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A portable blood pressure monitor utilizing Doppler ultrasound

George Philip Seifert
Iowa State University

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SEIFERT, GEORGE PHILIP

A PORTABLE BLOOD PRESSURE MONITOR UTILIZING DOPPLER ULTRASOUND

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A portable blood pressure monitor

utilizing Doppler ultrasound

by

George Philip Seifert

A Thesis Submitted to the

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INTRODUCTION

Both the layman's and clinician's ability to accurately measure blood pressure has increased with the introduction of computer controlled, automated blood pressure monitors. There have been a large number of these units introduced on the consumer market in the past few years. However, all of these units, at least the ones discovered by the author, use methods that fail to work on hypotensive subjects or in high noise environments. Thus, the usefulness of these units is limited.

This thesis describes in detail a battery powered, highly portable directional Doppler ultrasound system that, when used in conjunction with a microprocessor, promises to overcome the limitations of blood pressure monitors currently available. The system transduces opening and closing motions of the brachial artery under an occlusive cuff to a CMOS (Complementary Metal Oxide Semiconductor) compatible output to allow direct interfacing to a microprocessor.

Background

The most widely used and accepted method used in automated blood pressure monitors uses a piezoelectric microphone to detect Korotkoff sounds from the artery during cuff deflation. Systolic and diastolic pressures are detected in the traditional manner (Geddes, 1970), although there is some inconsistency between different kinds of units in the measurement of diastolic pressure (McDermot, 1980). Blood pressure monitors using the Korotkoff method are good indicators of daily fluctuations in blood pressure but their usage is not feasible with hypotensive subjects. The Korotkoff method fails on infants and hypotensive subjects because the microphone is not sensitive enough to detect the low amplitude sounds. The method also fails when high noise levels in the frequency range of Korotkoff sounds are present.

Oscillometry, the second most common method used by automated blood pressure monitors, senses pressure pulstions from an occlusive cuff. The oscillometric method has been shown to work well in determining mean arterial blood pressure. However, systolic and diastolic pressures must be extrapolated using the mean pressure (Looney, 1978) and have not compared favorably to direct measurements. This method also fails to work on hypotensive subjects.

While both of the above methods indirectly monitor the motion of the arterial wall, neither method is sensitive enough to pick up the small arterial wall motions associated with infants and hypotensive subjects. To overcome this problem, Ware (1965) proposed a more sensitive means of monitoring arterial wall motion using continuous wave diagnostic Doppler ultrasound. Doppler ultrasound is a technique in which low power (<50 mW/square cm), high frequency (>2 MHz) sound waves are transmitted toward the object whose movement is to be monitored. When the object moves, the transmitted signal will be Doppler shifted by an amount proportional to its velocity. Receiving circuitry detects this shift and suitably conditions the signal for further processing. Subsequent refinements of the ultrasound method by Ware and Laenger (1967), Kardon et al., (1967), and Stegall et al., (1968) showed that it was a viable alternative to the standard ausculatory technique. Further studies using the ultrasound method have shown its usefulness and accuracy on hypotensive subjects (Poppers et al., 1971), on infants (Zahed et al., 1971, Whyte et al., 1975), and in high noise situations (Kopczynski, 1974). It has also been shown that reconstruction of the entire arterial blood pressure wave is also possible (Ware et al., 1966).

Despite the considerable flexibility and accuracy of the ultrasound method, there has been only one attempt to automate the process of taking blood pressures. Although the unit (Arteriosonde™ by Roche Medical Electronics Inc.) has proven to be useful and accurate in the clinical situation (George et al., 1975, Hochberg et al., 1973), it is not highly portable and is only capable of presenting systolic and diastolic pressures to the user.

Objective

Recent advances in microelectronics make it possible to construct a blood pressure monitor using the ultrasound method in a considerably smaller space than is currently available. The increased processing capability of microprocessors will also make feasible the following:

1. Obtaining and storing systolic and diastolic pressure measurements throughout the day for future analysis. This would be possible on children, as well as adults, suspected of being hypertensive.
2. Obtaining blood pressures in an ambulance where ambient noise levels are too high to allow using the Korotkoff method.
3. Reconstruction of the entire arterial blood pressure wave.

However, before the above units can be built an ultrasound system capable of transducing the necessary information to a computer is needed. Therefore, the objective of this project is to develop a portable, battery powered ultrasound system with computer compatible outputs capable of detecting arterial wall motion under an occlusive cuff.

THE ULTRASOUND METHOD

The Doppler ultrasound method for obtaining systolic and diastolic arterial blood pressure measurement, or for reconstructing the entire arterial pulse wave, involves monitoring the motion of an artery under an occlusive cuff. The brachial artery is used because it is at the same level as the heart, thereby reducing hydrostatic effects. More importantly, the opening and closing moticns of the brachial artery are relatively easy to detect.

The output of a conventional Doppler system used to monitor arterial wall motion is shown in Figure 1. As the cuff pressure falls, the artery snaps open (arterial pressure exceeds cuff pressure) and closes again (cuff pressure exceeds arterial pressure). Cuff pressure equals systolic pressure when the arterial wall motion first splits into two distinct opening and closing phases and not when the first wall motion signals are sensed, as reported by Stegall et al., (1968). The opening and closing motions will continue to separate in time as the cuff pressure falls below systolic pressure until the closing motion occurs just before the opening motion. Cuff pressure equals diastolic pressure when these two motions first merge together.

The frequency range of the difference frequency signal (the difference between the transmitted and Doppler shifted frequencies) from the arterial wall for opening motion is

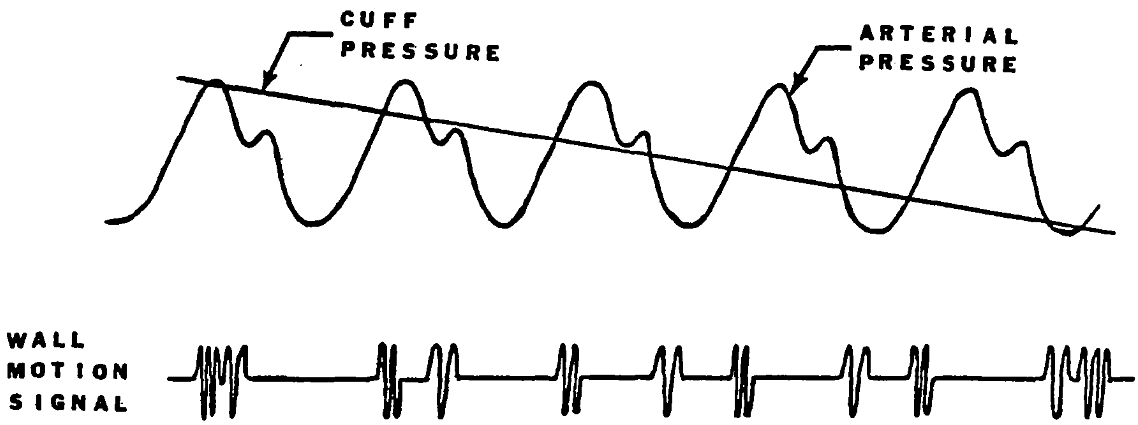


FIGURE 1. Output of a conventional Doppler used to monitor arterial wall motion

approximately 100-500 Hz and for closing motion is approximately 20-80 Hz for an 8 MHz system. These frequencies are directly proportional to the transmit frequency of the system. The frequency range that can be expected in systems using transmit frequencies other than 8 MHz can be predicted using the data given along with the Doppler equation (Wells, 1977). For example, the difference frequency signals that can be expected from a 2 MHz system will be 2/8 of those given above.

To the best of the author's knowledge, all Doppler systems that have been designed to monitor arterial wall motion use only frequency information to differentiate between the opening and closing phases. Unfortunately, there are problems with using only the frequency difference between opening and closing motions, especially if the signal is being computer processed. The main problem is determining the frequency of motion because the opening and closing phases last only a short period of time. The problem becomes increasingly worse as the transmit frequency is lowered (to accommodate large or heavy subjects) because there are fewer cycles to count within roughly the same amount of time. The wide variation in frequency from systole to diastole also makes differentiating between opening and closing phases difficult. This problem is compounded by the probability that the frequency range given for the opening and closing phases may not hold for every subject. To overcome these problems, a direction sensitive Doppler system has been developed which will increase artifact rejection capability and lessen the amount of software needed when using a computer to process the output signals.

A directional Doppler is capable of detecting whether a moving object, in this case the near wall of the brachial artery, is moving toward or away (referred to here as forward and reverse motion) from the ultrasound emitting transducer. The advantage of using a directional Doppler is that, in addition to the signal of Figure 1, two additional channels are obtained (which actually contain all the frequency information of the signal in Figure 1) that can be used to determine when the artery is opening and when it is closing (see Figure 2).

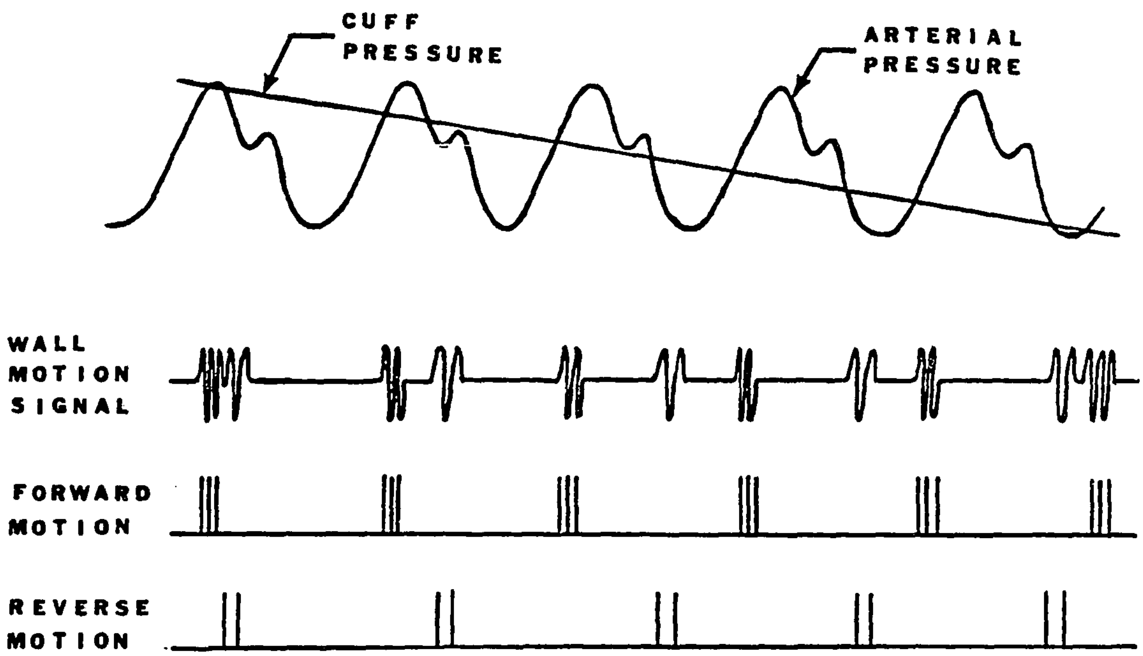


FIGURE 2. Additional information provided by using a directional Doppler

Cuff pressure equals systolic pressure when a distinct reverse motion follows just after a distinct forward motion (see RESULTS for further details). Cuff pressure equals diastolic pressure when the forward motion pulses are immediately preceded by the reverse motion pulses. Thus, no frequency information is necessary to determine when the artery is opening or closing, thereby eliminating the problems previously discussed.

SELECTION OF A SUITABLE DIRECTION SENSING METHOD

Separation of the received Doppler signal into forward and reverse components may be achieved by several different methods (Coghlan and Taylor, 1976). The two systems that were designed and tested use a technique called quadrature phase detection (Norgarrd, 1956). Quadrature phase detection was selected because it is very effective and does not require stable beat frequency oscillators or high Q band stop filters (thereby minimizing cost and design time) as do many other systems.

A system capable of simultaneously detecting near and far wall motion, similar to the blood flow monitor devised by Nippa et al., (1975), was tested first because the relative magnitudes of motion of the near and far walls were uncertain. Results of monitoring motion of the brachial artery under an occlusive cuff showed that the contribution of the far wall to the received signal was negligible, i.e., as the artery opened, forward motion only was recorded and as the artery closed, reverse motion only was recorded. Two possible explanations for the apparent lack of motion of the far wall follow. First, the received signal from the far wall is attenuated more than the signal from the near wall because it is farther away from the transducer, thus giving the impression that the motion of the far wall is relatively small. Second, the far wall probably does not undergo as much motion as the near wall because it is compressed against the humerus bone.

Since the main contribution to the received signal is from the near wall and because of the complexity of a system capable of detecting simultaneous forward and reverse motion, it was decided that a system similar to the one used by McLeod (1967), which can only detect net forward and reverse motion, would be more than adequate. McLeod's system uses a modification of the quadrature phase detection technique in which the outputs are digital pulses that appear in either a forward or reverse channel, but not simultaneously. Normally these outputs are integrated to provide an analog output, but for this project are left as digital pulses to be directly interfaced to a computer.

THE CIRCUITRY

Overview

A block diagram of the system that was developed is shown in Figure 3. The transmitter drives the transducer crystals and demodulators. The received signal, amplified by the tuned radio frequency (RF) amplifier, and the signals derived from the transmitter, are mixed to produce two identical outputs shifted 90 degrees relative to each other. These two signals are then fed to audio amplifiers which have identical bandpass and gain characteristics. Outputs from both channels are then applied to Schmitt triggers to produce a digital output.

The rest of the direction sensing circuitry is best explained by referring to Figure 4 in conjunction with Figure 3 (see also APPENDIX B). Arbitrarily taking the top channel in Figure 3 as a reference, the bottom channel will be either +/-90 degrees out of phase with respect to the top channel for forward or reverse motion.

When forward motion is present, the bottom channel will lead the top channel by 90 degrees and pulses will appear at the output of the top AND gate at the same frequency as the wall motion signal. When reverse motion is present, the bottom channel lags the top channel by 90 degrees and pulses will appear at the output of the bottom AND gate.

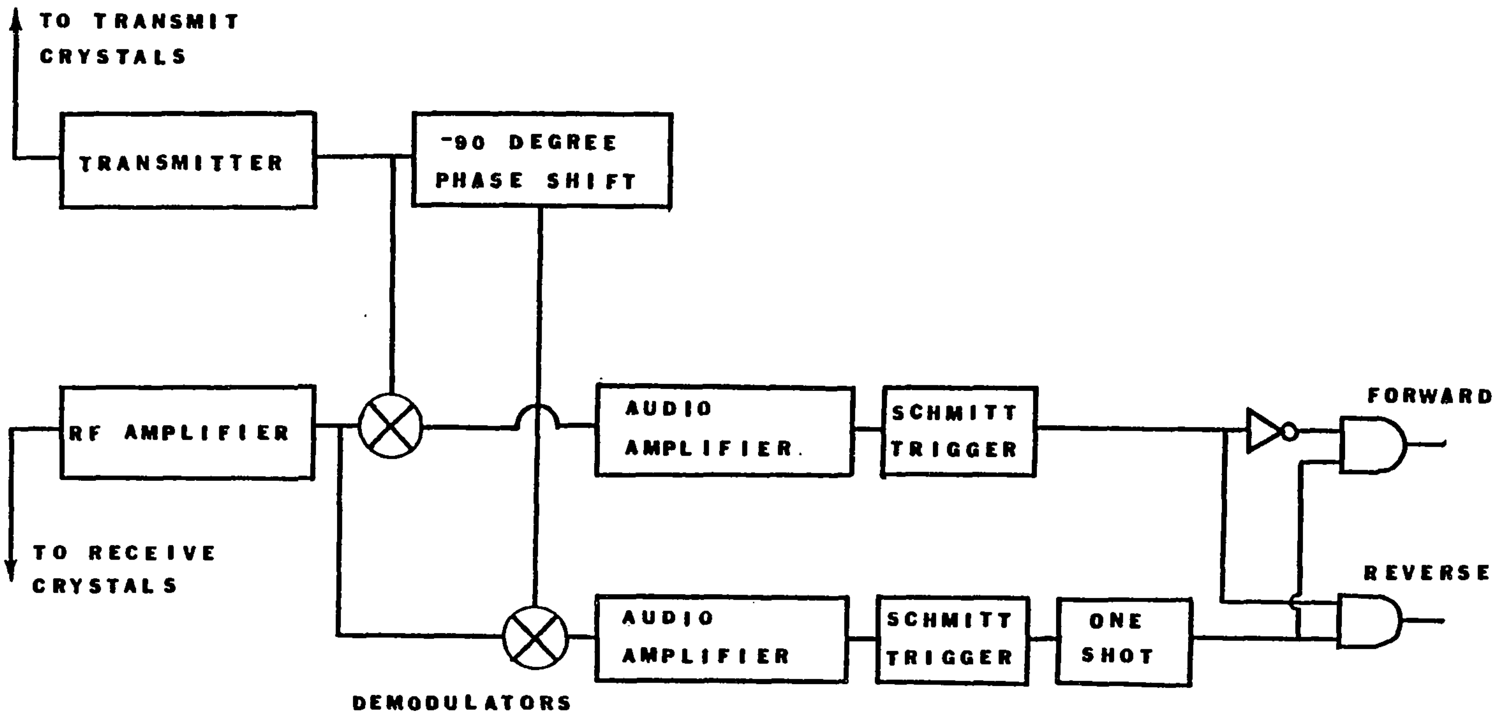


FIGURE 3. System block diagram

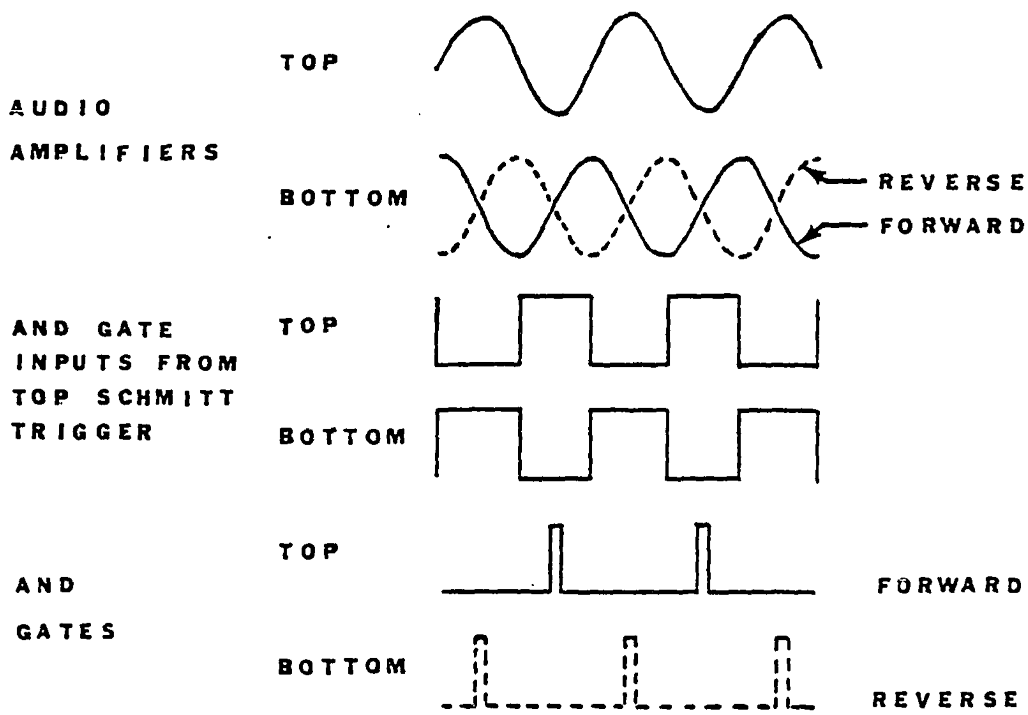


FIGURE 4. Extracting the forward and reverse motion information

Descriptions of each block in Figure 3 will be given in the following sections. Sufficient detail will be given to provide the user with a working knowledge of the system. All references in the following sections, unless otherwise specified, will be to Figure 5.

FIGURE 5. Ultrasound circuitry

R (K ohm)

1	1
2	.47
3	.82
4	10 (pot)
5,6,10,11	680
7,12	2.2
8,13	.39
9	10
14	10 (pot)
15	220
16	100
17	22
18	.33
19	120
20,30	2.2
21,31	.33
(22),(32)	150
23,33	56
24,34	1
(25),(35)	6.8
26,36	47
27,37	10
28,38	10
29,39	10
40	10
a,b,c	.1
d	.22

Note: Values of components whose numbers are in parentheses should be closely matched between channels.

C (Farads)

1	47p
2	.1m
3	100p
4	47p
5	10p
6	10p
7,10	.01m
8,11	.01m
9,12	.01m
13	.01m
14	47p
15	47p
16	.01m
17	.01m
18,24	220m
19,25	.1m
(20),(26)	.05m
21,27	.005m
(22),(28)	1m
23,29	.005m
30	.01m
a,c	.01m
b	100m
d	10m
e	10m
f	10m
g	.1m

D

1	1N4001
2,3,4	1N4148

Q

1	2N3904
2	2N5458
3,4,5	MFE131
(6),(7)	MFE131

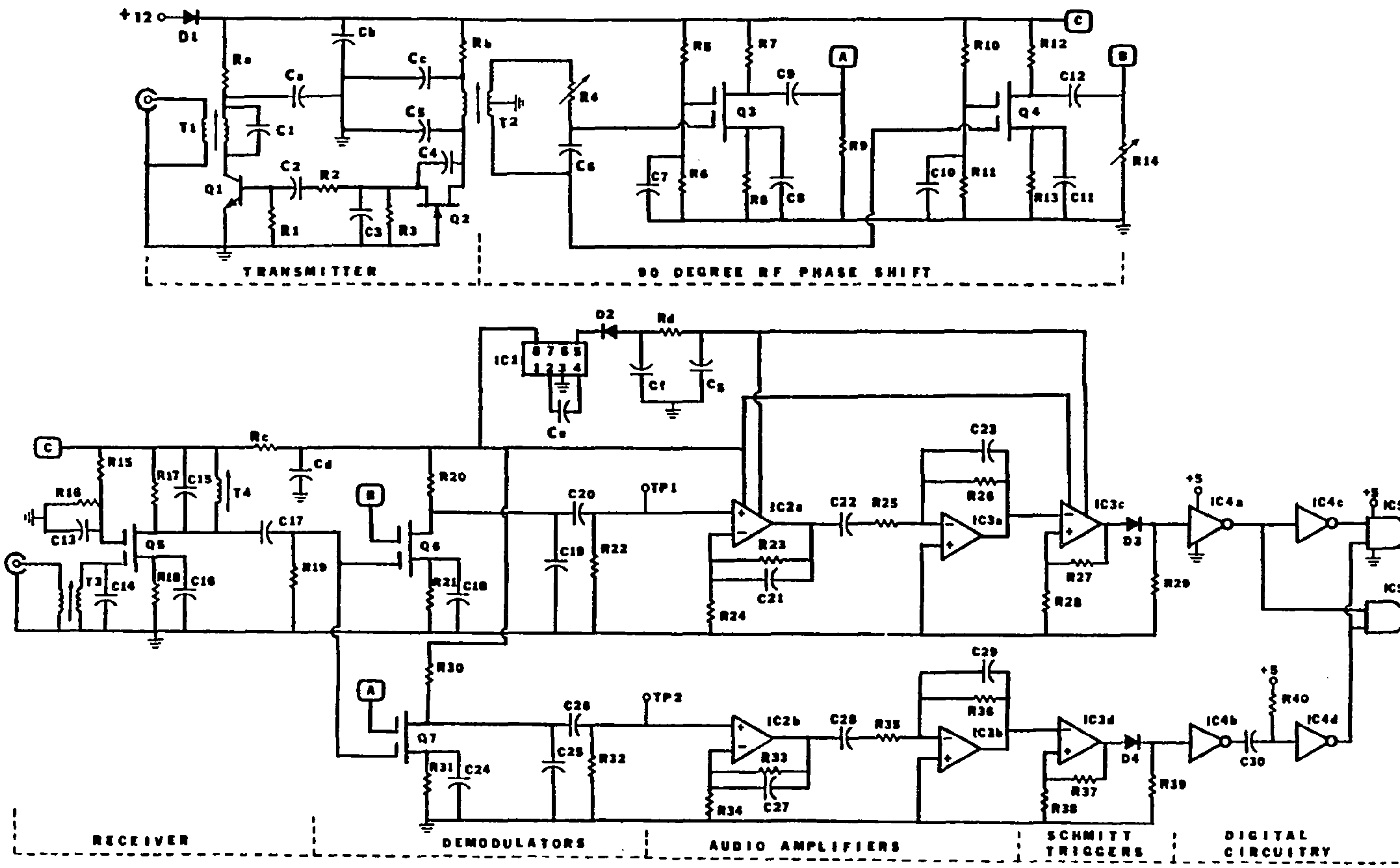
IC

1	ICL7660 Voltage Converter
2	1458 Dual Operational Amplifier
3	348 Quad Operational Amplifier
4	4049 Hex Inverting Buffer (CMOS)
5	4081 Quad AND Gate (CMOS)

T

1	primary 27 turns, secondary 6 turns
2	primary 27 turns, secondary 4 turns (center tapped)
3	primary 4 turns, secondary 27 turns
4	27 turns

All transformers wound with 41 guage wire on Amidon L-43 coil forms.


+12
D1
Ra
Cb
Ca
Cc
Rb
C5
T1
C1
T2
R4
C4
C2
R2
Q1
R1
C3
R3
Q2
C6
R5
R7
A
C9
Q3
C7
R6
R8
C8
R9
C
R10
R12
B
C12
Q4
R14
C10
R11
R13
C11
TRANSMITTER
90 DEGREE RF PHASE SHIFT
D2
Rd
IC1
8 7 6 5
1 2 3 4
Cf
Cg
Ce
Rc
Cd
R15
R17
C15
T4
R16
C13
Q5
C17
R20
R19
Q6
T3
C14
R18
C16
R21
C18
C20
TP1
C19
R22
IC2a
C22
R25
C23
R26
IC3a
IC3c
D3
+5
IC4a
IC4c
IC5a
IC5b
R23
C21
R24
R27
R28
R29
R30
C26
TP2
IC2b
C28
R35
C29
R36
IC3b
IC3d
D4
IC4b
C30
R40
IC4d
Q7
R31
C24
C25
R32
R33
C27
R34
R37
R38
R39
RECEIVER
DEMODULATORS
AUDIO AMPLIFIERS
SCHMITT TRIGGERS
DIGITAL CIRCUITRY

Transmitter

The transmitter consists of a variable oscillator and a class C amplifier (see Carr, 1978 for more information on class C amplifiers). C5 and the inductance in the primary of T2 are chosen to resonate at the series resonant frequency of the crystals chosen. R3 and C3 are chosen so that about 4 volts peak appears at the source of Q2. C4 provides positive feedback necessary to sustain oscillations. The signal from the source of Q2 is coupled to the base of Q1 through R2 and C2. Power output to the crystals may be changed by varying R2 or C2, thus altering the conduction angle of Q1. R1 is a bleed resistor which helps turn Q1 off. C1 and the inductance in the primary of T1 are chosen similar to T2 and C5. Transmit crystals are connected to the secondary of T1 through coaxial cable. The number of turns in the secondary of T1 will also vary the power output to the crystals.

90 Degree RF Phase Shifter

An all pass network (Budak, 1974), which is transformer coupled to the transmitter oscillator is used to generate two signals 90 degrees out of phase. These two signals are then used to drive the demodulators. A transformer with a small turns ratio is used to prevent loading of the oscillator.

The phase angle between the inputs to Q3 and Q4 is given by A=2*arctan(w*R4*C6), from which R4 and C6 are chosen to give A=90 degrees. These two signals are amplified to 2 volts peak by Q3 and Q4. Dual gate transistors (MFE 131) are used for Q3 and Q4 primarily because of the high gain capability of these transistors. R11 is left variable to compensate for small amplitude differences in the demodulators.

Receiver

Values of some components in the receiver are critical to preserve the 90 degree phase difference between the two channels. These components are C20,C22,C26,C28,R22,R25,R32, and R35. Close matching of these components between channels is recommended. Amplitude matching between channels is not nearly as critical as preserving the 90 degree phase difference, so careful selection of the remaining components is not necessary.

RF amplifier

The RF amplifier is a single stage, tuned amplifier which uses a dual gate MOSFET (Metal Oxide Silocon Field Effect Transistor) to reduce feedback capacitance (thereby reducing the possibility of oscillations). The receiver crystals are connected through coaxial cable to the primary of the input step up transformer T3. C14 and the inductance in the secondary of T3, as well as C15 and T4, are chosen similar to C5 and T2.

The gate 2 to ground voltage of Q5 is set to about 4 volts. This setting is noncritical but should be kept less than 5 volts to prevent oscillations. The RF gain (<30 dB) is kept lower than Doppler systems that use amplitude demodulators in order to minimize distortion in the demodulators.

Demodulators

The demodulators are dual gate MOSFETs configured as product demodulators (Roddy and Coolen, 1977, McKay, 1970) in which the transmitted and received signals are multiplied together to produce an audio frequency output (see APPENDIX B). In conventional continuous wave Doppler systems, these demodulators are not needed and a simple amplitude demodulator can be used. However, since two separate outputs 90 degrees relative to each other are needed, two separate demodulators must be used.

The output side of the demodulator transistors are biased as any audio frequency amplifier with high and low frequency cutoff values of 723 and 21 Hz, respectively. The gate 2 to ground voltage, which varies the transistor's transconductance, is set to 2 volts peak and is biased at o volts to allow maximum variation of the transconductance which in turn gives maximum sensitivity in the demodulators.

Unfortunately, due to the variability of MOSFET characteristics, the demodulator transistors (Q6 and Q7) need to be closely matched. The easiest way to select a transistor is to insert it in the circuit and monitor the drain to ground voltage. This voltage should be between 6 and 6.5 volts. The variation of the transconductance with the gate 2 voltage depends only slightly on the particular transistor used and is not a factor in selecting a transistor.

Audio amplifier and Schmitt trigger

The first audio amplifier stage (IC2) provides a gain of 567 and a high frequency cutoff of 568 Hz. The second audio amplifier stage (IC3) provides a gain of 47 and further high and low frequency attenuation with cutoff values of 677 and 23 Hz, respectively. These two amplifiers provide enough gain to cause saturation of the second stage when the artery is in motion. This causes no problem because the signal is fed to a Schmitt trigger. The Schmitt trigger trip points are about +/-5 volts which is sufficient to exclude any noise that may be propagated through the amplifiers. Before the signal is fed to the digital circuitry, the negative portion of the signal is clipped off by diodes D3 and D4.

An amplifier capable of driving headphones or a small speaker is shown in Figure 6. The input may be connected to the output of the second stage of either channel. This amplifier is useful when troubleshooting the system.

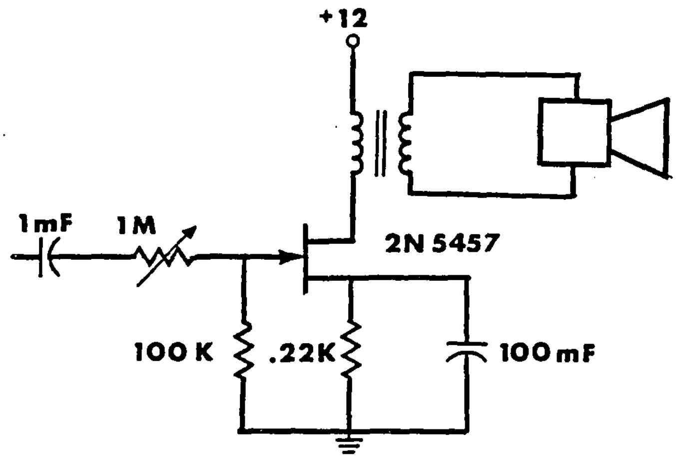


FIGURE 6. Headphone amplifier

Digital circuitry

For microprocessor compatibility, the digital circuitry is powered by 5 volts. IC4 is used to level translate the signal from the Schmitt trigger to a 5 volt CMOS compatible level. The complementary outputs as well as the one-shot pulse (.1 ms) are also provided by IC4. The output from IC4 is connected directly to the AND gates which are also CMOS.

Power

Power for the system may be supplied from any +12 volt D.C. source. Current drain is small (<30 mA) so that battery power is feasible. If the 12 volt supply is derived from the line voltage, the output should be well-filtered to remove 60 Hz noise because the audio amplifiers provide high gain in this frequency range. The 5 volts needed for the digital circuitry is easily derived from the 12 volt supply by a 5 volt regulator. This regulator should be connected just after diode D1. Connection of the regulator to any other point along the supply line may cause excessive voltage drops across the decoupling resistors. The negative supply voltage needed to power the operational amplifiers is also derived from the 12 volt supply by IC1. The chip itself draws very little current (<500 microamps) and provides a negative output voltage just slightly less than the negative of the positive supply used to power it.

TRANSDUCER DESIGN

The transducer houses the transmit and receive crystals and is the most critical component in the system. The transducer must be able to pick up motion from the artery from a variety of positions and must not interfere with the transmission of cuff pressure to the artery. Various configurations for the transducer were built and tested but met with limited success. These configurations and various design constraints will be given below.

The crystals (LTZ-5 material) were first glued to a 1/8 inch thick piece of Plexiglas for mechanical stability. The crystals were then mounted on a very thin sheet (1/32 inch) of Plexiglas and held in place with silicon rubber. This sheet provided a flexible yet strong base for the crystals and also provided good protection for the wires. An elastic strap with Velcro fasteners was then glued to the transducer so that it would be held securely to the arm.

The simplest transducer configuration used consisted of two crystals. This configuration works but is very difficult to position properly because as the cuff deflates the transducer moves with respect to the artery. So even though the transducer may be positioned properly before the cuff is inflated it will probably move as the cuff is inflated. Even with diverging lens attached to the crystals this configuration did not work satisfactorily.

The next step was to use an array of crystals as used by Ziedonis and Mount (1968), where two transmit and three receive crystals are used. The transducer lies perpendicular to the artery and essentially floods the area with ultrasound. Considerable improvement was obtained with this configuration but it still required some care in positioning. The major problem with this configuration was that pressure readings obtained were about 20 mmHg lower than those taken by the Korotkoff method. This was due to the thickness of the transducer. Adding diverging lens to the crystals only added to the thickness of the transducer and did not improve its ability to detect arterial wall motion when moved.

Some improvement over the above configuration was obtained when the transducer (still with two transmit and three receive crystals) was placed at about a 45 degree angle with respect to the artery. This allows more accurate transmission of the cuff pressure to the artery but as in the other configurations requires some manipulation to properly position. This configuration was used for data gathering purposes (see RESULTS). The crystals used were 2 x 5 mm and no lenses were used.

Obviously the ideal transducer still needs to be developed. The recordings shown in the results section should be used as a guideline to determine when acceptable results are obtained for any new transducer design. The number of crystals has no effect on the direction sensing capability or overall effectiveness of the system. It was believed at first that since the receive and also transmit crystals are connected in parallel, there might be some cancellations in the received signal. However, this was not the case because good results were obtained with multiple transducer designs.

RESULTS

Figure 7 shows a typical recording of the arterial wall motion obtained from the author's arm. Similar recordings were obtained from another normotensive subject and from a hypotensive subject. The top channel shows the forward motion signal, the middle channel shows the reverse motion signal, and the bottom channel shows the output of the second audio amplifier. This particular recording was obtained by first recording the outputs on a 4 channel tape recorder. The recording was then played back at 1/4 of the recording speed and recorded on a strip chart recorder (paper speed was 5 mm/s). It was necessary to do this because the strip chart recorder could not faithfully reproduce frequencies above approximately 200 Hz.

A pulse stretcher (see Figure 8) was connected to the outputs of the AND gates to convert the .1 ms pulses to 1.5 ms pulses in order to make the output pulses easier to record. Note that adding the pulse stretcher is not equivalent to making the pulses from the one-shot longer, as these pulses should be as short as possible (Webster, 1978).

The arrows in Figure 7 mark where cuff pressure equals systolic and diastolic pressure. The onset of pulses occurs while cuff pressure still exceeds systolic pressure. These pulses are caused by vibrations of the compressed arterial wall when the blood slams into it. Usually the first wall

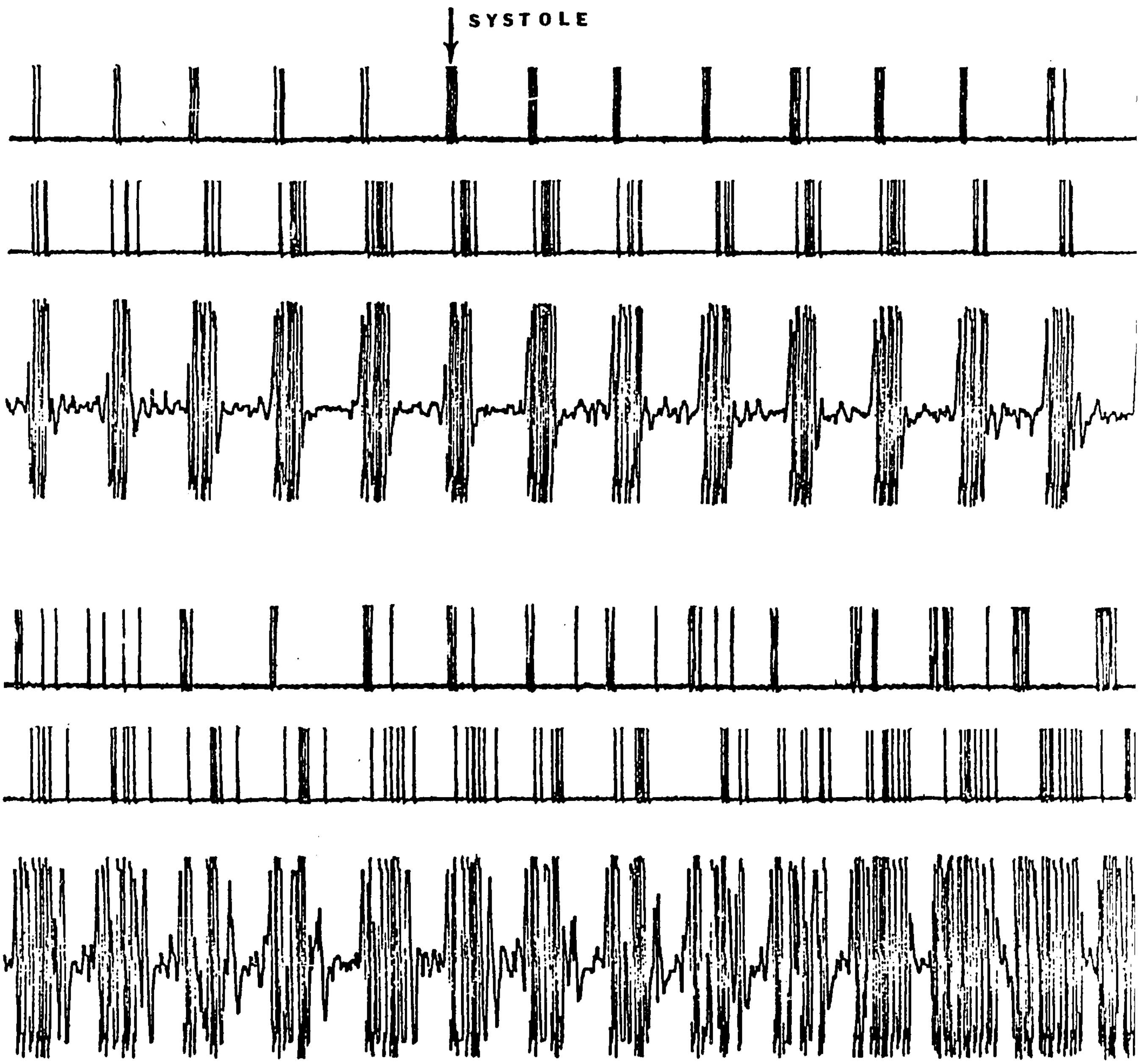
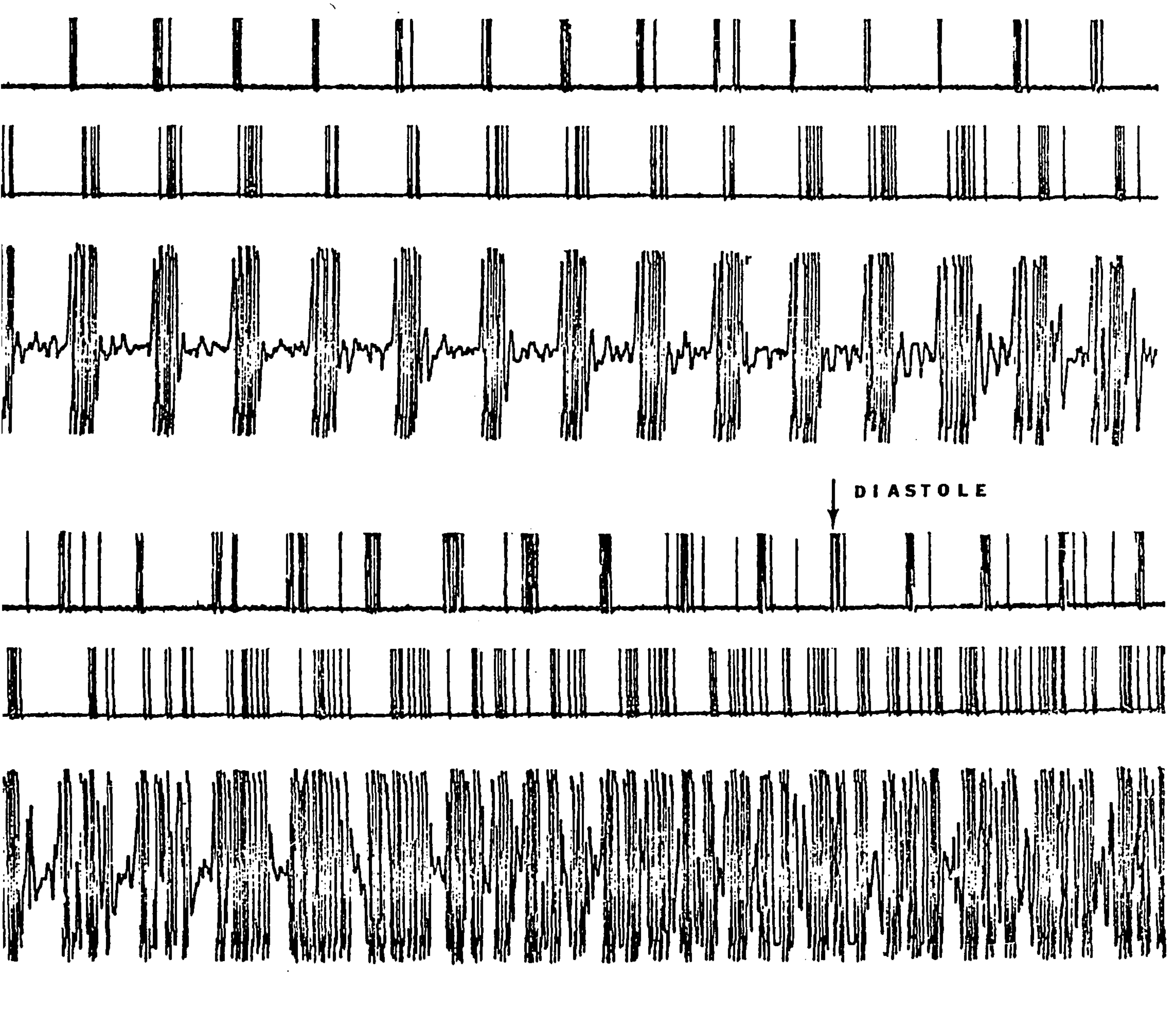


FIGURE 7. Recording of arterial wall motion



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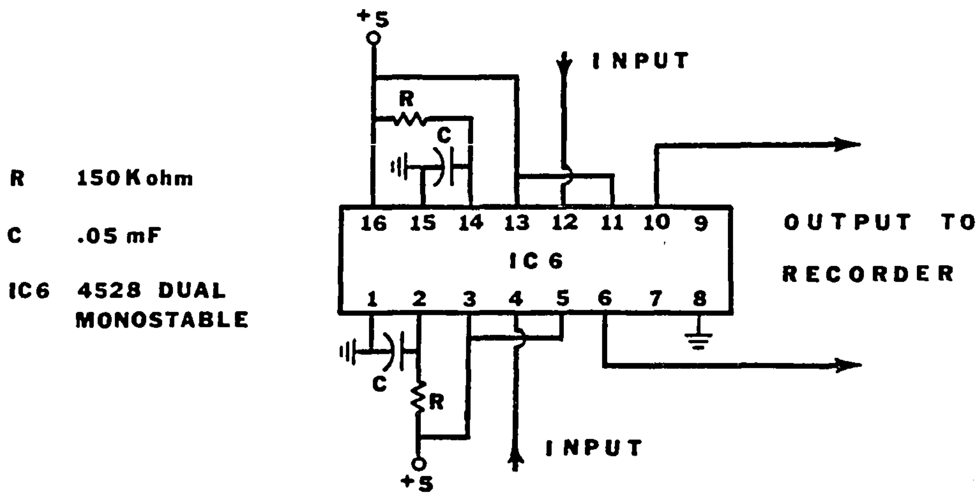

FIGURE 8. Pulse stretcher

motion signals are sensed when the cuff pressure is about 20 mmHg above the known systolic pressure. Of course, the actual pressure at which the first signals are sensed varies with different subjects. Nonetheless, it is important to realize that wall motion signals are sensed before cuff pressure falls to systolic pressure.

When the cuff pressure equals systolic pressure, the artery is allowed to open during systole. However, it is difficult when examining the bottom channel alone in Figure 7 to identify the transition between increased vibrations in the occluded artery and the actual opening of the artery.

The forward and reverse channels provide the additional information necessary to determine more precisely when the artery actually opens. Hence, systolic pressure can be more accurately determined.

Since the low frequency attenuation of the wall motion signal is only 40 dB/decade some motion artifact can be seen in Figure 7. Further low frequency attenuation could be obtained with hardware filters, but a more economical solution would be to let the computer used for processing the wall motion signal do the filtering.

Obviously, the actual arterial wall motion recording is much more complicated than the idealized signal shown in Figure 2. Once the cuff pressure nears the dicrotic notch pressure, the artery opens and closes twice during each heart cycle which causes the additional motion shown in the actual recording. Fortunately, this additional motion subsides as the cuff pressure nears the diastolic pressure, so that the merging of the opening and closing phases becomes relatively easy to detect.

The forward and reverse motion channel outputs, as well as the output from the second audio amplifier, may be used as audible outputs. Identifying the patterns in the audio signal is not difficult, but does require some practice.

To emphasize the usefulness of the directional Doppler ultrasound system, both the forward and reverse channel outputs and the output from the second audio amplifier were made audible by using the headphone amplifier (see Figure 6) and were used to determine the blood pressure measurements of 3 adult male subjects. Blood pressure measurements were also made using the Korotkoff method during the same cuff deflation. Systolic pressures were very difficult to determine when using the output from the second audio amplifier because the actual opening of the artery was indistinguishable from increasing vibrations of the occluded arterial wall, as previously discussed. However, when using directional information, systolic pressures were much easier to determine and compared favorably to measurements using the Korotkoff method. Diastolic pressures were also easier to determine using the directional outputs, but were not difficult to determine using the output from the second audio amplifier.

Figure 9 shows a comparison of blood pressure measurements taken using the Korotkoff method and the directional outputs from this system. Although the results show a good correlation between methods, more precise and accurate blood pressure measurements will be possible when this system is automated.

While excessive arm movement will cause incorrect readings with this system, false or misleading measurements can occur with any indirect blood pressure monitor when the

Subject	Trial	Ultrasound	Korotkoff
1	1	112/80	108/80
	2	112/80	110/80
2	1	130/80	127/77
	2	124/86	128/82
	3	124/84	128/78
3	1	110/78	110/78
	2	120/84	116/80
	3	120/84	120/80

Note: Pressures are given in mmHg and listed as systolic/diastolic.

FIGURE 9. Comparison of blood pressure measurements

tension of the muscles under the cuff is changed. For instance, a wrist flex can cause the cuff pressure to change by as much as 10 mmHg. Also, the actual blood pressure may be altered by movement of the arm. As long as the arm is held reasonably still, this system will work properly.

CONCLUSIONS

It has been shown that it is possible to obtain distinct opening and closing motions of the brachial arterial wall by using a direction sensitive Doppler. The additional information provided by using a directional Doppler seems to be well worth the small amount of extra circuitry needed. Although some work needs to be done on developing a less position sensitive transducer, the results are very encouraging that the system developed may be part of an accurate, reliable blood pressure monitor.

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APPENDIX A: ALIGNMENT AND ADJUSTMENT PROCEDURES

The following steps need to be taken only when a new transducer is used or if the system fails to work properly. Refer to Figure 5 throughput the procedure.

Initially, the 8 MHz oscillator must be fine tuned so that the crystals will operate at maximum efficiency. This can be done by monitoring the output to the crystals and adjusting variable inductor T2. The output will peak at the crystal's series resonant frequency. Variable inductor T1 should then be adjusted to obtain a maximum output while maintaining as clean a sinusoidal output as possible.

Next the receiver should be tuned to achieve maximum sensitivity. This can be done by either monitoring test points TP1 or TP2 on an oscilloscope or by listening to the audio output while swirling the transducer in a cup of water and at the same time tuning variable inductors T3 then T4.

After the transmitter and receiver are tuned the phase difference between the channels can be set. This is most easily done by monitoring test points TP1 and TP2 on an oscilloscope with X vs. Y capability. Swirl the transducer back and forth in a cup of water. Variable resistor R9 should be adjusted so that the oscilloscope output is either circular or ellipsoidal (but not tilted on axis) which means that the two channels will be 90 degrees out of phase. Variable resistor R14 can now be adjusted so that the output is as circular as possible. This method is somewhat crude but is very simple and effective.

The two channels do not have to be exactly 90 degrees out of phase and equal in amplitude for the system to work properly (Webster, 1978). However, the closer the two channels are to being 90 degrees out of phase when adjusted, the better the system will tolerate component value drift.

APPENDIX B: EQUATIONS RELATING TO THE PRODUCT DEMODULATORS

In the multiplying demodulators of Figure 5, the received RF signal is multiplied by two signals, 90 degrees out of phase, that are derived from the transmitter. Assuming that the demodulators are perfect multipliers, the resultant output can be determined as follows.

Let $2A\cos(w0+P)t$ and $2A\cos(w0+Q)t$ be the two signals derived from the transmitter that have some arbitrary phase relationship to the transmitter. Ideally P-Q should be 90 degrees. Also, let $C\cos(w0)t + L\cos(w0-wl)t + U\cos(w0+wu)t$ represent the received signal. Multiplication yields,

$$V = 2A\cos(w0+P)t\,[C\cos(w0)t + L\cos(w0-wl)t + U\cos(w0+wu)t]$$

$$= AC\,[\cos(2w0+P)t + \cos(P)t] + AL\,[\cos(2w0+P-wl)t + \cos(P+wl)t] + AU\,[\cos(2w0+P+wu)t + \cos(P-wu)t]$$

$$VQ = 2A\cos(w0+Q)t\,[C\cos(w0)t + L\cos(w0-wl)t + U\cos(w0+wu)t]$$

$$= AC\,[\cos(2w0+Q)t + \cos(Q)t] + AL\,[\cos(2w0+Q-wl)t + \cos(Q+wl)t] + AU\,[\cos(2w0+Q+wu)t + \cos(Q-wu)t] \;.$$

Bandpass filtering yields,

$$V = AL\cos(wl+P)t + AU\cos(wu-P)t$$

$$VQ = AL\cos(wl+Q)t + AU\cos(wu-Q)t \;.$$

For forward motion (toward the transducer),

$$V = AU\cos(P-wu)t$$

VQ= AU cos(Q-wu)t

ANGLE V - ANGLE VQ= P - Q .

For reverse motion (away from the transducer),

V= AL cos(P+wl)t

VQ= AL cos(Q+wl)t

ANGLE V - ANGLE VQ= Q - P .

If P - Q= 90 degrees then VQ leads or lags V by 90 degrees depending on the direction of motion.