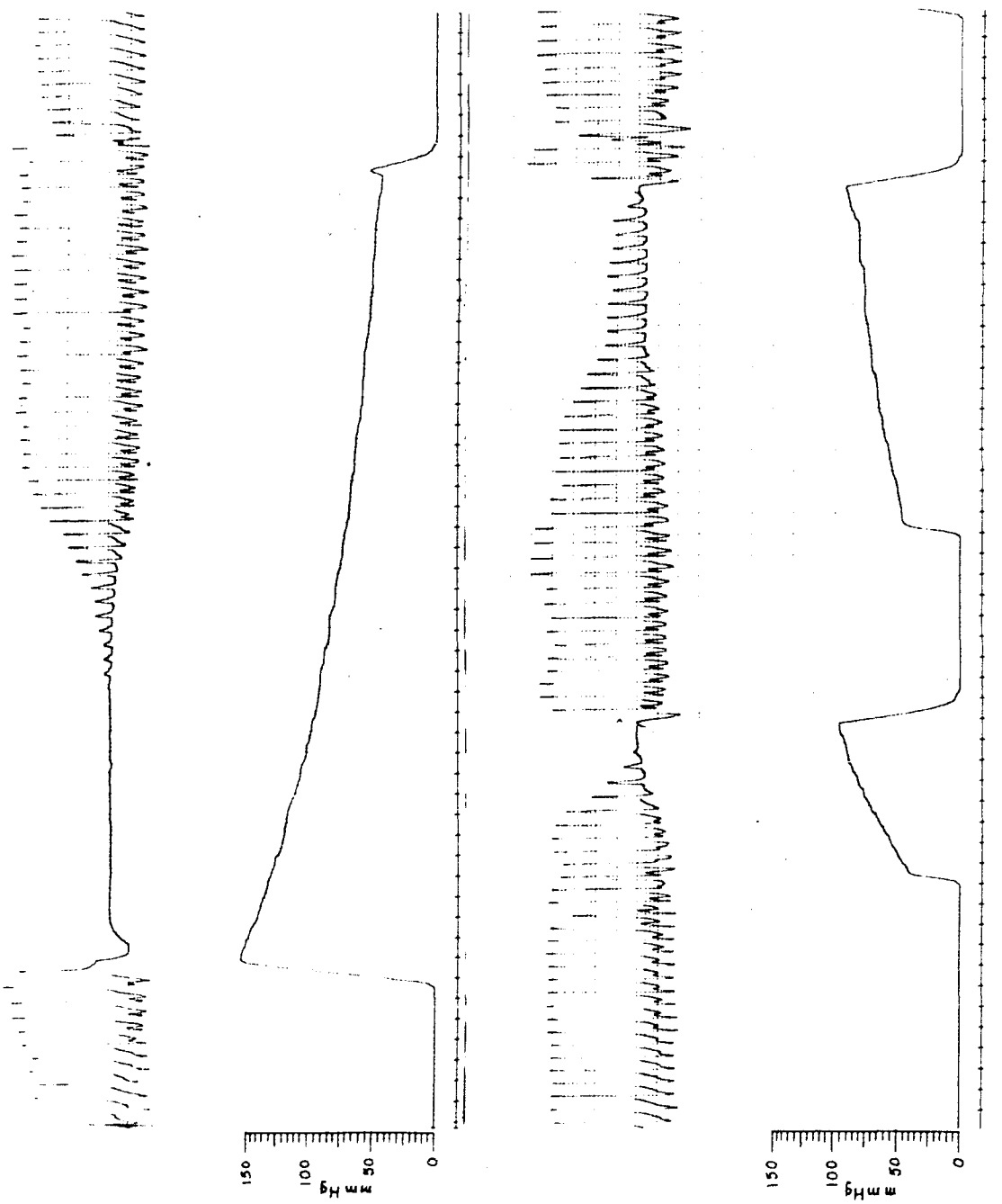


Figure 43. Pulse and occlusion pressure waveforms obtained during the manual mode of operation

Figure 44. Pulse and occlusion pressure waveforms obtained during the automatic mode of operation


150
100
50
0
mm Hg
150
100
50
0
mm Hg

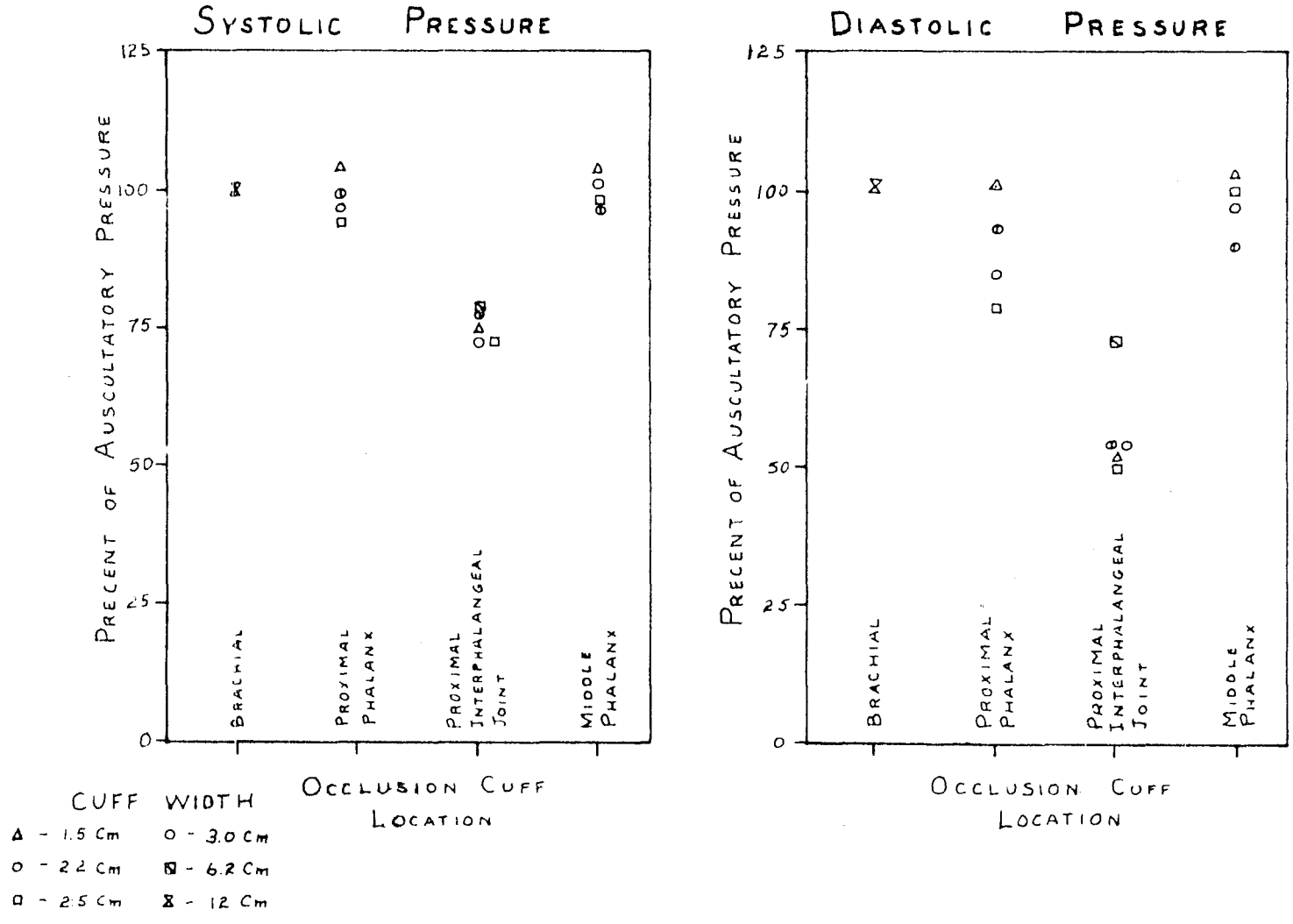
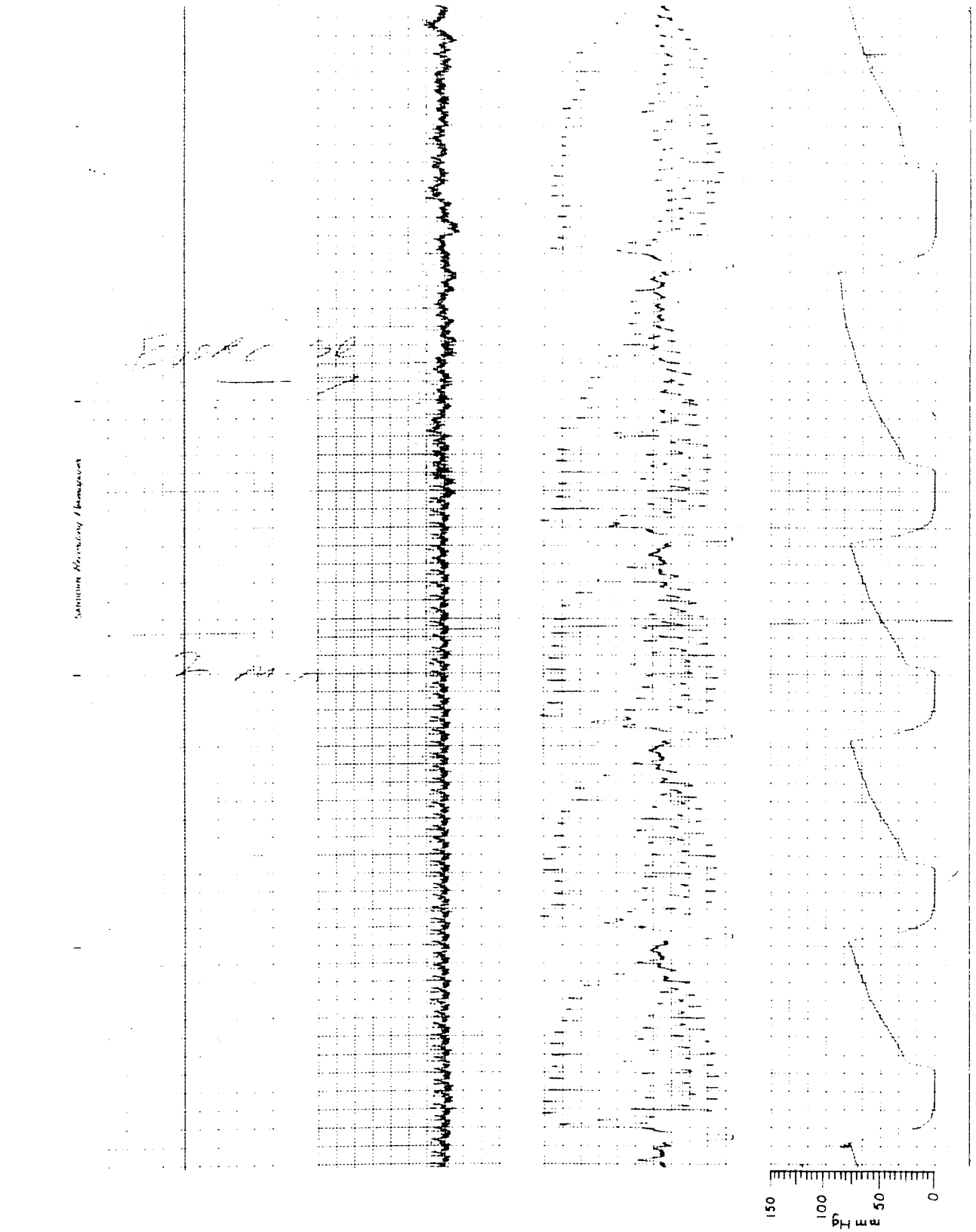


Figure 45. Pressure variation with occlusion cuff location

Figure 46. Electrocardiogram and blood pressure recording obtained during a treadmill exercise evaluation


150
100
50
0
mm Hg

XI. TABLES

Table 1. General specifications for an electronic blood pressure measuring system

Item	Specification
Operating modes	
Automatic	Blood pressure determined by suitable electronic circuitry on a cyclic basis. Operator adjustments should be kept to a minimum during measurement procedures
Manual	Pressure operation should be similar to technique used with auscultatory method
Measurement conditions	Both modes of operation should function during a standard treadmill ergometer test
Occlusion system	
Pressure range	An air pressure system with a pressure range of 0-300 mm Hg
Occlusion cuff	The system should be capable of operation with either brachial or digital cuff
Occlusion cycle	The occlusion system should have a minimum repetition rated one pressure determination per minute
Presentation of measurements	Blood pulse amplitude variations should be recorded on one channel with the corresponding occlusion pressure variations recorded on an adjacent channel

Table 1 (Continued).

Item	Specification
Blood pulse waveform	
Waveform characteristics	Pulse amplitude changes should be retained as detected by the pulse sensor. The frequency content of the waveshape does not have to be retained
Pulse rate	The system should operate for pulse rates of from 50 to 210 heart beats per minute

Table 2. Occlusion cuff color code for cuff width identification

Cuff color	Cuff width cm
Blue	1.5
Red	2.2
Pink	2.5
White	3.0
Purple	6.2

Table 3. Components for the pulse sensor selector switch

Component symbol	Component identification
PC_1	Photoconductive cell, Clavix Type, CL504-SL
LP_1	Subminiature lamp, Chicago Miniature Lamp, CM-666
LP_2	1 1/2 volt pilot lamp
LH	Lamp holder, L bracket type
DB	1 1/2 dry cell battery
S-1	3 loafer, 3 position switch adapted with 2 way air valve
J-1	9 pin subminiature female connector
J-2	4 pin female Amphenol connector
J-3	24 pin male Amphenol connector
P-2	4 pin recessed male Amphenol connector
R_1	1 kilohm resistor
R_2	1 kilohm potentiometer
R_3	470 kilohm resistor
R_4	190 kilohm resistor

Table 4. Operating controls

Operating control	Function
Pump on-off switch	Provides 115 volts AC power to control relays and program pressure valve; provides air pressure Note: Must be on to effect occlusion. Turn on only when pulse sensor unit is on subject
Pump pilot lamp	Indicates pump on
Mode selector	Provides plate voltage to control relays. Selects proper circuit for pressure control system. All control systems returned to stand-by and pressure vented in center position
Automatic	Provides plate voltage to, pulse, systolic, diastolic, and jump control circuits. Permits automatic measurement on programmed cycle
Manual	Provides plate voltage to jump control circuit. Provides rapid rise of pressure to preselected maximum pressure. Pressure is permitted to decrease at rate dependent upon manual pressure decay control setting Note: Spring return to center position
Minimum diastolic pressure	Selects jump pressure at start of automatic mode. Functions only with mode selector in automatic. Pressure jumps to either 50 or 80 mm Hg before beginning the diastolic rate of pressure increase

Table 4 (Continued).

Operating control	Function
Maximum manual pressure	Selects jump pressure at start of manual mode. Functions only with mode selector in manual. Pressure jumps to one of four values. 150, 200, 250, 300 mm Hg
Manual pressure decay control	Selects pressure decay rate in manual mode. Functions only in manual mode. Clockwise rotation decreases the rate of pressure decay Note: Multi-turn pressure needle-valve
Pulse sensor cable	Input from pulse sensor. Provides power to photoconductive cell and light source. Receives input pulse signal. Air connection provides pressure to pulse sensor occlusion cuff
Pulse sensor light intensity	Controls intensity of digital light source. Varies photoconductive cell input pulse amplitude Note: Use in conjunction with sensitivity to establish decoder calibration
Sensitivity	Controls sensitivity of first stage amplifier and input to the recorder driver-amplifier for the pulse waveform recording channel Note: Use in conjunction with light intensity to establish amplifier calibration

Table 4 (Continued).

Operating control	Function
Centering	Controls centering of pulse waveform recording channel
Air pressure DC output voltage	Provides output to DC input of a Sanborn ECG preamplifier to record the occlusion pressure waveform
Pulse sensor selector	Connects pulse sensor in use to pulse sensor cable. In the off position, provides for input termination to amplifier

Table 5. List of components for the differential preamplifier of the decoder unit

Component symbol	Component identification	Component symbol	Component identification
R_5, R_6	5 megohm resistor.	R_{27}, R_{28}	470 kilohm resistor*
R_7, R_8	1 kilohm resistor	R_{29}, R_{30}	2.7 kilohm resistor*
R_9	1.2 megohm resistor	R_{31}, R_{32}	1 megohm resistor
R_{10}, R_{11}	1.5 megohm resistor*	R_{33}, R_{34}	1 kilohm resistor
R_{12}	2.5 megohm resistor	R_{35}	100 kilohm resistor
R_{13}	150 kilohm potentiometer	R_{36}, R_{37}	150 kilohm resistor
R_{14}, R_{15}	470 kilohm resistor*	C_1, C_2	0.05 μf disk capacitor
R_{16}, R_{17}	2.7 megohm resistor*	C_3, C_4	100 $\mu\mu$f " "
R_{18}, R_{19}	1 megohm resistor	C_5, C_6	0.001 μf " "
R_{20}, R_{21}	1 kilohm resistor	C_7, C_8, C_9, C_{10}	0.05 μf " "
R_{22}	2.5 kilohm potentiometer	V_1	5751 electron tube
R_{23}	1.2 megohm resistor	V_2	5814 " "
R_{24}, R_{25}	1.5 megohm resistor*	V_3	5751 " "
R_{26}	33 kilohm resistor		

All resistors are 1/2 watt, 10% tolerance

*Matched to within 2%

Table 6. List of components for the systolic decoder circuit

Component symbol	Component identification	Component symbol	Component identification
R38	500 kilohm potentiometer	R50	22 kilohm resistor, 1 watt
R39	100 kilohm resistor, 1 watt	R51	4.7 kilohm resistor, 1 watt
R40	1 kilohm resistor, 1 watt	C_{11}, C_{12}, C_{13}	0.1 f molded paper capacitor
R41	100 kilohm potentiometer	C_{14}	12 $\mu\mu$f disk capacitor
R42	470 kilohm resistor	C_{15}	60 μf electrolytic capacitor
R43	47 kilohm resistor, 1 watt	C_{16}	1 μf electrolytic capacitor
R44	470 kilohm resistor	CR_1, CR_2	1N461 diode
R45	2.2 megohm resistor	CR_3, CR_4	1N91 diode
R46	56 kilohm resistor, 1 watt	V_{4A}	1/2 (5751) electron tube
R47	1 kilohm resistor, 1 watt	V_5	(5814) electron tube
R48	500 kilohm potentiometer	V_{6A}	1/2 (5814) electron tube
R49	500 kilohm resistor		

All resistors are 1/2 watt, 10% tolerance, unless otherwise noted

Table 7. List of components for the diastolic decoder circuit

Component symbol	Component identification	Component symbol	Component identification
R_{52}	1 megohm resistor	R_{69}	500 kilohm potentiometer
R_{53}	22 kilohm resistor, 1 watt	R_{70}	500 kilohm resistor
R_{54}	1 kilohm resistor, 1 watt	R_{71}	22 kilohm resistor, 1 watt
R_{55}	1 megohm potentiometer	R_{72}	4.7 kilohm resistor, 1 watt
R_{56}	10 kilohm resistor, 1 watt	C_{17}, C_{18}	0.1 μf paper capacitor
R_{57}	1 megohm resistor	C_{19}, C_{20}	60 μf electrolytic capacitor
R_{58}	10 kilohm resistor	C_{21}, C_{22}, C_{23}	0.1 μf paper capacitor
R_{59}	1 megohm resistor	C_{24}	12 $\mu\mu$f disk capacitor
R_{60}	100 kilohm resistor	C_{25}	60 μf electrolytic capacitor
R_{61}	1 kilohm resistor, 1 watt	C_{26}	1 μf electrolytic capacitor
R_{62}	100 kilohm potentiometer	CR_5, CR_6	1N461 diode
R_{63}	470 kilohm resistor	CR_7, CR_8	1N91 diode
R_{64}	47 kilohm resistor, 1 watt	V_{4B}	1/2 (5751) electron tube
R_{65}	470 kilohm resistor	V_{6B}	1/2 (5814) electron tube
R_{66}	2.2 megohm resistor	V_7	(5814) electron tube
R_{67}	56 kilohm resistor, 1 watt	V_8	(5814) electron tube
R_{68}	1 kilohm resistor, 1 watt		

Table 8. Maintenance alignment procedure

Part	Item	Procedure
I		General system inspection
	1.	Remove the decoder unit from the Sanborn recorder. Connect the maintenance patch cable between the decoder unit and the driver-amplifier. Remove the patch cable between the air pressure DC output voltage jack and the ECG DC input jack. Check for the proper interconnection of all other units
	2.	Place the mode selector switch in the center position and the pulse selector switch in the off position. The pump switch should be in the off position
	3.	Turn the master power switch for the Sanborn recorder to the on position. Turn on the power supply units for the decoder chassis and the ECG preamplifier chassis to be used with the pressure recording channel. Allow five minutes for equipment stabilization before proceeding with the alignment
	4.	Measure the voltage at all test points and compare the measurements with the voltage readings of Table 10 Note: If the voltage at the red test jack is not 45 ±1 volt, adjust the B+ voltage on the driver-amplifier to obtain this desired voltage
II		Pressure transducer circuit alignment
	1.	Connect a 0-300 mm Hg bellows type gauge to the variable pulse sensor air connection on the pulse sensor selector unit. Turn the pulse sensor selector to the variable position

Table 8 (Continued).

Part	Item	Procedure
		Note: The use of a gauge is recommended in consideration of the small volume of air required in the digital occlusion cuff. However, a mercury manometer may be used if compensation is made for the additional air volume involved
	2.	Rotate the sensitivity control of the ECG preamplifier for the occlusion pressure channel fully counter-clockwise. Place the attenuator on the 10 position. Disconnect the ECG patch cord from rear of the driver-amplifier for this channel. Connect a patch cord from the air pressure DC output voltage jack to the DC input jack of the ECG preamplifier
	3.	Set the maximum manual pressure control to 150. Turn the pump on
	4.	Depress and hold the mode selector switch in the manual position. Observe the increase in pressure on the gauge. Adjust R_{74} so that the pressure in the system rapidly increases to a maximum value of 150 mm Hg and then slowly decreases. If the rate of pressure decay may be controlled with adjustment of the manual pressure decay control
		Note: If the pressure continues to increase without reaching a maximum pressure below 300 mm Hg, R_{87} located in the control relay unit should be adjusted. To adjust the thyratron current, rotate R_{87} fully clockwise. Then slowly rotate R_{87} counter-clockwise while testing the performance of the pressure circuit. This item should be performed each time the thyratron, V_9, is replaced

Table 8 (Continued).

Part	Item	Procedure
	5.	Adjust the sensitivity and centering controls on the ECG preamplifier to produce a full scale displacement from a base line located 5 mm from the right margin of the recording channel each time the manual mode is activated
III		Decoder unit alignment
	1.	The mode of operation switch should be in the center position. The pump switch and the pulse sensor selector switch should be in the off position. Rotate the sensitivity and pulse sensor light intensity controls fully counter-clockwise
	2.	Adjust the centering control for a base line located approximately 15 mm from the right margin. Turn the sensitivity control over the full range and watch the recorder stylus. If there is any stylus motion, adjust the balance control, R22, until the motion disappears
	3.	Rotate R38 fully clockwise. Rotate R41 and R62 fully clockwise. Check to insure the pump is off. Place an oscilloscope high impedance test probe in pin 4 of the chassis test socket
		Note: Fully clockwise is the normal position of R34 for the pulse sensors provided. The oscilloscope will be used to monitor DC waveforms with a peak amplitude of 50 volts
	4.	Place a finger of a subject into the variable pulse sensor without using the occlusion cuff. Turn the pulse sensor selector to the variable position and the mode switch to automatic
	5.	Perform item 8 of the operating alignment of Table 10

Table 8 (Continued).

Part	Item	Procedure
	6.	Adjust R41 until the waveform observed on pin 4 with the oscilloscope returns to the zero base line when all digital pulses have been removed from the pulse recording. The digital pulses may be removed by occluding the finger of the subject proximal to the pickup sensor or by increasing the light intensity and decreasing the sensitivity until the pulse waveform diminishes. Return the mode switch to the center position Note: The purpose of adjusting R41 is to establish the minimum pulse which will effect the proper triggering of the multivibrator
	7.	Place an occlusion cuff around the finger of the test subject and connect the cuff to the variable air connector of the pulse sensor selector. Perform item 8 of the operating alignment if necessary and turn the recorder on at a speed of 1 cm per second
	8.	Rotate R55 fully counter-clockwise. Turn the pump switch on and place the mode selector in the automatic position. Adjust the systolic air orifice located on the top of the programmed relay valve unit to give a smooth rate increase of the systolic occlusion pressure. The suggested rate of systolic occlusion pressure is 10 mm Hg per second over the pressure range from 80 to 120 mm Hg. Rotate the recycle delay adjust fully counter-clockwise
	9.	Rotate R62 and R69 fully clockwise. Slowly rotate R55 clockwise while observing the recordin gs. The pressure should increase at the diastolic rate following the initial pressure jump and until 1.5 seconds after the start of the decrease in the amplitude of the pulse waveform. Adjust R62 and R69 to obtain the proper release of the diastolic circuit

Table 8 (Continued).

Part	Item	Procedure
		Note: The diastolic decoder contains circuits with charging time constants in excess of 1 minute and provision should be made for adequate charging of these circuits
	10.	Adjust the diastolic air orifice located on the top of the programmed relay unit to produce a rate of pressure increase of 5 mm Hg per second. Check the systolic rate of pressure increase and perform item III - 8 if necessary
IV		Automatic recycle alignment
	1.	Complete the operating alignment of Table 10. Leave the instrument operating in the automatic mode with a test subject
	2.	Rotate the recycle time adjustment, R89, located on the top of the control relay unit until the desired rest interval between occlusion pressure cycles has been obtained
	3.	Return the pulse sensor selector switch to the off position and the mode selector to the center position. Turn the pump off and remove the test subject from the pulse sensor. The unit should now be ready for use in blood pressure measurement

Table 9. Test socket voltage readings

Test point	Type of reading volts	Meter* reading volts	Remarks
Nine pin socket on decoder chassis			
Pin 1	DC	-54	Negative bias voltage
Pin 2	DC	-100	Negative power supply voltage
Pin 3	DC	0	-90 when V8 energized
Pin 4	DC	0	V_{5A}, pin 7
Pin 5	DC	0	V_{5B}, pin 2
Pin 6	AC	4	
Pin 7	AC	4	
Pin 8	DC	0	250 mode switch in automatic
Pin 9	DC	-93	
Decoder connector octal sockets of maintenance cable			
Pin 1	DC	250	Chassis and system ground
Pin 2	DC	0	
Pin 3	DC	0	Measure 4.2 volts between pins 3 and 4
Pin 4	DC	0	
Pin 5	AC	3.1	Measure 6.3 volts between pins 5 and 6
Pin 6	AC	3.1	
Pin 7	DC	13	
Pin 8	DC	13	
Red test points	DC	45 ±1	Bias voltage to driver-amplifier

*Readings were taken with a VTVM with the common lead attached to the chassis

Table 10. Operating alignment procedure

Item	Procedure
1	Connect the pulse sensor cable to the pulse sensor cable input jack. Insure the air fitting is properly connected. Place the mode selector switch in the center position and the pump switch off. Set the pulse sensor selector switch to the off position. Connect the patch cable between the air pressure output voltage to the DC input of a ECG preamplifier
2	Turn on the Sanborn recorder master power switch and the power supply switches for all channels to be used
3	Adjust the centering control of the decoder unit so that the base line is approximately 15 mm from the right margin of the recording channel
4	Place the attenuator switch of the ECG preamplifier to the 10 position. Adjust the centering control of the ECG preamplifier unit to be used for the occlusion pressure recording channel for a base line located 5 mm from the right margin

Note:

Attenuator switch setting	Corresponding full scale pressure reading
10	150 mm Hg
20	300 mm Hg

5	Place a finger of the test subject in a pulse sensor unit and turn the pulse sensor selector switch to the position corresponding to the pulse sensor in use
6	Turn the pump switch on, select a recording speed and start the recorder in motion

Table 10 (Continued).

Item	Procedure
7	Set the maximum manual pressure control to 150 and depress and hold the mode selector switch to the manual position. Observe the pressure recording and release the mode selector switch once the pressure has ceased to increase Note: If the recording tracing does not move over the full recording channel the maintenance alignment procedure for the pressure transducer circuit in Table 8 should be performed. The blood pressure measurement run may be completed and a pressure calibration performed at the completion of the test
8	Turn the decoder sensitivity control fully counter-clockwise and vary the pulse sensor light intensity control to obtain a maximum amplitude of the recorded pulse waveform. If the recorded pulse waveform exhibits power line (60 cycle) interference the subject ground should be checked. If the interference remains, the pulse sensor light intensity should be turned clockwise until the interference decreases. Adjust the sensitivity control to obtain a pulse recording amplitude of approximately 2 cm peak to peak amplitude Note: The maximum clockwise position of the pulse sensor light intensity control may cause some subject discomfort due to local thermal effects. Therefore, continued use of the maximum light intensity setting should be avoided
9	Select the desired setting of the minimum diastolic pressure control if the automatic mode of operation is to be used. If the manual mode of operation is to be used both the desired value of the maximum manual pressure selected and the manual pressure decay control should be turned clockwise. The pressure decay control should be varied to obtain the desired decay rate during the portion of the manual mode following the initial pressure jump

Table 10 (Continued).

Item	Procedure
	Note: Adjustment of the maximum manual pressure control or the manual pressure decay control is not required in the automatic mode of operation
10	Move the mode selector to the desired mode of operation. The selector must be held in the manual mode position and is spring loaded to return to the center position when released from manual. Return of the mode selector switch to the center position interrupts all measurements, and vents the occlusion cuff
11	At the completion of the measurement, the mode of operation switch should be returned to the center position and the pulse selector switch should be placed in the off position. Then the pump switch may be turned off and the subject removed from the pulse sensor

Table 11. Component list for the pressure transducer circuit

Component symbol	Component identification	Component symbol	Component identification
R73	5000 ohm resistor	R86	22 kilohm resistor
R74	5000 ohm potentiometer	R87	5000 ohm potentiometer
R75	1.2 megohm resistor	R88	47 kilohm resistor
R76	470 kilohm resistor	C27	0.3 μf disk capacitor
R77	1.5 megohm resistor	Pressure transducer	Bourns Model 504 transducer
R78	1 megohm resistor	V_9	2D21 thyratron tube
R79	39 kilohm resistor	J_4	balanced phono jack
R80	470 ohm resistor	S_2	SPDT rotary switch
R81	270 ohm resistor	S_3	SP4T rotary switch
R82	1000 ohm resistor	S_4	6 pole, 3 position mode switch, 1 position spring return to center position
R83	680 ohm resistor	Z_1	TI652 Zener diode
R84	470 ohm resistor		
R85	410 ohm resistor		

Table 12. Component list for the control relay unit, the supplementary negative power supply, and the programmed pressure valve unit

Component symbol	Component identification	Component symbol	Component identification
Control relay unit		Negative power supply	
K_1	SPDT 5000 ohm plate relay	T_1	Stancor PM8419 transformer
K_2	4PDT 5000 ohm plate relay	F_1	fuse, 3 amp 110 volt
K_3	SPDT 5000 ohm plate relay	R_{91}	2.5 kilohms, 10 watt resistor
K_4	3PDT relay; 110 vac coil	R_{92}	2.5 kilohms, 10 watt resistor
K_5	4PDT relay; 110 vac coil	C_{30}	10 μf 450 volts electrolytic capacitor
K_6	SPDT relay; 6.3 vac coil	C_{31}	20 μf 250 volts electrolytic capacitor
K_7	SPDT relay; 6.3 vac coil	C_{32}	0.05 μf disk capacitor
K_8	SPDT time delay relay, Amperite 6C16T	V_{10}	6 x 4 rectifier tube
R_{87}	5000 ohm potentiometer	V_{11}	OB2 voltage regulator tube
R_{89}	10 ohm potentiometer	J_9	7 pin miniature test socket
R_{90}	0.1 megohm resistor	J_8	7 pin large socket
TS	Terminal strip - 20 connections	P_{10}	110 vac power cord
J_5, J_6	11 pin socket	Programmed relay pressure valve	
J_7	2 pin female receptacle		
P_8	7 prong plug	K_9, K_{10}	Modified 24-pos. stepping relay
C_{28}, C_{29}	8 μf, 600 volt capacitor	P_6	11 pin plug
Z_2	1N757 Zener diode	Pump	Dyna pump, American Scientific Apparatus Company
Z_3, Z_4	1N1823 Zener diode	pressure valve	Nupro 2C-4C check valve
C_{33}, C_{34}	0.5 μf 400 volt capacitor		

Table 13. Output voltage characteristics of the supplementary power supply

Pin location on chassis output terminal	Output voltage volts	Output current	Remarks
1	-100 DC	40 ma	Less than 2% ripple voltage
2	System electrical ground		
3	6.3 AC	3 amp	
4	6.3 AC	3 amp	
5	115 AC	2 amp	Fused circuit
6	115 AC	2 amp	