The Estimation of Indirect Blood Pressure Using Photoplethysmography

by

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Submitted to the Department of Electrical Engineering and Computer Science on December 1, 1980 in partial fulfillment of the requirements for the Degree of Master of Science in Electrical Engineering.

ABSTRACT

A non-invasive blood pressure measurement method using an arm occlusion cuff and a reflective photoplethysmograph placed over a fingertip successfully measured systolic, but failed to measure diastolic pressure.

The plethysmograph monitored pulsatile blood flow, and both reflective and transmission pleths were equally effective. Pleth design included selection of the peak spectral characteristics (~700 nm) and determination of the optimal separation distance (~1 cm) for the matched LED and photoresistor. Various sites on the arm were studied.

The Lambert-Beer model for transmission of light through a limb is discussed, and is the basis for taking the natural log of the pleth output. The compressive nature of the natural log minimized signal differences between subjects of different skin characterisitcs. Subsequent processing included bandpass filtering and gain of several thousand.

As cuff pressure was reduced, systolic pressure corresponded to the value at which pulsatile blood flow was first detected. For systolic pressure, 77 trials were performed on 29 subjects, ages 19 to 64, with auscultatory values of 98 to 148 mm Hg. Subjects were black and white, both men and women. Error was defined to be the auscultatory value minus the photoplethysmography system result. Arithmetic mean error was +2.12 mm Hg; maximum error was +15 mm Hg; mean absolute error was 3.88 mm Hg. Acceptable error was defined to be +-5 mm at 80 mm Hg, +-10 mm at 160 mm Hg. Fitting data with a Gaussian curve predicts that the technique will yield acceptable results 83.3% of the time.

As cuff pressure was reduced, diastolic pressure was defined as the value at which the amplitude of the pulsatile blood flow signal leveled off. However, spectral analysis on the pleth signal, step response analysis for sudden cuff pressure release, increasing the pleth's receptive field, fingertip heating, and dual channel recordings with normal, slow, and pulsed cuff inflation, all failed to overcome fluctuations of the pulsatile blood flow that were due to systemic and/or local response mechanisms.

Thesis supervisor: Prof. Walter Olson, Herman von Helmholtz Associate Professor of Health Science and Technology and Electrical Engineering

To my dear parents --

Thank you for all your love, guidance, and support.

In loving memory of my dear grandparents (ז"ל)

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****

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My warmest thanks and best wishes go to my sister and brother-in-law, Diane and Chaim. I am both proud and honored to have you as my relatives and closest friends. May we share many more years together.

Finally, in reflecting over the past few years, no acknowledgment would be complete without a mention of The Boss, The King, The EE Nurd Club, and, of course, The Tang Gang. Isaac, special thanks for all your help and advice. Wishing all of you the very best.

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INTRODUCTION

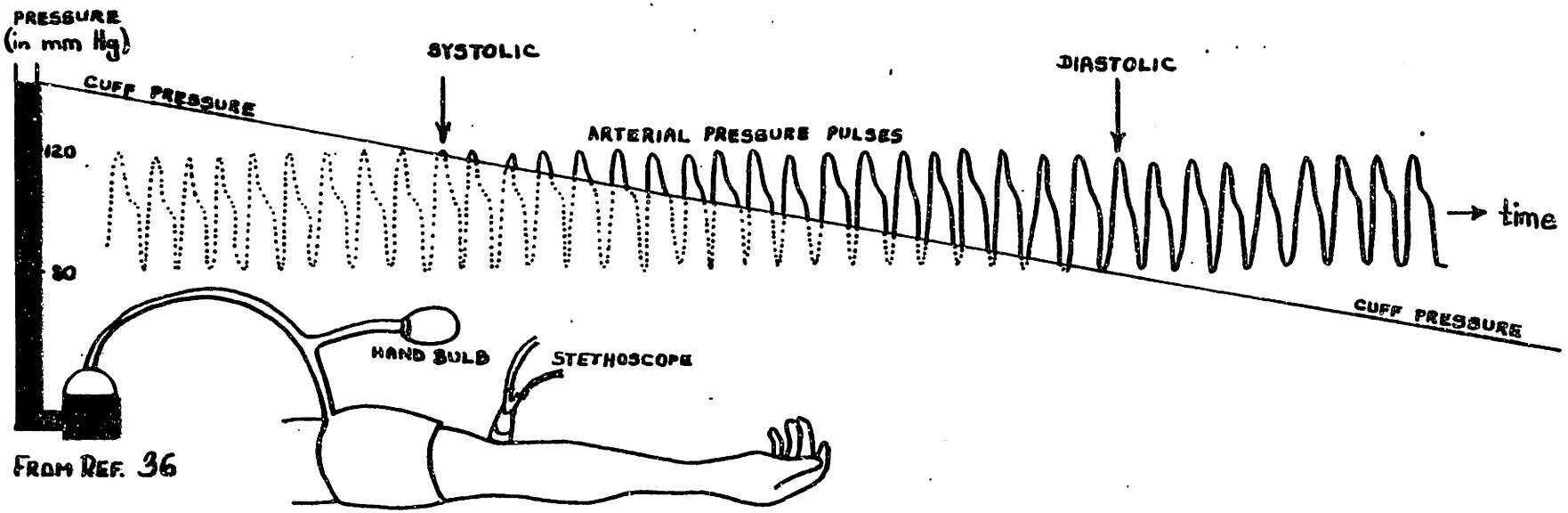
To establish the performance of the circulatory system, the medical profession would ideally like to know the extent of blood perfusion in various organs throughout a patient's body. Because this measurement is hard to obtain, a measure of blood pressure has become the data by which the elusive perfusion measurement has been estimated. Blood pressure yields information about heart function, the state of the circulatory system, and even gives an indication of the general mental/physical state of the patient (i.e. hypertension).

Blood pressure is a very general term. At any given time, the pressure in the circulatory system varies from place to place, with maximum pressure in the aortic area [30]. It is often necessary for doctors to know several pressures, especially those in the various heart chambers. In general though, when one goes for a physical exam, or for patients whose exact localized pressures are not necessary, blood pressure is interpreted as the pressure read through an occlusion cuff placed on the patient's arm. This yields an approximate measure of main arterial pressure [27], which is approximately aortic blood pressure [30]. It is important to note that these are approximate values [37,1,33]. The only way presently known to measure exact blood pressure is by inserting a catheter directly into the area of interest, i.e. an invasive measurement. Measurements using the occluding cuff have been widely accepted because of their non-invasive

nature, where non-invasive is defined to mean no penetration of the patient's skin.

In his book, Geddes [11] describes several techniques for non-invasive blood pressure measurement. Of these, there are three widely used methods:

1) The first is the auscultatory technique, depicted in Figure 1. It is the most widely used method, and is considered to be the best of the non-invasive techniques [35,12,37]. The occlusive cuff is inflated until its pressure exceeds the systolic level of the patient. The cuff pressure is then reduced at a rate of 2 to 3 millimeters of mercury per heartbeat. As the cuff pressure drops below the patient's systolic level, the blood flow through the partially occluded brachial artery of the arm results in the Korotkoff sounds, whose character changes as cuff pressure drops from systolic to diastolic levels. (Though the effecting mechanisms are not completely understood [33], the sounds are probably due to both the turbulence of the blood flow through the partially occluded artery and to artery wall vibration [32,34]). The medical profession has classified the different characteristics of the Korotkoff sounds into five phases (Figure 1). The onset of the sounds corresponds to the systolic value for the subject. There is disagreement among the medical profession as to what phase corresponds to the diastolic value. In general, the diastolic point corresponds to somewhere around the phase 4/phase 5 area [1,13,35,15,32,34].



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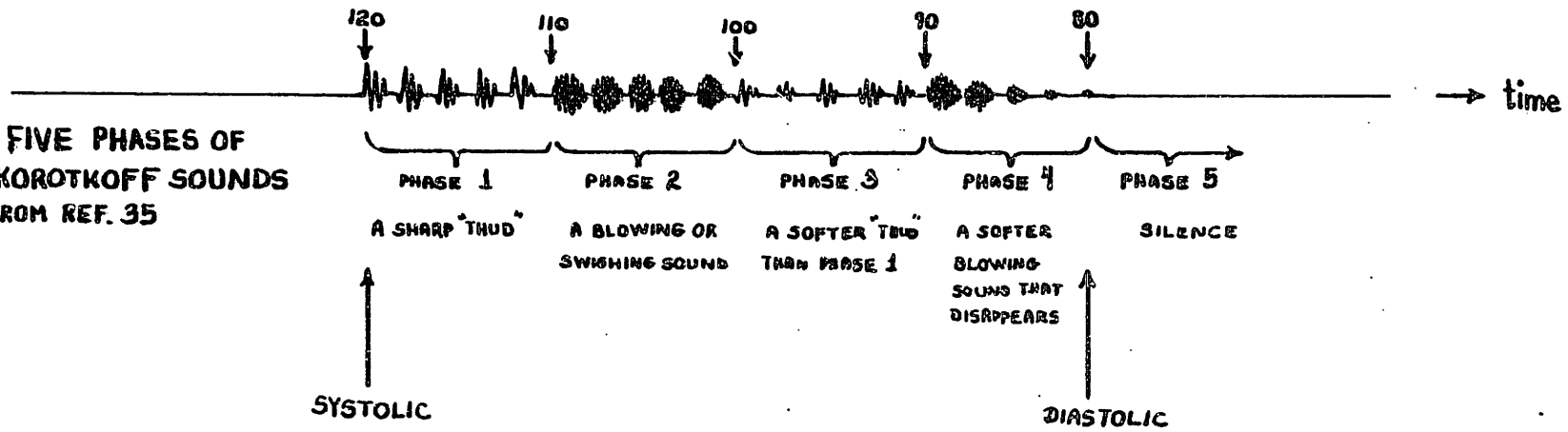

THE FIVE PHASES OF THE KOROTKOFF SOUNDS FROM REF. 35

FIGURE 1

The Korotkoff sounds can be heard with either a stethoscope or with a microphone placed on the skin over the brachial artery. Unfortunately, for children or hypotensive people, the sounds are not always discernable [13].

The auscultatory method is very dependent upon the positioning of the pickup device, whether it be a stethoscope or a microphone. The accuracy of the results are affected by the noise level in the surroundings at the time of the measurement and by the experience and subjectivity of the measurer [37].

2) The ultrasound technique, which is pictured in Figure 2. One uses an occlusion cuff inside of which is mounted an ultrasound transducer. The technique relies upon the Doppler frequency shift of the transducer signal, as affected by the magnitude of blood flow velocity and wall motion in the artery of interest, the brachial artery. For cuff pressures in between the subject's systolic and diastolic levels, the artery's opening and closing in response to the pulsatile blood flow is detected by the ultrasound detector (Figure 2). As cuff pressure increases towards the systolic point, the time interval between the artery opening and closing decreases. When the interval goes to zero, cuff pressure is at the subject's systolic pressure. When lowering cuff pressure towards the diastolic point, the time interval increases until the closing signal of one pulse coincides with the opening signal of the next one. This is the diastolic level [36].

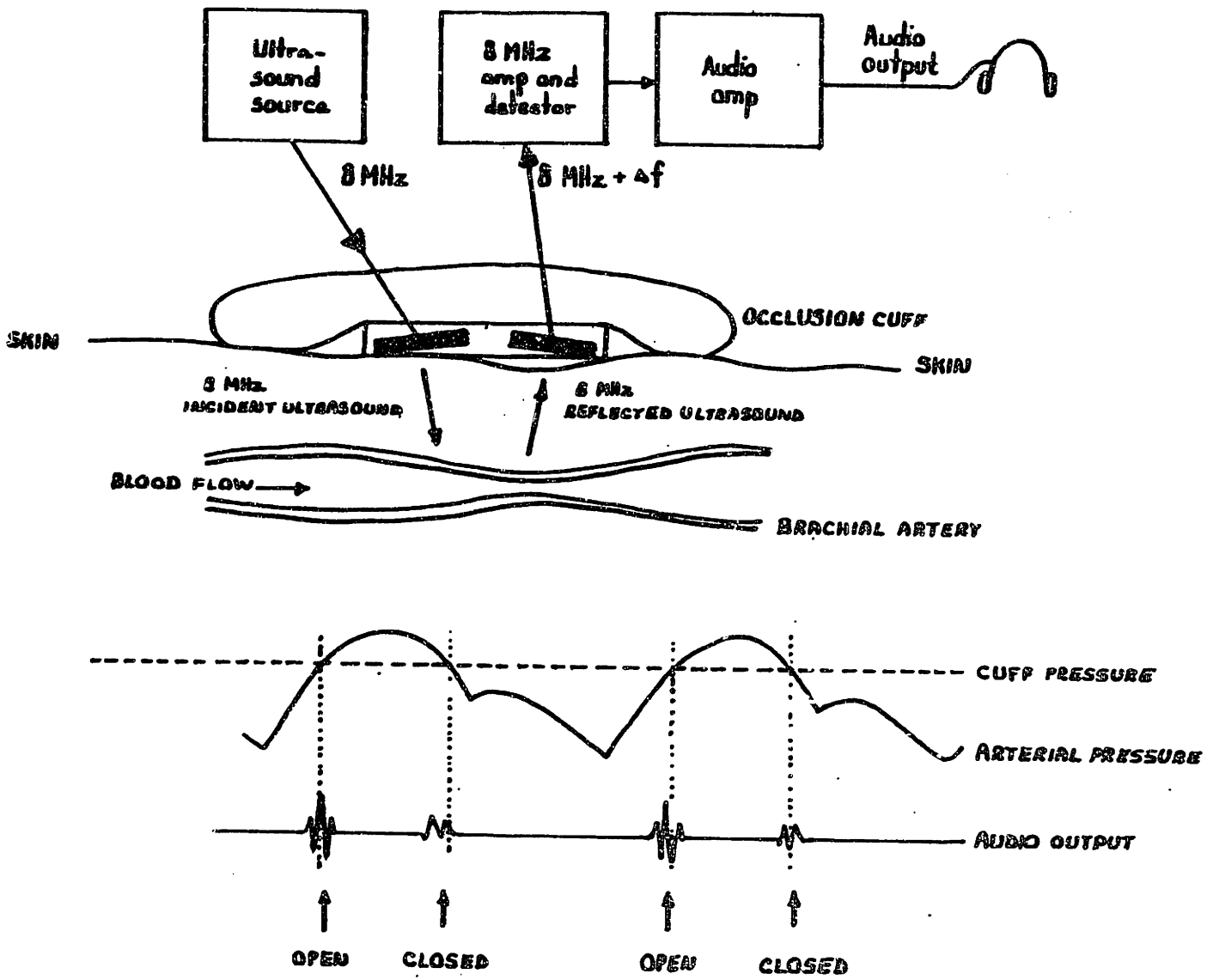

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FIGURE 2

While this technique avoids some of the problems of the auscultatory technique, one must contend with the increased cost, complexity, and high degree of position sensitivity of the ultrasound transducer [13]. The transducer must be positioned to focus the ultrasound signal on the brachial artery. In the past, acceptance of the technique was hindered by the "lack of understanding of the interaction between ultrasound and living tissue" [37]. However, more recent investigations have shown that the use of ultrasound as a diagnostic tool poses no hazards to patients [13].

3) The last technique is the tonometric method. Unlike the previous two techniques, once set it does not require a cuff to occlude blood flow. This method employs a pressure transducer pressing against the skin of the arm over the brachial artery and partially occluding it. The partial occlusion relieves circumferential tension in the artery wall and the remaining signal transmitted to the transducer is a direct result of blood pressure changes within the artery [2,36,3]. A standard auscultatory measurement is used to calibrate the system [2]. From then on, relative amplitude changes in the received transducer waveform are directly related to blood pressure changes.

The tonometric technique has much to offer because of its ability to provide for long term patient blood pressure monitoring. Unfortunately, it is very prone to positioning problems, and still relies on the auscultatory technique for its basic reference setting.

Each of these techniques has drawbacks and inaccuracies associated with it. My objective was to investigate the possibility of using either a reflectance or a transmittance optical plethysmograph and an occlusion cuff to measure systolic and diastolic blood pressure. The technique is portrayed in Figure 3. The optical plethysmograph and associated circuitry would monitor pulsatile blood flow in the brachial artery. The occlusion cuff would be inflated above the systolic level for the patient. The brachial artery would be totally occluded and no blood would be flowing further down the arm. The circuitry would reflect this as baseline noise. When cuff pressure dropped just below systolic, the first spurt of blood making it through the almost totally occluded artery would give rise to a short pulse on the circuitry output. The amplitude of these pulses would grow as cuff pressure is reduced and more blood passed through the widening cross-sectional area of the artery. The amplitude would level off when diastolic pressure was reached, since the artery would have been fully opened, and further reduction in cuff pressure would not affect blood flow in the blood vessels.

The use of photoplethysmography is not a new idea. A similar approach to measuring blood pressure was developed at the Mayo Clinic using a transmittance photoplethysmograph placed over the pina of the ear and occluding blood flow by means of a pressure bellows built into the pleth [13]. Photoplethysmography has been used to monitor pulsatile blood

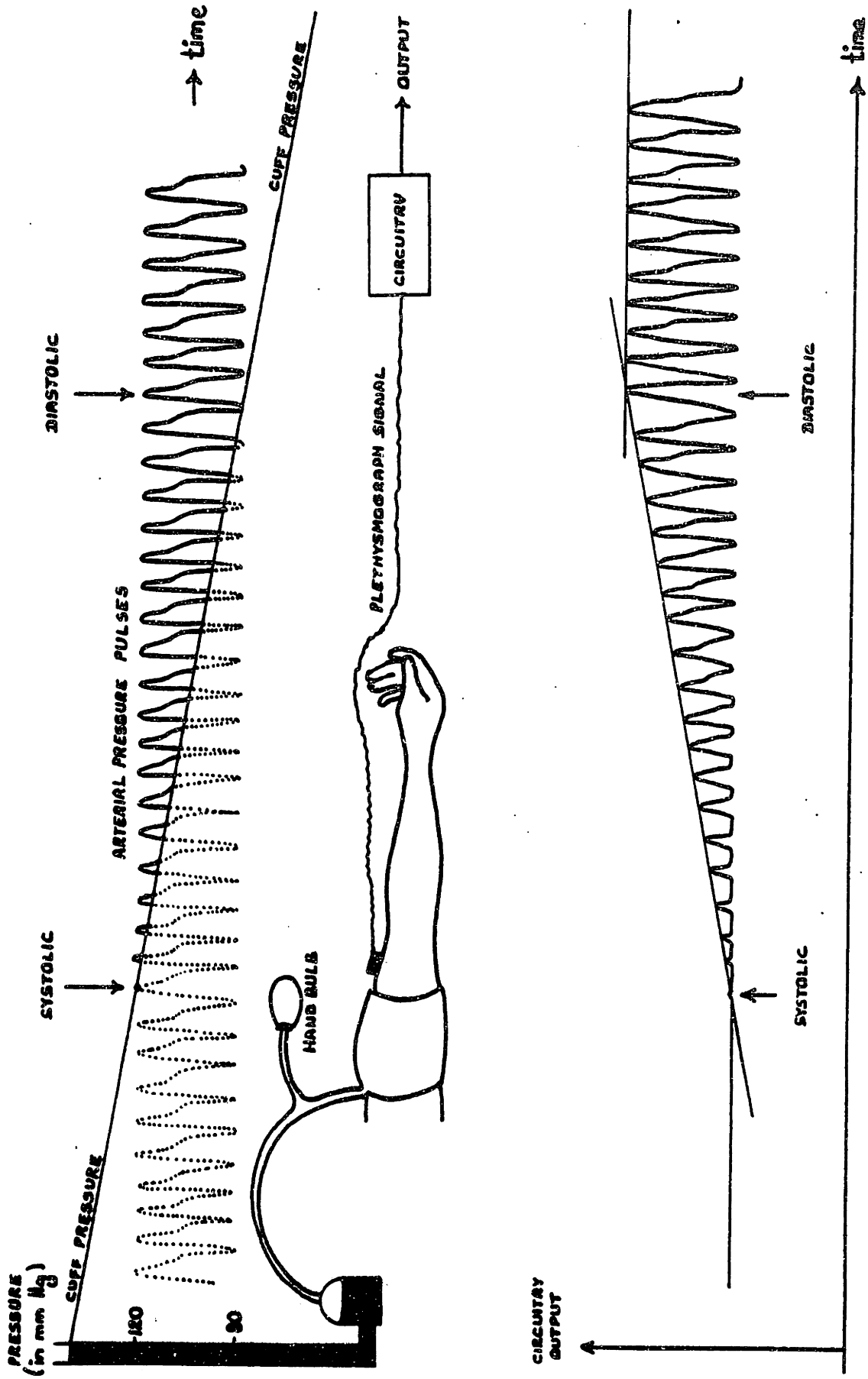

PRESSURE (in mm Hg)
CUFF PRESSURE
120
80
SYSTOLIC
DIASTOLIC
ARTERIAL PRESSURE PULSES
CUFF PRESSURE
time
HAND BULB
PLETHYSMOGRAPH SIGNAL
CIRCUITRY
OUTPUT
CIRCUITRY OUTPUT
SYSTOLIC
DIASTOLIC
time

FIGURE 3

flow into regions of the body [13], to monitor vascular status in trauma and burn victims [5], and indirectly to obtain heart rate [22] and relative perfusion [25].

THE BASIC PHYSIOLOGIC MODEL AND SIGNAL PROCESSING APPROACH

Let me first present Beer's Law for the transmission of light through a transluscent homogeneous substance:

Consider a transparent material of thickness D, containing a homogeneously dispersed light absorbing substance of concentration C, radiated with incident light Iin, and transmitting light Itrans (see Figure 4a). Two definitions apply:

a. The absorbance of the material as a function of the wavelength λ of the incident light, $A(\lambda)$ is:

$$A(\lambda) = -\ln (I_{trans}/I_{in}) = E(\lambda)CD \qquad (1)$$

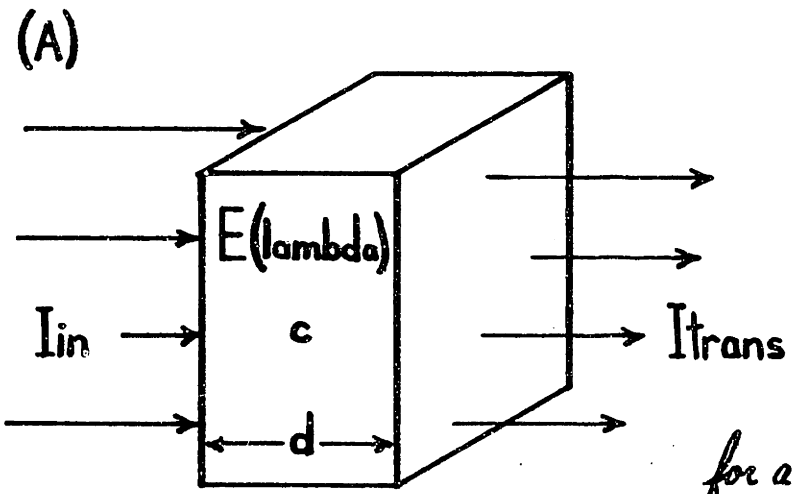
where $E(\lambda)$ is the wavelength dependent molecular extinction coefficient for the material.

b. The transmittance of the material as a function of the wavelength λ of the incident light, $T(\lambda)$ is defined to be:

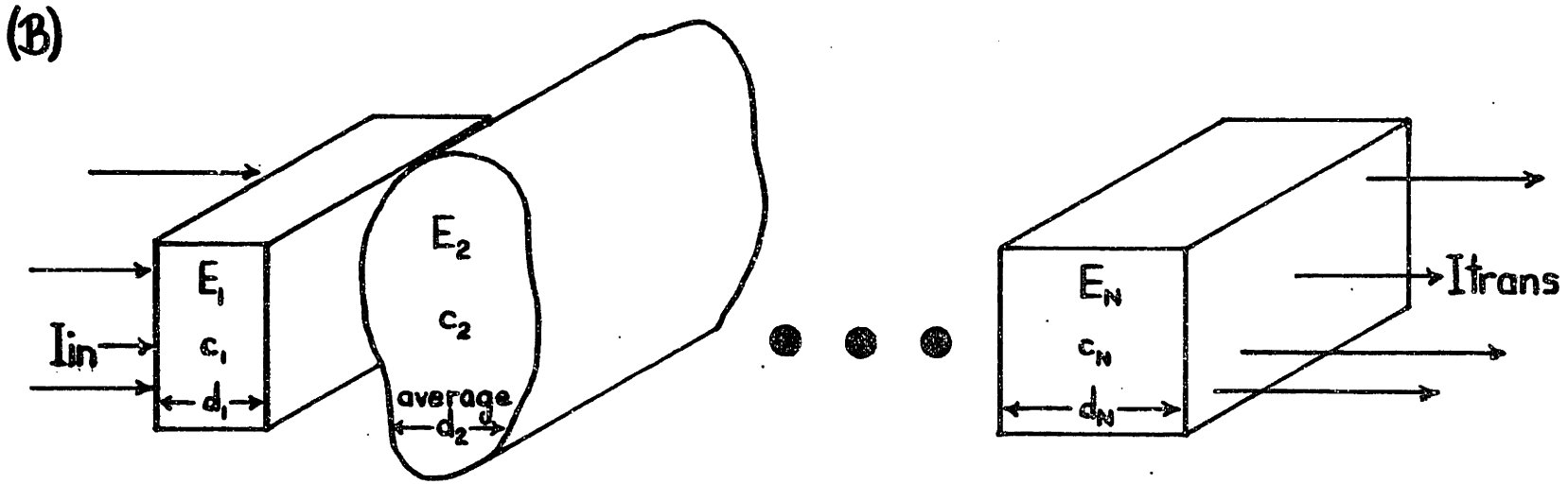
$$T(\lambda) = I_{trans}/I_{in} \qquad (2)$$

Manipulating relationships (1) and (2) yields:

$$T(\lambda) = \exp [-E(\lambda)CD] \qquad (3)$$



for a single absorber, $I_{trans} = [I_{in}]\, e^{-Ecd}$



for mutiple absorbers in series, $I_{trans} = [I_{in}]\, e^{-(E_1 c_1 d_1 + E_2 c_2 d_2 + \ldots + E_N c_N d_N)}$

Figure 4

Combining equations (2) and (3) into a more meaningful form for this discussion:

$$Itrans = (Iin)\ exp\ [-E(\lambda)CD] \qquad (4)$$

Equation (4) is one form of Beer's Law [18]. Next consider the case where multiple light absorbers are placed in series (Figure 4b). For a radiating light of Iin, Beer's Law states that for multiple light absorbers, the light transmitted through the last absorber will be:

$$Itrans = (Iin)\ exp[-(E1C1D1 + E2C2D2...+ EnDnCn)] \qquad (5)$$

where En is the wavelength dependent molecular extinction coefficient for the nth material.

The Lambert-Beer model of light absorbtion in a human limb [25] is based on the above development and the model of Figure 4b. Iin is the light impinging upon tissue, and the various light absorbing materials correspond to the different components of the limb, such as skin, blood vessels, fat, and blood [25].

To extend Beer's Law for application to a human limb, an assumption is made in the development of the Lambert-Beer model: The model treats the various materials in the limb as homogeneous light absorbers. It is most important to discuss the validity of this assumption, and I will review this

premise later. For now, under the assumption that Beer's Law can be extended to consideration of a human limb, let me continue with the development and show how this model can be used.

For a given wavelength of incident light, En(λ) for each absorbing material will remain constant. If we assume that the physical make-up of the light absorbing materials in a body does not change significantly over periods of time comparable to heart rate, then the concentration of light absorbing substance in each material, Cn, will remain constant. Consequently, equation (5) says that changes in Itrans will be due only to changes in the value of D for the various absorbing materials.

Most of the materials in a limb, such as skin and fat, have an average D that remains constant, but blood vessels do not. Since pulsatile blood flow results in cross-sectional area changes in blood vessels [25], the associated changes in the values of D for blood vessels will result in changes in Itrans.

Suppose we take the natural log of both sides of Equation (5):

$$\ln(Itrans) = \ln(Iin) - E1C1D1 - E2C2D2 \ldots - EnCnDn \qquad (6)$$

If D1 represents the varying thickness of a blood vessel experiencing pulsatile blood flow, while D2 through Dn are fixed average thicknesses of skin, fat, and other components

of a limb, we can take the derivative of Equation (6) with respect to D1 and see how the ln(Itrans) varies with changes in D1:

$$d[\ln(I_{trans})]/dD_1 = -E_1C_1 \qquad (7)$$

Equation (7) states the important result: Changes in the natural log of light that has been passed through a human limb will be linearly related to the average thickness D1 of the blood vessels inside.

We have used the Lambert-Beer model of a human limb to derive the above result. This development has used the following assumptions:

1) That the different components of a human limb act like independent homogeneous light absorbers, which obey Beer's Law of light transmission.

2) That the molecular extinction coefficients for each component of a limb, $E_n(\lambda)$, remain constant during the time pulsatile blood flow is to be monitored.

3) That the concentrations of light absorbing substances, Cn, in each component of a limb remain constant during the time pulsatile blood flow is to be monitored.

4) That the average thicknesses, Dn, of each component of a limb other than blood vessels, remain constant during the time pulsatile blood flow is to be monitored.

The signal output from the photoplethysmograph is a direct measure of Itrans. If one processes this signal by taking its natural log, the above development says that the AC component of the output will be linearly related to changes in the thicknesses of those blood vessels located within the plethysmograph's receptive field. The circuitry output will in effect be monitoring pulsatile blood flow within the vessels.

As explained in the introduction, to use photoplethysmography for my application, I needed a system whose output was in some way proportional to the pulsatile blood flow in blood vessels within the receptive field. Based on the development just presented, I processed the output of the plethysmograph by taking its natural log and using the AC component.

There is another incentive for processing the pleth signal with a ln function. The function alleviates some of the problem associated with using light on patients of different skin color and characteristics. Because of its mathematically compressive nature, the differences between the signal amplitudes obtained from a white versus a black patient are reduced. While it is true that the ln function will reduce the overall amplitude of the plethysmograph signal as

compared with linear processing, this difference is mainly manifested in the DC component of the signal, which I do not use.

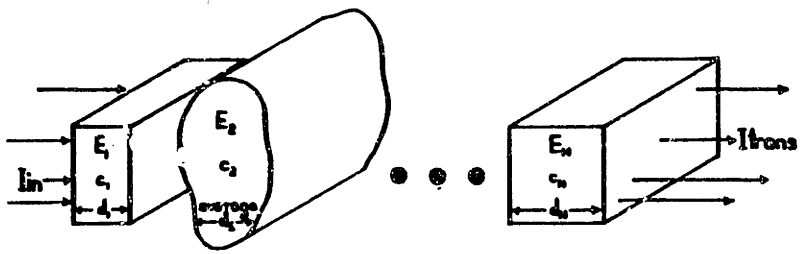
At this point there are some issues which should be addressed:

The development was based on a few assumptions, any one of which could be invalid. Those assumptions which require that the values of En and Cn remain constant may not be absolutely true, but it seems reasonable to assume that for a patient at rest, these values would remain approximately constant over the time span required to take a blood pressure measurement.

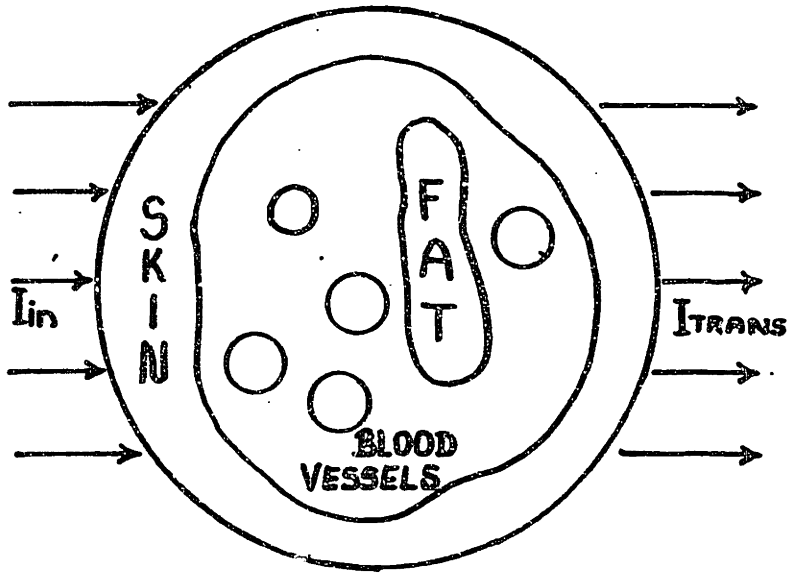
The main point of contention is the basic assumption that the components of a human limb are homogeneous light absorbers, and that they obey Beer's Law (Figure 5). Actually, this question can be narrowed down further. If one accepts the premise that the components of a human limb other than the blood vessels will not change their light transmission characteristics during a blood pressure measurement interval, then one can say that these components will transmit light in accordance with some relationship (not necessarily known or understood today) that will remain constant. Given this, we can focus our attention on the changing blood vessels only. It is important to know whether Beer's Law applies to whole blood, for if it does not, our whole Lambert-Beer model is invalid.

The use of Beer's Law as the cornerstone of photoplethysmographic applications is supported in the

CAN WE MAKE THIS JUMP?



FOR HOMOGENEOUS LIGHT ABSORBERS,
Itrans IS RELATED TO Iin BY
BEER'S LAW FOR MULTIPLE ABSORBERS.



CROSS SECTION OF A HUMAN LIMB

IS Itrans RELATED TO Iin BY BEER'S LAW
FOR MULTIPLE ABSORBERS ?

Figure 5

literature [25,6]. However, numerous articles on the relationship between impinging and transmitted light through whole blood conclude that Beer's Law does not apply; rather that complicated scattering theories govern the transmission and reflection of light in the human environment [19,18,20,41,23]. The evidence that Beer's Law does not apply to a physiologic application is strong. Yet its use as cited above [25,6] indicates that the law may be applicable, at least as a first approach, to modelling the physiology of a limb.

When dealing with the often changing physiology of a human limb, the lack of knowledge about the exact internal changes over time make it difficult to derive exact measurements and draw firm conclusions about the validity of a model. In the end, I chose to process the photoplethysmograph's signal by taking the AC component of its natural log, as per the Lambert-Beer model described above. Even if the whole model is invalid, the compressive nature of the natural log function would be useful in my work. It should be noted that in the final analysis, I need only to monitor pulsatile blood flow, not to derive any detailed quantitative information from the signal, for which exact relationships governing its source would need be known.

A good discussion of what one "sees" with photoplethysmography is contained in Gorelick's M.S. thesis [13], "Transducer Development for Blood Pressure Measuring Device." As mentioned in my introduction,

photoplethysmography has been used to monitor pulsatile blood flow for numerous applications, with the light source/detector pair placed on various physical locations on the subjects [13,25,22,5].

One can question the use of the natural log as the signal processing technique. One can also question the validity of the assumptions made by the Lambert-Beer model. But the results support this most basic observation: Light that has impinged upon a human limb and that has been received by a photodetector, does contain information about pulsatile blood flow within the receptive field. In view of the fact that my system did indeed output signals indicative of pulsatile blood flow, I feel justified in making this conclusion: The use of the natural log to process the signal was not detrimental to my research, despite evidence that the model, based on which that function was selected, may be invalid.

THE SIGNAL PICK-UP - THE PLETHYSMOGRAPH

A photoplethysmograph consists of a light source and a photodetector. In order to design a simple and durable pleth, suitable for use in a hospital environment, numerous design considerations must be taken into account.

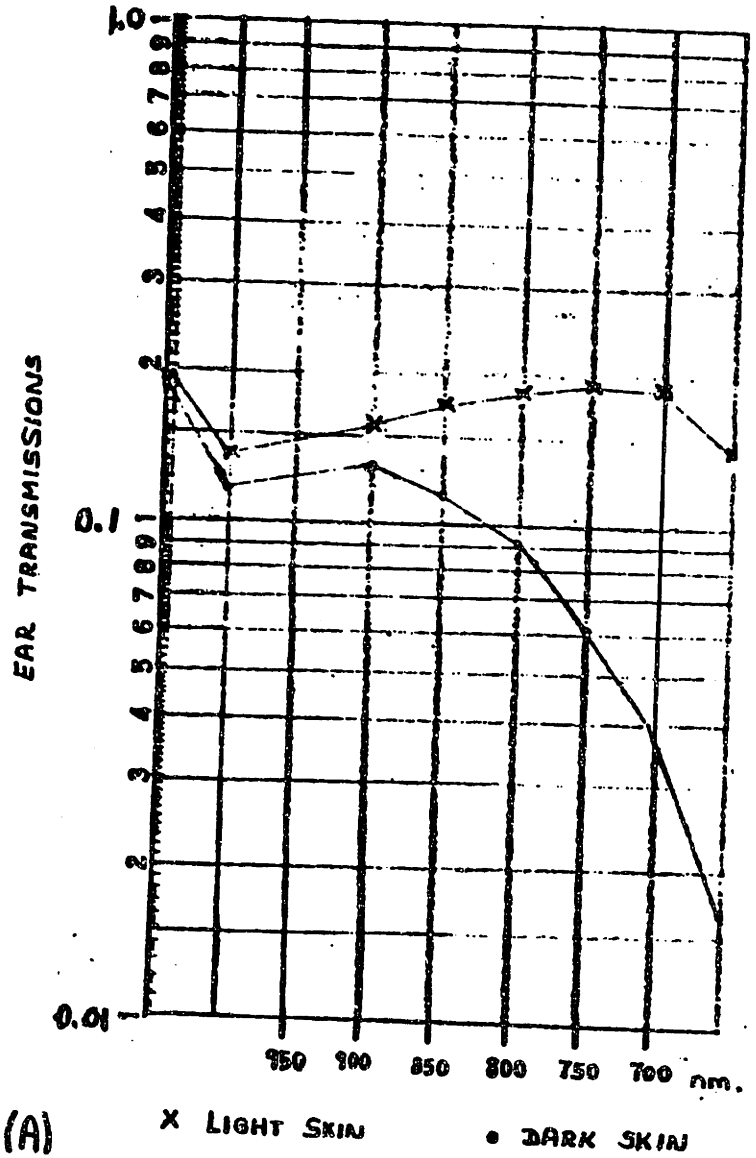
First in the design of a plethysmograph is the selection of a light source and its emission frequency. In the previous section, the molecular extinction coefficient of a material was mentioned. Every material, in this case body tissue, blood vessels, and blood, has some light absorption characteristics, which are frequency dependent. For my application, the following three constraints apply:

1) We are interested in pulsatile blood flow, regardless of the oxygen content of the blood. Consequently, the frequency of the light used should be chosen to equalize absorption by hemoglobin and oxyhemoglobin.

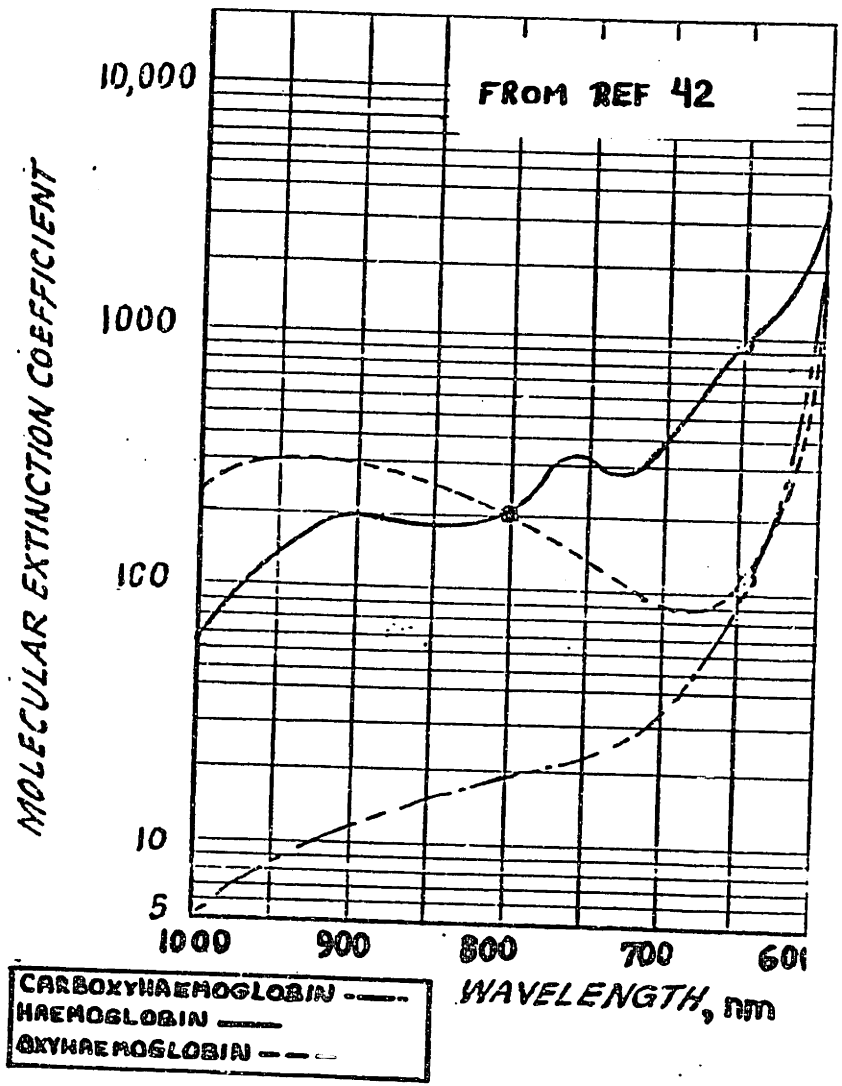
2) One must contend with light loss due to absorption in skin tissue. To minimize differences in signal loss from patient to patient, the light should be equally absorbed by skins of different color.

3) Commercial availability of parts must be considered. For optimal system peformance, the spectral response of the photocell should match the spectral characteristic of the light source.

Figure 6 contains two sets of relationships. 6a shows light transmission versus wavelength for light and dark skin. 6b shows molecular extinction coefficient versus wavelength


EAR TRANSMISSIONS
1.0
0.1
0.01
950
900
850
800
750
700
nm.
X LIGHT SKIN
• DARK SKIN

(A)


FROM REF 42
MOLECULAR EXTINCTION COEFFICIENT
10,000
1000
100
10
5
1000
900
800
700
CARBOXYHAEMOGLOBIN
HAEMOGLOBIN
OXYHAEMOGLOBIN
WAVELENGTH, nm

(B)

FIGURE 6

for various blood pigments. At an approximate wavelength of 800 nanometers, hemoglobin and oxyhemoglobin absorb light equally [42,13,38]. Light absorption differences due to skin color decrease with higher wavelengths.

One finds that there are no photodetectors with peak sensitivity at 800 nanometers. There are photoresistors at approximately 700 nanometers, and photodiodes at 900 and 940 nanometers. Figure 7 shows a comparison of molecular extinction coefficients and light transmission for these three wavelengths. Both 900 and 940 nanometers would be suitable because they minimize differences in absorption due to skin color while maintaining insensitivity to oxygen content of the blood.

900 and 940 nanometers are part of the infrared spectrum. Since flesh transmits more light as the frequency enters the infrared region [7], and in-vitro, light transmittance becomes more independent of oxygen concentration in the blood at infrared frequencies [21], either 900 or 940 nanometers should be used.

Much of my work was based on a research report written by Larry Nielsen of the Hewlett Packard Medical Products Group [25]. His work dealt with the development of a photoplethysmograph for perfusion indication. Because he has used an emission frequency of 700 nanometers, and had parts readily available for construction of a pleth, I used 700 nanometers as my operating frequency. In retrospect, I would use 900 or 940 were I to do the research over, because these

Data from Figure 5

MOLECULAR EXTINCTION COEFFICIENT

light wavelength	oxyhaemoglobin	haemoglobin
700 nm	90	400
900 nm	300	200
940 nm	300	150

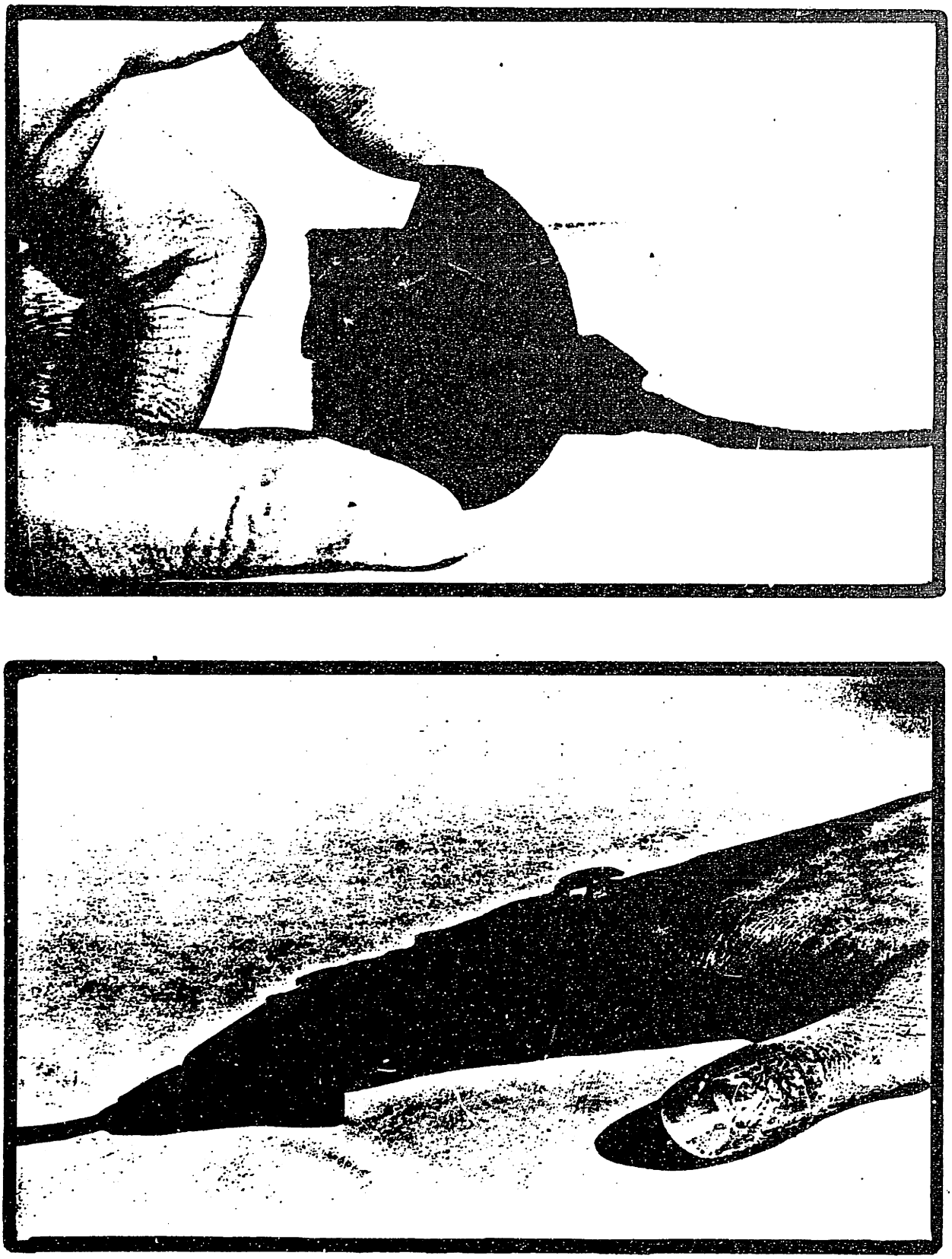
LIGHT TRANSMISSION

light wavelength	dark skin	light skin
700 nm	.04	.2
900 nm	.13	.16
940 nm	.12	.14

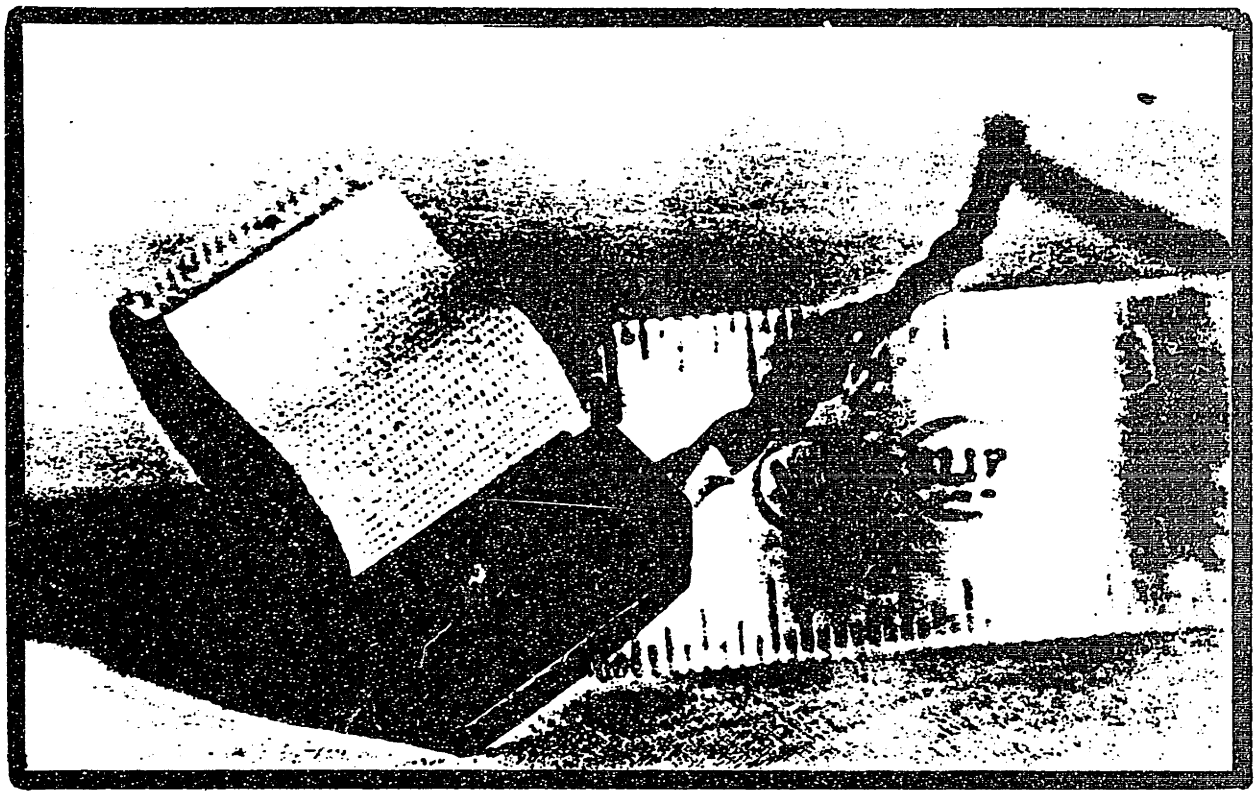
FIGURE 7

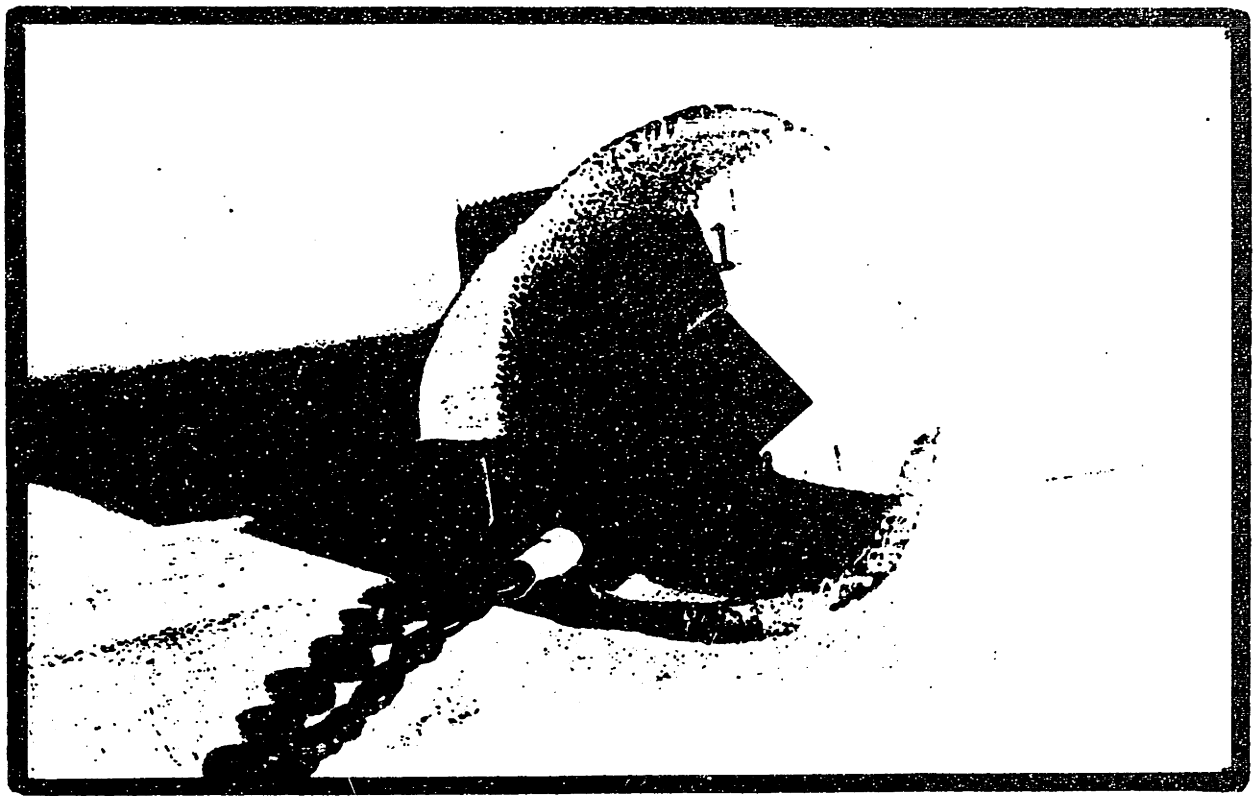
wavelengths would be in keeping with the constraints listed above. One should keep in mind that the possibility exists that my choice of wavelength/frequency (700 nanometers) may have resulted in some oxygen concentration dependent results.

The plethysmographs that were designed and used are shown in Pictures 1 and 2. The light source is an LED, used because its low heat output makes it suitable for direct application to a patient's skin. The LED is a Hewlett Packard 1990-0580 with peak spectral emission at 700 nanometers, driven by 18 milliamps DC (output power approximately 40 milliwatts). The photoresistor is a Clairex 904L, with peak spectral response at 690 nanometers. Figure 8 shows the close overlap of wavelength curves for the LED and photoresistor. To emphasize the importance of matching the spectral responses of the light source and photodetector, see Figure 9. It shows my circuitry's output for both a matched and an unmatched set. The waveforms will be explained later. At present, note the differences in baseline noise. As a result of the mismatch between source and photocell, the photoresistor sees less light and its resistance rises to the megohm range. The capacitance between noise sources (such as flourescent lights) and a person sitting in an average room is roughly 220 picofarads. Using this figure, the impedance between a noise source and a subject, at 60 Hz, is 1/wc or about 12 megohms. This provides for coupling of noise into the circuitry as previously shown (Figure 9), and as modelled in Figure 10. For a matched source and photoresistor, the



The Transmission Photoplethysmograph Picture 1





The Reflective Photoplethysmograph Picture 2

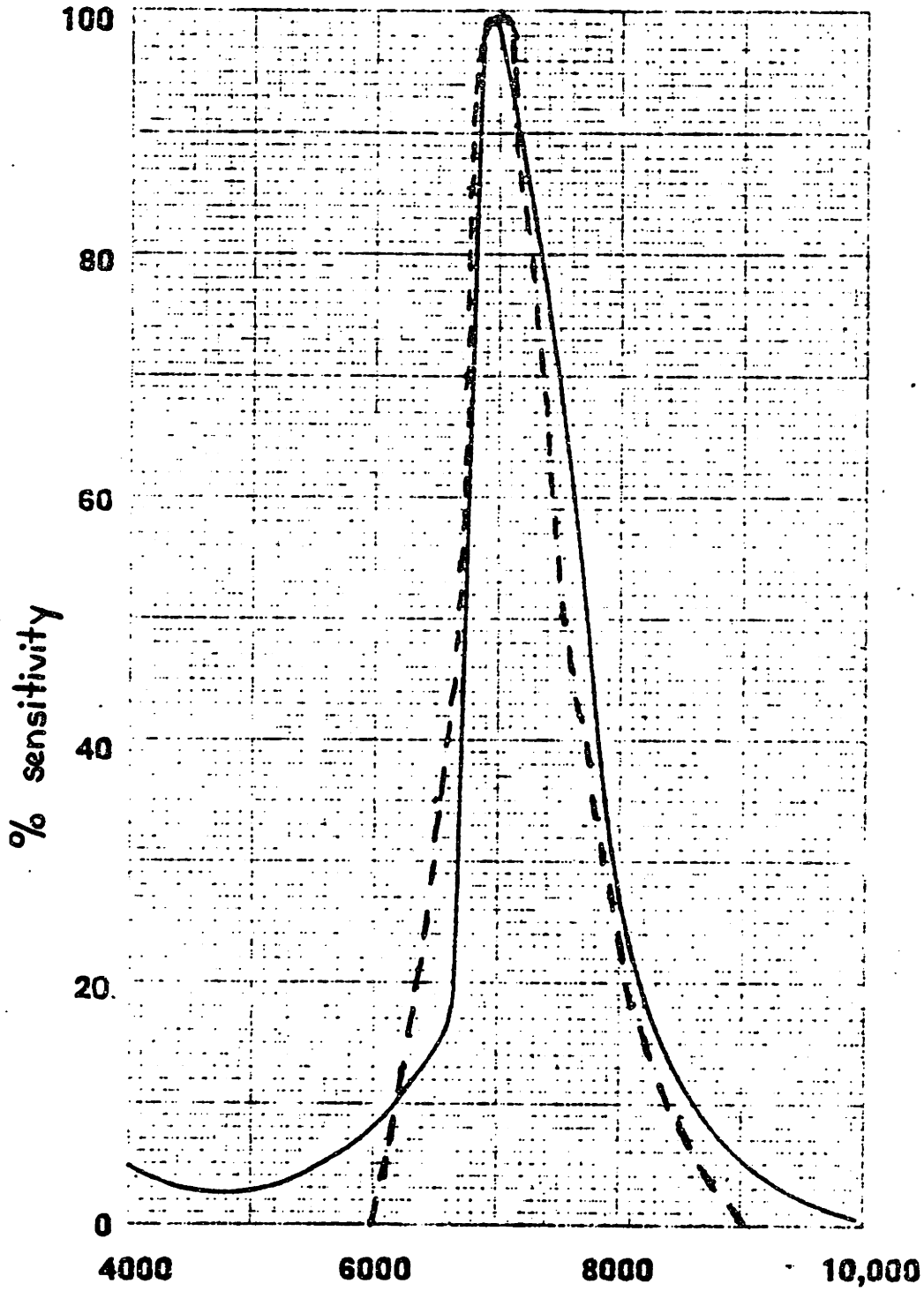
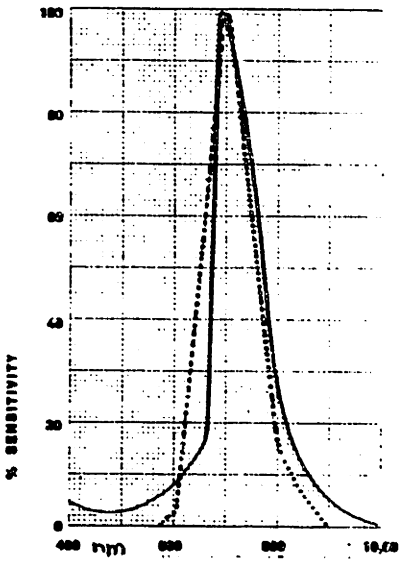
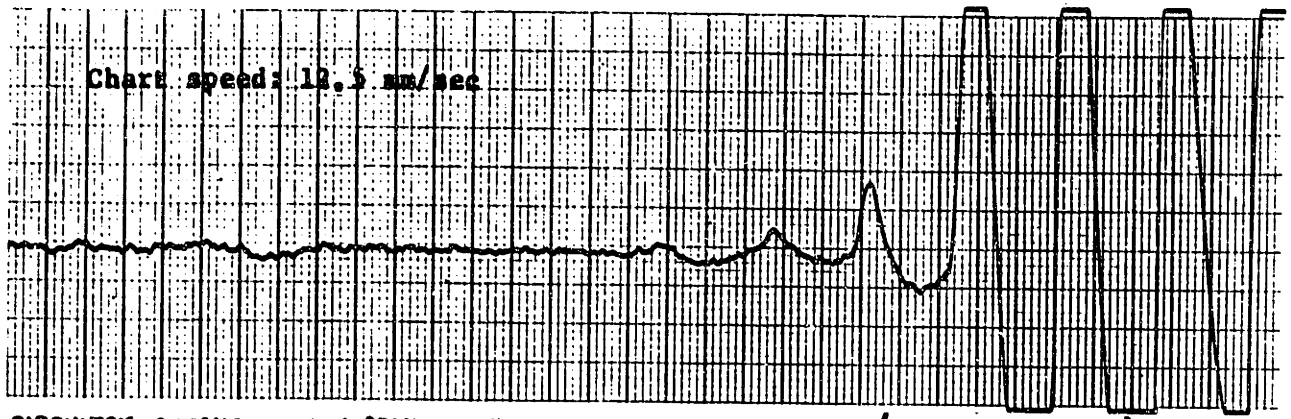

100
80
60
40
20
0
% sensitivity
4000
6000
8000
10,000
ANGSTROMS
HP 1990-0580 RED LED
CLAIREX 904L

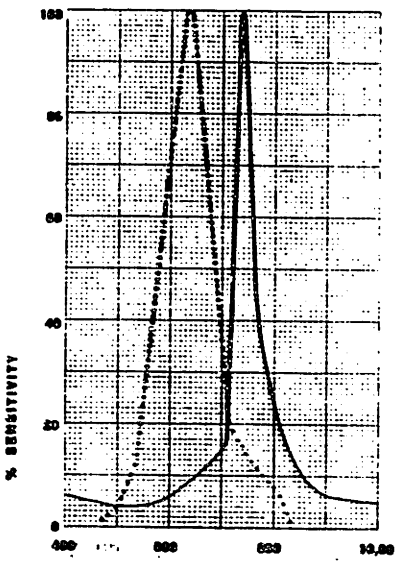
FIGURE 8

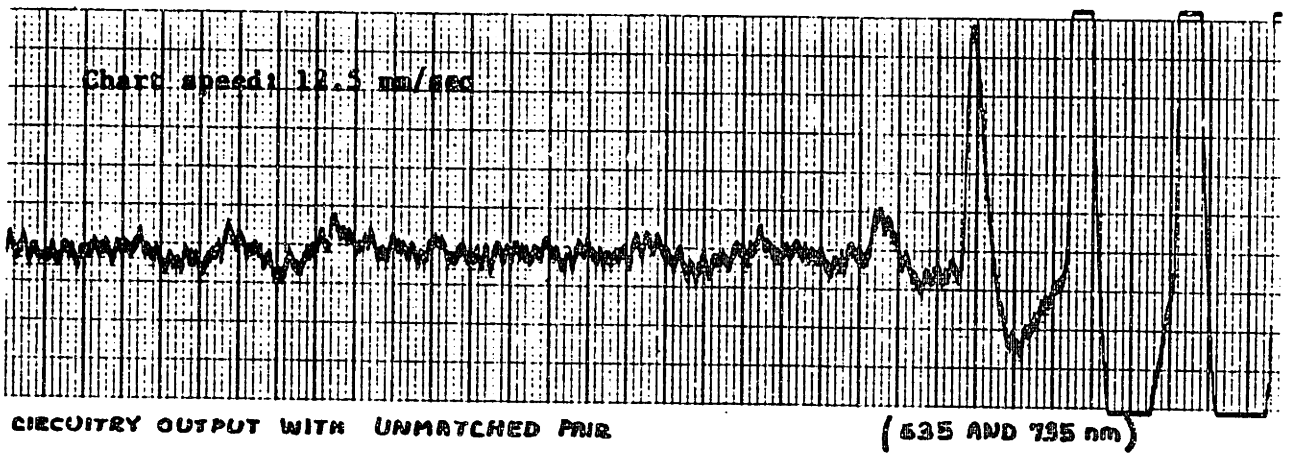


... LED — PHOTORESISTOR



CIRCUITRY OUTPUT WITH MATCHED LED AND PHOTORESISTOR (700 AND 690 nm)





CIRCUITRY OUTPUT WITH UNMATCHED PAIR (635 AND 795 nm)

Same gain for both recordings.

FIGURE 9

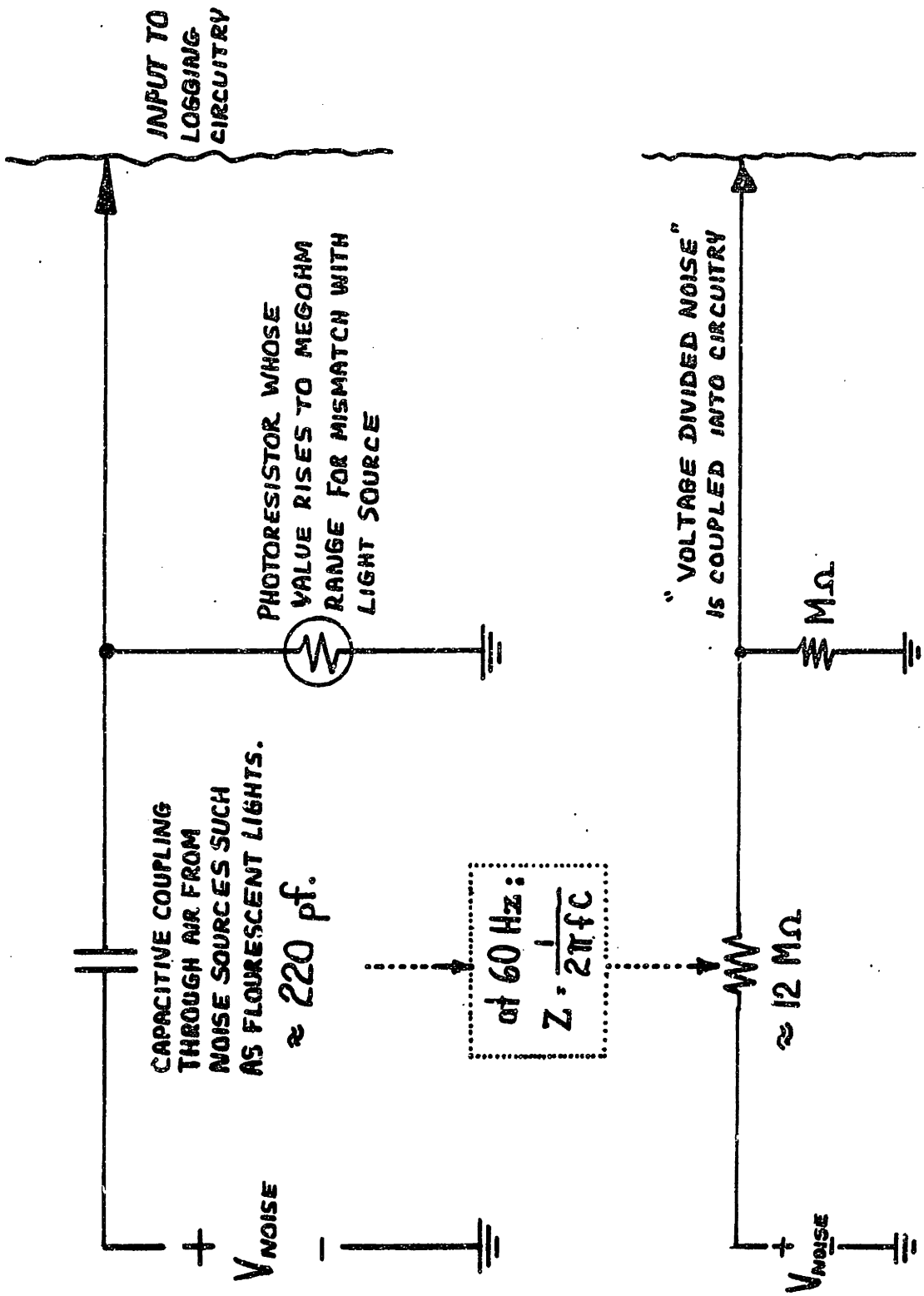
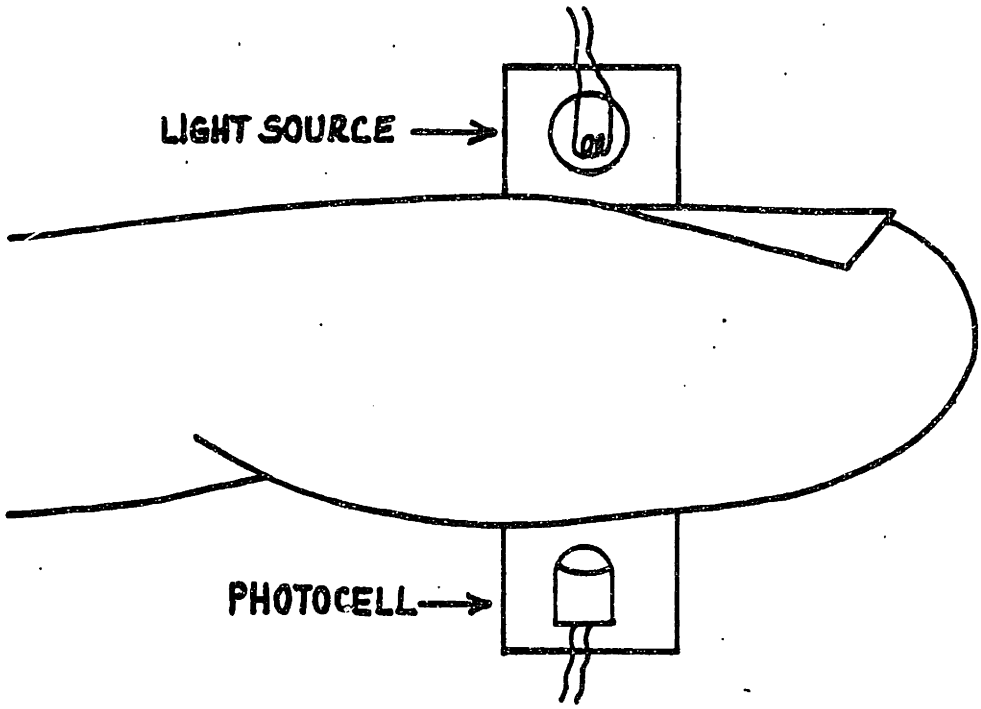

INPUT TO LOGGING CIRCUITRY
PHOTORESISTOR WHOSE VALUE RISES TO MEGOHM RANGE FOR MISMATCH WITH LIGHT SOURCE
CAPACITIVE COUPLING THROUGH AIR FROM NOISE SOURCES SUCH AS FLOURESCENT LIGHTS.
≈ 220 pf.
at 60 Hz: Z = 1/2πfC
≈ 12 MΩ
"VOLTAGE DIVIDED NOISE" IS COUPLED INTO CIRCUITRY
MΩ
+ V NOISE -
+ V NOISE

FIGURE 10

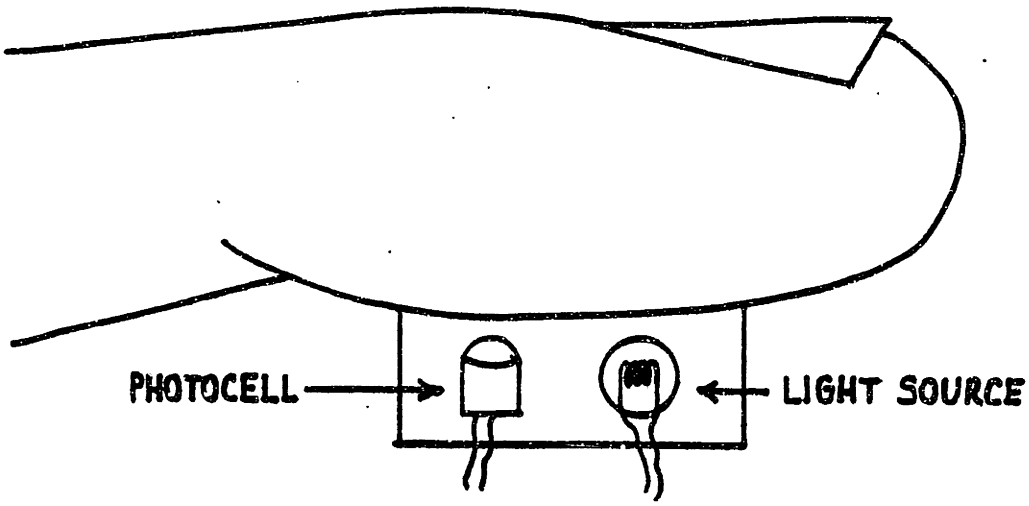
photocell resistance is on the order of 10 to 20 kohms.

Having chosen the components for the plethysmograph, the next considerations deal with geometrical placement of the source and photocell. At this point, one must understand the physiological limitations placed on the photoplethysmography technique. If we are to monitor pulsatile blood flow, we must know what sites on the body provide a useful signal when placed in the receptive field of a photoplethysmograph. The Lambert-Beer development indicates that pulsatile blood flow can be monitored anywhere that an elastic blood vessel lies in the receptive field of the plethysmograph. One would expect that placing the pleth on a limb over an artery would provide a good, clean signal.

When the light source and photocell are placed side by side on the same side of the limb, the signal received will be that input light reflected off the site of interest. When the source and photcell are placed on opposite sides of the limb, the light detected will be that transmitted through the site of interest (Figure 11). Initially, one might be tempted to pick transmission over reflection, under the assumption that since all the received light had passed through the limb, there would be more useable information contained in the received light. One might expect that much of the received light in reflection would just have bounced off the skin surface and would contain no information about the state of the underlying blood vessels. Two types of tests were done to compare the two methods, using a fingertip as the test



A TRANSMISSION PHOTOPLETHYSMOGRAPH IN PLACE ON A FINGER



A REFLECTIVE PHOTOPLETHYSMOGRAPH IN PLACE ON A FINGER

FIGURE 11

site (Pictures 1,3). In the first, circuitry output was recorded with one type of pleth and then with the other. The subject was not disturbed except for the short time interval required to change from one plethysmograph to the other. Comparative amplitude was the evaluating criteria for this test. Representative data is shown in Figure 12. In most cases, transmission yielded a higher amplitude signal than reflection.

The second test was to determine whether the entire system performed "better" with transmission. Given that transmission yielded a larger amplitude output signal, did the entire system pick off systolic and/or diastolic blood pressure points with more accuracy? To determine this, a series of trials were done on different subjects. The details of how the system measures systolic blood pressure will be given in a later section. Suffice it to say that for each trial, systolic blood pressure was determined using one type of pleth, and then shortly afterwards with the other type. To avoid any possible physiologic factors affecting the data, the order of the two measurements was mixed throughout the trials. If transmission yields a more sensitive system, one would expect to obtain a higher reading for systolic pressure with transmission than with reflection. The results are listed in Figure 13. Transmission did not improve system performance. By the evaluating criteria, one could even say that reflection is better. The most useful conclusion is that either will perform adequately, and other criteria should be

The Reflective Photoplethysmograph in place on a fingertip.

Picture 3

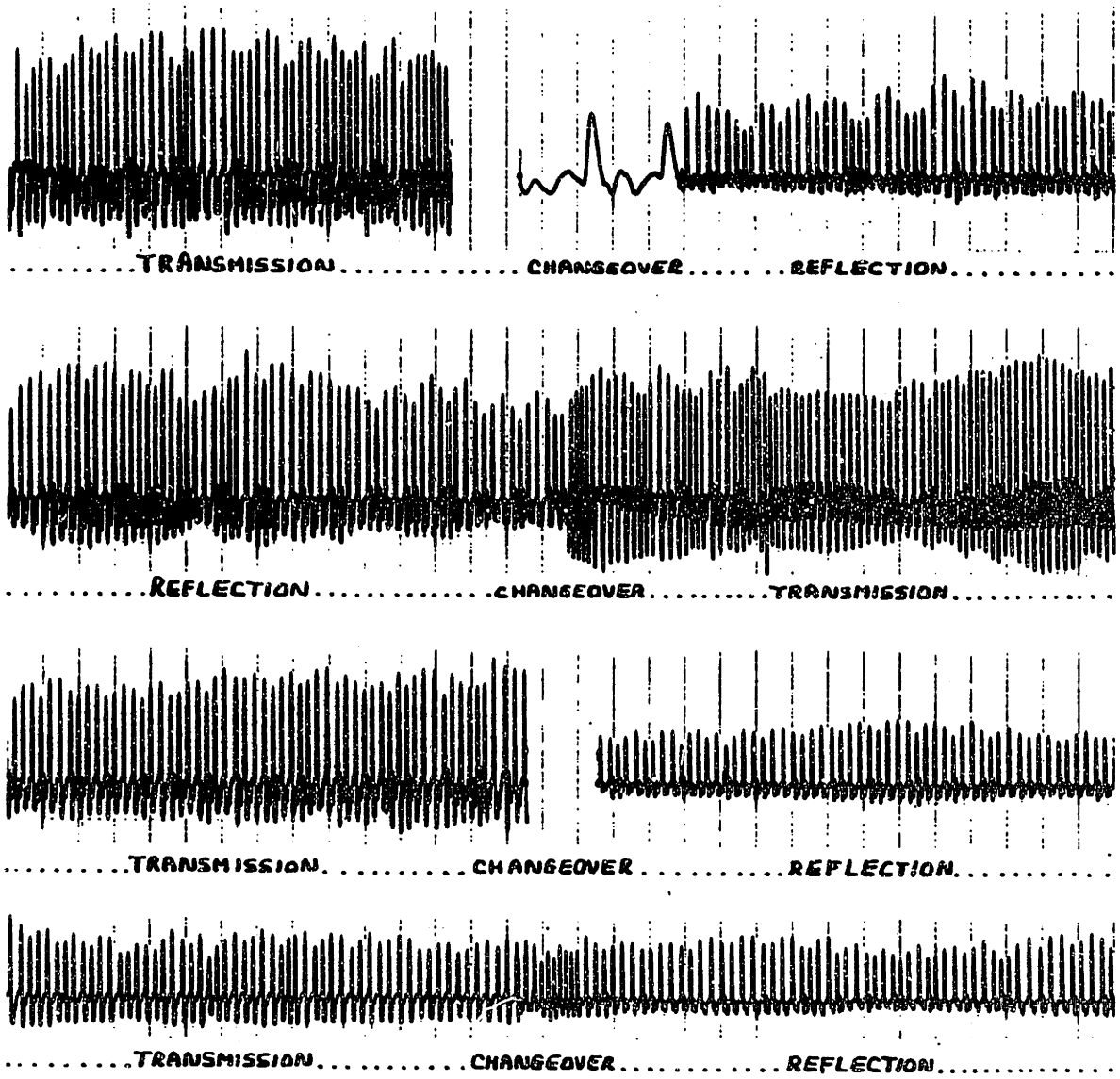


CHART RECORDER SPEED IS 1.25 mm/sec

FIGURE 12

Systolic blood pressure

trial	transmission (a)	reflection (b)	difference = a-b
1	128 mm Hg	130 mm Hg	-2 mm
2	112 mm Hg	114 mm Hg	-2 mm
3	107 mm Hg	107 mm Hg	none
4	114 mm Hg	114 mm Hg	none
5	126 mm Hg	131 mm Hg	-5 mm
6	118 mm Hg	120 mm Hg	-2 mm
7	120 mm Hg	120 mm Hg	none
8	125 mm Hg	125 mm Hg	none
9	117 mm Hg	125 mm Hg	-8 mm
10	109 mm Hg	112 mm Hg	-3 mm
11	101 mm Hg	98 mm Hg	+3 mm

..... average difference over these 11 trials is -1.6 mm Hg .....

Figure 13

considered in deciding which plethysmograph to use.

Since either a reflective or a transmission plethysmograph will perform adequately, the choice of which to use depends upon the body site on which the pleth will be placed. The pleth could be placed on the arm, just below the occlusion cuff and over the brachial artery. This would allow for precise monitoring of blood flow or lack thereof as cuff pressure was changed. It would also allow for a good comparison of the optical technique with the auscultatory one, which monitors blood flow in the brachial artery.

As it turned out, the arm was too thick for the transmission plethysmograph to work. All the intermediate tissue and muscle in the arm adversly affected the reflective plethysmograph pickup. In fact, these factors rendered the optical pickup on the upper arm useless. No useable signal was obtained. A higher intensity light source might "probe" the arm with more success. However, significantly more light would require use of an incandescent bulb which would generate too much heat for use on or near a patient's skin.

In search of a suitable pickup site, the plethysmograph was moved further down the arm. It was hoped that the thinner wrist area would contain less muscle, fat, and other interferences, and would yield a useable signal. Nothing but noise was detected with the transmission pleth. With reflection, one could discern on some subjects the basic waveform of pulsatile blood flow in the radial artery. The signal was noisy and inconsistent, not useful or reliable for

use in further experiments.

Given those sources who claim that capillaries do expand in response to pulsatile blood flow [22,14,13], an attempt was made to use the capillaries near the skin's surface in the arm right below the cuff, by pinching the skin and using the transmission plethysmograph. Only noise was seen at the circuitry output.

The final place tried, and the site used in all subsequent testing, was the fingertips of the hands (Pictures 1,3). My research showed that one can obtain a good, clean signal, though the signal source is not exactly understood. The literature is vague. Many of the articles I have read explain that plethysmography monitors pulsatile blood flow by detecting increases in the quantity of blood within the finger. These articles are not clear about what exactly is being monitored: artery stretching, capillary stretching, overall finger pulsatile expansion, or something else [16,26, 30,25,10]. Some sources believe that pulsatile capillary expansion is being monitored [22,14], while others state that capillaries do not undergo pulsatile expansion in response to blood flow [30]. It is possible that one is monitoring the arteriovenous anastomoses in the fingers, since these vessels are elastic [30].

The factors and problems that affect the choice of pickup site on the arm would also apply to the leg and foot. The pickup site must be well perfused and thin enough for the weak pleth light source to pervade the test site and be

"modulated" by the pulsatile blood flow within. The ear would also meet these requirements. As a pickup site it is of no use to the occlusion cuff/blood pressure measurement. However, monitoring pulsatile blood flow in the ear gives an indication of overall perfusion of the head, which is important data for physicians to have. Figure 14 shows the relative amplitudes of circuitry output for signals picked off the pinna of the ear and a fingertip using a transmission pleth.

For completeness, note that one need not use only the fingertip. A good signal, though of lower amplitude, can be obtained by placing either type of plethysmograph further up the finger, as shown in Figure 15.

The plethysmograph used in the experiments was the reflective type (shown in Pictures 2,3), unless otherwise noted. I found it more convenient and comfortable to use. The transmission type, shown in Picture 1, is a Hewlett Packard 14385A plethysmograph, used in the HP 47205A Perfusion Indicator. It was designed to fit over the pinna of the ear, and squeezed the fingertips of larger fingers too much to be comfortable.

The effects of varying the spacing between the LED and the photocell were investigated. One might expect that the closer the light source to the photocell, the more intense the received signal would be. Within limits, this is not the case. Figure 16 shows three representative trials from different subjects. In trials A and B, the left trace was

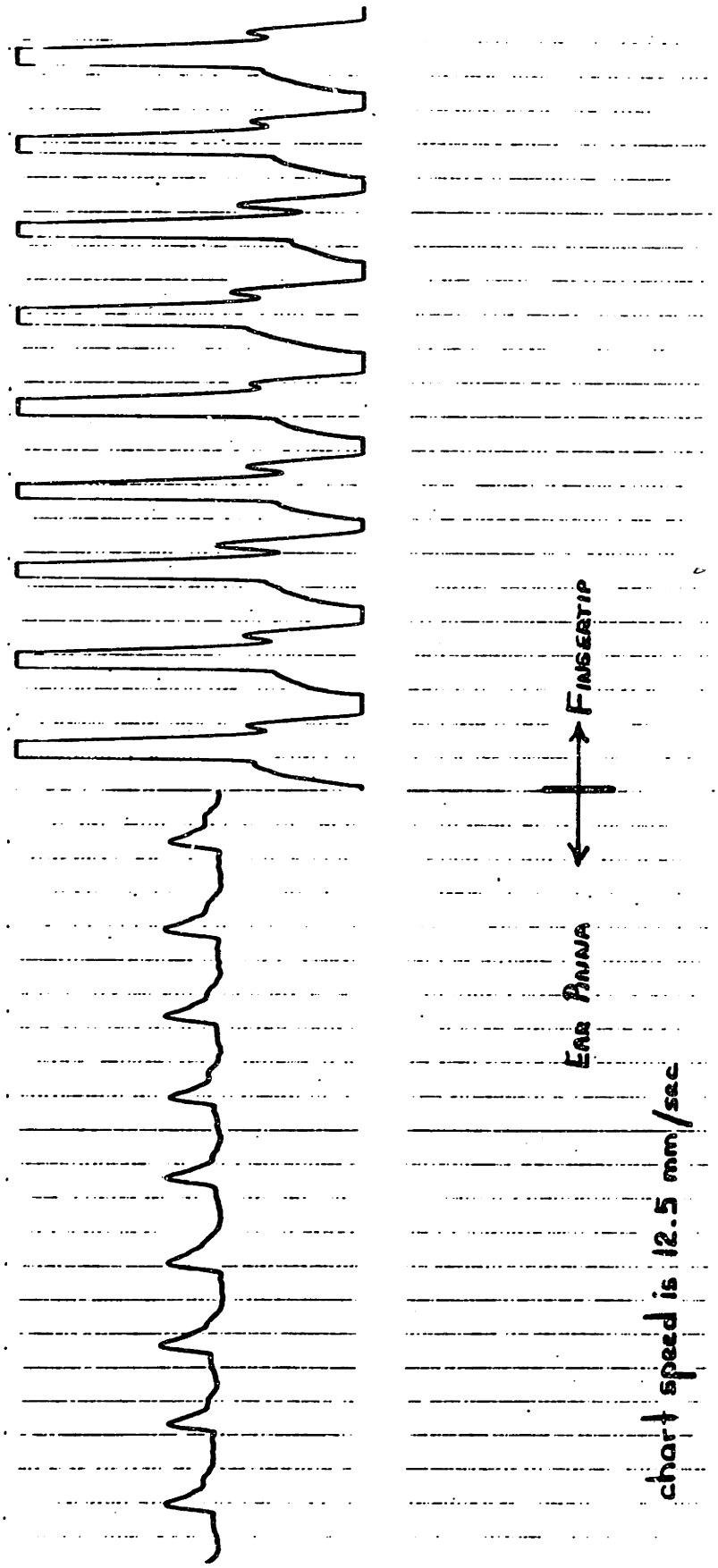
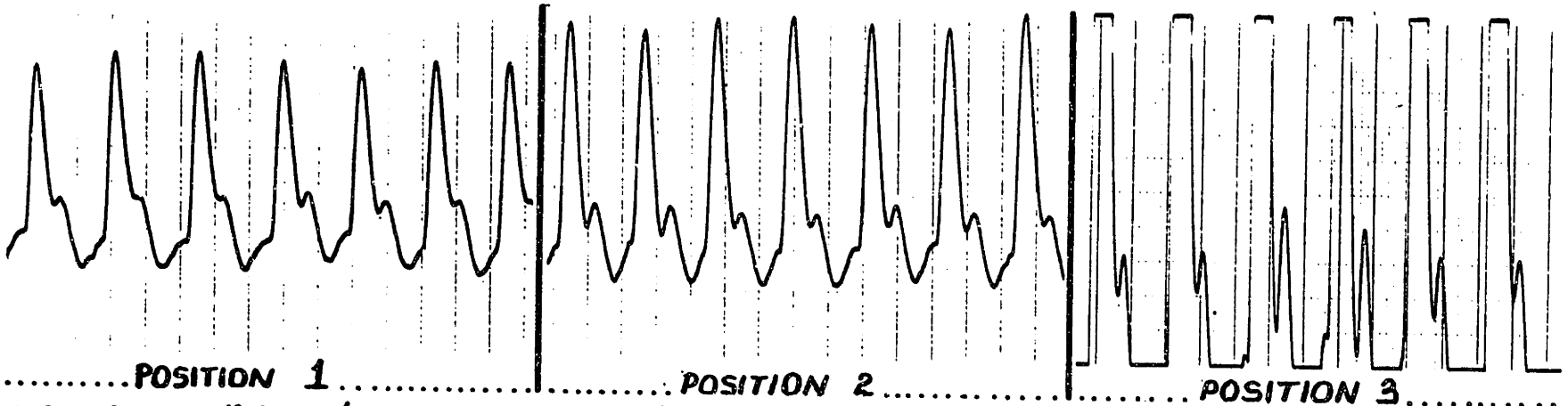


FIGURE 14


POSITION 1
POSITION 2
POSITION 3
CHART SPEED IS 12.5 mm/sec

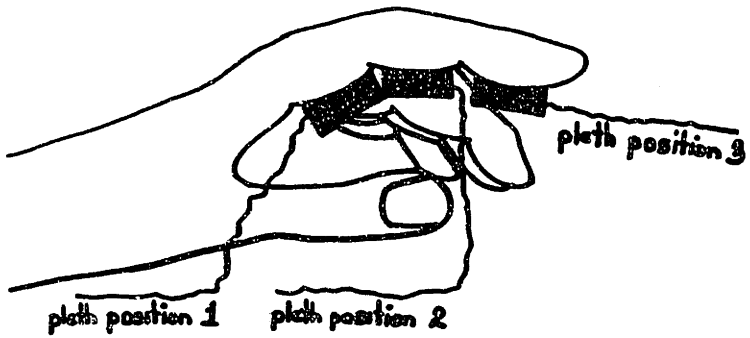
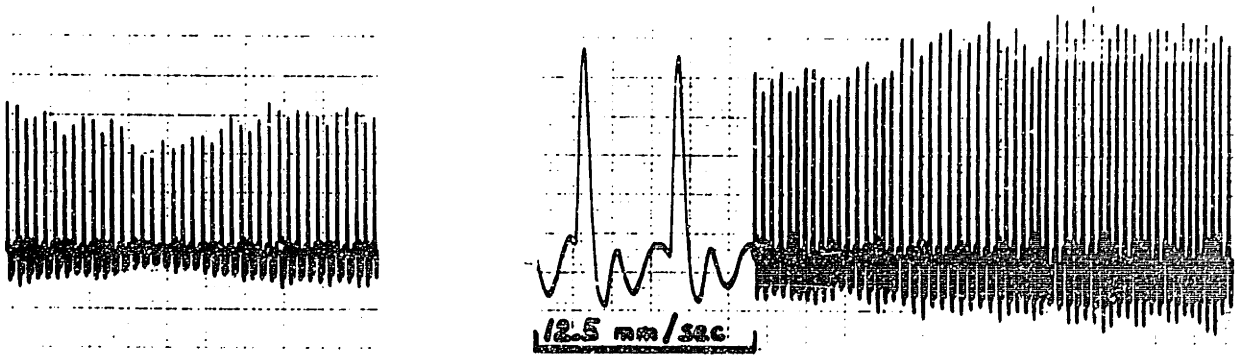
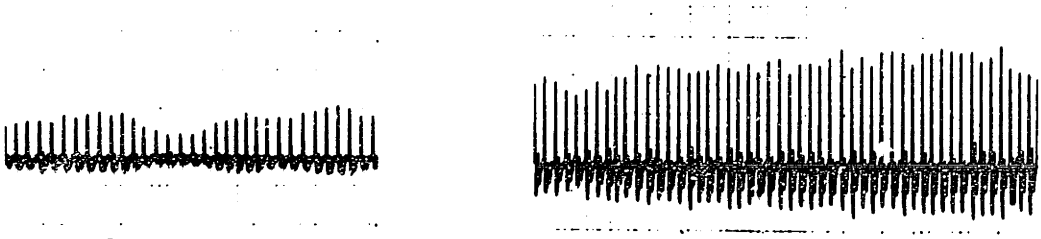

pleth position 3
pleth position 1
pleth position 2

FIGURE 15

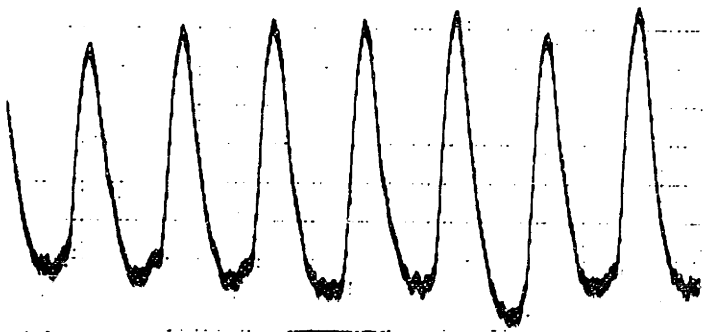


TRIAL A : LED AND
PHOTOCELL SIDE BY SIDE ..... SPACED APPX. 1 cm APART



TRIAL B: LED AND
PHOTOCELL SIDE BY SIDE.....SPACED APPX. 1cm APART

CHART SPEED IS 1.25 mm/sec
UNLESS OTHERWISE NOTED.



TRIAL C: SPACED APPX. 2.5 cm APART
12.5 mm/sec

FIGURE 16

made with the LED and photocell placed side by side (center to center separation of 0.51 cm). For the right trace, the LED and photocell were approximately 1.02 cm apart. Ignore the details of the waveform, but note the overall increase in amplitude as the plethysmograph components were separated. The photocell has a "field of sensitivity". As the light source is moved further away, more of the input light that has been scattered by obstacles in its path is received by the photocell. Consequently, within limit, the signal amplitude is larger. Too much separation causes too little light to reach the photocell, resulting in increased noise and decreased signal amplitude. Trial C, as compared with the clean waveform of trial A of Figure 16, shows the increased noise that accompanies excessive separation.

Component spacing affects overall pleth size. The smaller the pleth, the easier it is to securely attach it to the finger. This is advantageous because the pleths are very subject to motion artifact.

The reflective pleth used has a spacing of 1.02 cm between LED and photoresistor. A clean, high amplitude signal is obtained, yet overall pleth size is only 2.54 cm by 1.14 cm by 1.40 cm.

THE SIGNAL PROCESSING CIRCUITRY

The plethysmograph is the means by which the physiological signal is input to the processing system. Now the processing circuitry will be presented. Its purpose is to convert the physiological input into meaningful information for the user.

The circuitry can be divided into three main parts; the natural log circuitry, bandpass filtering, and amplification.

The Natural Log Circuitry

One of the simplest circuits that performs the natural log function is shown in Figure 17. The following relationships apply:

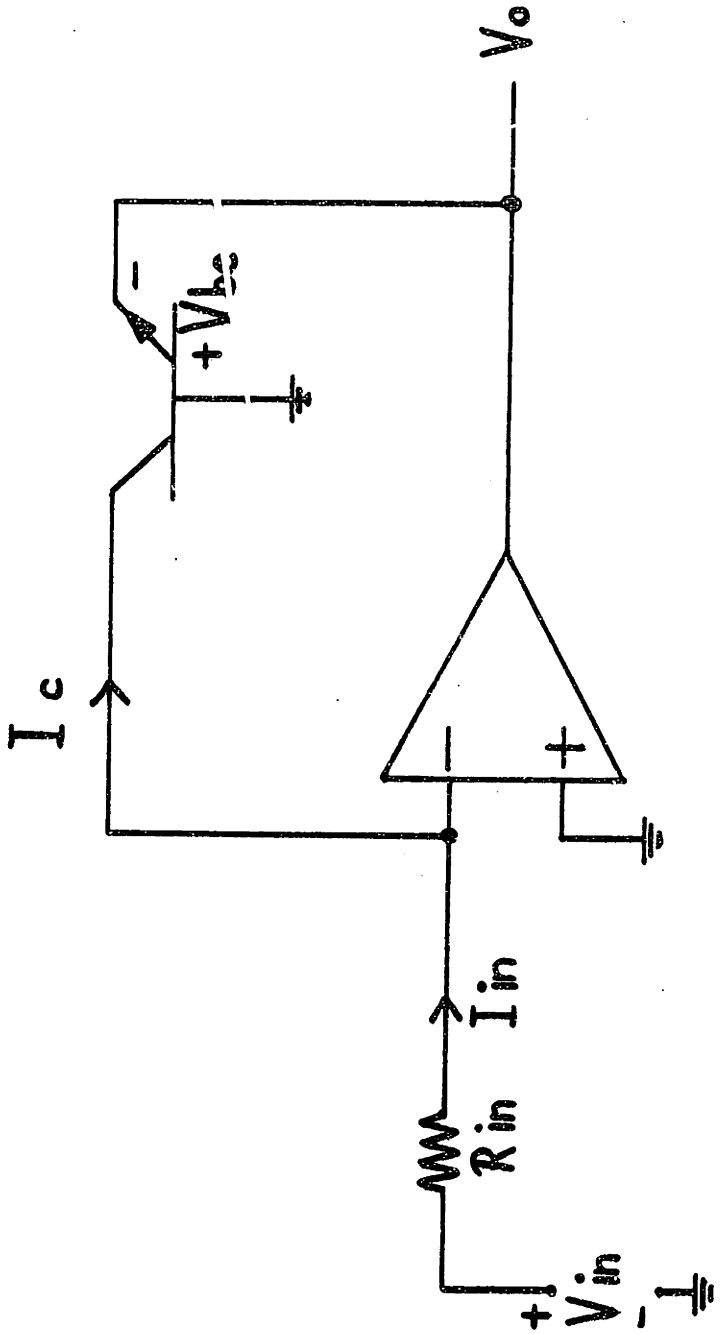
a. $V_{be} = -V_o$ (8)

b. $I_c = I_s [\exp(qV_{be}/KT) - 1]$ where I_s is a constant, the saturation current for a transistor, on the order of 10E-13 amperes. Generally, $I_c/I_s \gg 1$ and (b) can be manipulated and simplified:

$$V_{be} \sim (KT/q) \ln(I_c/I_s) \quad (9)$$

c. The op amp adjusts V_o so that V_{be} biases the transistor to conduct the input current through it.

Consequently, $I_c = I_{in}$ (10)



for $V_{in} >$ ⏚

FIGURE 17

Putting all these relationships together allows one to solve for Vo: $Vo \sim -(KT/q)\ln Iin + (KT/q)\ln Is$

$$Vo \sim -(KT/q)\ln Iin + Constant \qquad (11)$$

At room temerature (293 degrees Kelvin), with K equal to 1.38E-23 joules per degree Kelvin, q equal to 1.6E-19 coulombs, and Is equal to approximately 10E-13 amperes, (KT/q) is approximately 0.025 volts. The constant in equation (11) is about -0.70 volts. Hence Vo equals:

$$Vo \sim (-.025 \text{ volts}) \ln(Iin) - 0.70 \text{ volts.} \qquad (12)$$

The output of this current driven device provides the natural log of Iin modified by constants. With Rin equal to 10 Kohms, tests were done to verify circuit performance. Figure 18 shows the results. There is a fairly constant error for low current, attributable to inaccuracy in calculating the (KT/q)ln Is constant. Note how dependent the constant and (KT/q) are on the value of temperature, and the dependency of the constant on Is which varies from transistor to transistor.

To use this design for the optical plethysmograph logging circuit I would hook the photoresistor into the circuit in place of Rin. For a fixed Vin, Rin fluctuations would result in Iin fluctuations. Because of the constraint that the HP 14385A transmission plethysmograph has one side of the photoresistor tied to ground, the circuit would have to be

Vo FOR VARIOUS VALUES OF Iin

Vo ~ $-(kT/q)\ln(Iin) - .70$ for T= 293 degrees Kelvin
Is = 10E-13 amperes

Iin	(a) observed Vo	(b) calculated Vo	error=a-b
1 mamp	-.66 volts	-.53 volts	-.13
0.95 mamps	-.67 volts	-.52 volts	-.15
0.90 mamps	-.67 volts	-.52 volts	-.15
0.85 mamps	-.67 volts	-.52 volts	-.15
0.80 mamps	-.67 volts	-.52 volts	-.15
0.70 mamps	-.66 volts	-.52 volts	-.14
0.60 mamps	-.66 volts	-.51 volts	-.15
0.50 mamps	-.65 volts	-.51 volts	-.14
0.40 mamps	-.64 volts	-.50 volts	-.14
0.30 mamps	-.63 volts	-.50 volts	-.13
0.20 mamps	-.61 volts	-.48 volts	-.13
0.10 mamps	-.58 volts	-.47 volts	-.11
--) Results for the expanded region of .001 to .095 mamps (--			
0.095	-.583 volts	-.466 volts	-.117
0.09	-.582 volts	-.465 volts	-.117
0.085	-.580 volts	-.463 volts	-.117
0.08	-.578 volts	-.462 volts	-.116
0.075	-.576 volts	-.460 volts	-.116
0.07	-.573 volts	-.458 volts	-.115
0.065	-.570 volts	-.456 volts	-.114
0.06	-.568 volts	-.454 volts	-.114
0.055	-.564 volts	-.452 volts	-.112
0.05	-.560 volts	-.450 volts	-.110
0.045	-.556 volts	-.447 volts	-.109
0.04	-.551 volts	-.444 volts	-.107
0.035	-.546 volts	-.441 volts	-.105
0.03	-.541 volts	-.437 volts	-.104
0.025	-.535 volts	-.432 volts	-.103
0.02	-.531 volts	-.427 volts	-.104
0.015	-.531 volts	-.419 volts	-.112
0.01	-.538 volts	-.409 volts	-.129
0.005	-.520 volts	-.392 volts	-.128
0.001	-.484 volts	-.351 volts	-.133

FIGURE 18

modified to work properly, and is shown in Figure 19. Since an op amp always tries to keep both its inputs at the same voltage level, the negative input would move to -Vref. Iin would have the proper direction and the circuitry would work. With the configuration of Figure 19, Vbe equals -Vref - Vo. The circuitry analysis is the same as before, the basic concept being that Iin (which equals Vref/Rpleth) must equal Ic. The final relationship is:

$$Vo \sim -Vref - (KT/q)\ln(Vref/Rpleth) + (KT/q)\ln Is \qquad (13)$$

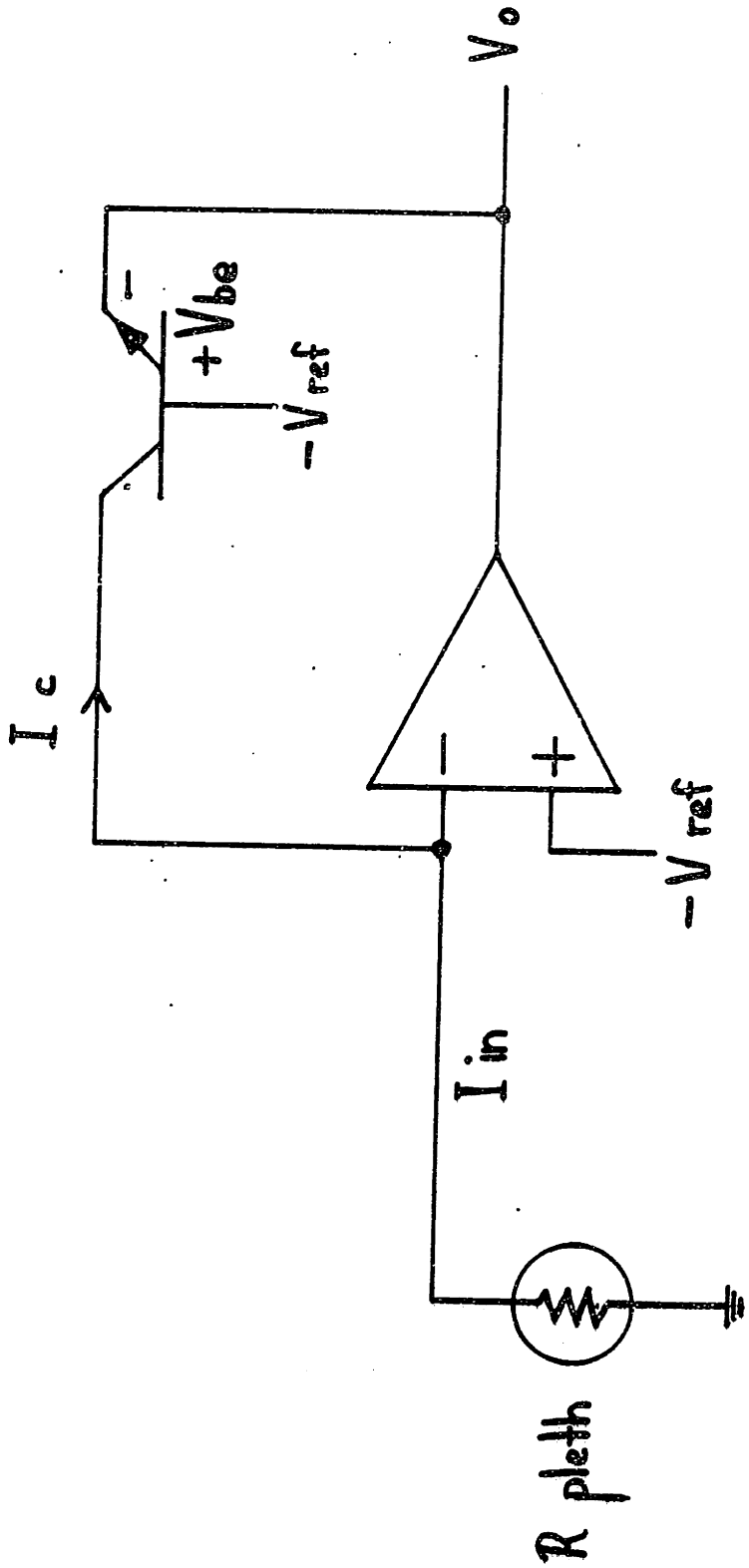
For my purpose, this circuit had two shortcomings. It was dependent on the value of Is, and it lacked any gain in front of the (KT/q)ln term. The final circuit used avoids these shortcomings. First refer to Figure 20. Shown is a natural log circuit using two transistors and a constant current source I. Vo equals V2 - V1. As before, V1 is approximately (KT/q)ln(Iin/Is1). V2 is approximately (KT/q)ln(I/Is2). Using these relationships one solves for Vo:

$$Vo \sim -(KT/q)\ln[(Iin/I)(Is2/Is1)] \qquad (14)$$

If the transistors are a matched pair, Is1 and Is2 will be equal and Vo will be independent of the saturation currents.

Figure 21 shows the final natural log circuit used in the research. The two transistors are on a single chip (a matched pair) so the Is values for each will be approximately equal.

FIGURE 19



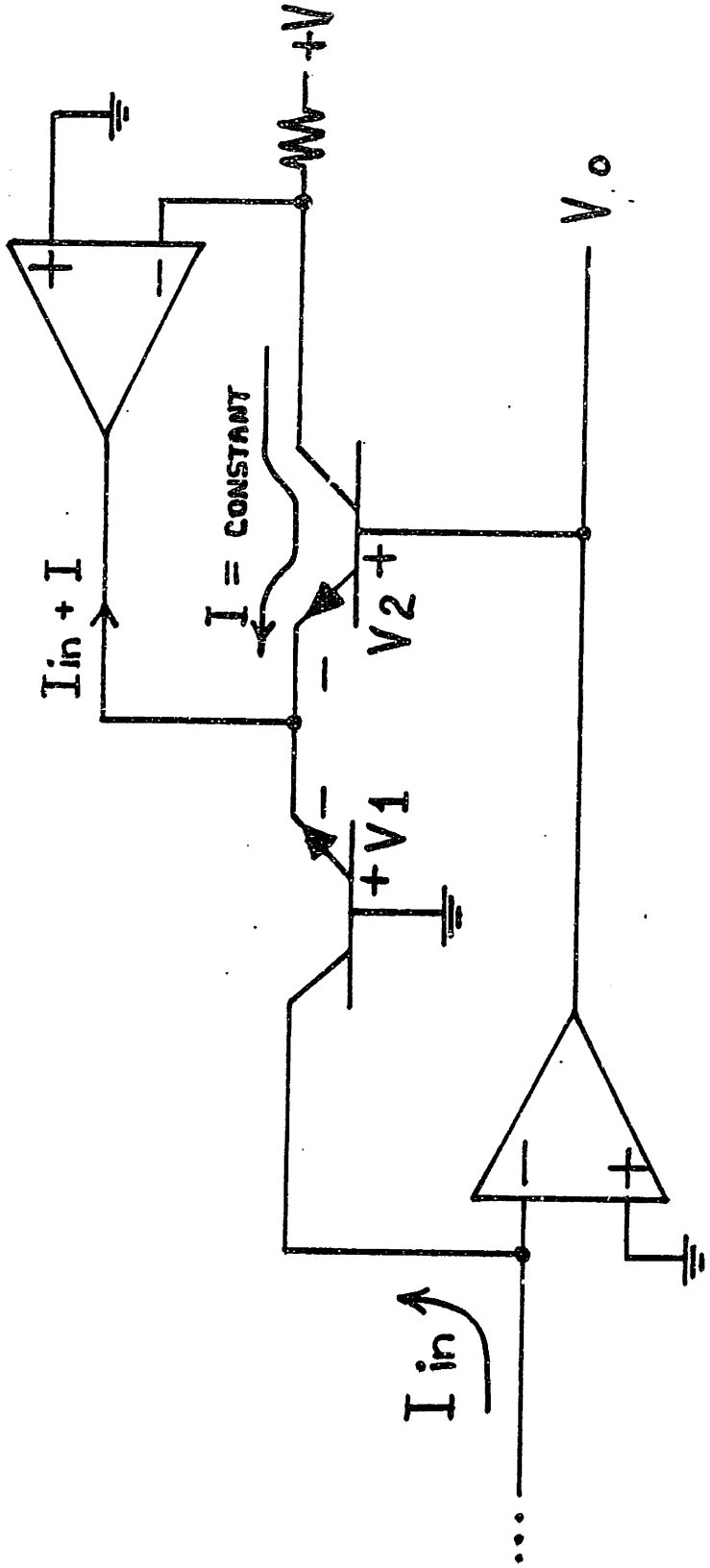


FIGURE 20

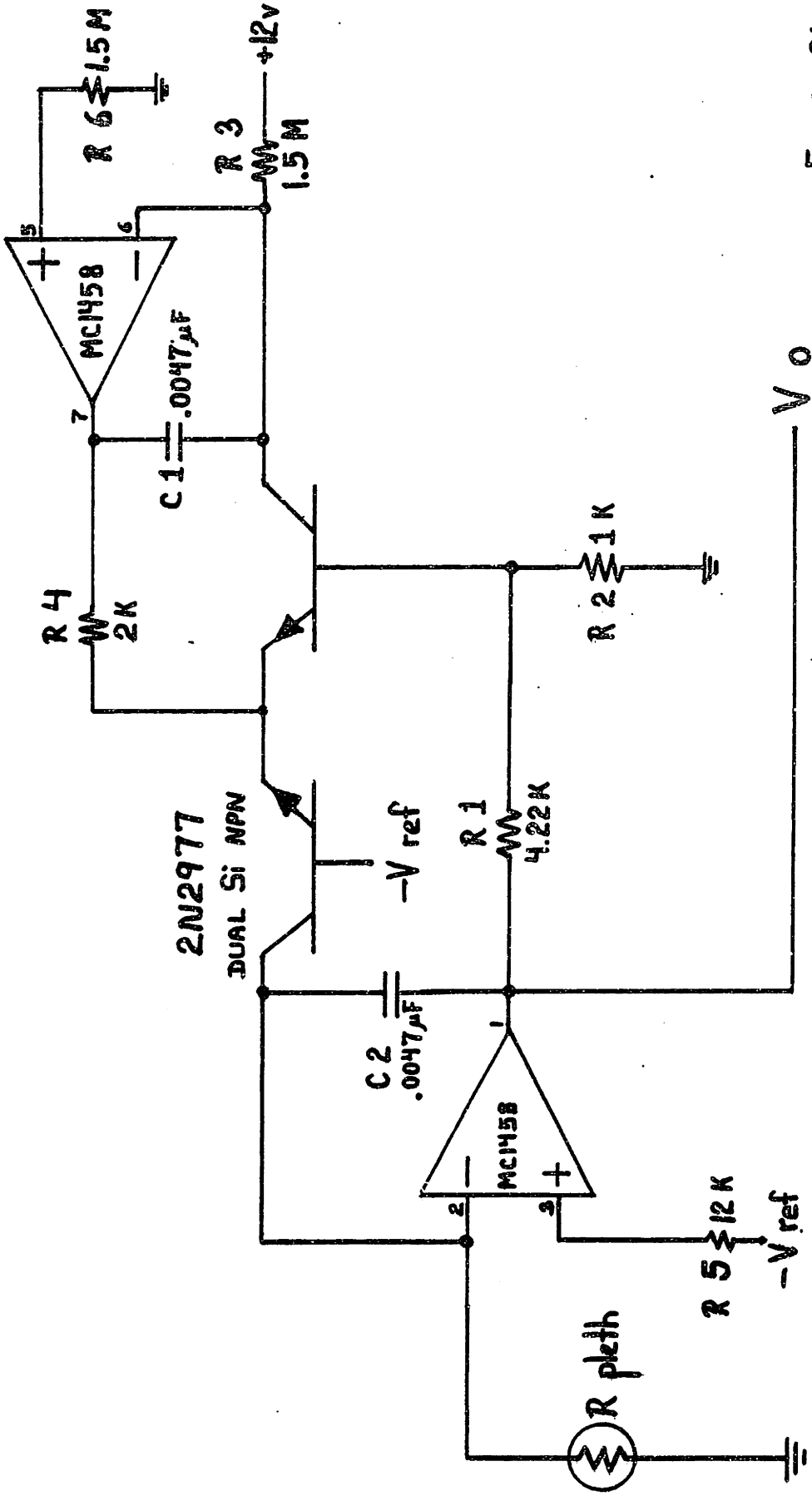

R 6
1.5 M
+12v
R 3
1.5 M
MC1458
C 1
.0047μF
R 4
2K
R 2
1 K
V o
2N2977
DUAL Si NPN
-V ref
R 1
4.22K
C 2
.0047μF
MC1458
R 5
12 K
-V ref
R pleth

FIGURE 21

The two capacitors, C1 and C2 are used to compensate the op amps. The 12 volts and R3 set the constant current source I to 8 microamps. R5 and R6 serve to match resistances seen looking out of the inputs of their respective op amps. R1 and R2 provide for a gain of 5 as compared with the previous circuits. This circuit's transfer function is:

$$Vo \sim -[(R1+R2)/R2]\,[(KT/q)\ln(Iin/I) + Vref] \qquad (15)$$

By making I small, the value of Vo is increased. The exact value of Vref is not critical, though it must be stable since it directly affects the value of Iin. Vref was set to 1 volt. When placed on a fingertip, the photoresistor has an average value of roughly 10-20 Kohms, and Iin will be less than 0.1 milliamp. This minimal amount of current avoids unnecessary self-heating of the photoresistor. To implement the -1 volt for -Vref, a zener controlled reference voltage was designed as explained in Appendix 1.

The output voltage of the natural log circuitry is dominated by the -[(R1 + R2)/R2] Vref term which (for the component values used) pulls Vo down to -5.22 volts. The fluctuations of Vo around this baseline due to pulsatile blood flow are minute. The amplitude of the AC signal component of Vo rarely exceeds 2 millivolts. Nevertheless, it is this AC component which is of interest to us.

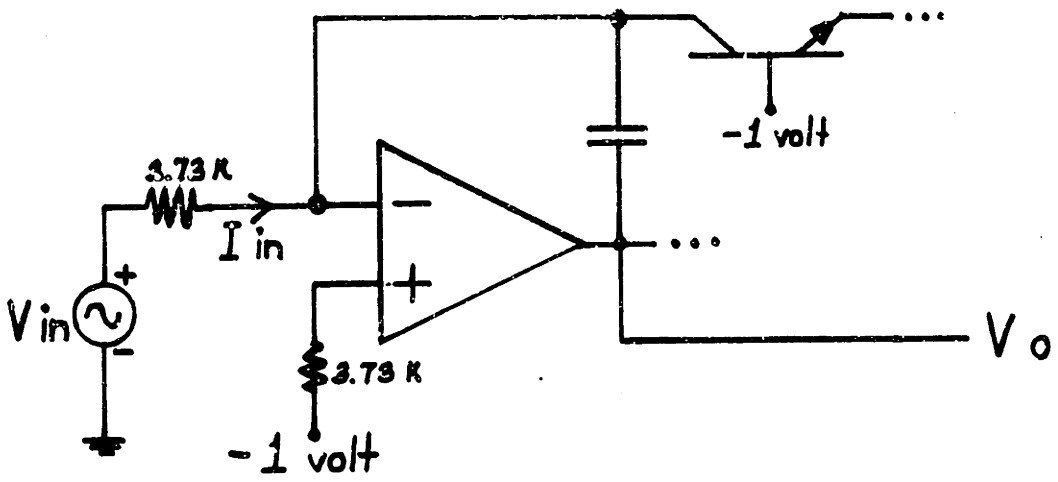
As pictured in Figure 3, the pressure at which the first circuitry output pulse occurs is the systolic blood pressure

for the individual. The first pulse's amplitude is very low. High gain is necessary both to discern it with accuracy and repeatability, and to provide a system output capable of driving a chart recorder. Raising the AC component of the ln circuit output from millivolts to volts requires filtering out the DC level and gain of a few thousand.

Filtering

The filtering used creates a passband for the natural log circuit output. DC is filtered out to allow for high amplification of the important AC component. High frequencies are filtered out to reduce noise on the system output.

The passband cutoffs are selected to bracket the physiologic signal frequency spectrum. Determination of the corner frequencies is a multiple step process. First, one must verify that the natural log circuitry itself has a flat frequency response, i.e. that it does not in any way "color" the physiologic signal spectrum. The natural log circuit was configured as shown in Figure 22. The expression for Vo (Eqtn 15) was explained in the previous section. Also shown is a sample calculation of Vo for a sine wave input swinging between -0.2 and -0.8 volts. Figure 23 shows the observed peak to peak values of Vo as a function of Vin's frequency. The observed values at low frequency are close enough to the calculated one of Figure 22 to indicate proper circuit operation. The circuit exhibits flat frequency response out to about 1 kHz. Provided that the physiologic signal does not



$$V_o = -(5.22)(.025)\ln\left[(I_{in})\left(\frac{1.5\,M\Omega}{12\,v}\right)\right] - 5.22 \text{ volts}$$

for V in swinging between -0.2 and -0.8 volts...

with V in = -0.2 volts, $I_{in} = \left[\frac{-0.2+1}{3.73\,k}\right] = 2.145 \cdot 10^{-4}$ amps

and V_{o1} will be -5.65 volts

with V in = -0.8 volts, $I_{in} = \left[\frac{-0.8+1}{3.73\,k}\right] = 5.362 \cdot 10^{-5}$ amps

and V_{o2} will be -5.47 volts

Peak to peak output voltage is $|V_{o1} - V_{o2}| = \boxed{0.18 \text{ volts}}$

FIGURE 22

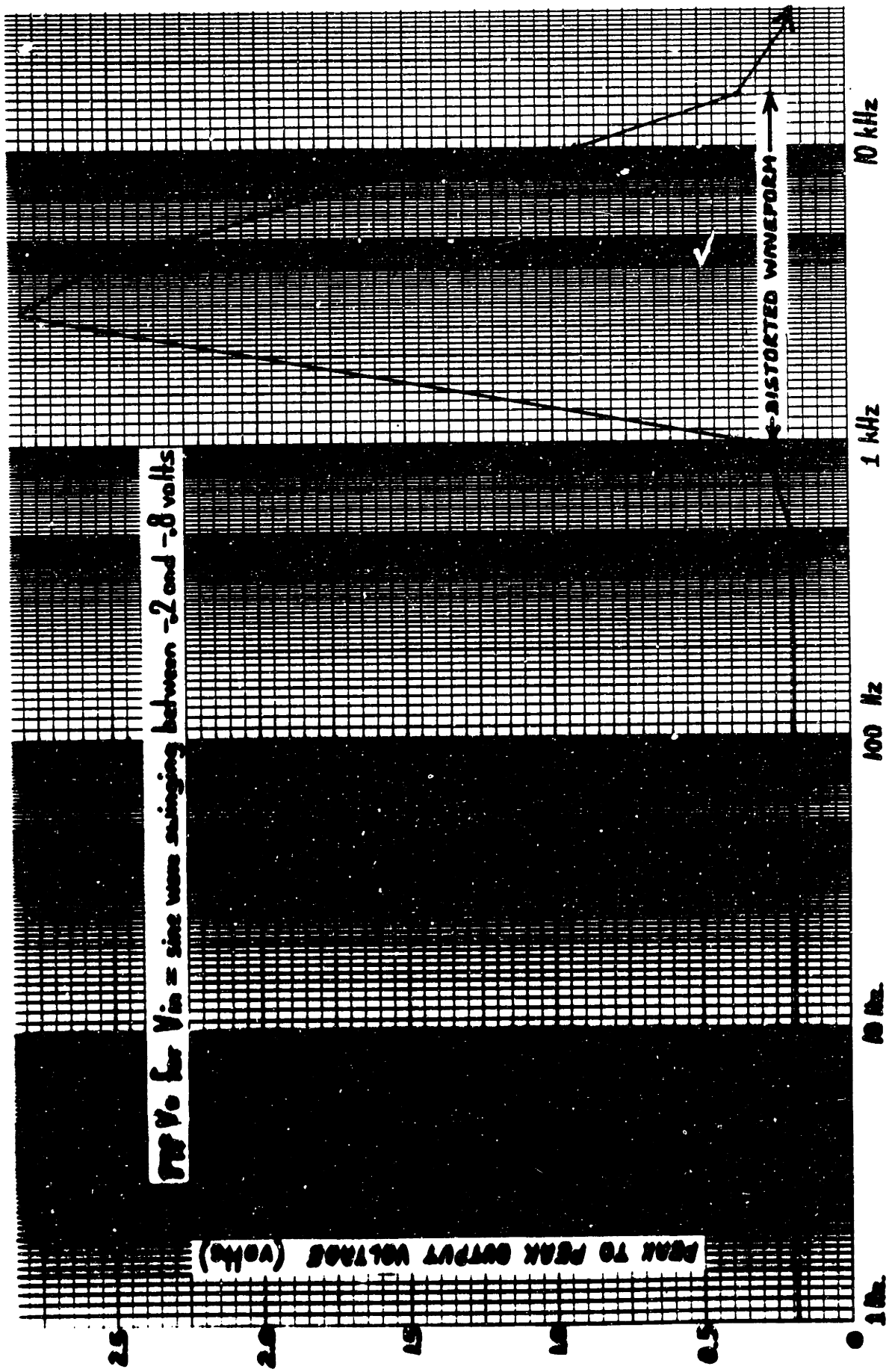

10 kHz
1 kHz
100 Hz
10 Hz
1 Hz
DISTORTED WAVEFORM
PEAK TO PEAK SUPPLY VOLTAGE (volts)

FIGURE 23

have frequency components above 1 kHz [8], the frequency spectrum of the natural log circuitry output will reflect that of the physiologic signal. Next, we determine that spectrum.

Using a Hewlett Packard 3582A spectrum analyzer, an analysis was done on the output of the natural log circuit. To determine the frequency components of the baseline noise, a spectrum analyzer recording was made with the plethysmograph off the finger and covered. The results are shown in Figure 24. The prominent noise is near DC, at 60 Hz, and at its harmonics. (Harmonics are not shown in figure.)

Next the plethysmograph was placed on the fingertip. The subject's hand was covered to avoid interference from external light. Figure 25 shows representative results. The frequency spectrum of the physiologic signal is from DC to about 15 to 20 Hz [8]. Based on this, the cutoff for the low pass filter was set to be approximately 15 Hz. The peaks in the 0 to 25 Hz recording appear to be decaying harmonics of the first peak. The first peak is at 0.75 to 1 Hz, which corresponds to the heart rate of the subject.

As previously mentioned, the input signal must be AC coupled into the gain stage(s) to facilitate high AC gain of several thousand. Initially, the corner frequency for this high pass filter was 0.15 Hz. Figure 26a shows the low frequency wandering that still appeared on the system output. Figures 26b and c show system output when this cutoff was moved to 0.7 and 5 Hz respectively. Either of these two

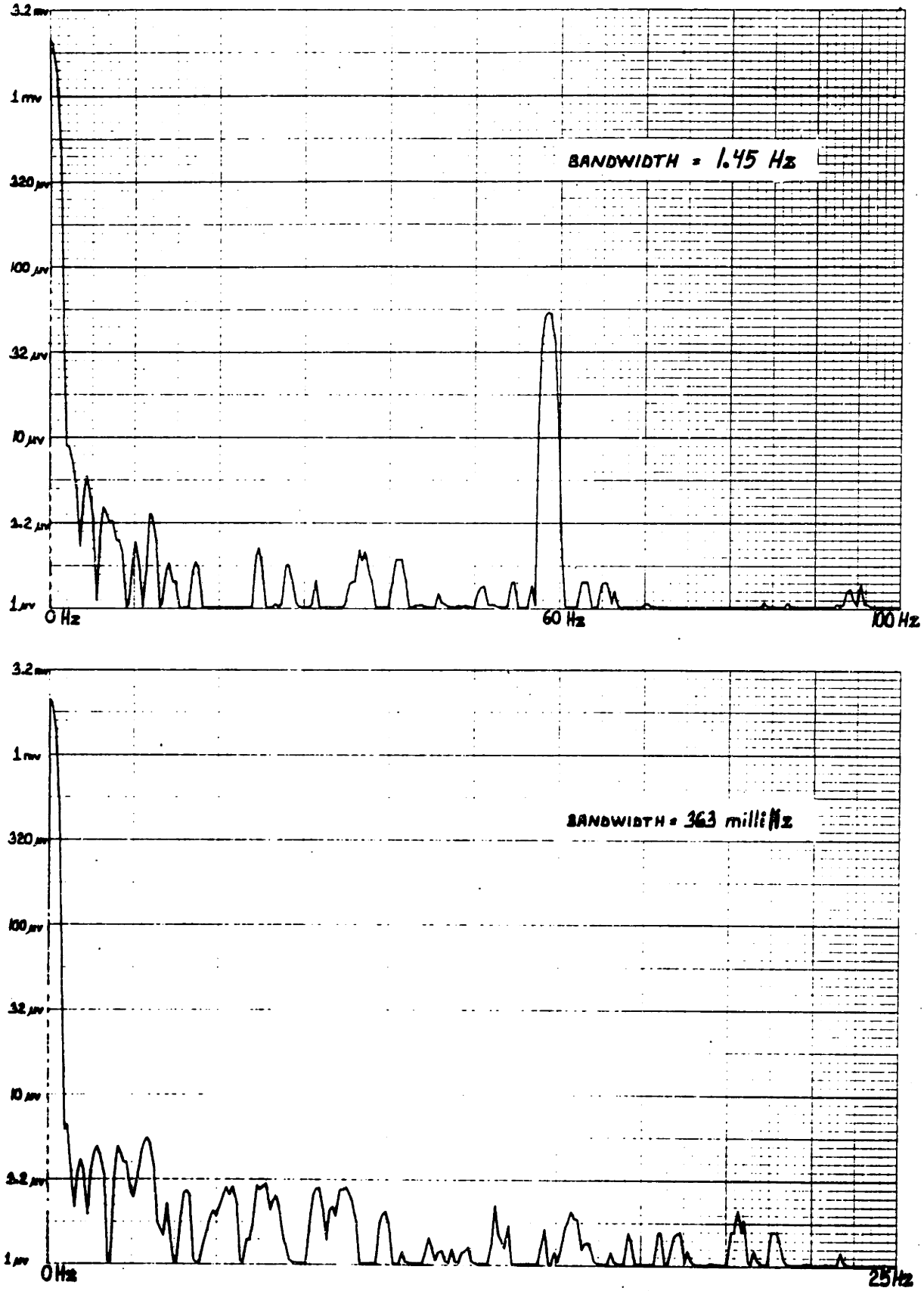
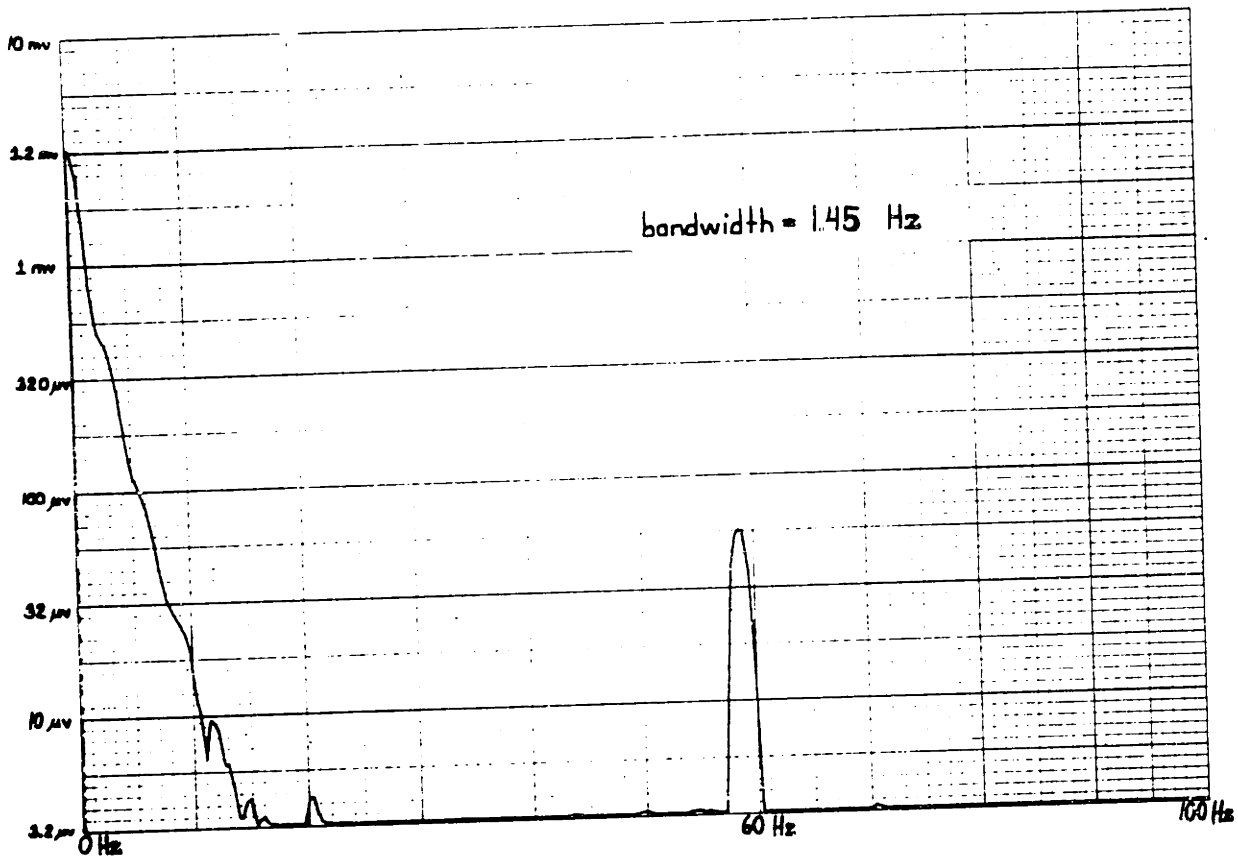

3.2 mv
1 mv
320 μv
100 μv
32 μv
10 μv
3.2 μv
1 μv
0 Hz
60 Hz
100 Hz
BANDWIDTH = 1.45 Hz
3.2 mv
1 mv
320 μv
100 μv
32 μv
10 μv
3.2 μv
1 μv
0 Hz
25 Hz
BANDWIDTH = 363 milliHz

FIGURE 24


10 mv
3.2 mv
1 mv
320 μv
100 μv
32 μv
10 μv
3.2 μv
0 Hz
60 Hz
100 Hz
bandwidth = 1.45 Hz

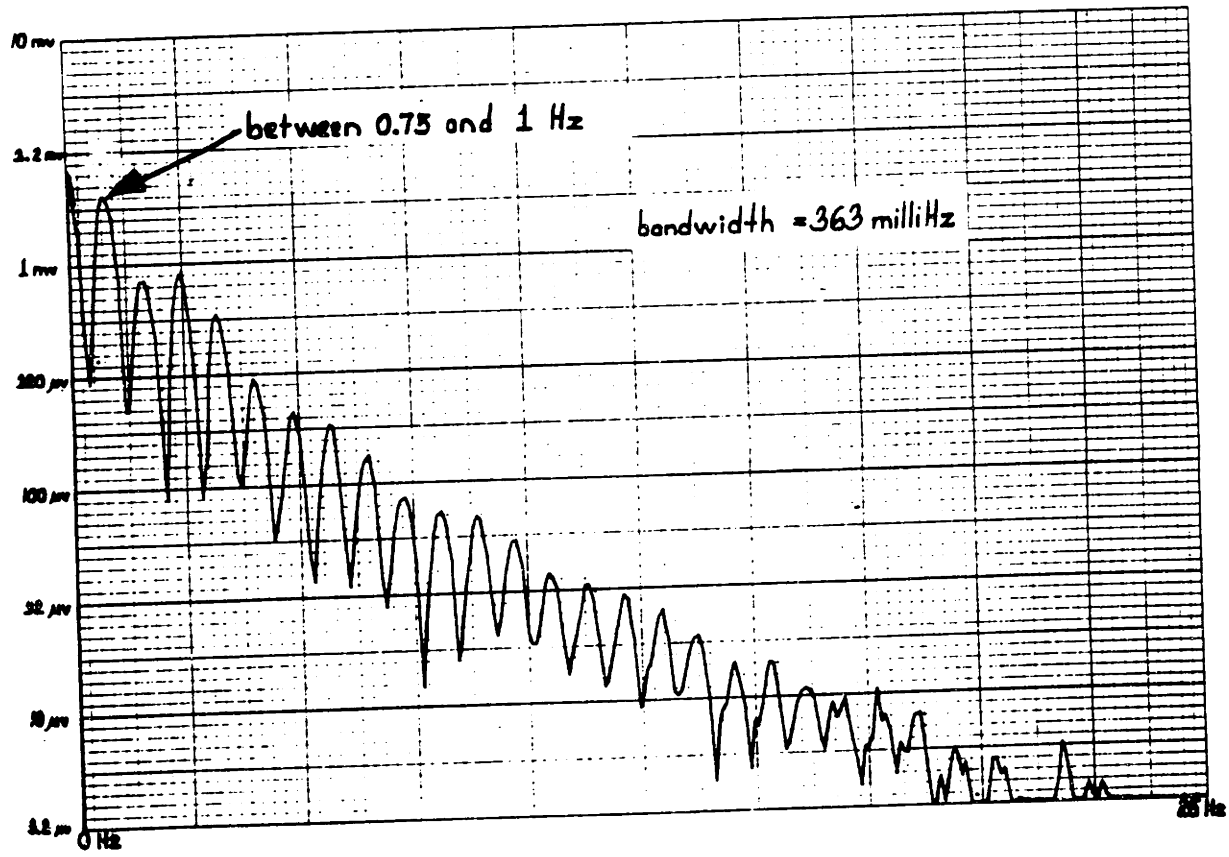
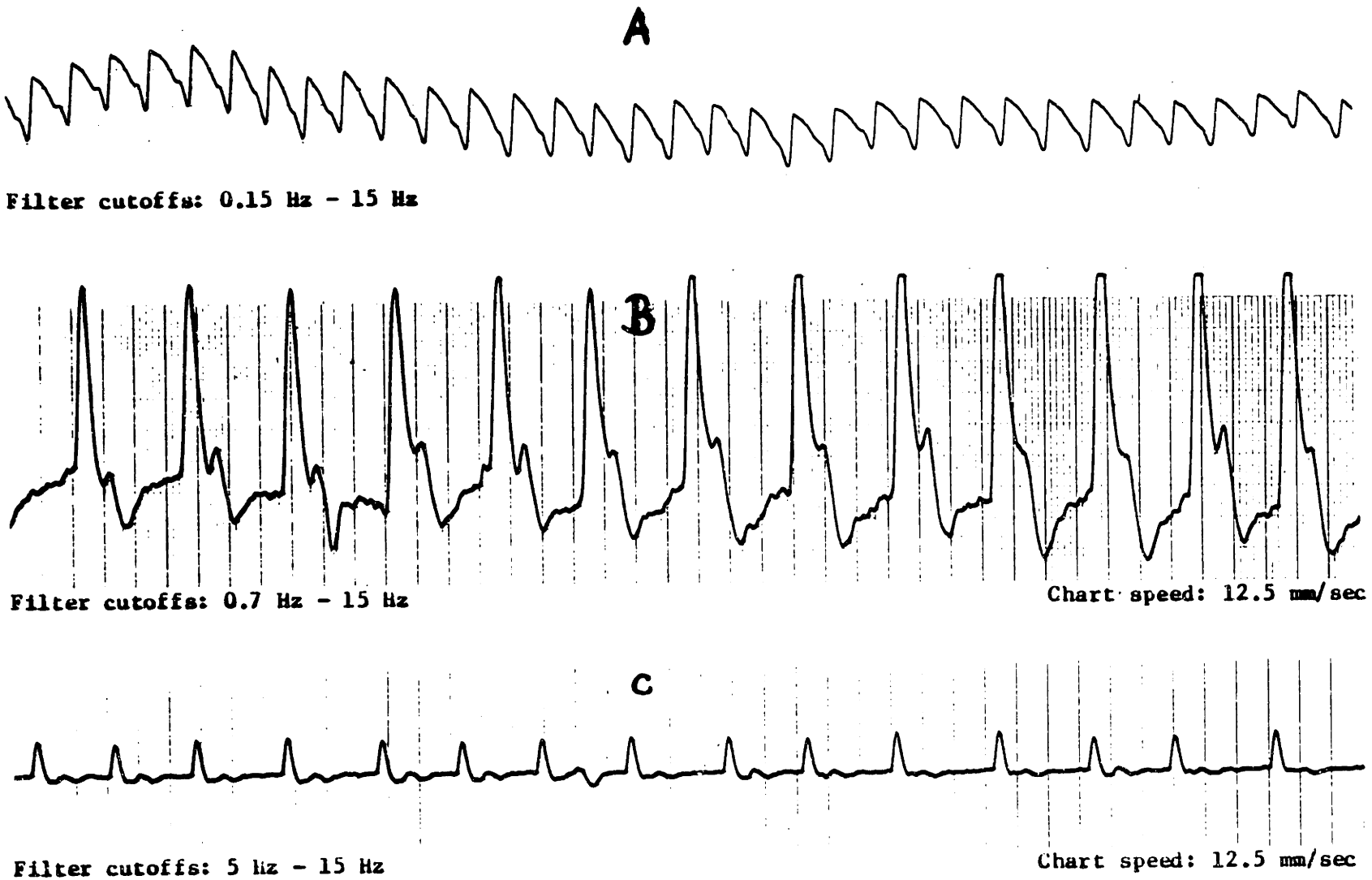

10 mv
3.2 mv
1 mv
320 μv
100 μv
32 μv
10 μv
3.2 μv
0 Hz
between 0.75 and 1 Hz
bandwidth = 363 milliHz

FIGURE 25



Note: These recordings were made at different times. Consequently, the relative amplitudes of the signals are insignificant.

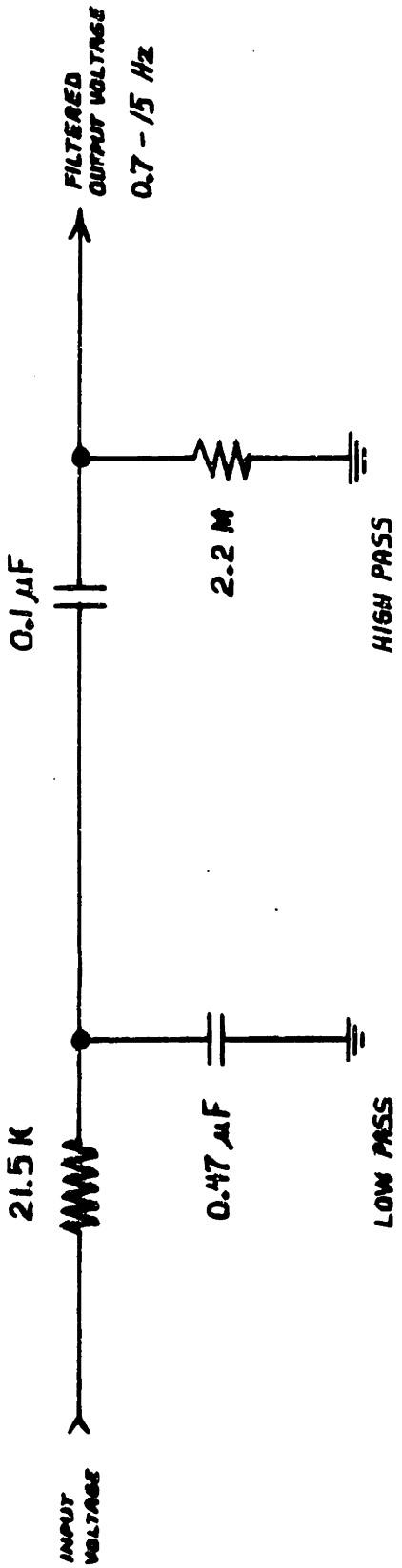
FIGURE 26

frequencies would be suitable. The 0.7 Hz filter attenuates the pulse amplitude less than the 5 hz filter. Therefore, the 0.7 Hz filter was used for systolic measurements, since maximum discernability of the first output pulse was desired. Diastolic blood pressure corresponds to the point where the circuit output amplitude levels off (Figure 3). For this determination, since it is important that the baseline remain very stable, the 5 Hz cutoff was used.

Figure 27 shows the circuitry implementation of the filters. Ideally, one would need a buffer in between the low and high pass for each to act independently. In practice, a buffer is not necessary. The high pass 2.2 M resistor was selected to be approximately one hundred times larger than the low pass 22 K. As a result, in the passband, (where the input voltage basically sees just the two resistors) almost all of the input voltage appears as output voltage across the 2.2 M. This holds for the 0.7 to 15 Hz cascade where the corner frequencies are greater than a factor of ten apart. Interaction is greater in the 5 to 15 Hz combination, and the passband has less than the ideal unity gain. However, the cascade still works well enough for my application.

Amplification

The choice of filter resistor values so as to eliminate the need for a buffer presents a new consideration. The filtered output voltage of the natural log circuit is the input to the gain stage(s), as shown in Figure 28. The op amp


INPUT VOLTAGE
21.5 K
0.47 µF
LOW PASS
0.1 µF
2.2 M
HIGH PASS
FILTERED OUTPUT VOLTAGE
0.7 - 15 Hz

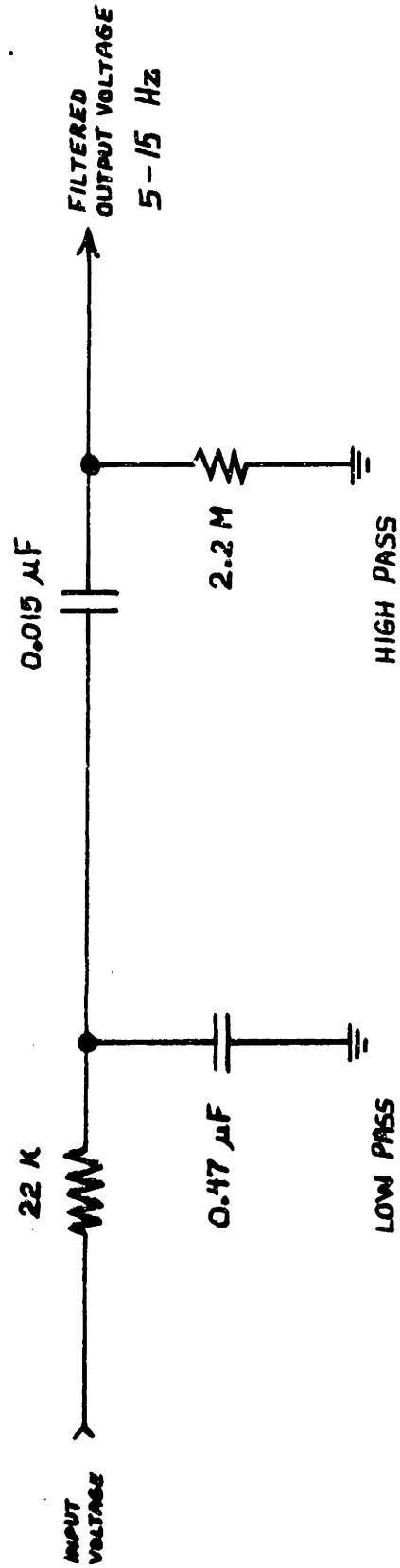

INPUT VOLTAGE
22 K
0.47 µF
LOW PASS
0.015 µF
2.2 M
HIGH PASS
FILTERED OUTPUT VOLTAGE
5 - 15 Hz

FIGURE 27

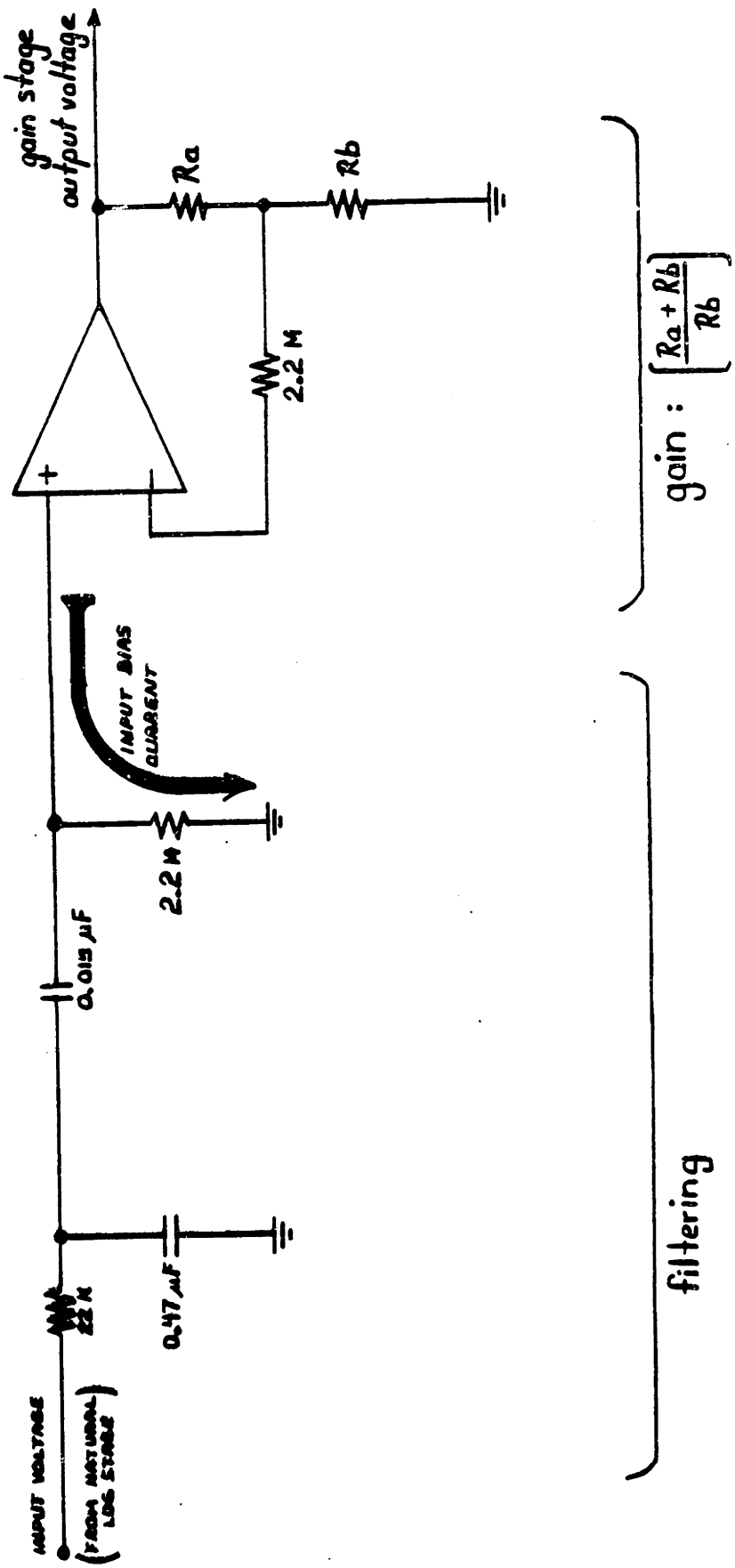

gain stage output voltage
Ra
Rb
2.2 M
+
INPUT BIAS CURRENT
2.2 M
0.015 µF
0.47 µF
22 K
INPUT VOLTAGE
(FROM NOTCHING LOG STAGE)
gain : [(Ra + Rb)/Rb]
filtering

FIGURE 28

input bias current flows through the 2.2 M resistor. For a typical op amp (LM 301), this current can be up to 250 nanoamps, producing an offset voltage of up to 0.5 volts. An amplification stage gain of about 20 or more would pin the op amp output. Consequently, the op amps used in the gain stages are National LF 355's. Their input bias current of less than 50 picoamps produces less than 0.1 millivolts offset voltage. Even with extremely high stage gain this offset would have minimal effect on the gain stage output voltage.

For systolic blood pressure determination, the filtered output of the natural log circuit was amplified about 7000 times. This allowed one to discern the first circuitry output pulse. Further cuff pressure reduction below the subject's systolic level yielded a larger amplitude physiologic signal whose peaks pinned the gain stage op amp output against the supply voltage. Since determination of diastolic pressure relied on observing the complete waveform amplitude of the circuitry output (Figure 3), gain was dropped to about 700.

The amplifier configuration was shown in Figure 28. The 2.2 Mohm resistor on the negative op amp input reduces offset voltage error due to input bias currents. Gain is determined by Ra and Rb. One could design the systolic gain of 7000 into one stage. However, Ra would have to be very large and Rb very small. High resistance resistors have poor temperature stability and large value tolerances. A single stage high gain amplifier has smaller bandwidth than a cascade of a few lower gain stages. Consequently, the amplification section

was broken down into a cascade of a few lower gain stages. Both systolic and diastolic processing circuitry share the natural log circuit, a bandpass filter (0.7 to 15 Hz), and a gain stage of approximately 60. To eliminate any noise introduced by this amplification, this output is again bandpass filtered (0.7 to 15 Hz) and amplified by 110 to produce the systolic system signal. The first stage output is also bandpass filtered (5 to 15 Hz) and amplified by 12 to produce the diastolic system output.

There is nothing sacred about these gain values. They yield useable signals for my research. Increasing systolic gain further does not improve system performance and increases output noise.

The natural log circuit, filters, and amplifiers have now been presented and explained. The total circuit is shown in Figure 29.

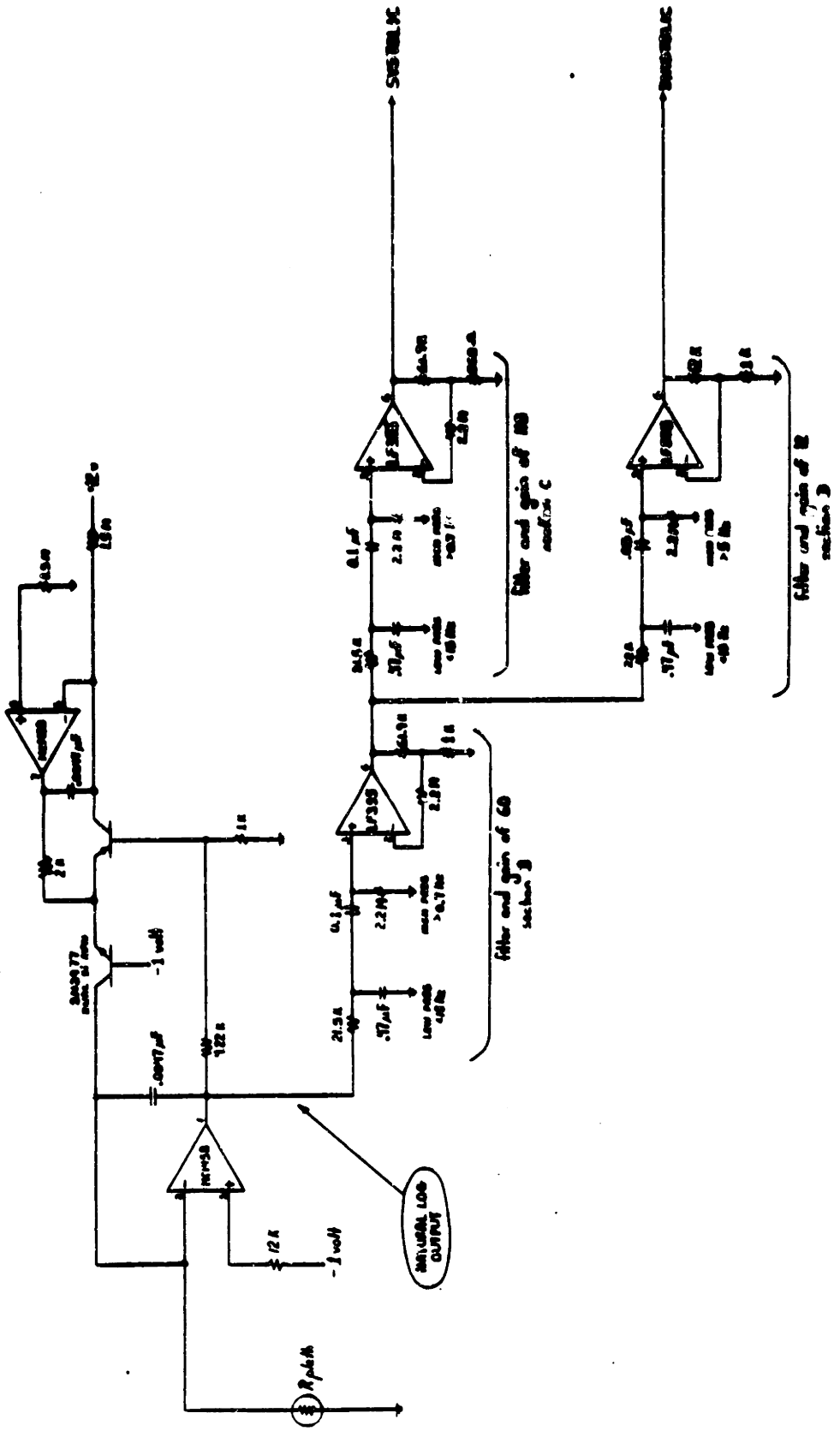


FIGURE 29

RESULTS FOR SYSTOLIC BLOOD PRESSURE

To evaluate the performance of the optical plethysmograph technique, a system was set up as shown in Figure 30. The reflective plethysmograph was positioned on the subject's fingertip, and the hand covered with black cloth to eliminate any interference from external light. The received physiologic signal was processed by the circuitry (Figure 29). Systolic circuitry output was displayed by both an oscilloscope and by one channel of a dual channel recorder. The second channel simultaneously displayed occlusion cuff pressure. Usually the recorder was calibrated to display pressure from 0 to 140 mm Hg full scale. For increased accuracy and readability when measuring systolic blood pressures, the pressure "expander" (explained in Appendix 2) allowed the recorder to display 80 to 160 mm Hg full scale.

Systolic values corresponded to the pressure at which the first circuitry output pulse occurred. (Figures 31a through e show some representative recordings.) While this measurement was being taken, an auscultatory reading was made using a Baum mercury manometer and a stethoscope. My results are compared with the auscultatory technique, because of its wide clinical acceptance as the standard of non-invasive blood pressure measurement.

77 trials were performed on 29 subjects, ranging in age from 19 to 64. Auscultatory systolic values ranged from 98 to 148 mm Hg. While most of the subjects were white males, women

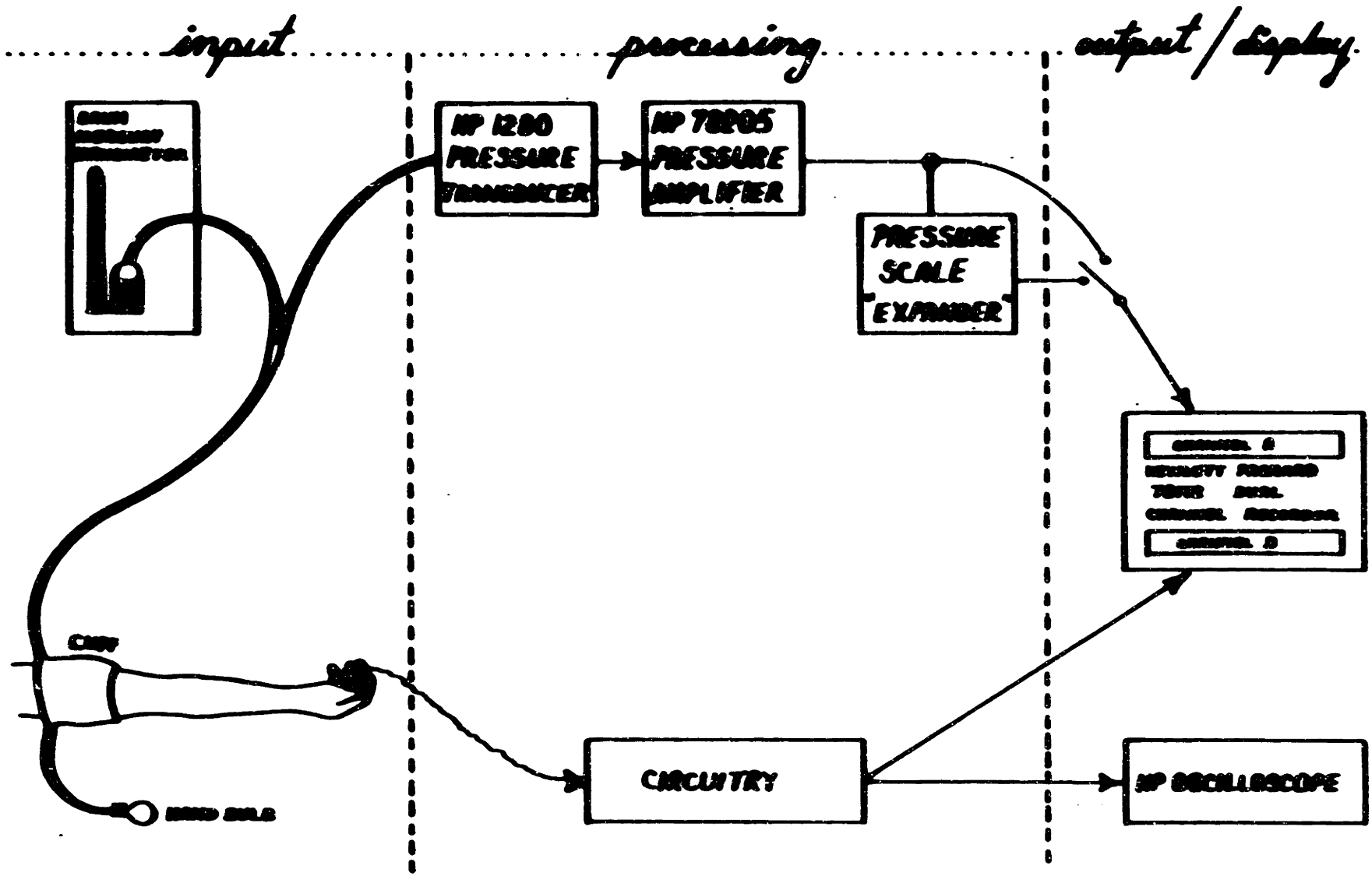
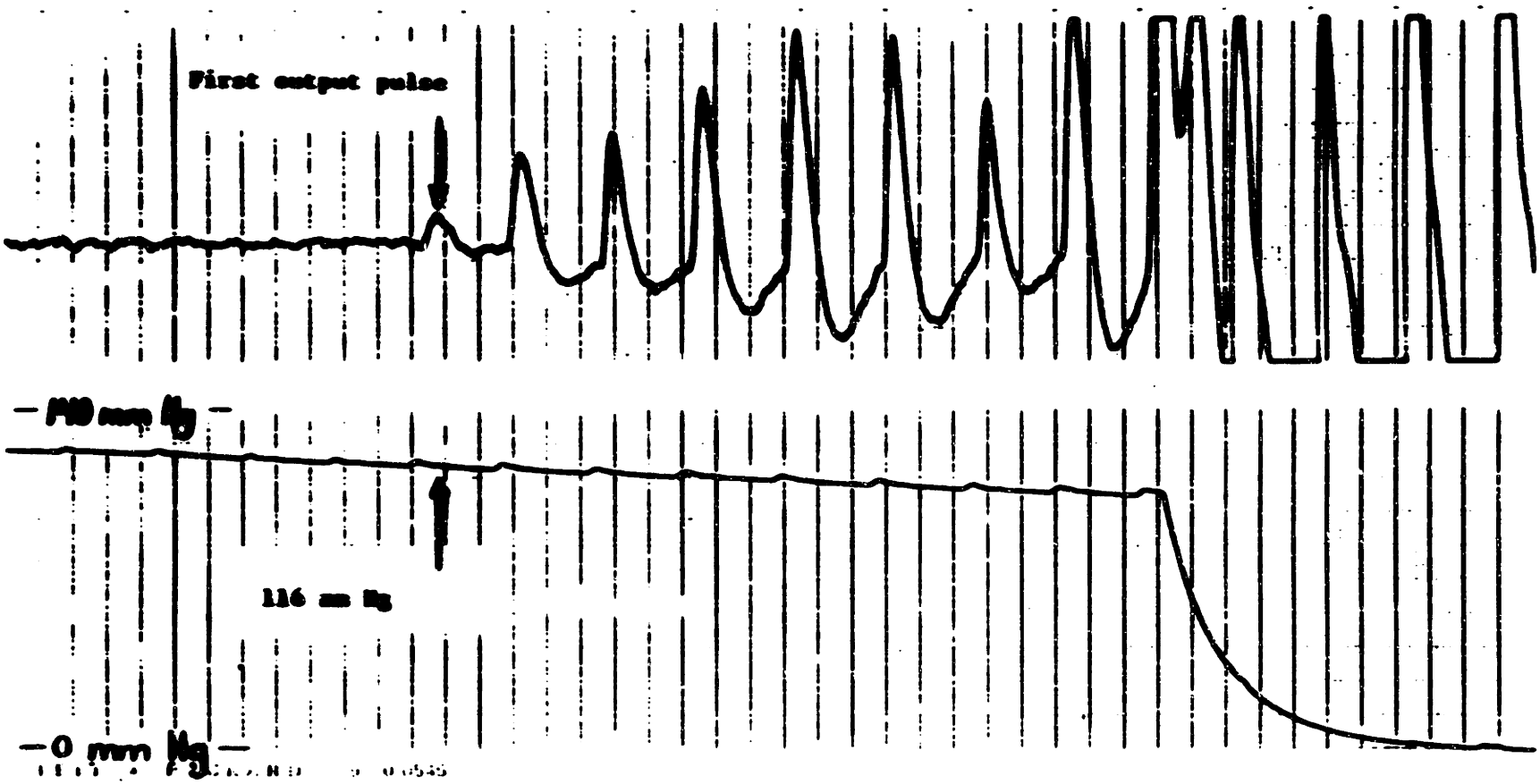

input
processing
output / display
HP 1280
PRESSURE
TRANSDUCER
HP 78205
PRESSURE
AMPLIFIER
PRESSURE
SCALE
EXPANDER
CUFF
CIRCUITRY
HP OSCILLOSCOPE

FIGURE 30



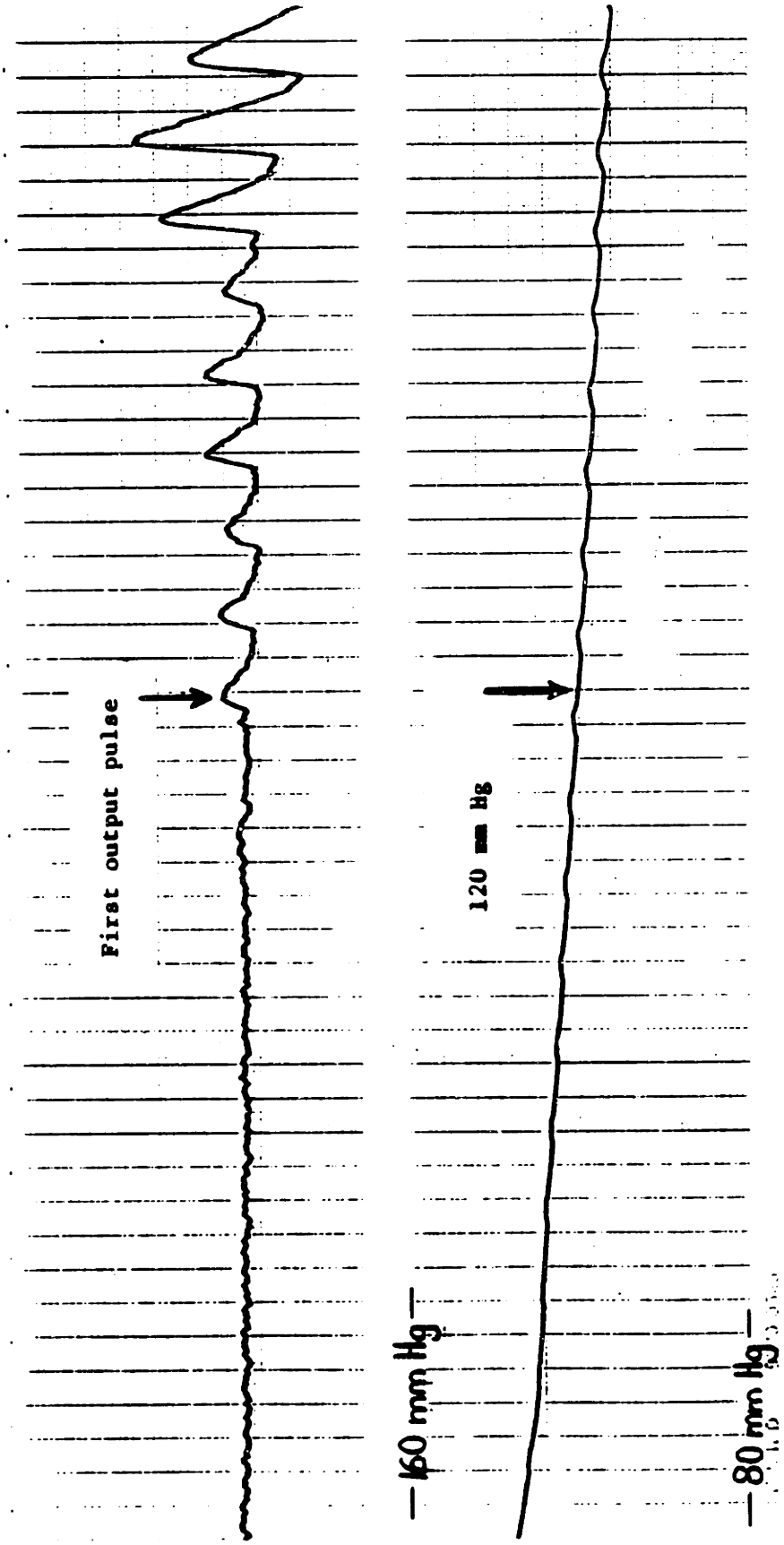
Subject: White male

Auscultatory result: 115 mm Hg

Error: 1 mm Hg Error = Auscultatory value - Photoplethysmography system result

Chart speed: 12.5 mm/sec

Figure 31a



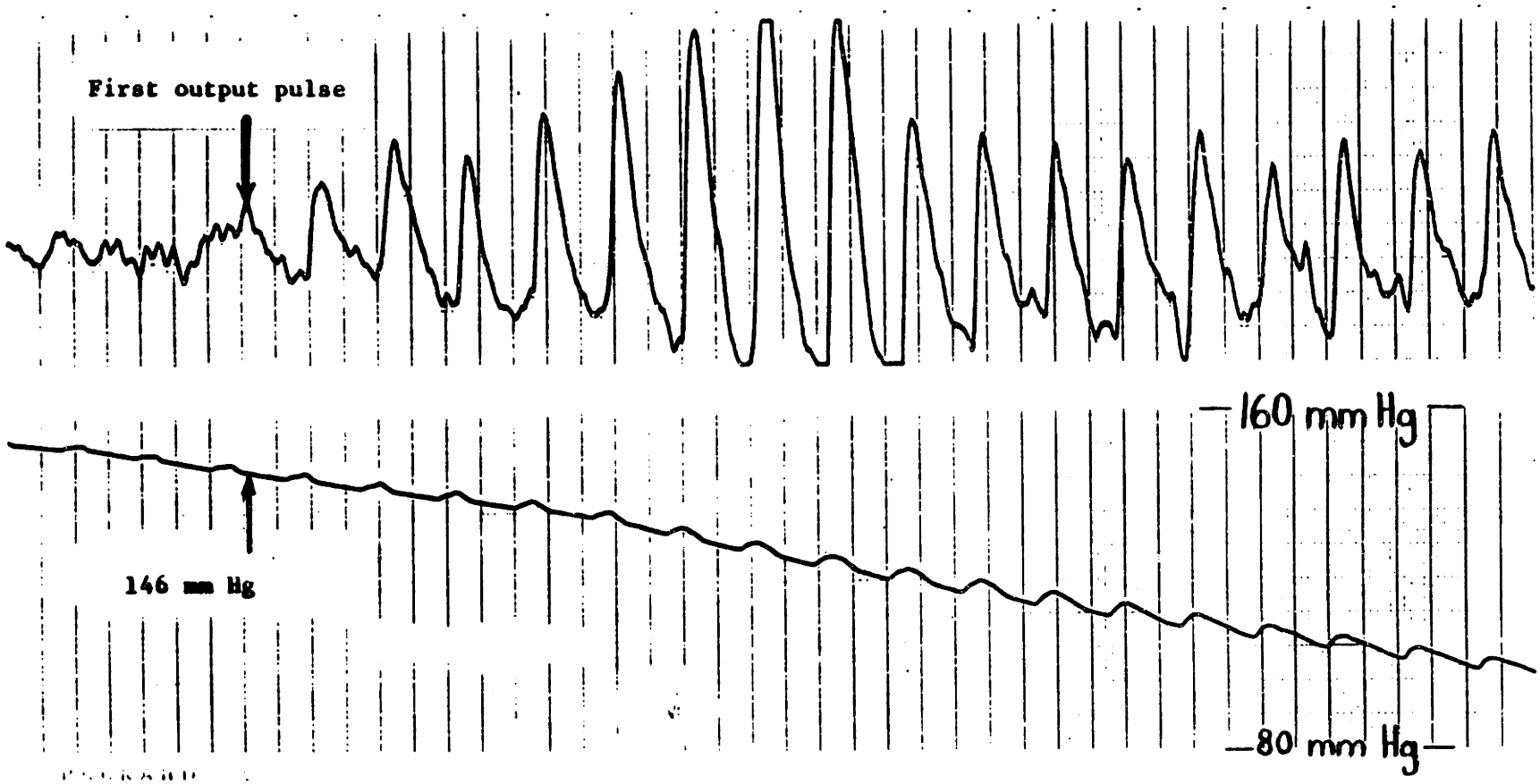
Subject: White female

Auscultatory result: 122 mm Hg

Error: 2 mm Hg

Chart speed: 12.5 mm/sec

Figure 31b



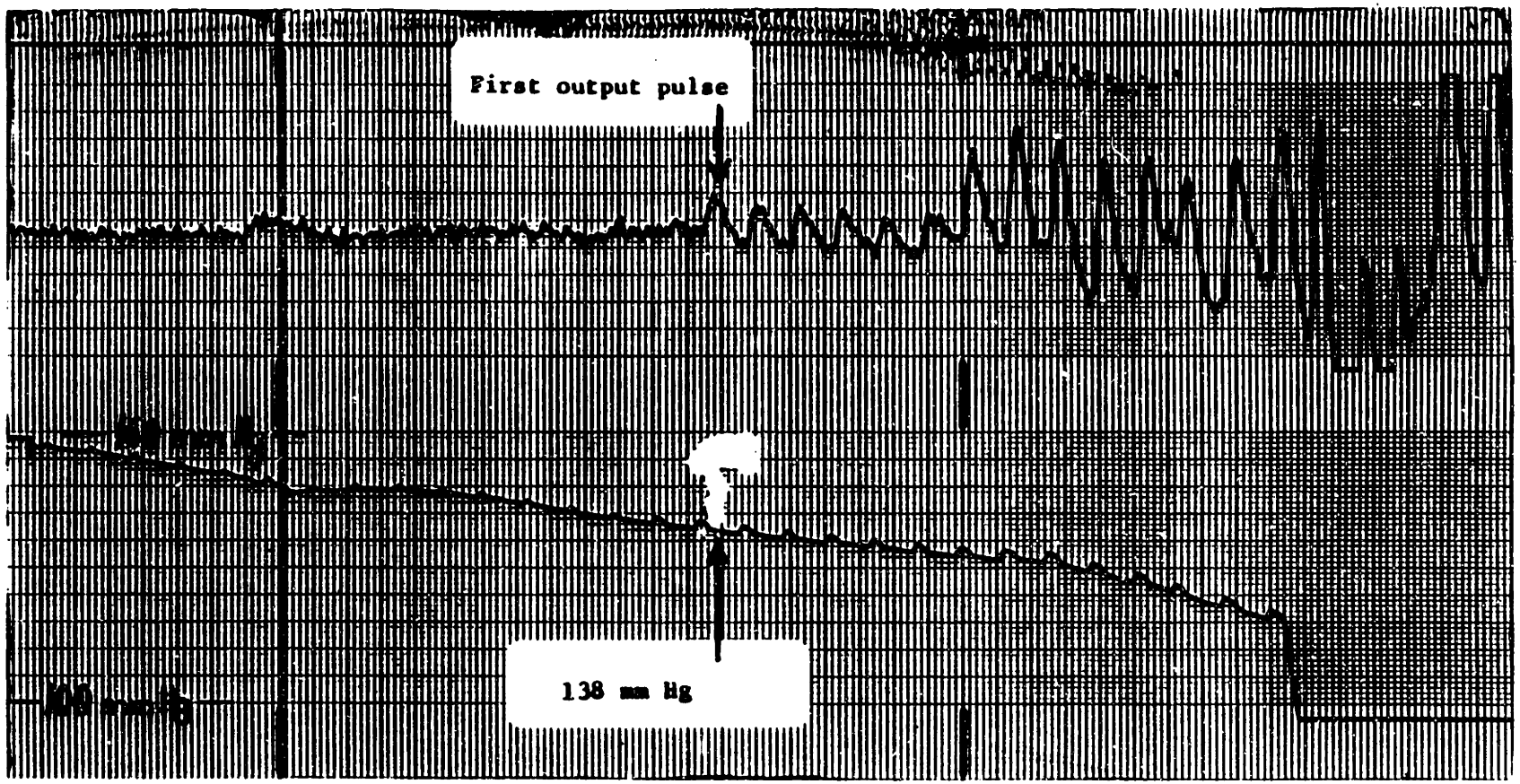
Subject: White male

Auscultatory result: 148 mm Hg, highest value recorded in experiments.

Error: 2 mm Hg

Chart speed: 12.5 mm/sec

Figure 31c



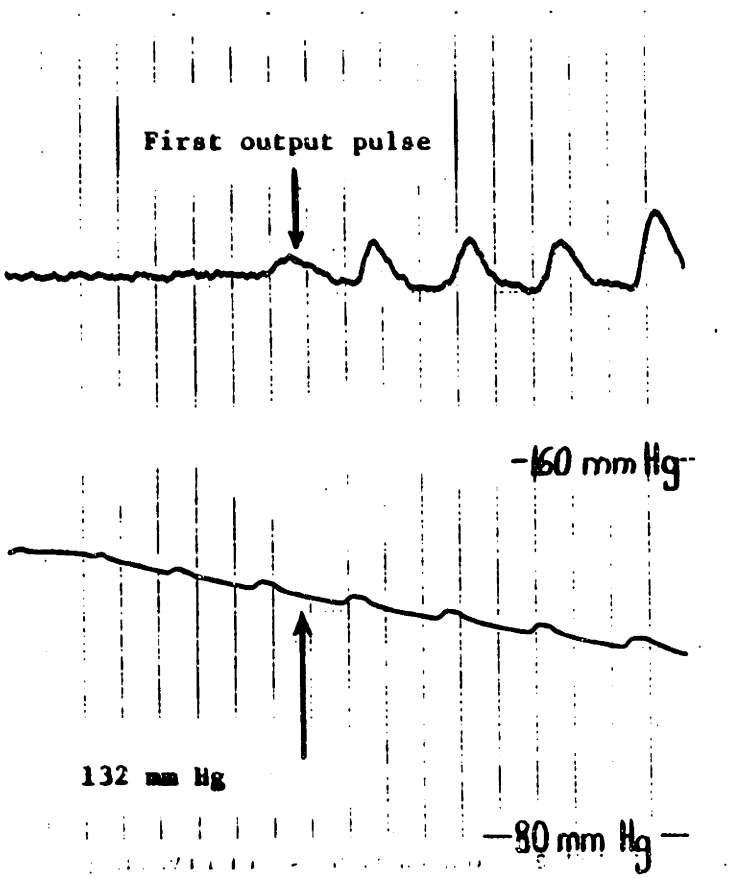
Subject: White male, 64 years old. Oldest subject tested.

Auscultatory result: 140 mm Hg

Error: 2 mm Hg

Recording made on 8 channel recorder, not on Hewlett Packard 78172

Figure 31d

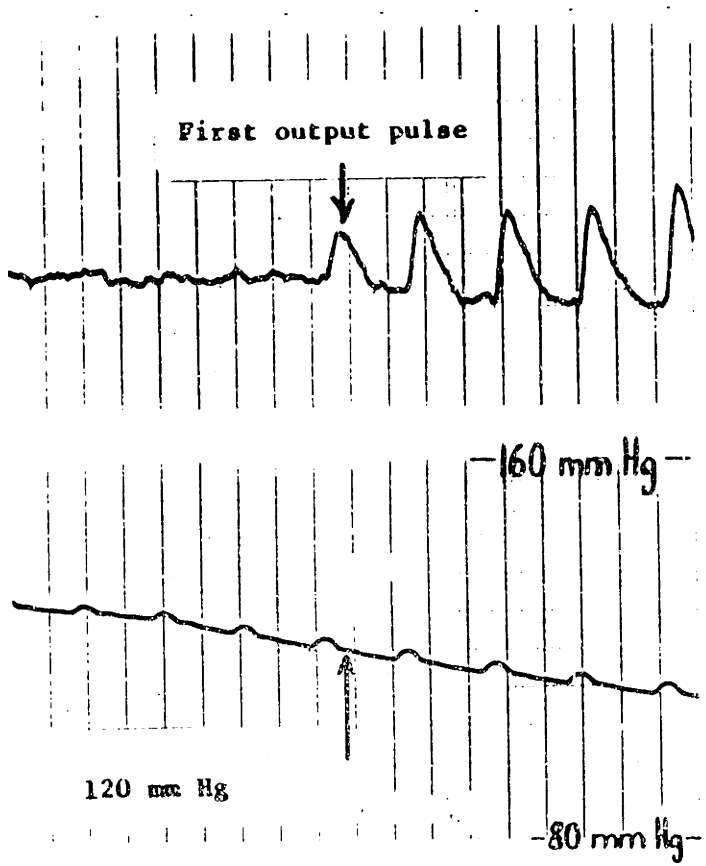


Subject: Black male

Auscultatory result: 134 mm Hg

Error: 2 mm Hg

Chart speed for both recordings: 12.5 mm/sec



Subject: Black male

Auscultatory result: 120 mm Hg

Error: none

Figure 31c

and blacks were also included. The data are listed in Figures 32a and b. Error was defined to be the auscultatory value minus the photoplethysmography system result. Figures 33 and 34 show the data spread around the ideal zero error point. The arithmetic mean for the error is 2.12 mm Hg. Maximum error was 15 mm Hg. Assuming a Gaussian distribution for the error, the standard deviation was 4.76 mm Hg. Mean absolute error was 3.88 mm Hg.

SYSTOLIC BLOOD PRESSURE RESULTS

	(a) Auscultatory Value	(b) System Result	Error (a-b) in mm Hg	Error in %	Notes
1.	126 mm Hg	122	4	3.17	A
2.	120	116	4	3.33	
3.	112	98	14	12.50	
4.	118	115	3	2.54	
5.	118	106	12	10.17	
6.	118	106	12	10.17	
7.	110	106	4	3.64	
8.	120	118	2	1.67	
9.	120	113	7	5.83	
10.	110	95	15	13.64	F
11.	108	106	2	1.85	F
12.	112	106	6	5.36	F
13.	112	104	8	7.14	F
14.	110	105	5	4.55	F
15.	114	118	-4	-3.51	F
16.	114	116	-2	-1.75	F
17.	122	123	-1	-0.82	M
18.	122	118	4	3.28	M
19.	132	132	0	0	M
20.	110	101	9	8.18	F,M
21.	104	104	0	0	F,M
22.	104	108	-4	-3.85	F,M
23.	122	120	2	1.64	M
24.	108	106	2	1.85	F,M
25.	132	132	0	0	F,M
26.	122	127	-5	-4.10	M
27.	122	122	0	0	M
28.	110	118	-8	-7.27	F,M
29.	112	112	0	0	
30.	128	129	-1	-0.78	
31.	118	123	-5	-4.24	F
32.	114	120	-6	-5.26	F
33.	118	112	6	5.08	
34.	118	116	2	1.69	
35.	116	112	4	3.45	
36.	116	116	0	0	
37.	124	123	1	0.81	
38.	123	120	3	2.44	
39.	122	126	-4	-3.28	
40.	120	120	0	0	
41.	119	122	-3	-2.52	F
42.	122	115	7	5.74	F

..... data continued in Figure 30b .....

Notes: A = Auscultatory reading taken by HP Clinic nurse.
M = Auscultatory reading taken by former ICU manager.
F = Subject was female.

Figure 32a

SYSTOLIC BLOOD PRESSURE RESULTS - continued -

	(a) Auscultatory Value	(b) System Result	Error (a-b) in mm Hg	Error in %	Notes
43.	120 mm Hg	120	0	0	
44.	120	122	-2	-1.67	
45.	118	119	-1	-0.85	
46.	115	116	-1	-0.87	
47.	116	118	-2	-1.72	
48.	144	148	-4	-2.78	
49.	140	138	2	1.43	
50.	111	104	7	6.31	F
51.	114	112	2	1.75	
52.	134	132	2	1.49	B
53.	130	122	8	6.15	B
54.	130	120	10	7.69	B
55.	126	121	5	3.97	
56.	120	120	0	0	
57.	120	109	11	9.17	
58.	120	120	0	0	B
59.	122	120	2	1.64	F
60.	122	118	4	3.28	F
61.	148	146	2	1.35	C
62.	138	136	2	1.45	
63.	126	125	1	0.79	
64.	142	138	4	2.82	
65.	126	122	4	3.17	B
66.	122	120	2	1.64	B
67.	124	117	7	5.65	
68.	123	119	4	3.25	
69.	102	101	1	0.98	F
70.	104	104	0	0	F
71.	104	99	5	4.81	F
72.	98	100	-2	-2.04	F,D
73.	105	98	7	6.67	F
74.	102	101	1	0.98	F
75.	106	110	-4	-3.77	F
76.	101	109	-8	-7.92	F
77.	108	109	-1	-0.93	F

Additional notes

B = Subject was black
C = Highest auscultatory value recorded in experiments.
D = Lowest auscultatory value recorded in experiments.

Figure 32b

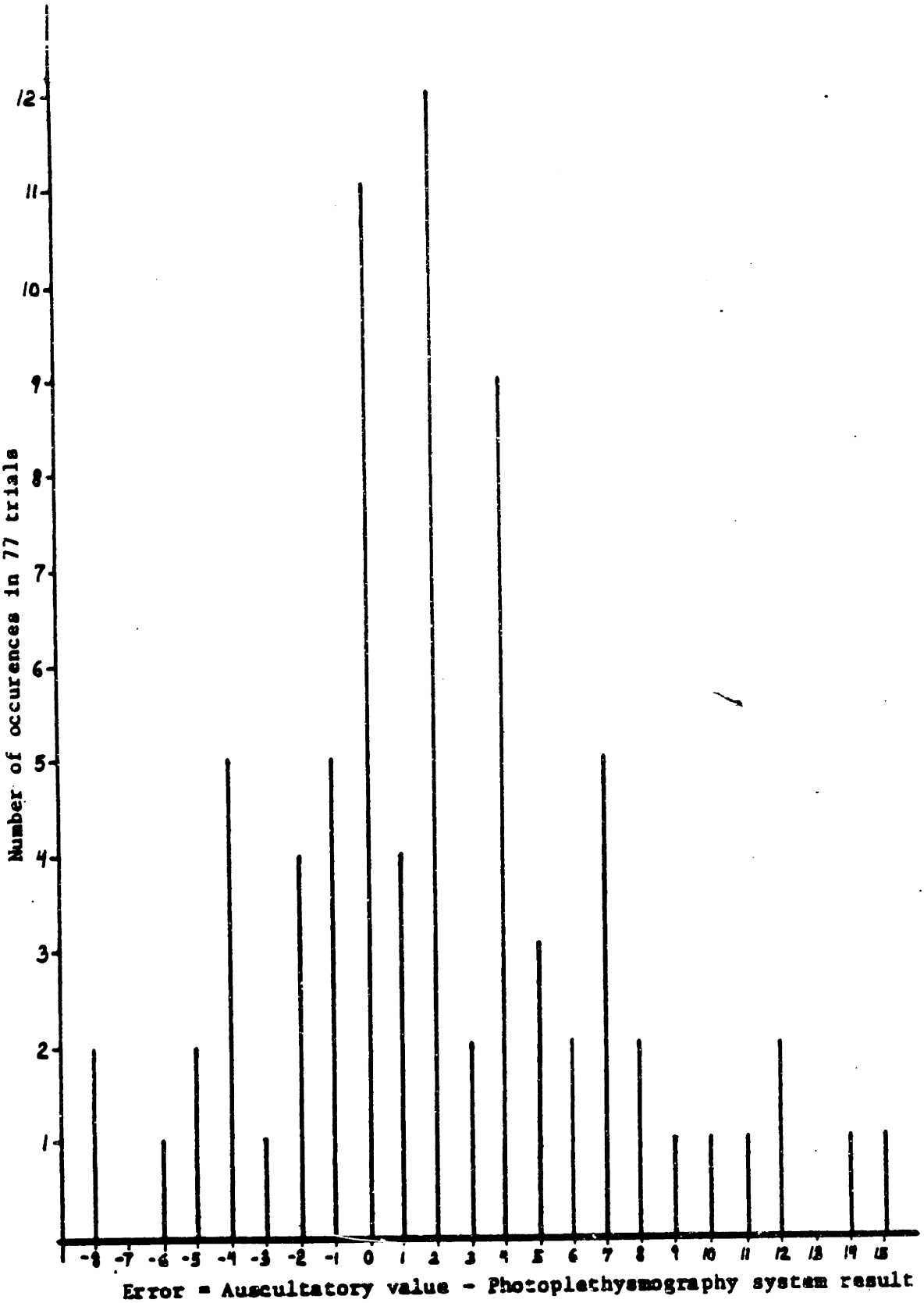

Number of occurences in 77 trials
12
11
10
9
8
7
6
5
4
3
2
1
-8 -7 -6 -5 -4 -3 -2 -1 0 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15
Error = Auscultatory value - Photoplethysmography system result

FIGURE 33

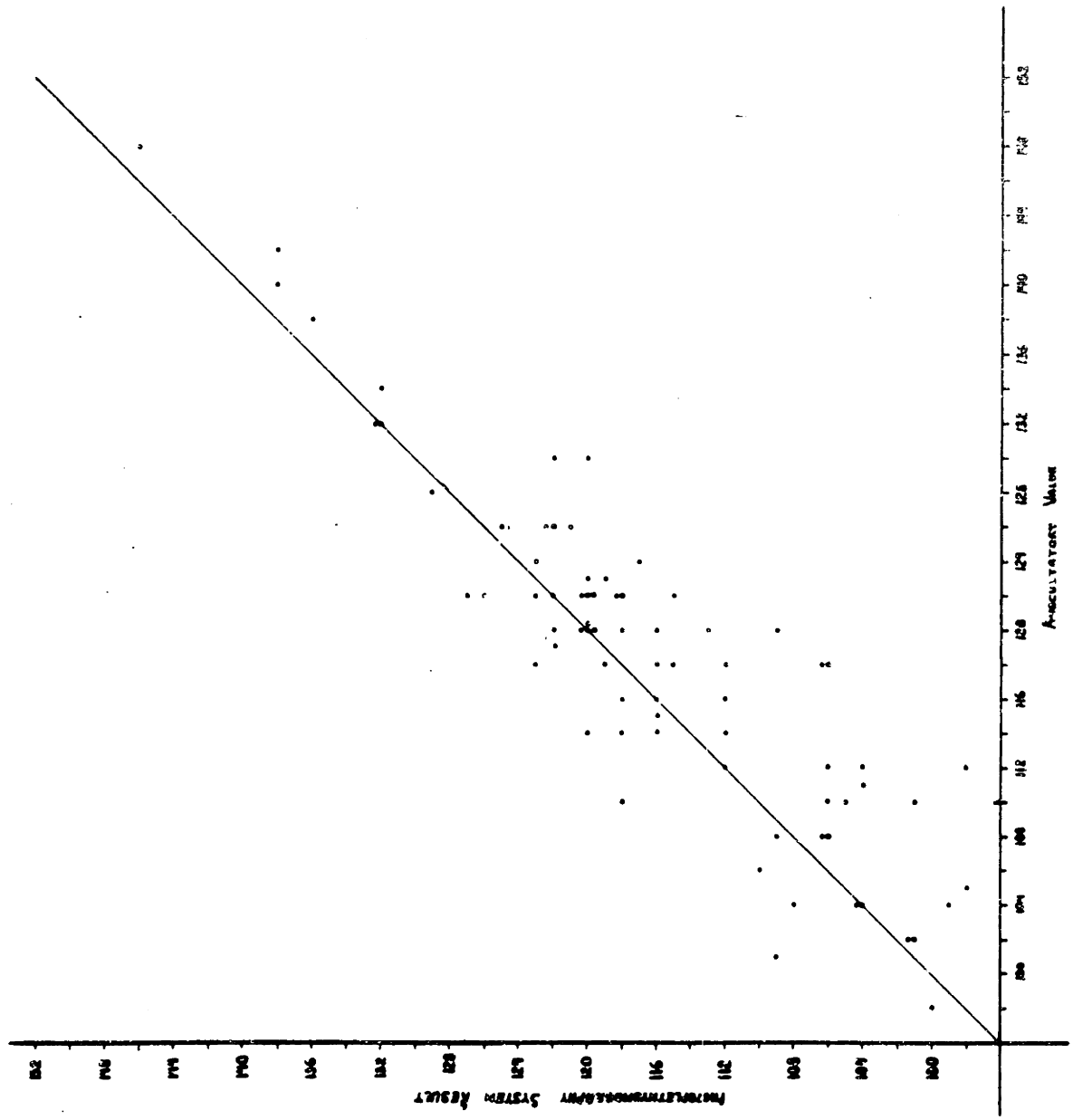


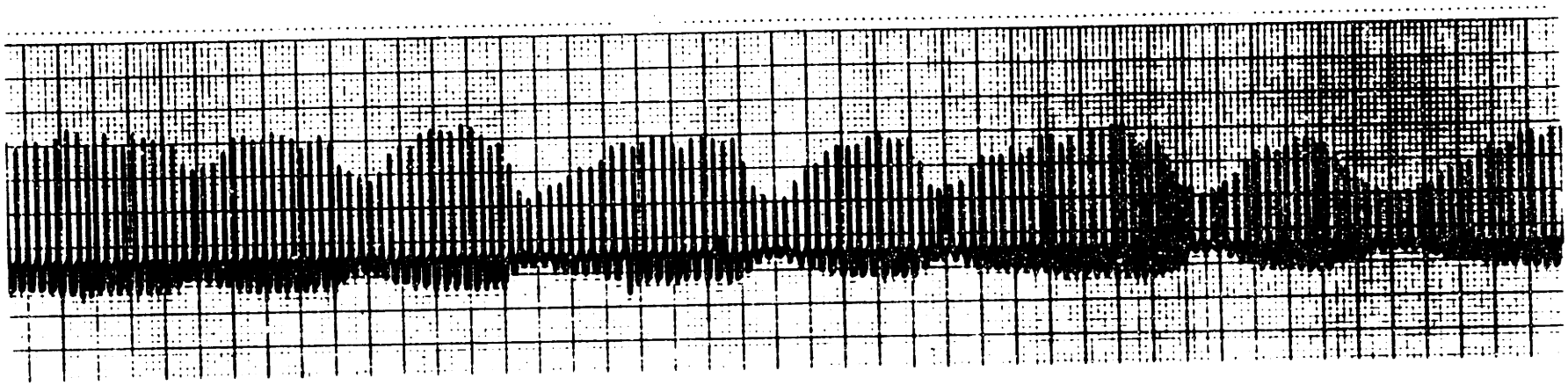
FIGURE 34

RESULTS FOR DIASTOLIC BLOOD PRESSURE

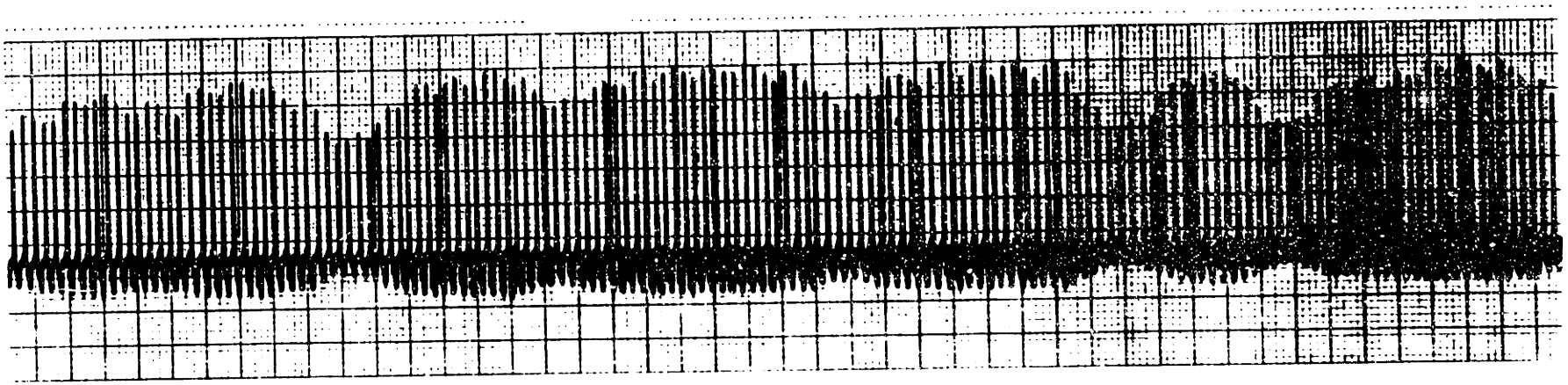
The Approach and the Problem

With the plethysmograph placed on the fingertip, Figure 3 depicts the approach to measuring diastolic blood pressure. The idea is based on the assumption that the amplitude of the pulsatile blood flow to the fingertips levels off when the brachial artery is fully opened. Physiologists have found that internal regulatory systems do vary the amount of blood that reaches different parts of the body [30]. However, I assumed that for a subject at rest, these fluctuations would be slow compared to the time required for a blood pressure measurement, and that the system output would indeed show a levelling to a constant amplitude.

This assumption was not correct. Figure 35 shows recordings made from subjects at rest without any occlusion cuff. I expected to see constant amplitude outputs. Instead, note both the large degree of variance in the pulsatile blood flow to the fingertip, and the small time intervals over which these changes take place (even within 15 seconds). These variations appeared in every recording made. The waveshapes and amplitudes varied inconsistently both as a function of time for a given individual, and from subject to subject. These unexpected variations rendered the initial diastolic approach ineffective, since the constant level amplitude which I had expected to reach did not exist. A check of related literature revealed similar observations by



Trial 1: Chart speed is 1.25 mm/sec



Trial 2: Chart speed is 1.25 mm/sec

Figure 35

others [38,40].

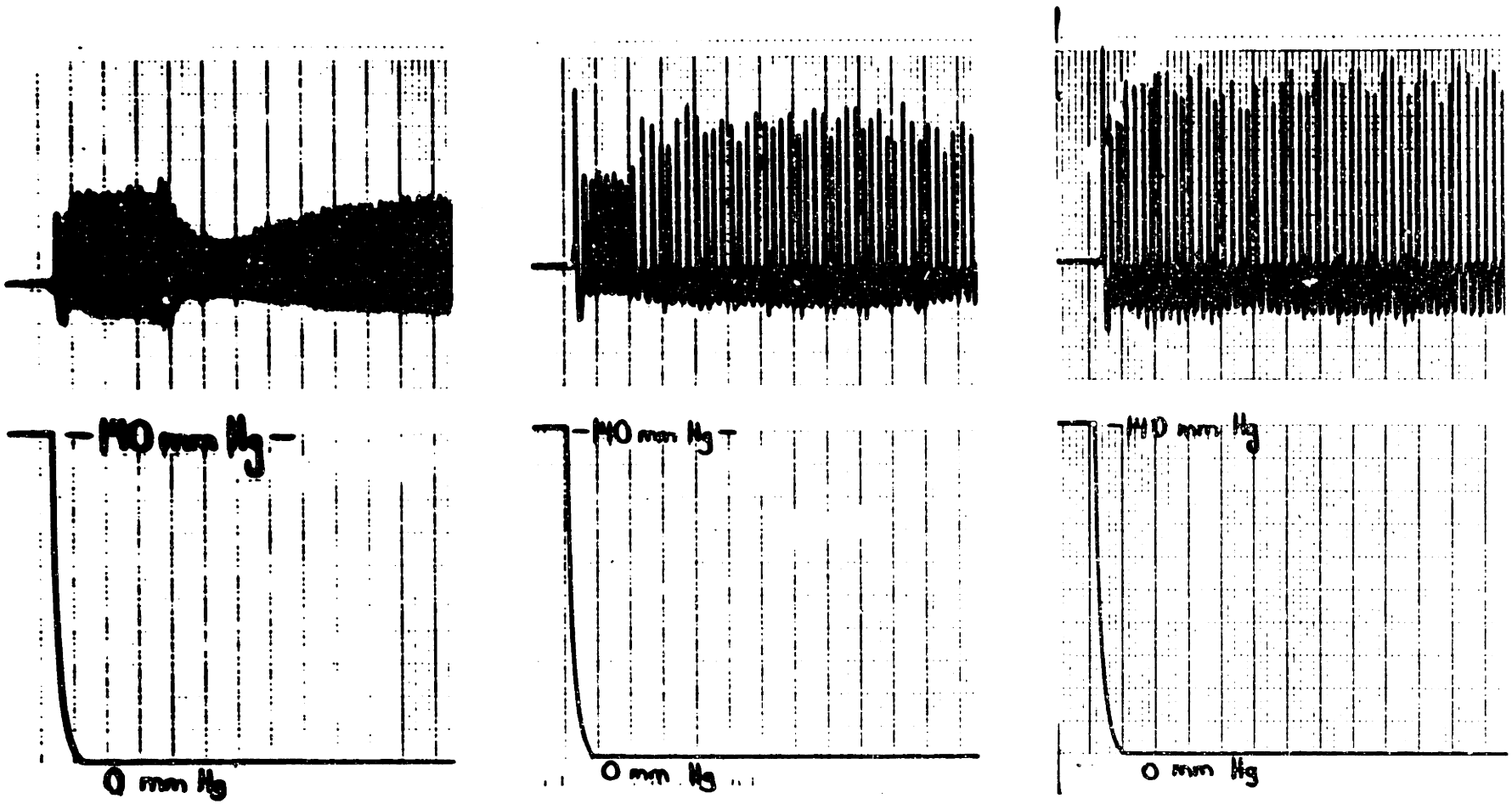
The Spectrum Analysis

I tried to circumvent the fluctuation problem by seeking another way to detect diastolic blood pressure. In the hope that either a lone frequency component or a unique spectral shape accompanied the diastolic point, a spectrum analysis was done on the output of the natural log circuitry output prior to the filtering stage (HP 3582A spectrum analyzer) as occlusion cuff pressure was dropped from the subject's systolic to below diastolic pressures. Neither condition was observed.

The Step Response

It was thought that knowledge of the physiologic step response to blood flow might add insight to the functioning of the circulatory system and to the determination of diastolic blood pressure in particular.

The occlusion cuff was pumped above the subject's systolic level, occluding all blood flow to the fingertips. Cuff pressure was then quickly dropped and the resulting system output recorded. Figure 36 shows some representative recordings. In Figure 36a, the initial signal peak is due to blood surging into the fingertips. Following is a period of increased perfusion during which the regulatory mechanisms dilate the capillaries, maximizing oxygen delivery to the



A: Trial 1
Chart speed: 1.25 mm/sec

B: Trial 2
Chart speed: 1.25 mm/sec

C: Trial 3
Chart speed: 1.25 mm/sec

Figure 36

previously oxygen deprived extremity. Unfortunately, this behavior is not common to all the recordings (Figures 36b,c). This inability to detect a generalized step response indicated that this was not a good approach to accomplish my goals.

Local Versus Systemic Control: Introduction

The failure of the previous two approaches leads to the following conclusion: If the photoplethysmography technique is to work for diastolic blood pressure determination, one must either deal with, or eliminate, the waveform variations.

Blood circulation throughout the body is regulated by both local and systemic control mechanisms. For example, pre-capillary sphincters play an important role in the former, while variable ventricular filling is one of the latter's mechanisms. To deal with the waveform variations, one must determine whether they are local or systemic in origin.

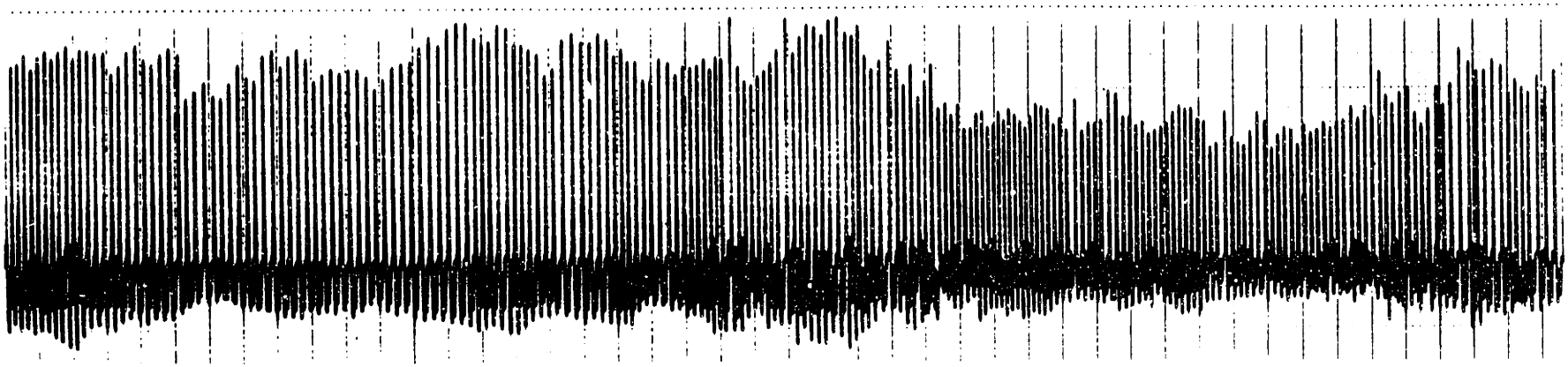
Local Versus Systemic Control: The Transmission Approach

Initially, the fluctuations were attributed to local fingertip capillary control mechanisms. It was thought that the variations were due to some number of capillaries within the plethysmograph receptive field constricting and dilating in response to local stimuli. Based on this, including more capillaries within the pick-up's receptive field might

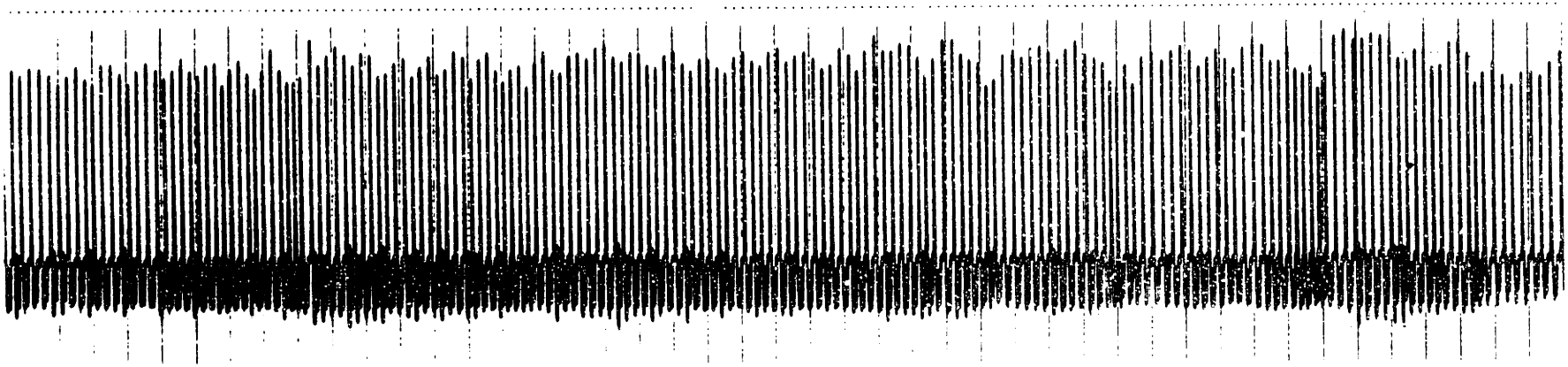
decrease the overall fluctuations, due to cancellation and averaging of these individual opening/closing effects. The simplest way to include more capillaries in the receptive field is to switch from the reflective to the transmission plethysmograph, where the received light must have passed through all the tissue in the fingertip. Figure 37 shows results with the transmission pleth placed over the fingertip and without any occlusion cuff. It appeared that the degree of amplitude fluctuation had decreased, though variations were still evident. (One must be careful in evaluating these recordings, because of the high degree of physiologic variation that occurs in a subject over even short periods of time). This attempt had not cured the variation problem.

Local Versus Systemic Control: The "Light Box"

Increasing the receptive field was taken to the extreme in the "Light Box", shown in Figure 38 and Picture 4. The "Light Box" was designed to monitor both the light transmitted through, and reflected off, all five fingers. Because the light source and photocell were spaced several inches apart, with the entire hand in between, incandescent bulbs were used instead of LED's to provide an adequate signal to noise ratio. Recordings were made (without occlusion cuff) and representative results are shown in Figure 39. The fluctuations were not eliminated.



Trial 1: Chart speed = 1.25 mm/sec



Trial 2: Chart speed = 1.25 mm/sec

Figure 37

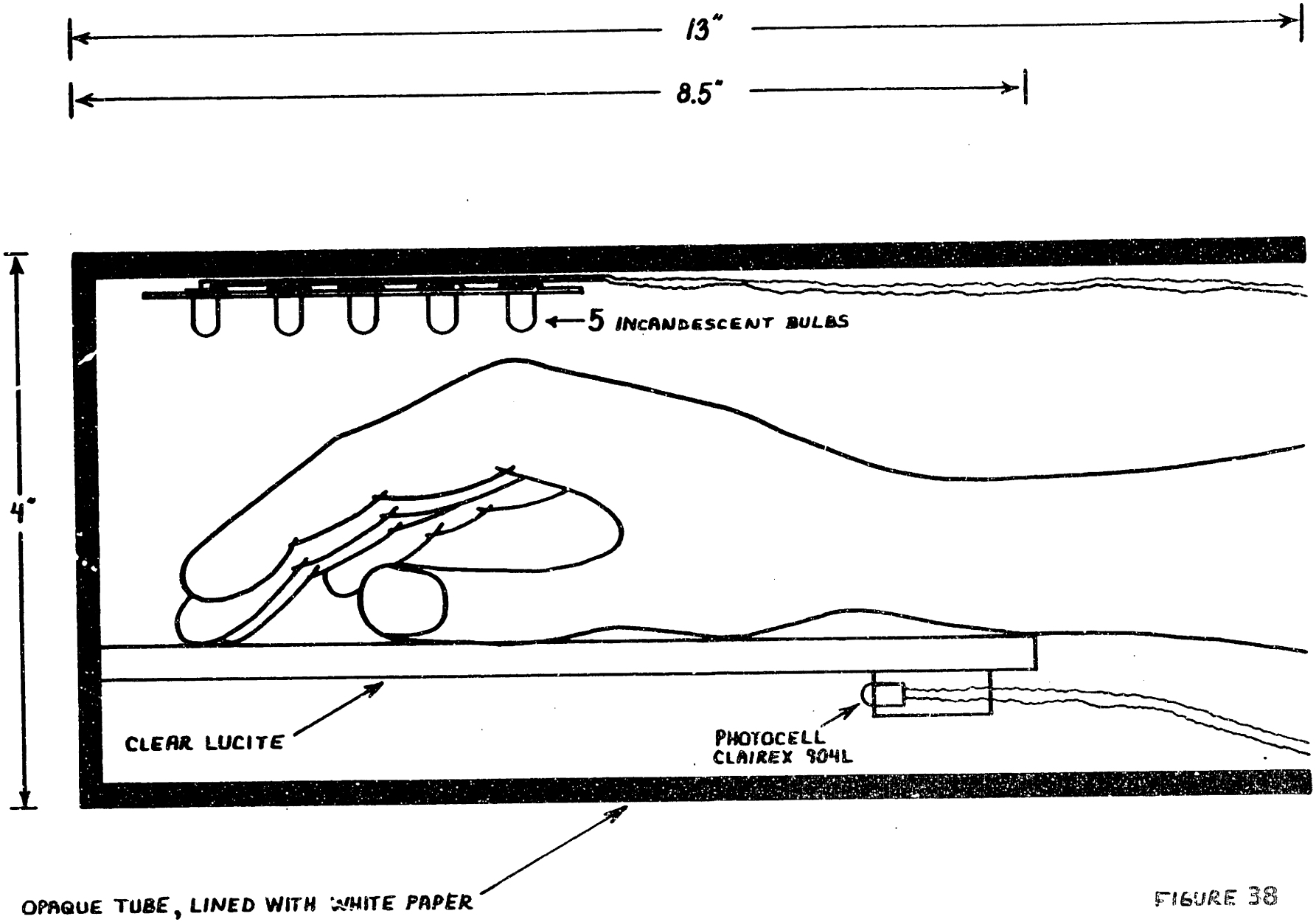
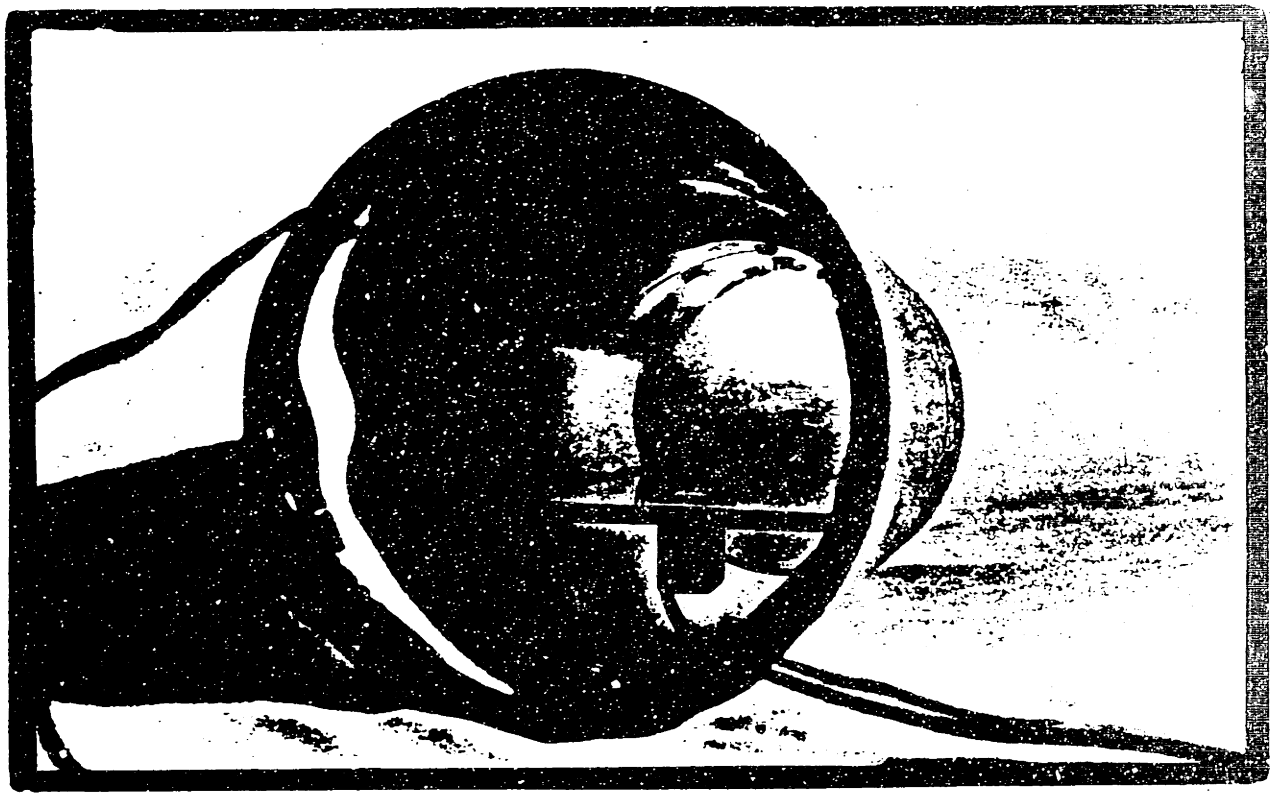
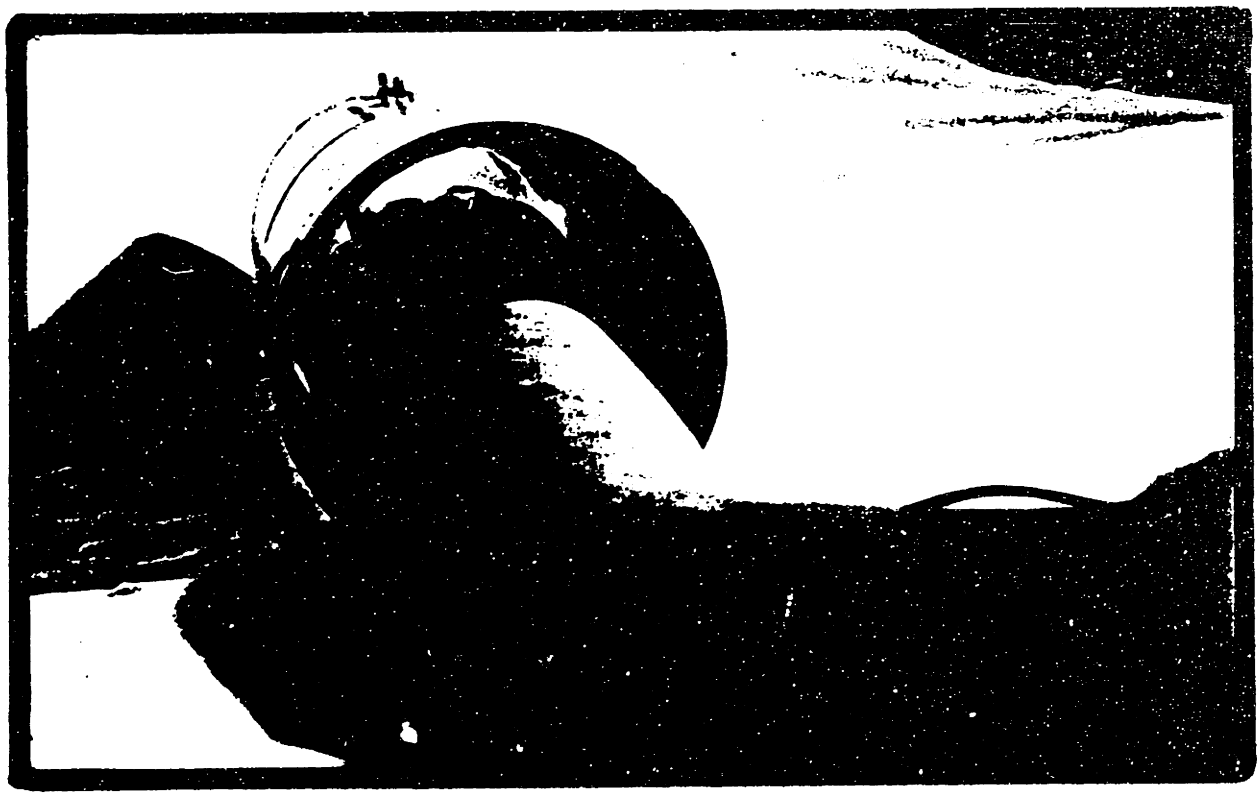
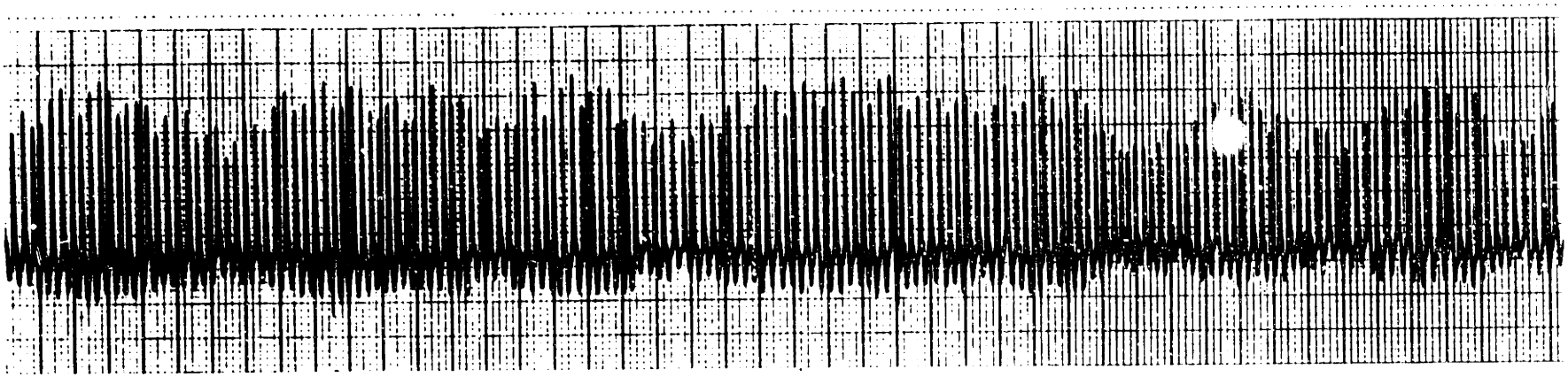

13"
8.5"
4"
5 INCANDESCENT BULBS
CLEAR LUCITE
PHOTOCELL
CLAIREX 904L
OPAQUE TUBE, LINED WITH WHITE PAPER

FIGURE 38

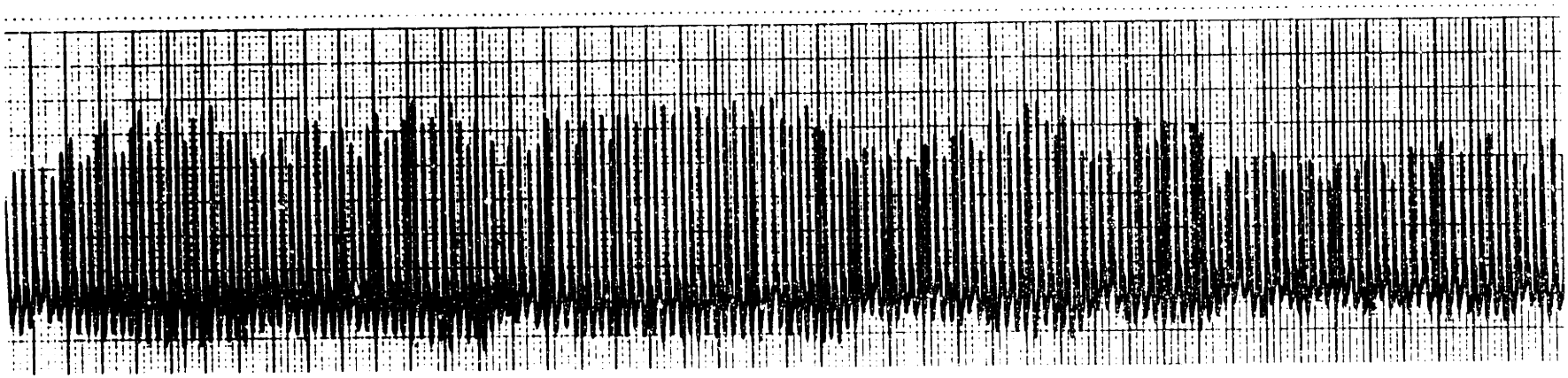




The "Light Box" and its placement over an arm. Picture 4



Trial 1: Chart speed: 1.25 mm/sec



Trial 2: Chart speed: 1.25 mm/sec

Figure 39

Local Versus Systemic Control: Heat

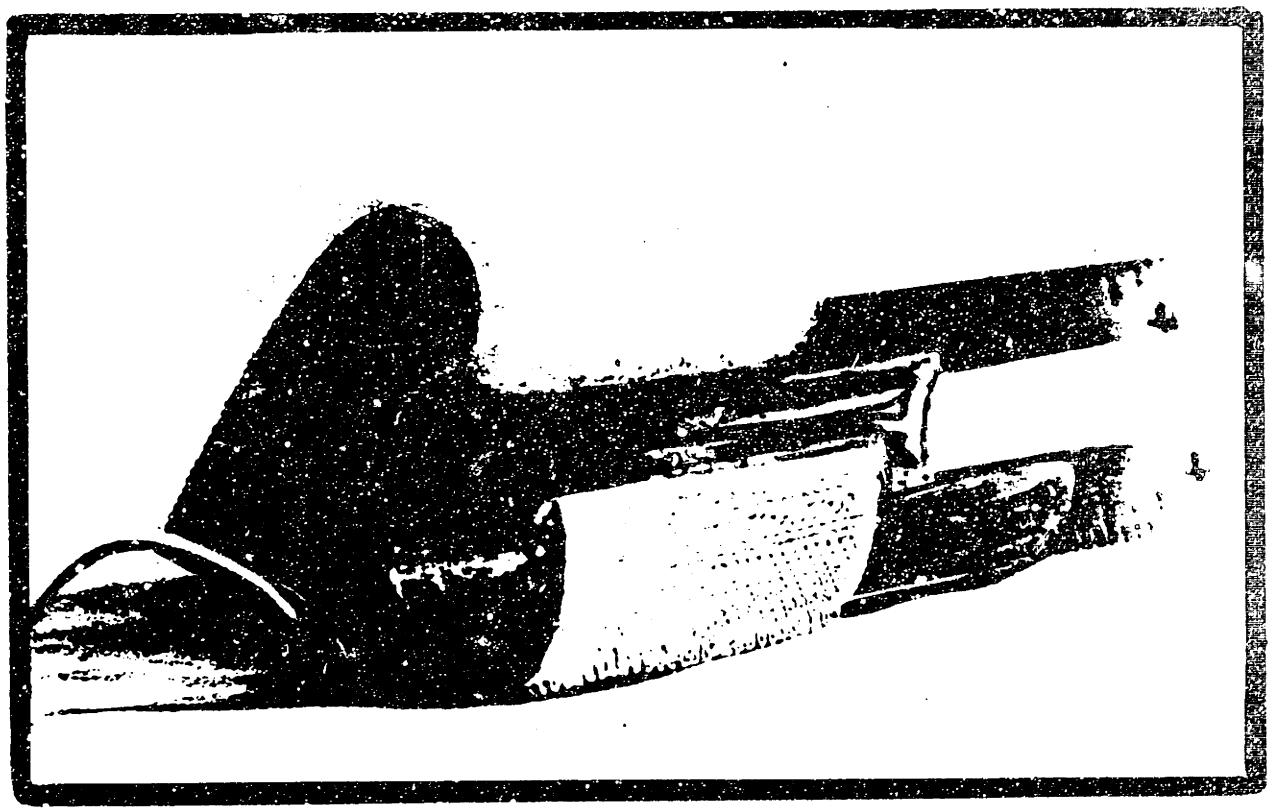
The last attempt at eliminating the fluctuations, under the assumption that they were local in origin, was to use heat. It was hoped that heating the finger (on which the plethysmograph was placed) would cause localized capillary dilation, reducing the constriction/dilation activity and resulting in a level amplitude signal. The "Finger Heater" (Pictures 5,6) warmed the finger while recordings were taken. Figure 40 shows that heat had no effect on reducing the variations.

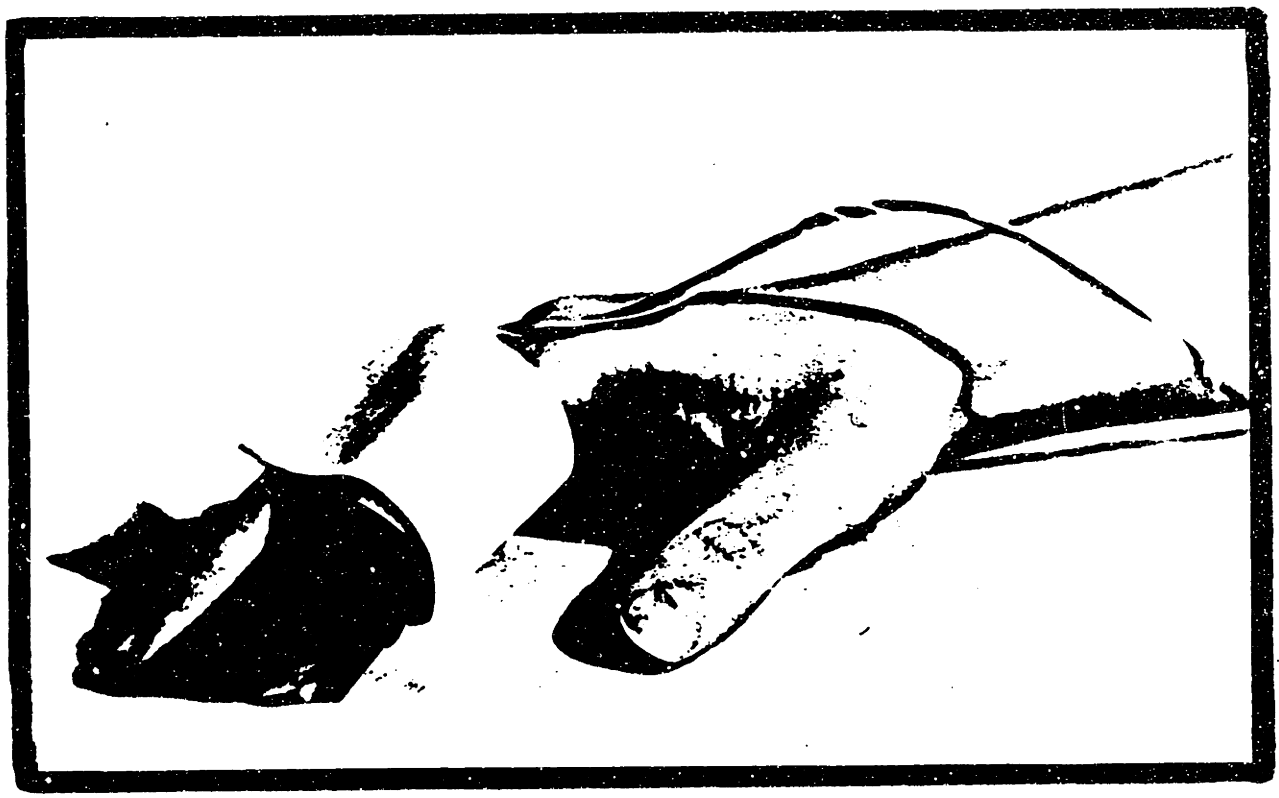
Local Versus Systemic Control: Dual Channel Recordings

Contrary to my previous assumption, the failure of the previous three approaches indicated that the waveform fluctuations were probably systemic in origin.

To verify systemic control, a second circuit was built with an identical reflective plethysmograph, natural log circuit, initial filter and gain of 60 stage (section B of Figure 29). The output of this gain stage went into a filter/gain stage like section D of Figure 29, with variable gain of 0 to 20.

The two diastolic circuit outputs were fed into the dual channel recorder. Recordings were made with the plethysmographs on two fingertips of the same hand, and with a plethysmograph placed on a fingertip of each hand. The variable gain circuit was adjusted so that both output

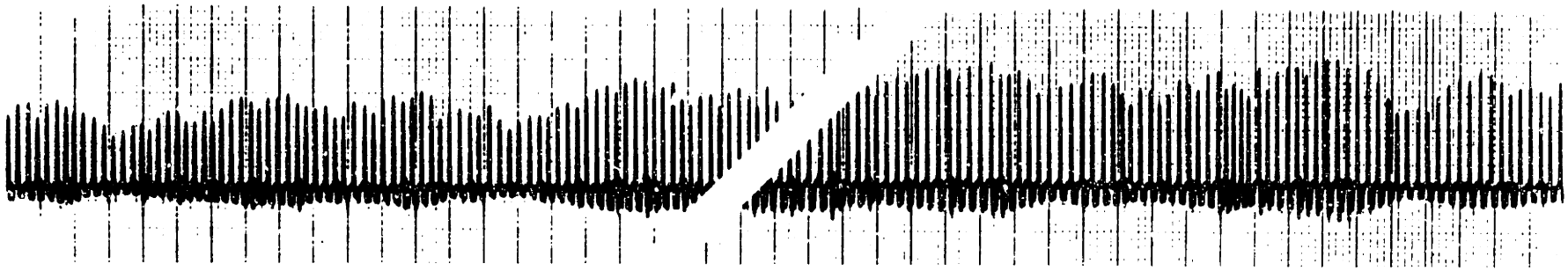




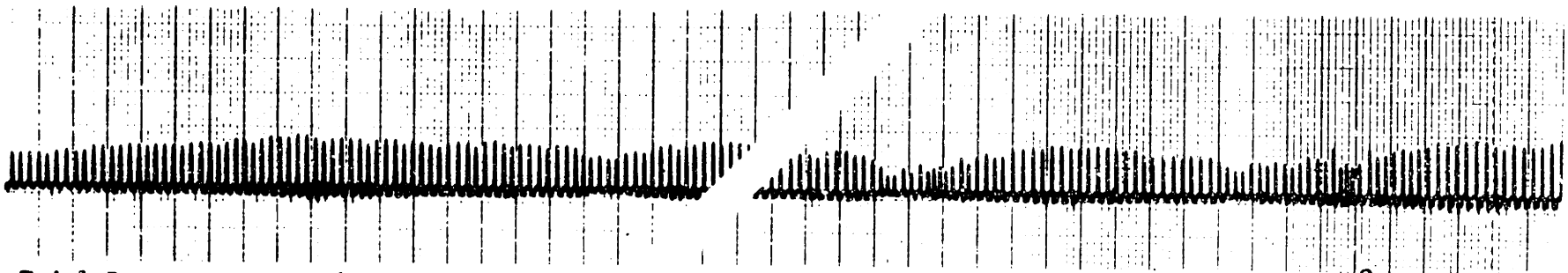
The "Finger Heater" and its placement. Picture 5

The "Finger Heater" and the pleth in place on a finger.

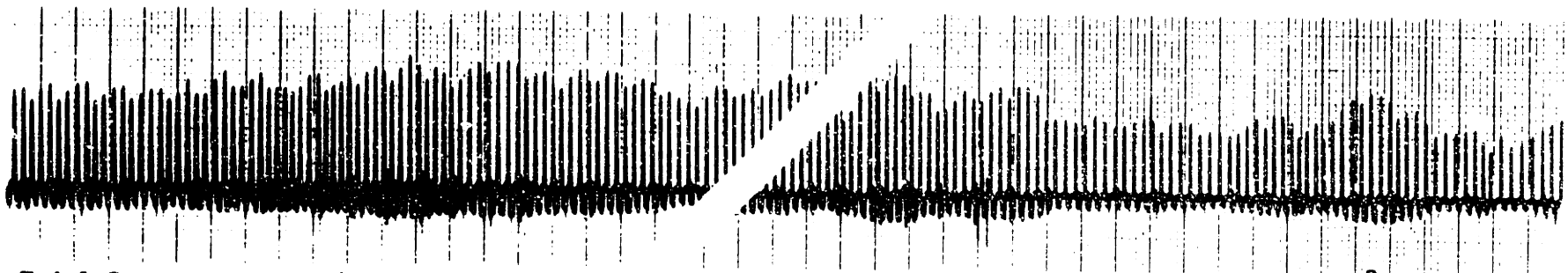
Picture 6



Trial A..... no heat ..... heater temperature 115° F .....



Trial B..... no heat ..... heater temperature 115° F .....



Trial C..... no heat ..... heater temperature 115° F .....

Figure 40

signals had approximately the same amplitude. Systemic control would be indicated by a mutual tracking of waveform variations on both channels. Representative results are shown in Figures 41a through d. It is clear that the waveform fluctuations are due to systemic circulatory controls.

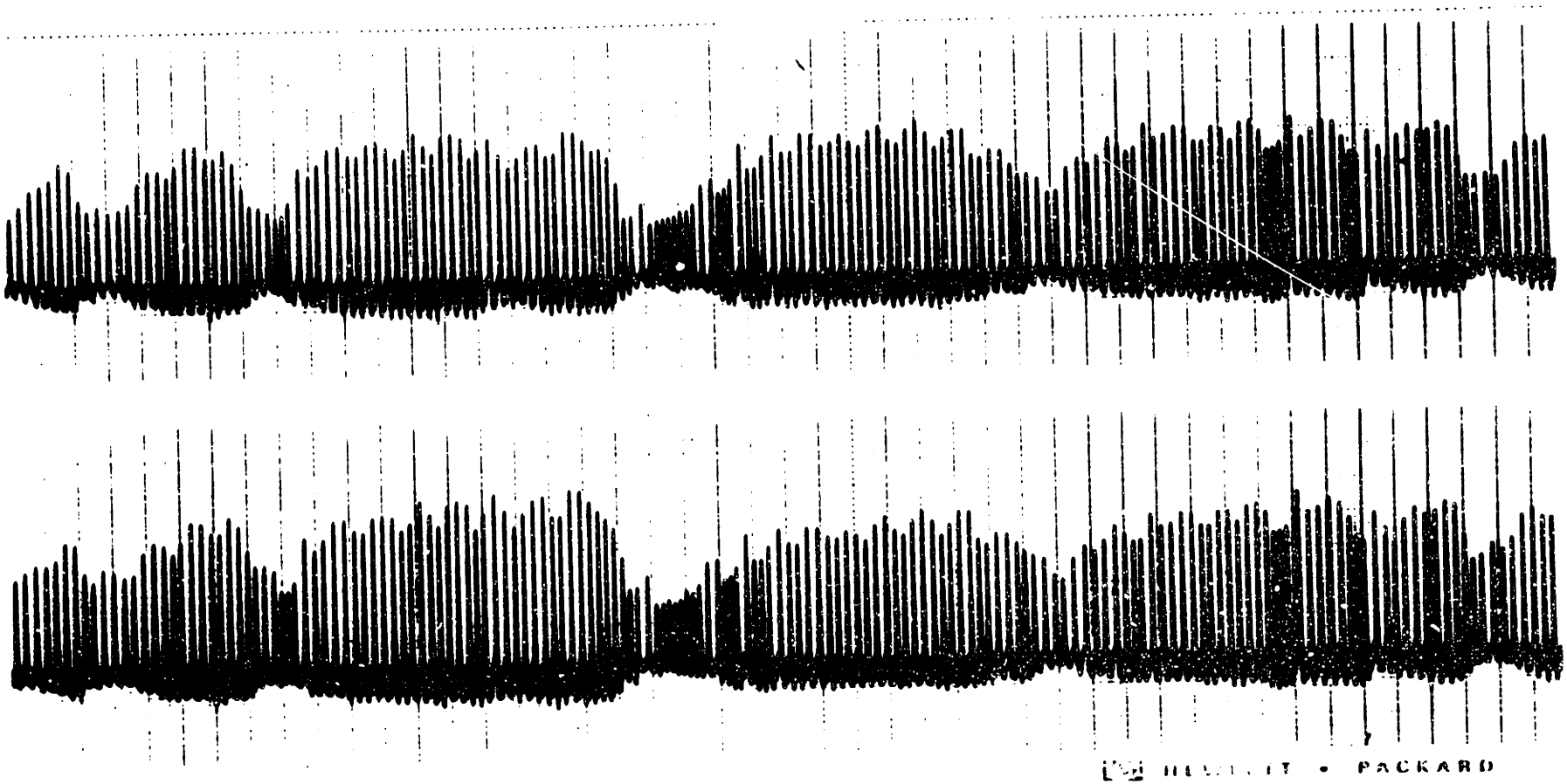
Dual Channel Recordings: The Idea

Given that the waveform fluctuations were primarily due to systemic controls, it seemed possible that diastolic blood pressure could be obtained with dual channel recordings.

Figure 42 depicts the technique. The first channel displayed one diastolic circuit's output for an uncuffed arm. This was the "reference" signal. The other diastolic circuit's output for the second, uncuffed arm was displayed simultaneously on the second channel. The gain on one circuit was adjusted to that both waveforms had the same amplitude with the cuff uninflated.

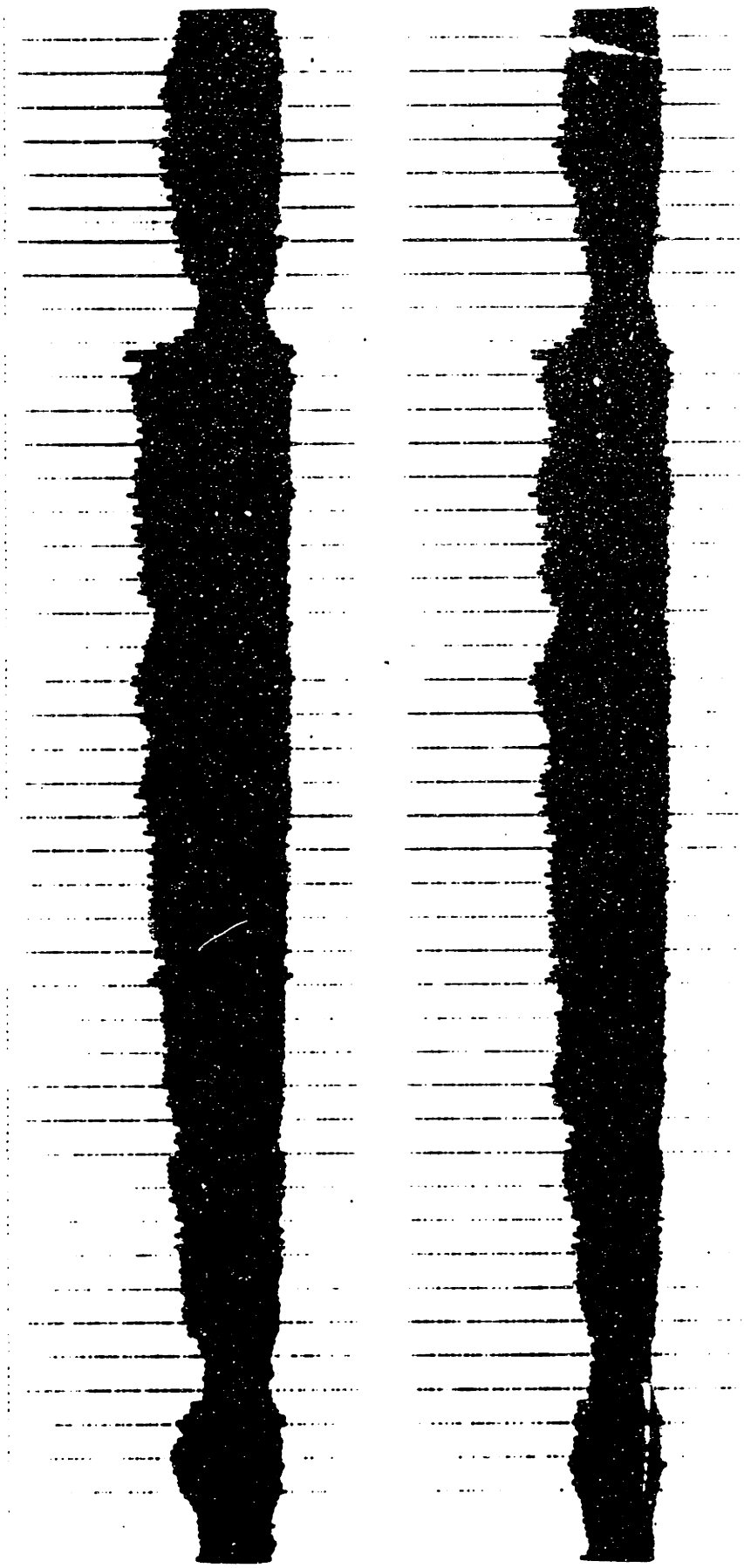
The cuff was inflated above the subject's diastolic level, partially occluding the blood vessels underneath, reducing the amount of blood flow to the fingers, and decreasing the amplitude of the cuffed arm signal. As in the original approach, cuff pressure was dropped slowly below the subject's diastolic pressure. The result no longer corresponded to the point at which the output amplitude levelled off, but rather to the pressure at which the cuffed arm's output resumed tracking of the reference signal.

The technique as described above failed to work. I



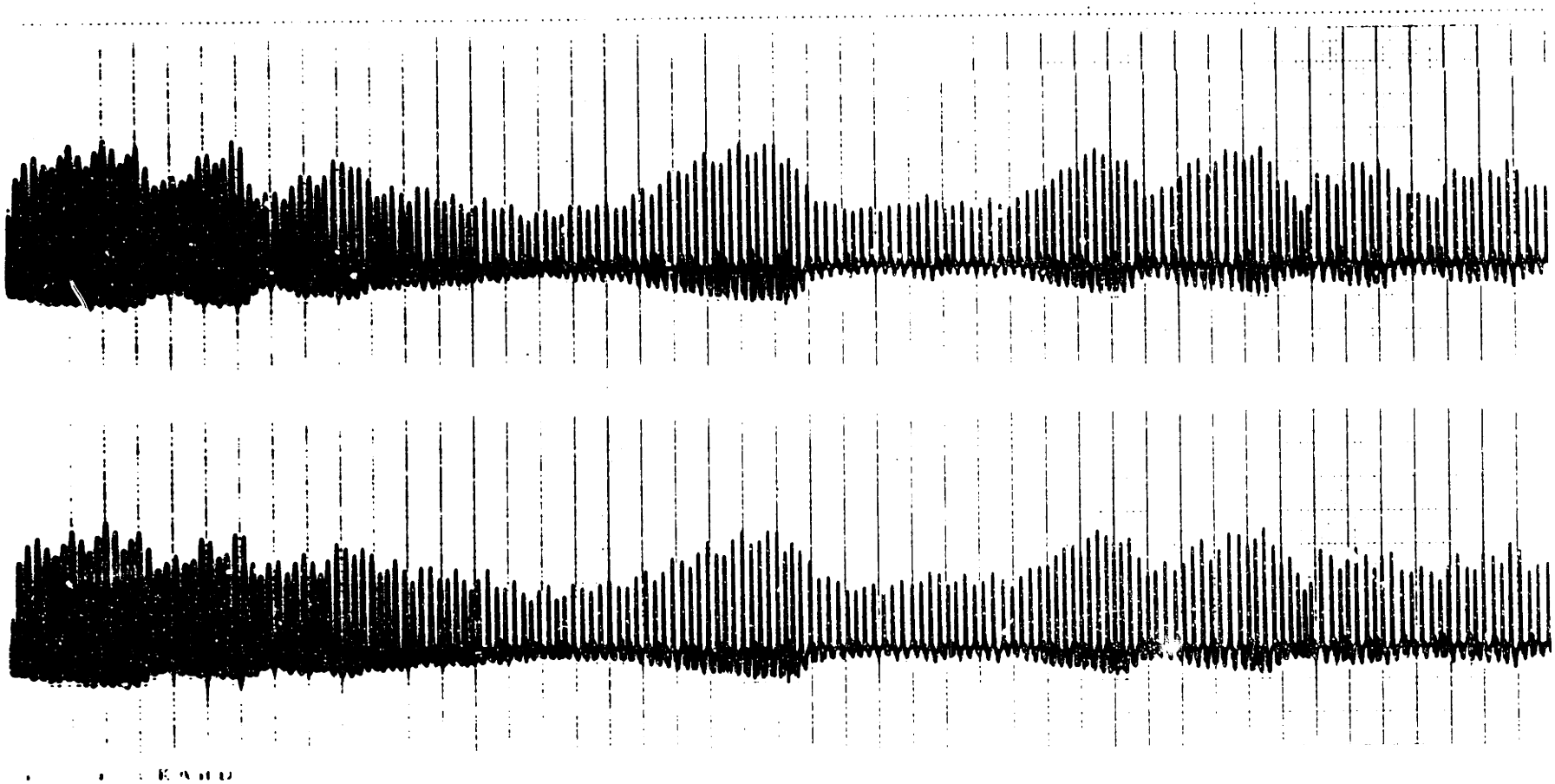
Subject A: Recordings taken off two fingers on same hand.
Chart speed: 1.25 mm/sec

Figure 418



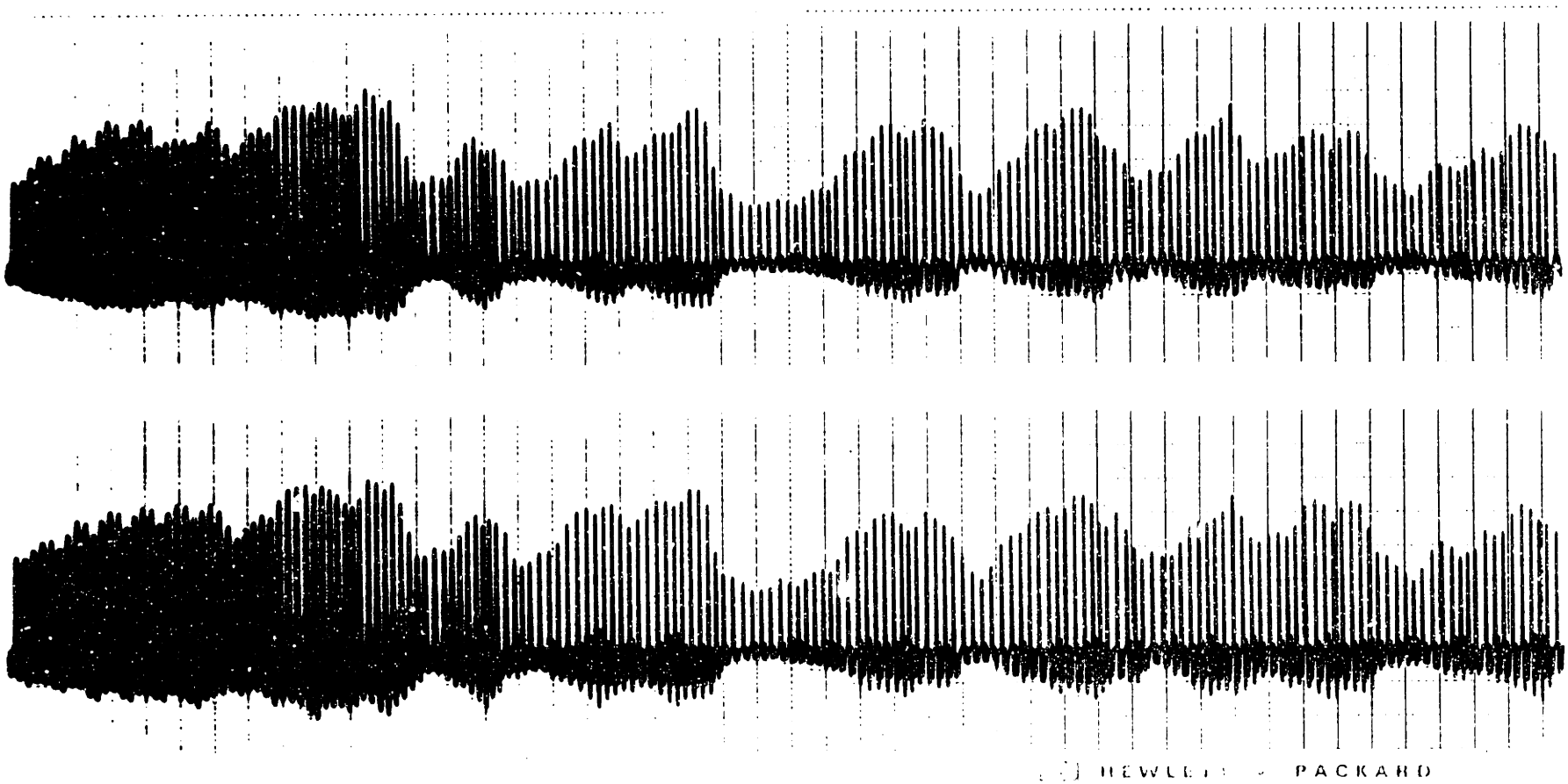
Subject R: Recordings taken off two fingers on same hand.
Chart speed: 1.25 mm/sec

Figure 41b



Subject C: Recordings taken off one finger of each hand.
Chart speed: 1.25 mm/sec

Figure 41c



Subject D: Recordings taken off one finger of each hand.
Chart speed: 1.25 mm/sec

Figure 41d

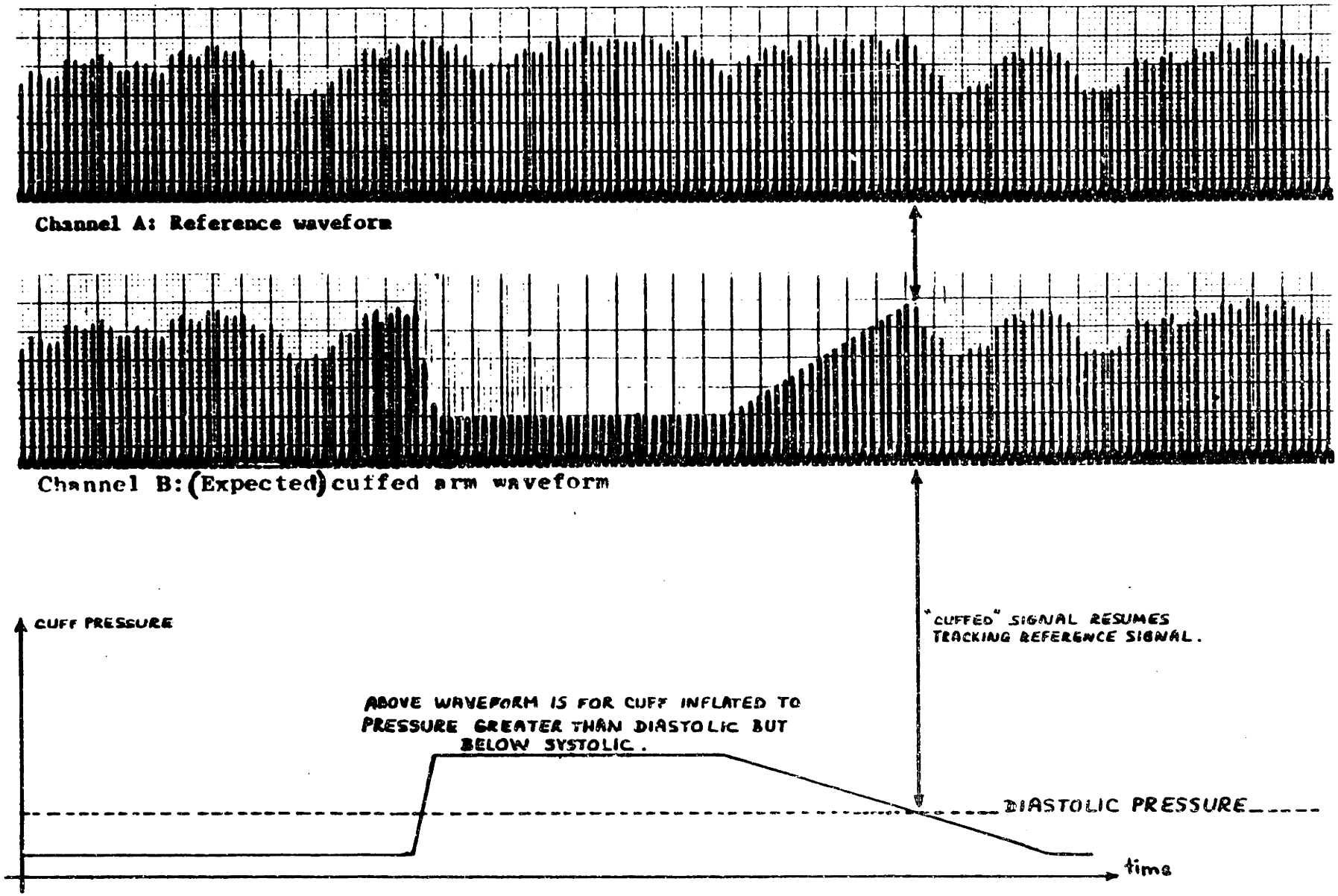

Channel A: Reference waveform
Channel B: (Expected) cuffed arm waveform
CUFF PRESSURE
"CUFFED" SIGNAL RESUMES TRACKING REFERENCE SIGNAL.
ABOVE WAVEFORM IS FOR CUFF INFLATED TO PRESSURE GREATER THAN DIASTOLIC BUT BELOW SYSTOLIC.
DIASTOLIC PRESSURE
time

Figure 42

attribute the failure to the effects of local circulatory controls in the fingers which, although not evident in all the results, dominated the recorded behavior. Figure 43 shows the problem. Pumping up the occlusion cuff suddenly to above the subject's diastolic pressure produces an oxygen deficiency in the cuffed arm's fingers. Local controls cause the blood vessels within to dilate. Consequently, cuff pressure reduction results in a marked increase in perfusion of those fingers. The cuffed arm signal overshoots the reference, and does not resume tracking until the oxygen level returns to normal. By this time, cuff pressure has been dropped way below the diastolic level, and the technique has failed.

Dual Channel Recordings: Slow Cuff Inflation

The previous attempt demonstrated that oxygen deprivation, whether due to rapid cuff inflation, or to pressures above the subject's diastolic level, or both, must be avoided for the dual channel technique to work. This criterion was incorporated into the next approach used, a dual channel recording with slow occlusion cuff inflation.

Cuff pressure was raised slowly (approximately 100 mm Hg per minute) and the two channels monitored. Unlike the previous approach, the effects of the rising pressure edge were of interest. Theoretically, the two channels would track each other until the diastolic point was reached, corresponding to the pressure at which the cuffed arm signal

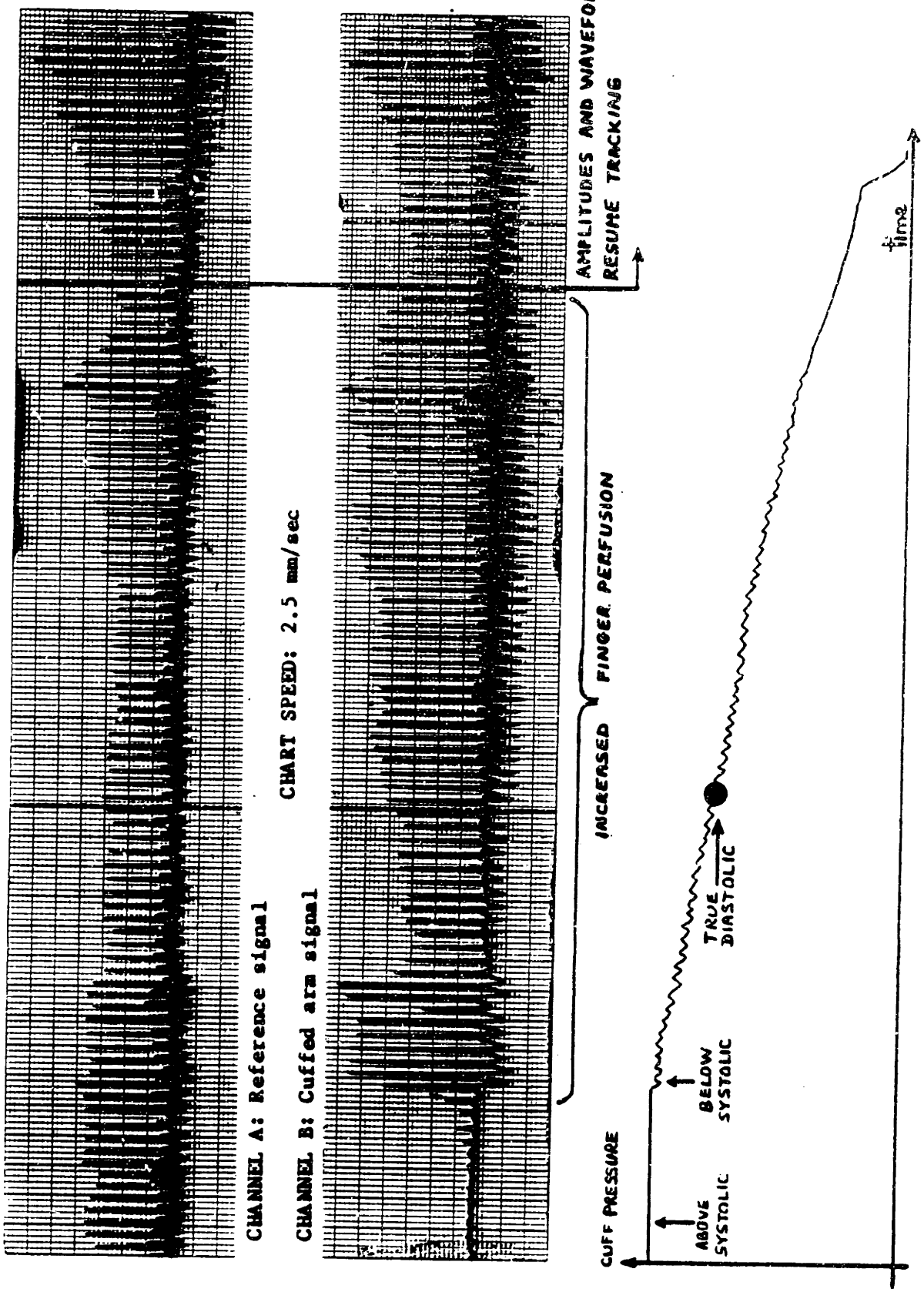

CHANNEL A: Reference signal
CHANNEL B: Cuffed arm signal
CHART SPEED: 2.5 mm/sec
AMPLITUDES AND WAVEFORMS RESUME TRACKING
INCREASED FINGER PERFUSION
CUFF PRESSURE
ABOVE SYSTOLIC
BELOW SYSTOLIC
TRUE DIASTOLIC
time

Figure 43

amplitude would begin to decrease. It was hoped that the slow inflation rate would allow the physiology to adapt to the increasing pressure. The rising pressure edge allowed the determination to be made without having to significantly exceed the diastolic pressure. Thus oxygen deprivation was minimized.

Unfortunately, this technique also failed to work. Once cuff pressure exceeds venous pressure, the subject's venous blood return path is occluded. Numerous heart beats ensue before the slowly rising pressure exceeds the diastolic point, and the pressure ascertained. Venous pooling causes a buildup of blood in the arm's blood vessels, resulting in stretched vessel walls and increased circuitry output amplitude (Figure 44) Consequently, rather than tracking the reference signal and then decreasing in amplitude, the cuffed arm signal exceeds the reference until cuff pressure close to systolic is attained, and the technique fails.

Dual Channel Recordings: Using Pressure Pulses

The previous attempts taught that dual channel recordings to obtain diastolic blood pressure would not work if either oxygen deprivation of the fingers or venous pooling of the arm occured.

The final idea was to use short cuff pressure pulses in conjunction with two channel recordings. First, gain would be adjusted so that both signals had the same amplitude for no applied cuff pressure. The occlusion cuff would be quickly

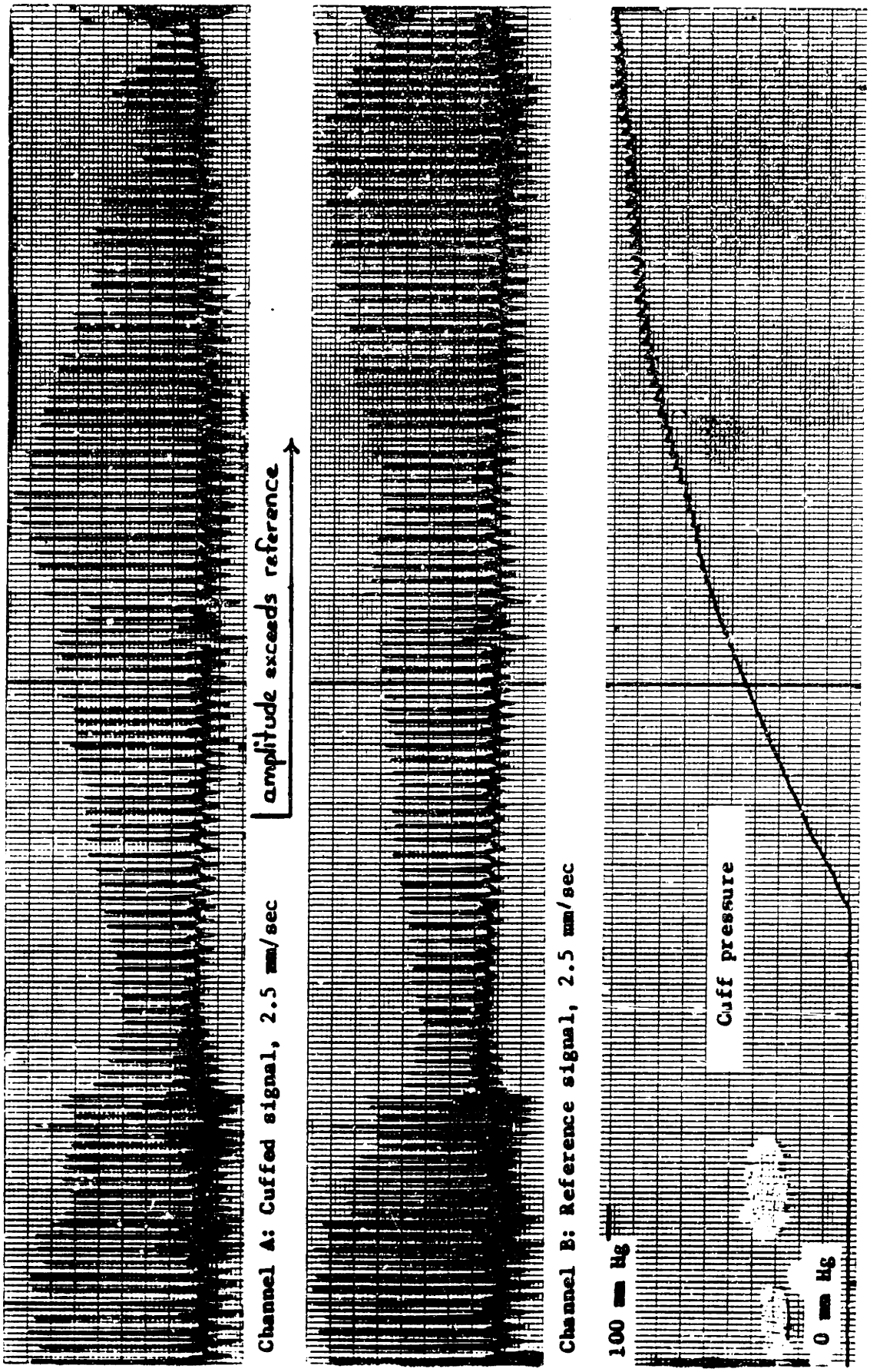

Channel A: Cuffed signal, 2.5 mm/sec
amplitude exceeds reference
Channel B: Reference signal, 2.5 mm/sec
100 mm Hg
Cuff pressure
0 mm Hg

Figure 44

inflated (within a heartbeat) to a set pressure. Within a heartbeat or two, a comparison of the two signal amplitudes would be made and cuff pressure released. The short duration of the pressure would avoid both aforementioned conditions. Equal signal amplitudes would indicate that the cuff had no effect on the arm's circulatory state. After waiting a few heartbeats to insure minimal physiological disturbance, the cuff would be re-inflated to a higher pressure, a signal comparison performed and pressure released again. This procedure would be iterated until the cuffed arm signal's amplitude was less than the reference. The corresponding pressure would be the subject's diastolic value.

The test set-up used to provide these pressure pulses is shown in Figure 45. The pressure difference between successive inflations was 10 mm Hg. To simplify the mechanics of the experimentation, a "Finger Cuff" (Picture 7) was used instead of the usual upper arm occlusion cuff. The two plethysmographs were placed on adjacent fingers of the same hand. Previous experiments with systolic blood pressure had shown that values obtained with the finger cuff varied by only a few millimeters of mercury from those obtained with the standard cuff. This discrepancy would not affect the evaluation of the pressure pulse technique. (Finger cuffs have also been used in other research of similar nature [13,16]).

An interesting and unexpected observation was made, shown in Figure 46. The quick release of cuff pressure caused a

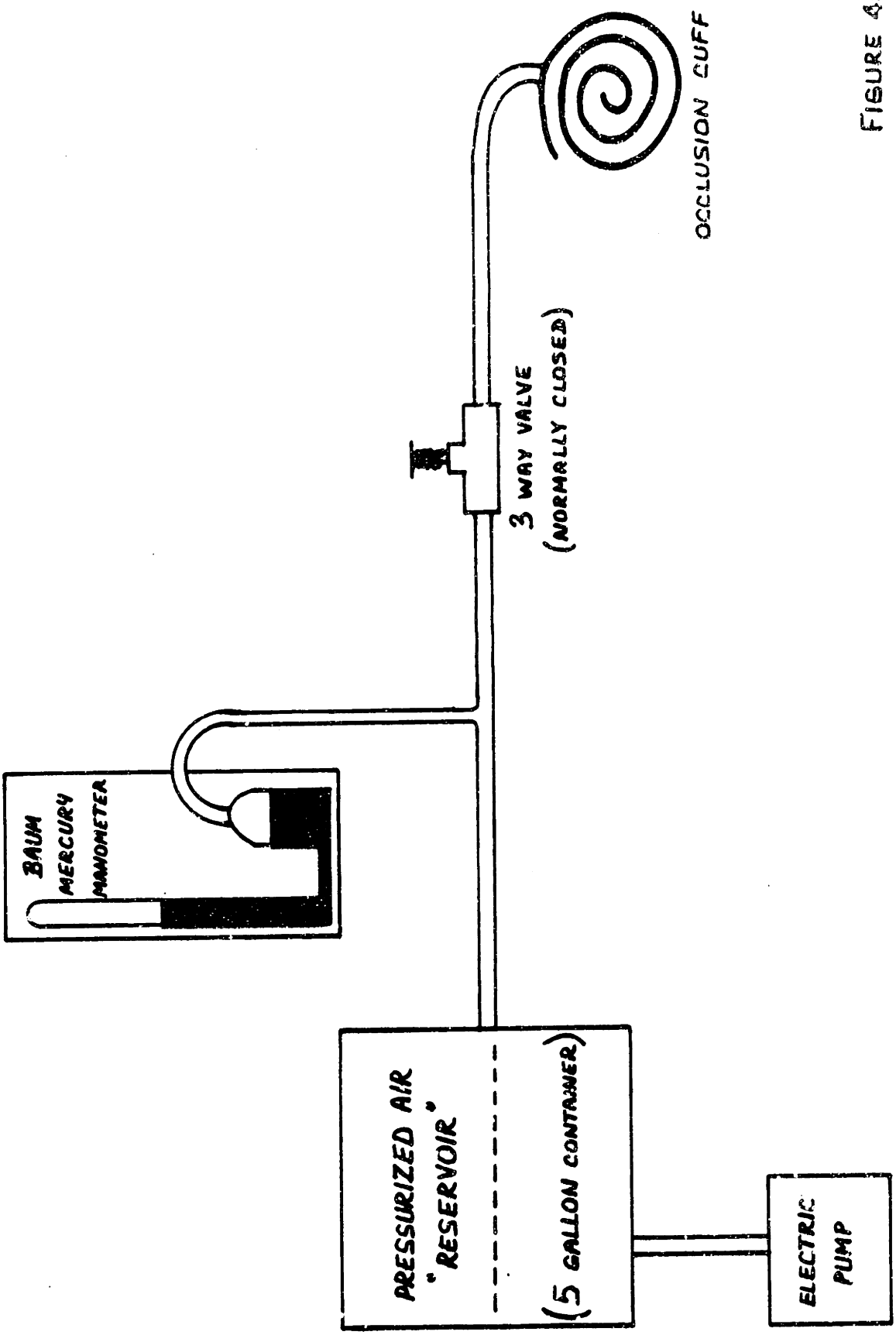
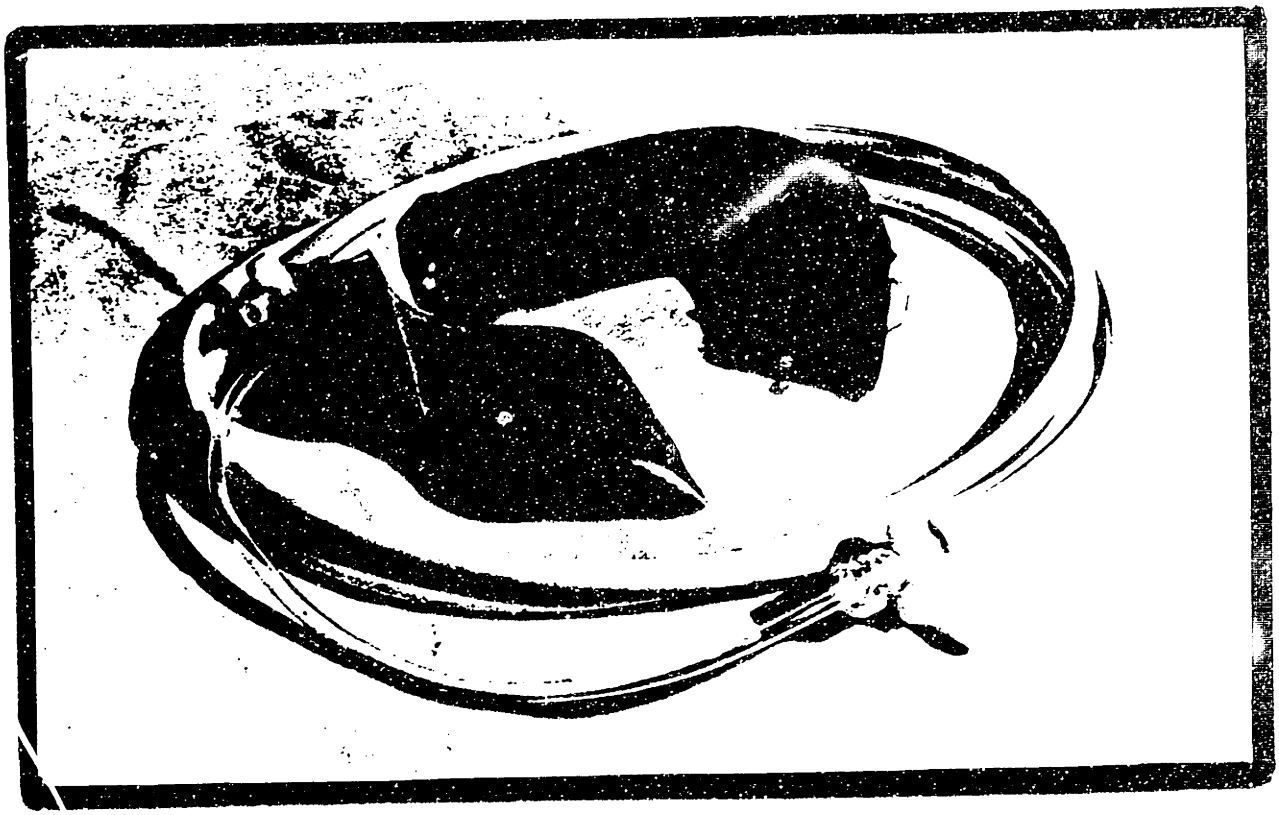
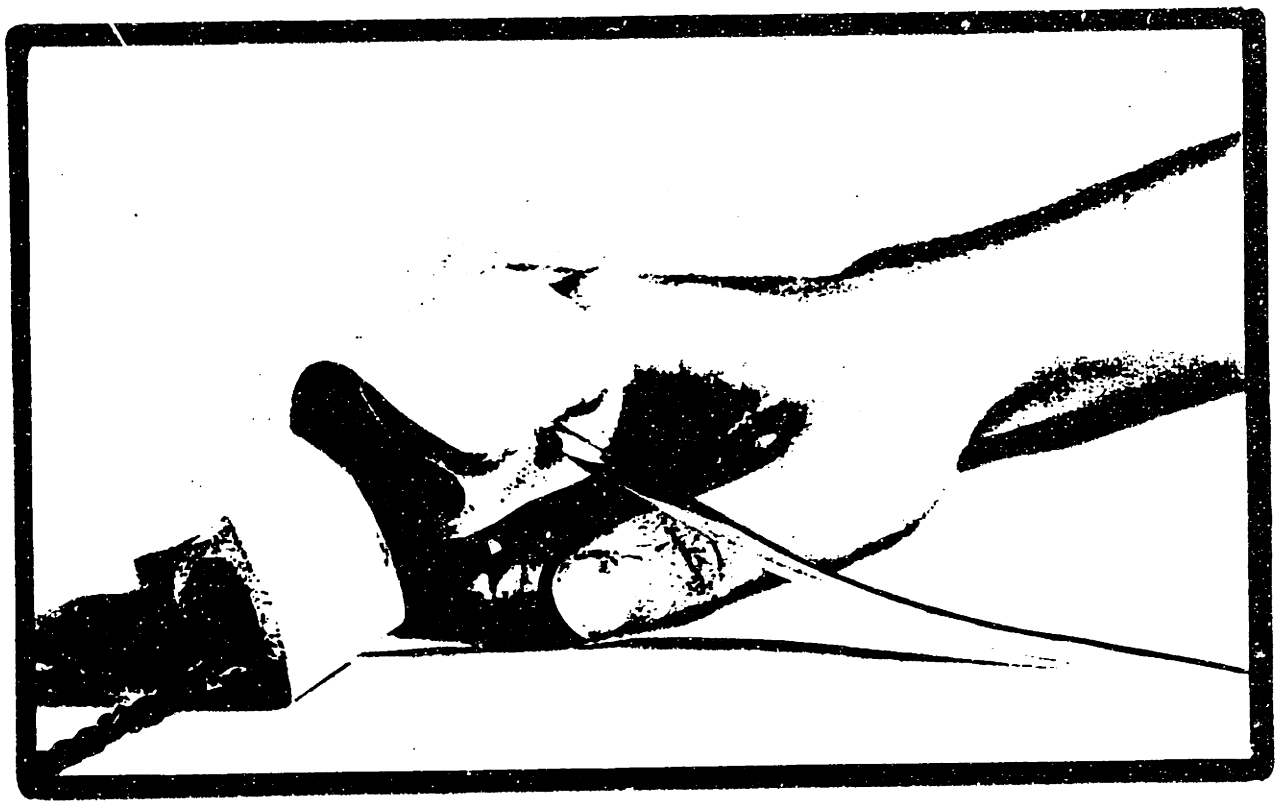
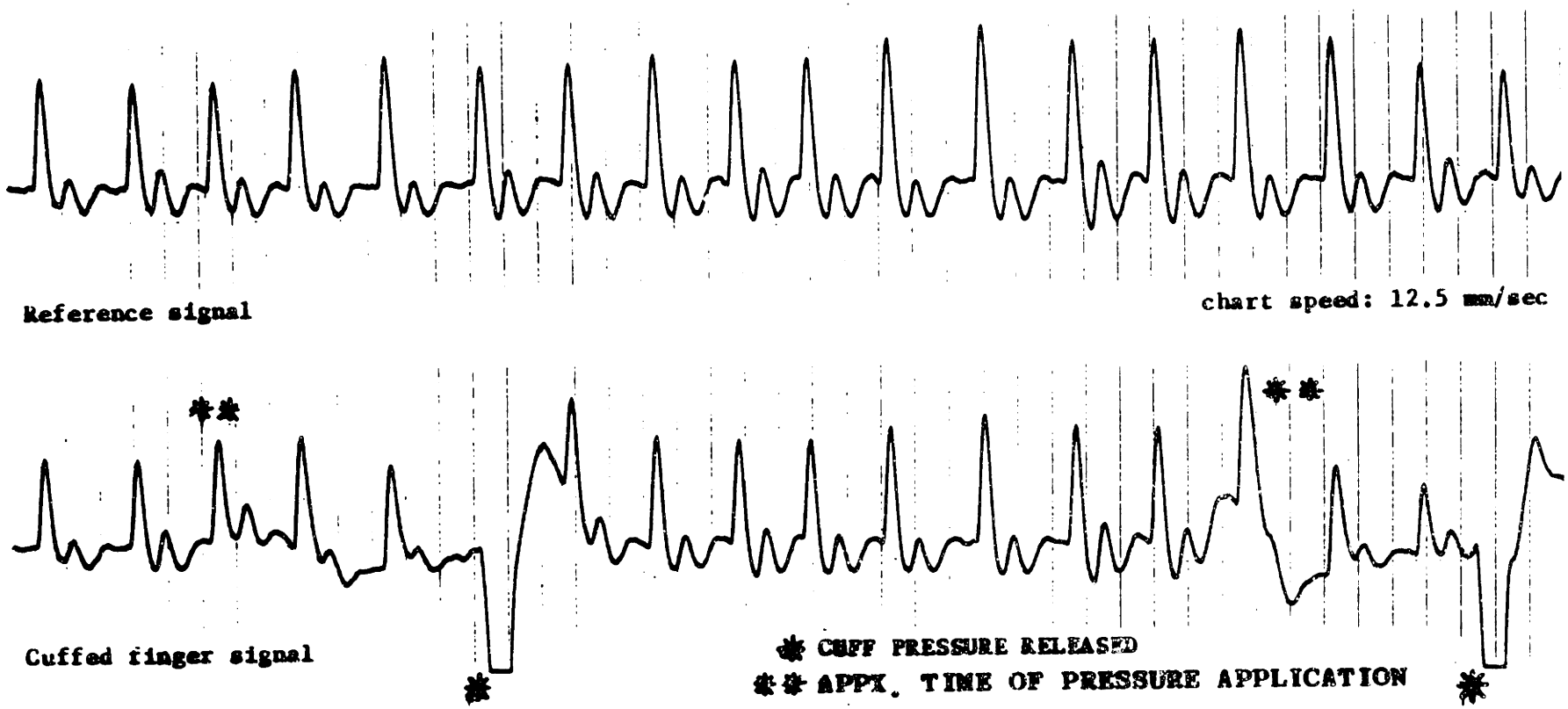

BAUM
MERCURY
MANOMETER
3 WAY VALVE
(NORMALLY CLOSED)
OCCLUSION CUFF
PRESSURIZED AIR
"RESERVOIR"
(5 GALLON CONTAINER)
ELECTRIC
PUMP

FIGURE 45





The Finger Cuff and its placement. **Picture 7**



When cuff pressure is applied ..... and when it is released .....

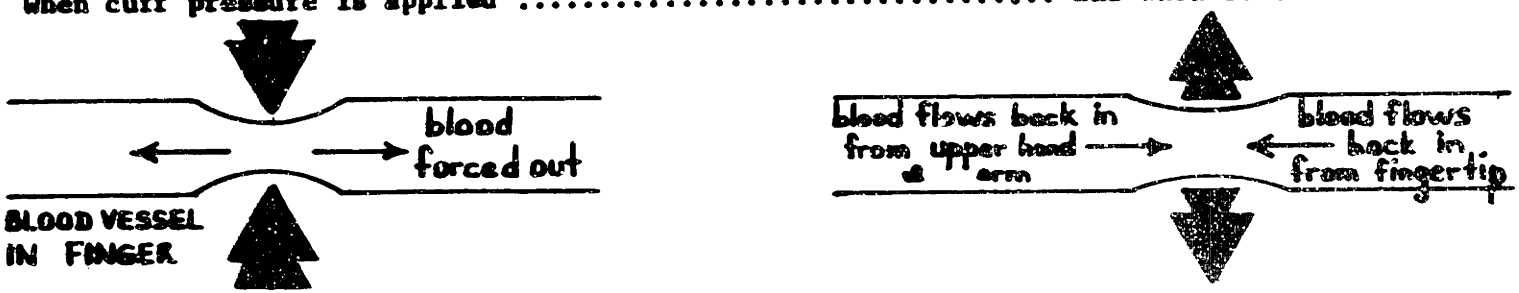


Figure 46

marked undershoot on the circuitry output. This effect was noted even at pressures as low as approximately 20 mm Hg. It appears that when cuff pressure is applied, blood in the fingers is forced away from the cuff area (Figure 46). Upon pressure release, blood flows back into the cuff area. Some of this blood flows from the fingertips, temporarily decreasing the quantity of blood there and causing the observed undershoot.

Initial results showed promise. Figures 47a through c show a series of recordings made as diastolic pressure was determined for a subject. The undershoots indicate when cuff pressure was released, but they are not used for the amplitude comparison. The cuff was inflated for approximately two heartbeats duration, and the corresponding output pulses are the ones of interest. Referring to Figures 47a and b, neither 20 or 40 mm Hg produced noticeable effects on the cuffed finger. The amplitude of the cuffed signal is noticeably lower than the reference with 60 mm Hg (Figure 47c). The actual diastolic value for the subject was approximately 55 mm Hg.

Further trials showed that the technique is too physiology dependent to be useful. For subjects with poor finger perfusion, whose reference signals were of low amplitude to start with, even cuff pressures as high as 90 mm Hg produced no noticeable effect on the output signal, when in fact the diastolic values ranged around 50 to 60 mm Hg (Figure 48). Even for well perfused fingers, it was like a guessing game

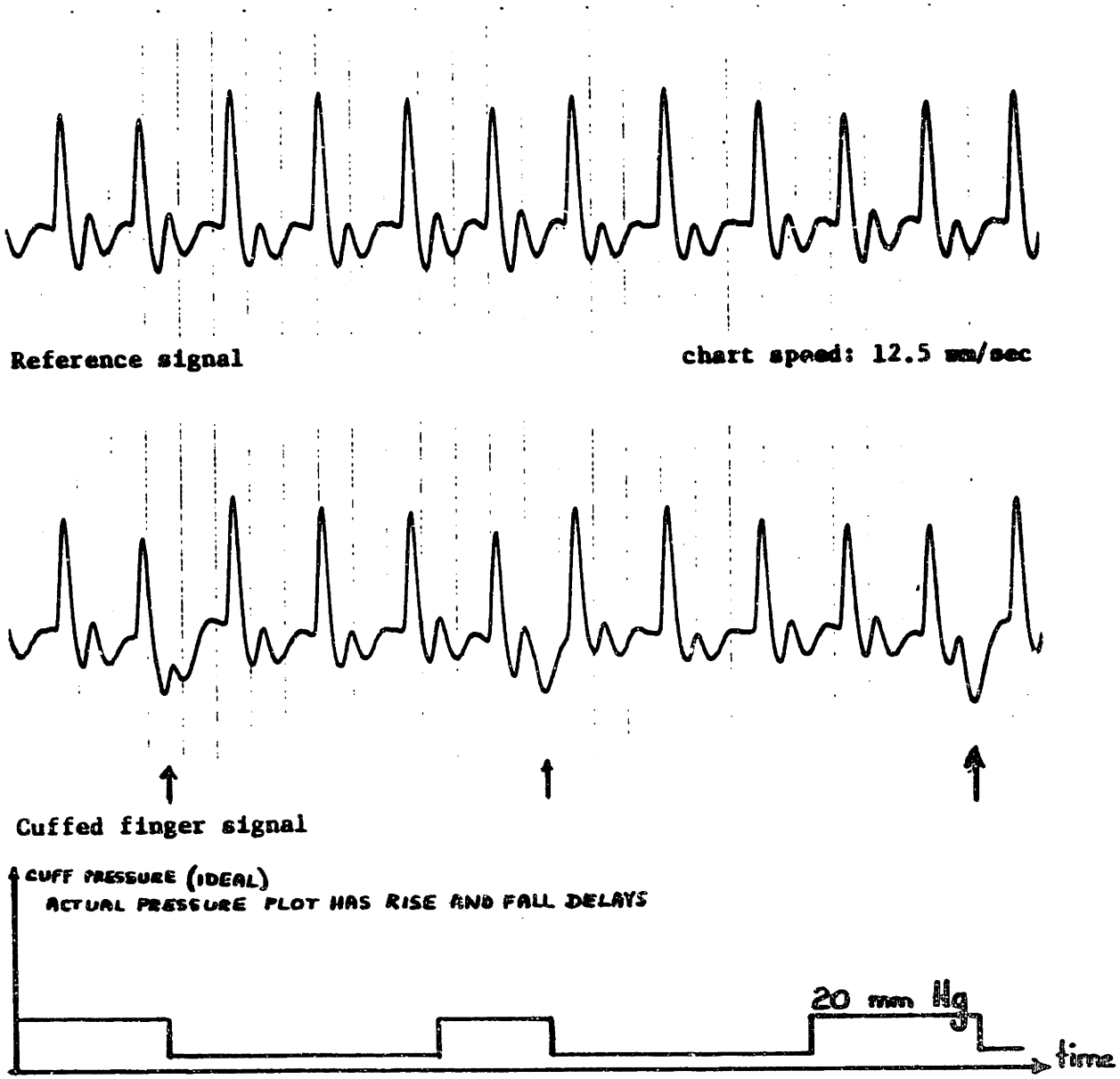

Reference signal
chart speed: 12.5 mm/sec
Cuffed finger signal
CUFF PRESSURE (IDEAL)
ACTUAL PRESSURE PLOT HAS RISE AND FALL DELAYS
20 mm Hg
time

Figure 470

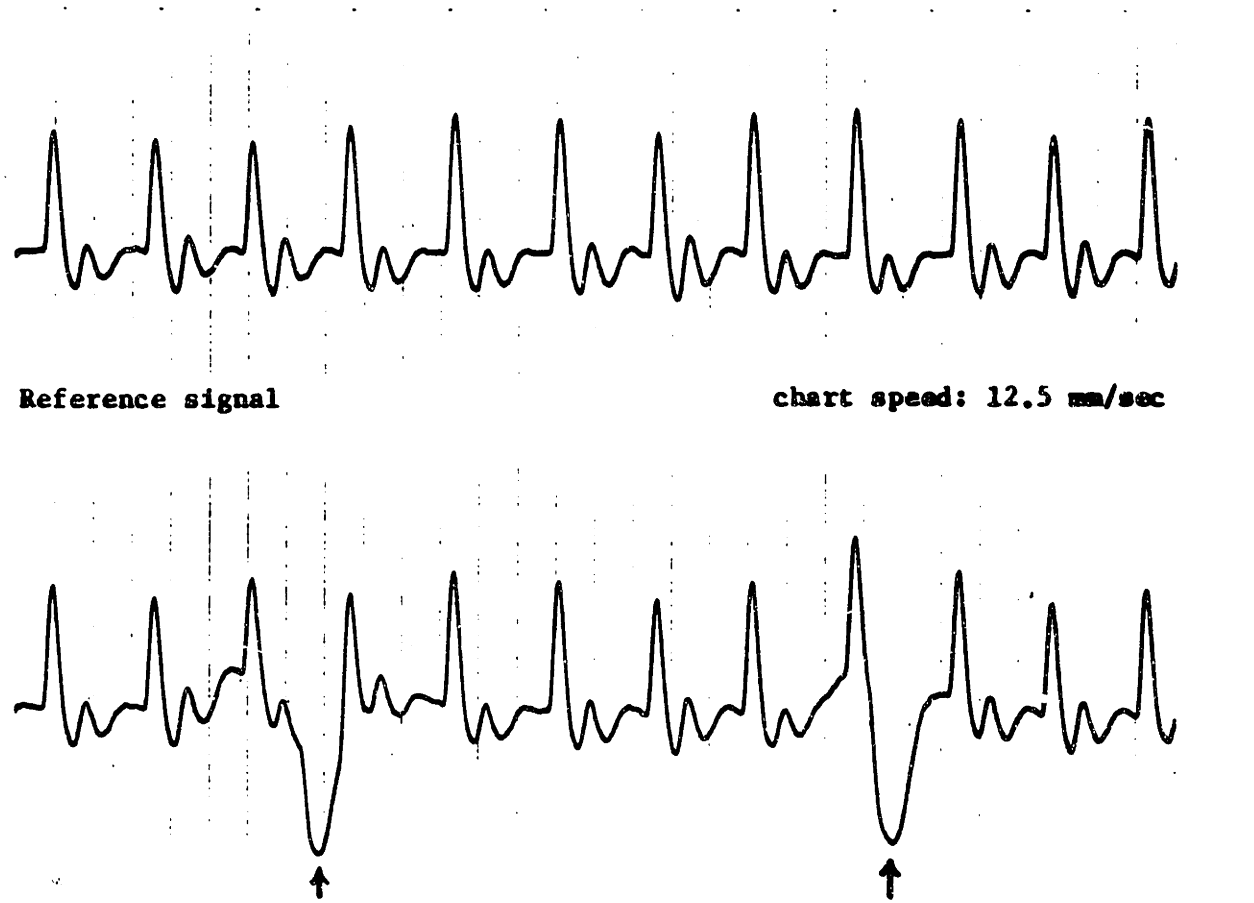

Reference signal
chart speed: 12.5 mm/sec
Cuffed finger signal
CUFF PRESSURE (IDEAL)
40 mm Hg
time

Figure 47b

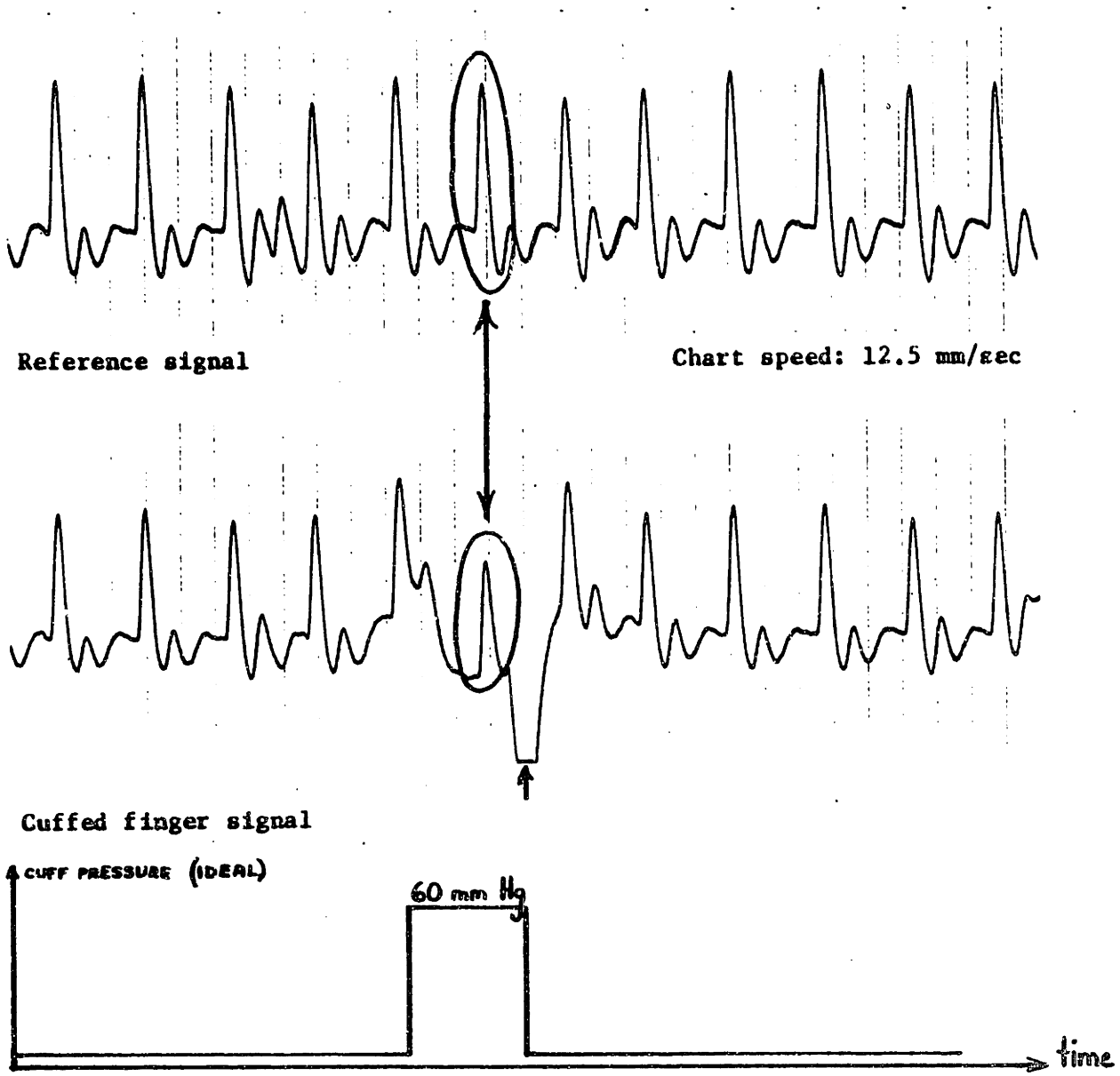
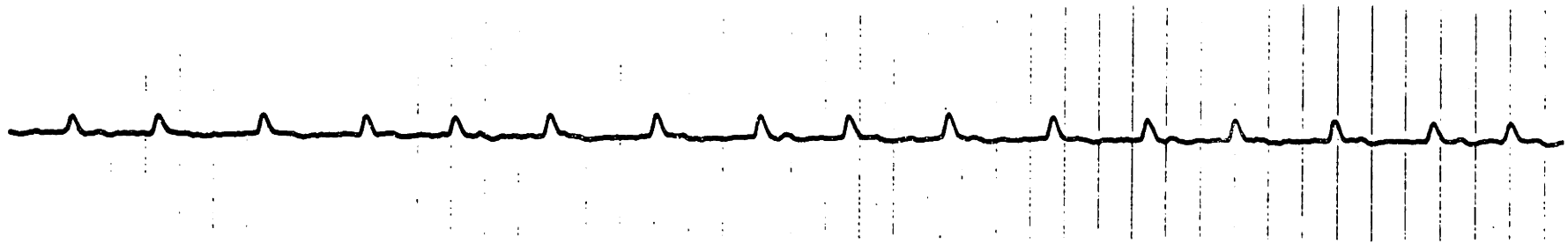
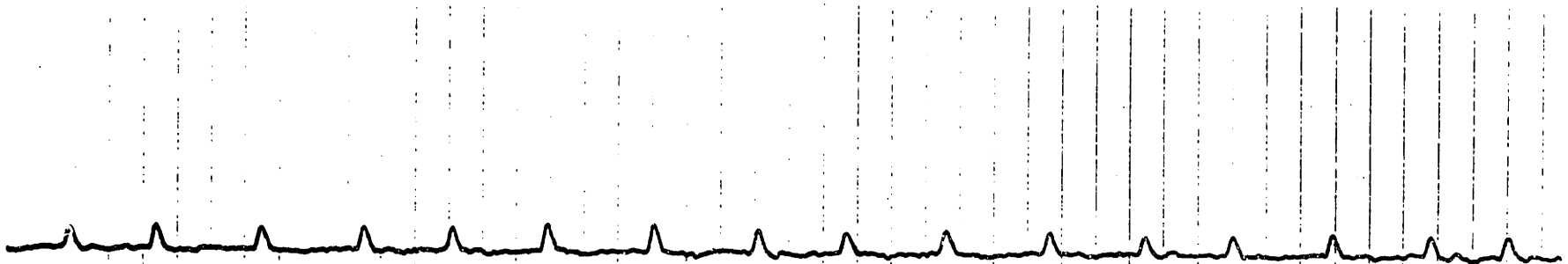

Reference signal
Chart speed: 12.5 mm/sec
Cuffed finger signal
CUFF PRESSURE (IDEAL)
60 mm Hg
time

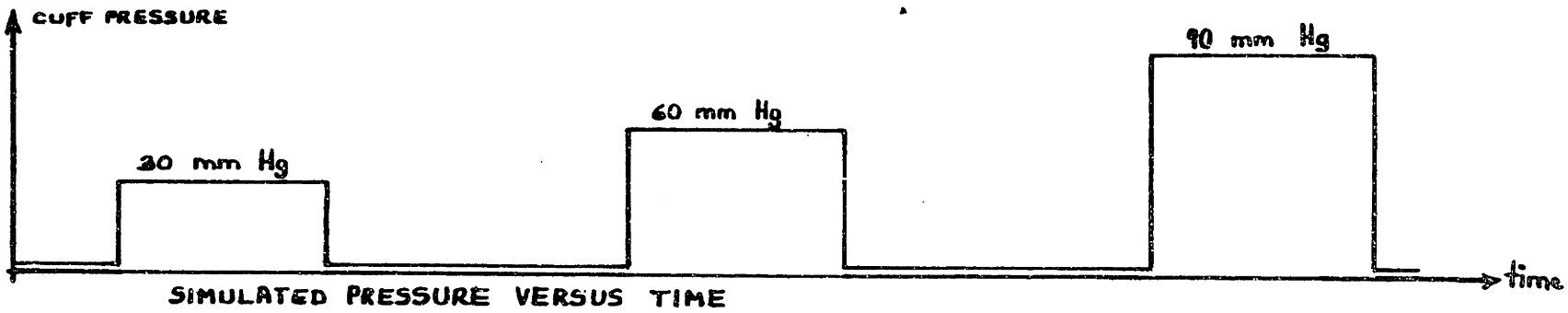
Figure 47c



Reference signal - poorly perfused finger



Cuffed finger signal - poorly perfused finger



SIMULATED PRESSURE VERSUS TIME

Figure 48

to determine whether noted amplitude differences were due to cuff effect or natural finger to finger perfusion differences over time.

A summary of the approaches and problems related to ascertaining diastolic blood pressure is shown in the flowchart of Figures 49a,b, and c.

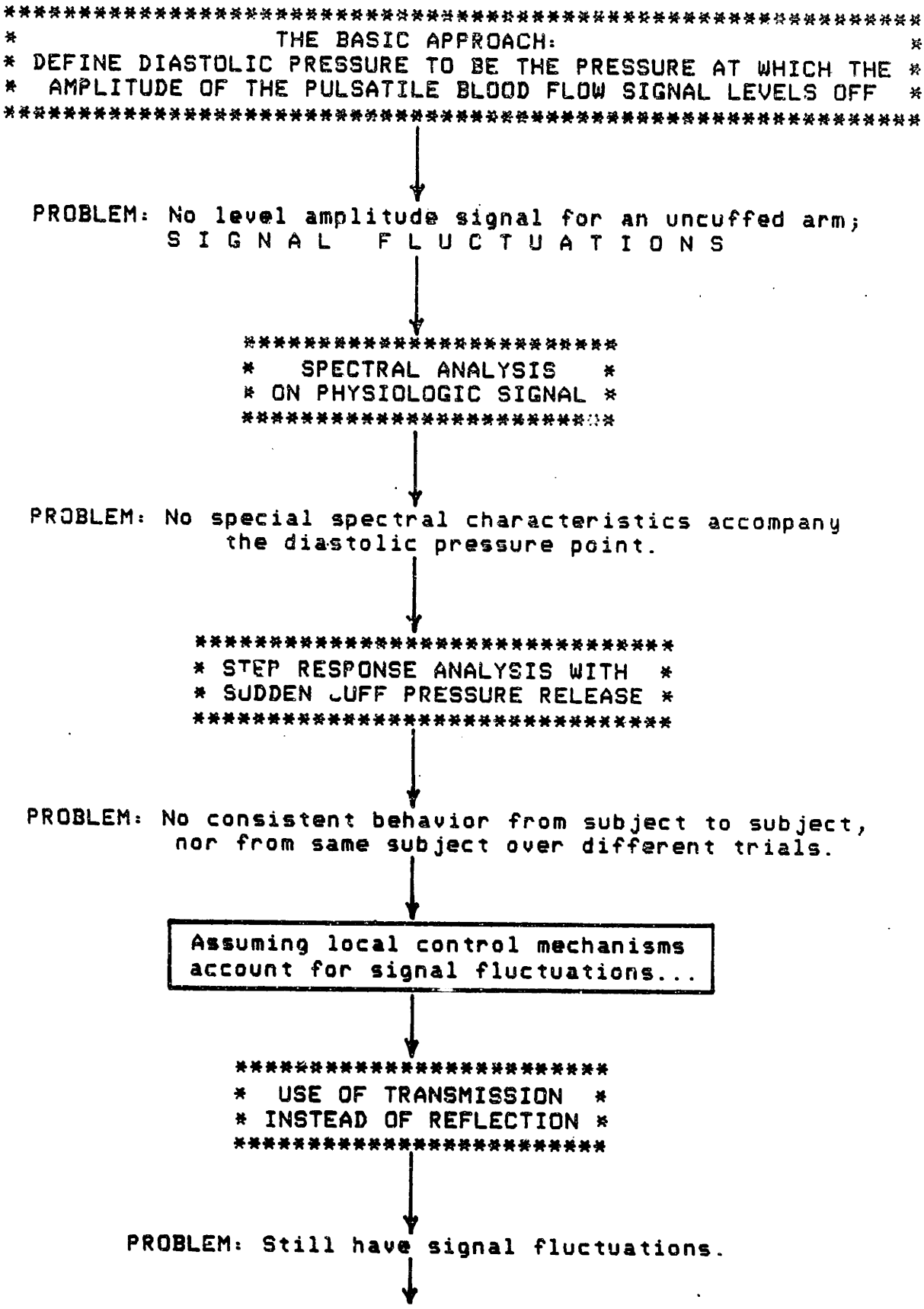

THE BASIC APPROACH:
DEFINE DIASTOLIC PRESSURE TO BE THE PRESSURE AT WHICH THE AMPLITUDE OF THE PULSATILE BLOOD FLOW SIGNAL LEVELS OFF
PROBLEM: No level amplitude signal for an uncuffed arm; S I G N A L F L U C T U A T I O N S
SPECTRAL ANALYSIS ON PHYSIOLOGIC SIGNAL
PROBLEM: No special spectral characteristics accompany the diastolic pressure point.
STEP RESPONSE ANALYSIS WITH SUDDEN CUFF PRESSURE RELEASE
PROBLEM: No consistent behavior from subject to subject, nor from same subject over different trials.
Assuming local control mechanisms account for signal fluctuations...
USE OF TRANSMISSION INSTEAD OF REFLECTION
PROBLEM: Still have signal fluctuations.

FIGURE 49a

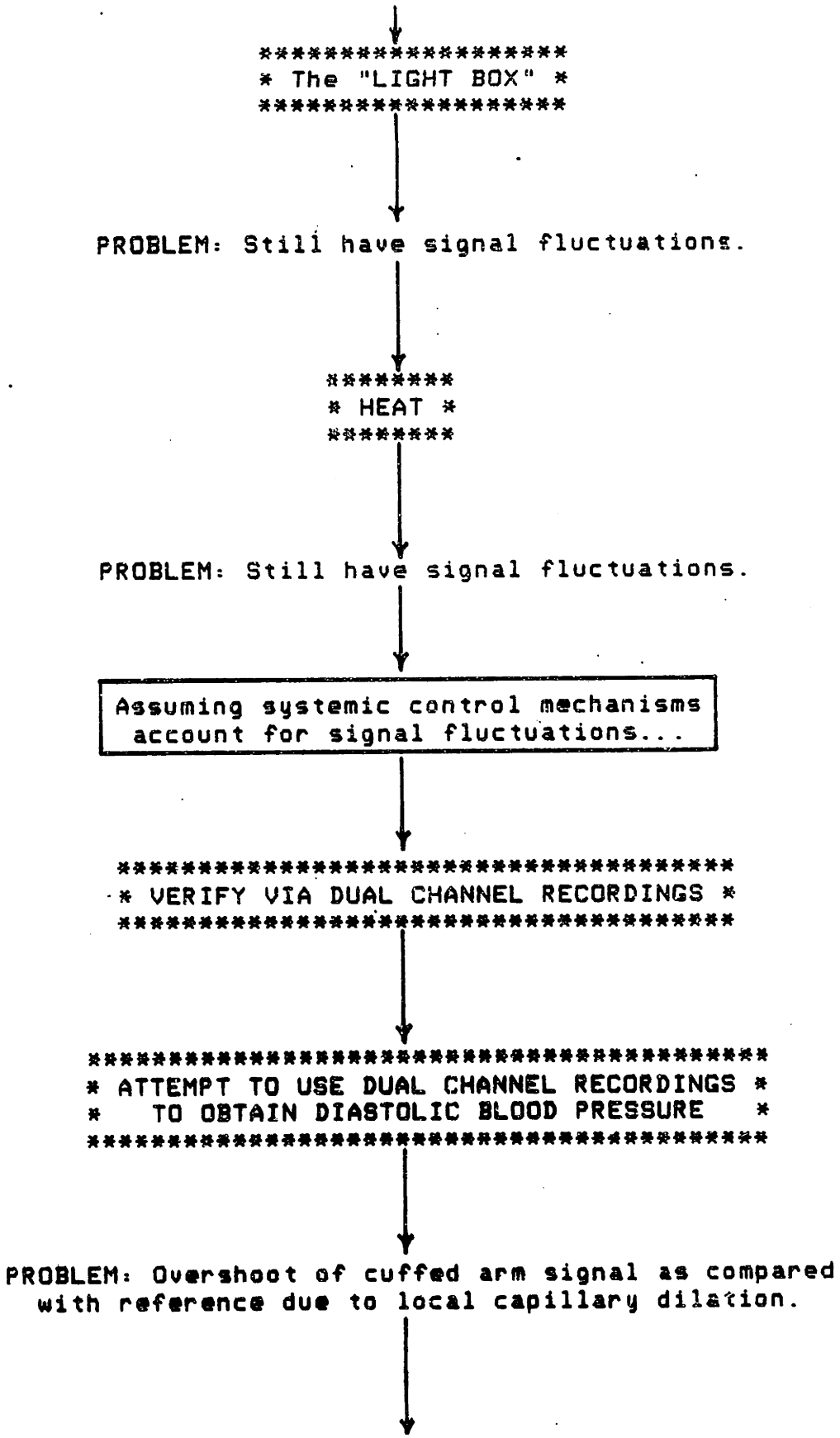

* The "LIGHT BOX" *
PROBLEM: Still have signal fluctuations.
* HEAT *
PROBLEM: Still have signal fluctuations.
Assuming systemic control mechanisms account for signal fluctuations...
* VERIFY VIA DUAL CHANNEL RECORDINGS *
* ATTEMPT TO USE DUAL CHANNEL RECORDINGS *
* TO OBTAIN DIASTOLIC BLOOD PRESSURE *
PROBLEM: Overshoot of cuffed arm signal as compared with reference due to local capillary dilation.

FIGURE 49b

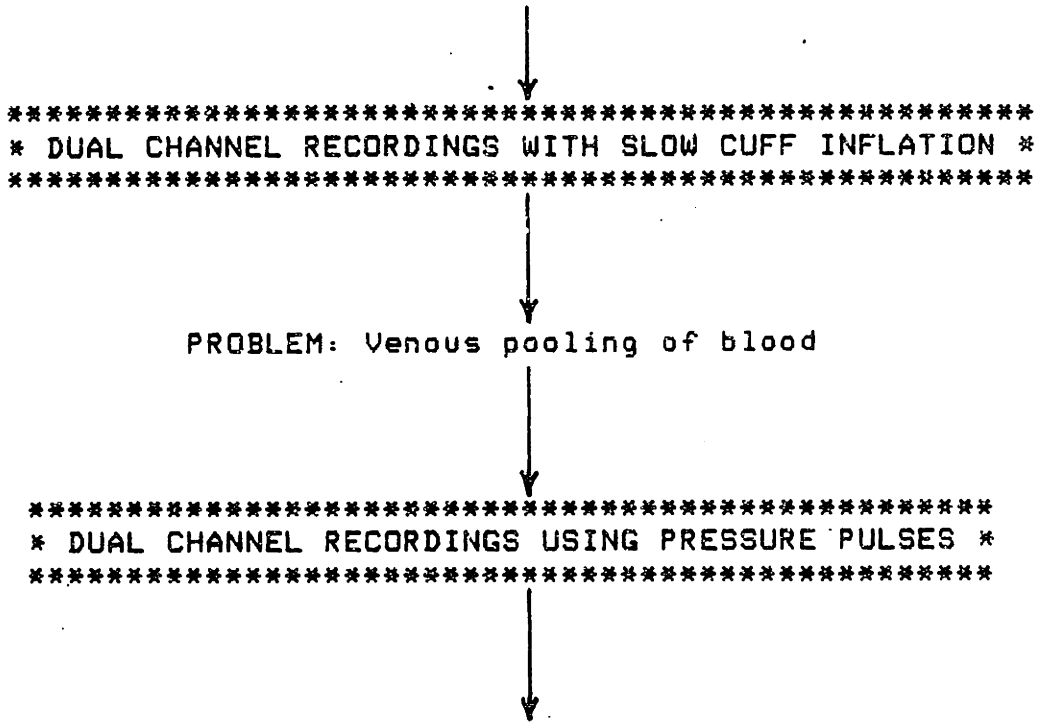

* DUAL CHANNEL RECORDINGS WITH SLOW CUFF INFLATION *
PROBLEM: Venous pooling of blood
* DUAL CHANNEL RECORDINGS USING PRESSURE PULSES *
PROBLEM: Inconsistent results. Technique fails to work with poor perfusion.

FIGURE 49c

DISCUSSION
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When evaluating the performance of a device or a technique, one must first define the criteria to be used in the evaluation. Presently used non-invasive blood pressure measurement techniques do not yield true aortic blood pressure. Techniques such as the auscultatory method do not even yield exact arterial blood pressure at the test site. The results are approximations, generally within a few millimeters of mercury. In many cases the medical profession has chosen to use these values instead of direct ones because they are so much simpler to obtain and entail less risk to the patient. Consequently, any new non-invasive technique should be compared to a widely used non-invasive method, rather than to a direct value.

Probably the widest used non-invasive technique is the auscultatory method, and it is with its results that the photoplethysmography approach is compared. According to a nursing specialist in critical care (Maryanne Schreiber, Marketing Specialist, Hewlett Packard Medical Products Group, former head of the Cardiothoracic Intensive Care Unit at Mount Sinai Hospital, New York City), the following discrepancy between a new technique and the auscultatory one for systolic blood pressure values would be acceptable:

An error of plus or minus 6.25%, which corresponds to:

+- 5 mm Hg at 80 mm Hg

+-10 mm Hg at 160 mm Hg

84.4% of the systolic blood pressure results met this

requirement. A more conservative estimate of performance is obtained when the error distributiuon is fitted with a normal (Gaussian) curve (Appendix 3). Given the mean error of 1.77%, and the standard deviation of 4.15%, the Gaussian distribution function predicts that the technique will meet the plus/minus 6.25% specification 83.3% of the time. I would conclude that the photoplethysmography technique could be accepted by the medical community as a means of obtaining non-invasive systolic blood pressure.

However, the photoplethysmography technique fails to estimate diastolic blood pressure. I believe that the main obstacle is the lack of knowledge about 1) the nature of the signal received with photoplethysmography and 2) the physiology within the test site and how to control it while making a non-invasive measurement. Indeed, once we ascertained that the basic observed fluctuations in the signal were due to systemic control, all the other problems encountered were due to local control mechanisms responding to our measurement attempts. If one could control or arrest them during the measurement interval, the dual channel approach might work to obtain a valid diastolic value.

One final note about the signal fluctuations. Numerous people have speculated that these fluctuations might be respiration related. Referring to Figure 35, one will note that the fluctuations occur over time periods as long as 35 seconds. Because of this, I doubt that the effect is directly related to respiration.

My general reaction to the work has been to gain an appreciation of the amount of knowledge we do not know about the functioning of the human system. Having reviewed the various techniques and ideas used or tried by others to obtain non-invasive blood pressure values, and having the experiences of my research, I believe that the solution to non-invasive pressure lies not with technological developments yet to be made, but rather with increased knowledge about the physiology of the human body.

REFERENCES
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APPENDIX 1

A ZENER REGULATED VOLTAGE REFERENCE FOR -1 VOLT

In the interest of those reading this thesis who have not had experience in circuitry design, this appendix should explain the steps in the design of a zener regulated voltage reference.

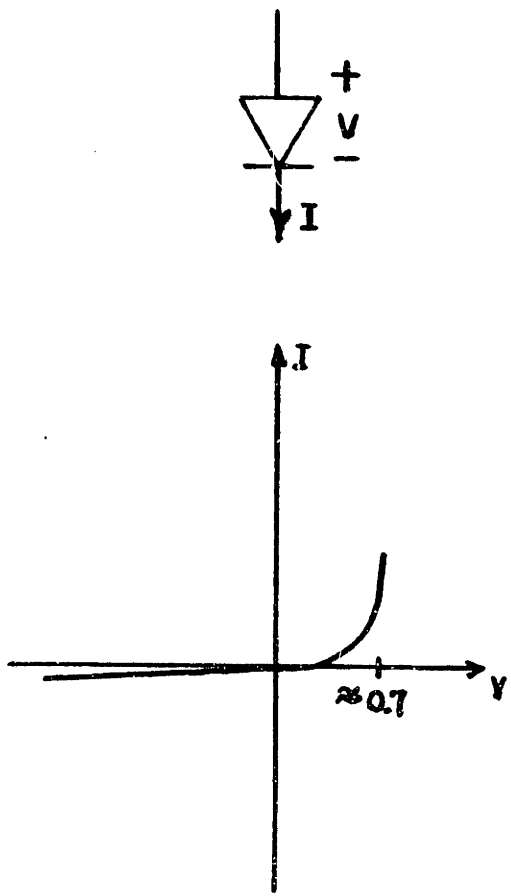
The current-voltage relationship for a regular diode is shown in Figure A1-1a. The relationship for a zener diode is shown in Figure A1-1b. The special aspect of a zener is its behavior for negative voltage. When wired in the configuration shown in Figure A1-1c, the zener's characteristics will hold the voltage below the resistor to Vz. Zeners are usually used in this reverse voltage configuration.

Theoretically, a zener with -Vz across it should be able to conduct infinite current. In practice, manufacturers will specify a value for current through the zener at which Vz is guaranteed. This current is usually labeled Iz, and the circuit designer selects other component values so as to have close to Iz amps passing through the zener most of the time.

There are four basic steps to the design. Refer to Figure A1-2a. First select a zener whose Vz is the desired voltage. Next, the maximum current to be drawn by the load, Il, must be calculated. Third, calculate the value for R:

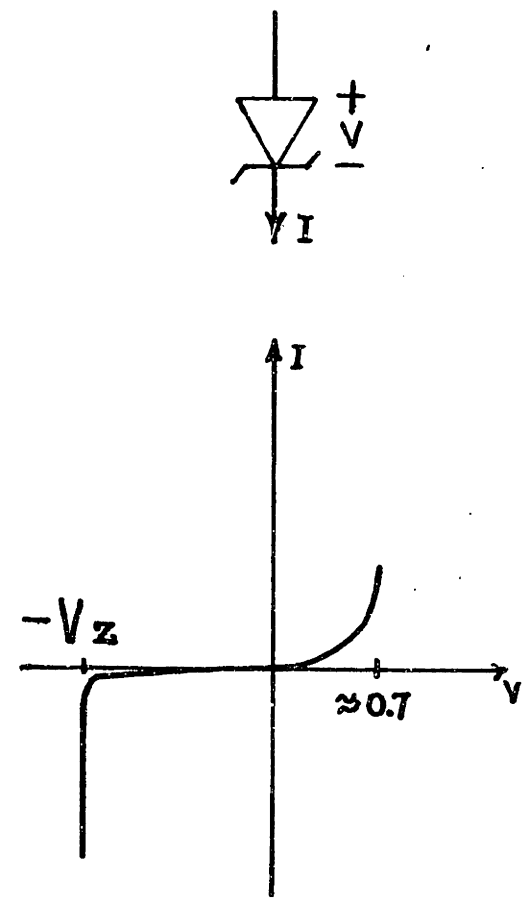
$$R = (Vs - Vz)/(Iz + Il)$$

for a standard diode



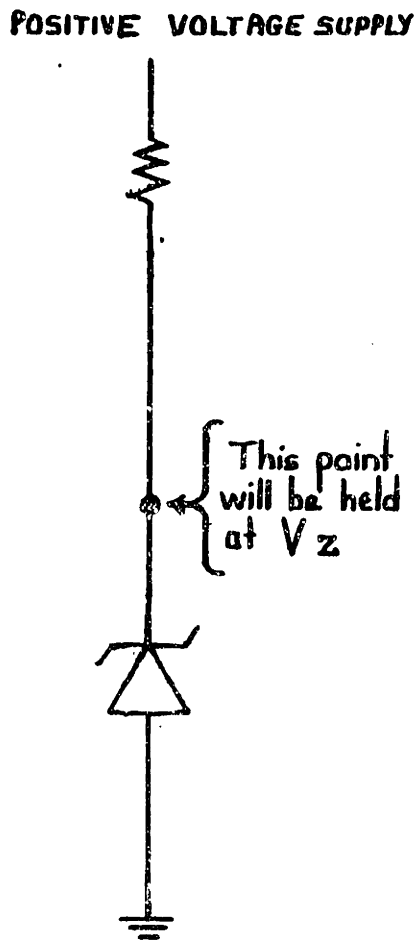
(a)

for a zener diode



Vz is set by manufacturer

(b)



(c)

FIGURE A1-1

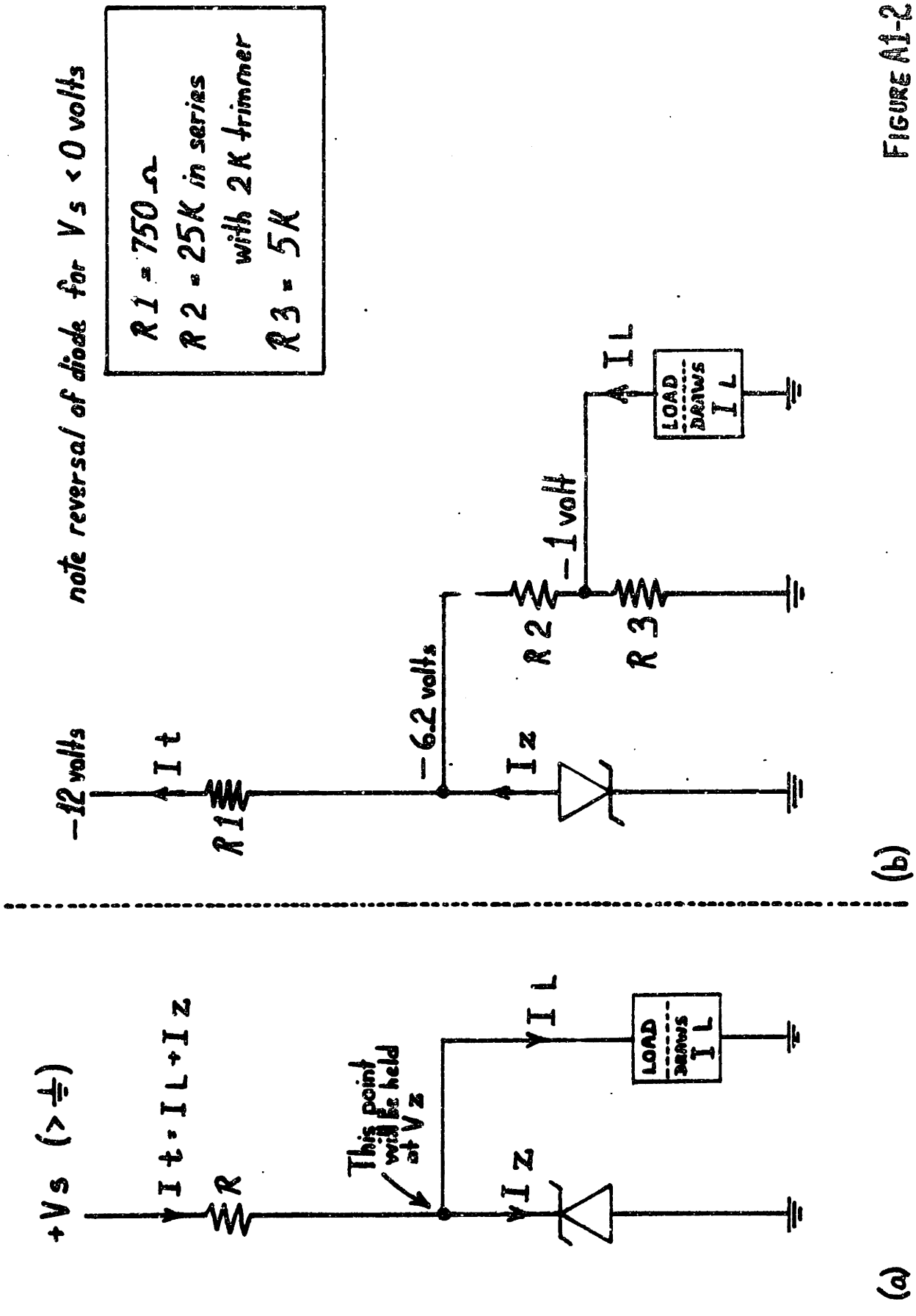

+Vs (> ±)
It = IL + Iz
R
This point will be held at Vz
Iz
IL
LOAD DRAWS IL
(a)
note reversal of diode for Vs < 0 volts
R1 = 750 Ω
R2 = 25K in series with 2K trimmer
R3 = 5K
-12 volts
It
R1
-6.2 volts
Iz
R2
-1 volt
R3
IL
LOAD DRAWS IL
(b)

FIGURE A1-2

The fourth step is a check. If by accident the load is disconnected, I1 which would normally travel to the load will also pass through the zener. The zener would now have to dissipate more power:

Power dissipated without load = Vz(Iz + I1)

One must select a zener whose maximum power dissipation capability exceeds the worst case value calculated above.

For the -1 volt reference needed for this research project, Vs equals -12 volts. I would select a 1 volt zener, but zener diodes below about 5 volts are unreliable and inaccurate. Consequently, a different value zener and a voltage divider are used. I chose a 6.2 volt zener because of its low cost and low temperature coefficient (JEDEC Number 1N827). The wiring configuration is shown in Figure A1-2b. Figure 21 shows that I1 will feed a transistor base and an op amp input. The collector current on the transistor will be less than 1 mamp. Assuming a beta of 100 for the transistor, the base current will be less than 0.01 mamps. Looking up the specs for the MC14385 op amp, the input impedance is greater than 300 Kohms. With the -1 volt reference, the input current will be less than 0.004 mamps. Total load current therefore will be less than 0.01 + 0.004 mamps:

I1 ~ 0.014 mamps

Referring to Figure A1-2b, the current through R2 and R3 should be large enough so as to make I1 negligible. Given this, a simple voltage divider relationship can be used to calculate the values for the resistors. A reasonable value is 0.2 mamps. which is greater than ten times I1. Solving for R2 and R3:

Voltage divider relationship: $[R3/(R2 + R3)]6.2 = 1$

Setting desired current: $6.2 \text{ volts}/(R2 + R3) = 0.2 \text{ ma}$

The resulting values are shown in Figure A1-2b. Now the value for R1 can be calculated:

$It = Iz + 0.2 \text{ mamps} + .014 \text{ mamps}$

$= 7.5 \text{ mamps for this diode} + 0.2 + .014$

$\sim 7.7 \text{ mamps}$

$R1 = (12 - 6.2)/7.7 \text{ mamps} \sim 750 \text{ ohms.}$

If the load were disconnected, the current flowing through the zener would rise from 7.5 mamps to 7.5 plus .014 mamps. This change is so insignificant that no power dissipation problems arise.

APPENDIX 2

PRESSURE SCALE "EXPANDER"
****

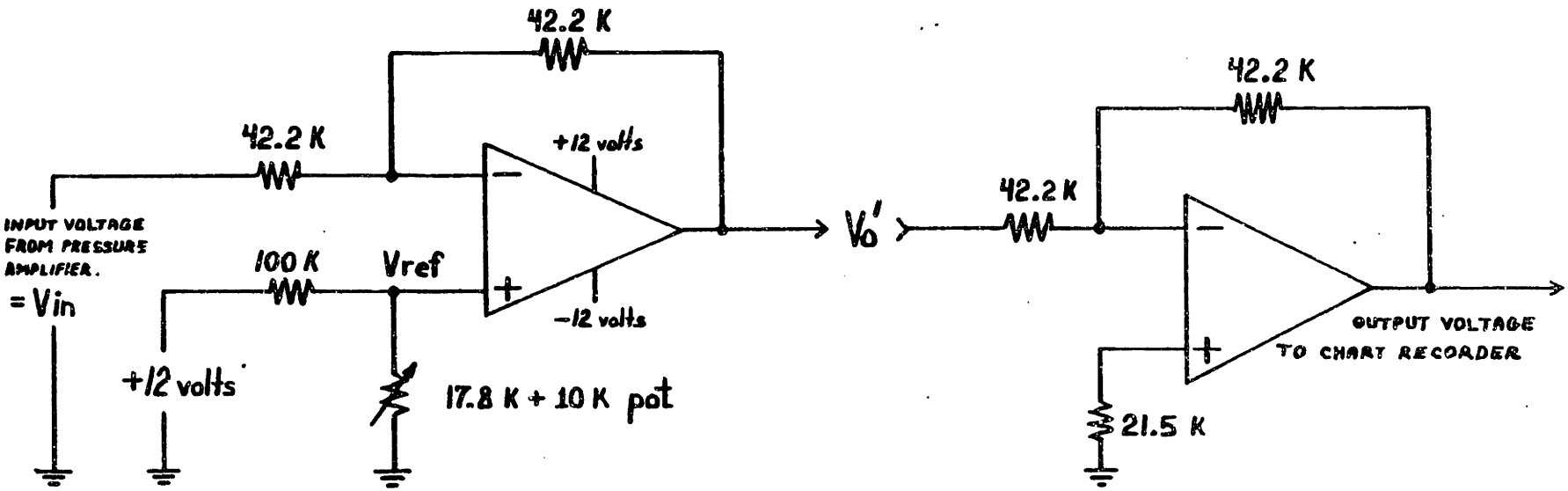
For most of the experiments, the recorder's gain was set so that full scale stylus deflection corresponded to occlusion cuff pressure of 140 mm Hg (Figure 30). For some systolic blood pressure measurements, it was necessary to be able to record pressures higher than 140. Readjusting the recorder so that the paper grid would display 0 to 160 mm Hg would decrease the accuracy with which the operator could discern the pressure value. The pressure scale "expander" modifies the recorder input so that the grid zero line represents 80 mm Hg and full scale corresponds to 160 mm Hg cuff pressure.

Initially, with zero volts input, one adjusts the chart recorder pressure channel so that the trace is on the zero line of the paper grid. On the 78172 recorder, this adjustment fixes a physical limit on the downward deflection of the stylus. Consequently, the trace will coincide with the grid zero line regardless of whether the input is ground or negative voltage.

For 160 mm Hg pressure into the i280 transducer, the 78205 pressure amplifier outputs 8.2 volts. Given a linear relationship between input pressure and output voltage, the pressure amp outputs 4.1 volts for 80 mm Hg pressure. By subtracting 4.1 volts from the pressure amp output, the recorder will see zero volts for 80 mm Hg, and 4.1 volts for 160 mm Hg. One adjusts the recorder gain so that 4.1 volts

input causes a full scale reading. The pressure channel now reads 80-160 mm Hg.

The "expander", shown in Figure A2-1, performs this subtraction. The 10K potentiometer provides for fine adjustment. The resistors that set Vref are chosen to satisfy the voltage divider relationship and to keep the current through them low (about 10E-4 amps). The 42.2 K value is used because it keeps the resistances seen looking out of the positive and negative op amp inputs approximately equal.



$V_o' = -V_{in} + 2\,V_{ref}$

Vref is set to be 2.05 volts

Output voltage to chart recorder is: **Vin − 4.1 volts**

FIGURE A2-1

APPENDIX 3
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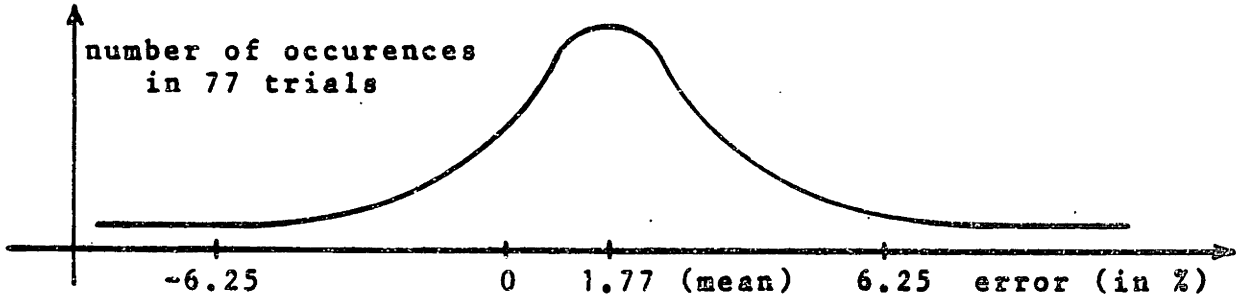
IF THE ERROR DISTRIBUTION IS FIT WITH A NORMAL (GAUSSIAN) CURVE,



THEN THE PROBABILITY THAT A TRIAL WILL RESULT IN AN ERROR WITHIN PLUS OR MINUS 6.25% IS GIVEN BY:

$$\text{Probability} = F\left[\frac{\text{upper limit - mean}}{\text{standard deviation}}\right] - F\left[\frac{\text{lower limit - mean}}{\text{standard deviation}}\right]$$

where F(z) is the normal distribution function:

$$F(z) = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{z} e^{-\frac{1}{2}t^2} dt$$

6.25% CORRESPONDS TO AN ERROR OF: 5 mm Hg at 80 mm Hg
10 mm Hg at 160 mm Hg

$$\text{Probability} = F\left[\frac{6.25 - 1.77}{4.15}\right] - F\left[\frac{-6.25 - 1.77}{4.15}\right]$$

$$= F(1.08) - F(-1.93)$$

$$= F(1.08) - 1 + F(1.93)$$

$$= 0.8599 - 1 + 0.9732$$

$$= 0.8331$$

THIS MEANS THAT THERE IS AN 0.1669 PROBABILITY THAT A TRIAL WILL RESULT IN AN ERROR GREATER THAN PLUS OR MINUS 6.25%