Microcomputer controlled blood pressure monitor

using

Doppler ultrasound

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by

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INTRODUCTION

An accurate blood pressure measurement is important for many diagnostic, surgical, and patient-care procedures. Systems have been designed and developed to measure systolic pressure accurately, but diastolic pressure measurement has always been difficult. This may be attributed to the fact that the diastolic pressure point does not have a significant signal variation either in terms of wall motion or flow; therefore, measurement of diastolic pressure demands a great amount of sensitivity from the measurement system and consequently the design becomes critical. The possibility of noisy environment or significant patient movement puts a greater restriction on the system design. The subject under test could be hypotensive or be an infant, and this could change the measurement parameters considerably. System specification should take care of this variation without drop in measurement accuracy. Obviously, it is difficult for a system to work accurately under such varied conditions and hence a tolerance factor is added to the results. In an operator-assisted measurement there is the additional problem of variability between operators.

Despite all the above mentioned defects, present day measurement techniques are fairly accurate, and coupled with the physician's expertise, are adequate for many situations, but their accuracy is not sufficient for many diagnostic and surgical procedures. This has prompted the search for, and design of, better and more reliable methods for measuring blood pressure.

A microcomputer, or a microprocessor, could be incorporated into the measurement system to substitute for the human operator and also to eliminate most of the measurement errors. Automatic methods for recording systolic and diastolic pressures can be particularly valuable for patient monitoring, for mass screening, and for eliminating the variability between operators. This thesis describes an automated blood pressure system. Specific attention is given to the inflation and deflation of the cuff, the processing of blood pressure data, and the incorporation of a microcomputer into the system.

Current trend in instrumentation is towards providing fast and accurate signal processing, with specific implications to 24 hour data acquisition and processing situations. The advent of single-chip microcomputers have made all this a feasible proposition, and in addition, provided for portability and battery powered operation. Thus one could envision a blood pressure monitor, using ultrasound, incorporating the above features. In addition, the microcomputer allows for better processing capability and increased data storing ability.

An ambulatory blood pressure monitor could be of great help in trend studies. Blood pressure data recorded by sampling over a 24 hour period could be obtained without the subject having to visit the hospital. An automatic cuff inflation and deflation procedure and battery powered operation provide most of the features. A microcomputer incorporated into the system will make the trend recording entirely feasible. The advent of single-chip microcomputers, with their own memory and input/output communication features, can greatly enhance the possibilities of storing blood pressure data recorded over a longer period of time.

A single-chip microcomputer (made by Zilog Inc., Intel Corp., or National Semiconductor Corp.) can provide both real-time and batch (stored) data processing. The architecture of a single-chip microcomputer is ideally suited for instrumentation and control applications. Input/output ports can serve to accept data, to output data, and also to control external operations. On-board memory is small, but most single chip microcomputers support external memories (up to 64K bytes) without appreciable increase in the associated hardware. On-board timers support timed operations. Most of the programming could be achieved in BASIC, a user-friendly language. Ease with which they could be interfaced to external systems has made them very useful for instrumentation applications.

Both real-time and stored data processing can be performed on the microcomputer. An electronic pressure transducer, connected to the cuff, can be interfaced through an A/D converter to the microcomputer to achieve real-time data processing. The microcomputer can provide blood pressure values as the cuff pressure is reduced from above systolic pressure to values below diastolic pressure. Alternately, the output from the signal acquisition circuit could be saved in the form of digital pulses in the memory. At the end of each cycle, the microcomputer can process the frequency contents of the audio signals and try to identify systolic and diastolic pressure points.

An ambulatory monitor need be small in size and light in weight. A single-chip microcomputer system can be incorporated into such a monitor and not add to its size and bulk. Thus, without much of an increase in cost (single-chip microcomputers are inexpensive) and with no additional discomfort other than carrying a small, fairly lightweight unit on one's person, the automatic ambulatory blood pressure monitor provides appreciable enhancement in diagnostic values.

In the present thesis, conventional Doppler technique is used for measuring the blood pressure. The brachial artery is subjected to occlusion by a cuff inflated and deflated automatically. This measurement method should provide reliable operation in a high noise environment and in significant patient movement situations. In addition, accurate results could be obtained from infants and hypotensive adults. The design emphasizes portability and battery powered operation. An attempt was made to record and process 24 hour blood pressure data. Efforts were directed towards providing a modem link between the patient-strappable ambulatory monitor and the hospital central processing unit. The single-chip microcomputer made by Zilog Inc. (referred as Z8671 MCU) is used in this thesis.

Since an ultrasound method was designed to transduce the signal compatible with a computer, this section will begin with a description of the design and implementation of a circuit for transducing the arterial wall motion signal. This will be followed by a procedure for automatic inflation and deflation of the cuff, and finally, a discussion of the interfacing of the computer compatible outputs of the transducing circuitry to a microcomputer.

Doppler ultrasound is used as the measurement technique in the system designed. A piezoelectric crystal is used to transduce wall motion signals from the brachial artery which is under an occlusive cuff. These signals are processed and digitized by a CMOS (complementary metal oxide semiconductor) circuit to be compatible with a microcomputer. The microcomputer operates on the digitized signal to determine the systolic and diastolic pressure values. Emphasis of the design is towards portability, automation, and data storing ability.

The block diagram representation for the system is as shown in Figure 1.

This thesis has been divided into the Doppler signal acquisition system and the Doppler signal processing system.

The Doppler signal acquisition system, described in the first half of the thesis, comprises of the ultrasonic transducer detecting the wall motion signals and the associated circuitry that converts the wall motion signals into audio signals and digital pulses. Audio signals after amplification are fed to a speaker to identify systolic and diastolic pressures from the sounds associated with them. Pressure reading on the mercury manometer is correlated with the speaker output. The digital pulses separate in time as cuff pressure is varied between systolic and diastolic pressure values. At systolic pressure, the pulses are closely packed, while, at diastolic pressure they are well apart.

The Doppler signal processing system, described in the second half of this thesis, comprises of a pressure transducer that converts the cuff pressure into voltage. This voltage is converted into digital values by an A/D (analog-to-digital) converter. Digitized values are input to a

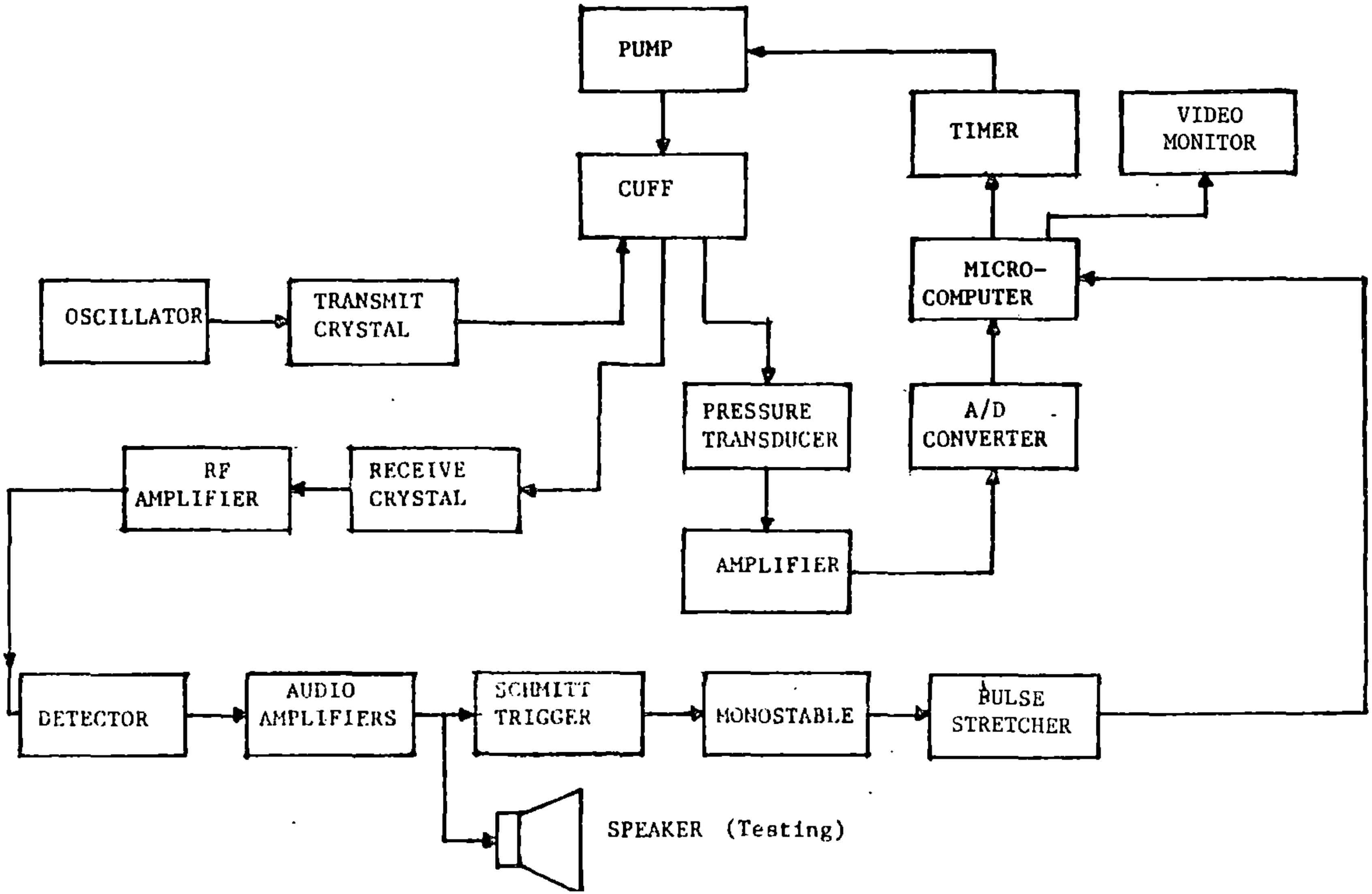


Figure 1. System block diagram

microcomputer. Digital pulses from the signal processing circuit of section I are input to the microcomputer which is programmed to measure the interval between pulses, as cuff pressure is decreased. Systolic and diastolic points are identified by the microcomputer by determining the shortest and the longest intervals between pulses, respectively. A CRT linked to the microcomputer displayed the blood pressure values.

A timing diagram is provided in Appendix A. Appendices C, D, E, and F give a description of the software used in this project.

The circuits desribed in the Doppler signal acquisition system are given as an alternative means of inflating and deflating the cuff automatically, in the absence of a microcomputer. Automatic inflation and deflation of the cuff using a microcomputer is described in the Doppler signal processing system.

Each of these sections will be described in detail in the following pages.

LITERATURE REVIEW

Background

Reliable monitoring of blood pressure in the clinical situation demands that a variety of requirements be met by the system. Perhaps the most important among these is the ability to obtain accurate data from the patient. These data should be readily obtainable from patients with different body characteristics, various health conditions, and in places of high acoustic noise and electromagnetic interference. Direct arterial pressure recordings are limited by the need for arterial puncture, for special and sterile apparatus, and the vagaries of damping in an hydraulic system. As a consequence, the indirect measurement of blood pressure is used in most clinical situations.

Earlier indirect methods

Several methods have been used to determine the state of flow in the artery and these methods have enabled the indirect calculation of systolic and diastolic blood pressure values. Most commonly used method in routine clinical measurements of blood pressure is sphygmomanometry (auscultatory method). This method makes use of an occlusive cuff applied to an arm and positioned over the brachial artery. The procedure involves the detection of the change in character of sounds produced by the pulsatile flow through the constriction in the artery caused by the occlusion of the artery (Cobbold, 1974). Ware and Anderson (1966) reported that the Korotkoff sounds, detected by a stethoscope or a crystal microphone transducer placed on the artery, change in character and intensity as the cuff pressure is reduced. They correlated the appearance and disappearance of these sounds with cuff pressure to systolic and diastolic pressures. They performed spectral analysis of the Korotkoff sounds on five adult males at rest, during exercise, and immediately after exercise. Sounds were recorded using a piezoelectric crystal microphone placed on the brachial artery at the distal edge of the cuff. They classified each of the recorded sounds into four phases: the first clearly discernible sound as phase 1, the first muffled sound as phase 4, the last unmuffled sound as phase 3, and the sounds intermediate between phase 1 and phase 3 as phase 2. Systolic and diastolic pressures were measured as the beginning of phase 1 and beginning of phase 4, respectively.

This technique is sensitive to changes in the environment, and to the cuff size; hence, it gives different results under different conditions. Conventional sphygmomanometry fails to give accurate blood pressure measurements in infants or hypotensive adults. Ware et al. (1967) reasoned that small or constricted blood vessels produce very weak Korotkoff sounds and that the microphone (stethoscope) is not sensitive enough to pick up these low amplitude signals. They reported successful blood pressure recordings on infants and on adults in hypotensive shock using the ultrasound method.

Seelinger and Hoebel (1969) found that artifacts produced by patient movement significantly affected the accuracy of the pressures obtained by sphygmomanometry. They adopted a new technique of performing the measurements with increasing cuff pressure. A cuff was applied to the upper arm, and a microphone, instead of a stethoscope, was fastened to the cuff. The microphone featured a built-in amplifier with emitter follower that eliminated the artifacts caused by cable movement. They used the Korotkoff sound method to determine the blood pressure and reported fairly accurate results.

Ambient noise also impairs the utility of the auscultatory technique. Snyder (1968) reported that the procedure is inaccurate in high noise environments due to the presence of signals in the frequency range of the Korotkoff sounds. Instead of depending on the production of sounds from the arterial wall and blood turbulence, he suggested that the movement of the arterial wall be monitored using ultrasound.

Finally, in clinical situations, the Korotkoff sound method is highly dependent on the hearing capability of the operator. This was demonstrated by the findings of Silverstein (1974), who reported that the variability between operators could lead to inconsistency in the measurement of blood pressure values. In his study, he had used the Korotkoff sound method on hypotensive adults and found that the determination of data identifying hypotension and hypertension is dependent on such subjective factors as the examiner's auditory acuity for both amplitude and frequency, and the reflexive capacity to correlate the auscultatory sounds with the pressure thus determined. He designed a device for the automatic measurement of blood pressure, eliminating the necessity for operator assistance.

Thus, the auscultatory method, suggested by Korotkoff in 1905, has been widely used in routine clinical measurements. Most of the shortcomings associated with this method have been identified and efforts have been made to rectify them.

The second most commonly used method is oscillometry. This involves subjecting an artery to counterpressure and observing for oscillations as the counterpressure is varied. Systolic pressure is signalled when, during the reduction of the counterpressure, the pulses appear; diastolic pressure is indicated by the point of maximum oscillations (Geddes, 1970). But the endpoint determinations are debatable and their accuracy is questionable. This was indicated by the conflicting findings of different investigators. Studies have shown that, while systolic pressure could be determined from the oscillometric recordings, for the most part the diastolic pressure was not detectable. The reports of investigators give an insight into the shortcomings of this method. Experimental verification of the systolic and diastolic endpoints led to conflicting reports. Studies conducted on dogs using the oscillometric method showed that, even with the counterpressure above systolic pressure, oscillations were observed. This was contradictory to the belief that the appearance of the oscillations marked the systolic endpoint. Investigators also found that at the point of maximum oscillations, the counterpressure was not exactly equal to the diastolic pressure. This contributed to the measurement errors.

As late as 1975, Hartley et al. (1975) reported that oscillometry was not accurate for the determination of diastolic endpoint. Characterization of the diastolic pressure with the point of maximum oscillations was disproved by the occasional occurrence of two maxima and on other occasions by the inconsistency of the oscillations.

One adaptation of the oscillometric method did provide better correlation with the direct method (Geddes, 1970). In one case, the pinna of the ear was placed between a pressure capsule and a light source. With no pressure applied to the capsule, the pulsatile arterial blood flow varied the optical density of the light path, giving a pulsatile output from the photocell. By inflating the capsule to a pressure above systolic pressure, blood was squeezed out of the ear. Now as the pressure was reduced, the pulsations appeared when the pressure fell below systolic pressure. With a continued reduction in the capsule pressure, the amplitude of oscillations increased and the diastolic pressure was marked by the point of maximum oscillations. The values correlated well with those obtained from other methods.

Doppler Ultrasound Method

While the auscultatory and oscillometric methods are simple and sufficiently accurate for most purposes, they both suffer from the disadvantage that they fail to give accurate pressure values for hypotensive patients and infants. Ultrasound has been found to be accurate and free from the errors associated with the earlier methods.

Theory

Doppler ultrasound is a technique in which low power (50 mW/sq.cm.), high frequency (2 MHz and above) sound waves are transmitted towards an object (arterial wall in this case) the movement of which is to be monitored. A reflector, moving in the transmitted ultrasonic beam, will

Doppler-shift some of the sound. Shifted and unshifted sounds are mixed at the receiving transducer to produce a beat frequency f1, which is in the audible range and is given by the relation

$$f1 = (2Fs \times \cos A \times V) / C \qquad (1)$$

where Fs = the transmitted frequency, V = the target (arterial wall) velocity, C = the velocity of sound in the medium, and A = the angle between the transducer and the direction of flow.

Doppler shift is the sum of two parts. The first can be found by noting that to an observer moving with the flow velocity v the transmitter crystal is moving away, and its frequency Fs appears as a frequency F given by

$$F = Fs / ((1 + V \times \cos B) / C) \qquad (2)$$

where B is the angle between the transmitting crystals and the direction of flow.

The second part can be found by observing that the scattering of ultrasound in the fluid is equivalent to the transmission of a frequency F from a virtual source within the fluid. Thus, to an observer fixed at the receiver crystal, the frequency F will appear as a frequency Fr given by

$$Fr = F (1 - ((V \times \cos C) / C)) \qquad (3)$$

where C is the angle between the receiving crystals and the direction of flow.

From equations (2) and (3), the Doppler frequency change can be written as

$$f1 = Fs - Fr = F ((V \times \cos B) / C)) + ((V \times \cos C / C)) \qquad (4)$$

In equation (4), if B = C (transmitting and receiving crystals at the same angle with the direction of flow), then,

$$f1 = (2F \times V \times \cos (B)) / C \qquad (5)$$

Equation (5) is same as equation (1).

In deriving the above equation, scattering from a single region of fluid flow moving with a velocity v was considered. In practice, there exists a distribution of velocities ranging from zero up to a maximum value. Thus the Doppler-shifted signal is not a single frequency, but a band of frequencies. Furthermore, the transmitted and received signals are over a range of angles and not confined to a single angle, as indicated in equation (5). Both these factors cause the received signal to consist of a spectrum of frequencies extending from Fs to just beyond Fs + f1, where f1 is the Doppler shift frequency corresponding to peak velocity.

Equation (5) is actually of the form

$$\pm f1 = (2F \times V \times \cos (B)) / C$$

where the minus and plus signs have been introduced to account for flow towards and away from the transducer, respectively. Thus, the frequency shift is directly proportional to the flow velocity and the signs indicate the direction of flow.

Measurement methods

Ware et al. (1967) found that a method which can sense arterial wall motion directly, using the Doppler ultrasound principle, is sensitive enough to measure systolic and diastolic blood pressures even in infants and hypotensive adults. They used a pocket-portable 8 MHz Doppler ultrasound detector to sense the arterial wall motion signals. The detector consisted of a power oscillator, a low noise amplifier detector, an audio frequency amplifier driving a pair of headphones and two lead zirconate titanate transducers. This method was based on the principle that a reflector moving in the transmitted ultrasound beam will Doppler-shift some of the sound waves and the shifted and unshifted signals when mixed at the receiving transducer produce a beat frequency in the audio range. They reported that as the cuff pressure was decreased below systolic pressure, distinct opening and closing signals appeared. Further, as the cuff pressure was decreased, these two signals separated in time until a closing signal merged with an opening signal. They characterized systolic pressure as the point when the first opening and closing signals appeared and the merging of the two signals as the diastolic pressure. They reported successful measurement of systolic and diastolic pressures on infants and on adults in clinical shock. Their results were within 2 to 4 mmHg to those measured directly by catheter. In addition, the Doppler device driving a headphone could give signals amidst high noise levels.

The same year, Ware and Laenger (1967) accomplished accurate blood pressure measurements using the Doppler ultrasonic principle even when the subject was in extremely high environmental sound levels, in high gravitational fields, experiencing conditions of reduced peripheral blood flow, and a newborn infant. They observed the arterial wall motion and found that distinctive opening and closing signals were detectable with each cardiac cycle. They determined values for the velocity and duration of the opening and closing signals. The brachial artery opening motion had a duration of approximately 100 to 150 ms (6 to 10 Hz), reached peak velocities varying from a few mm/sec, when cuff pressure was near systolic pressure, to several cm/sec when cuff pressure was near diastolic pressure. Arterial closing motion lasted longer than the opening motions.

In the system described by Stegall et al. (1968), Doppler ultrasound was used to determine the state of closure of an artery under an occlusive cuff. They used two piezoelectric crystals, one connected to the 8 MHz oscillator and acting as the transmitter of ultrasound, and the other connected to a narrow band amplifier, functioning as the receiver. Both the crystals were attached to the bottom of the cuff. With this arrangement, they reported successful measurements of both systolic and diastolic pressures on infants and hypotensive patients. Their results were corroborated by the findings of Poppers et al. (1971) on hypotensive patients, Lagler and Duc (1980) and De Swiet et al. (1980) on infants, and Snyder (1968) on subjects in high noise environments.

Doppler ultrasound method was used to record the mechanical behavior of the artery under an occlusive cuff by Massie et al. (1968). They obtained a spectrum of Doppler frequencies corresponding to the artery opening and a different set of frequencies relating to the closing of the artery. But the closing signals were found to vary from person to person. Gray and Hatke (1968) demonstrated that measurement of blood pressure by indirect means can be accomplished with ultrasonic detection of the opening and closing of an artery under a cuff. They used frequency spectra to correlate the opening and closing signals to systolic and diastolic pressures.

Snyder (1968) reported using arterial wall movement to measure systolic and diastolic pressures. He used a standard cuff around the upper arm, and a transducer on the underside of the cuff transmitted and received ultrasonic energy. Systolic and diastolic pressures were identified as the points at which the rapid arterial wall motion (related to the opening of the artery) just begins and stops, respectively.

Hartley et al. (1975) reported that Doppler-shift ultrasound may be used to sense arterial wall velocity distal to the occlusive cuff, and the wall velocity patterns could be characterized to determine systolic and diastolic pressures. They used the Arteriosonde 1010 (Hoffman-La Roche, Inc.) to accomplish the measurement. This system consisted of an 8 MHz oscillator driving a piezoelectric crystal transducer to produce ultrasonic waves. The reference and Doppler-shifted reflected signals were detected by a diode and amplified through a low frequency band pass amplifier. The signal was subsequently used to amplitude modulate a 450 Hz tone that drove a speaker. Using fast Fourier transforms (FFT) on the reflected signals, they reported that frequencies between 5 and 35 Hz accompanied by an increase in amplitude, indicated systolic pressure. Diastolic pressure corresponded to a decrease in amplitude in frequencies from 35 to 85 Hz. Results of this method closely matched with those obtained simultaneously from the direct and operator assisted methods.

Thus, the application of Doppler ultrasound for the indirect determination of blood pressure was proved feasible. The overriding reasons for the success were: accurate results with infants and hypotensive adults, a better signal-to-noise ratio and sensitivity (Hochberg et al., 1974), unaffected by environmental disturbances, and independent of the auditory acuity of the operator.

Doppler signal processing techniques

Although the measurement of blood pressure by Doppler ultrasonic principle was proved to be feasible, there still remained the difficult task of processing the complex Doppler signals.

Mount et al. (1968), found that the reflected signal was phase modulated in proportion to the artery wall displacement, and frequency modulated in proportion to artery wall velocity. They suggested a correlation for the determination of arterial blood pressures from the modulated signals. Gray and Hatke (1968) suggested that Doppler reflected signals could be processed by the choice of frequency selective networks. They reported that the primary signal from the artery is lower in amplitude, but higher in frequency. Thus, any nonlinearity occurring in the receiver due to large amplitude signals from such things as transducer crosstalk and bone return could be eliminated using a frequency selective network. This could be achieved by having a filter at the carrier frequency of proper bandwidth in the radio frequency (RF) amplifier of the receiver. This technique has the advantage of providing uniform amplitude artery motion signals in the presence of variable amounts of crosstalk carrier signal from bone or transducers.

Later, Massie et al. (1970) described a method for filtering the detected Doppler signals to discriminate against low frequency artifact information. Reliable determinations of blood pressure were performed using an EKG (electrocardiogram) gating circuit. Here, an EKG gate, referenced on the QRS complex, provided the time interval during which the pressure pulses reaching the cuff were taken as valid signals. All others were rejected as artifacts. This resulted in greater artifact rejection and enhanced signal-to-noise ratio.

Hartley et al. (1975) used a diode to detect the reference and Doppler-shifted reflected signals and then amplified them through a low frequency band pass amplifier. The diode acted as the rectifier, converting any audio information present in the RF signal into pulsating dc voltages. This signal was used to amplitude modulate a 450 Hz tone, driving a speaker. The FFT was used on the signal to determine the frequency content and to relate that to the systolic and diastolic blood pressures. Baker and Simmons (1976) suggested that phase lock techniques be used for tracking the motion of the arterial wall. They reported that when a suitably shaped sound beam was directed at the arterial wall, signals would be reflected. The phase of these signals with respect to the transmitted signal indicated the position of the wall.

Baker et al. (1976) used a new ultrasound method to relate the phase difference between the transmitted and received signals with the wall motion of the artery. A servo or phase lock loop was formed to cause the transmitted phase to track the received phase. It was found that the detector output was zero when the phase difference was 90 degrees. They used the principle that the relative phase of a signal back scattered from a blood vessel wall depends on the distance between the transmit-receive crystals and the vessel wall. Thus, when motion occurs, the relative phase is different from 90 degrees, causing an error signal. The correction factor was related to the distance and hence to the motion of the arterial wall towards and away from the transducer. With the relation between the arterial wall motion, both toward and away, and systolic and diastolic pressures already established, Baker et al. (1976) reported successful blood pressure measurements. A limitation of this method is that only the change in distance could be measured and not the absolute distance itself. Also, a stable and accurate transducer design was essential for the measurement.

Later, a Doppler ultrasonic instrument, designed by Mcleod et al. (1976) used the Doppler shift of reflected signals for the determination of vessel wall motion. The Doppler system responded only to moving targets and thus eliminated the adverse influence of stationary echoes. The phase information between the transmitted and received signals was used to detect wall motion. Observation of the wall motion required resolving the walls from the underlying tissues and the precise location of the reflected signals. The phase changes between the transmitted and reflected signals were measured and the changes were correlated to the wall motion. Two noteworthy features of the method are the noncriticality of transducer placement and increased signal-to-noise ratio.

Coghlan and Taylor (1980) designed and constructed a real-time spectrum analyzer to convert raw Doppler-shift signals to an easily interpreted form. The Doppler-shifted waveform was reduced to an image (sonogram) and useful parameters or indices were derived from the sonogram. A digital computer used the derived indices to calculate blood pressure, flow, and wall compliance. Rittgers et al. (1980) used the digital fast Fourier transformation to analyze the Doppler ultrasound signals. Digital FFT was chosen because it provided high resolution, real-time analysis and was capable of processing over a wide range of input frequencies.

A brief discussion of this method is essential because of its success and accuracy. The overall system consisted of a bidirectional Doppler unit whose outputs of audio shifted frequencies were filtered and frequency shifted to produce a combined directional signal suitable for single channel spectrum analysis. This signal was amplitude controlled by an autogain circuit prior to being presented to the real time spectrum analyser. Frequency translation was accomplished by single sideband modulation techniques where the forward and reverse signals were shifted about a selected carrier frequency. This constituted an upper sideband forward flow signal and a lower sideband reverse flow signal about a suppressed carrier. Rittgers et al. (1980) reported using this system for identifying stenoses of both extremities. With systolic and diastolic pressure points qualified by a unique combination of forward and reverse signals, this method could be applied to the measurement of blood pressure.

Doppler ultrasound method also permits the reconstruction of the complete blood pressure waveform. This was confirmed by the findings of Ware and Anderson (1966). Zeidonis and Mount (1968) devised a multiple transducer arrangement that made the transducer placement less critical. The feasibility and the improvements made in the measurement procedure, pointed towards the design and development of an automated system. The following methods have been successfully used in the analysis of Doppler ultrasound signals: audio evaluation, analog outputs from zero-crossing detector circuits, decomposition into its component frequencies using narrow-band filters, swept frequency analysis, time compression spectrum analysis, and digital fast fourier transformation.

Directional Doppler Method

As an enhancement on conventional Doppler method, the directional Doppler method was adopted. Unlike the former, the directional Doppler technique could distinguish between the positive and negative frequency shifts. This method helped in detecting the arterial wall motion towards and away from the sensing element. One of the main limitations of the conventional Doppler devices was their inability to discriminate between flow towards and away from the transducer. This was found to be very serious because veins and arteries often lie adjacent to one another with blood flowing in opposite directions. When the vessels are resonated, the

Doppler signal would contain a mixture of arterial and venous signals. This was observed by Baker and Watkins (1967), who reported that the conventional ultrasonic Doppler devices could not distinguish the distance to a moving interface. This was also shown by Mcleod (1967). He found that simple Doppler flowmeters were unable to sense direction, due to their inability to distinguish between positive and negative frequency shifts. This eventually led to the design of systems capable of detecting the direction of motion of the arterial wall.

Directional Doppler signal processing techniques

Two methods were introduced to separate forward and reverse blood flow signals. One was the quadrature phase detection and the other was phase rotation. Both the techniques are used in radio communication. Earlier, Mcleod (1967) observed that a phase shift introduced prior to detection of Doppler signals made possible the identification and separation of positive and negative shifts. This phase shift was introduced by the addition of carrier components. He applied the quadrature phase detection to separate the Doppler shifted sidebands from the carrier frequency. After detection, the phase separated forward and reverse Doppler signals were sent to stereo headphones or zero crossing detectors. The second method referred to as phase rotation produces two independent audio outputs, one for forward and the other for reverse Doppler signals. This is a very effective method, but unfortunately the circuits involved are very complex and hence very difficult to build and adjust.

Use of the directional Doppler method by Kelsey et al. (1969) to detect the pharyngeal wall displacement confirmed the usefulness and accuracy of this method. Further studies by Baker and Simmons (1976) revealed that this technique could be used to distinguish between the motion of the arterial wall towards and away from the transducer.

For the project described in this thesis, conventional Doppler method was chosen over directional Doppler method for reasons of cirucit simplicity. Also, the results from conventional Doppler method have been within acceptable limits.

THE ULTRASOUND METHOD

The output from a conventional Doppler system is as shown in Figure 2. If the vessel wall is moving, the reflected signal will be Doppler shifted in frequency by an amount proportional to the instantaneous wall velocity.

The cuff is inflated to well-beyond systolic pressure, thereby causing the brachial artery to collapse and stop the blood flow. As the cuff is deflated, no signal is heard as long as the cuff pressure exceeds the systolic pressure. But just as the cuff pressure falls below systolic pressure, a distinct thumping sound is heard that is characterized by two distinct opening and closing phases. Systolic pressure is denoted by the occurrence of a closing phase following an opening phase, accompanied by a high frequency audio signal. Opening of the artery generates a high frequency audio signal, while the closing generates a low frequency signal. Further deflation causes a time separation between the opening and closing signals, first to increase and then to decrease. Merging of the two signals is accompanied by a definite change in the audible character of the signal and the cuff pressure at this point is a measure of the diastolic pressure (Ware and Laenger, 1967).

In the directional Doppler method, the directional information is preserved by electrically separating the reflected ultrasonic waves which have been shifted up in frequency from those with a downward shift (Baker and Watkins, 1967). It has been seen that a phase shift introduced prior to detection makes possible the identification and separation of positive and negative Doppler shifts (Mcleod, 1967). Such a phase shift changes

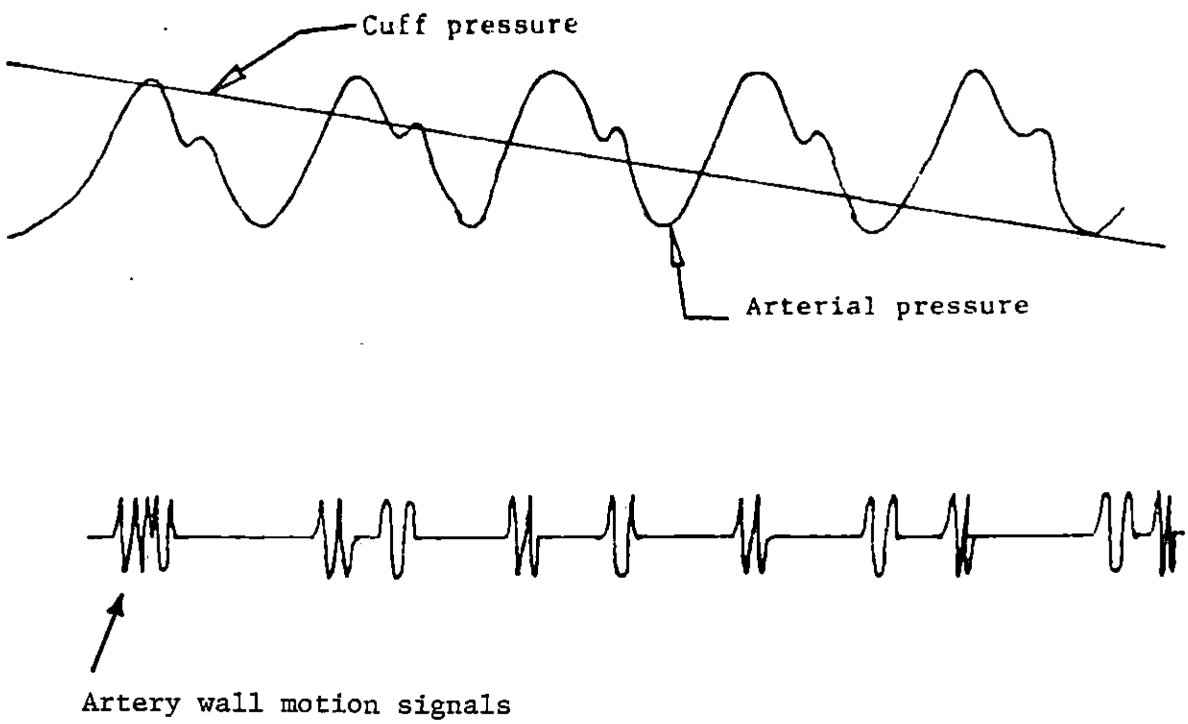


Figure 2. Conventional Doppler output

sign when flow reverses and is produced by the addition of carrier components (Direction Doppler Demodulation). Extracted audible Doppler shift frequencies corresponding to motion towards the transducer appear at one output; those for the motion away from the transducer at another (Rittgers et al., 1980). Output signals from directional Doppler instruments can be presented or analyzed in the same way as the Doppler output signals from simple nondirectional devices, by means of earphones, chart recorders, etc.

In this case, the movement of the arterial wall towards the transducer and away from it could be distinguished from one another. Designating the wall motion signal towards the transducer as forward, while that away from it as reverse, the directional Doppler generates an extra channel of information. Systolic and diastolic pressure points have been characterized by a sequence of opening and closing signals which are the same as the motion of the arterial wall towards and away from the transducer, respectively. Thus, the forward and reverse signals could be used to identify systolic and diastolic pressure points. It should be noted that no frequency information is necessary for the determination of systolic and diastolic pressure values.

In this project, the conventional Doppler ultrasound method was used to determine systolic and diastolic blood pressures. 2MHz was the transmitted frequency and reflected signals were in the ranges 50 to 125 Hz and 8 to 25 Hz for opening and closing motions of the arterial wall, respectively. The brachial artery was subjected to occlusion. This vessel was because it is accessible and could be subjected to occlusion without difficulty, and since it is on level with the heart, any measurement errors due to hydrostatic effects were eliminated. Since, systolic and diastolic pressures are related to a specific sequence of the arterial wall opening and closing signals, they could be deduced from the frequency analysis of the received signals.

When a beam of ultrasonic energy is directed at a moving surface, part of the sound waves are reflected back. The reflected beam is phase modulated in proportion to the artery wall displacement and frequency modulated in proportion to the artery wall velocity. In practical situations, however, the receiving transducer sees the directly reflected energy combined with multipath reflections caused by frequency distortions. In addition, a certain amount of unmodulated energy is added as well. Thus, the received signal is a combination of the carrier frequency (transmitted frequency) and the Doppler shift frequencies. Since the artery motion is the fastest motion in the arm, slower signals from bone motion and skin reflection can be eliminated by inserting a filter of proper bandwidth in the radio frequency (RF) amplifier of the receiver. These signals are then amplified and limited to produce a constant output that is independent of the input amplitude. This signal processing technique provides uniform amplitude artery motion signals in the presence of variable amounts of crosstalk signals from bones or transducer.

In the method used, a 2.2 MHz oscillator/transmitter drives a piezoelectric crystal transducer to produce the ultrasound. Carrier and Doppler shifted signals are detected by a diode and amplified through low frequency lowpass filters. A diode acts as a rectifier and if there is any audio frequency information present on the RF signal (in the form of amplitude modulation), the detector functions to convert the audio voltages into pulsating dc voltages. The most commonly used circuit is the envelope detector and it produces an output proportional to the envelope of the input wave. The detector consists of a diode, a resistor, and a capacitor. During the positive cycle of the input voltage, the diode acts as an "ON" switch, allowing the capacitor to charge to the peak RF input. During the negative half of the RF cycle, the diode is "OFF", but the capacitor holds the positive charges received, so the output voltage remains at the peak positive value of the RF signal. A resistor provides the discharge path for the capacitor during the downward swing in modulation. The amplitude modulated signal on being passed through lowpass filters provides the frequency spectrum related only to the opening and closing signals of the arterial wall.

2.2 MHz was chosen as the operating frequency for two reasons. First, the operating frequency is a function of the average artery diameter and for a 4 to 5 mm diameter artery, the suitable frequencies have been predicted to be between 2 to 10 MHz. Next, Hartley and Strandness (1969) have shown that the insertion loss increased with frequency from 2 to 10 MHz in both normal and diseased areas. Conventional Doppler method was chosen for circuit simplicity. To reduce the loading effects of frequency dependent RC network impedances, a FET input operational amplifier could be used.

The output from the detector, after filtering, is fed to a Schmitt trigger to get a square wave. This signal is rectified to cut off the negative cycle and is used to trigger a monostable multivibrator. Pulses of constant width and amplitude (between 0 and 5 V) are provided by the one-shot. The final digital circuitry is CMOS and serves to couple the signal processing stage to a microcomputer.

DOPPLER SIGNAL ACQUISITION SYSTEM

Electronic Circuitry

A block diagram of the Doppler signal acquisition system is as shown in Figure 3. The oscillator drives the transducer crystal. The received modulated signal (a mixture of RF and audio signals) is amplified, demodulated and fed to an audio amplifier (2 stages) with low pass characteristics. This audio output is fed to a Schmitt trigger an The subsequently to a buffer. The audio output is also fed to a speaker. Two sequential timers control the inflation and deflation durations of the cuff. A pump and a valve control the flow of pressurized air to the cuff. One of the timers sets the cuff inflation period by enabling power to the pump, while the second brings about the deflation of the cuff and also provides for a break period before starting yet another cuff cycle.

For the description in this thesis, the system will be divided into a cuff inflation and deflation section, an RF section, an audio amplifier and Schmitt trigger section, and a digital section. Each of these sections can be further subdivided in order to describe the individual circuits.

Cuff Inflation and Deflation Circuits

The principal design consideration was to have the cuff inflate and deflate automatically, without any operator assistance. In a broader perspective, this would allow it to be incorporated into an ambulatory

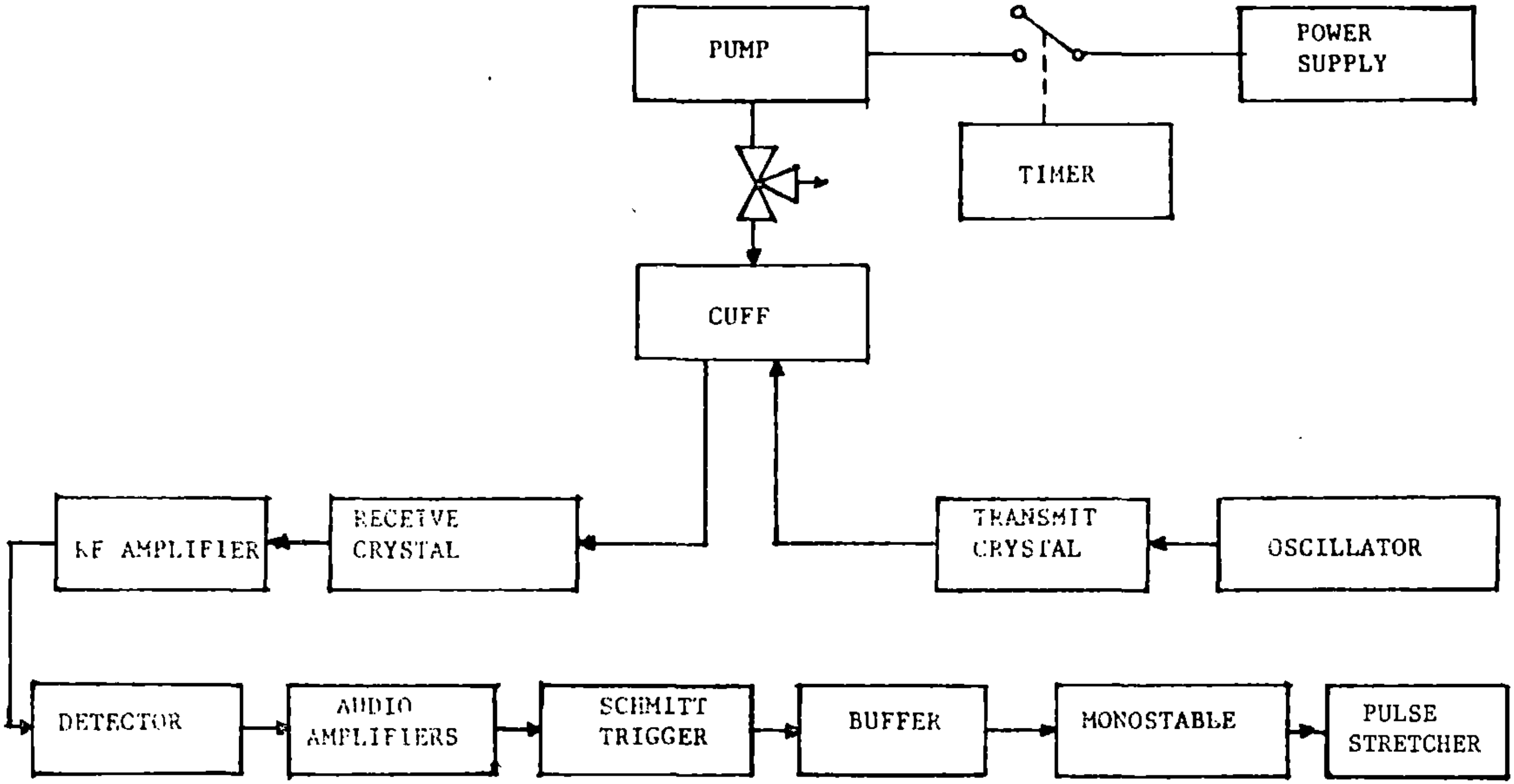


Figure 3. Doppler signal acquisition system block diagram

blood pressure monitor.

The system was designed to have three events in each cycle. First, a short inflation period, followed by a longer deflation period and finally a long break before starting next cycle. The above mentioned features were achieved through two timers, a valve, and a pump.

The two timers were connected in sequence and were set to run for 10 seconds and 9 minutes, respectively. A valve controlled the flow of pressurized air from the pump to the cuff. One of the ports of the valve was exposed to atmospheric pressure, thus providing for a constant bleed. A pump supplied pressurized air at a constant rate.

A circuit diagram of the timers is as shown in Figure 4a. At the start, timer IC1 is triggered and remains HIGH for t1 seconds. During this supply to the pump is enabled and this causes the cuff to inflate. The duration of t1 seconds is set such that at the end of t1 seconds, the cuff is inflated to a pressure well above systolic pressure. Timer IC2 remains LOW during this period. At the end of t1 seconds, timer IC1 goes goes LOW and triggers timer IC2 to go HIGH. This turns off the pump and and stops inflation of cuff. The cuff deflation operation is started. The cuff needs to be deflated at a rate of 2 to 3 mmHg/sec for any transducer to detect the changes associated with the blood flow in the artery. IC2 is HIGH long enough to achieve this deflation rate and in addition, provide the necessary break before starting next cycle. At the end of interval t2, timer IC2 goes LOW and triggers timer IC1 to go HIGH, restoring power to the pump and starting next cuff inflation cycle.

Figure 4. Timer to control cuff inflation and deflation

a) Circuit diagram

b) Photograph of pump with valve

R	K(ohms)	C	(Farads)
1	47	1	1000p
2	.150	2	.01m
3	.150	3	20m
4	680	4	1000p
5	47	5	.01m
6	.150	6	50m
7	6800	7	50m

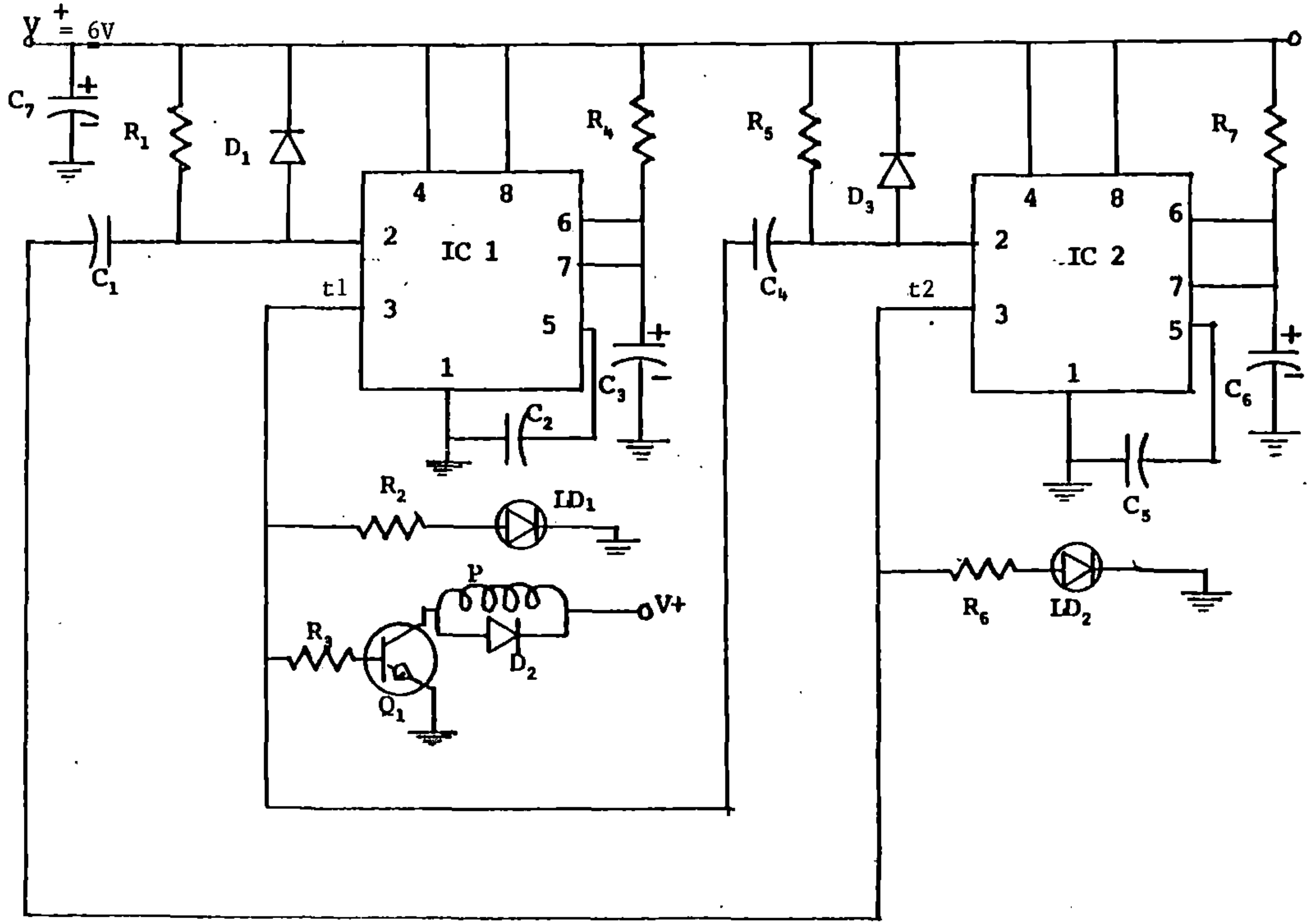
D
1,2,3 1N4148

LD
1,2 Light emitting diodes

Q
1 D44C3

IC
1,2 555 Micropower timers

P
Romega 080 Metering Gas Pump


V + = 6V
C7
R1
D1
IC 1
4
8
6
7
2
3
5
1
t1
C1
C2
C3
R4
R2
LD1
P
V+
R3
Q1
D2
C4
R5
D3
IC 2
t2
C5
C6
R7
R6
LD2

(a)

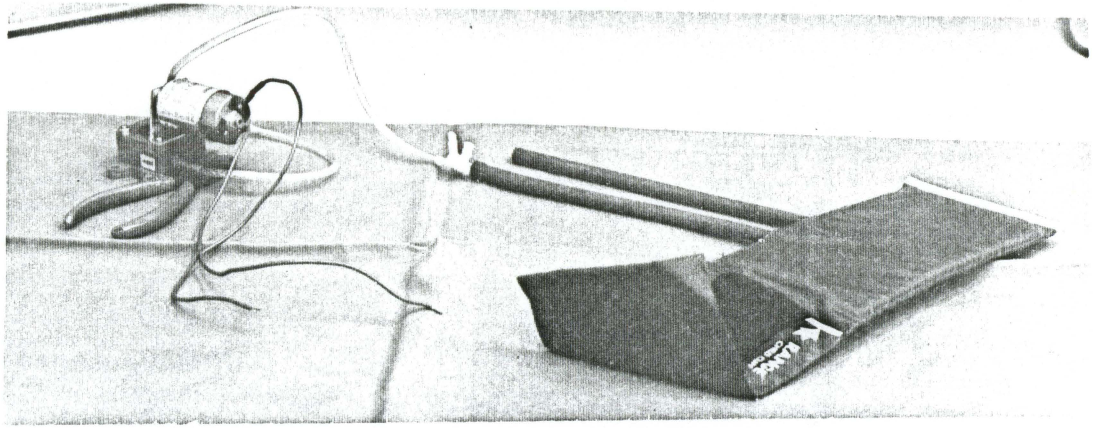


Figure 4. Continued (b)

R4-C3 provide the timing interval for timer IC1 and R7-C6 do the same for timer IC2. LD1 and LD2 serve as indicators for testing the durations t1 and t2, respectively. The valve, with its bleed port, helps the cuff inflation and deflation operation by providing a constant deflation rate. It is a mechanical valve not requiring any power. Pump P, which is very small, requiring 1.2 W for its operation. Timers are CMOS (Exar Integrated Systems, 1978) and hence consumed very little power. A power transistor Q1 switches the power supply to the pump. The design is directed towards battery powered operation and compactness.

Physical dimensions of the pump are such that the pump could be incorporated as part of an ambulatory blood pressure monitor. The pump, Romega model Ø8Ø manufactured by Romega Corporation (W. Germany), weighs 22Ø grams and works on a +1Ø V power supply. Although the current drawn by the pump is 12Ø mA, it could be operated on a +1Ø V battery since the pump is switched on for a small fraction of time (1Ø seconds in every 1Ø minutes). A soundproof casing made around the pump could eliminate the discomfort caused by vibrations and noise. The characteristics of the pump are ideally suited for being part of an ambulatory blood pressure monitor with automatic inflation and deflation of the cuff.

The bleed port of the valve is manually adjusted and is set prior to the measurement procedure. It is such as to provide a rapid inflation rate and a slow deflation rate (2 to 3 mmHg per second). A few trials with inflation and deflation of the cuff will enable the correct valve setting to be achieved. Thus, once set at the start, the valve need not be readjusted. A guideline for setting the valve is that during inflation the cuff pressure should rise to 180 to 190 mmHg before timer IC1 goes low and that during deflation the cuff pressure should decrease at the rate of 2 to 3 mmHg every second.

Figure 4b shows the photograph of the cuff inflation and deflation system with the pump and the valve.

RF Circuits

This section describes the oscillator, the receiver/RF amplifier, and the detector circuits.

Oscillator

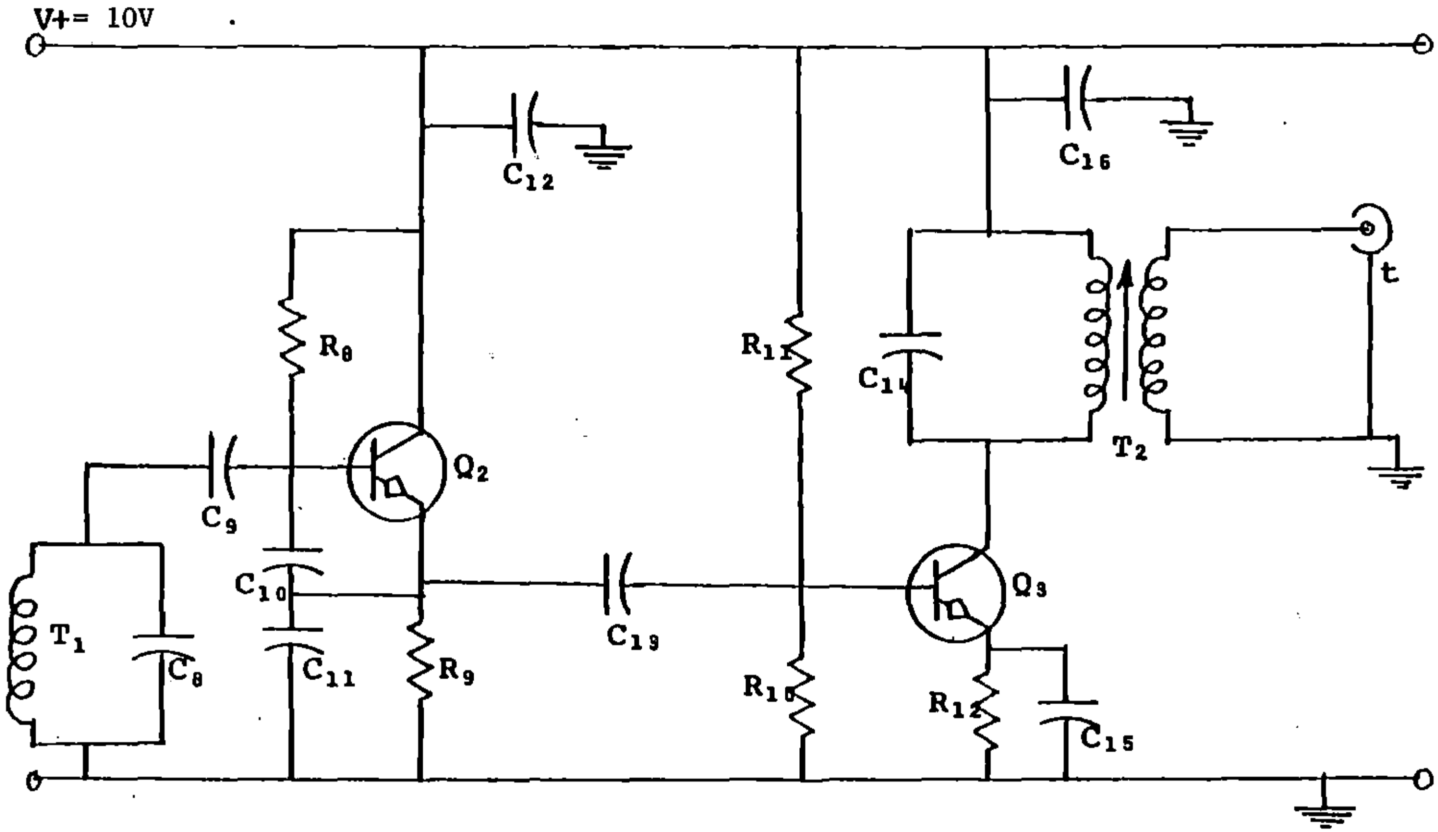
The Oscillator consists of a Colpitts oscillator (Carr, 1978) and a Class C amplifier (Lenk, 1977). A circuit diagram is as shown in Figure 5.

A Colpitts oscillator uses a tapped capacitor voltage divider to form the feedback path (C10 and C11 in this case). The tank circuit consists of a variable inductance T1 and capacitors C8, C9, C10, and C11. Capacitor C8 and inductance T1 are designed to resonate at the resonant frequency of the crystal transducer (2.2 MHz in this case). Capacitor C9 serves to couple the ac voltage developed across the tank circuit to the base of the transistor Q2. This ac voltage provides the necessary bias across the base emitter junction of Q2. Capacitors C10 and C11 give the positive feedback path between base and emitter of Q2, necessary to sustain oscillations. In general, capacitor C10 is much larger than C11. Capacitor C12 isolates the power supply from the RF signals.

Figure 5. Oscillator

R	K(ohms)	C	(Farads)	Q	
8	68	8	770p	2	2N4410
9	.780	9	.1m	3	2N3904
10	47	10	1680p		
11	47	11	568p	T	
12	.400	12	.05m	1	Primary, 21 turns
		13	.1m	2	Secondary, 5 turns
		14	800p		Primary, 21 turns
		15	.01m		
		16	.1m	t	Transmit crystals of the transducer

T1 and T2 wound on Ferrite cores.


V+= 10V
C_{12}
C_{16}
R_8
R_{11}
C_{14}
t
Q_2
C_9
T_2
C_{10}
T_1
C_8
C_{11}
R_9
C_{13}
Q_3
R_{10}
R_{12}
C_{15}

The Colpitts oscillator was chosen due to its greater stability at high frequencies. In this circuit, the total capacitance is made up of capacitors C8 through C11, inclusive, but mechanical and thermal effects can be limited to a single capacitor C8. If C8 is only a small portion of the overall capacitance creating the resonant condition, then changes in C8 due to thermal problems have but a small effect on the frequency of oscillation.

Output from the emitter of Q2 is coupled to the base of Q3 through a capacitor C13. Resistor R11 provides the pull up for the base of Q3, while R1Ø and R12 together provide the necessary current gain. Capacitor C14 and the inductance in the primary of T2 are chosen similar to T1 and C8. Transmit crystals are connected to the secondary of transformer T2 by coaxial cables. The number of turns in the secondary of T2 controls the power power to the transmit crystals.

A Class C amplifier offers greater stability, but a lesser gain and lower input impedance than the Class A and Class B amplifiers. One major advantage in using the Class C amplifier is that the circuit characteristics such as voltage and current gain, as well as the input and output impedances, are dependent upon circuit values and not on the transistor beta. Although nonlinear, the Class C amplifier provides a sine wave output because of the resonant action of the output tank circuit. Finally, the efficiency of the Class C amplifier, in terms of power output, is very high.

Receiver/RFamplifier

This section describes a receiver and an RF amplifier. See Figure 6. The receiver is a step-up transformer T3 receiving the reflected signals from the receiver crystals. Capacitor C17 and the inductance in the primary of T3 are chosen to resonate at the resonant frequency of the transducer crystals. This is true for capacitor C18 and the inductance in the secondary of T3 as well. The receiver crystals of the transducer are connected by coaxial cables to primary of the input step-up transformer. The output from the receiver is modulated and has audio frequencies on the carrier radio frequency and the signal amplitude is of the order of millivolts. Receiver output is presented to the RF amplifier.

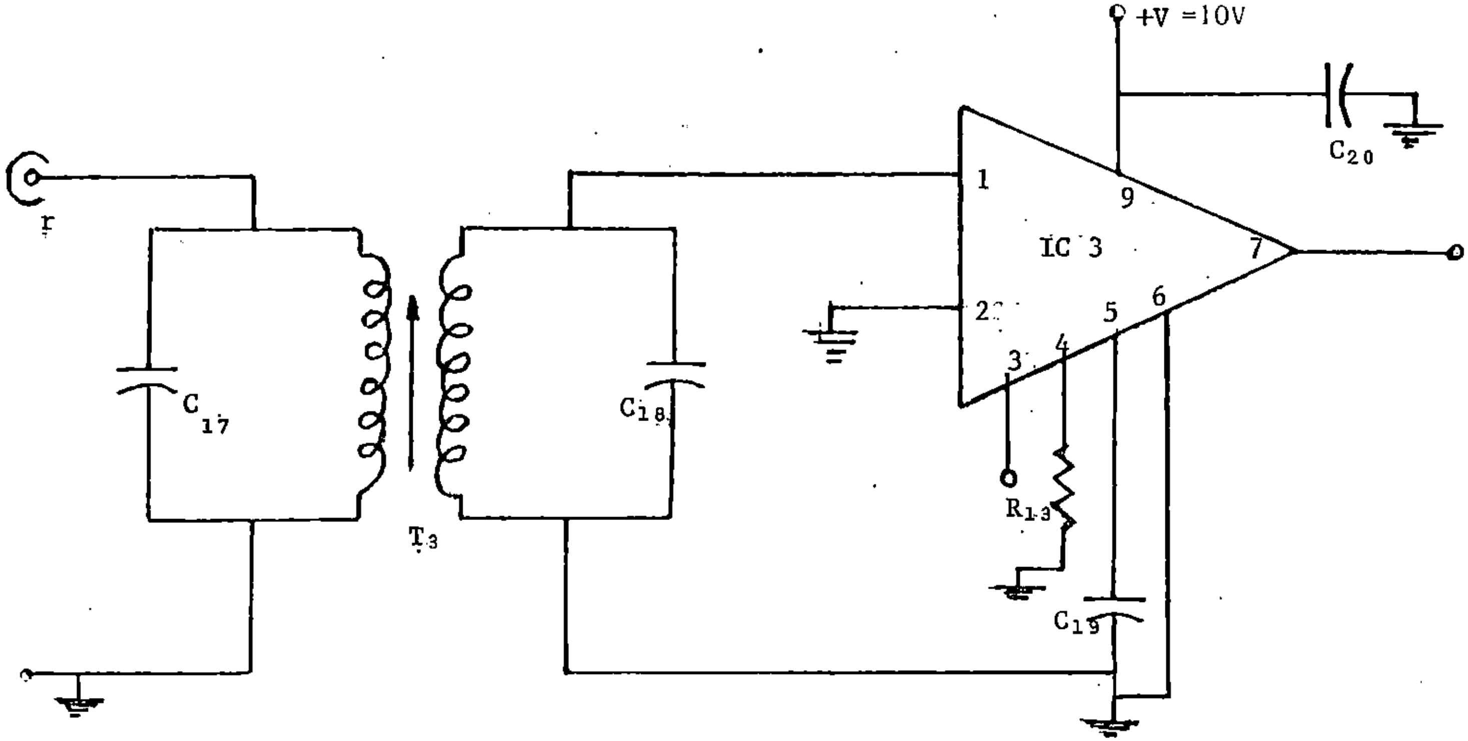
The RF amplifier (a video amplifier) has a gain of 34dB at a frequnecy of 40 MHz. Resistor R13 between pin 4 and ground determines the gain of the amplifier and capacitor C19 to ground from pin 5 determines frequency response of the amplifier.

The output from the amplifier is presented to the detector circuits. This stage provides a gain of 50 to 100 and sufficiently raises the level of the signal reaching the detector circuit.

Detector

This circuit helps to recover the information signal in the envelope from the modulated wave. The envelope detector provides an output voltage proportional to the envelope of the input wave. A circuit diagram is as shown in Figure 7a.

Diode D4 acts as a rectifier, demodulating any audio information



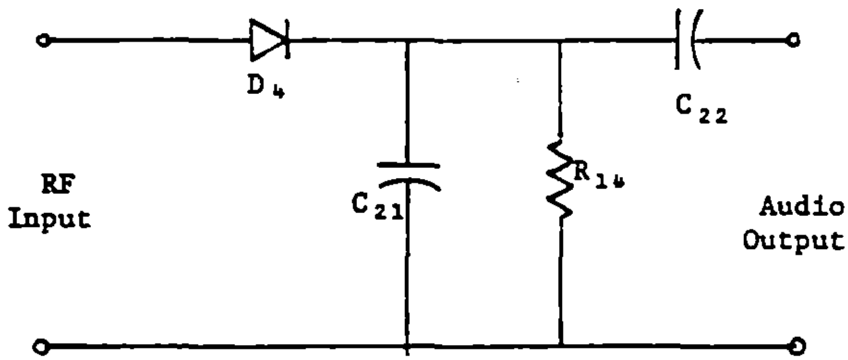
R	K(ohms)	C	(Farads)	IC	
13	.05	17	.004m	3	MC1552G Video Amplifier
		18	800p		
		19	.033p	T3	Primary, 5 turns Secondary, 21 turns Both wound on Ferrite cores
		20	.1m		
				r	Receive crystals of the transducer

Figure 6. Receiver/RF amplifier

present in the RF signal. Capacitor C21 charges to the peak value detected by the diode and resistor R14 provides the discharge path for the capacitor. Capacitor C22 helps to remove any dc offset present on the audio signals to be presented to the following audio circuits.

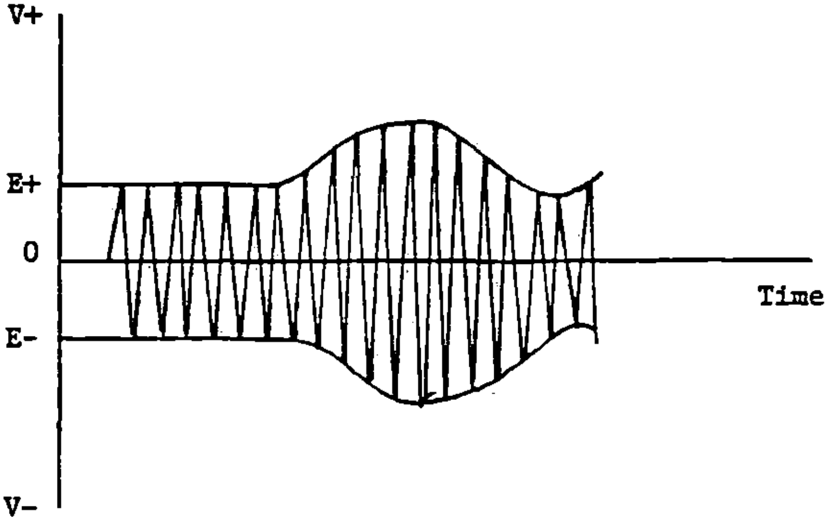
The signal received from the artery under occlusion is frequency modulated and consists of audio frequencies superimposed on the carrier RF signal. The diode detector is basically a rectifier, so its output is a dc signal that varies at the same rate and in the same manner as the modulating signal. Capacitor C21 is used to remove the RF variations that form the diode's output pulsations. Diode D4 acts as an "ON" switch when the input voltage is positive, allowing the capacitor to charge up to the RF input. During the negative half of the cycle, the diode is "OFF", but the capacitor holds the charge already received, so that the output voltage remains at the peak positive RF value. Resistor R14 enables fast discharge of the capacitor during the downward swing of the modulated signal. The capacitor is given a value that is large enough to filter out the RF signal, but is not high enough to attenuate the audio signals. The modulated input waveform and the demodulated output waveform are shown in Figure 7 (b and c).

Design considerations involved in deciding the component values are important (Roddy and Coolen, 1977). Typically, the voltage at the input to the detector is of the order of 1 to 2 V; therefore, the maximum reverse voltage rating of the diode need not exceed 2 V. Also, because of the low signal voltage available, a low forward voltage drop is desired for the diode. Finally, the current rating of the diode need not exceed 1 to 2 mA.



R	K(ohms)	C	(Farads)	D	
14	150	21	.1m	4	1N4148
		22	10m		

(a)



(b)

Figure 7. Detector
a) Circuit diagram
b) Modulated input
c) Output waveform

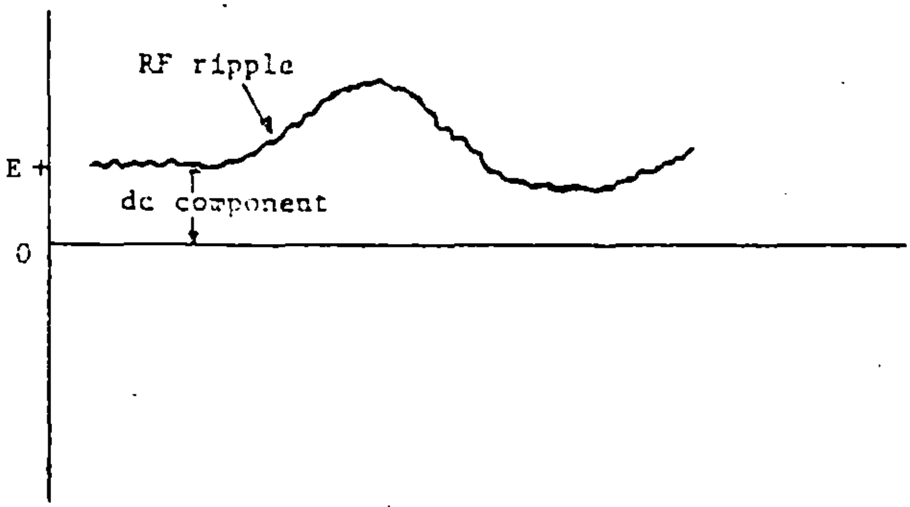
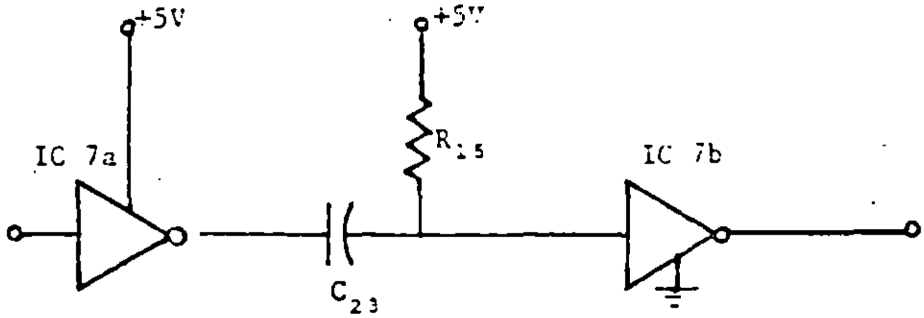


Figure 7. Continued (c)



R	K(ohms)	C	(Farads)	IC	
15	10	23	.01m	7a,7b	4049 Hex Inverting Buffer

Figure 8. Digital circuits

The capacitor should be large enough to provide little reactance at RF signal frequency, but small enough to provide a short RC time constant so that the audio signals will fully charge the capacitor. Typically, reactance of the capacitor should be around 50 ohms and the RC time constant should not be greater than 250 us. Ideally, the value of the resistance should match the impedance of the transformer secondary. At the same time, the resistor acts as the output load impedance of the detector and hence, should match the input impedance of the following audio stage. A trade-off is made between these two levels, not forgetting the restriction imposed by the requirement that the RC time constant does not exceed 250 us.

Output from the detector (amplified) can be fed to a speaker. This can be used to troubleshoot system. Figure 8 will be discussed later.

Audio Amplifiers and Schmitt Trigger

This section describes the two audio amplifier stages and the Schmitt trigger (Coughlin and Driscoll, 1982). Figure 9 gives the circuit diagram for the audio amplifiers and the Schmitt trigger.

The audio amplifier consists of two stages, each providing gain and frequency attenuation. The first stage provides a gain of 50 and a high frequency cutoff of 150 Hz. The second audio amplifier stage has a gain of 50 and high frequency cutoff of 150 Hz. These cutoff frequencies were chosen because the opening and closing signals from the arterial wall lie

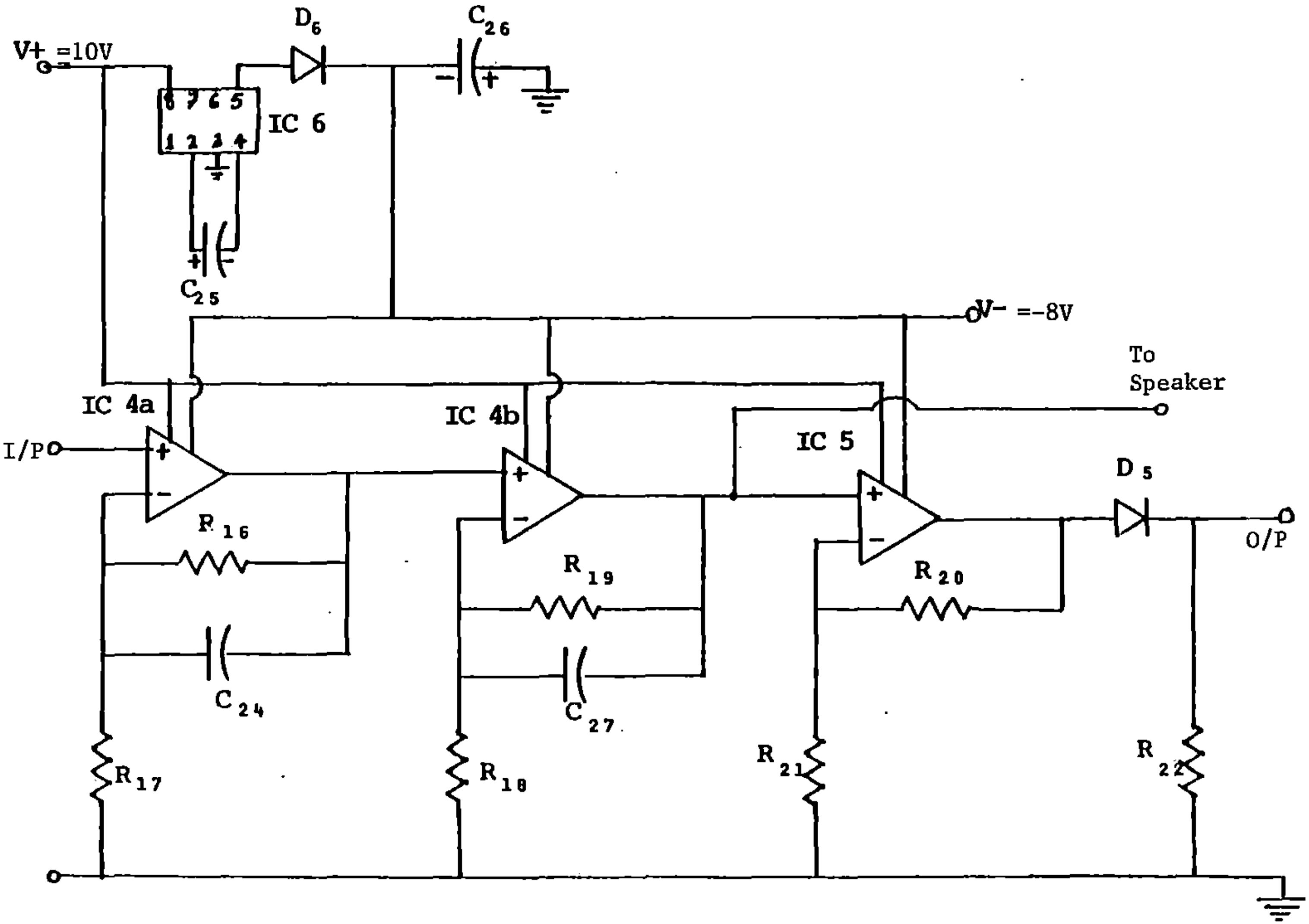
Figure 9. Audio amplifiers and Schmitt trigger

R	K(ohms)
16	1500
17	.470
18	.470
19	1500
20	6.8
21	6.8
22	6.8

C	(Farads)
24	470p
25	10m
26	10m
27	470p

D	
5,6	1N4148

IC		
4a,4b	MC1458	Dual Operational Amplifier
5	LM348	Quad Operational Amplifier
6	ICL 7660	Voltage Converter



V+ =10V
D6
C26
IC 6
C25
V- =-8V
To Speaker
IC 4a
IC 4b
IC 5
I/P
D5
O/P
R16
R19
R20
C24
C27
R17
R18
R21
R22

in the region of 0-150 Hz for a 2.2 MHz carrier frequency. The two amplifier stages provide sufficient gain to saturate the Schmitt trigger.

The output of the audio amplifier is fed to a speaker. Speaker output is used to calibrate the cuff pressure. Digital output from the microcomputer is verified against the sounds heard over the speaker.

The Schmitt trigger (IC5) trip points are set to +/- 5 V and the output from the previous stage is sufficiently high to trip the Schmitt trigger diode D5 clips off the negative portion of the signal before it is fed to the buffer inverter.

Digital Circuits

A circuit diagram for a buffer/inverter is as shown in Figure 8. The circuits are CMOS. The first stage of the buffer inverter helps to level translate the Schmitt trigger output to a 0 and +5 V compatible signal. The next stage acts as a one-shot to provide pulses of 0.1 ms duration. Finally, the output in the form of pulses between 0 and 5 V can serve as inputs to a microcomputer.

A pulse stretcher circuit helps to increase the width of the pulses before being input to the microcomputer. This is done in order to enable the software (used for measuring the interval between pulses) sufficient time between the rise and fall of the pulses.

DOPPLER SIGNAL PROCESSING SYSTEM

The Microcomputer

Taking into view the features expected of a microcomputer to be incorporated into an ambulatory monitor, the Zilog Incorporated single-chip microcomputer Z8 (referred as Z8671 MCU) was chosen. The architectural features of the Z8671 MCU offered easy programming and interfacing to external circuits.

The Z8671 MCU is an 8-bit microcomputer with an on-board BASIC interpreter in a 2K ROM (read only memory) and an on-board user memory in a 2K RAM (read/write memory). The Z8671 MCU architecture allows it to serve in either memory or input/output intensive applications.

The Z8671 MCU contains 2K bytes of ROM, 32 input/output lines and 144 bytes of random access memory. The internal RAM is a register file composed of 124 general purpose registers, 1 serial input/output register, 15 status control registers, and 4 input/output registers. A conventional terminal consisiting of a keyboard and a CRT (cathode ray tube) is linked to the microcomputer.

As many as 32 pins of the 40 pin package are input/output related. The 32 input/output lines are grouped into 4 ports and treated internally as 4 registers. They are software configurable for either input or output. While ports 0 and 1 serve as multiplexed data/address bus, ports 2 and 3 can be used for input/output applications. Port 3 under program control, can be configured as special control lines for handshake and port 2 can be programmed as input to accept data, or as output to transmit data. Serial data are received through bit Ø of port 3 and transmitted through bit 7 of port 3.

The Z8671 MCU supports 15 BASIC commands. In addition, it has 47 assembly language instructions, 9 addressing modes, and 6 maskable interrupts. The BASIC/DEBUG interpreter on the Z8671 MCU allows machine language programs to be written and executed through subroutine calls in in BASIC. Two parameters can be passed to the subroutine through the subroutine call and the subroutine can return a value to be used by the calling BASIC program. This provision greatly enhances the capabilites of the microcomputer allows interfacing to high speed peripherals.

Finally, the Z8671 MCU provides two on-chip 8-bit counter/timers. Both counter/timers are independent of the processor instruction sequence, which relieves software from time-critical operations such as interval timing or event counting. Counting mode and starting/stopping sequences of the counter/timers can be under software control. For longer time intervals, the two counter/timers can be cascaded as a single unit.

In this thesis, the Z8671 MCU is configured for input/output intensive applications. Ports Ø and 1 are retained as a multiplexed data/address bus. Port 2 is configured as input to accept digital data and port 3 is configured for general purpose operation. Some of the port 3 lines interface with the A/D converter, while two others control the cuff inflation timer and accept digital pulses from the signal acquisition system.

For the most part, BASIC is used for programming and machine language routines are used only for faster applications. Appendices B, C, and D provide a listing for the software used.

Electronic Circuitry

A block diagram of the Doppler signal processing system is as shown in Figure 10. A pressure transducer converts the cuff pressure into an equivalent output in volts. The voltage output of the pressure transducer is converted into digital values in hexadecimal code by the A/D converter. Digitized pressure values are then input to the microcomputer.

The microcomputer controls the operation of the A/D converter by providing the start convert, address latch enable, and output enable signals. The timer, responsible for providing the cuff inflation duration, is also controlled by the microcomputer. Output of the signal acquisition system is also input to the microcomputer.

Each component of the system block diagram will be described in detail in the following sections.

Pressure Transducer and Amplifiers

The Pressure transducer and amplifiers circuit diagrams are as shown in Figure 11.

The transducer is a monolithic gage pressure transducer (LX0603GB) made by National Semiconductor Corporation. Monolithic pressure

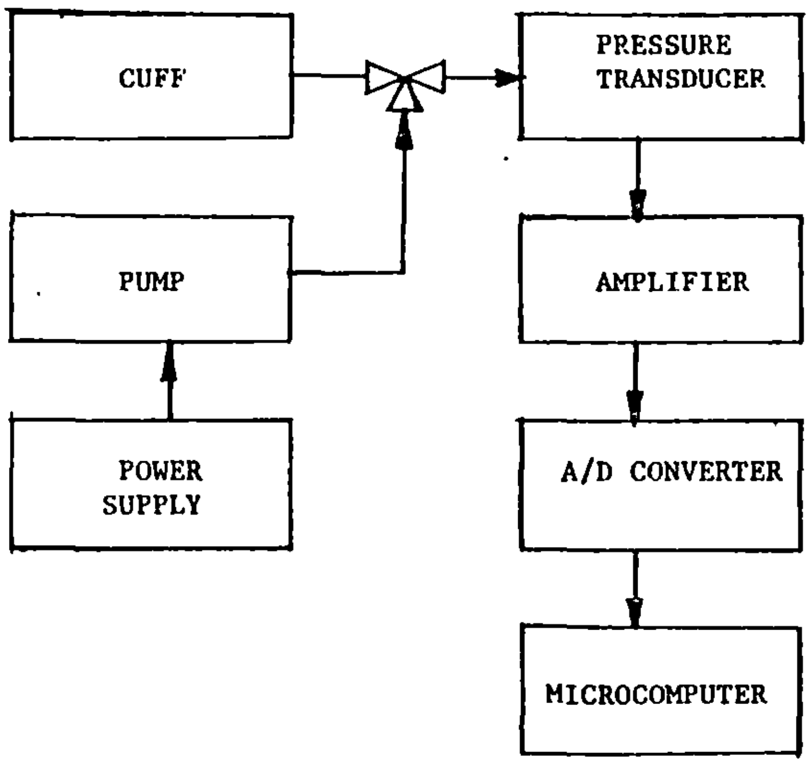


Figure 10. Doppler signal processing system block diagram

transducers are piezoresistive integrated circuits which provide an output voltage proportional to the applied pressure. These devices are temperature compensated with respect to sensitivity and are designed to be very useful in applications requiring battery power or compatibility with microprocessors. Their low cost per unit also reduces the design cost.

In the LXØ6Ø3GB pressure transducer, pin 5 is the supply pin, pin 2 is grounded, and pins 3 and 4 are the outputs. Transducer output signals, pins 3 and 4, are directly taken from a Wheatstone bridge. Pin 3 is the positive signal output and goes positive when pressure is increased. Pin 4 is the negative signal output going negative when pressure is increased. Transducer output changes by 2 mV for every psi (pounds per square inch) change in applied pressure. The applied pressure, in this case, is the cuff pressure varying between Ø and 2ØØ mmHg (Ø to 4 psi). Thus the transducer output requires amplification. Also, the A/D converter, further down, works on a Ø to 5 V input and therefore, the transducer output needs to be scaled.

The amplifier consists of two stages: one for removing the transducer offset and the other for amplifying the transducer output. The principal problem encountered in calibration and scaling of a transducer is the interaction between offset and span parameters. By making the offset and span independent of each other, easy calibration and simple scaling can be achieved. The use of two amplifier stages effectively reduces the offset-span interaction.

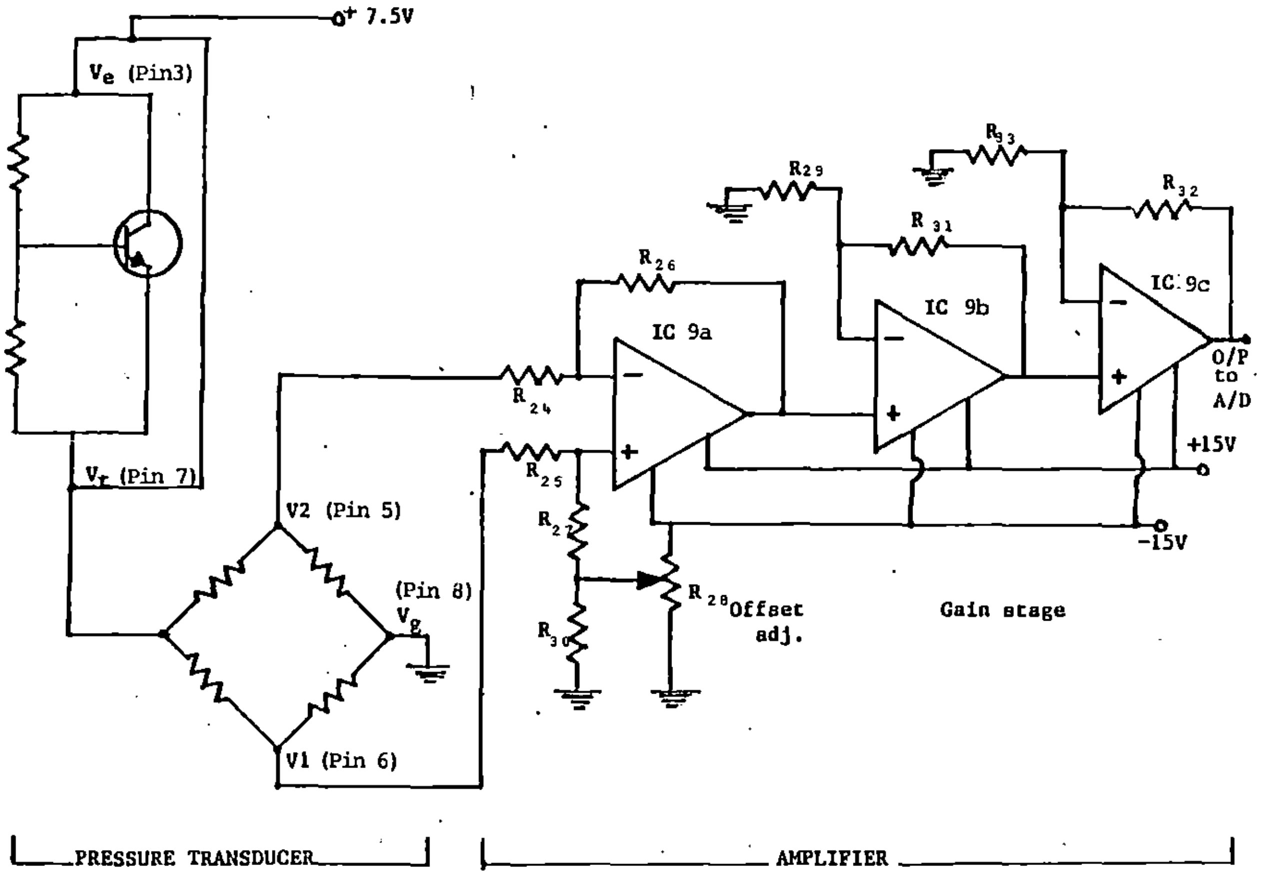
In Figure 11, IC9a provides the offset stage with R28 being the offset adjustment potentiometer. IC9b and IC9c form the gain stages providing

Figure 11. Pressure transducer and amplifiers

R	K(Ohms)
24	22
25	22
26	330
27	330
28	10
29	1.2
30	22
31	22
32	270
33	2.7

IC	
9a, 9b, 9c	LM324 Low Power Operational Amplifier

PT	
LX0603GB	Monolithic Gage Pressure Transducer



+ 7.5V
V_e (Pin3)
V_r (Pin 7)
V2 (Pin 5)
(Pin 8)
V_g
V1 (Pin 6)
R_{24}
R_{25}
R_{26}
R_{27}
R_{28}
R_{29}
R_{30}
R_{31}
R_{32}
R_{33}
IC 9a
IC 9b
IC 9c
O/P to A/D
+15V
-15V
Offset adj.
Gain stage
PRESSURE TRANSDUCER
AMPLIFIER

gains of 19 and 13, respectively. Under this configuration, the pressure transducer provided a linear voltage output for changing input cuff pressures. Transducer output voltage levels (after amplification) for changing cuff pressures is as shown in Figure 12. As can be seen from Figure 12, transducer output varied between Ø and 5.Ø8 V for cuff pressures between Ø and 19Ø mmHg.

Figure 13 shows a plot of the pressure transducer output voltage level versus the cuff pressure. As can be seen, the plot is a straight line indicating a linear transducer response. The three trials are results of pressure measurements performed by three different individuals.

Analog to Digital Conversion

The analog-to-digital converter interface to the microcomputer is as shown in Figure 14.

The A/D converter is the ADCØ8Ø8 chip made by National Semiconductor Corporation. The ADCØ8Ø8 converter is a monolithic CMOS device with an 8-bit A/D converter, 8-channel multiplexer, and microprocessor compatible control logic. The converter works on a single 5 V power supply and has a conversion time of 1ØØ us. Easy interfacing to microprocessors is provided by the latched and decoded multiplexer address inputs and latched TTL outputs.

The pressure transducer output is fed to input channel Ø, selected by grounding address pins 23, 24, and 25. This address is latched into the decoder on the low-to-high transition of the address latch enable signal

Cuff Pressure (mmHg)	Voltage (Volts) Trial 1	Trial 2	Trial 3
190	5.08	5.07	5.06
180	4.75	4.80	4.76
170	4.49	4.50	4.55
160	4.21	4.23	4.23
150	3.95	3.90	3.98
140	3.69	3.73	3.72
130	3.39	3.46	3.46
120	3.16	3.18	3.17
110	2.90	2.89	2.88
100	2.67	2.65	2.65
90	2.37	2.36	2.38
80	2.11	2.11	2.16
70	1.79	1.84	1.88
60	1.47	1.56	1.51
50	1.25	1.29	1.27
40	0.80	1.04	1.01
30	0.63	0.74	0.76
20	0.50	0.51	0.51
10	0.20	0.22	0.21
0	0	0	0

Figure 12. Pressure transducer output

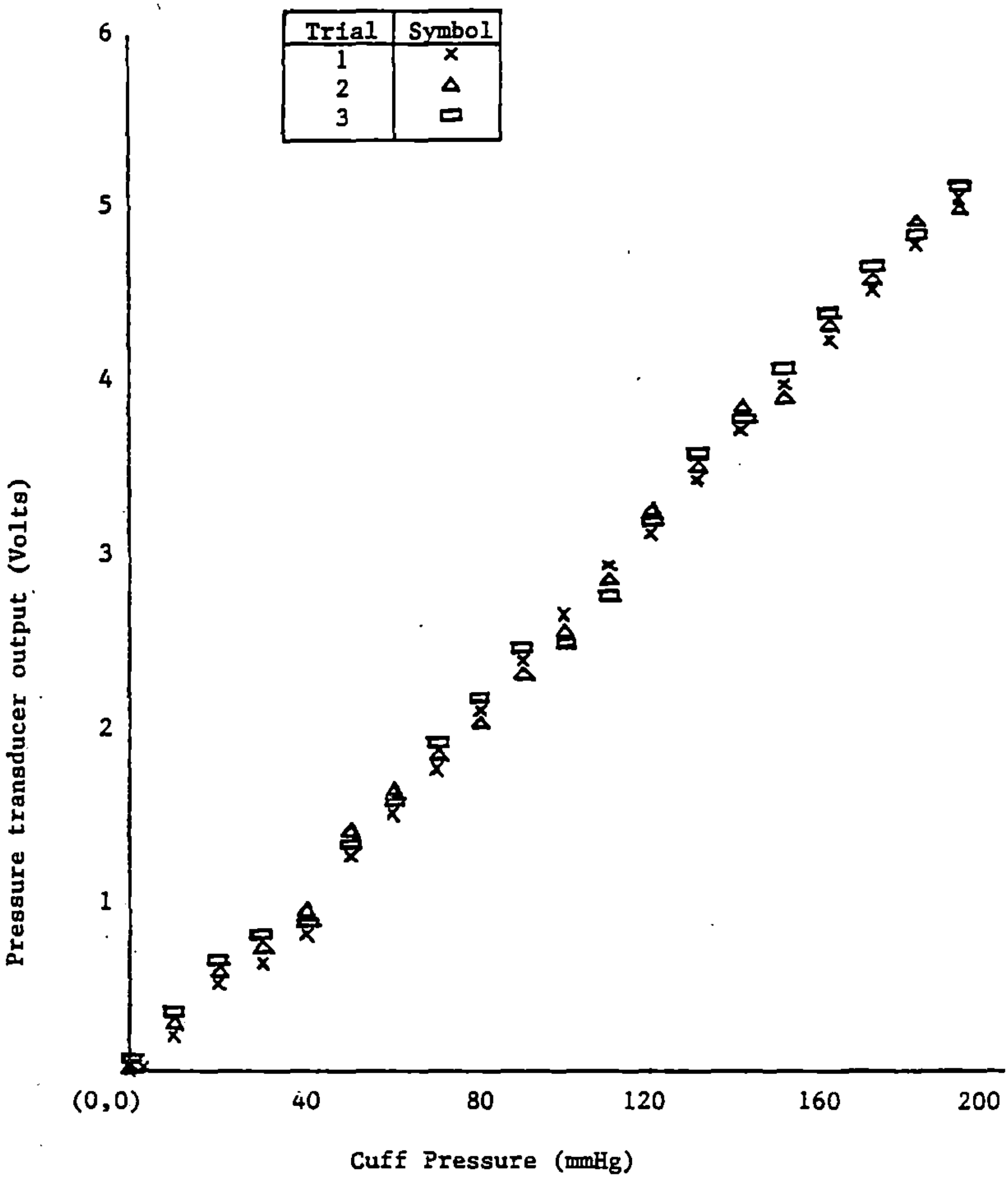


Figure 13. Pressure transducer output versus cuff pressure

(ALE). The converter is interfaced to the microcomputer through ports 2 and 3 of the microcomputer.

Port 2 bits Ø through 7 serve to input the digital output of the converter to the microcomputer. Control signals are provided by port 3 lines. Port 3 bit 3 senses the end of conversion signal (EOC), bit 4 provides the start conversion (START) and address latch enable (ALE) signals, and bit 5 provides the output enable (OE) signal. The converter clock frequency is 5ØØ KHz and is obtained from an external oscillator.

The A/D converter output changes by a bit for every 2Ø mV change in the input signal. The A/D converter output (in hexadecimal code) for different cuff pressures is as shown in Figure 15. Data were recorded through three trials performed by three different individuals. A plot of the digital output of the A/D converter versus the cuff pressure for three different trials is as shown in Figure 16.

Appendix F gives a listing for the BASIC program and the machine language subroutine called by the BASIC program. The programs are responsible for performing analog-to-digital conversion and for outputting the converted digital data to the microcomputer. The machine language subroutine starts at location hex ØAØØ. The BASIC program starts at location hex Ø8ØØ.

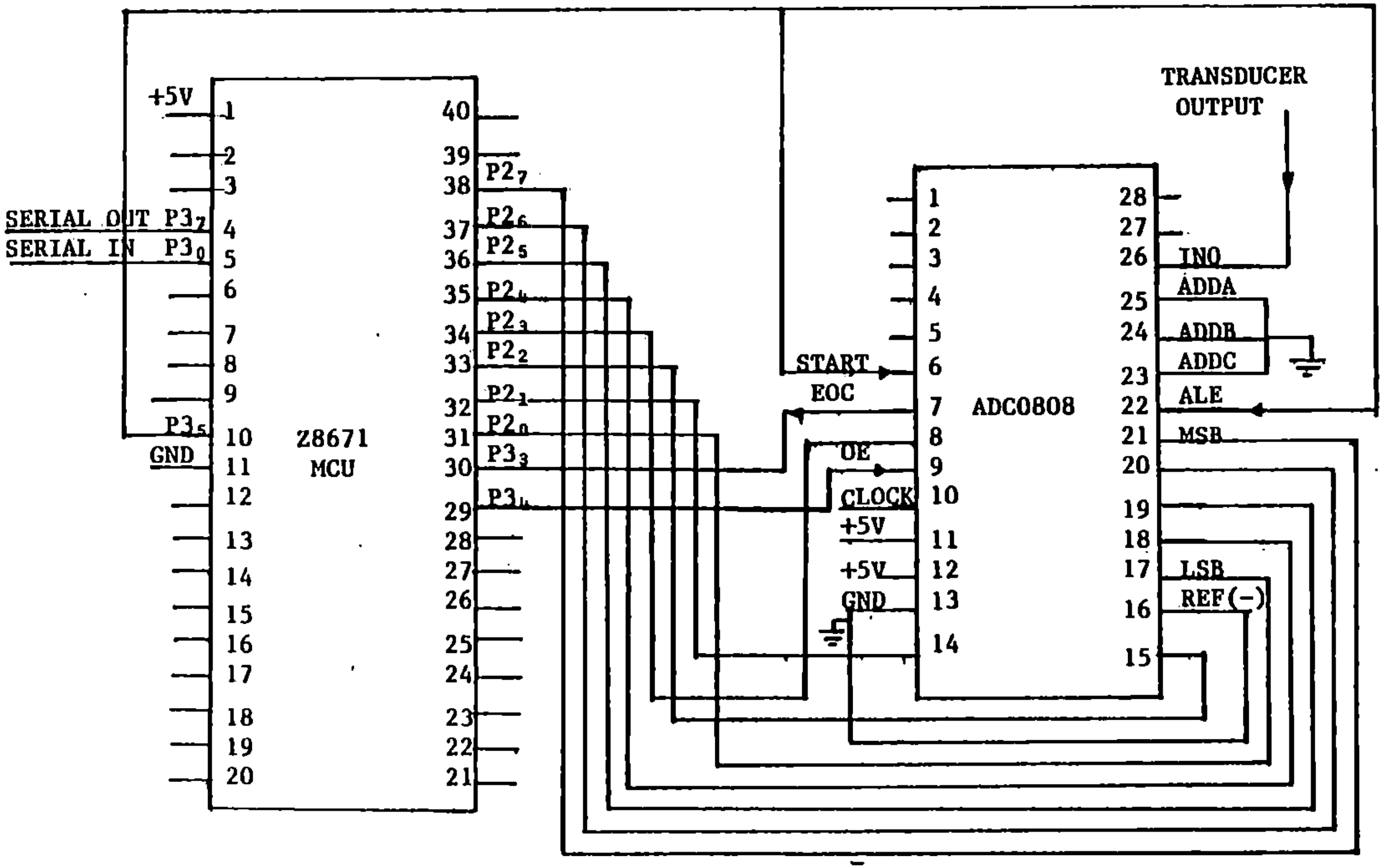


Figure 14. A/D interface to microcomputer

Cuff Pressure (mmHg)	Digital output (Hex) Trial 1	Trial 2	Trial 3
190	FF	FF	FF
180	FF	FF	FF
170	FA	FA	F9
160	EC	ED	E9
150	DF	DA	D9
140	CD	CC	CA
130	B8	B9	BD
120	AC	AC	AA
110	9E	9E	9C
100	89	8C	8F
90	7D	7E	7F
80	6C	6E	6C
70	5B	5B	5B
60	59	59	58
50	4C	4B	49
40	3E	3D	3C
30	29	2D	2C
20	1A	1A	1B
10	0B	0A	0B
0	08	08	08

Figure 15. A/D output

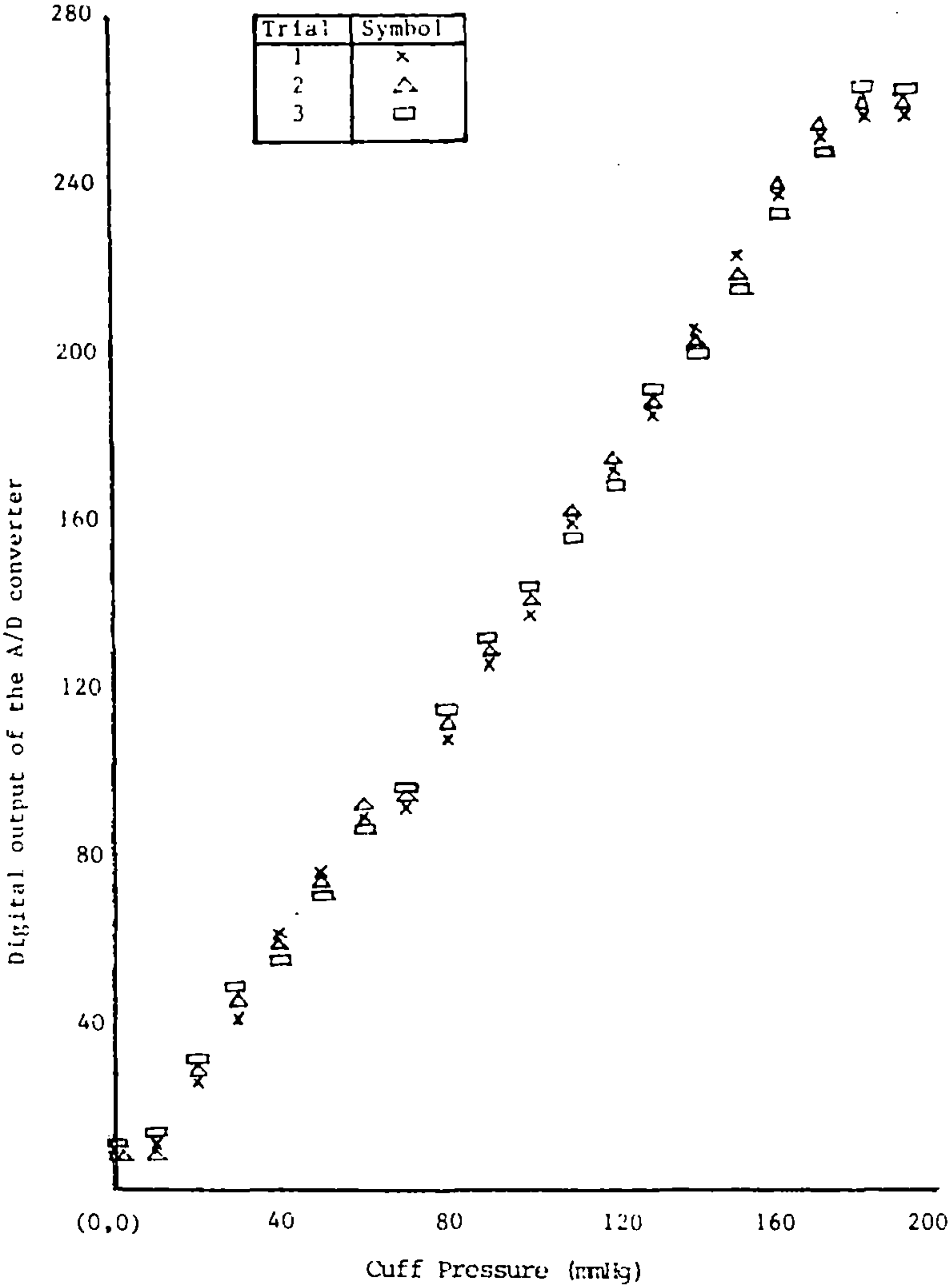


Figure 16. Digital output of the A/D converter versus cuff pressure

Microcomputer Interface to Cuff Inflation/Deflation and Signal Acquisition Circuits

The block diagram representation for the microcomputer interfacing to cuff inflation/deflation and signal acquisition circuits is as shown in Figure 17.

A timer controls the inflation duration of the cuff by switching on and off the power supply to the pump. A detailed circuit diagram and description for the timer is given in the signal acquisition section of this thesis. Port 3 bit 6 of the microcomputer provides the trigger for the timer at the start of every cuff inflation cycle. The BASIC program in Appendix B includes the delay between each cuff cycle. During the deflation of the cuff, the output of the signal acquisition system in the form of digital pulses is input to the microcomputer through port 3 bit 1.

As the cuff pressure is decreased from above systolic pressure, the output pulses start appearing just above systolic pressure. These pulses separate in time as the cuff pressure is reduced from systolic to diastolic pressure. The microcomputer measures the interval between the pulses and stores the value at specific memory locations. At the end of deflation, before starting next inflation cycle (a duration of 7 to 8 minutes), the microcomputer compares the intervals and arrives at the smallest and longest intervals. Since the pulses are closest to each other at systolic pressure and most separated at diastolic pressure, smallest and longest intervals are identified as systolic and diastolic pressure points, respectively. The microcomputer can output the intervals associated with

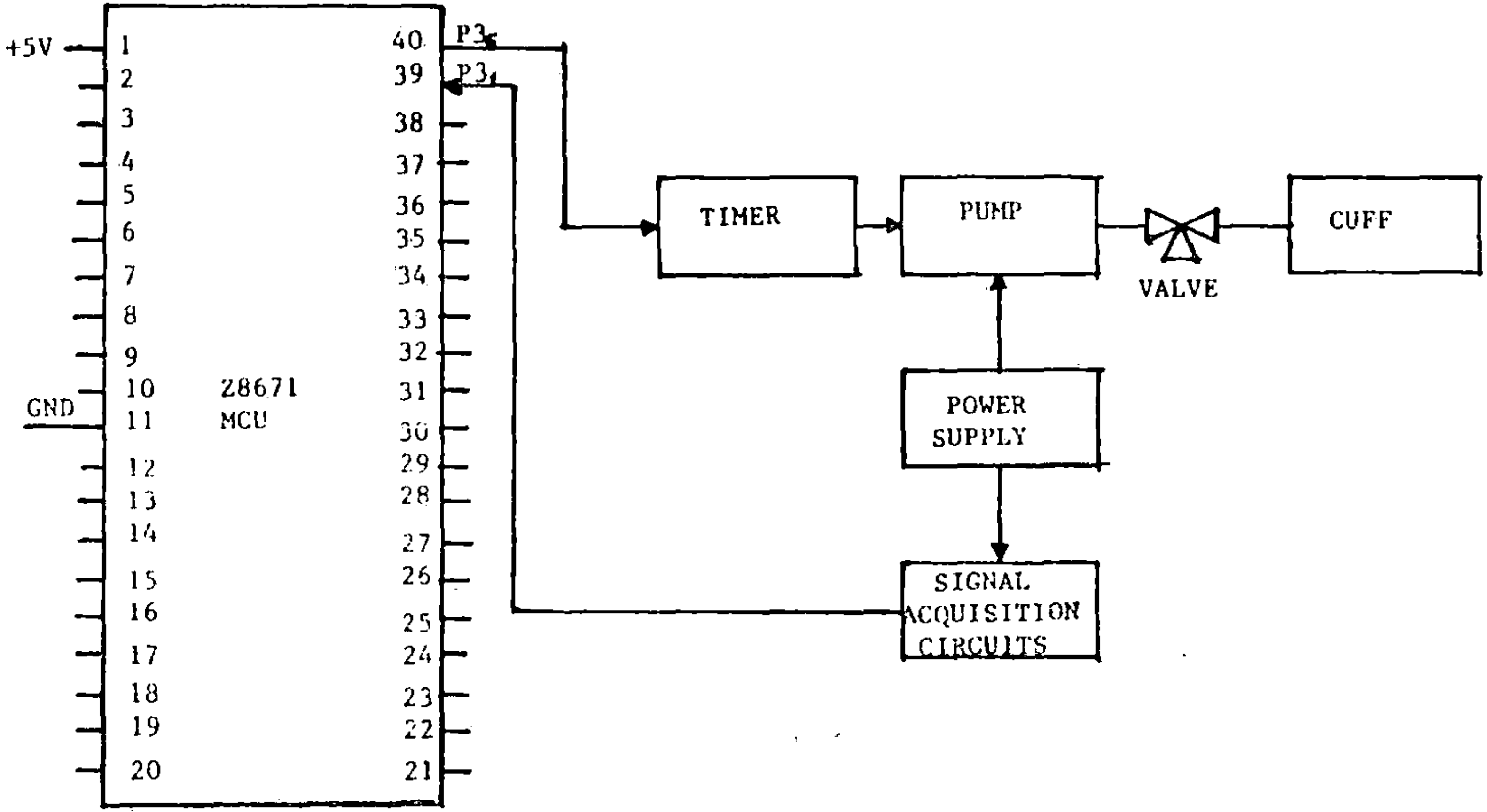


Figure 17. Microcomputer interface to cuff inflation/deflation and signal acquisition cicruits

each of them.

This processing is not real-time. A statistical approach involving a table relating the two pressure values with the associated pulse intervals (the frequency content) can be used with the microcomputer output to determine the blood pressure of a subject.

Appendix D gives a listing for the BASIC program for cuff inflation and deflation and Appendix E the machine language routine for pulse interval measurement.

SYSTEM POWER

Power is derived from a +10 V dc source. The pump draws the most current (120 mA). Since the pump is in operation for a very small fraction of the time (10 seconds in every 10 minutes), the entire system could be battery powered. Digital circuits require +5 V for their operation and this could be derived from the +10 V source by using a +5 V regulator. If the power is drawn from the ac line, care needs to be taken in preventing the 60 Hz signals from entering the audio circuits because the audio amplifiers have a very high gain in the frequency range between 0 and 150 Hz. This is achieved by using adequate filters. The negative supply for ICs 4 and 5 is derived from IC6 which is a voltage converter. IC6 provides a negative supply which is slightly less than the positive supply (Intersil, 1981).

The microcomputer works on +12 V, -12 V, and +5 V power supplies. The microcomputer is driven from the ac supply regulated to required voltages by regulators. Positive and negative power for the amplifiers is derived from a dual power supply. The pressure transducer is driven by a +10 V dc source. Finally, the A/D converter requires +5 V for its operation and this could be derived from the +10 V source by using +5 V regulator.

RESULTS

The pressure transducer was calibrated against a mercury manometer. Transducer output was observed for decreasing cuff pressures. The observations are as shown in Figure 12. The transducer output was observed for linearity and offset. Offset adjustment potentiometer helped to provide Ø V output from the transducer for Ø mmHg input pressure. The transducer response for decreasing cuff pressures is as in Figure 13.

The A/D converter output for decreasing cuff pressures is as shown in Figure 15. Simultaneously, the pressure transducer output was also observed. The plot shown in Figure 16 establishes a correlation between the cuff pressure, the A/D converter output, and the pressure transducer output. This plot will be used later in this section to determine systolic and diastolic pressure values.

The A/D converter output was input to the microcomputer through port 2. Output of the Doppler signal acquisition system, in the form of pulses, was passed through a pulse stretcher circuit before being input to the microcomputer through port 3 bit 1. Software in Appendices C, D, E, and F controlled the above mentioned operations.

At the start of each measurement cycle, the timer enabled the power supply to the pump, thereby causing the inflation of the cuff. The microcomputer provides the trigger for the timer through port 3 bit 6. Port 3 bit 6 is programmed to be an output line. The timer was preset to be "ON" for a duration of 1Ø seconds. At the end of inflation, the cuff starts to deflate and continues to deflate for the next 9 minutes. The program in Appendix D provides for both the time period required for the deflation of the cuff and also for the delay between each cuff cycle (duration of 9 minutes). During the deflation of the cuff, output of the signal acquisition system in the form of digital pulses is input to the microcomputer through port 3 bit 1.

The machine language program in Appendix E measured the interval between the digital pulses. Systolic and diastolic pressure points were identified as the points having the shortest and the longest interval between pulses, respectively. Simultaneously, the digitized values from the A/D converter for the systolic and diastolic pressure points were observed. A terminal (keyboard and CRT) was linked to the microcomputer. Figure 16 was used to arrive at the equivalent blood pressure values in mmHg. Measurements were performed on five different subjects (4 males and 1 female) in the age group 20 to 30 years. Three trials were performed on each subject by three different individuals. The blood pressure values are as reported in Figure 19. The A/D converter output in the form of digitized pressure values were displayed on the CRT linked to the microcomputer.

As the cuff pressure is decreased from above systolic pressure, the output pulses start appearing just above systolic pressure. These pulses separate in time as the cuff pressure is reduced from systolic to diastolic pressure. The microcomputer waits for the arrival of the pulses and on detection of the pulses starts measuring the interval between them. After measuring the interval between a pair of pulses and storing them at specific memory locations, the microcomputer transfers control to the digitizing routine. The microcomputer returns to the interval measurement routine after storing away the digitized pressure values obtained from the A/D converter. Repetition of the above procedure is done till no pulses are observed at the input to the microcomputer. At the end of deflation, before starting next inflation cycle (a duration of 8 minutes), the microcomputer compares the intervals and arrives at the shortest and longest intervals. Since pulses are closest to each other at systolic pressure and most separated at diastolic pressure, shortest and longest intervals are identified as systolic and diastolic pressure points, respectively.

Blood pressure values obtained from the microcomputer were verified against the Korotkoff and the Doppler ultrasound methods. Results are as shown in Figures 18 and 19. Three separate procedures were adopted to arrive at these results. All the procedures involved the automatic inflation of the cuff (wrapped around the upper arm) to 190 mmHg and observations being done while the cuff was deflated. The deflation rate was 2 to 3 mmHg per second. Measurements were performed on five different subjects (4 males and 1 female) in the age group 20 to 30 years.

To start with, a correlation was established between the Korotkoff method and the digital output of the microcomputer. This was achieved by relating the sounds heard over a stethoscope, placed over the brachial artery, with the microcomputer output. Systolic pressure was noted as the point when the sounds appear clear on the stethoscope and diastolic pressure as the point just before the sounds start disappearing. The mercury

Subject	Trial	Doppler ultrasound systolic/diastolic (mmHg)	Korotkoff systolic/diastolic (mmHg)
1	1	109/80	111/78
	2	109/78	110/80
	3	110/78	110/80
2	1	108/76	108/78
	2	108/80	108/82
	3	108/78	108/80
3	1	122/88	120/84
	2	122/86	122/88
	3	118/88	120/88
4	1	110/76	108/74
	2	110/74	108/76
	3	108/76	108/72
5	1	103/62	102/60
	2	102/60	104/60
	3	102/62	102/62

Figure 18. Pressure recordings using Doppler ultrasound and Korotkoff methods

Subject	Trial	A/D output (Hex)	Systolic/Diastolic Pressure (mmHg)
1	1	9C/6A	110/78
	2	9E/6E	110/80
	3	9A/6E	108/80
2	1	9B/6D	108/80
	2	9F/6B	112/78
	3	9C/6A	110/78
3	1	AF/7C	122/88
	2	AD/78	122/86
	3	A8/7B	118/88
4	1	9D/69	108/76
	2	9D/69	108/76
	3	9D/67	108/72
5	1	8D/59	102/60
	2	8E/59	104/60
	3	8D/5A	102/62

Figure 19. Digitized pressure values

manometer reading for each of the sounds was also observed.

Blood pressure corresponding to each digital value was calculated from the plot in Figure 16. This was compared with the mercury manometer.

A correlation between the audio output from the Doppler method and the digital output from the microcomputer was achieved. Audio output, after amplification, was fed to a speaker. As the cuff was deflated, the ultrasonic transducer, placed against the artery, picked up the arterial wall motion signals. For a 2 MHz system, the audio frequencies are in the range Ø to 15Ø Hz. Sounds start appearing at cuff pressures just above systolic pressure. These audio signals are specific to the pressures and hence need a practiced ear for detection and distinction. A thumping sound heard over the speaker was identified as the systolic pressure and the muffling of the sounds as the diastolic pressure. Digital output of the microcomputer for shortest and longest intervals between pulses corresponding to the occurrence of the above mentioned speaker outputs were noted down.

Three trials were performed on each subject by three different individuals. Systolic and diastolic blood pressure values were obtained from Figure 19. As the audio output from the speaker could not be quantized, the author performed several trials to get familiar with the nature of the sounds to distinguish them from other unwanted signals.

A correlation between the Korotkoff and the ultrasound methods was established. The output from the audio amplifier was presented to a speaker to identify the Korotkoff sounds. A stethoscope on the brachial artery served as the reference for comparison. The cuff was inflated automatically to a pressure of 190 mmHg. Deflation rate was set at 2 to 3 mmHg.

Manometer reading corresponding to systolic and diastolic pressures were observed. Sounds heard over the stethoscope were taken as the reference. The speaker output was analyzed against the known systolic and diastolic pressure values. This was done by observing the reading on the manometer corresponding to each audio output from the speaker and correlating them with the known systolic and diastolic pressures. Audio signals, with some training and practice, were distinguishable into the four phases already defined (see literature review). By training and practice, it is implied that the listener needs to have a sense of perception for the sounds. This is not easy to achieve unless practiced a few times. Both systolic and diastolic pressure points are identified by unique audio signals.

Sounds (due to the turbulence of blood) are produced when an artery is occluded and slowly released. These sounds, when heard over a stethoscope, help to identify systolic and diastolic pressures. But to relate the occurrence of the specific sounds with the pressure reading on a mercury manometer, the listener should be familiar with the sounds themselves and also should have the ability to differentiate them from other unwanted signals. These two come out of practice.

In the present situation, audio input to the speaker is a mixture of frequencies in the range 0 to 150 Hz. Signals from the artery are in the range 0 to 25 Hz and 50 to 125 Hz. There is the presence of the 60 Hz noise in the audio signal because of the high gain of the audio amplifiers in the Ø to 15Ø Hz frequency range. As the cuff pressure is decreased, at one point the intraarterial pressure equals the external pressure. At this point, the artery snaps open and closes and this produces a characteristic audio signal. The pressure reading on the mercury manometer corresponding to the audio signal was taken as the systolic pressure value. As the pressure was decreased further, the intraarterial pressure exceeds the external cuff pressure and blood starts flowing freely. Sounds continue to be heard as the cuff is further deflated. At one point during the deflation, the sounds appear muffled and further down are not heard at all. The reading of the manometer corresponding to the point at which the sounds get muffled was taken as the diastolic pressure. The systolic pressure was not so difficult to identify since the audio signal at this point is fairly loud. But identification of the diastolic pressure point required familiarity and a sharp sense of hearing on part of the listener. The author performed several trials with the stethoscope before performing measurements based on the audio signals from the speaker. Also, the blood pressure values related well with the Korotkoff method.

Figures 18 and 19 show the blood pressure values obtained from each of the three methods. The observations were within reasonable limits and comparable to each other. The point to be mentioned here is that on every subject, each trial was performed by a different individual. This is of importance because the audio outputs are very subjective and vary with the hearing ability of the listener. Familiarity with the sounds also plays an important role in the accuracy of measurement. This way, the above mentioned sources of measurement errors could be overcome to a great

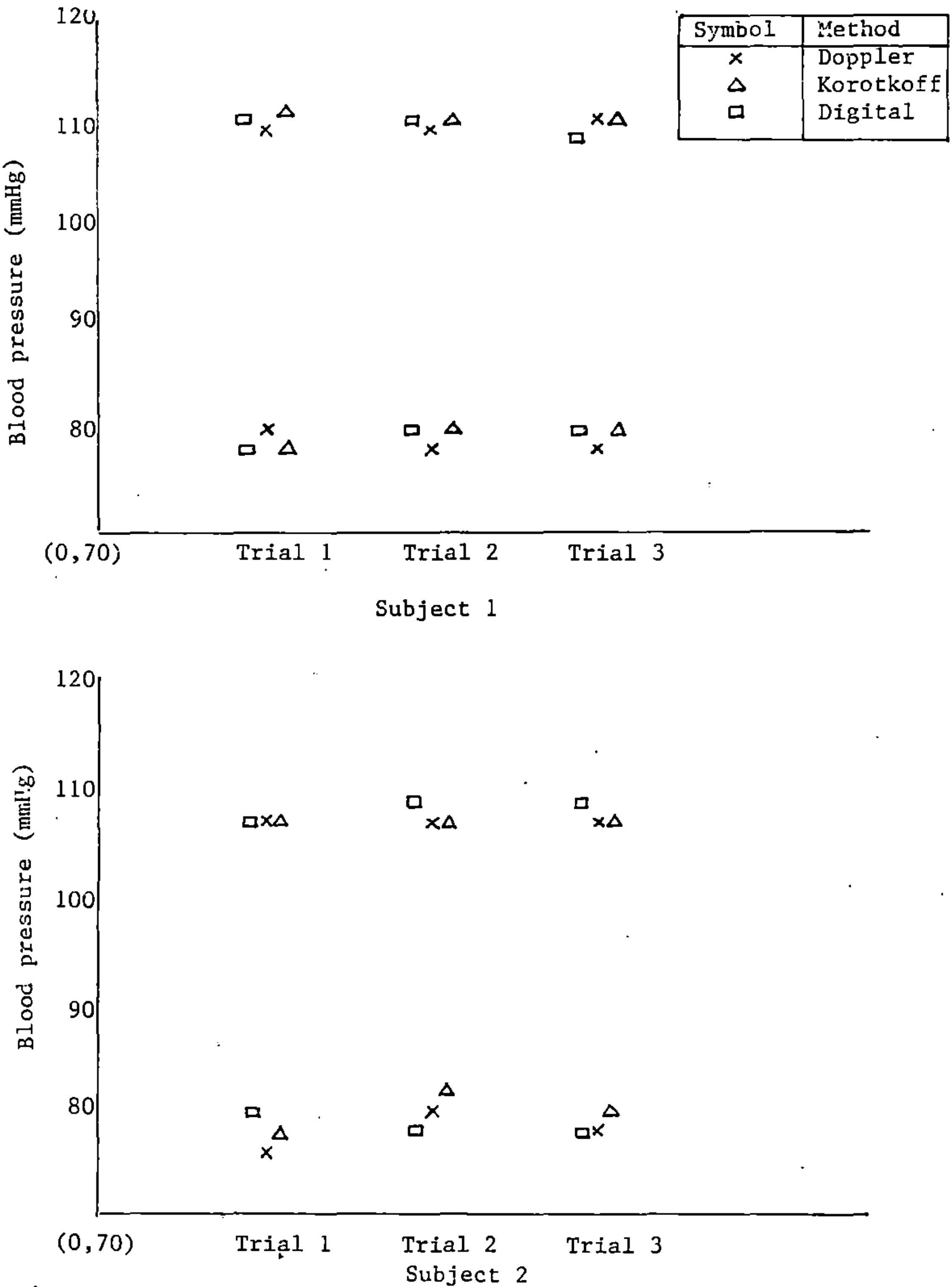
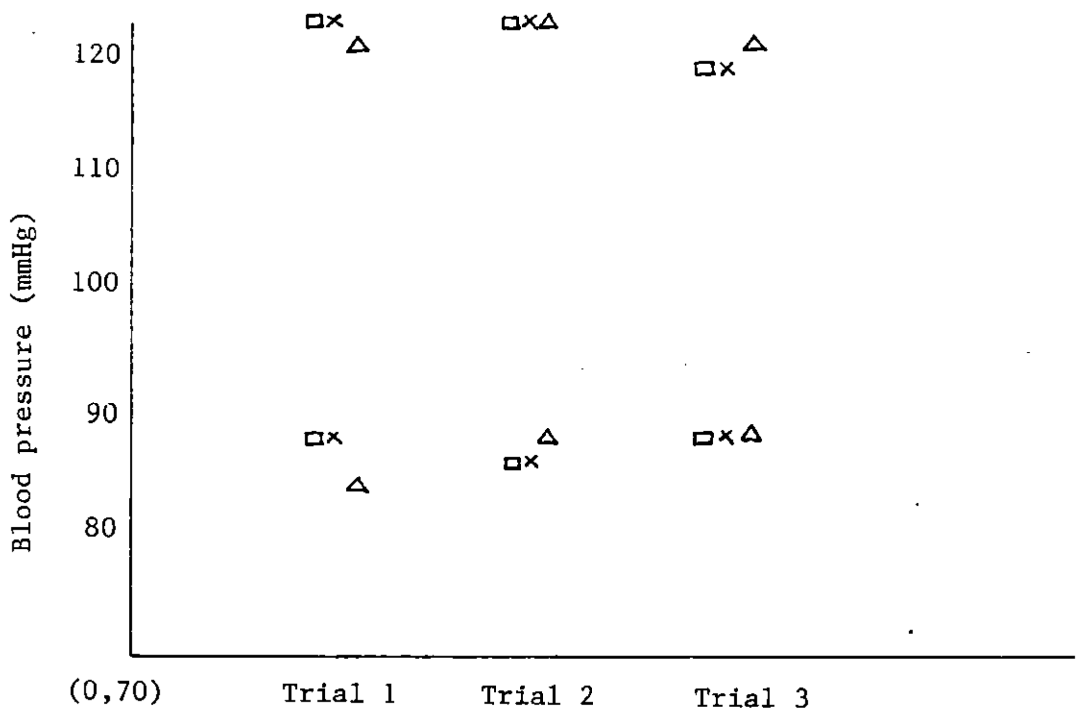
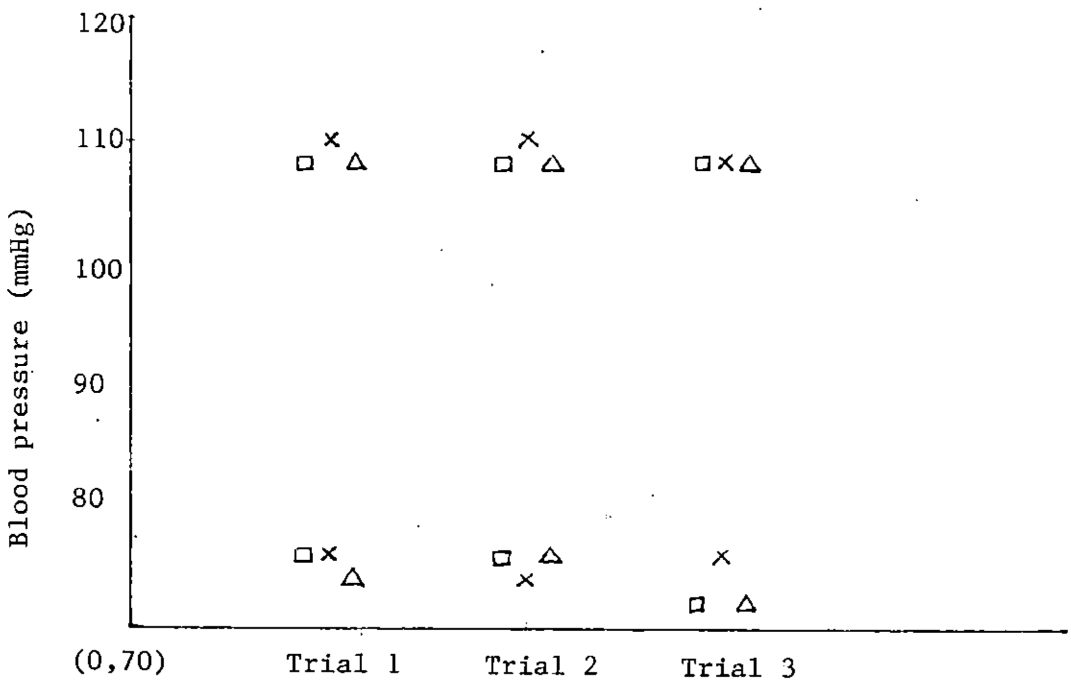


Figure 20. Pressure recordings using Doppler ultrasound versus Korotkoff methods



Subject 3



Subject 4

Figure 20. Continued

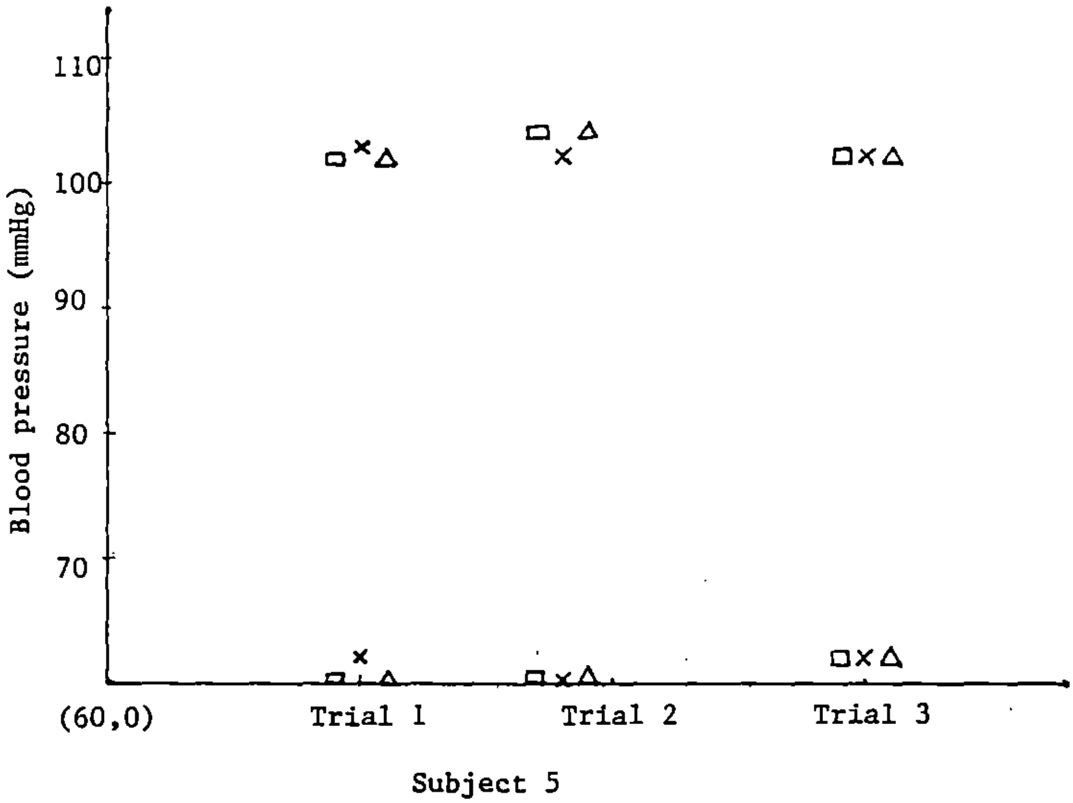


Figure 20. Continued

extent.

A correlation between the three results is as shown in Figure 20. The plots indicate the systolic and diastolic pressure values for each trial and for every subject. As can be seen, the three results compare within acceptable limits.

A striking feature of this system is the usage of 2.2 MHz ultrasound frequency. Most commercial systems, to the best of the author's knowledge, use transducers in the frequency range of 8 to 10 MHz. Also, most of the research, to the best of the author's knowledge, has been done with frequencies in the range between 8 and 10 MHz. For a beam of constant cross section, the power decays exponentially because of absorption of heat in the tissue. The absorption coefficient is approximately proportional to the operating frequency. This implies that the signal attenuation decreases with a decrease in operating frequency. Thus, the choice of a lower operating frequency (2.2 MHz in the present case) calls for lesser power from the oscillator driving the transmitter crystals. In the present case, the target of interest is the arterial wall and not the red-blood cells suspended in the blood. Therefore, the power scattered back from the red-blood cells is not of importance. The power scattered back from the red-blood cells is proportional to the fourth power of the operating frequency and hence demands for higher operating frequencies from systems depending on the power back scattered from the red-blood cells. This was another reason behind the choice of a lower (2.2 MHz) operating frequency. Using the plot in Figure 16 as the reference, systolic and diastolic pressures can be obtained just by noting the digital values corresponding to the various audio outputs from the speaker. Thus, once the system is calibrated, neither a manometer nor a stethoscope, is required.

A photo of the Doppler signal processing system is as shown in Figure 21. This system includes the microcomputer, the A/D converter, and the pressure transducer.

For achieving the most efficient measurement, certain procedures are to be followed very carefully. The oscillator and the receiver are to be carefully tuned to the carrier frequency. The received audio signals are a mixture of frequencies and distinguishing the required arterial wall motion signals from the rest requires practice. A trained ear is necessary to identify the sounds with the arterial blood pressure and the the reasons for the same were discussed earlier in this section. The transducer floods the area around the artery with ultrasound and to get the best possible reflections, the transducer has to be properly positioned on the artery. Proper positioning could be achieved through practice. Excessive arm movement causes variation in the results and does give misleading results. But this is true with any indirect blood pressure measurement method. As long as the arm is reasonably still, the method functions properly.

Audio signals appear at pressures higher than the systolic pressure. Also, the diastolic pressure point could lie between the phases 4 (muffled sounds) and 5 (no sound) and this could give erroneous results. Observation of the frequency spectrum related to the arterial wall opening and closing does provide a more reliable method of identifying the diastolic pressure point.

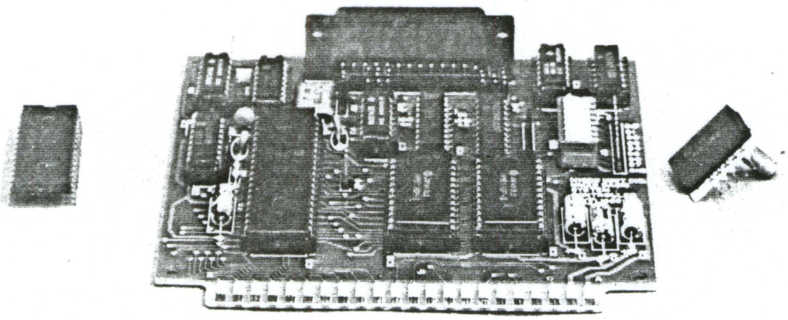


Figure 21. Microcomputer, A/D converter, and pressure transducer photograph

The microcomputer register pointer should be initialized properly. Precaution should be taken in not overwriting the BASIC program with machine language routine. The initialization values for each of the pointer registers is shown on top of each of the software routines.

CONCLUSION

Rresults obtained prove the feasibility of using the conventional Doppler for measuring blood pressure and the possibility of incorporating a micrcomputer into the system. Results also prove the successful usage of a 2.2 MHz Doppler ultrasound for the detection of the arterial wall motion.

The successful incorporation of a microcomputer into the system has enhanced the diagnostic value of the method. The microcomputer digital output correlated with the pressure recordings done using the ultrasound Doppler method. Although a manometer was used to calibrate the A/D converter output initially, actual measurement procedures do not need the stethoscope and the manometer. By correlating the audio output with the digitized values from the microcomputer, one could arrive at a subject's blood pressure. Speaker and stethoscope were used for the verification of the microcomputer output against the Doppler ultrasound and Korotkoff methods. Successful and fairly accurate blood pressure values obtained by noting the digitized values with the shortest and longest intervals between pulses, proved the feasibility of incorporating a microcomputer. This totally eliminated the involvement of a human operator in the measurement procedure. The plot in Figure 16 can be used as a reference for future measurements.

The microcomputer and the stored software controlled the inflation and deflation of the cuff. A delay between each cuff cycle was also provided by the microcomputer. Thus, the human operator responsible for inflation and deflation of the cuff in earlier methods was eliminated. Cuff pressure was sensed by an electronic pressure transducer and digitized by an A/D converter. A digital equivalent for each value of the cuff pressure was input to the microcomputer. An ultrasonic probe sensed the arterial wall motion signals and these signals were converted to digital pulses. A correlation between the A/D output and the digital pulses enabled the determination of the systolic and diastolic pressure values of the subject under test. The plot in Figure 16 served as the reference. This system no longer relied upon the hearing acuity of the operator. Also, there was no influence of environmental noise on the measurement accuracy.

All the measurements were performed on subjects in the age group 20 to 30 years. Also, the tests were performed to measure the subject's blood pressure at a given instant. But the microcomputer can assist in storing data obtained from measurements performed over a length of time.

The data storing ability of the microcomputer could be used for recording and saving blood pressure data over a longer period of time. These data could be used for trend studies and predictions.

Results from the conventional Doppler, the successful incorporation of a single-chip microcomputer for controlling and data processing, indicate the advent of an automated, ambulatory, and battery powered blood pressure unit.

SCOPE FOR FUTURE STUDIES

The idea behind this project was to design an ambulatory blood pressure monitor using the ultrasound Doppler method.

Arterial wall motion signals have been successfully transduced and digitized by a microcomputer. A plot of the digitized values versus the cuff pressure serves as the reference for determining the systolic and diastolic pressures.

Future studies should be along methods for recording blood pressure data over a longer period of time. The incorporation of a microcomputer into the system can provide the memory for saving the data. The design should involve the time recording and also an ability to record abnormal values and the time of its occurrence. Such provision would help to to identify subject's action at the time of occurrence of the abnormal event. This could assist in diagnosing the reason behind the rise or fall in blood pressure. In the present thesis, measurements were performed on selective subjects in the age group 20 to 30 years. A better measure of the capability of the system is the performance of tests on many subjects and of different age groups.

Trend study could help in relating an individual's daily routine with the trends in blood pressure. Such a study could identify probable cardiac cases. Diagnostic value of the system could be significant for subjects recovering from cardiac attacks. Their blood pressure could be studied and suitable therapy suggested.

The ambulatory nature of the system eliminates the need for frequent visits to the hospital. Battery powered operation and light weight components can reduce the discomfort of carrying the unit on one's person.

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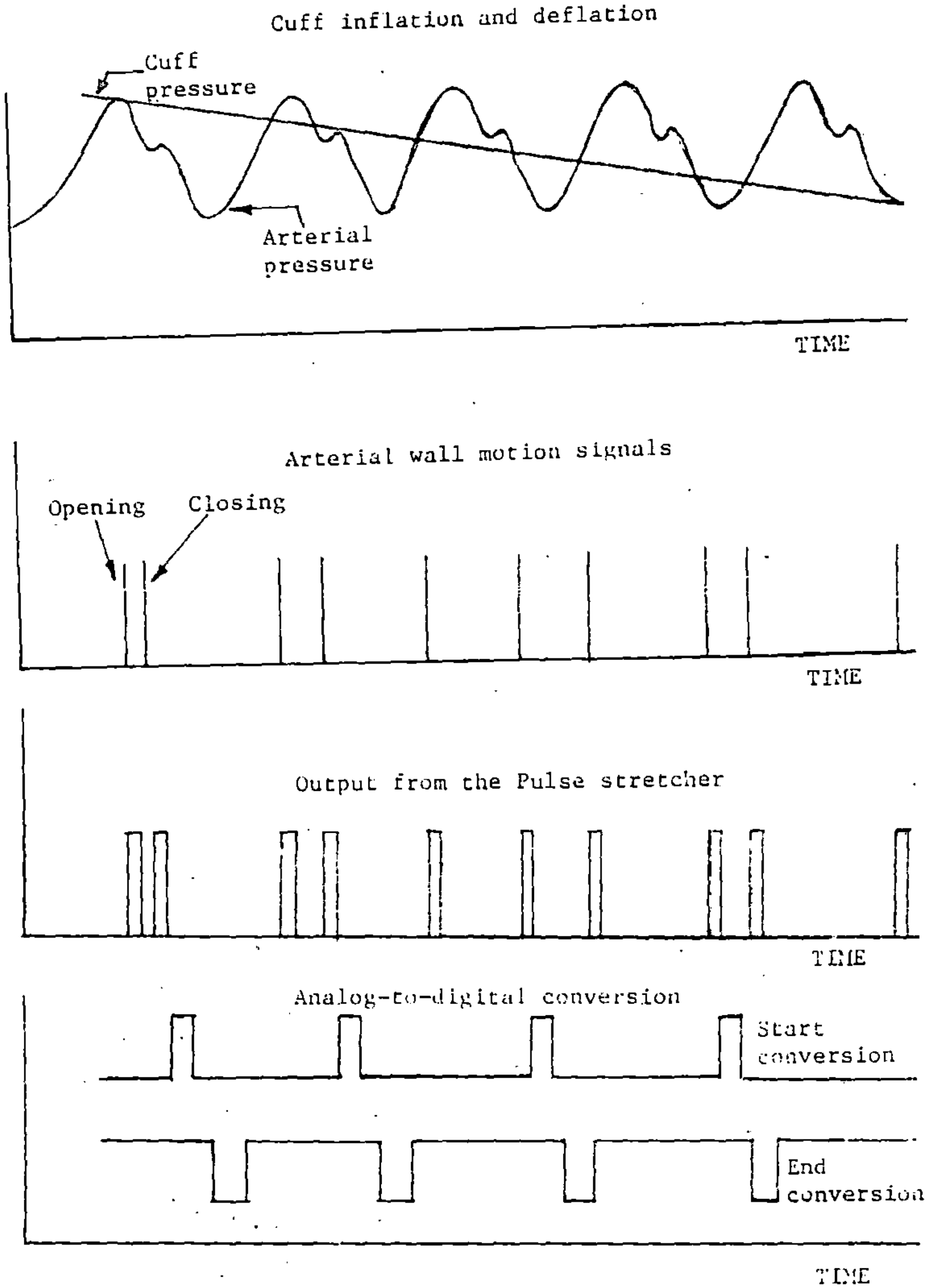
APPENDIX A: TIMING DIAGRAM



APPENDIX B: TRANSDUCER DESIGN

A good transducer design is very important to the accuracy of measurement; consequently, an appropriate design must meet many requirements. Therefore, the principle behind the operation of the prefabricated piezoelectric transducer used in this work and the criteria behind the transducer design are worth mentioning.

In diagnostic applications, ultrasound is both generated and detected by the piezoelectric effect. This phenomenon involves a conversion between mechanical and electrical energies. Piezoelectric materials have the property that when subjected to deformation they provide an electric field. The converse is also true. The best known naturally occurring crystals that exhibit the piezoelectric effect are the quartz crystals. There are many synthetic crystals, lead zirconate titanate, for example, that are strongly piezoelectric and are used commonly in ultrasonic diagnosis.

Narrow beams of ultrasound are usually required in ultrasonic diagnosis. Such a beam is often generated by a piezoelectric material which is electrically excited by means of two electrodes, one on each surface. If an alternating voltage is applied between the electrodes, the piezoelectric effect causes a synchronous variation in the thickness of the transducer. It has been shown that a transducer operating in the thickness mode generates a well-collimated beam of longitudinal waves if the width of the transducer is large compared to one wavelength (Mcdicken, 1976).

The size of the individual transducers is selected on the basis of the necessary ultrasonic beam width, electrical impedance, and to maintain reasonable cross-section to thickness ratio of the transducer to minimize the generation of shear waves. For example, a rectangular transducer crystal provides a fairly evenly distributed ultrasonic energy, while a round one spreads the energy in one plane only.

Physiological parameters also have a bearing on the transducer design. Features such as location of the artery, its size and distensibility, and its relationship to the bone complicate transducer design. These features affect the beam shape, distribution, size, and location.

Transducer thickness is a function of the wavelength of the exciting signal. When the thickness is one-half a wavelength at the frequency of the electrical excitation, the stresses responsible for the crystal deformation reinforce, leading to resonance. For a transducer to function more efficiently, its mechanical resonant frequency needs to be matched with the frequency of the exciting signal. This results in large amplitude displacements and high sensitivity (Newhouse et al., 1980).

The important areas to be considered in a transducer design are crystals and crystal matching, focusing, backing and coating, and isolation.

Theoretically, using separate crystal types as the transmitter and the receiver is optimal, but because of their differing aging parameters, a single crystal is used. Also, using a single crystal type for the transmitter and the receiver gives better sensitivity and time stability. Resonant matching of crystal transmitter and receiver is necessary for optimum coupling of mechanical and electrical characteristics.

In some applications, it is desirable for the transducer to operate at maximum efficiency. This requires that the electrical energy be transferred with a minimum of loss from the transducer to the load and vice versa. In such a case, the transducer is backed by air, to minimize the loss in its mounting. This has proved optimal for Doppler ultrasonic applications, both theoretically and experimentally (Mcdicken, 1976). Further improvements can be achieved by matching the characteristic impedance of the transducer to that of the loading medium by means of a matching layer attached to the transducer. This matching layer should be of an intermediate characteristic impedance and of thickness equal to one-quarter of a wavelength. Epoxy is the most commonly used matching material.

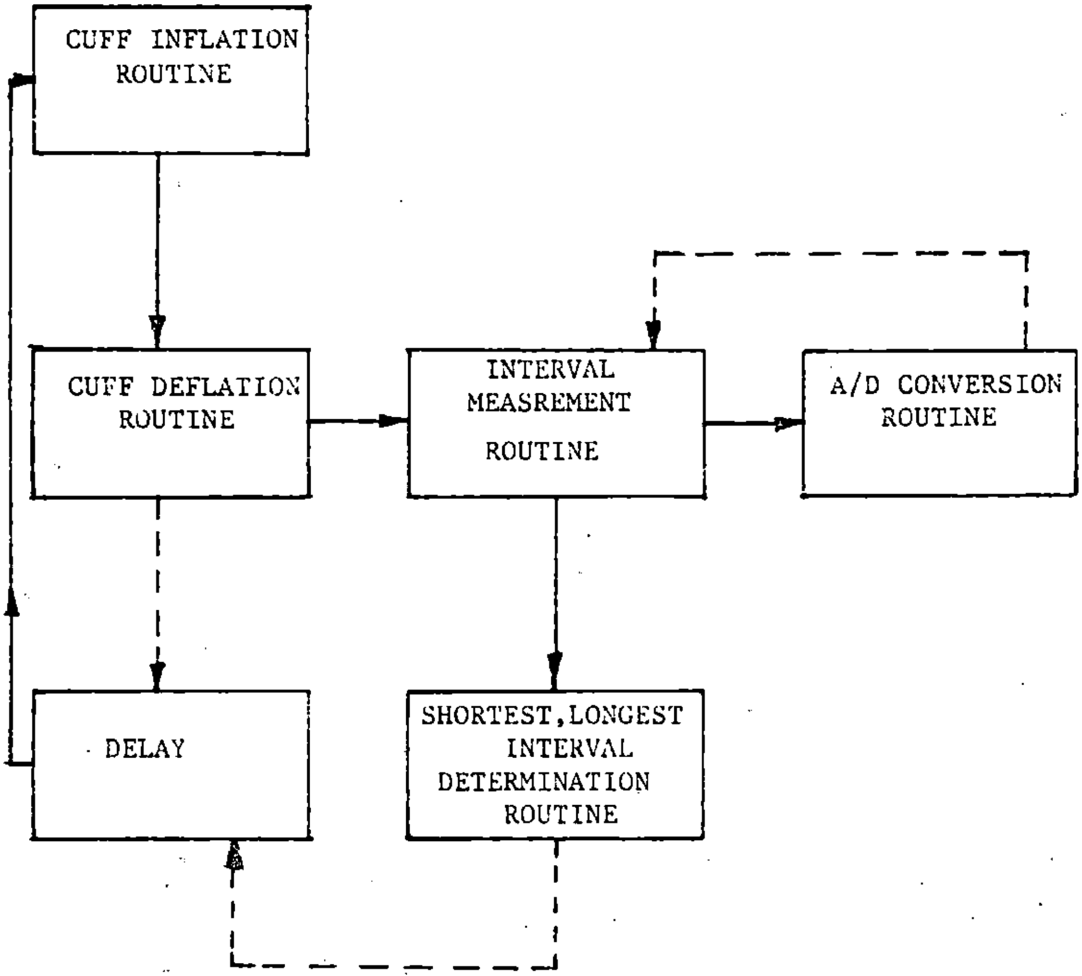
Another important design criterion is the crystal damping. High effeciency transducers are the ones with minimal damping at the resonant frequency. This causes a drastic drop in transducer sensitivity to frequencies just beyond the resonant frequency. Thus, the response of the transducer becomes dependent on the frequency. If the damping is increased, the response becomes less dependent on the frequency and the transducer responds to energy pulses of very short duration. In this case, the energy is spread over a spectrum of frequencies. The transducer has a wider frequency response because of the increase in damping. Mechanical damping is provided by a block of highly absorbant material, like particles of tungsten, attached to the rear surface of the transducer (Mcleod, 1967).

The transducer probe is isolated from the case by an ultrasonic insulator (rubberized cork). This minimizes the coupling of ultrasonic energy between the case of the probe and the transducer and eliminates the ringing of the case.

A transducer could have a single transmitting and a single receiving crystal. In such a case, the placement of the probe on the artery is very critical. Accurate transducer placement is achieved by trial and error. But a multiple element transducer, wherein there are separate and more than one transmitting and receiving crystals, eliminates the criticality of transducer placement. Typically, this arrangement floods the artery with ultrasonic energy. Although the reflected signals are a mixture of wanted signals and unwanted stationary echoes, proper filtering eliminates the unwanted signals. But each of these elements need be both electrically and acoustically isolated from each other.

Finally, a low-loss transmission of ultrasonic energy to the skin is achieved by a coupling medium. This coupling gel must have certain properties. It should permit long-term monitoring without drying out, be easy to remove without irritation of the skin, and be nontoxic.

APPENDIX C: SOFTWARE ROUTINE LINKAGE SCHEMATIC

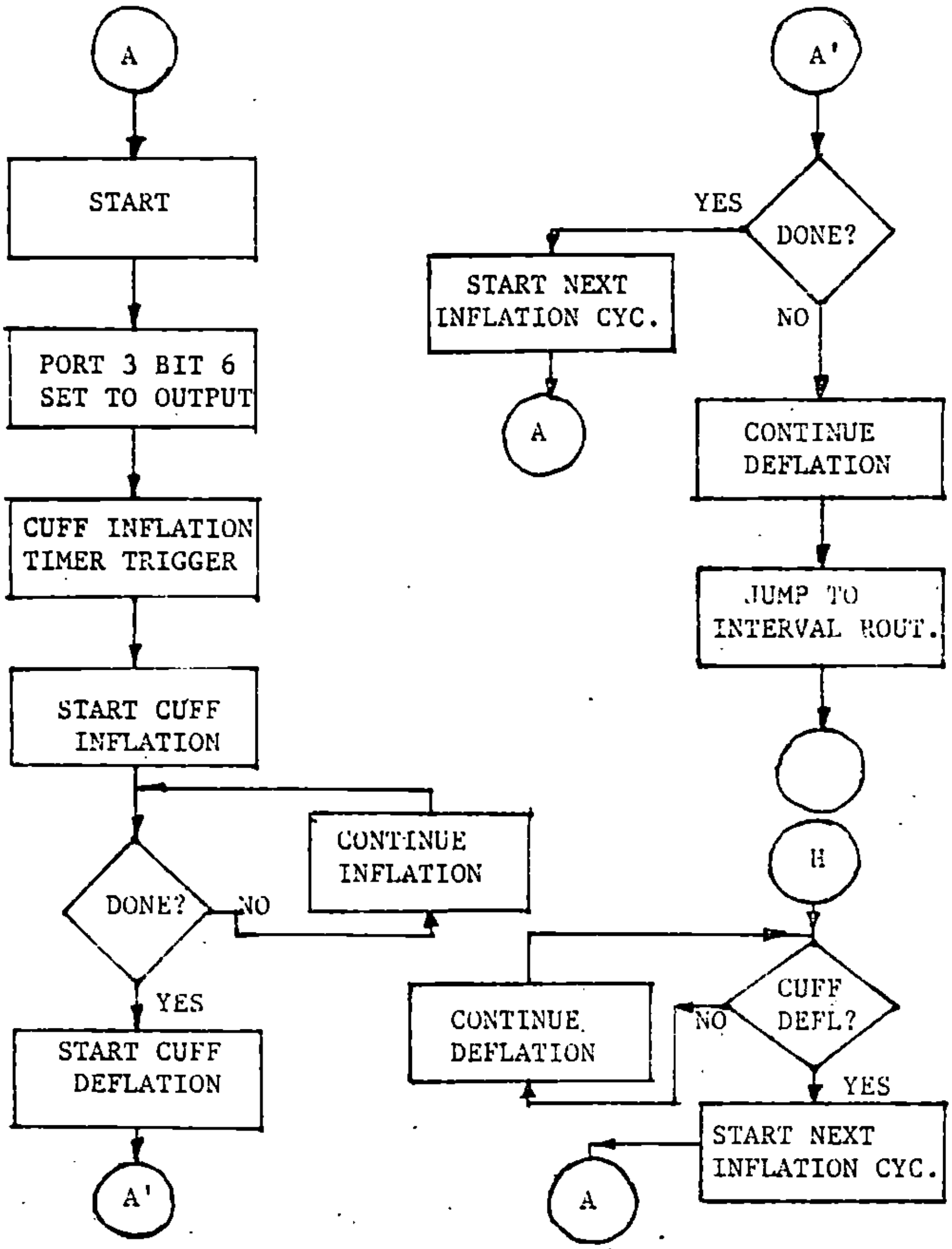


——— Link between independent functional blocks

– – – – – – Link from within a independent functional block.

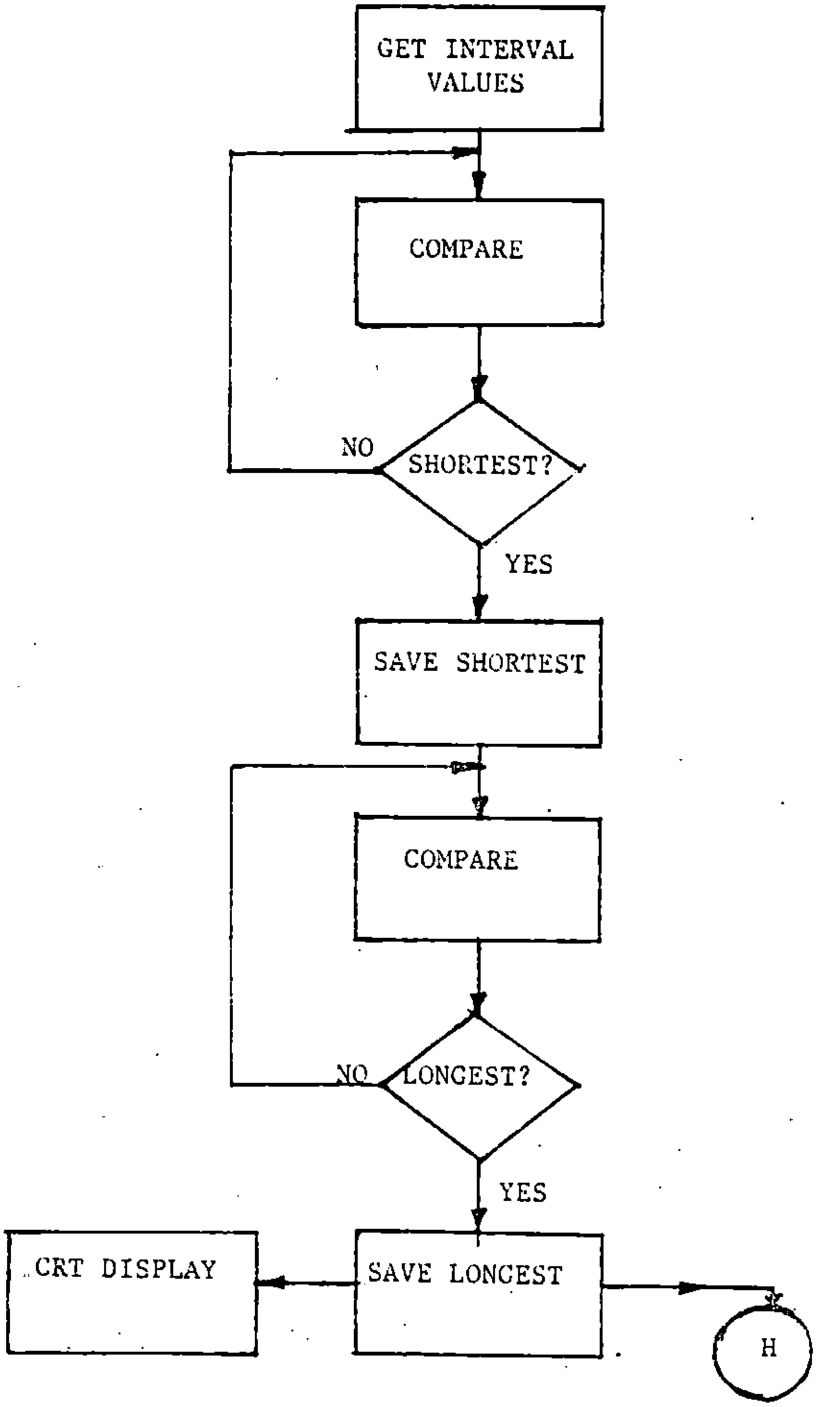
APPENDIX D: CUFF INFLATION/DEFLATION ROUTINE

FLOW DIAGRAM



SHORTEST AND LONGEST INTERVALS DETERMINATION

FLOW DIAGRAM



BASIC PROGRAM FOR CUFF INFLATION/DEFLATION

```
Register pointers: R8-9 = %900
                   R4-5 = %921
                 R10-11 = %1720
                   R6-7 = %1720
```

```
10   REM Screen Erasure and Beep invoke procedure
20   REM Port3 bit6 set to output
30   REM Port3 bit6 drives the pump
40   REM Pump on for 10 seconds and off for 9 minutes
50   REM Center character on screen
60   A = 23
70   PRINT " "
80   REM Erase screen ; print blank character
90   A = A - 1
100  IF A < > 0 THEN 70
```

```
150  PRINT " "," "," ","CUFF INFLATION TO START AFTER BEEP"
160  REM Character centered: GOSUB 400
170  REM Beep invoked after delay
180  REM Delay : GOSUB 500
190  PRINT " ": REM For beep press CTRL G between qoutes
200  REM Set port3 bit6 to output
210  @247 = %41; @3 = %40
220  PRINT
230  PRINT
240  PRINT
250  PRINT" "," "," "," ","CUFF INFLATION STARTED"
260  REM 5 Seconds delay: GOSUB 500
270  @3 = 00: REM Timer trigger
280  GOSUB 600
290  REM Inflation delay 10-12 seconds
300  GOSUB 500
310  PRINT " ": REM For beep press CTRL G between quotes
320  GOTO 700
400  I = 10
410  PRINT " "
420  I = I - 1
430  IF I < > 0 THEN 410
440  RET
```

```
500  B = Ø
510  B = B + 1
520  IF B < > 300 THEN 510
530  RET
```

```
600  K = Ø
610  K = K + 1
620  IF K < > 800 THEN 610
630  PRINT
640  PRINT
650  PRINT
660  PRINT " "," "," ","WAIT FOR BEEP TO START DEFLATION"
```

```
700  PRINT
710  PRINT
720  PRINT
730  PRINT " "," "," "," ","START DEFLATION OF THE CUFF"
740  REM Machine language routine for Systolic and Diastolic
750  REM pressure points identification
760  GOTO 1000
770  PRINT
780  PRINT
790  PRINT
791  PRINT " "," "," ","WAIT FOR BEEP TO STOP DEFLATION"
792  GOSUB 500
793  PRINT " ": REM For beep press CTRL G between quotes
794  PRINT
795  PRINT
796  PRINT
797  PRINT " "," "," ","DEFLATION OF CUFF IS COMPLETE"
```

```
800  PRINT
810  PRINT
820  PRINT
830  PRINT " ","ERASE SCREEN BEFORE STARTING NEXT INFLATION CYCLE"
840  REM One cycle of inflation and deflation ic complete
850  REM Return to start
860  GOTO 60
```

```
900  Y = 0
910  Z = 0
920  Z = Z + 1
930  REM Delay for deflation
940  IF Z < > 100 THEN 920
950  Y = Y + 1
960  IF Y < > 300 THEN 910
970  REM Sound beep to mark end of deflation
980  RET
```

```
1000 X = 34 : REM Count storage starting location
1010 J = 0  : REM Pulse counter initialise
1020 @247 = %41 : REM Port3 bit1 set to input
1030 @3 = 00 : REM Port3 register initialise
1040 GO@%0B00,J,X : REM Count interval between pulses routine
1060 X = 34
1070 J = 0
1080 PRINT @X
1090 X = X + 1
1100 J = J + 1
1110 IF J < > 12 THEN 1080
```

```
1200 IF @34 - @35 < 0 THEN 1300
1205 REM Identify longest and shortest intervals
1210 IF @34 - @36 < 0 THEN 1400
1220 GOTO 1270
1230 PRINT "@34 IDENTIFIES SYSTOLIC POINT"; PRINT @34
1234 PRINT "@36 IDENTIFIES DIASTOLIC POINT"; PRINT @36
1235 PRINT @35
1240 GOSUB 900
1245 GOTO 770
```

```
1250 print "@3 IDENTIFIES SYSTOLIC POINT"; PRINT @34
1254 PRINT "@35 IDENTIFIES DIASTOLIC POINT"; PRINT @35
1255 PRINT @36
1260 GOSUB 900
1265 GOTO 770
1270 IF @35 - @36 < 0 THEN 1250
1280 GOTO 1230
```

```
1300 IF @35 - @36 < Ø THEN 1400
1310 GOTO 1360
1320 PRINT "@35 IDENTIFIES SYSTOLIC POINT"; PRINT @35
1324 PRINT "@34 IDENTIFIES DIASTOLIC POINT"; PRINT @34
1325 PRINT @36
1330 GOSUB 900
1335 GOTO 770
1340 PRINT "@35 IDENTIFIES SYSTOLIC POINT"; PRINT @35
1344 PRINT "@36 IDENTIFIES DIASTOLIC POINT"; PRINT @36
1345 PRINT @34
1350 GOSUB 900
1355 GOTO 770
1360 IF @34 - @36 < Ø THEN 1340
1370 GOTO 1320
```

```
1400 IF @34 - @35 < Ø THEN 1430
1410 PRINT "@36 IDENTIFIES SYSTOLIC POINT"; PRINT @36
1414 PRINT "@35 IDENTIFIES DIASTOLIC POINT"; PRINT @35
1415 PRINT @34
1420 GOSUB 900
1425 GOTO 770
1430 PRINT "@36 IDENTIFIES SYSTOLIC POINT"; PRINT @36
1434 PRINT "@34 IDENTIFIES DIASTOLIC POINT"; PRINT @34
1435 PRINT @35
1440 GOSUB 900
1445 GOTO 770
```

APPENDIX E: INTERVAL MEASUREMENT ROUTINE

Mnemonic	Address Hex	Data Hex	Bytes	Comments
LD R45,R21	ØBØØ	E4	Ø3	J into R45
	ØBØ1	15		
	ØBØ2	2D		
LD R46, #Ø6	ØBØ3	E6	Ø3	Number of pulses
	ØBØ4	2E		to be counted
	ØBØ5	Ø6		into R46
CP R46, R45	ØBØ6	A4	Ø3	Done with the
	ØBØ7	2D		number of pulses?
	ØBØ8	2E		
JR Z	ØBØ9	6B	Ø2	Yes; exit
	ØBØA	28		
LD R4Ø, #ØØ	ØBØB	E6	Ø3	Initialize counter
	ØBØC	28		low byte
	ØBØD	ØØ		
LD R41, #ØØ	ØBØE	E6	Ø3	Initialize counter
	ØBØF	29		high byte
	ØB1Ø	ØØ		
LD R42, #Ø2	ØB11	E6	Ø3	Test bit for
	ØB12	2A		port3 status
	ØB13	Ø2		
CP R42, R3	ØB14	A4	Ø3	Port3 bit1 high?
	ØB15	Ø3		
	ØB16	2A		
JR NZ	ØB17	EB	Ø2	No; wait for high
	ØB18	FB		
CP R42, R3	ØB19	A4	Ø3	Yes; port3 bit1
	ØB1A	Ø3		low?
	ØB1B	2A		

JR Z	ØB1C	6B	Ø2	No; wait for
	ØB1D	FB		port3 bit1 low
INCW RR4Ø	ØB1E	AØ	Ø2	Yes; Increment
	ØB1F	28		16 bit counter
CP R42, R3	ØB2Ø	A4	Ø3	Port3 high?
	ØB21	Ø2		
	ØB22	2A		
JR NZ	ØB23	EB	Ø2	No; continue
	ØB24	F9		count
LD @R19, R41	ØB25	F5	Ø3	Count low byte
	ØB26	29		into @34
	ØB27	13		
INCW RR18	ØB28	AØ	Ø2	Next location
	ØB29	12		
LD @R19, R4Ø	ØB2A	F5	Ø3	Count high byte
	ØB2B	28		into @35
	ØB2C	13		
INCW RR18	ØB2D	AØ	Ø2	Next location
	ØB2E	12		
INCW RR2Ø	ØB2F	AØ	Ø2	Next pulse
	ØB3Ø	14		
JR	ØB31	8B	Ø2	Return to start
	ØB32	CD		
RET	ØB33	AF	Ø1	Exit from subroutine

APPENDIX F: DIGITAL OUTPUT SUBROUTINE

```
Register pointers:  R8-9   = %800
                    R4-5   = %821
                    R6-7   = %1720
                    R10-11 = %1720
```

BASIC PROGRAM

```
10  REM. Program for A/D conversion
20  REM Set port2 to input
30  @246 = 255
40  REM Set port3 to general purpose mode
50  @247 = %41
60  REM Machine language program for A/D conversion
70  REM Starting address of machine language subroutine at 0A00
80  GO@%0A00
90  REM output digital data
100 REM Port2 reads in digital data
110 PRINT HEX(@2)
120 REM Return to start
130 GOTO 80
```

MACHINE LANGUAGE SUBROUTINE FOR A/D CONVERSION

Mnemonic	Address Hex	Data Hex	Bytes	Comment
LD R3, #%30	0A00	E6	03	Enable Start
	0A01	03		Convert, ALE
	0A02	30		and OE
AND R3, #08	0A03	56	03	EOC low?
	0A04	03		
	0A05	08		
JR NZ	0A06	EB	02	No; Wait
	0A07	FB		Yes; proceed
AND R3, #08	0A08	56	03	EOC high?
	0A09	03		
	0A0A	08		
JR Z	0A0B	6B	02	No; Wait
	0A0C	FB		Yes; proceed
LD R3, #%10	0A0D	E6	03	Enable OE
	0A0E	03		Read data onto
	0A0F	10		port2
RET	0A10	AF	01	Output data and return to start