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Study of an Automatic Sphygmomanometer System

James B. Morgan

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STUDY OF AN AUTOMATIC SPHYGMOMANOMETER SYSTEM

BY

JAMES B. MORGAN

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This thesis is approved as a creditable and independent investigation by a candidate for the degree, Master of Science, and is acceptable as meeting the thesis requirements for this degree. Acceptance of this thesis does not imply that the conclusions reached by the candidate are necessarily the conclusions of the major department.

Thesis Adviser Date

Head, Electrical Engineering Department Date

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ABSTRACT

A simple and dependable method of automatically measuring blood pressure indirectly would be of great help to the medical profession. An electronic system to implement the measurement of blood pressure using a double cuff method is presented. The scheme is to determine diastolic pressure from the amplitude of the difference signal obtained by subtracting the pulse wave observed in one cuff from that of the other, and to determine systolic pressure from the amplitude of the pulse wave observed in the cuff furthermost from the heart. Although the system is electronically practical, the results obtained from this system show that the difference signal does not give a dependable indication of the diastolic pressure point. The method of determining systolic pressure is workable.

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CHAPTER I

INTRODUCTION

One of the most common diagnostic procedures in modern medicine is the measurement of blood pressure. Coronary malfunctions, arterial diseases, congenital cardiovascular irregularities, pulmonary and endocrine disorders, as well as disorders of the nervous system, may be diagnosed with blood pressure records.[1]

Basically, the measurement of blood pressure involves determination of the limits of pressure variation occurring in the arteries due to the pumping action of the heart. The peak arterial pressure, the systolic pressure, occurs during contraction of the heart; the lowest arterial pressure, the diastolic pressure, occurs during relaxation of the heart.

Direct methods for measuring blood pressure, which involve inserting a cannula into the artery, are accurate but undesirable for normal clinical use. Indirect methods, in which the measurement process is entirely external to the body, are generally used.

Almost all indirect methods for measuring blood pressure use an inflatable cuff wrapped around the arm of the subject. Because the tissues of the arm act as a fluid media, any pressure in the cuff is transmitted equally in all directions to the part of the arm encircled by the cuff. Little resistance to compression is offered by the arterial walls, so whenever the cuff pressure exceeds the instantaneous arterial pressure, the arterial walls collapse and the artery is occluded. If the cuff pressure is above the systolic pressure of the subject, the artery is occluded throughout the entire cardiac cycle.

In the auscultatory method, the most generally used method, the systolic and diastolic end points are determined by listening for characteristic pulse sounds in the right brachial artery. When the measurement is performed manually using this technique, an operator inflates the cuff until the brachial artery is completely occluded. The cuff is then slowly deflated while the operator listens with a stethoscope to the brachial artery below the cuff for sounds which identify the two end points. He notes from a manometer the pressures at which these characteristic sounds are heard. The accuracy of this method is very dependent on the hearing acuity and experience of the operator. Because of this human factor an error of up to 8 mmHg may be expected in both end point readings under normal conditions.[2]

In an effort to make the measurement of blood pressure simpler to perform, more reliable, and more accurate, many attempts have been made to automate the procedure. Here again, the most commonly used method is the auscultatory method. A microphone is placed over the brachial artery below the cuff and the microphone output signal is passed through sensing circuitry which detects when systolic and diastolic pressures are reached. Several problems are encountered in automating the auscultatory method. Microphone placement is critical, patient movement and background noise sometimes cause erratic operation, and often the electronic circuitry has to be readjusted for each subject.[2]

Although it is not inconceivable that all the above problems associated with automating this method could be overcome, searches are constantly being made for relationships other than characteristic sounds which could be used to identify the end points. These searches concentrate on characterizing the diastolic pressure, which is not nearly so sharply defined as is the systolic pressure.

In 1960, Marx reported a method of measuring blood pressure which uses two pneumatic cuffs placed adjacent to each other on the upper arm.[3] In this method the diastolic pressure of the subject is determined by subtracting the pressure pulsation signal in the lower (distal) cuff from the pulsation signal in the upper (occluding) cuff. Marx reported that when the static pressure in the cuffs is below diastolic pressure this difference signal is very small. As the cuff pressure passes through the diastolic pressure of the subject, however, the difference signal amplitude increases very rapidly. The onset of a large difference signal can therefore be used to identify the diastolic point. The systolic pressure is the minimum pressure at which no pulses are received in the distal cuff.

Marx's explanation for the difference signal is that below diastolic pressure the artery is virtually unobstructed and the pulses arrive at the two cuffs almost simultaneously and with the same shape. There is thus no appreciable difference signal. As the cuff pressure passes through diastolic pressure, the pulse begins to encounter resistance in passing through the occluding cuff. The pulse signal from the distal cuff begins to lag behind the pulse signal from the occluding cuff, resulting in a momentary difference signal coinciding with the rising edge of the occluding cuff pulse.

The oscillometric* method has potential advantages over the auscultatory method. There are fewer inherent problems which arise in implementing the oscillometric method. The problems of microphone placement and background noise are nonexistent in the oscillometric method. Furthermore, it is likely that the oscillometric method could be implemented using simpler circuitry than is required for the auscultatory method, because in the latter the circuitry must, in essence, perform a frequency analysis of an audio signal. The simpler the circuitry is, the more reliable and economical it will be.

Marx had a test apparatus built to measure blood pressure using his oscillometric technique. The apparatus reportedly worked, but it was not practical as a clinical sphygmomanometer. To be practical as a clinical sphygmomanometer, a system should possess the following properties:

1. Accuracy
2. Simplicity of operation
3. Portability
4. Reliability
5. Quietness of operation
6. Economic practicality.

A study of Marx's apparatus (which was available as a reference for this thesis) and its scheme of operation indicated that there was a possibility that many of its shortcomings might be overcome by a

*Oscillometric--pertaining to the measurement of oscillations or pulsations.

well-designed system using solid-state electronics, especially integrated circuits, along with miniature pneumatic components.

All the above being considered, it is the objective of this thesis to determine the feasibility of developing an automatic sphygmomanometer system which uses the oscillometric method described to detect diastolic and systolic pressures, and which possesses properties that make it suitable for clinical use.

CHAPTER II

THE PNEUMATIC SYSTEM

The sphygmomanometer requires two subsystems, a pneumatic system and an electronic system. The pneumatic system is the central part of the sphygmomanometer, and the electronic system is a peripheral system which controls and monitors the pneumatic system. Once a pneumatic system has been completely defined, many characteristics of the electronic system will have been determined. Refer to Fig. 2-1.

Two pneumatic cuffs are needed--the occluding cuff on the upper arm and the distal cuff on the lower arm. The cuffs used are of the size and type used by Marx.[3] The occluding cuff is considerably larger than the distal cuff to make the pressure pulsations felt in each cuff approximately the same in magnitude.

The cuffs are to be inflated from a common source simultaneously. However, the pulses originating in either cuff must not be allowed to cross over to the other cuff. In other words, there must be a low pass coupling between the two cuffs. This is accomplished by placing a needle valve in the line with each cuff and placing an air capacitor between the two valves.

A pressure source is needed. Inflating the cuffs directly from a small air compressor is ruled out because pulsations in the compressor output would interfere with the operation of the system. Instead, an air compressor is used to pressurize a supply tank prior to initiating the measurement cycle. The supply tank provides a pressure source which is completely free from ripple. Some kind of pressure switch

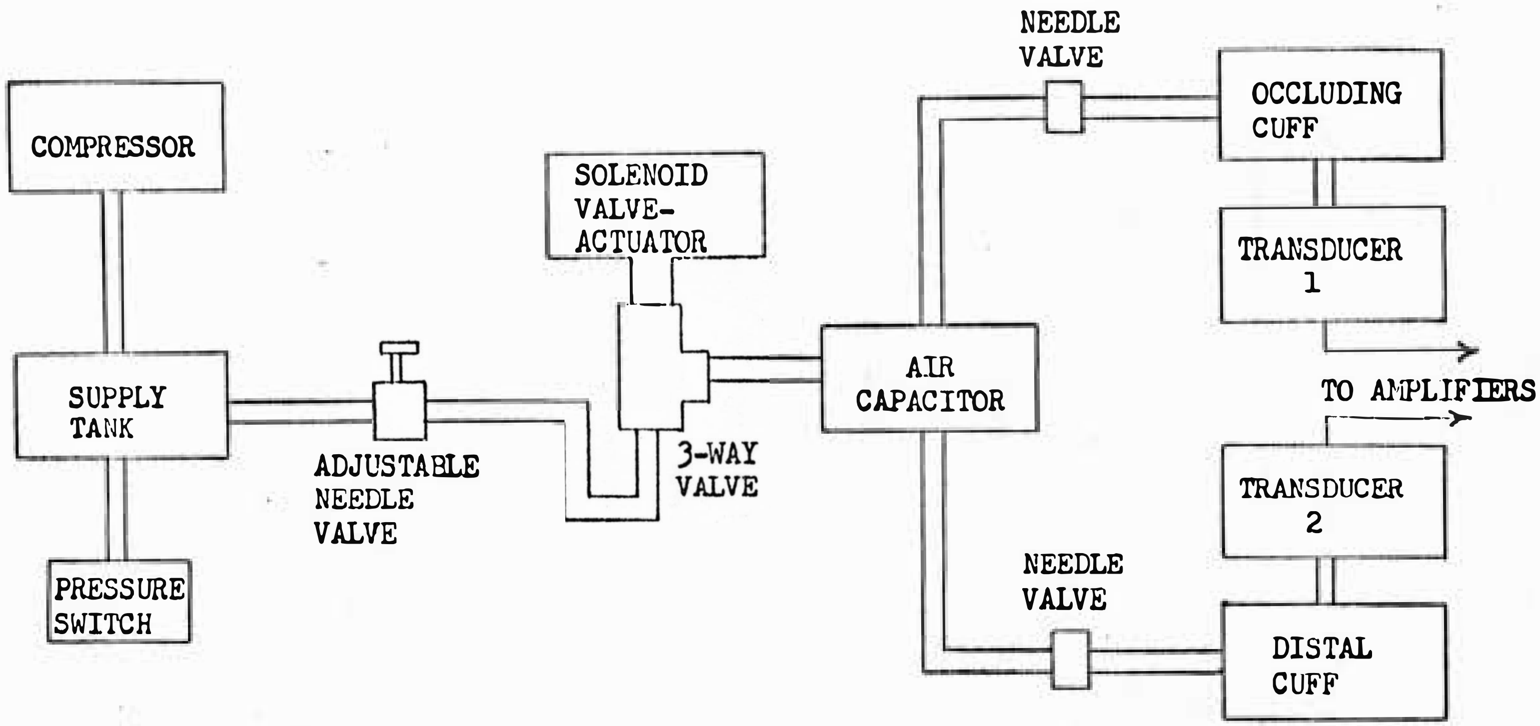

Fig. 2-1. The pneumatic system.

must be used to signal when the tank has sufficient pressure to carry out a measurement cycle. For a 100 cu. in. tank, 500 mmHg pressure proved to be sufficient.

The inflation-deflation scheme chosen determines a great deal about how complicated the sphygmomanometer system becomes. For instance, Marx used a four-step cycle. The diastolic pressure was determined as the cuffs were being inflated slowly. The cuffs were then rapidly inflated to about 250 mmHg to occlude the artery, then slowly deflated until systolic detection was completed. Finally, the cuffs were deflated rapidly to atmospheric pressure.

A simple pneumatic system is essential in order for the sphygmomanometer to be practical. Pneumatic parts are inherently bulky and heavy as well as expensive. The best scheme for this application is to obtain both pressure measurements as the cuffs are being slowly inflated and then rapidly deflate the cuffs. Besides minimizing the number of pneumatic components needed this scheme minimizes patient discomfort because the cuffs need never be inflated above the systolic pressure of the patient.

To control the inflation of the cuffs, a three-way solenoid actuated valve is used. This valve is placed in the system so that when the solenoid is activated the cuffs are coupled to the supply tank and when the solenoid is not activated the cuffs are coupled to the atmosphere. It is therefore necessary to control only one solenoid to control cuff inflation and deflation.

A gage pressure transducer on each cuff monitors the pressure in that cuff. An alternative would have been to use one differential transducer to obtain the difference signal pneumatically, but this proved to be much more costly than using two gage pressure transducers and obtaining the difference signal electronically.

The speed of cuff inflation is controlled by the adjustable needle valve in the supply line. Too rapid an inflation rate will make it impossible to get sufficiently accurate readings, but, on the other hand, the inflation rate must not be so slow that the measurement cycle takes excessively long.

CHAPTER III

THE ELECTRONIC SYSTEM

The electronic system must control and monitor the pneumatic system. It must switch the compressor and the solenoid on and off at the proper time, and detect and record the diastolic and systolic pressures. The electronic system can be separated into five sections according to function. They are:

1. Pneumatic control circuit
2. Amplifier circuit
3. Diastolic detection circuit
4. Systolic detection circuit
5. Pressure readout circuit.

A general concept of the system is developed before any of these sections is developed in detail. The functional block diagram of this system concept is shown in Fig. 3-1.

The pneumatic control circuit ensures that there is sufficient pressure in the supply tank before allowing the measurement cycle to begin. As soon as the systolic reading has been taken, the cuffs are automatically deflated. It is also desirable that the operator be able to deflate the cuffs and terminate the measurement cycle at any time during the cycle. This circuit therefore requires a start input, an interrupt input, an input from a pressure switch on the supply tank, and an input from the systolic detection circuit. It has two outputs--one controlling the compressor and one controlling the solenoid.

It is desired to have digital readouts of the two end point

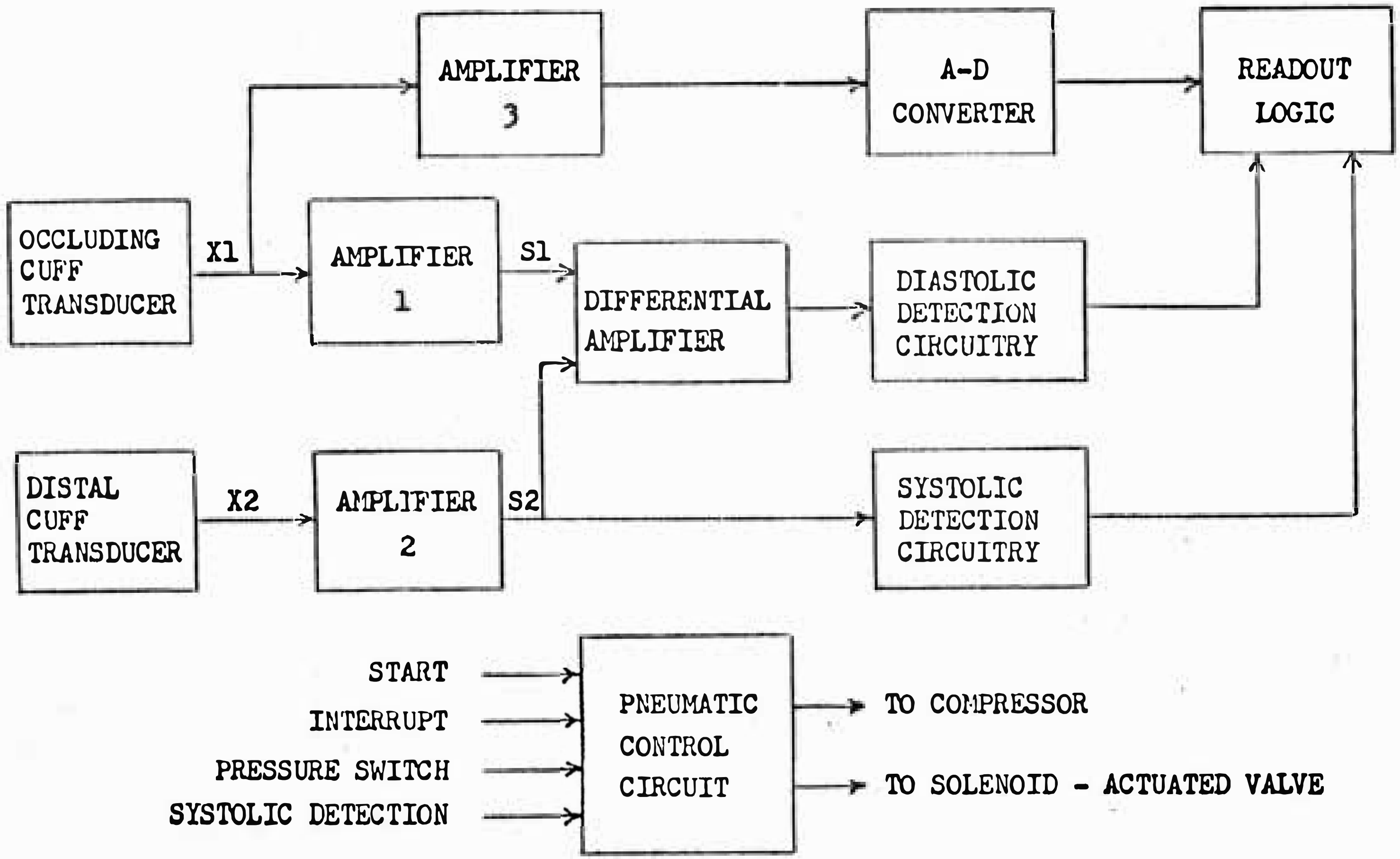

Fig. 3-1. Electronic system block diagram.

pressures. The output of the occluding cuff transducer is scaled by amplifier 3 so that the output of the analog-to-digital (A-D) converter is a digital representation of the cuff pressure in mmHg in binary-coded decimal (BCD) form. The readout logic consists of storage registers, decoders, and the digital displays. The storage registers follow the converter output until a hold signal is received from their respective detection circuits.

The diastolic detection circuit recognizes from the difference signal S1-S2 when the diastolic pressure has been reached. According to Marx, this signal should be of very small amplitude until the diastolic point is approached, and should increase in amplitude very rapidly as the diastolic pressure is passed. When the amplitude of S1-S2 exceeds a certain level, then, the diastolic detection circuit sends a hold signal to the diastolic readout circuit.

The systolic detection circuit detects the cessation of pulses coming from the distal cuff. It sends a hold signal to the systolic detection circuit when the amplitude of S2 drops below some threshold level.

Amplifiers 1 and 2 are AC coupled. Due to variations expected in the transducer outputs X1 and X2, the gain of these amplifiers must be regulated. If the amplifier outputs S1 and S2 are not the same amplitude as one another at all times, it is possible that the diastolic detection circuit might be triggered falsely before any relative shift occurs in the distal pulse. An automatic gain control is therefore incorporated into amplifiers 1 and 2. After diastolic detection, however, the gain of amplifier 2 must be held constant. It must not be allowed to compensate for the expected decrease in X2 as systolic pressure is reached.

Each of the sections will now be developed in detail.

3-1. PNEUMATIC CONTROL CIRCUIT.

The necessary sequence of operation of the pneumatic control circuit is as follows:

1. Recognize start signal.
2. Start compressor if pressure switch on supply tank is open. (Assume normally open switch with some hysteresis.)
3. Shut off compressor and activate solenoid to inflate cuffs when pressure switch closes. If the pressure switch is closed when the start signal is received, this step is initiated without first starting the compressor.
4. Deactivate solenoid to deflate cuffs immediately after the systolic reading has been taken. Also deflate the cuffs if the operator chooses to interrupt the measurement cycle.

The circuit designed to carry out these operations is shown in Fig. 3-2. The circuit uses 5 volt NAND logic to control triacs which do the necessary switching. The inputs to the logic are defined as:

$\overline{\text{PRESSURE SWITCH}}$--the input from the pressure switch

$\overline{\text{START}}$--the input from the start switch

$\overline{\text{INTERRUPT}}$--the input from the interrupt switch

$\overline{\text{SYSTOLIC DETECTION}}$--the input from the systolic detection circuit.

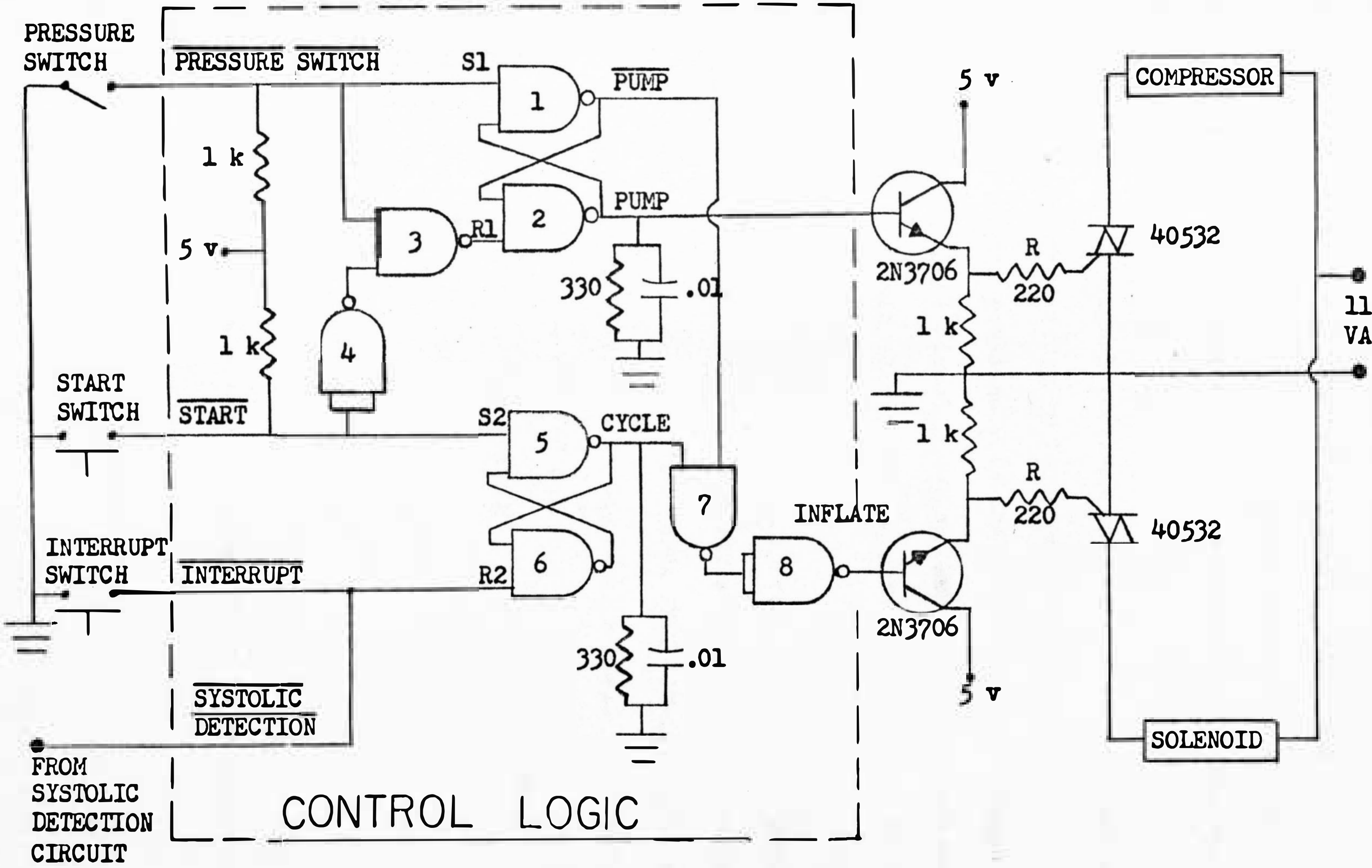

Fig. 3-2. Pneumatic control circuit.

The start and interrupt switches are both normally open momentary pushbuttons. The systolic detection circuit output goes low when systolic pressure has been reached. By the definition of positive logic, the following definitions of signals received by the pneumatic control circuit also hold:

PRESSURE SWITCH--signal received when the pressure switch closes

START--signal received when the start switch closes

INTERRUPT--signal received when the interrupt switch closes

SYSTOLIC DETECTION--signal received when systolic detection occurs.

Two NAND latches are used as storage elements. The outputs of these latches are defined as:

PUMP--signal indicating that the compressor should be running

CYCLE--signal indicating that a START signal has been received and that neither an INTERRUPT nor a SYSTOLIC DETECTION signal has been received since the START signal. A measurement cycle is to be carried out.

The latch inputs are defined as follows:

S1--the set input which sets $\overline{\text{PUMP}}$

R1--the reset input which sets PUMP

S2--the set input which sets CYCLE

R2--the reset input which sets $\overline{\text{CYCLE}}$.

The INFLATE signal indicates that the solenoid is to be activated to inflate the cuffs.

The logic equations to be implemented are:

1. $S1 = \overline{\text{PRESSURE SWITCH}}$

2. $R1 = \overline{\overline{\text{PRESSURE SWITCH}} \cdot \text{START}}$

3. $S2 = \overline{\text{START}}$

4. $R2 = \overline{\text{INTERRUPT} + \text{SYSTOLIC DETECTION}}$

5. $\text{INFLATE} = \overline{\overline{\overline{\text{PUMP}} \cdot \text{CYCLE}}}$

Equations 1 and 3 are implemented by direct wiring. It is assumed that the SYSTOLIC DETECTION signal will be the output of a transistor-transistor logic (TTL) circuit. Although this signal will be normally high, it can be shorted to ground without any chance of harming the TTL. Equation 4 is therefore implemented by tying the $\overline{\text{INTERRUPT}}$ and the $\overline{\text{SYSTOLIC DETECTION}}$ inputs together. R2 goes low when either an INTERRUPT or a SYSTOLIC DETECTION signal is received. Equation 2 is implemented by gate 3 with gate 4 supplying the START signal. Equation 5 is implemented by gates 7 and 8.

An emitter follower is used to provide current gain to trigger the triacs. The value of the resistors R is determined from the equation

$$R = \frac{V(1) - V_{be} - V_g}{I_g}$$

where

$V(1)$ = logical one voltage level

V_{be} = voltage on the triac gate when the triac is conducting

I_g = triac gate current needed for dependable triggering.

For a silicon transistor and a 40532 RCA triac this becomes

$$R = \frac{(3.8 - 0.6 - 1.0)\text{v}}{10 \text{ ma}} = 220 \text{ ohms}$$

The .01 uf capacitor and the 330 ohm resistor on the outputs of gates 2 and 5 prevent these outputs from setting up high when V_{cc} is first applied to the circuit. Thus neither the solenoid nor the compressor will turn on until a START signal is received.

3-2. AMPLIFIER CIRCUIT.

The amplifier circuitry consists of the DC amplifier, the AC amplifier from each transducer, and the differential amplifier. The SN72709 integrated circuit operational amplifier (op amp) is used for all of these except the automatic gain control (AGC) stage of the AC amplifiers. The SN72709 has a differential input, a single-ended output, and has an input which allows adjustment of the offset voltage.

The required gain K of the DC amplifier is determined from the output voltage S of the transducer per mmHg and from the resolution* R of the A-D converter by the relationship

$$K = R/S.$$

The transducer used has an output of .050 mv/mmHg (S = 0.05 mv) and the resolution of the A-D converter used is 10 mv. The gain required in this application is therefore 200. This gain is obtained using the standard feedback configuration as shown in Fig. 3-3, which gives a differential gain of $K = R_f/R_{in}$. The values of the frequency compensation elements are obtained from the manufacturer's recommendations.[4]

The devices which are used in obtaining the pressure reading--the transducer, the op amp, and the A-D converter--all can be expected to have some offset voltage. It is therefore necessary to provide a way

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*Resolution is defined as the voltage represented by a 1 in the least significant decimal position in the output of the A-D converter.

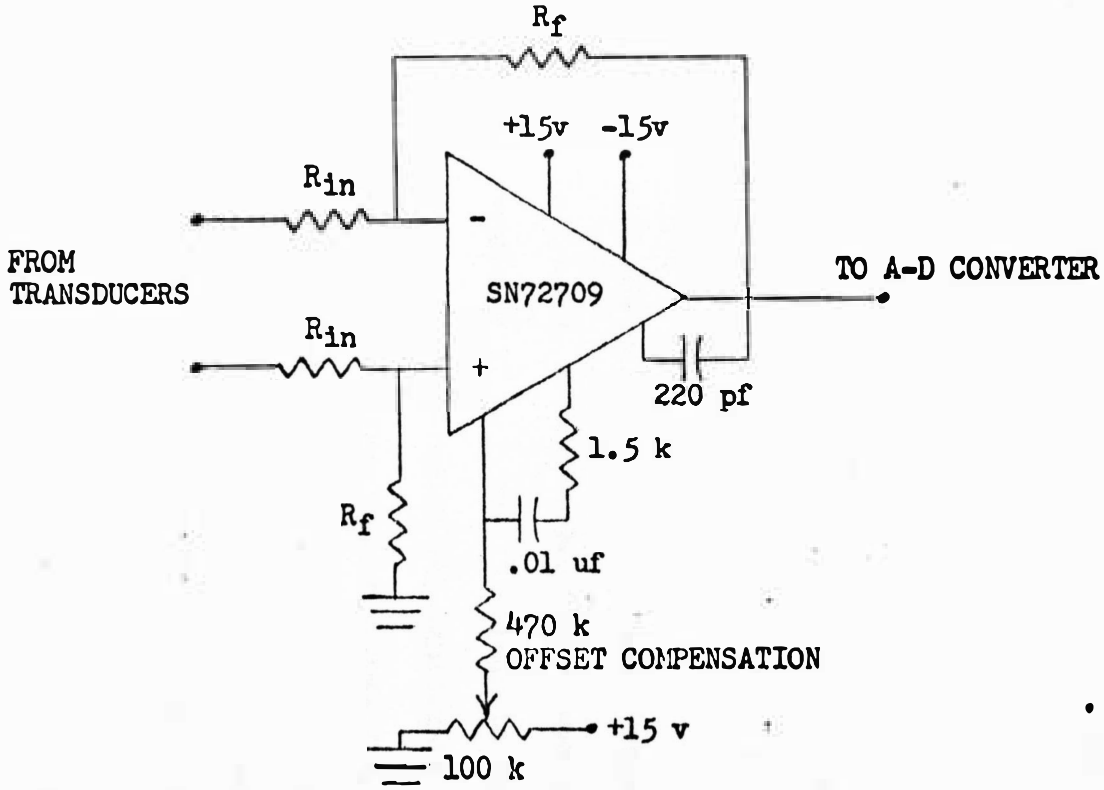

Fig. 3-3. DC amplifier.

to zero the reading when there is no pressure in the cuffs. The offset adjust circuit shown does this very effectively. Adjusting the potentiometer adjusts the bias current on one of the transistors internal to the op amp, thus adjusting the DC output level.

It is known that the amplitudes of the transducer outputs X1 and X2 will vary with the tightness of the cuffs on the arm, the physical characteristics of the arm, the size of the cuffs, and the pressure in the cuffs.[3] This implies that, in order for level detection to be a dependable way of locating the end point pressures, the gains of the AC amplifiers must be regulated.

It is also known that the maximum pulse amplitude which can be expected in the cuffs is 3 mmHg, with a corresponding maximum pulse amplitude out of the transducer of 0.15 mv. In order to get an approximate idea of the amount of gain required of the amplifiers, it is assumed at this point that the amplifier output signal should have an amplitude of at least 3 volts. The minimum gain required of the AC amplifiers is therefore 20,000. The maximum gain of the amplifiers is not so straightforward to compute. The signal amplitude out of the transducers is zero at 0 mmHg and increases gradually as the cuffs are inflated. The maximum gain required of the circuit therefore depends on the pressure at which it is decided to begin the measurement cycle. This pressure must be chosen as the minimum pressure at which an acceptable signal level is received from the transducers, which should be about 20 mmHg.[2] Due to lack of data on the magnitudes of pulse amplitudes which may be expected at 20 mmHg, the maximum gain is left as a variable to be determined.

The frequency band which must be passed by the amplifiers is not precisely known at this point. A few observations can be made, however. The lower cutoff frequency can be made low enough to pass the frequency of the lowest heartbeat which might be encountered, i.e. about 0.1 Hz. This may be lower than necessary but would ensure that all lower frequencies would be passed. The upper cutoff frequency cannot be chosen quite so easily. The higher frequency content of the pulsations is not known, but it would be advantageous if the upper cutoff frequency could be set below 60 Hz. Otherwise one might expect interference from 60 Hz. pickup to be a significant problem. The upper cutoff frequency is also left as a variable to be determined. The schematic for each of the AC amplifiers is the same and is shown in Fig. 3-4.

The first stage provides a gain of 1,000. The upper cutoff frequency of the amplifiers is determined by the parallel combination of the 10 Meg resistor and the capacitor C which provide the feedback for this stage. This upper cutoff frequency will be $1/20\pi C$ Hz. where C is in microfarads.

The second stage has provision for AGC. The 2N3860 transistor provides the amplification. The 2N3820 field-effect transistor (FET) in series with the 15 k resistor supplies a constant emitter current, so that the dynamic loading and the gain of the circuit are determined almost entirely by the value of the resistor R4 and the drain-to-source resistance of the 2N2498 FET.[5] The resistor R4 is used to limit the maximum gain of the amplifier. With R = 0, the gain of the AGC stage

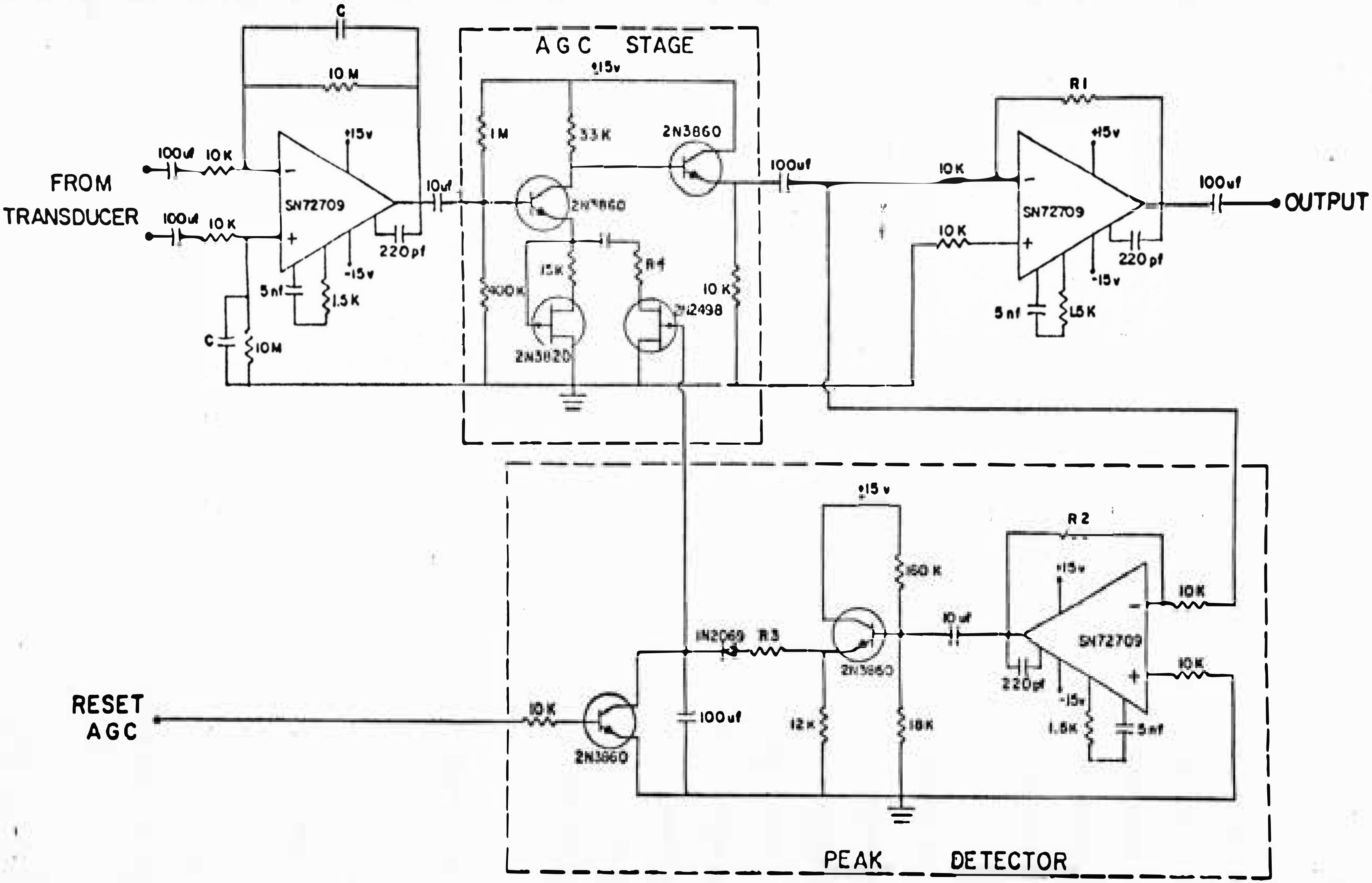

Fig. 3-4. AC amplifier.

varies from 60 to 1 as the voltage on the gate of the 2N2498 is varied from 0 to 5 v. The value of R4 is left to be determined.

The peak detector voltage is independent of the amplifier output amplitude. The gain of the op amp in the peak detector circuit must be set (by choosing the appropriate value for R2) so that the 2N2498 will be operating in its control region. The value of R1, which determines the gain of the output stage, must be chosen according to the voltage levels at which the detection circuitry is set to trigger.

An emitter follower is placed in the peak detector circuit to provide a high impedance looking from the op amp so that low frequencies will not be attenuated. A resistor R3 is placed in series with the diode to make the AGC circuit less susceptible to artifacts. The value of R3 is chosen such that several pulses of a given voltage amplitude must be received before the peak detector output will approach that voltage. If this precaution were not taken, a single artifact could reduce the gain of the amplifiers enough to cause malfunction of the system. The value of R3 is also left to be determined.

The reset signal for the AGC is obtained from the output of gate 7 in the Pneumatic Control Circuit. This signal goes high whenever the solenoid is not activated, and holds the peak detector output voltage at zero. When cuff inflation begins, this signal goes low and allows the peak detector to build up a voltage at its output.

The differential amplifier is shown in Fig. 3-5. It has a differential gain of unity with $R_f = R_{in} = 10$ k. Its output is the occluding cuff signal minus the distal cuff signal.

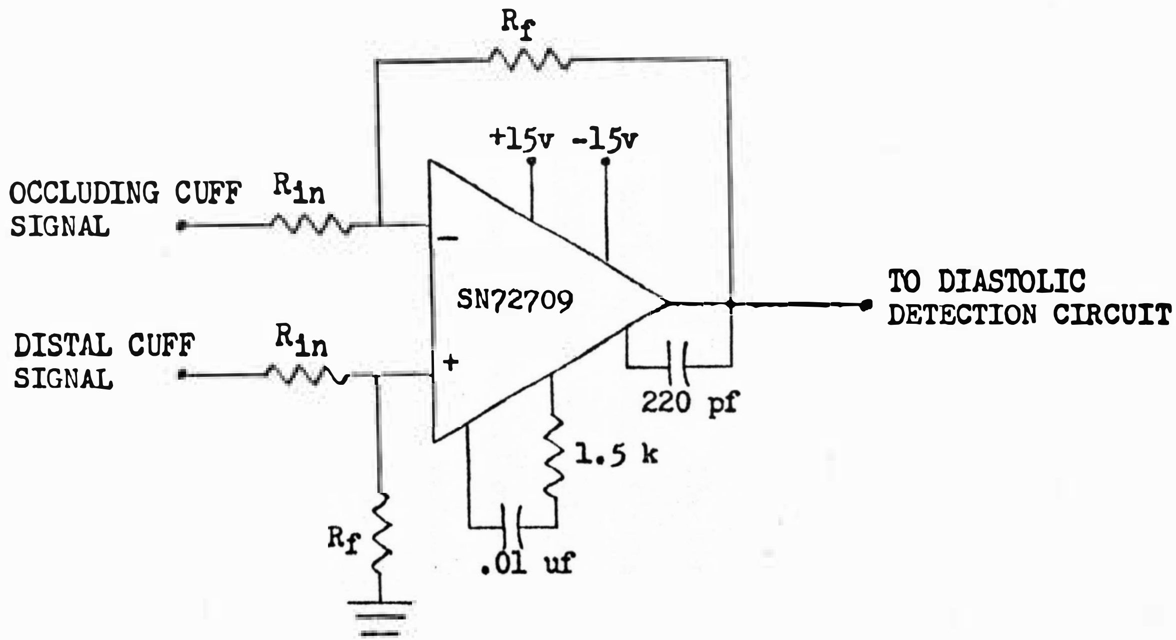

Fig. 3-5. Differential amplifier.

3-3. DIASTOLIC DETECTION CIRCUIT.

The diastolic end point will be indicated by the appearance of a large signal amplitude at the output of the differential amplifier. Due to the presence of artifacts, however, it is not possible to assume that diastolic pressure has been reached when the first large pulse is received from the differential amp. Rather, it is necessary to check a series of pulses to see if they coincide with the heartbeat. If the pulses originate from the heartbeat, then a given number will be received within a given span of time. The circuit of Fig. 3-6 is designed to check for the authenticity of pulses using this principle. It uses two TTL one-shots and a shift register composed of three D-type edge triggered flip-flops. The SN74121 one-shot used has a B input which can be used for level detection, and has both positive (Q) and negative ($\overline{Q}$) going output pulses available. The output pulse width is set by connecting the proper values of resistance and capacitance across the timing inputs to the one-shot.[6] The SN7474 D-type flip-flop transfers data from the D input to the Q output when the clock goes from a logical 0 to a logical 1. It has a clear input which when held low will cause the Q output to remain at a logical 0.

The trigger level of the one-shot is 1.5 v. This voltage is considered reasonable for detection of diastolic pressure. The restriction it imposes on the AC amplifiers is that their output amplitudes must never be allowed to differ by an amount approaching 1.5 volts or the circuit will trigger falsely. It also assumes that the pulse amplitude of the amplifiers is near 3 volts so that at diastolic

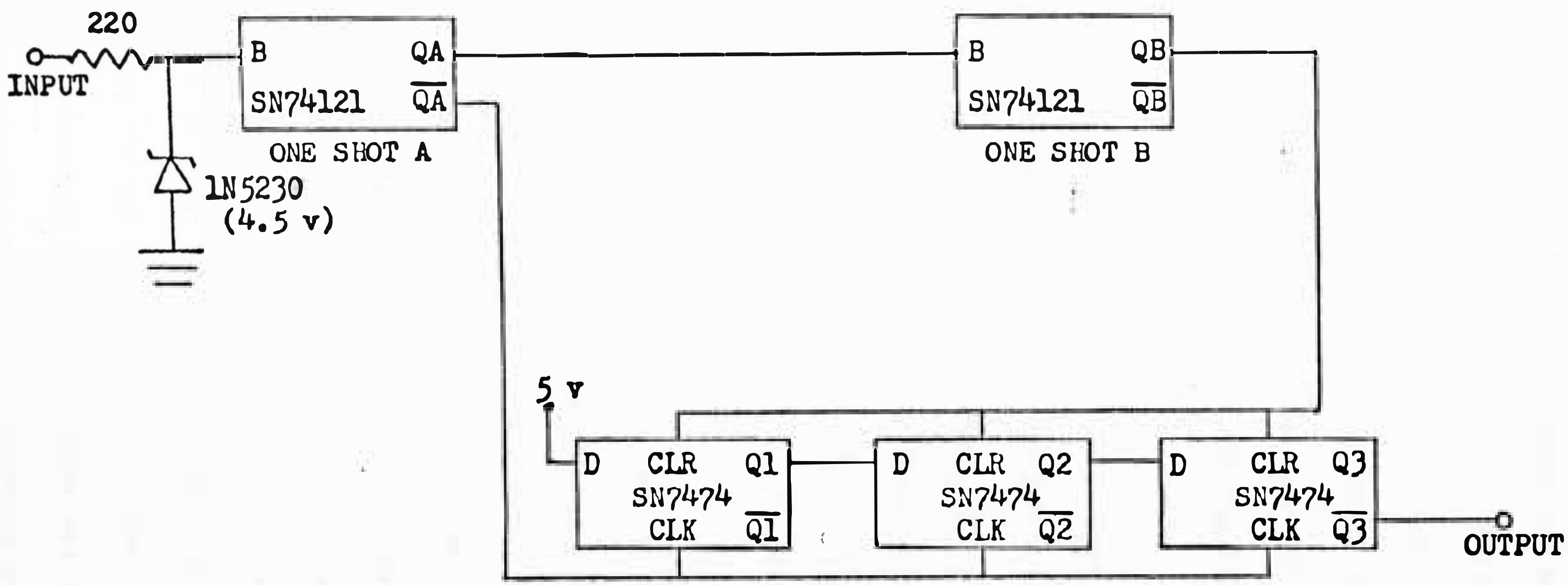

Fig. 3-6. Diastolic detection circuit.

pressure the difference signal will indeed exceed the 1.5 volt threshold. Any tendency for the circuit to trigger either too early or too late can be compensated by adjusting the output level of the amplifiers.

The sequence of operation of the circuit is now described. One-shot A puts out a pulse when its input rises above the 1.5 volt detection level. This in turn triggers one-shot B causing QB to go high removing the clear from the flip-flops. The pulse width of QA is much shorter than that of QB. When one-shot A returns to its normal state $\overline{QA}$ goes from a logical 0 to a logical 1 and clocks the flip-flops. This transfers a logical 1 to Q1. If two more pulses are detected by one-shot A before QB returns to a logical 0, the output of the circuit Q3 will go low, signifying that diastolic pressure has been reached. If two more pulses are not received during the pulse time of one-shot B, QB will clear the shift register before the output of the circuit changes state, and the cycle will repeat.

The output pulse width of one-shot A is set at 0.3 seconds. The TTL one-shot ignores the input while it is in its triggered state. This circuit therefore will not count a pulse which occurs within 0.3 seconds after the last pulse counted. This sets an upper limit on the maximum heart rate for which this system may be used at 200 beats/minute. The output pulse width of one-shot B is set at 3.3 seconds. This puts an upper limit on the pulse period which will be accepted. If the time between pulses is greater than 1.5 seconds, one-shot B will reset the flip-flops before three pulses are counted. A lower limit on the heart rate for which this system may be used is therefore set at 40 beats/minute. The heart-rate limits are easily adjusted by changing the pulse widths of the one-shots.

A 4.5 volt zener diode is used on the input to one-shot A (which comes from 15 volt circuitry) to protect its input transistors from possible overvoltages. The 220 ohm resistor is to limit the current out of the differential amp in the event its output should rise above 4.5 volts.

3-4. SYSTOLIC DETECTION CIRCUIT.

The systolic detection circuit must determine when pulses disappear from the distal amplifier, and signal the systolic readout to hold the systolic pressure reading. A circuit which does this is shown in Fig. 3-7.

This circuit is essentially a retriggerable monostable. As long as pulses are received at its input, its output remains at logical 1. Some set time interval after the last pulse is received the output of the circuit goes to logical 0.

This particular circuit uses a detection level of 1.2 v. Before systolic pressure is approached in the measurement cycle the input to the circuit is a pulse train with amplitude greater than 1.2 v. Each time a pulse is received the one-shot puts out a low level pulse which brings Q3 out of saturation and allows C to charge to 4.5 volts through Q4. The one-shot pulse width is set at 0.3 seconds, but is not critical. The purpose of the one-shot is to make the circuit independent of the time which each pulse remains above the detection level. Emitter follower Q5 provides a high impedance looking from C. When the output of the one-shot returns to its normal state, Q3 is again saturated and

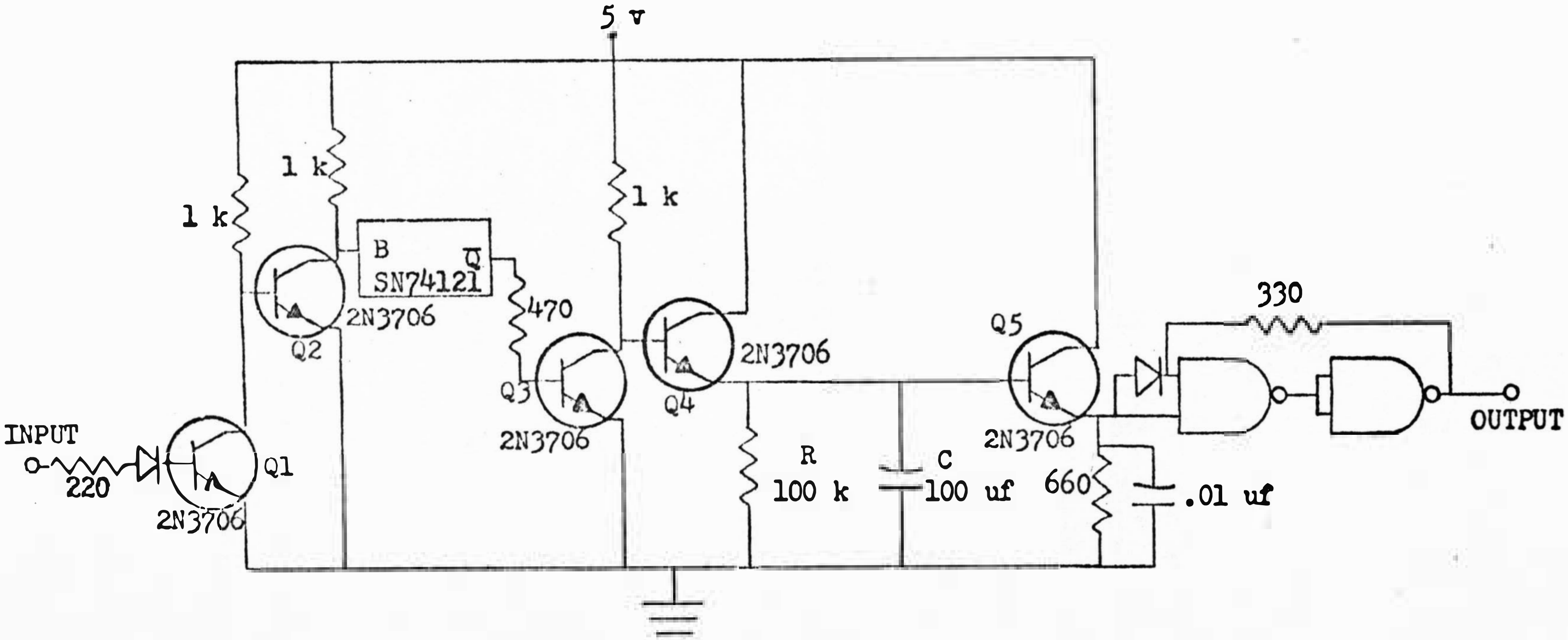

Fig. 3-7. Systolic detection circuit.

the base to emitter diode on Q4 becomes reverse biased. C discharges slowly until the next pulse comes along and then recharges to 4.5 v through Q4.

The NAND gate configuration is a Schmitt trigger with a trigger level of 1.5 v. Its output goes from a logical 1 to a logical 0 when the voltage at its input drops past the 1.5 v level. The Schmitt trigger input follows the capacitor voltage through the emitter follower. The RC time constant is such that the capacitor will not discharge very far between pulses. Only if two consecutive pulses are missed will the capacitor discharge enough to cause the Schmitt trigger output to go low and signal systolic detection.

The 220 ohm resistor is used to limit the current from the distal cuff amplifier. The input to the one-shot is protected from over-voltages by a 4.5 v zener diode. The .01 uf capacitor prevents high frequency oscillations from occurring in the circuit.

The 1.2 v detection level was chosen at a level which was considered high enough that pulsations crossing over from the occluding cuff and amplifier output noise would not cause false triggering.

3-5. PRESSURE READOUT CIRCUITRY.

The readout circuitry includes the A-D converter and all the storage, decoding, and latching circuitry necessary to obtain digital pressure readings. The outputs of the AC amplifiers are undependable before 20 mmHg cuff pressure is reached, so this circuit must include a way to prevent any reading from being held before this pressure is reached.[2]

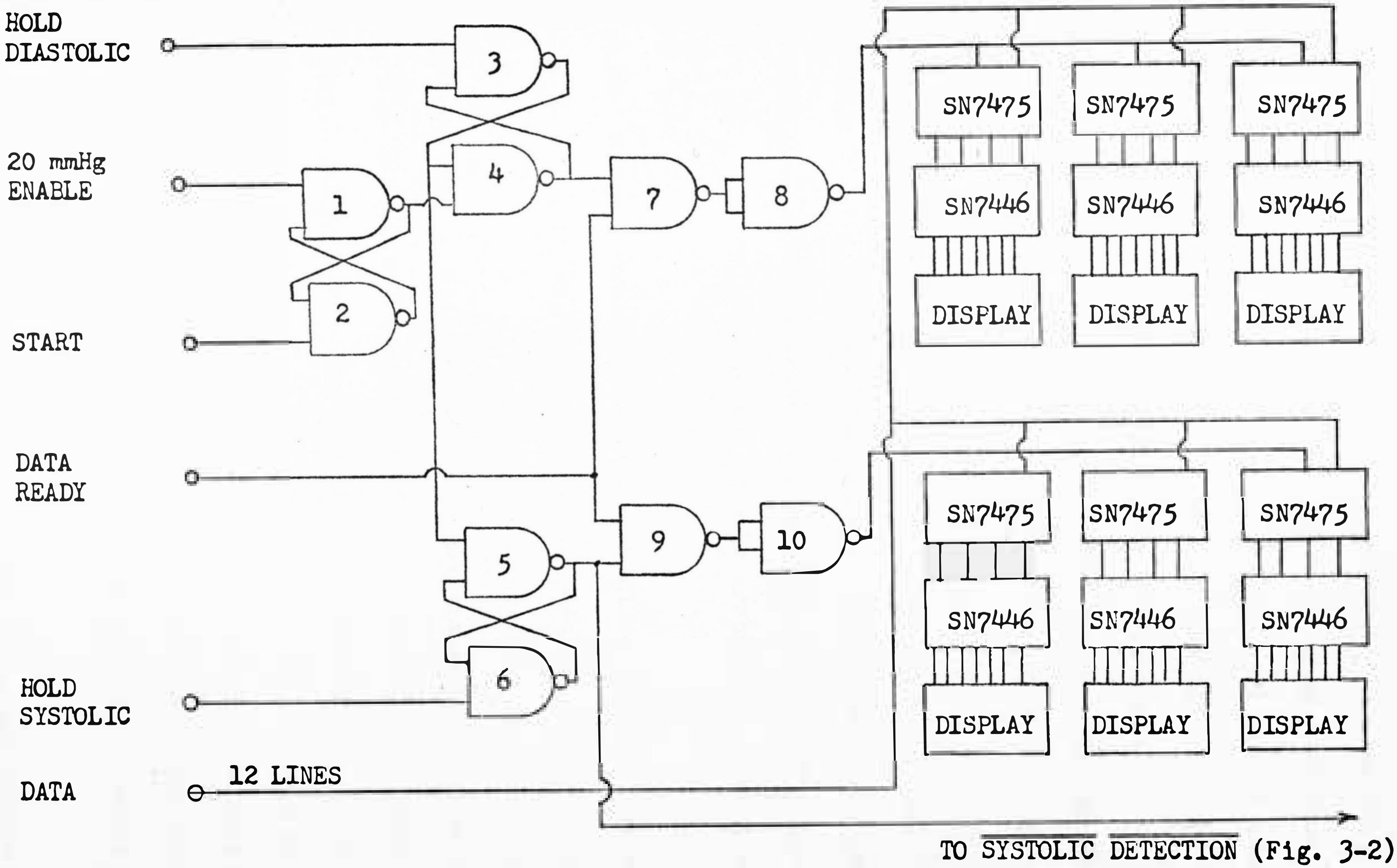

Fig. 3-8. Pressure readout circuitry.

The circuit used is shown in Fig. 3-8. The A-D converter has BCD outputs and also has a data ready signal which goes high when the BCD outputs constitute a valid reading. A storage register is used for each readout. Each storage register consists of three SN7475 4-bit bistable latches. When the clock input to these registers is high, the outputs follow the inputs; when it goes low the data at the outputs is stored. The output from the registers goes to SN7446 BCD-to-seven-segment decoder-drivers which drive the seven segment displays.

Gates 3 and 4 make up the diastolic latch; gates 5 and 6 make up the systolic latch. The latch composed of gates 1 and 2 is used to disable the diastolic and systolic latches when the cuff pressure is below 20 mmHg. The source of the 20 mmHg enable signal will be discussed later.

When the start switch is momentarily closed, the output of gate 1 is set low. As long as this is low the diastolic register is forced to follow the data. This in turn forces the systolic register to follow the data because the systolic latch is not enabled until the diastolic reading has been locked in and the output of gate 3 goes high. At 20 mmHg a low input to gate 1 enables the diastolic latch. When the diastolic hold signal goes low the diastolic pressure reading is locked in and the systolic latch is enabled. When the systolic hold signal goes low the systolic pressure reading is locked in. The output of gate 5 is also sent to the pneumatic control circuit to cause the cuffs to be deflated, and the cycle is complete.

The data ready signal is used to enable the clock to the registers through gates 7 and 9 so that the output of the registers always constitutes a valid reading.

A signal which goes from a logical 1 to a logical 0 at 20 mmHg must be obtained from somewhere. There are several ways to do it, but the simplest is to obtain it from the output of the SN7475 in the tens position of either register. The complement of the next-to-the-least significant digit in the tens position, which is available directly from the SN7475, first goes from a 1 to a 0 when the decimal output of the register gets to 20. This digit is always a valid signal, and can therefore be used directly as the enable signal without incorporating any additional logic.

CHAPTER IV

RESULTS AND CONCLUSIONS

4-1. AMPLIFIER PARAMETERS.

The amplifier design left several component values as unknowns. (See Section 3-2 and Fig. 3-4.) These will now be determined.

The minimum gain required has been established as 22,000. R1 is chosen as 220 k, giving the output stage a gain of 22. When the gain of the AGC stage is reduced to unity, the gain of the amplifier is therefore 22,000.

The value of R2 must be chosen so that 2N2498 FET in the AGC stage is operating in its control region, and so that the gain of the AGC stage is reduced to one when the amplitude of the amplifier output signal reaches the 3-volt level. The 2N2498 gate voltage which reduces the gain to unity varies with the particular FET used. The value of R2 can be calculated as

$$R2 = (R1)(V_{gc} + 1.2 \text{ v})/V_{out}$$

where

V_{gc} = the gate voltage in volts which reduces the gain of the AGC stage to unity. This must be determined by testing each 2N2498 separately.

V_{out} = the maximum amplitude in volts which is desired out of the amplifier.

The 1.2 volt quantity accounts for the diode drop in the peak detector circuit. For a V_{gc} of 3 volts, a V_{out} of 3 volts, and a corresponding

R1 of 220 k, R2 is equal to 300 k.

The maximum gain of the AGC stage was limited to 10 by setting R4 equal to 3.3 k. This in turn limits the maximum amplifier gain to 220,000. This is considered the practical limit because of the prominence of noise and artifacts in the amplifier output at gains exceeding 220,000.

The value of R3 must be determined somewhat intuitively. The duration of the pressure pulse occurring with each heart beat is approximately 0.02 seconds. However, the upper part of the pulse is of considerably shorter time duration, say roughly .01 seconds. The smaller the value of R3, the more accurately it detects the magnitude of each pulse. However, R3 must not be so small that a single artifact will reduce the gain of the circuit to a level at which the system cannot function properly. Also the pressure in the cuffs will be increasing slowly, so no rapid change in the pulse amplitude is expected. A relatively slow time constant is indicated. Satisfactory results were obtained with R3 equal to 8 k, giving the peak detector a time constant of 0.08 seconds.

The upper cutoff frequency of the amplifier is determined by the value of C on the first stage. The time constant of the feedback components on this stage must be small enough that the rapidly rising front edge of the pulse wave will not be distorted. Recordings taken on a Brush recorder showed that the rise time of the pulse wave is on the order of 0.03 seconds. The RC time constant is made a factor of 10 smaller to insure that the rising edge of the pulse is not attenuated, i.e. 0.003 seconds. The value of C is thus 300 pf.

4-2. CIRCUIT MODIFICATIONS.

Because of the experimental nature of this system it is desirable that a simple method of adjusting the threshold levels of the diastolic and systolic detection circuits be included. The exact threshold levels for each of the detection circuits will have to be determined by clinical testing. The circuit shown in Fig. 4-1 was therefore added at the inputs to each of the two detection circuits. It is an operational amplifier with a gain that can be adjusted by adjusting the 20 k potentiometer. The gain is adjustable from 0.5 to 2.5. The detection level of the diastolic detection circuit is therefore adjustable from 0.7 v to 3 v; the detection level of the systolic detection circuit is adjustable from 0.5 to 2.4 v.

The 20 mmHg level for enabling the detection circuitry was found to be too low. The signal is too noisy and unreliable at that pressure. It is desirable to delay the enable to 40 mmHg, as the signal is almost always strong by the time this pressure is reached in the cuffs. This would still allow measurement of almost all diastolic pressures. If the diastolic pressure does happen to lie below 40 mmHg, the diastolic detection circuit will be receiving a pulse train as the enable signal is received and will therefore cause the diastolic reading to be held soon after the 40 mmHg level is passed. The system can then complete the measurement cycle in normal fashion.

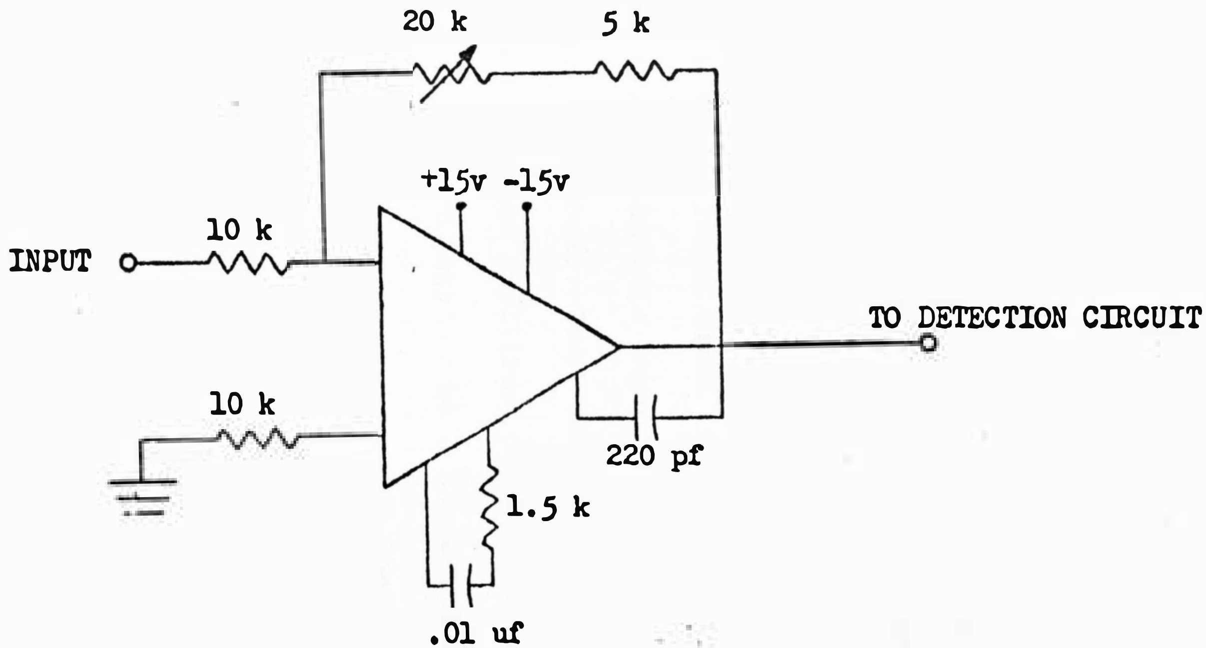

Fig. 4-1. Detection level adjust.

4-3. SYSTEM PERFORMANCE.

Fig. 4-2(a) shows a recording of the amplifier outputs as the pressure in the cuffs increases from 60 to 95 mmHg. Amplifier parameters are as determined in Section 4-1. Channel A is the occluding cuff amplifier output. Channel B is the distal cuff amplifier output. Channel C is the differential amplifier output. Channel D is the cuff pressure.

The pulse amplitudes are well matched and relatively constant throughout the pressure range shown. The difference signal amplitude (the signal is inverted in this recording) is seen to increase quite slowly throughout the range. There is no point at which a sudden change is noted in the difference amplitude. This is in contrast with the results published by Marx, which show a large increase (about 100%) in the difference amplitude as the pressure in the cuffs passes from 5 mmHg below diastolic pressure to 5 mmHg above diastolic pressure. However, Marx's system had no AGC. A close look at his recordings indicates that a significant part of the change in the difference signal which is observed is due to 1) the fact that both pulse amplitudes are increasing as the cuff pressure increases and 2) the fact that the occluding pulse amplitude is increasing faster than the distal pulse amplitude. Either of these two factors alone would cause an increase in the difference amplitude.

The differencing process itself amplifies small changes in the amplitude of either pulse. For example, if some signal A has an amplitude of 11 units, and some signal B in phase with A has an amplitude of

10 units, the difference signal A-B will be 1 unit in amplitude. If signal A increases to 12 units and B remains unchanged, the difference signal will increase to 2 units. A 9% increase in A results in a 100% increase in the difference signal. The same argument holds if A leads B time-wise.

This effect was observed in recordings with this system in which the pulse amplitudes did not remain as well controlled as those shown in Fig. 4-2(a). The difference signal was found to be greatly affected by mismatches in the amplifier outputs. In test runs of the system on several subjects the diastolic trigger occurred anywhere from 40 mmHg to 100 mmHg. These results indicate the need for a much more precise AGC--one which continuously adjusts the pulse amplitude to some reference voltage. This can be accomplished by replacing the peak detector circuit of Fig. 3-4 with the circuit of Fig. 4-3. With this circuit a charge is added to the capacitor each time the pulse amplitude exceeds the pulse amplitude limit (the reference voltage) which in turn reduces the gain of the AGC stage. The signal amplitude at the amplifier output is therefore constantly adjusted downward toward the reference voltage level. The pulse amplitude is selected at 3 volts as was done in the former AGC scheme. This scheme also has the advantage that the signal amplitude is virtually independent of variations in the control characteristics of the 2N2498 FET's.

Fig. 4-2(b) shows the recordings of amplifier outputs from 95 mmHg of cuff pressure out through systolic pressure. It is the continuation of Fig. 4-2(a). The systolic point is much more obvious than is the

Fig. 4-2(a). Amplifier outputs.

Cuff Pressure: 60 mmHg to 95 mmHg.
Chart Speed: 25 mm/sec or 5 DIV/sec.

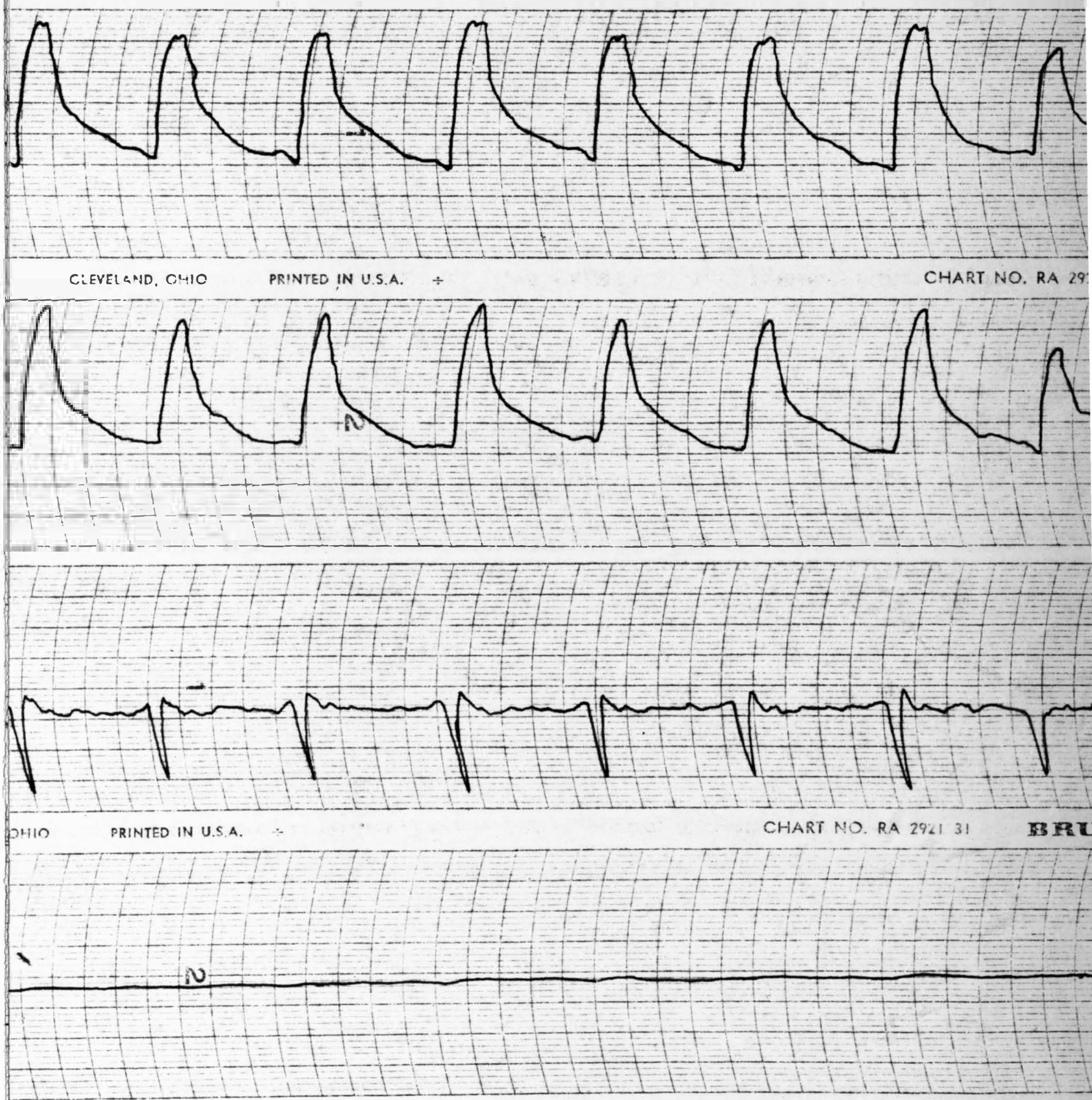
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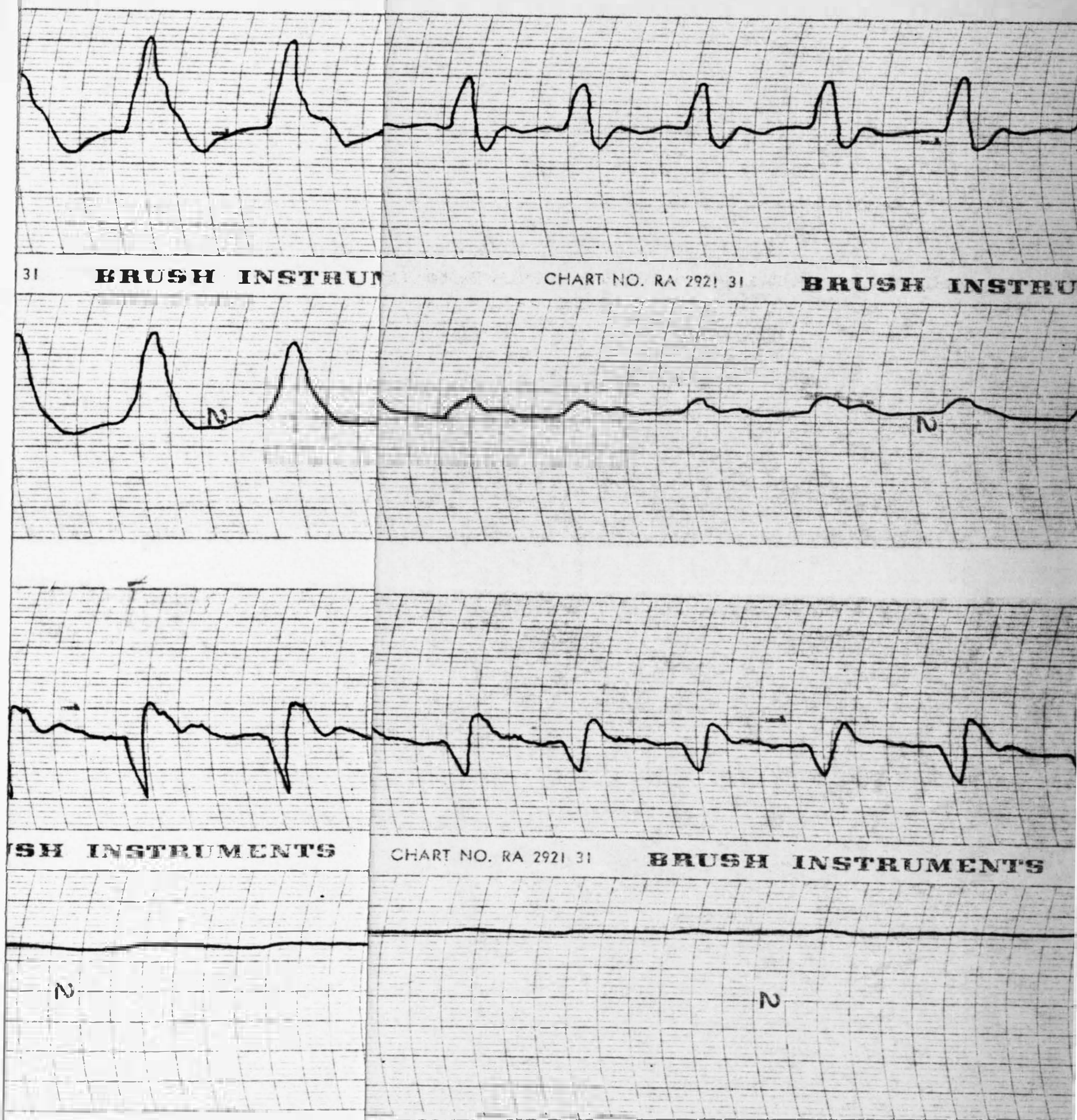
Fig. 4-2(b). Amplifier outputs.

Cuff Pressure: 95 mmHg through systolic.
Chart Speed: 25 mm/sec or 5 DIV/sec.



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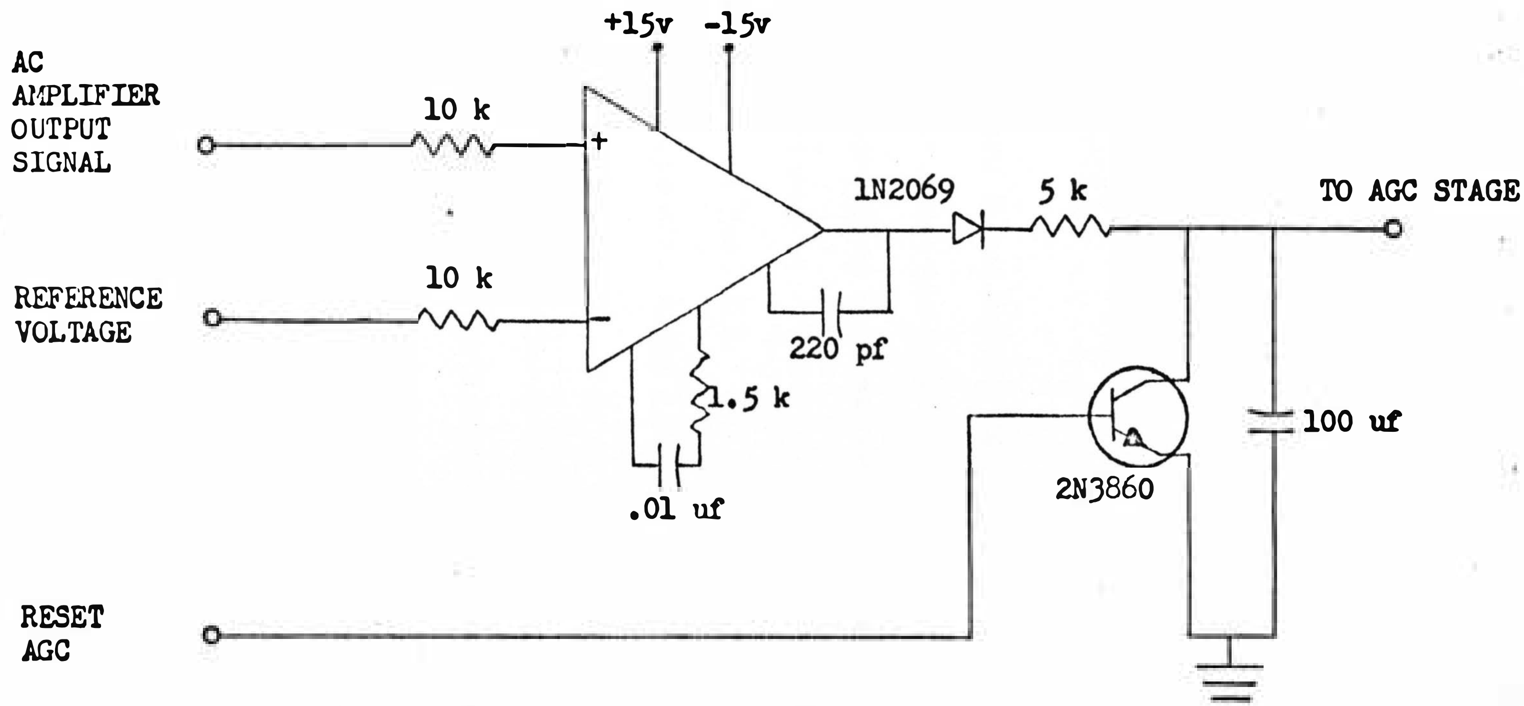

Fig. 4-3. Improved AGC scheme.

diastolic point. The pressure at which the distal pulse amplitude levels off to a small amplitude depicts systolic pressure. The small pulse amplitude which is observed above systolic pressure is due to crosstalk between the cuffs. In test runs of the system the systolic trigger occurred much more dependably and consistently than did the diastolic trigger.

4-4. CONCLUSIONS.

The system described in this thesis meets most of the requirements for an automatic sphygmomanometer. It is simple to operate. The use of solid-state circuitry and few mechanical parts lends itself to a system with high reliability. The system is compact and weighs about ten pounds, so it is portable. The two sources of audio noise, the compressor and the solenoid, could be acoustically isolated to insure quietness of operation. The total parts cost of the system would be under $600, so it could be manufactured to sell competitively.

The measurement of systolic pressure is straightforward and presents no problem. Clinical tests would have to be run to establish the exact threshold for the systolic detection circuit.

In order to make the difference signal more predictable from one subject to the next, a precise AGC circuit is required on each AC amplifier. However, even with well controlled and closely matched pulse amplitudes, the difference is grossly affected by factors other than the passing of the diastolic pressure point. The difference signal amplitude does not appear to be highly correlated with the diastolic pressure of the subject. This fact makes the scheme unsatisfactory for use in an automatic system. Because Marx did not use an automatic gain control on his amplifiers his results are somewhat misleading.

In an effort to obtain an explanation for the presence of a difference signal at low cuff pressures, it was found that it is incorrect to assume that the pulses arrive at the cuffs nearly simultaneously below diastolic pressures. A pulse velocity in the arm of 5 meters per second is not uncommon.[7] This implies that a lag of 0.02 seconds can be expected between the pulses in the cuffs, whose upstream edges are about 10 centimeters apart, even at low cuff pressure. This delay is significant compared to the rise time of the pulse wave, and indicates that a difference signal might be expected at any cuff pressure.

These observations along with the results obtained from this system place in question the very theory on which this scheme for measuring diastolic pressure is based. It is recommended that additional basic research be done to obtain information about the physiological effects involved in this measurement process before any further attempt is made to implement it automatically.

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