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NASA CR-86048

N 68 34548

PHYSIOLOGICAL MONITORING TECHNIQUE USING UNATTACHED SENSORS

By

Mylen Fitzwater, dePaul Corkhill, Earl Jackson,
Paul Halvorson, and Werner Sepper

March 1968

Prepared under Contract No. NAS 12-121
and Supplementary Agreement #1

PHILCO-FORD CORPORATION
Western Development Laboratories
Palo Alto, California 94303

Electronics Research Center
NATIONAL AERONAUTICS AND SPACE ADMINISTRATION

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ABSTRACT

This final report describes the work performed under Contract No. NAS 12-121. The objective of this work was to develop techniques using unattached sensors for measuring Lead I Electrocardiograph, Impedance respiration, Impedance pulse, Thoracic sounds, and Galvanic skin response. Brief definitions of these measures are as follows:

Lead I Electrocardiograph - Heart biopotential waveforms obtained from electrodes placed on each arm.

Impedance respiration - The electrical body impedance changes which produce a waveform as a function of thoracic volume changes during inspiration and expiration. Impedance measured from palm to palm.

Impedance pulse - The electrical body impedance changes which produce a waveform as a function of blood pulse related body volume changes. Impedance measured from palm to palm.

Thoracic sounds - Sounds both within and below the human auditory frequency range. These sounds were picked up by a microphone placed in the mid-dorsal position.

Galvanic skin response - Changes in body resistance measured by impressing a direct current from palm to palm. The response, which is normally psychophysiological, occurs after a short delay following an emotion producing stimulus, and persists for a few seconds.

Two unattached sensor systems for recording the above variables have been built. The system is described in detail.

Studies were performed to determine the measurements taken with the unattached sensor monitoring system compared with conventional methods. Feasibility studies were performed to determine the capability of deriving additional information from characteristics of the obtained measures; these include:

1) heart rate determination from ECG and pulse channels,
2) emotional impedance changes comparable to galvanic skin response,
3) myographic (muscle potential) level from ECG signals.

Results obtained using palm-to-palm dry unattached electrodes are comparable to conventional methods. However, further circuit design changes are planned based on validation studies recently performed in the University of Minnesota Biophysics Laboratories. The developed techniques may be applicable to monitoring astronauts.

SECTION 1
INTRODUCTION

In future extended man-space missions, it is planned that the crew members live and perform their intra-vehicle duties in a "shirt sleeve environment" i.e., without the encumbrance of wearing a space suit. Under conditions of a large space vehicle and crew member mobility it becomes advantageous to disencumber the personnel further by not requiring the body harness and electronics as required in current operations.

At present, attached electrodes of either the electrolyte paste or spray-on dry type are utilized. Under prolonged wear some skin reactions have been noted. In addition, electrodes have become loose and leads have broken interfering with, or ceasing measurements. Further, where body mounted electronics are utilized, the astronaut must make a direct wire plug-in to the space vehicle or make battery changes when biotelemetering is used.

The research and development described in this report was undertaken to ascertain the capability of obtaining physiological monitoring information with unattached sensors, palm-to-palm, with the following constraints:

- No attachments, straps, adhesives were permitted.
- Information was to be taken from the body without any special preparation or procedures, e.g., no paste or electrolytes, no abrasion of body surfaces.

The specific objective was to conduct research leading to the development of electronic instrumentation for electrocardiographic and body impedance change measurements without attaching or implanting sensors or electrodes. This effort was based upon modifying the basic breadboard technology previously developed by Philco-Ford under independent funds. This preliminary development was known as the instrumented monitoring chair and later, the MediScreen.

General philosophy of the undertaking was to minimize the number of sensing devices and to endeavor to achieve more than one output physiological measurement from a given body contact source.

SECTION 2

DESCRIPTION OF EQUIPMENT FOR THE UNATTACHED SENSOR SYSTEM

The design of the equipment described in this section was based on the following assumptions about the intended end use:

- The body is to be in casual contact with the sensors but none are to be attached.
- The number of sensors is to be kept at a minimum, with multiple physiological data sensed with single sensors, where possible, and with data separation taking place in the electronics.
- Sensors are to be fixed to a surface normally in contact with the subject during the course of his routine activity. If shirt-sleeved astronauts were to be monitored, sensors could be placed in his body couch or they could be placed on control levers, wheels, knobs, or other fixtures in the spacecraft which the astronaut normally handles. (An office side chair was used for this study since it conformed to the above requirements.)
- Only short-duration sampling is required to assess the health of the individual; therefore, one could afford to be satisfied with successful monitoring during only the short periods of time the subject would normally be in good contact with the sensors. (It was suggested that switching circuitry might later be developed to control only the recording of selected good data.)

The instrumentation developed prior to and during the study differed from conventional apparatus principally in the ability to separate multiplexed physiologic data from a minimum number of casual contact sensors. The use of single sensors for acquiring more than one parameter and the measurement of the electrocardiogram with unattached sensors were not new ideas. Geddes, et al, (Reference 1) previously reported the recording of respiration and ECG with common electrodes.

Chairs with salt water troughs on the arm rests were used in early electrocardiographic systems. The ECG bio-potential was sensed from the immersed forearm of the subject. Chair-mounted electrodes have been used previously in ECG screening studies (Reference 2). Since our study was performed, the Missouri Regional Medical Program Office announced their use of chair-mounted ECG electrodes (Reference 3).

To summarize, the system to be described here is unique because of its capability to separate a large number of physiological parameters detected by a minimum number of transducers in dry contact with the body.

2.1 GENERAL DESCRIPTION

The system for monitoring with unattached sensors consists of:

a. Two palmar dry electrodes and a microphone as sensors,

b. a set of solid-state modular signal conditioners designed to separate, filter, and amplify each channel of physiological information, and

c. a strip-chart recorder.

The system block diagram is shown in Figure 2-1.

The sensors and the signal conditioning package were mounted in a conventional office chair to demonstrate feasibility of the system. The modular electronics package and associated transducers could have been mounted in a reclining chair, dental chair, tilt table, or other convenient fixture which conformed to the anthropometric requirements.

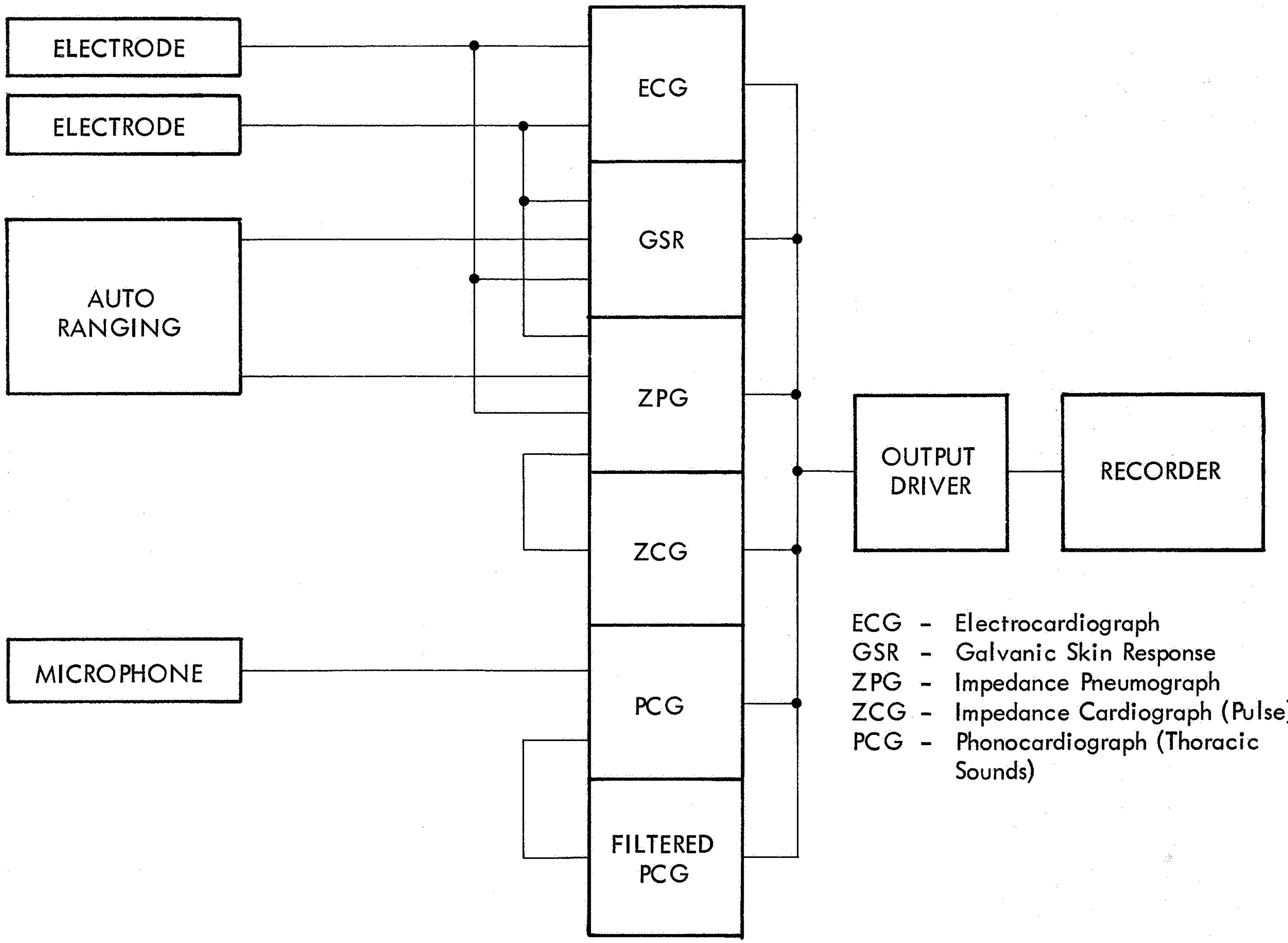

Figure 2-1 System Block Diagram

2.2 ELECTRODES

In a study performed by Philco-Ford prior to this contract, a number of electrode materials were compared to determine the best dry electrode material for this project. Although we were aware of the basic design problems for conventional electrodes (i.e., polarization, noise, etc.), the fact that we were interested only in short duration data allowed us certain freedom in selecting the material. (Electrode designers of ICU and CCU long-duration-monitoring-systems have not had this luxury.) The materials listed in Table I were tested:

TABLE I
RATING OF ELECTRODE MATERIALS

Electrode Material Tested	Rating Scale
1. Aluminum, braid	3
2. Aluminum, smooth	5
3. Chrome-plated copper	5
4. Conductive plastic	5
5. Copper, braid	1
6. Copper, smooth	2
7. Gold-plated, copper braid	1
8. Gold-plated, copper	2
9. Nickle-plated copper	5
10. Silver-bearing epoxy	4
11. Silver-plated copper	2
12. Silver-plated copper braid	1
13. Stainless steel	5
14. Wire bristles, steel	3

*Rating Scale: 1 - most desirable material
2 - good
3 - fair
4 - poor
5 - unusable

The materials listed in Table I were tested free of corrosion, oil, or surface dirt. Although materials such as copper were useful when clean, their susceptibility to an insulating layer of oxidation made them undesirable for long-life use. Gold-plated copper was selected for use in the described system because of its relatively high signal-to-noise ratio, relatively low base line drift (polarization), and immunity to corrosion.

Anthropometric considerations dictated mechanical placement of the electrodes. The electrodes were kept short to minimize RFI susceptibility and were placed within a comfortable reach of adults and most children.

Surface area contact required to produce adequate monitoring information has been found to be greater or equal to one square inch per palm. Electrodes adopted for the system provided at least six square inches of surface area per palm to assure that casual resting of the hands on the electrodes would exceed the minimum contact requirements. Figure 2-2 shows the gold-plated electrodes mounted on an office chair with the signal conditioning electronics and the microphone mounted in place.

See Section 3 for report of recent studies on electrical characteristics of electrode/skin interface.

2.3 SYSTEM ELECTRONICS

Physiological data sensed by the transducers falls into the following categories:

a.	Bio-potential:	ECG Electrocardiogram
b.	Bio-resistance:	GSR - low frequency changes of skin resistance BSR - baseline skin resistance (d. c)
c.	Bio-impedance:	ZPG - low frequency changes in impedance respiration measured at 100 kHz ZCG - 0.8 to 5-Hz changes in impedance due to central blood pulse measured at 100-kHz
d.	Thoracic Sounds:	PCG - sounds in the 20-2000-Hz range FILTERED PCG - sounds below 20-Hz range

Separation, filtering, and amplification of these data are described in the following paragraphs.

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Figure 2-2 Medical Monitoring Chair with Gold Plated Electrodes

2.3.1 Electrocardiograph Channel

To prove the feasibility of obtaining an electrocardiogram (Lead I) from unattached electrodes, the configuration shown in Figure 2-3 was developed. The electrodes were stainless steel and the amplifier was selected for high performance characteristics. Work was initiated to design a small, low cost amplifier that would operate with a two-electrode system. Lack of sufficient common mode noise (60 Hz) rejection and gain limitations prevented successful operation of most off-the-shelf modular amplifiers.

With the advent of low cost, high performance operational amplifiers, a circuit was developed that satisfied the initial requirements. (Refer to Figure 2-4.) An obvious feature of this circuit is its low input impedance. Thus, to provide the necessary low frequency response, very large input capacitors had to be used. This circuit was advantageous for a two electrode (no ground) ECG in that it reduced 60 Hz noise to an acceptable level by loading common mode noise to ground. High differential gain in the amplifier returned the signal to the desired output level. Although the interamplifier bandwidth (.15 - 100 Hz) was wide enough, the following standard objections to this unconventional design remained to be solved:

- Effect of skin/electrode contact resistance changes on calibration.
- Effect of baseline drift due to leakage of the large input capacitors.
- Effect of baseline drift due to electrode polarization.
- Effect of distortion due to skin/electrode capacitance coupling.

To provide an ECG with a two-electrode system comparable to a standard clinical ECG, the above basic problems had to be solved. Electrode polarization was considered to be less of a problem since only short term applications were planned. Since the described system amplifies only one ECG lead, the conventional problem of multilead bridge loading was not applicable. The effects of variations in contact resistance, and subject source resistances was eliminated by using a series 1-millivolt calibration. This arrangment required that the subject be

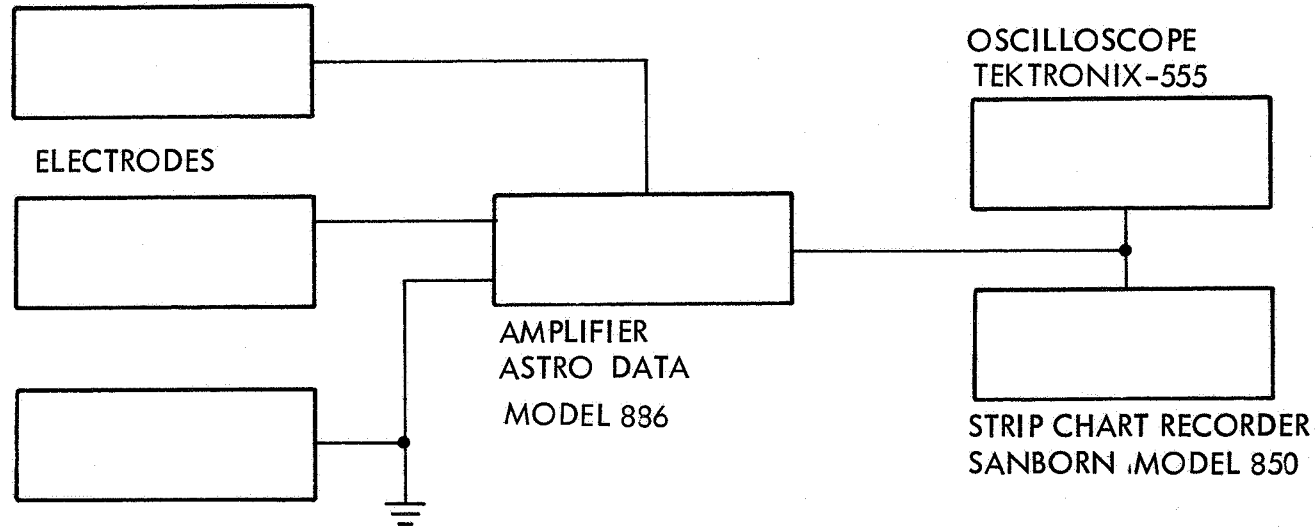


Figure 2-3. Electrocardiograph Block Diagram

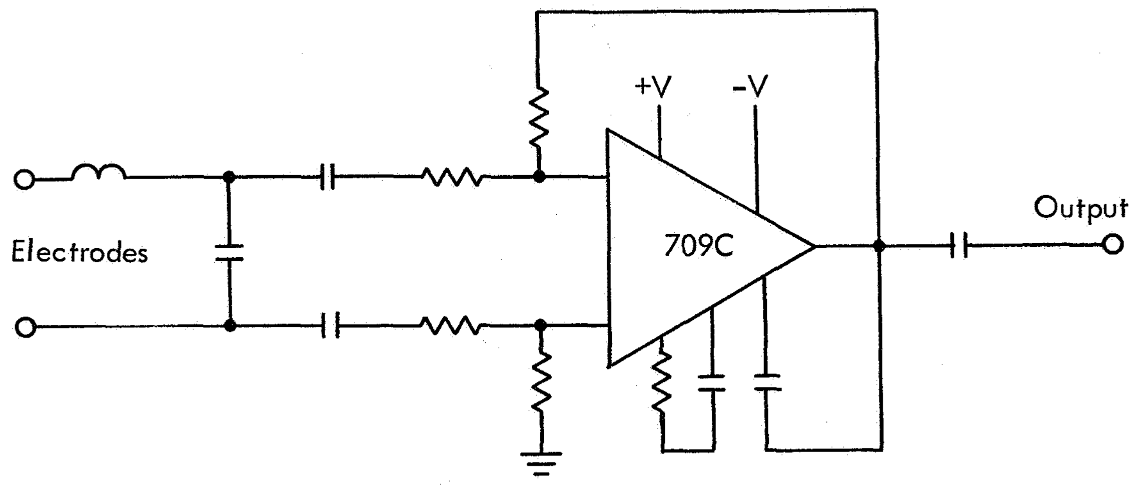


Figure 2-4. First Operational Amplifier Configuration

in contact with the electrodes for calibration, but it allowed the system gain to be adjusted to give a standard calibration (1-mV = 2 cm) recorder deflection. This calibration is also a contact resistance compensation adjustment.

The low frequency response was obtained through the design of a three-stage amplifier. The first stage is a buffer amplifier that is direct coupled for good low frequency response, incorporates low input impedance with high common mode rejection to eliminate noise, and has low gain (30 dB) to keep the amplifier from saturating. The second stage is a field effect transistor (FET) source follower that has high input impedance to allow the use of low value capacitors and still maintain good low frequency response. This stage has a low output impedance to match the input impedance of the third stage. The third stage is a standard operational amplifier with a gain of 1200. The circuit diagram is shown in Figure 2-5.

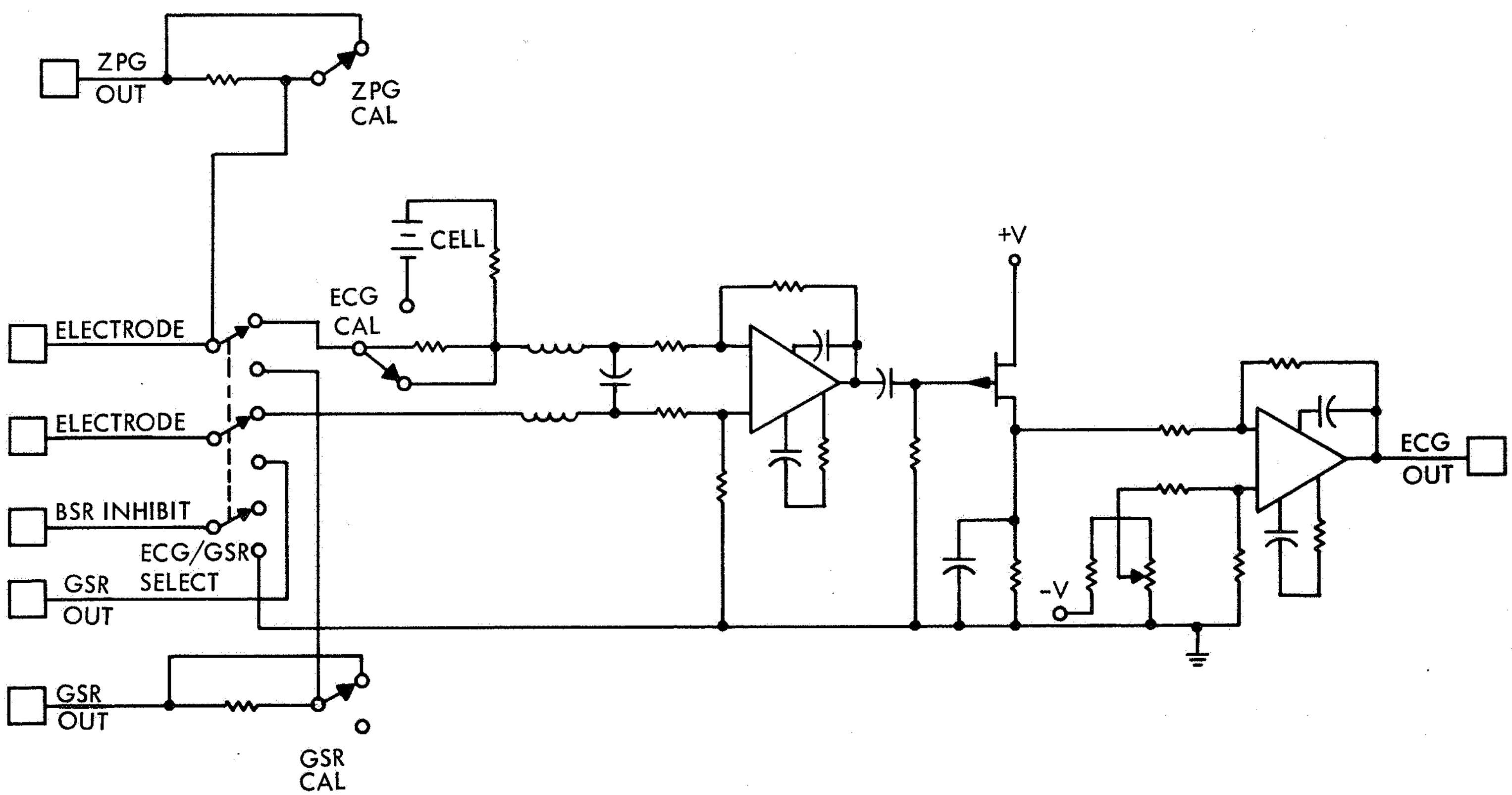


Figure 2-5 Present ECG Circuit

Recent validation studies conducted in the Biophysics Laboratories of the University of Minnesota have shown the electrode/skin interface to be much more capacitive than was assumed during the course of this reported study. Although it is difficult to visually detect distortion in the clinical data presented in Sections 3 and 4 of this report, the fact that computer analyses show significant capacitance at the electrode interface suggests that distortion is theoretically possible with the described circuit. To assure elimination of phase distortion over the entire 0.15-100 Hz data band, high input impedance will be required. Rather than relying entirely on common mode rejection, as was the case with high impedance amplifiers used early in the study, a design employing the "driven body" principle is being developed for future equipment.

2.3.2 Galvanic Skin Response

To monitor additional parameters from the same electrodes, a galvanic skin response (GSR) channel was added to the system. This channel in conjunction with ZPG, ZCG, and peripheral pulse measurements was to form the basic unit of a psycho-physiological monitoring system. The first GSR circuit (Figure 2-6) was a battery-powered bridge with manual nulling. Although this configuration produced good results, the system required constant operator attention. In addition it was found that, because of the dc voltage applied to the electrodes by the GSR bridge, ECG could not be operated simultaneously.

Design objectives for the GSR circuit included a bridge power source that would not require periodic replacement, automatic bridge nulling, an output signal for the baseline value, and a constant dc current source. The requirement for a constant current was eliminated first when it was found that the electrode area variance from subject to subject made measuring and maintaining a constant current impractical. Because of this problem, a constant voltage bridge was employed (Figure 2-7). The use of diodes to provide the constant voltage also eliminated the need for a battery. To achieve automatic bridge nulling, thermistor and field effect transistor (FET) feedback circuits were tried. With appropriate feedback and time constants, these circuits provided adequate nulling, but over only a very narrow range.

With the advent of the autoranging circuit (see paragraph 2.3.4), the capability was available for operating of the constant voltage bridge anywhere along its range without saturating the amplifier. This circuit incorporates calibrated range with a series calibration pulse for the signal within each range. The autoranging circuit also produces signals that, when summed, provide baseline resistance (BSR) as shown in Figure 2-8.

The present GSR circuit consists of a diode bridge, a range zero adjustment, and a summing amplifier, seen schematically in Figure 2-9.

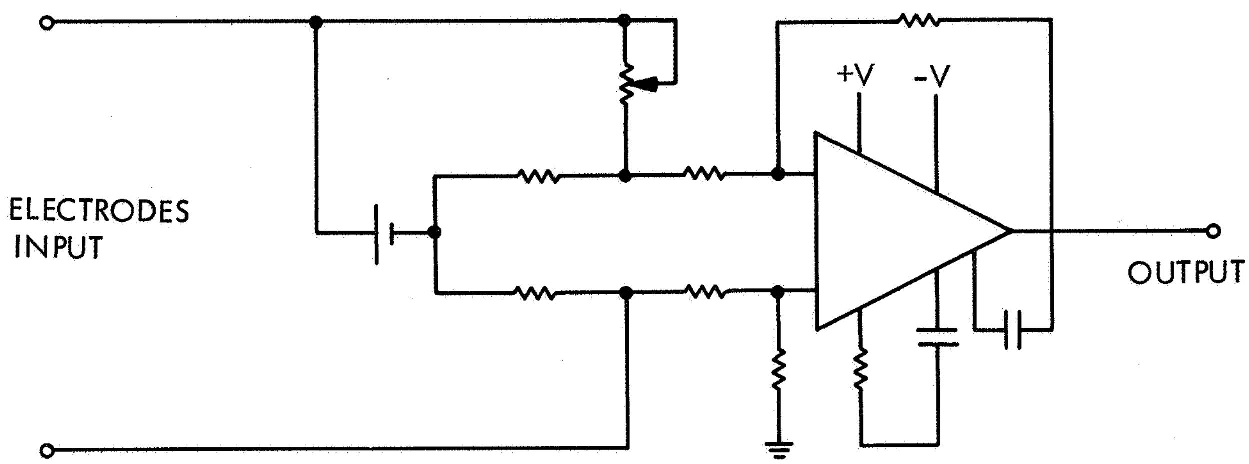


Figure 2-6. First GSR Circuit

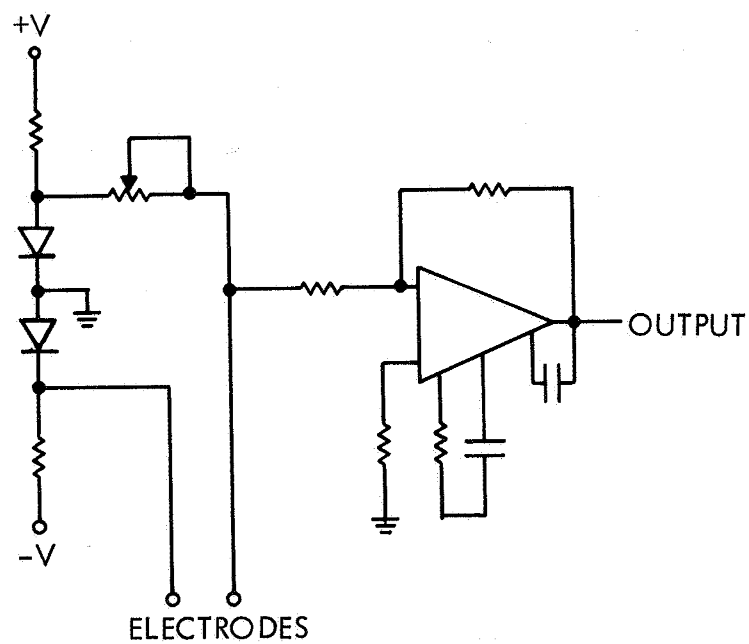


Figure 2-7. Constant Voltage GSR Bridge and Amplifier

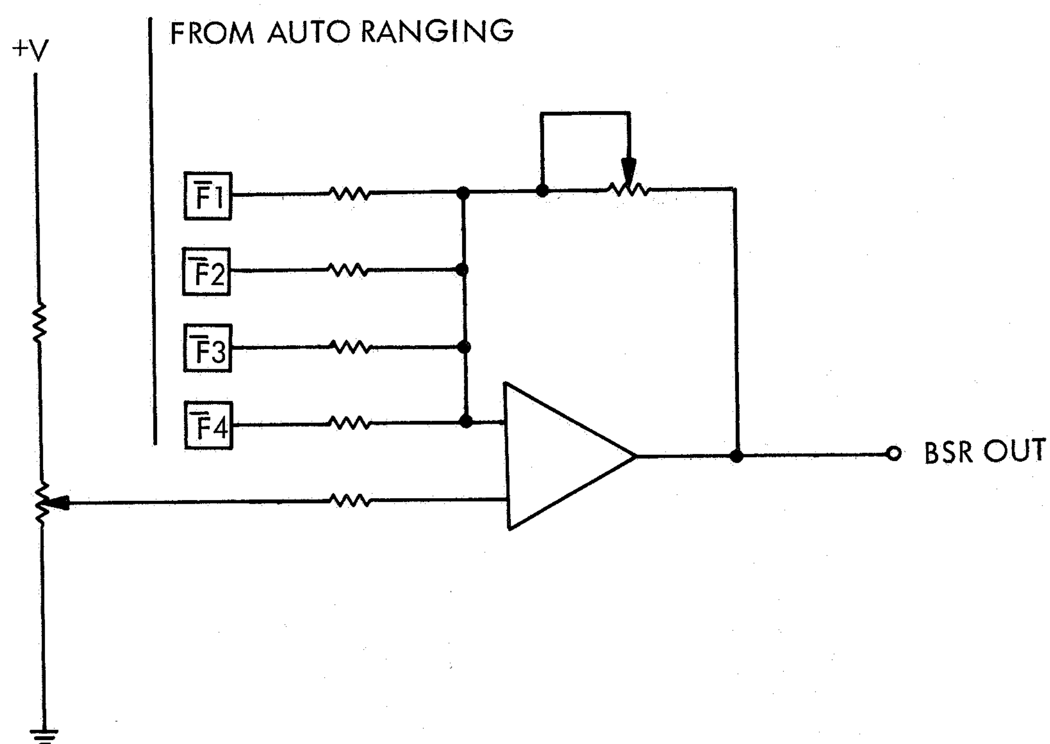


Figure 2-8. BSR Circuit

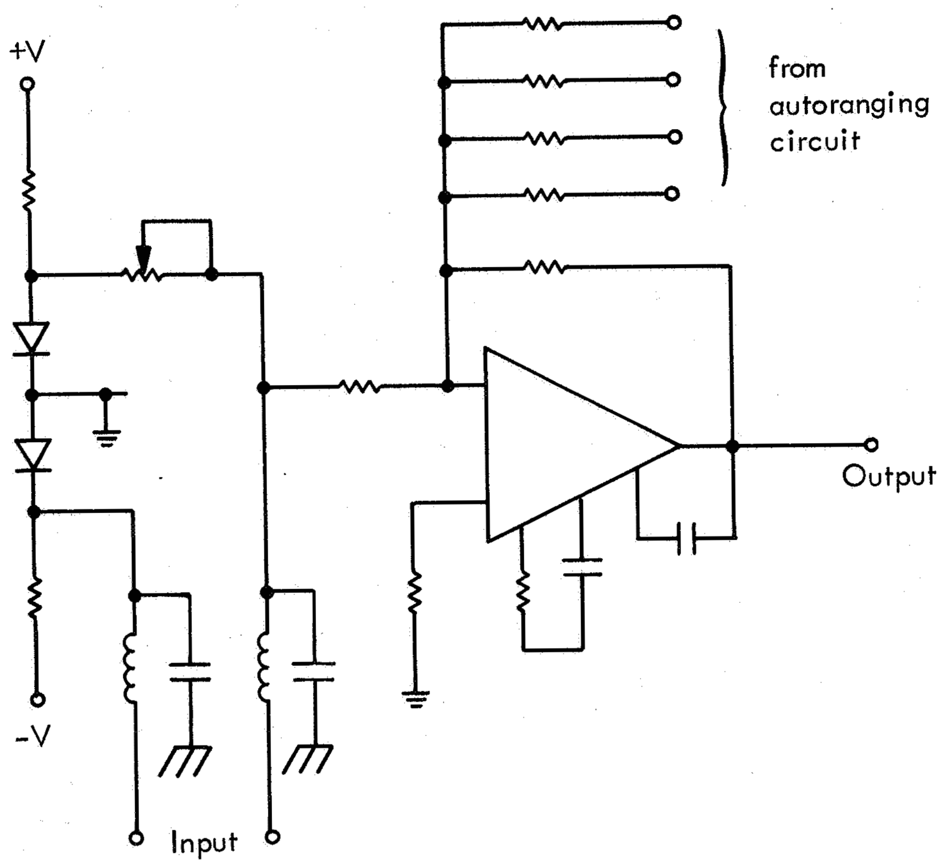


Figure 2-9. Present GSR Circuit

2.3.3 Impedance Respiration and Pulse

The operation of an impedance pneumograph (ZPG) and an ECG amplifier from a single pair of attached electrodes had been demonstrated (Reference 1). To achieve similar results using unattached electrodes, a commercial ZPG unit was purchased and connected as shown in Figure 2-10. This arrangement served the function; with proper filtering and gain, a pulse type waveform was obtained. This system, using unattached sensors, was limited by the following:

- Excessive Baseline Drift
- Inadequate Range
- Manual Adjustment

A program was initiated to develop a ZPG that would eliminate range and drift problems and that would require no manual adjustment. The first problem resolved was that of inadequate range, accomplished by use of a free running multivibrator for the energizing source. (See Figure 2-11.)

The problems of drift and manual adjustments were eliminated by the use of an automatically offset operational amplifier (autoranging), discussed in paragraph 2.3.4. The square wave unit provided more than adequate range; however it also produced distortion on the pulse waveform (ZCG) output. It was determined that this distortion could be eliminated only by use of a single frequency energizing source. The first sine wave ZPG (Figure 2-12) required an operational amplifier oscillator followed by a transistor driver and transformer-coupled electrodes. Operated with the autoranging capability, this configuration provided excellent recordings of ZPG.

Following this development, the circuit was redesigned to reduce the parts count, hence the cost. The present configuration (Figure 2-13) consists of a transistor inductor capacitor (LC) oscillator, a driver for the electrodes, a detector, a dc level shifter, and a summing amplifier. The oscillator is set at 100 kHz, the frequency selected for the best sensitivity with both the respiration and the pulse type waveforms. The driver is used to reduce the load on the oscillator and to provide range.

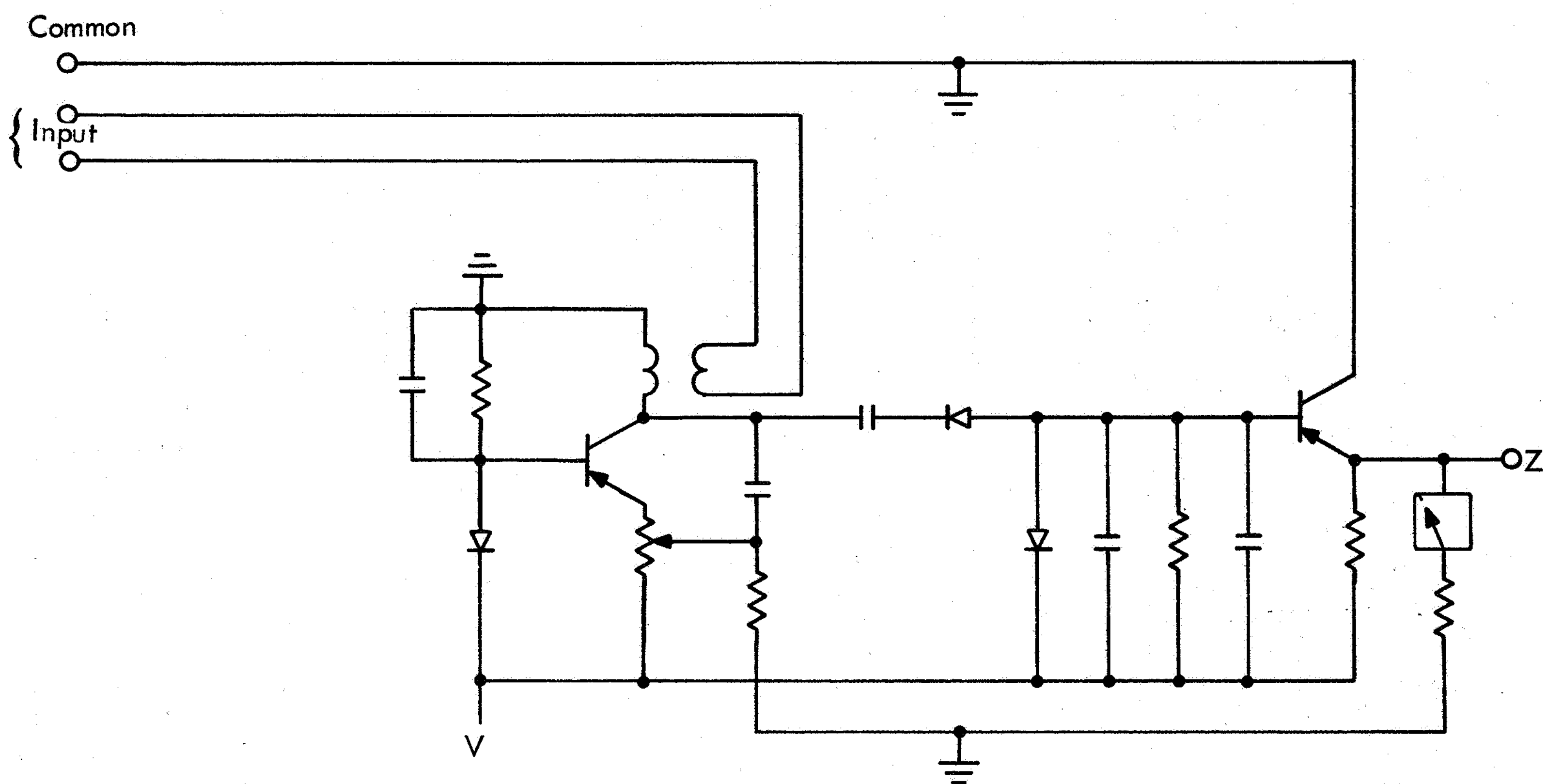


Figure 2-10. Commercial Impedance Pneumograph

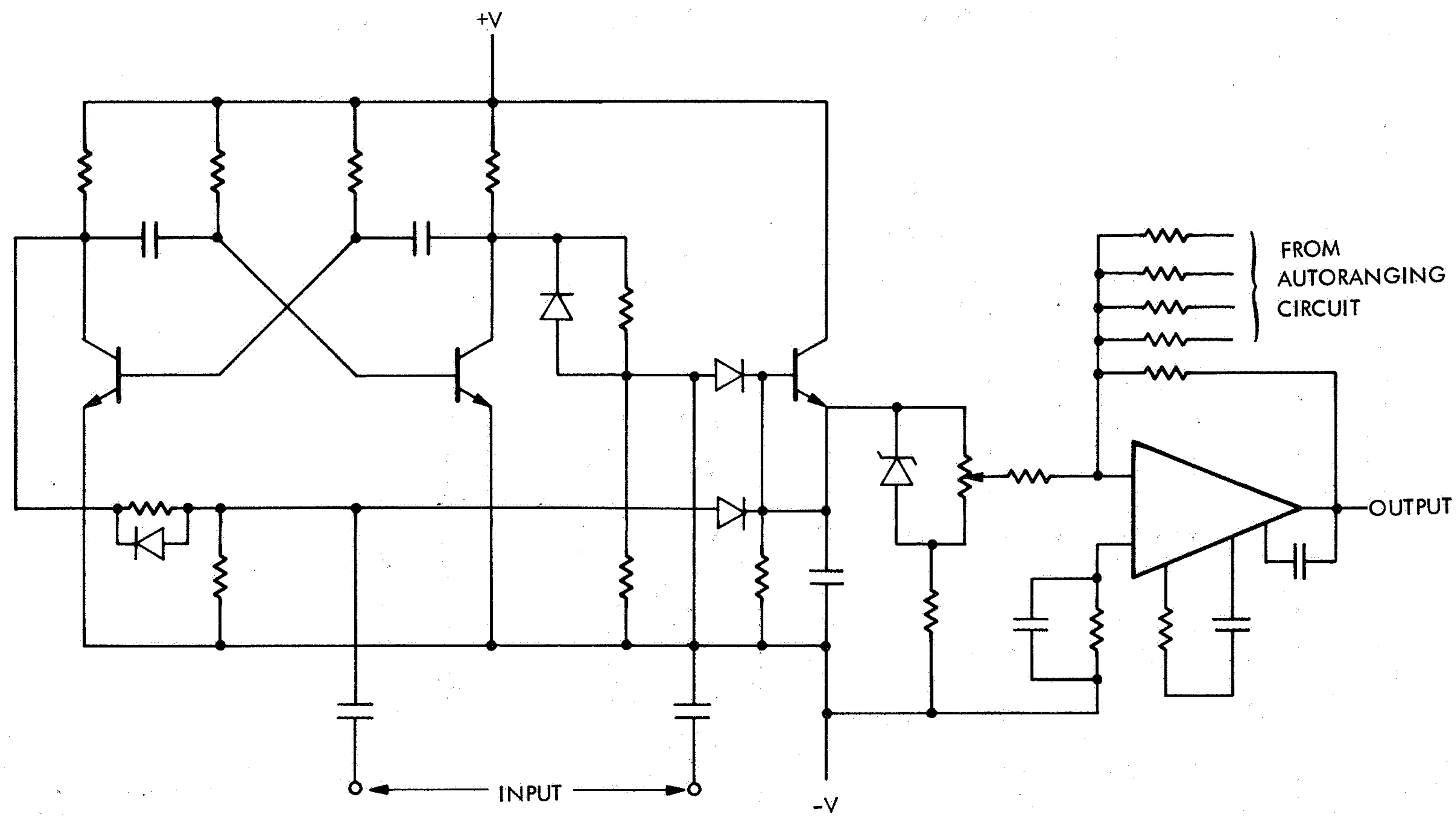


Figure 2-11 ZPG Using Square Wave Energizing Source

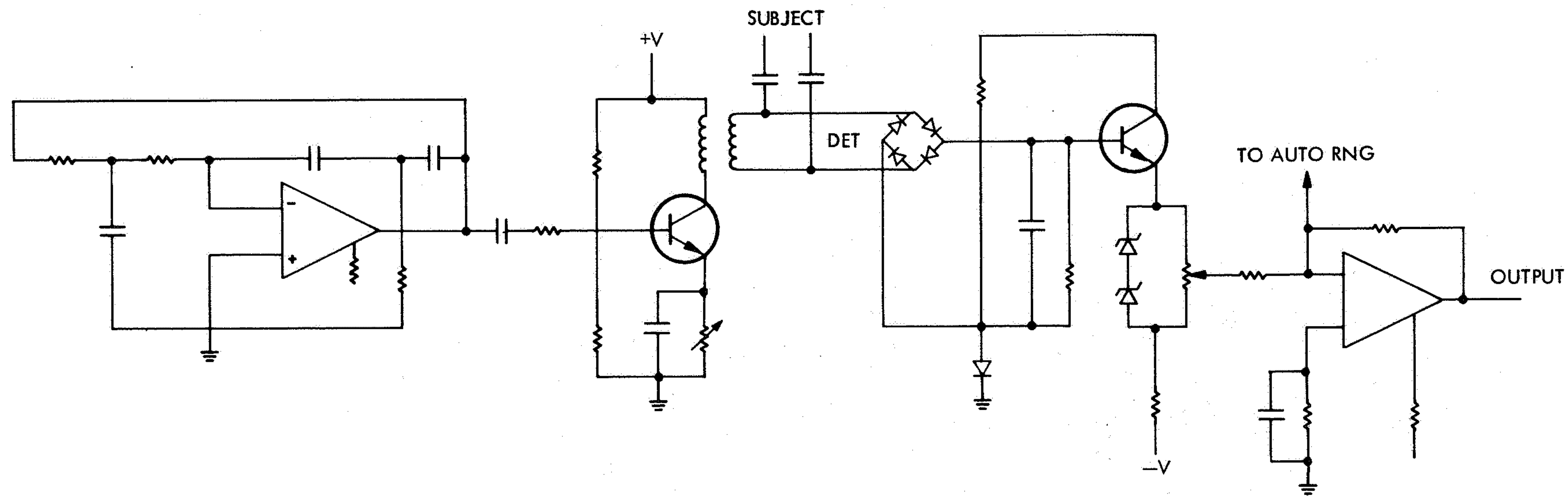


Figure 2-12 Sine Wave ZPG Circuit

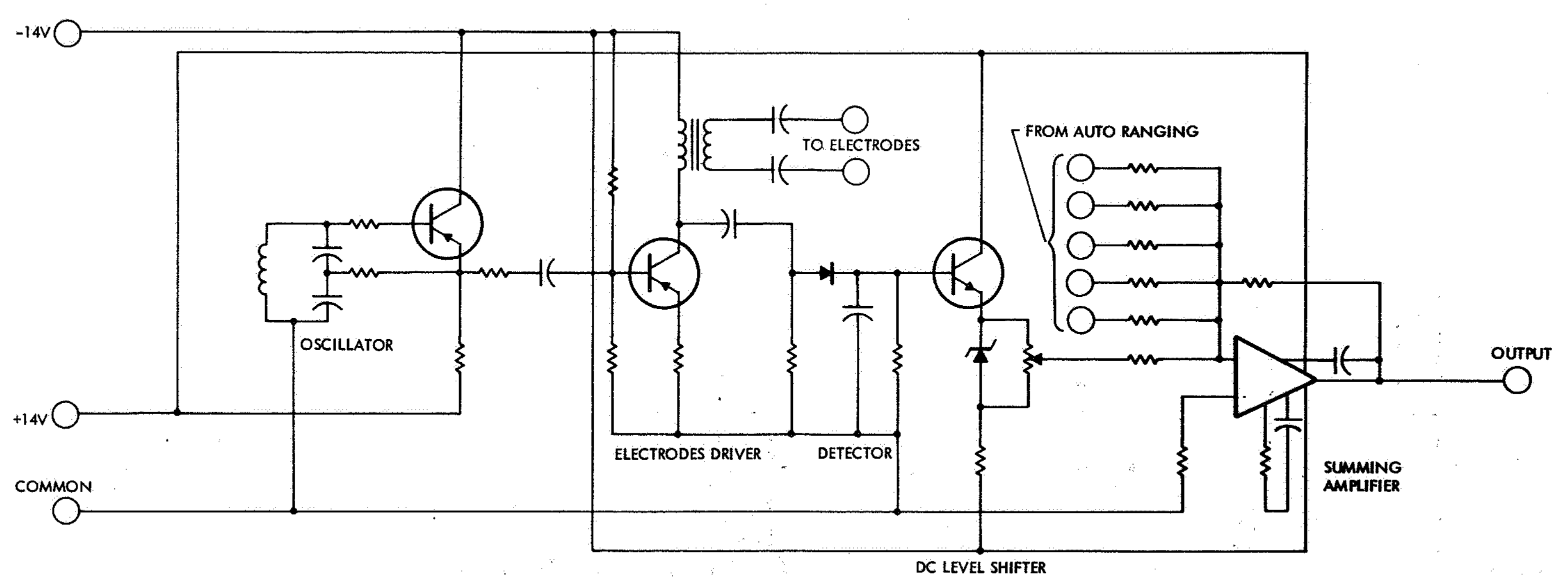


Figure 2-13 Present ZPG Circuit

The sensitivity of the system is high enough to use a half wave instead of the full wave detector that had been used in previous systems. The dc level shifter is set to provide a zero Vdc level to the summing amplifier at one end of the operational range. The summing amplifier provides first, amplification for the signal; and second, a summing point for the dc level of the signal and the output of the autoranging.

2.3.4 Autoranging Circuit

A major problem of a system using unattached sensors, no skin preparation, and no conductive paste is baseline drift. This problem is most pronounced in the ZPG and GSR circuits. Various methods of analog nulling were tried, but because of the low frequency of the signals (dc to 0.8 Hz), analog nulling was inadequate.

The problem was solved by using an operational amplifier to amplify both the signal and drift. With the operational amplifier in the inverting mode, a dc voltage can be applied to the summing junction any time the operational amplifier reaches saturation to offset it back into operating range. This method allows the use of a digital circuit to provide the offset voltage. The first circuit tried was constructed of discrete components and consisted of two Schmitt triggers, (Figure 2-14a) to sense positive and negative saturation, a four-bit up-down counter (Figure 2-14b) to provide 15 offset voltages, a bipolar transistor clock circuit, and a +5 Vdc regulator circuit (Figure 2-14c).

To facilitate construction, the circuit was redesigned with a number of improvements incorporated. Figure 2-15 shows the improved autoranging circuit. The 4-bit up-down counter was replaced by a 5-bit down counter constructed of integrated circuits. This alteration increased the number of offset voltages from 15 to 31. Additionally, by counting in only one direction the gate circuits were eliminated, greatly simplifying the circuitry. Counting in one direction only was no disadvantage since the counter could cycle in a few milliseconds. The loss of milliseconds of data is insignificant. Additionally, the Schmitt trigger circuits were replaced by single transistor switches. The +5 V regulator was simplified to a single-transistor circuit and the bipolar transistor clock was replaced by a unijunction transistor clock circuit to reduce the parts count, the construction time, and the cost.

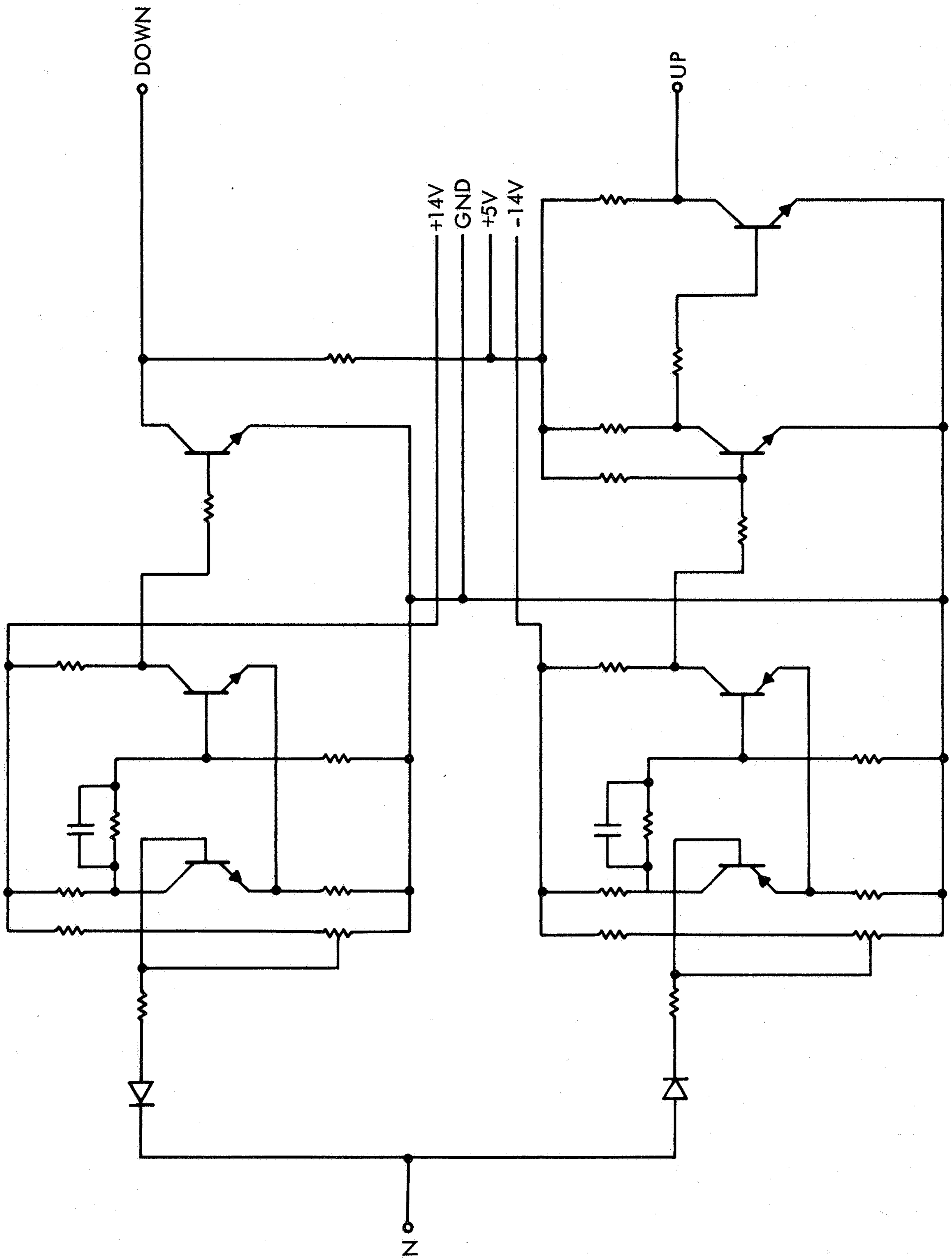


Figure 2-14a Schmitt Trigger Circuit

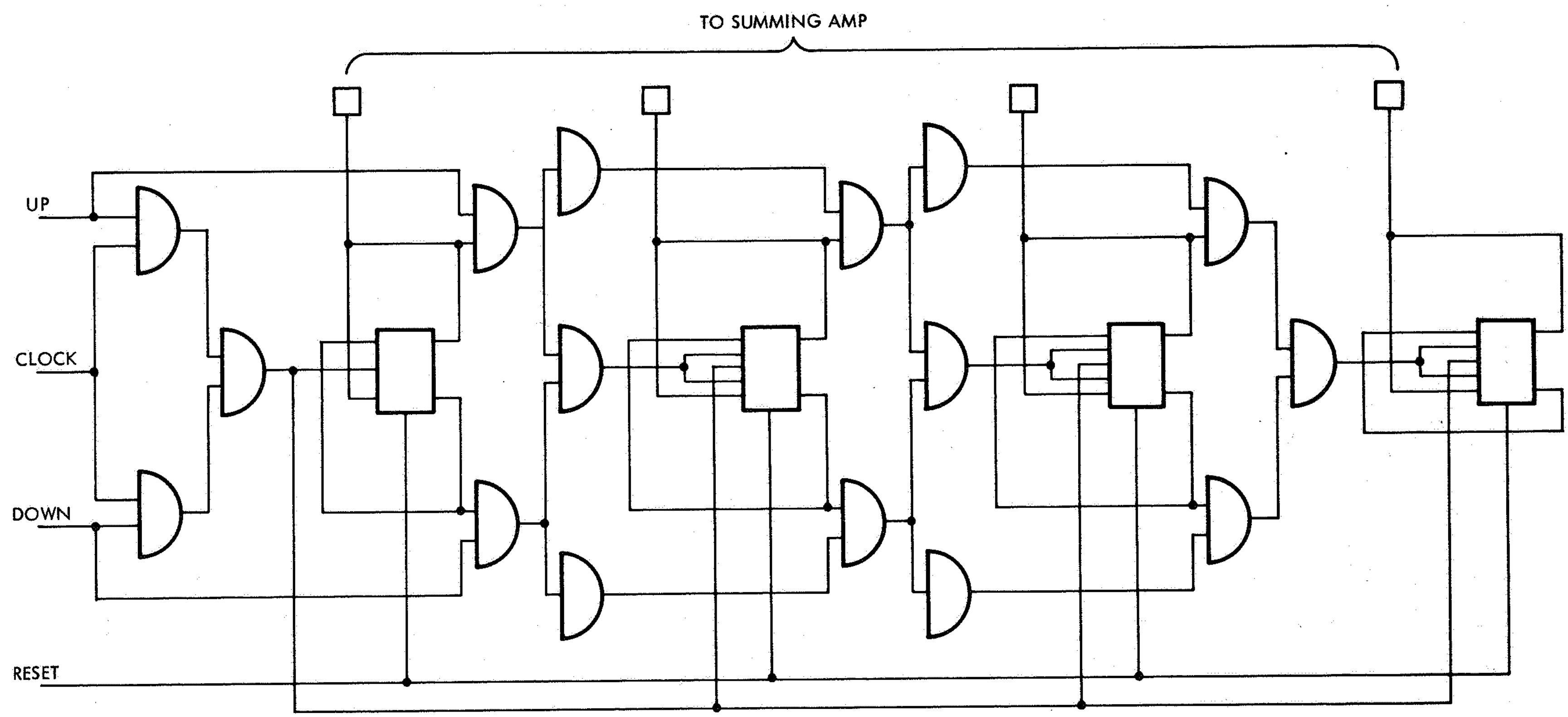
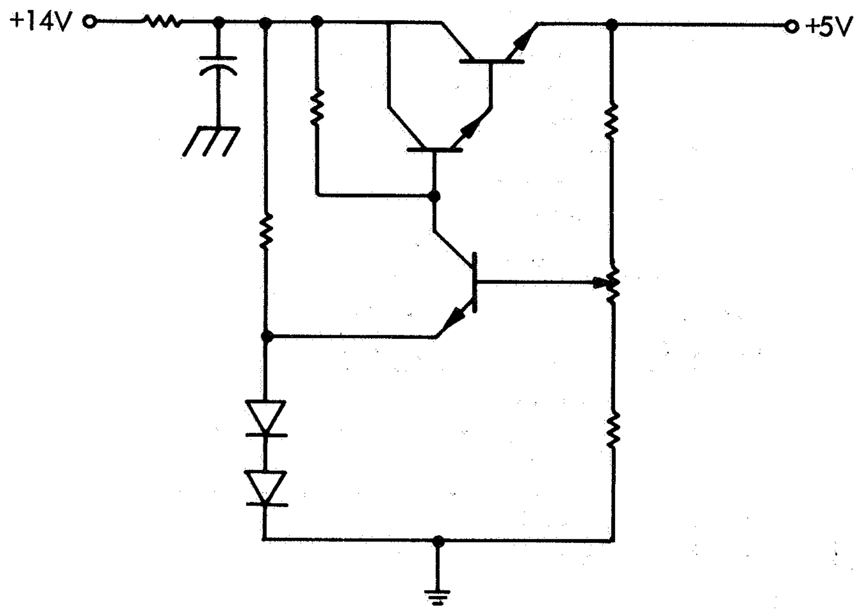
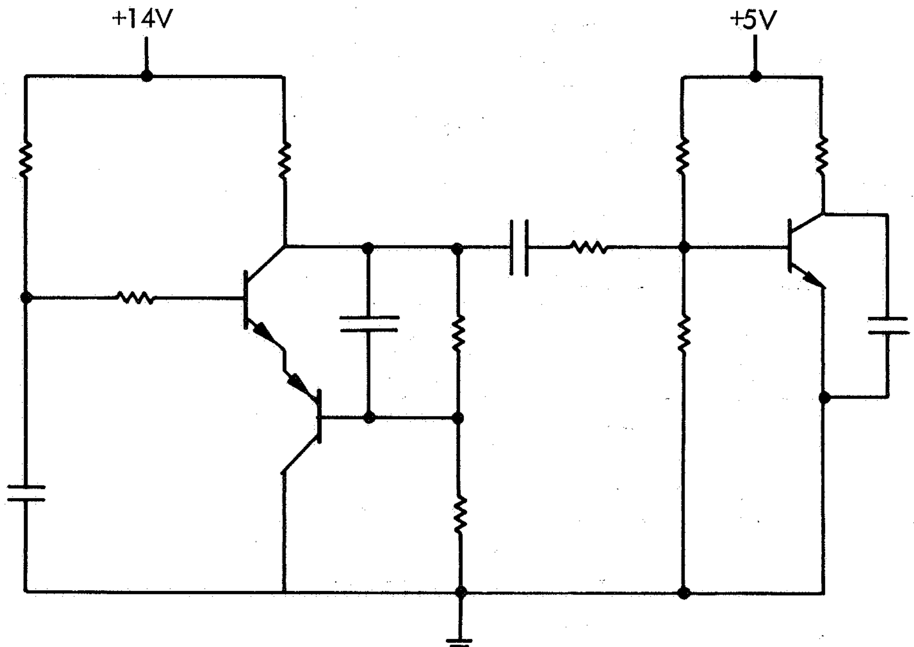


Figure 2-14b 4-Bit Up-Down Counter



5V REGULATOR



CLOCK CIRCUIT

Figure 2-14c Regulator and Clock Circuit

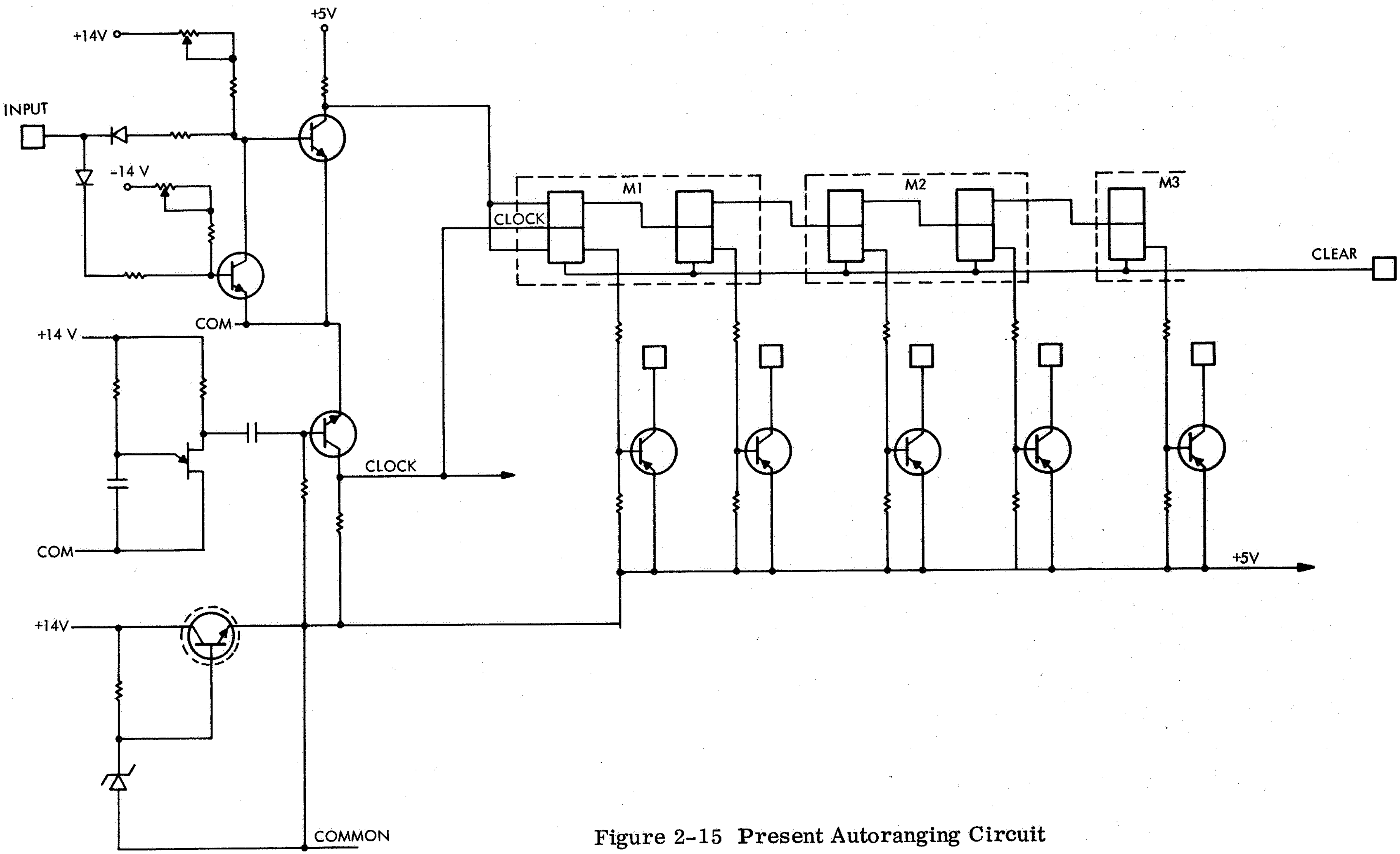


Figure 2-15 Present Autoranging Circuit

2.3.5 Thoracic Sound Channels

A microphone (crystal type) was mounted in the back of the chair to detect thoracic sounds. The first configuration consisted of the microphone, (hard mounted to the chair), a lab instrumentation amplifier, and a recorder. Superimposed on the signals from this configuration was noise picked up on the MIC cable. In addition, the signal level was dependent on the pressure applied by the subject. The lab amp was replaced by an integrated circuit amp (see Figure 2-16) and the microphone was mounted with foam rubber. This modification eliminated dependency on body pressure, but the foam rubber allowed the microphone to become canted, preventing good contact. A mount was designed using spring loading to keep the MIC straight and also to keep a constant pressure on the subject. To eliminate the noise, an FET pre-amplifier was mounted inside the microphone mount. To get the best signal, the location of the microphone was critical and no one location was found to give satisfactory results with all subjects.

Anthropometrically, the chair frame did not provide enough flexibility for wide ranges of body shapes and sizes. The thoracic sound channels were most affected by this problem. To accommodate both children and adults, it may be necessary, in the future, to provide several microphones located at different levels over the back and switched in according to the subject's body size. Alternatively, the chair back could be designed for easy access to the microphone so that alternate microphone placement positions could provide the necessary adjustability for anthropometric differences.

It now appears that the data detected with the microphone mounted in the chair back is useful only for timing information. For heart sounds, better results would be obtained by using an external microphone applied at standard monitoring positions.

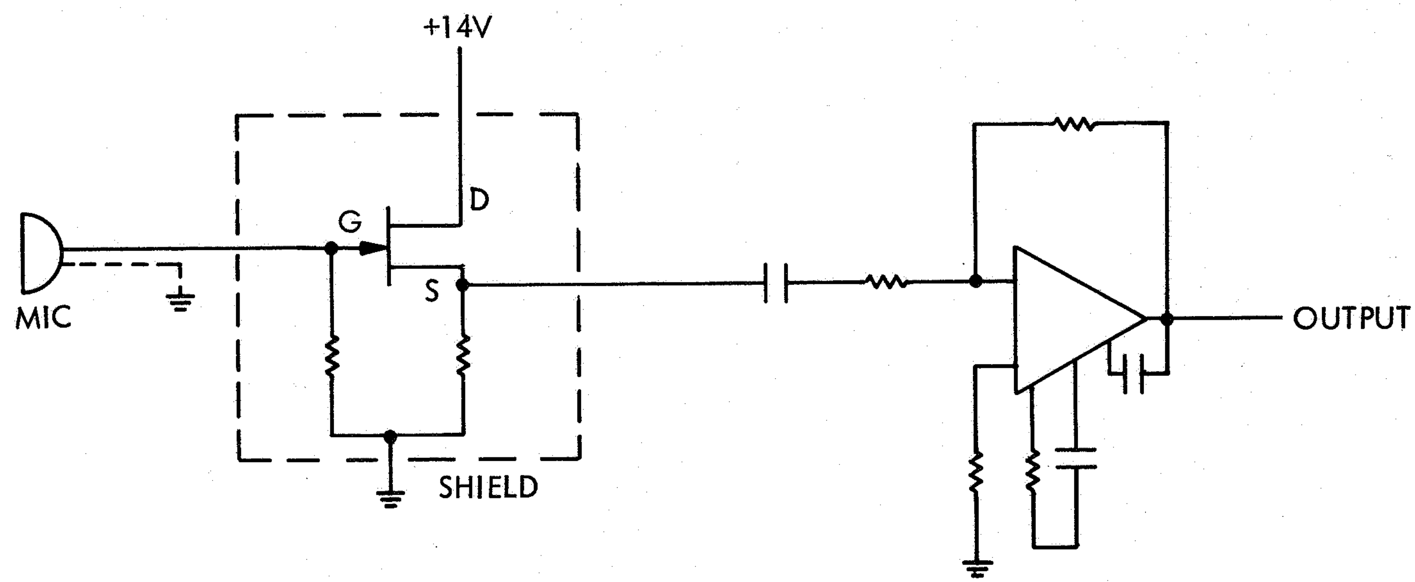


Figure 2-16. Thoracic Sound Circuit

2.3.6 Additional Data Output

The following additional time related data can easily be extracted from the analog information processed by the above signal conditioners:

a. Heart rate (ECG R-R interval/time)
b. Respiration rate (ZPG breath-breath interval/time)
c. Pulse wave velocity (PCG first heart sound - peripheral pulse event/time)
d. Electromechanical delay (ECG R-wave - PCG first heart sound time interval)

A simultaneous measurement of ECG and respiration provides a unique opportunity to eliminate the normal sinus arythmia from the beat-by-beat heart rate measurement. The respiration waveform can be used to trigger the measurement of the ECG R-R interval for measurement of instantaneous heart rate.

2.4 STRIP CHART RECORDER

The standard four-channel recorder is of the heated-pen type (Figure 2-17). The top row of control knobs is used for selecting the appropriate recorder channel. With ten positions for each channel, forty types of data may be recorded, permitting any four to be presented simultaneously. The second row of knobs controls the position of the pen within its paper channel; the bottom row controls pen heat. Each column of four controls is paired with a recording pen--in order from left to right. The recorder is manufactured by Mechanics for Electronics, Inc., Cambridge, Mass. (Recorder Model 4M-3, amplifier Model 655, power supply Model M-11.) These components were repackaged by Philco-Ford with added function-select switches.

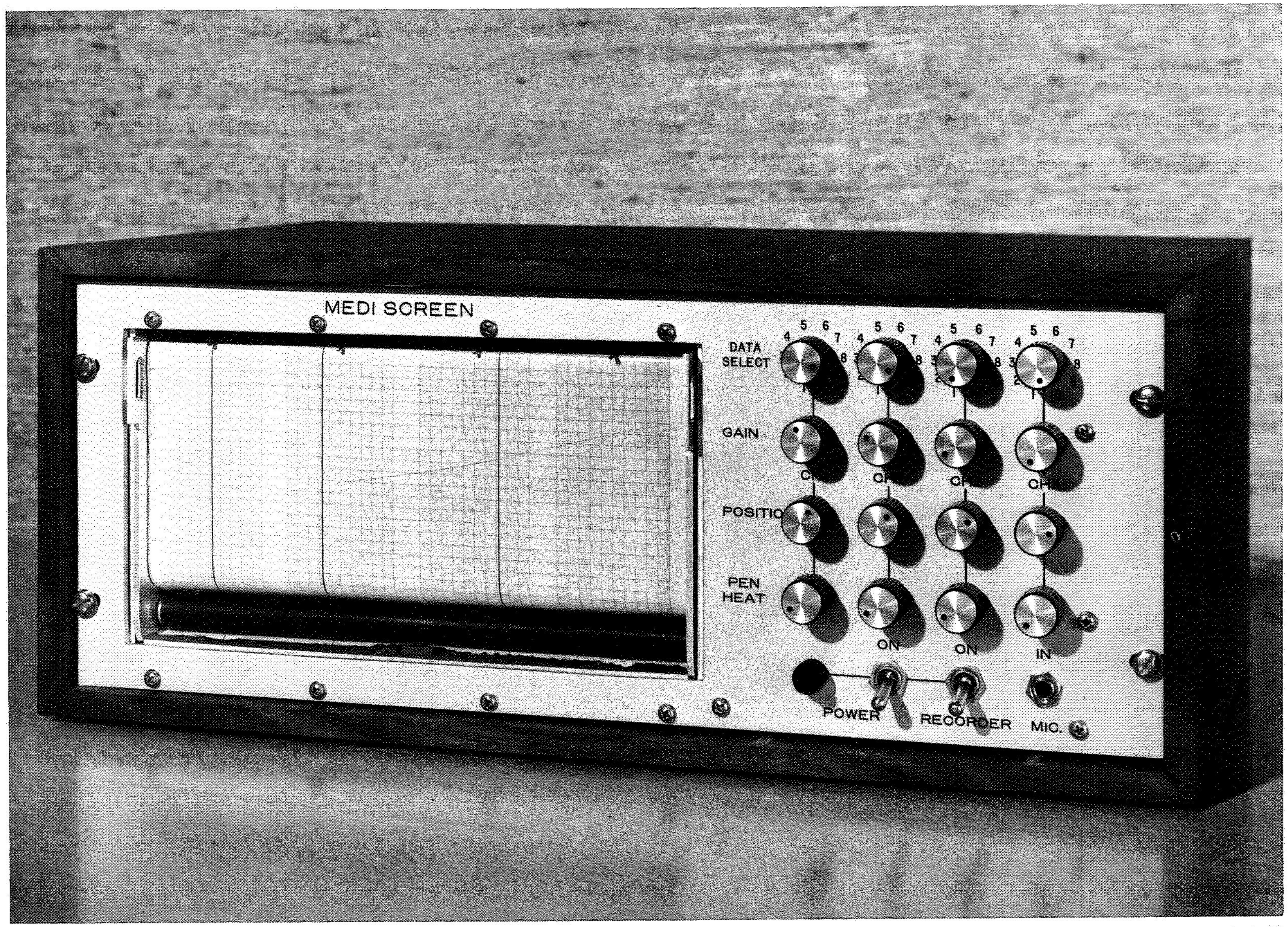


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Figure 2-17 Unattached Sensor System Strip Chart Recorder

2.5 SYSTEM TECHNICAL SPECIFICATIONS

Since common sensors are used to detect the data which are separated in multiple signal conditioners, one must be cautious in how he describes the system specifications, particularly the input characteristics. The minimum input impedance on any given channel affects the operation of all other signal conditioning channels. The minimum input impedance on the system described is differentially 8 kilohms and has a common mode input impedance of approximately 44 megohms. (Specified minimum input impedances are those impedances presented across the electrodes to driving sources operating at the frequencies specified below.) Circuit design changes are planned to increase significantly the differential input impedance, as discussed in paragraph 2.3.1, of this report.

2.5.1 Signal Conditioner Specifications

Specifications for signal conditioning are listed in Table II.

2.5.2 Strip Chart Recorder Specifications

The described system is designed to operate with any compatible recorder meeting AHA specifications.

TABLE II SIGNAL CONDITIONER SPECIFICATIONS

Signal Conditioner	Frequency Response (Signal Conditioner Outputs)	Data Lag (See Section 3)	Recovery Time (±5% of Actual - from Instant of Contact)	Input Range (With Auto-Ranging)	Output Range	Output Impedance	Calibration (Manual Pushbutton Switch)
ECG (Electrocardiogram)	0.15 - 100 Hz		20 sec (Max)	0 - 10 mV (Adjustable)	± 10 V	600 Ω	1 mV in series with electrode
GSR (Galvanic Skin Response)	0 - 50 Hz		0.5 sec (Max)	1 - 60 kΩ (Baseline)	± 10 V	600 Ω	100 Ω in series with electrode
ZPG (Respiration)/ w/ 100 kHz Carrier	0 - 0.8 Hz	9 ms	0.5 sec (Max)	330 - 760 Ω	± 10 V	600 Ω	3 Ω in series with electrode
ZCG (Central Blood Pulse)/w 100 kHz Carrier	0.8 - 5 Hz	150 ms	0.5 sec (Max)	330 - 760 Ω	± 10 V	600 Ω	
Thoracic Sounds	20 - 2000 Hz 20 Hz Low Pass		0.5 sec (Max) 0.5 sec (Max)		± 10 V ± 10 V	600 Ω 600 Ω	

Power Requirements - 115 V, 60 Hz, 50 watts (Max)

SECTION 3
STUDIES COMPARING DEVELOPED TECHNIQUES AND CONVENTIONAL METHODS

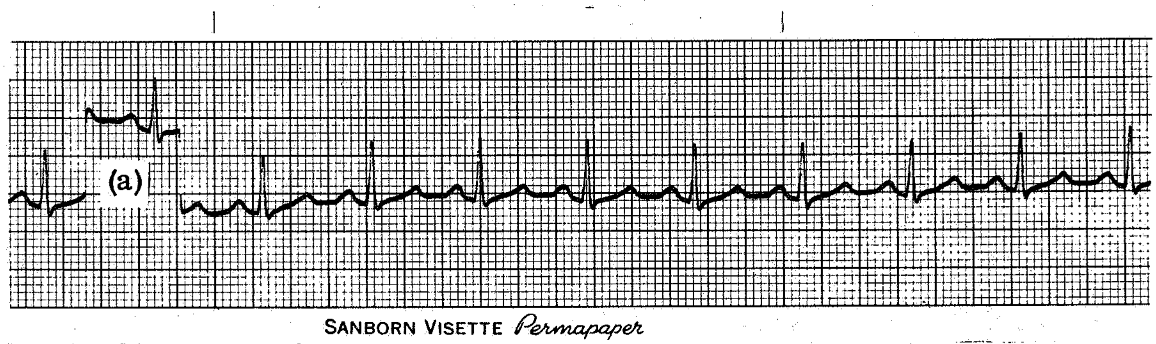
Electrocardiograph Lead I

Conventionally, electrocardiogram Lead I is obtained by measuring the heart generated potentials from electrodes placed on the volar wrist surfaces of each arm. For these measurements, the skin surface is normally cleaned with alcohol; then either ECG electrolyte paste is rubbed into the skin or an electrolyte pad is put in place. This procedure is followed by placing a rectangular ECG electrode over each area and attaching it in place with a rubber strap stretched over the electrode and around the wrist. In contrast, the unattached approach requires only that the subject place the palmar surface of each hand on the respective gold electrodes. No cleaning, skin surface preparation, electrolyte, or fastening is used.

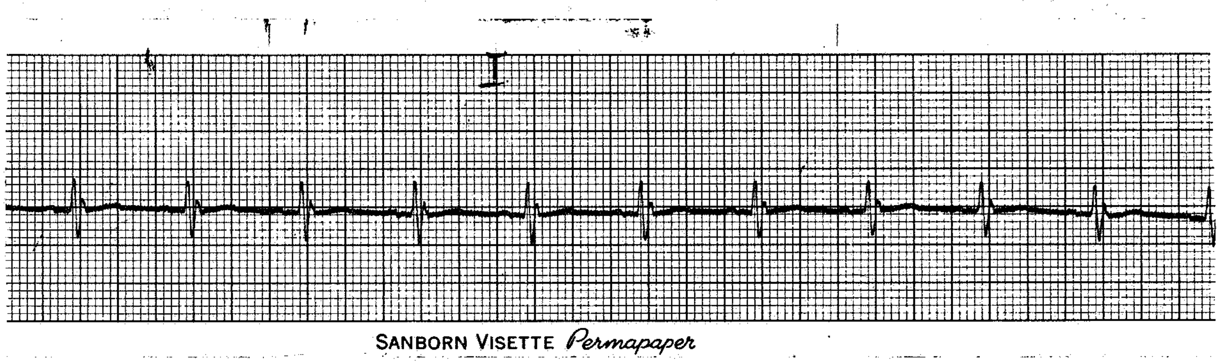
The task of this study was to determine the comparability of the two methods and, as required, to modify the unattached sensor circuitry in order to maximize the comparability of waveform characteristics.

The most desirable comparison of the two sensing approaches would be to obtain simultaneous recordings of the same heart potentials. The simultaneous approach was attempted, but resulted in cross interference. "QRS" complex alteration is quite pronounced with depressed "T" and "P" waves on the Sanborn Visette as shown in Figure 3-1. Amplitude depression is noted for the entire ECG waveform except the "S" wave on the monitoring system, as shown in Figure 3-2.

Since the simultaneous method was not applicable, the sequential method was used. A sample record was taken by conventional method (attached electrodes and a Sanborn Visette Electrocardiograph Model 300) followed by a sample record taken by the unattached method.



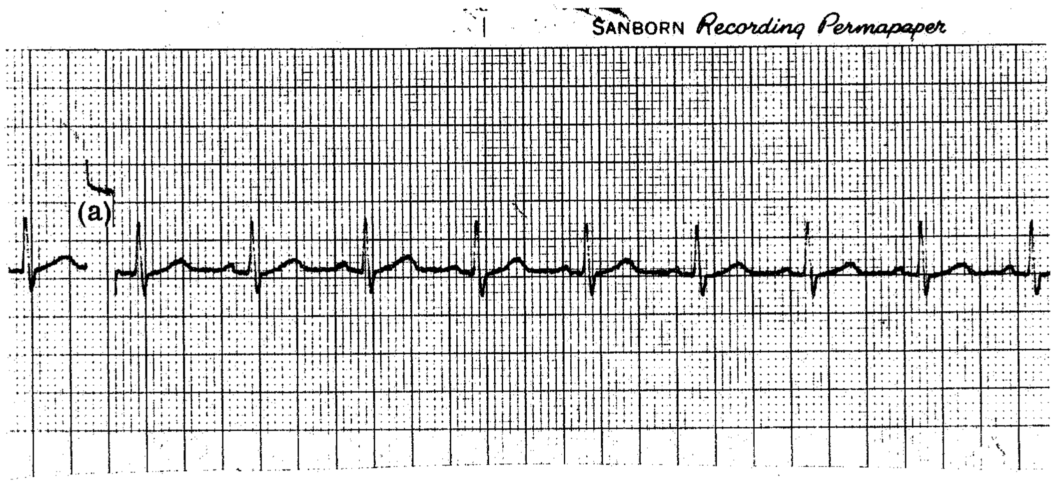
Normal ECG Lead I, Sanborn Visette Alone (Attached Electrodes)



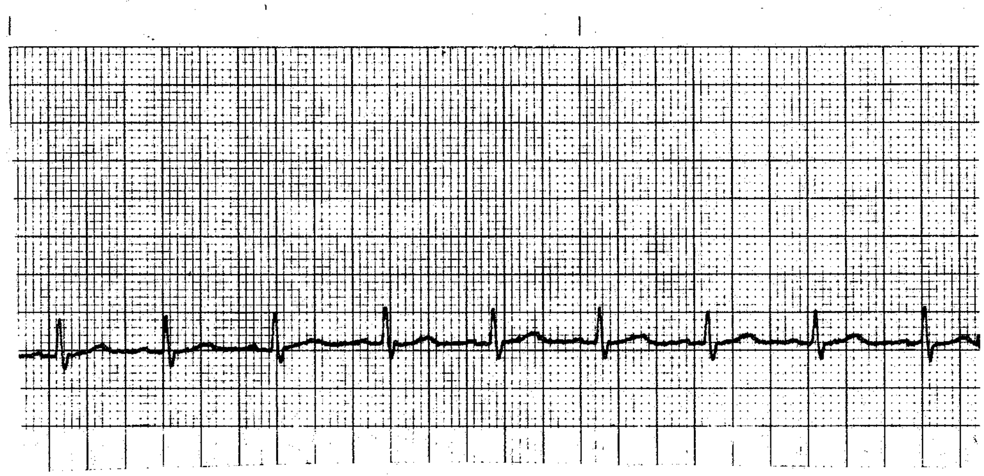
ECG Lead I Waveform Alteration on the Sanborn Visette during Simultaneous Operation of Sanborn Visette (Attached Electrodes) and Hands Resting on Unattached Electrodes

Figure 3-1 Example of ECG Waveform Alteration on Visette under Simultaneous Recording (Same Subject)

(a) Calibration, 2 cm = 1 mV; recording speed, 25 mm/sec



Normal ECG Lead I, Monitoring System Alone



ECG Lead I Waveform Alteration on the Unattached Monitoring System during Simultaneous Operation with Visette

Figure 3-2 Example of ECG Waveform Alteration on the Monitoring System under Simultaneous Recording (same Subject)

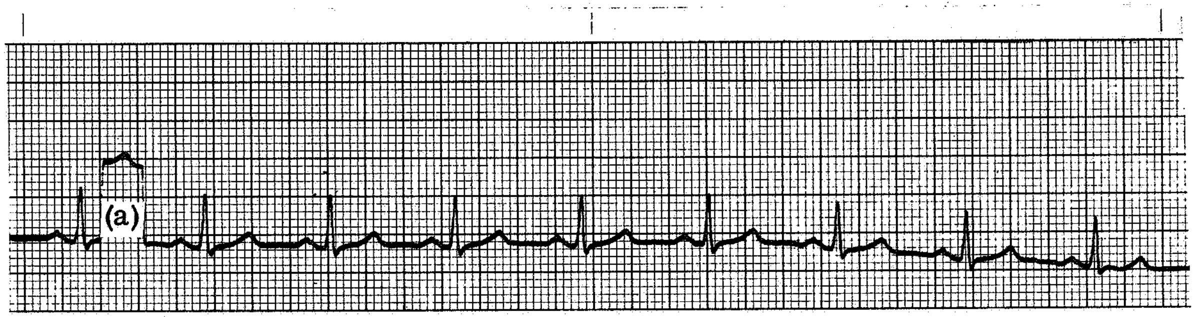
(a) Calibration, 2 cm = 1 mV; recording speed, 25 mm/sec

Characteristics of the monitoring system are presented in Section 2. Figure 3-3 illustrates an example of typical results on both the Visette and the monitoring system.

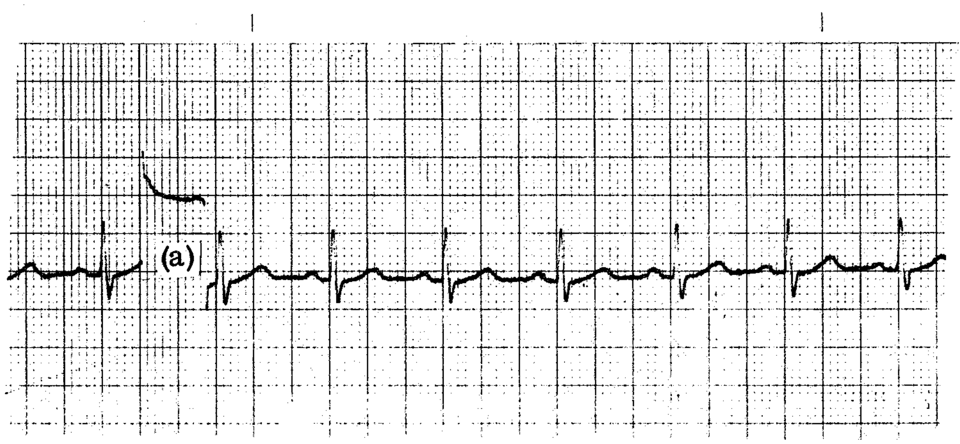
Careful study of the monitoring system ECG record, shown as the lower trace in Figure 3-3, indicates some S-wave distortion. This is explained by the RC coupling between palm and electronics. As was discussed in Section 2, the amplifier was designed for a frequency response of 0.15 - 100 Hz and it was assumed that the palm/electrode interface was mostly resistant at that low frequency. (Although the amplifier input impedance was low, amplitude distortion was to be cancelled by series calibration.) As was shown by measurements at the Biophysics Laboratories, University of Minnesota, undetected distortion was still possible, due to the surprisingly significant reactance (capacitive) component at the electrode/skin interface. The capacitance was measurable at near dc frequencies by computer analysis employing the instrumentation shown in Figure 3-4.

Figure 3-5 shows samples of the computer printout tab run, on which are noted significant phase angles, even at the very low frequencies. (Please note that the data as tabulated list only the characteristics of the skin/electrode interface and do not represent characteristics of any other part of the unattached monitoring system.) To check the computer and data acquisition system for self-induced phase angles, the skin/electrode interface was simulated with a pure resistance of approximately 5.5 kilohms. From the computer printouts shown in Figure 3-6, one can see that the computer was correctly analyzing phase angles since insignificant angles were detected when analyzing the resistor simulator.

Figure 3-7 shows the time behavior of skin/electrode impedance for two subjects. Note that the lower trace is fairly stable with time, while the upper trace shows rapid increase in impedance during the first four minutes and then drops toward the original level. This is perhaps explained by the fact that the more stable subject was very familiar with the monitoring system and probably registered less initial GSR. Note, however, that the phase angles at the measurement frequency of 20 Hz were nearly the same for both subjects.



Normal ECG Lead I, Sanborn Visette



Normal ECG Lead I, Unattached Sensor Monitoring System

Figure 3-3 Example of Comparative ECG Lead I Records on Same Subject.

(a) Calibration, 2 cm = 1 mV; Recording Speed, 25 mm/sec

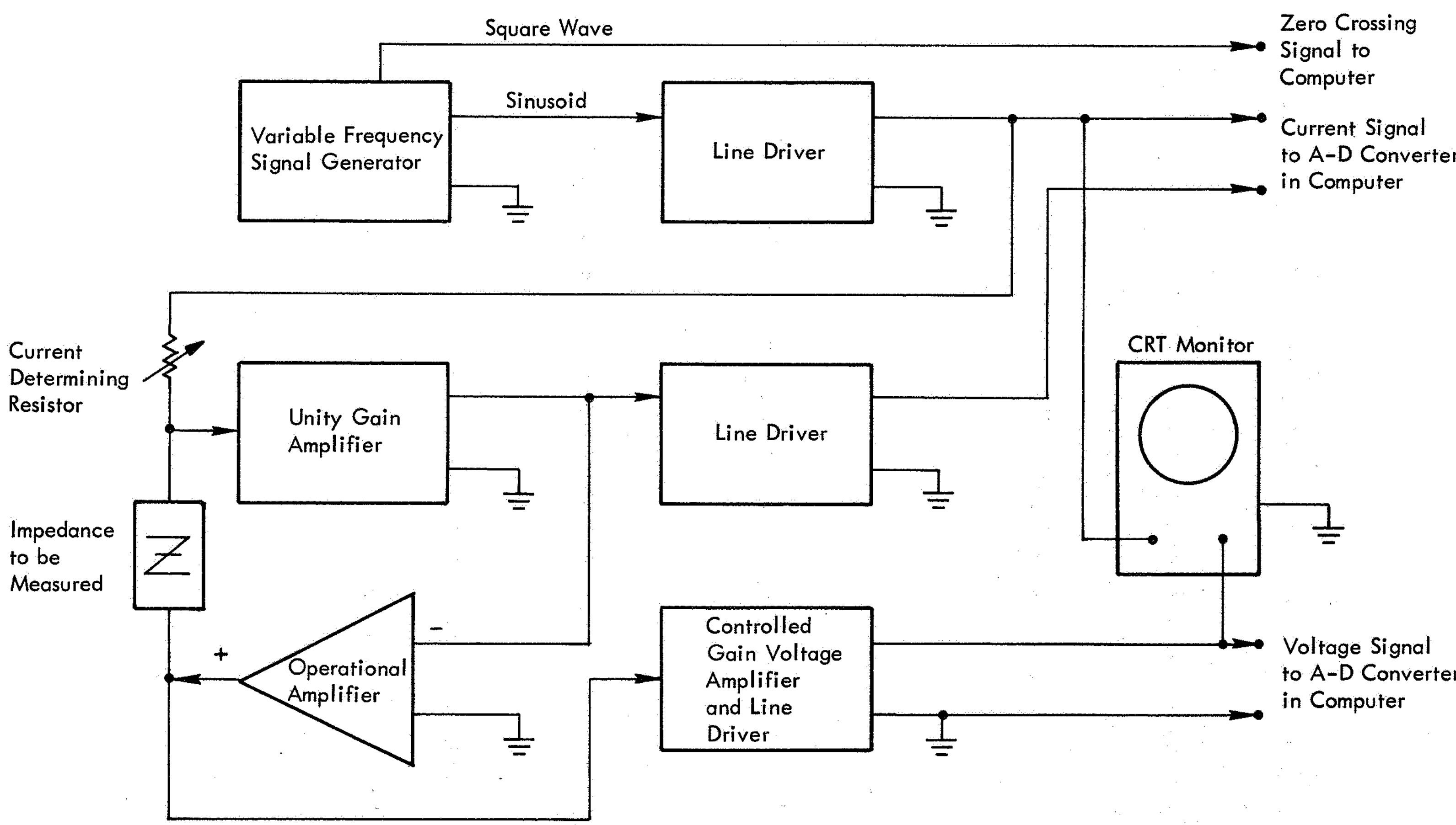


Figure 3-4 On-Line Impedance Measuring System (Operating in Biophysics Laboratory, University of Minnesota)

SUBJECT - E.J.

```
INPUT   = 1
05/16/68
TIME = 15:33 HRS   26.6 SECS
FREQUENCY = +    .096 CPS
V(0)        = -    .010 DC
V(1)        = +    .125 RMS
V(2)/V(1) = +    .761 %
V(3)/V(1) = +    .584 %
I(0)        = -    .000 MA
I(1)        = +    .023 MA
I(2)/I(1) = +    .047 %
I(3)/I(1) = +    .049 %
Z           = +5448.05 OHMS
THETA       = -   2.391 DEGREES
R(S)        = +5443.38 OHMS
X(S)        = +227.380 OHMS
```

```
INPUT   = 1
05/16/68
TIME = 15:34 HRS   59.9 SECS
FREQUENCY = +    .049 CPS
V(0)        = -    .015 DC
V(1)        = +    .125 RMS
V(2)/V(1) = +    .103 %
V(3)/V(1) = +    .424 %
I(0)        = +    .000 MA
I(1)        = +    .023 MA
I(2)/I(1) = +    .000 %
I(3)/I(1) = +    .070 %
Z           = +5420.17 OHMS
THETA       = -   2.329 DEGREES
R(S)        = +5415.69 OHMS
X(S)        = +220.342 OHMS
```

```
INPUT   = 1
05/16/68
TIME = 15:38 HRS   03.3 SECS
FREQUENCY = +    .018 CPS
V(0)        = -    .010 DC
V(1)        = +    .130 RMS
V(2)/V(1) = +   2.135 %
V(3)/V(1) = +   1.394 %
I(0)        = +    .000 MA
I(1)        = +    .023 MA
I(2)/I(1) = +    .037 %
I(3)/I(1) = +    .030 %
Z           = +5635.23 OHMS
THETA       = -   1.989 DEGREES
R(S)        = +5631.83 OHMS
X(S)        = +195.652 OHMS
```

```
INPUT   = 1
05/16/68
TIME = 15:30 HRS   31.0 SECS
FREQUENCY = +    .498 CPS
V(0)        = -    .[illegible]21 DC
V(1)        = +    .[illegible] RMS
V(2)/V(1) = + 41.291 %
V(3)/V(1) = + 25.623 %
I(0)        = -    .000 MA
I(1)        = +    .000 MA
I(2)/I(1) = + 52.300 %
I(3)/I(1) = +   7.022 %
Z           = +4945.96 OHMS
THETA       = +   6.131 DEGREES
R(S)        = +4917.67 OHMS
X(S)        = -528.278 OHMS
```

```
INPUT   = 1
05/16/68
TIME = 15:31 HRS   17.8 SECS
FREQUENCY = +    .484 CPS
V(0)        = -    .010 DC
V(1)        = +    .121 RMS
V(2)/V(1) = +    .363 %
V(3)/V(1) = +    .332 %
I(0)        = -    .000 MA
I(1)        = +    .023 MA
I(2)/I(1) = +    .066 %
I(3)/I(1) = +    .007 %
Z           = +5268.84 OHMS
THETA       = -   2.156 DEGREES
R(S)        = +5265.10 OHMS
X(S)        = +198.265 OHMS
```

```
INPUT   = 1
05/16/68
TIME = 15:32 HRS   15.0 SECS
FREQUENCY = +    .199 CPS
V(0)        = -    .012 DC
V(1)        = +    .126 RMS
V(2)/V(1) = +    .801 %
V(3)/V(1) = +    .261 %
I(0)        = +    .000 MA
I(1)        = +    .023 MA
I(2)/I(1) = +    .083 %
I(3)/I(1) = +    .038 %
Z           = +5461.21 OHMS
THETA       = -   1.918 DEGREES
R(S)        = +5458.15 OHMS
X(S)        = +182.823 OHMS
```

Figure 3-5 Example of Computer Printout. Notice Significant Phase Angles (θ)'s at Very Low Frequencies

```
INPUT   = 1
05/16/68
TIME = 15:42 HRS  2[illegible] SECS
FREQUENCY = +   .[illegible] CPS
V(0)        = -   .[illegible]9 DC
V(1)        = +   .127 RMS
V(2)/V(1) = +   .113 %
V(3)/V(1) = +   .139 %
I(0)        = +   .[illegible] MA
I(1)        = +   .023 MA
I(2)/I(1) = +   .037 %
I(3)/I(1) = +   .035 %
Z           = +5502.11 OHMS
THETA       = +   .059 DEGREES
R(S)        = +5502.11 OHMS
X(S)        = -  5.700 OHMS

INPUT   = 1
05/16/68
TIME = 15:47 HRS  29.1 SECS
FREQUENCY = + 19.586 CPS
V(0)        = -   .009 DC
V(1)        = +   .127 RMS
V(2)/V(1) = +   .155 %
V(3)/V(1) = +   .174 %
I(0)        = -   .000 MA
I(1)        = +   .023 MA
I(2)/I(1) = +   .111 %
I(3)/I(1) = +   .005 %
Z           = +5508.76 OHMS
THETA       = -   .105 DEGREES
R(S)        = +5508.75 OHMS
X(S)        = + 10.169 OHMS

INPUT   = 1
05/16/68
TIME = 15:48 HRS  15.3 SECS
FREQUENCY = + 49.902 CPS
V(0)        = -   .003 DC
V(1)        = +   .121 RMS
V(2)/V(1) = +  6.810 %
V(3)/V(1) = +  6.651 %
I(0)        = +   .000 MA
I(1)        = +   .022 MA
I(2)/I(1) = +  6.755 %
I(3)/I(1) = +  6.590 %
Z           = +5501.15 OHMS
THETA       = -   .087 DEGREES
R(S)        = +5501.14 OHMS
X(S)        = +  8.354 OHMS

INPUT   = 1
05/16/68
TIME = 15:49 HRS  29.3 SECS
FREQUENCY = + 47.007 CPS
V(0)        = -   .005 DC
V(1)        = +   .130 RMS
V(2)/V(1) = +  4.616 %
V(3)/V(1) = +  4.416 %
I(0)        = +   .000 MA
I(1)        = +   .023 MA
I(2)/I(1) = +  4.632 %
I(3)/I(1) = +  4.403 %
Z           = +5501.37 OHMS
THETA       = -   .105 DEGREES
R(S)        = +5501.36 OHMS
X(S)        = + 10.087 OHMS

INPUT   = 1
05/16/68
TIME = 15:50 HRS  11.9 SECS
FREQUENCY = + 44.061 CPS
V(0)        =
05/16/68
TIME = 15:50 HRS  19.8 SECS
FREQUENCY = + 44.046 CPS
V(0)        = -   .009 DC
V(1)        = +   .126 RMS
V(2)/V(1) = +   .459 %
V(3)/V(1) = +   .139 %
I(0)        = -   .000 MA
I(1)        = +   .023 MA
I(2)/I(1) = +   .380 %
I(3)/I(1) = +   .166 %
Z           = +5492.96 OHMS
THETA       = -   .066 DEGREES
R(S)        = +5492.96 OHMS
X(S)        = +  6.390 OHMS

INPUT   = 1
05/16/68
TIME = 15:51 HRS  15.9 SECS
FREQUENCY = +449.141 CPS


V(0)        = 1
05/16/68
TIME = 15:51 HRS  26.5 SECS
FREQUENCY = +439.885 CPS
V(0)        = -   .009 DC
V(1)        = +   .126 RMS
V(2)/V(1) = +   .307 %
V(3)/V(1) = +   .162 %
I(0)        = -   .000 MA
I(1)        = +   .023 MA
I(2)/I(1) = +   .369 %
I(3)/I(1) = +   .125 %
Z           = +5499.91 OHMS
THETA       = -   .781 DEGREES
R(S)        = +5499.40 OHMS
X(S)        = + 74.995 OHMS

INPUT   = 1
05/16/68
TIME = 15:52 HRS  08.7 SECS
FREQUENCY = + 48.855 CPS
V(0)        = -   .010 DC
V(1)        = +   .125 RMS
V(2)/V(1) = +   .954 %
V(3)/V(1) = +  1.135 %
I(0)        = -   .000 MA
I(1)        = +   .022 MA
I(2)/I(1) = +   .980 %
I(3)/I(1) = +  1.100 %
Z           = +5505.51 OHMS
THETA       = -   .069 DEGREES
R(S)        = +5505.50 OHMS
X(S)        = +  6.706 OHMS
```

Figure 3-6 Results of System Check Using 5.5k Resistor as Z-Simulator

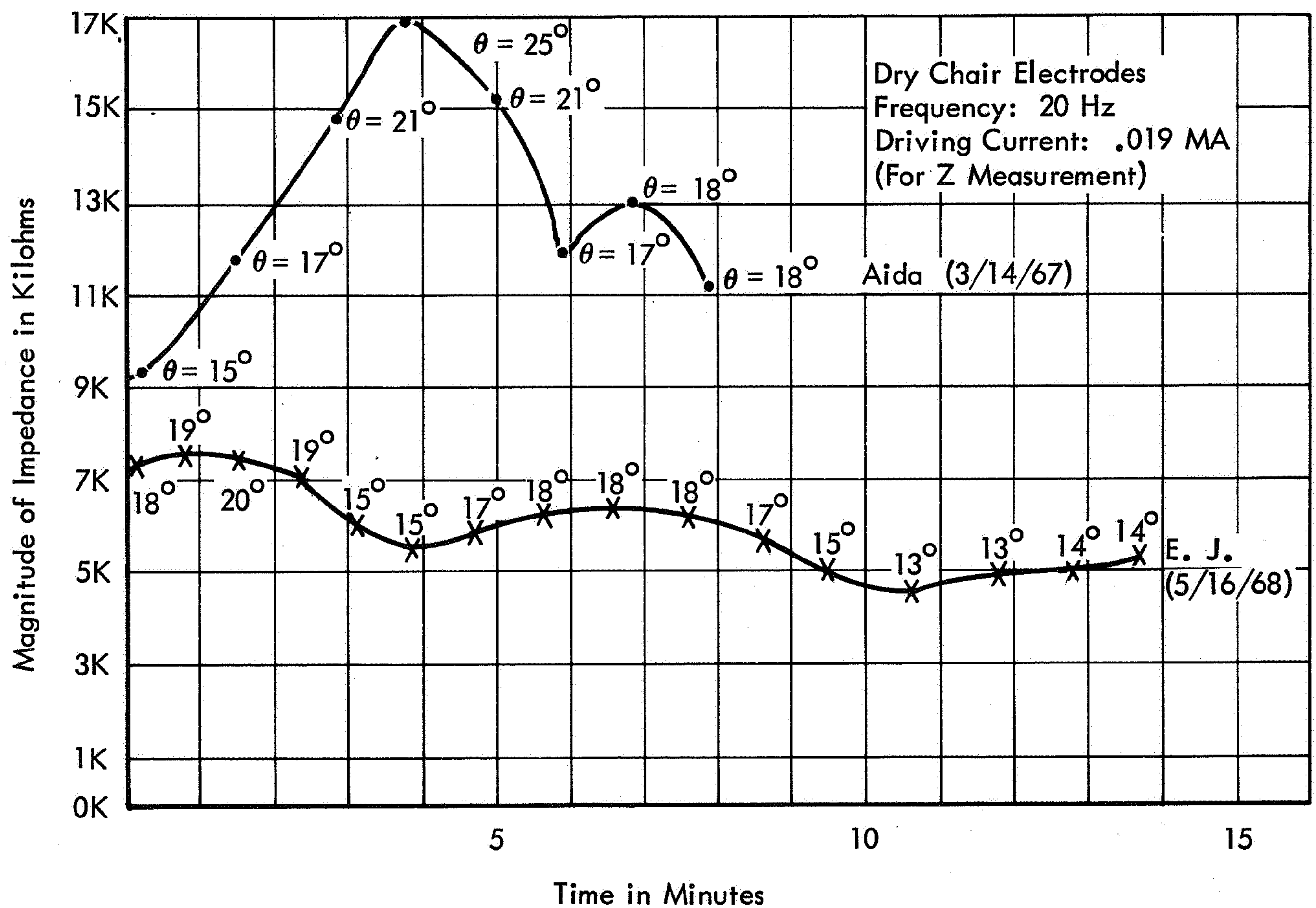


Figure 3-7 Time Behavior of Skin Impedance for Two Subjects

Figure 3-8 shows skin impedance vs. driving current for the same two subjects. Notice how the impedance immediately begins to drift upward for subject AIDA and then is driven down as the driving current is increased. This suggests that the rise in impedance shown in Figure 3-7 might also have been due to skin/electrode polarization, since the polarization apparently can be reversed by increasing the driving current, as shown in Figure 3-8. Notice again that the impedance drift was more for subject AIDA.

Figure 3-9 shows the classic semi-circular plot of series skin reactance vs. series skin resistance. The curve is plotted for frequencies ranging from 1/2 Hz to 450 Hz. A similar plot is shown in Figure 3-10, with the time sequence of skin impedance starting at 500 Hz, dropping to 0.1 Hz, and returning to 500 Hz. Note that for subject T.J., series reactance increases with time.

Figure 3-11 shows skin impedances for several subjects, with and without paste on the chair electrodes. As may be expected, the impedance drops for a given subject when paste is applied, but the phase angles remain approximately the same.

Comparison of ECG Results. - After studying the performance of the unattached monitoring system and comparing it with conventional ECG measuring techniques, the following conclusions have been drawn:

1. With the circuit modifications recommended in Section 2, Lead I ECG should be comparable in waveform and amplitude.

2. The baseline noise level is comparable when the subjects are relaxed.

3. The baseline noise of the monitoring system ECG is muscle "noise".

4. Skin preparation and the application of electrolytic pads or paste are not required to obtain a usable ECG Lead I from the palmar surfaces.

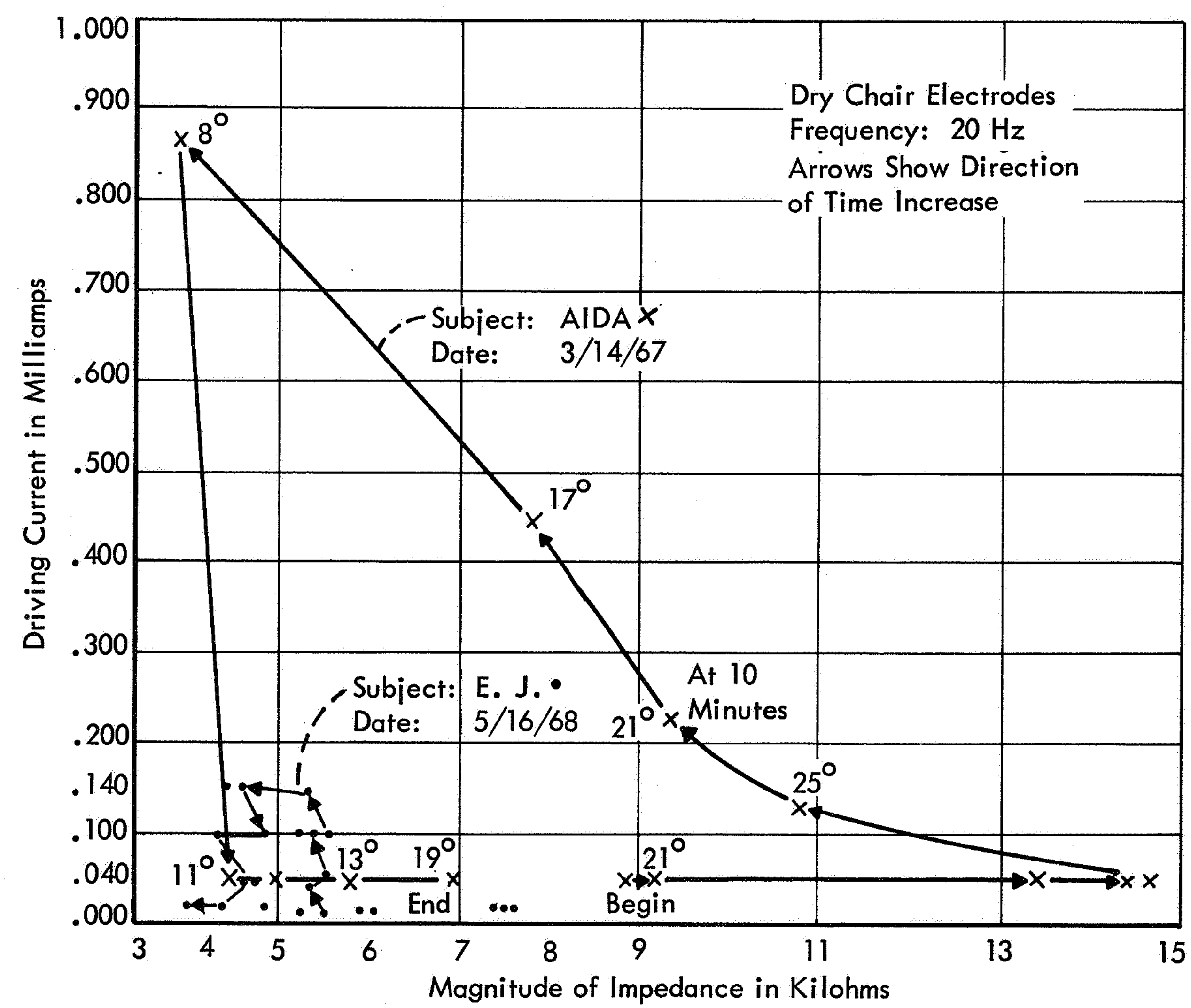


Figure 3-8 Skin Impedance vs Driving Current for Two Subjects

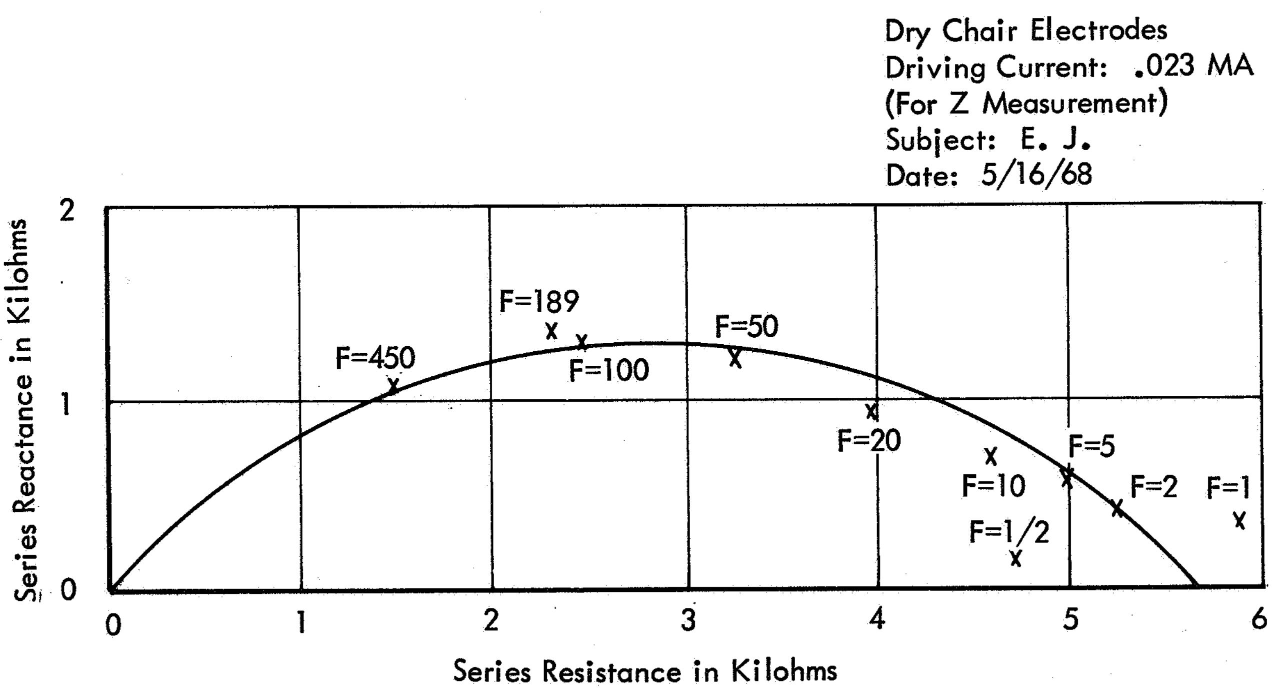


Figure 3-9 Geometric Mean Value of Skin Impedance

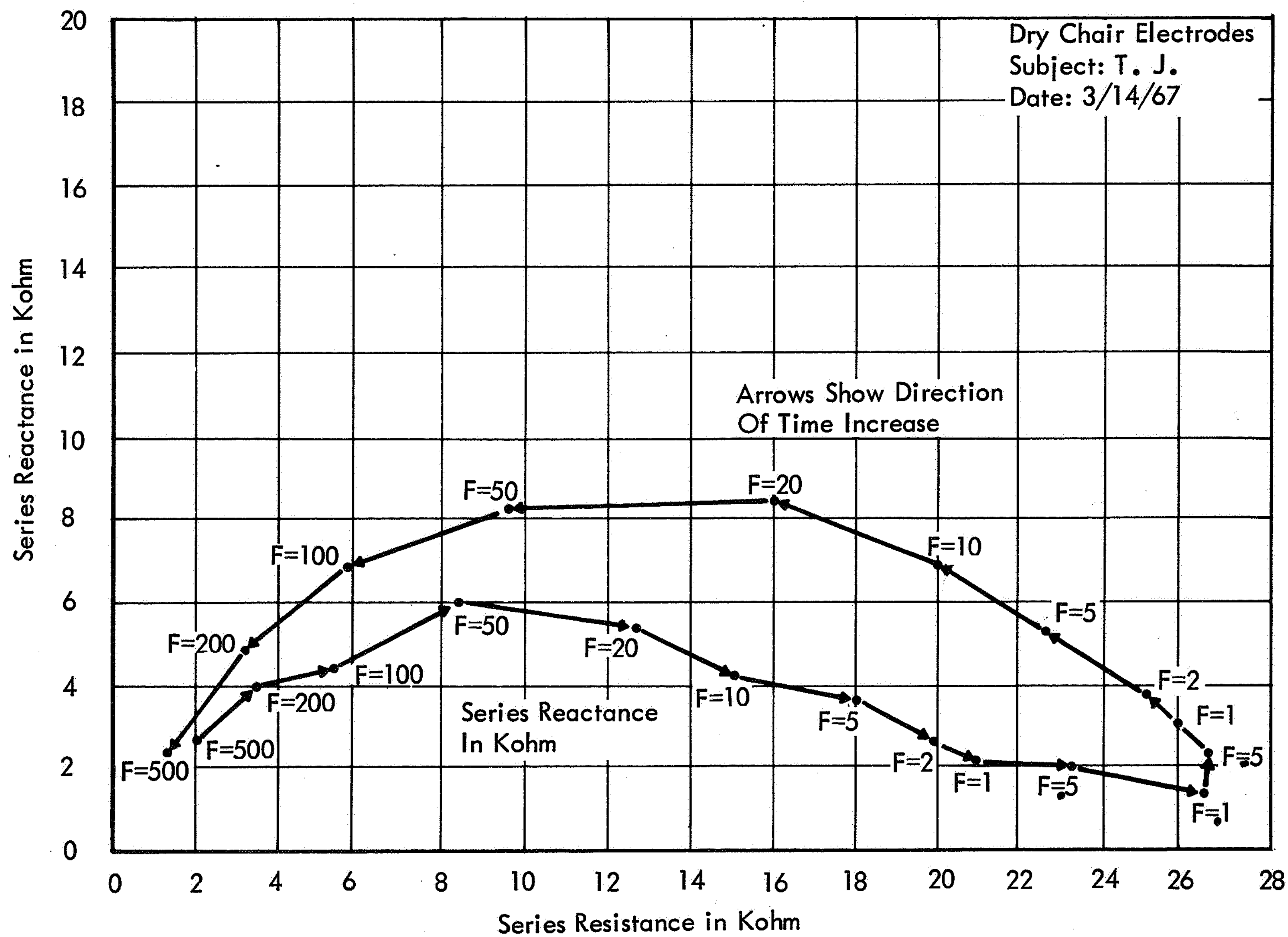


Figure 3-10 Skin Impedance Time Loop

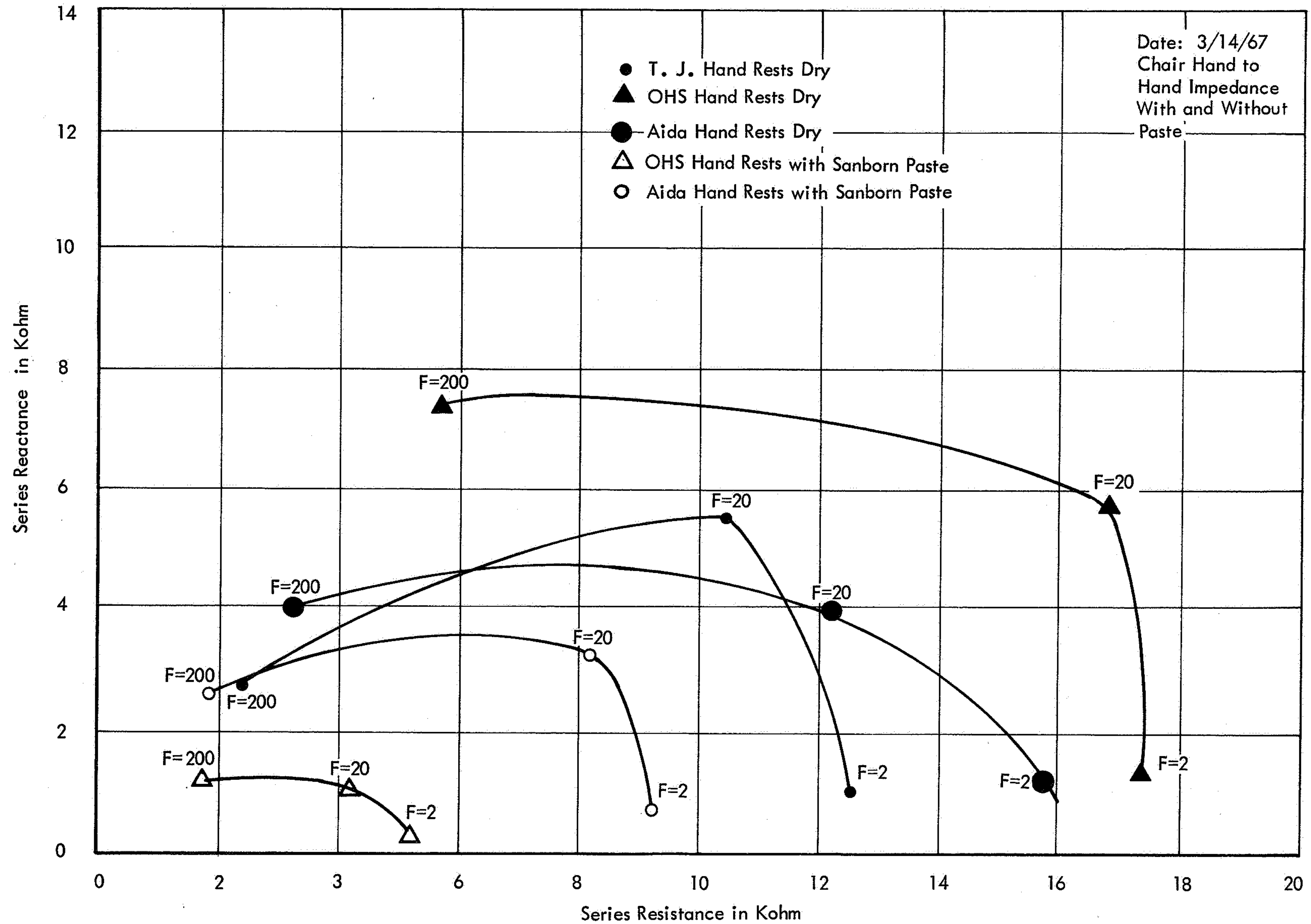


Figure 3-11 Skin Impedance With and Without Paste

5. As a consequence of the above, whenever intermittent or periodic monitoring is required, the unattached sensor approach seems applicable.

Impedance Respiration

Transthoracic body impedance has been successfully used in Project Gemini and other space and ground projects to monitor respiration. This measurement has been accomplished by attaching an electrode under each arm in the mid-axillary region.

As one of the prime body functions monitored during spaceflight, it became important to ascertain the effectiveness of monitoring impedance respiration by the unattached sensor approach. Therefore, this study was performed to compare mid-axillary impedance respiration measurement with palm-to-palm respiration measurement.

It was found that two impedance circuits which were identical could not be operated on the same subject simultaneously. When both 100-kHz signals would be impressed across the body (one from palm-to-palm, the other transthoracic), cross-interference was encountered.

The method of comparison utilized was to compare each impedance respiration technique against clinical spirometer records. The spirometer used was a Collins METABOLEX. This unit was modified by attaching a resistance slide-wire to the bellows support and attaching a sliding contact to the ink pen. The adaptation permitted direct, linear slide-wire resistance measurements to be calibrated against the actual respiratory volume.

Outputs of the spirometer waveforms and the impedance waveforms were simultaneously recorded on a dual channel strip chart recorder for comparison.

Figure 3-12 presents an example of the mid-axillary impedance respiration waveform compared with the spirometer waveform. Figure 3-13 shows an example using the same subject, of the palm-to-palm impedance respiration waveform

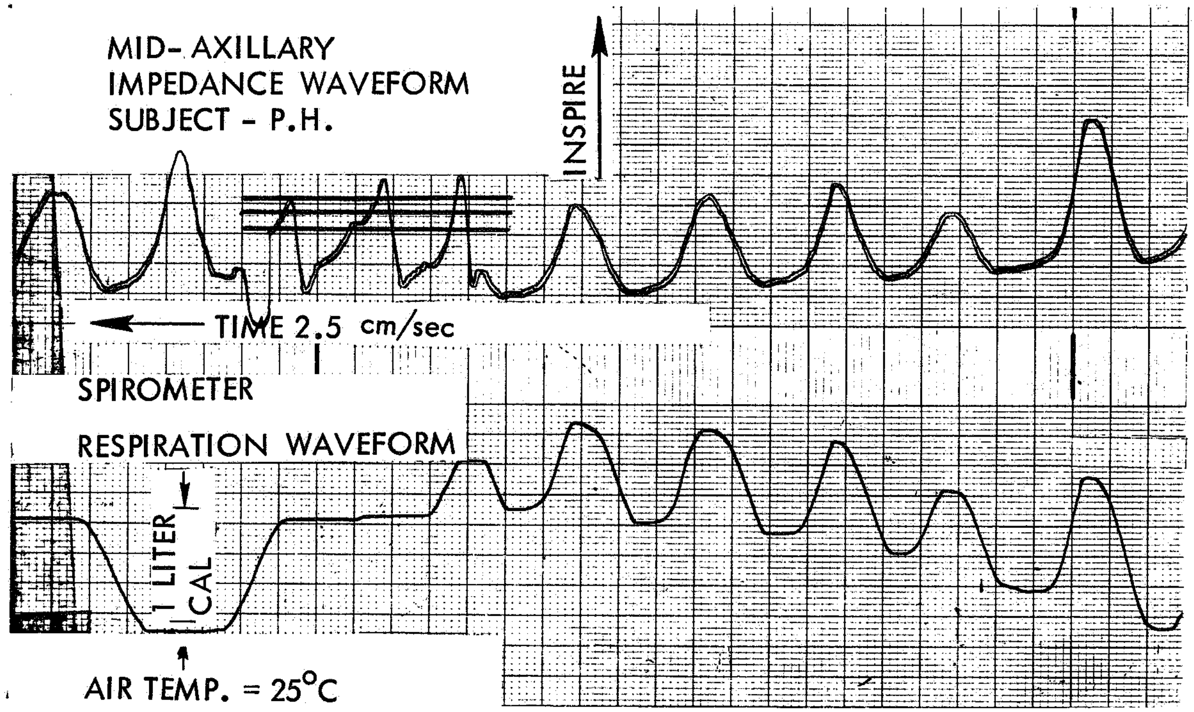


Figure 3-12 Example of Simultaneous Recordings of Mid-Axillary and Volumetric Respiration Waveforms

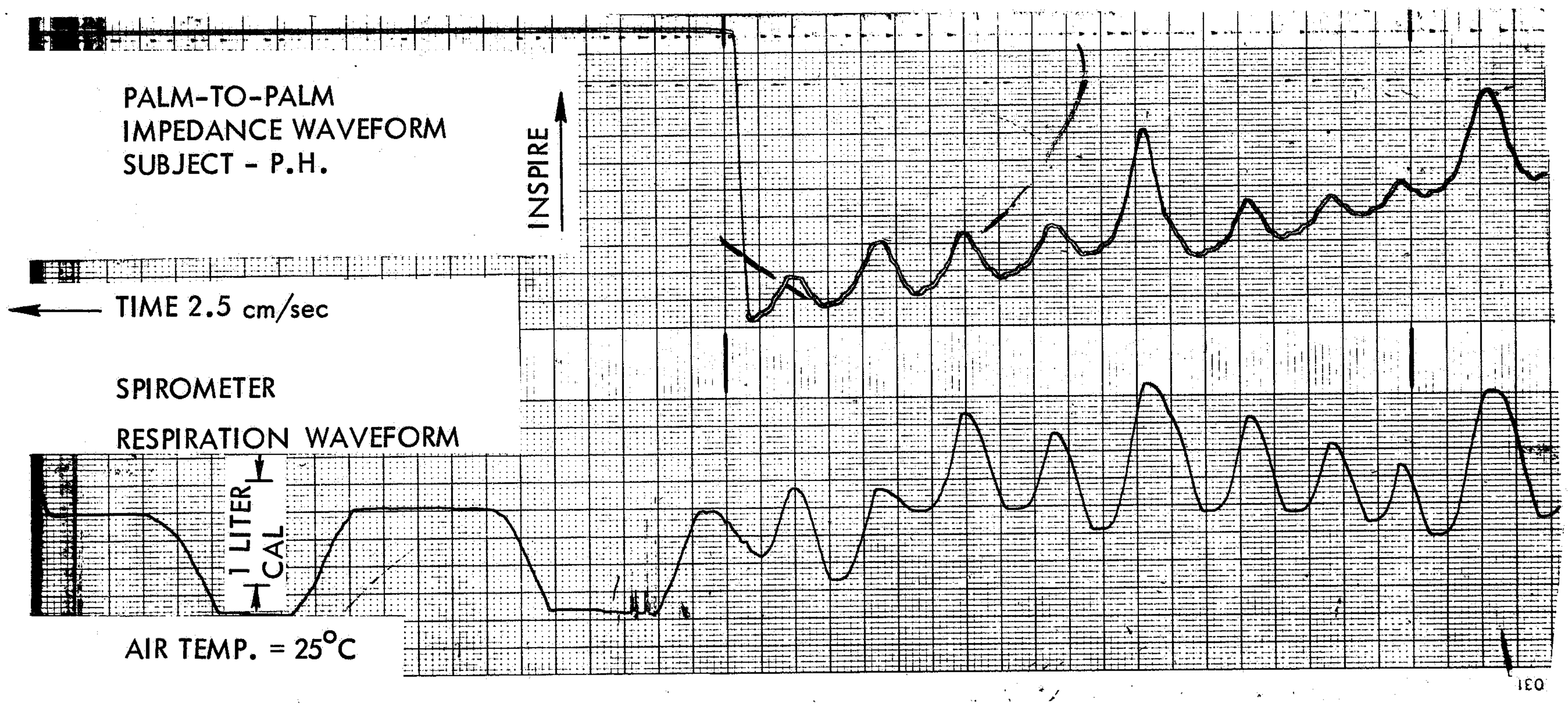


Figure 3-13 Example of Simultaneous Recordings of Palm-to-Palm (Unattached Sensor) and Volumetric Respiration Waveforms

compared with the spirometer waveform. In each of these figures, time runs from right to left and was recorded at 2.5 cm/sec. The subject was asked to take a deeper breath from time to time.

Results of impedance respiration study. - Results of this study are:

1. Phase lag of both the impedance and spirometer waveforms are approximately equal.

2. Both palm-to-palm and mid-axillary impedance waveform closely approximately the spirometer waveforms in contour.

3. Mid-axillary respiration amplitudes are more linearly proportional to the spirometer amplitudes than are the palm-to-palm amplitudes.

4. More baseline drift was consistently noted in the palm measurements than in mid-axillary measurements.

The important conclusions concerning the palm-to-palm impedance respiration are:

1. The unattached palm-to-palm impedance respiration method produces respiration waveforms comparable to the spirometer.

2. The respiration signals obtained are adequate for rate counting (pneumotachometry).

3. For quantitative volumetric purposes, the unattached impedance technique will require amplitude calibration scaling for each subject since impedance change per liter is a function of body proportions.

Impedance Pulse

Early in this study, it was determined that an impedance pulse waveform similar to that reported by Nyboer (Reference 4), existed also on the palm-to-palm impedance measurements. Circuitry was developed to isolate this pulse information from respiration information. As stated in Section 2, impedance respiration is obtained with a low pass of dc to 0.8 Hz. The impedance blood pulse is obtained with filter circuitry having a band pass of 0.8 to 50 Hz and additional amplification.

It had been hypothesized that the palm-to-palm impedance pulse is a central pulse (i.e., thoracic rather than distal-finger), since the pulse recordings continue when both of the subject's arms are simultaneously cuffed to a pressure above the systolic blood pressure. This study was undertaken to ascertain the timing sequence which exists between the occurrence of the palm-to-palm impedance pulse and the pulse as sensed simultaneously at the finger.

Since it was known that a phase lag existed in the impedance pulse circuitry, it was important to select a method of sensing the finger pulse which would have no phase lag. The method selected was to use a small photo-cell in conjunction with a light source and filter.

The impedance phase lag was determined by using two identical photo cells; one driving a strip chart channel directly, the second driving the impedance pneumograph-pulse circuits. The circuit for lag determination is shown in Figure 3-14. Each photo cell was driven by pulses from the same finger.

The method for comparing the impedance pulse with the finger pulse was to obtain simultaneous, dual-channel recordings of the two waveforms (see Figure 3-15).

<u>Impedance phase lag study results.</u> - Figure 3-16 shows the results of that experiment. The artery was occluded to provide a common timing index to both photo cells; then the first pulse after occlusion provided the timing mark. Noting that the two pulse curves differed markedly in contours, the two photo cells were reversed to see if their geometry, with respect to the finger, affected the pulse shape. Figure 3-17 shows this reversal. No significant change of wave shape was found.

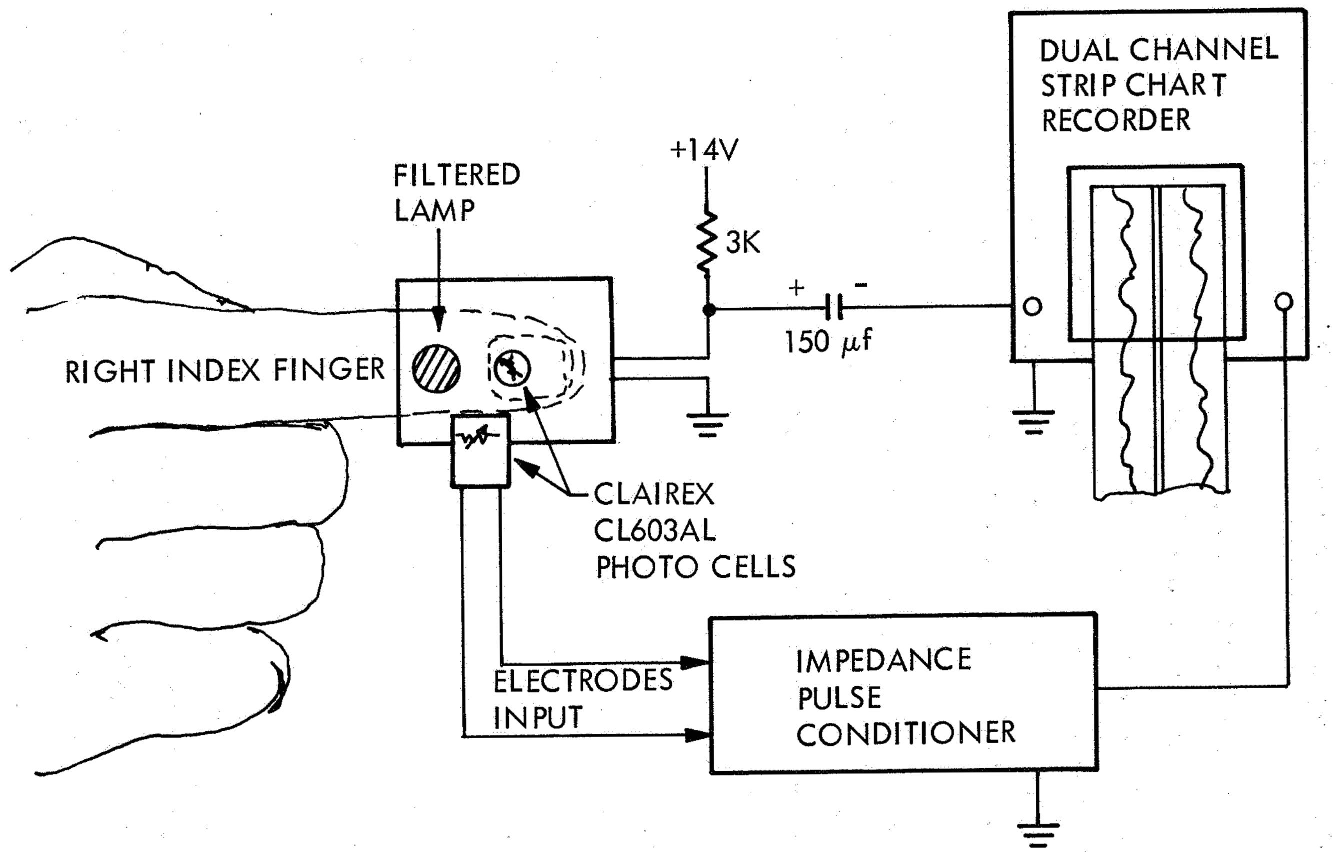


Figure 3-14 Instrumentation for Experiment Determining Phase Lag in Z-Pneumograph and Z-Pulse Electronics

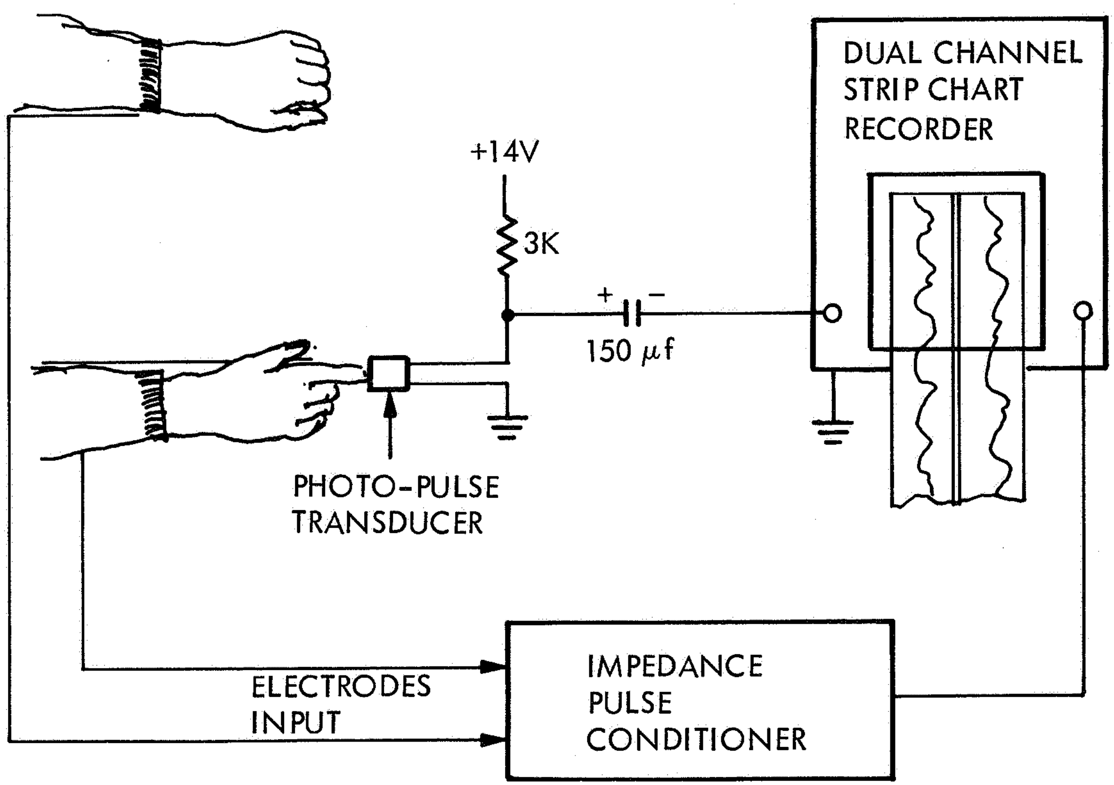


Figure 3-15 Instrumentation for Comparing Palm Impedance Pulse Waveform with Photo-Pulse Transducer Waveform

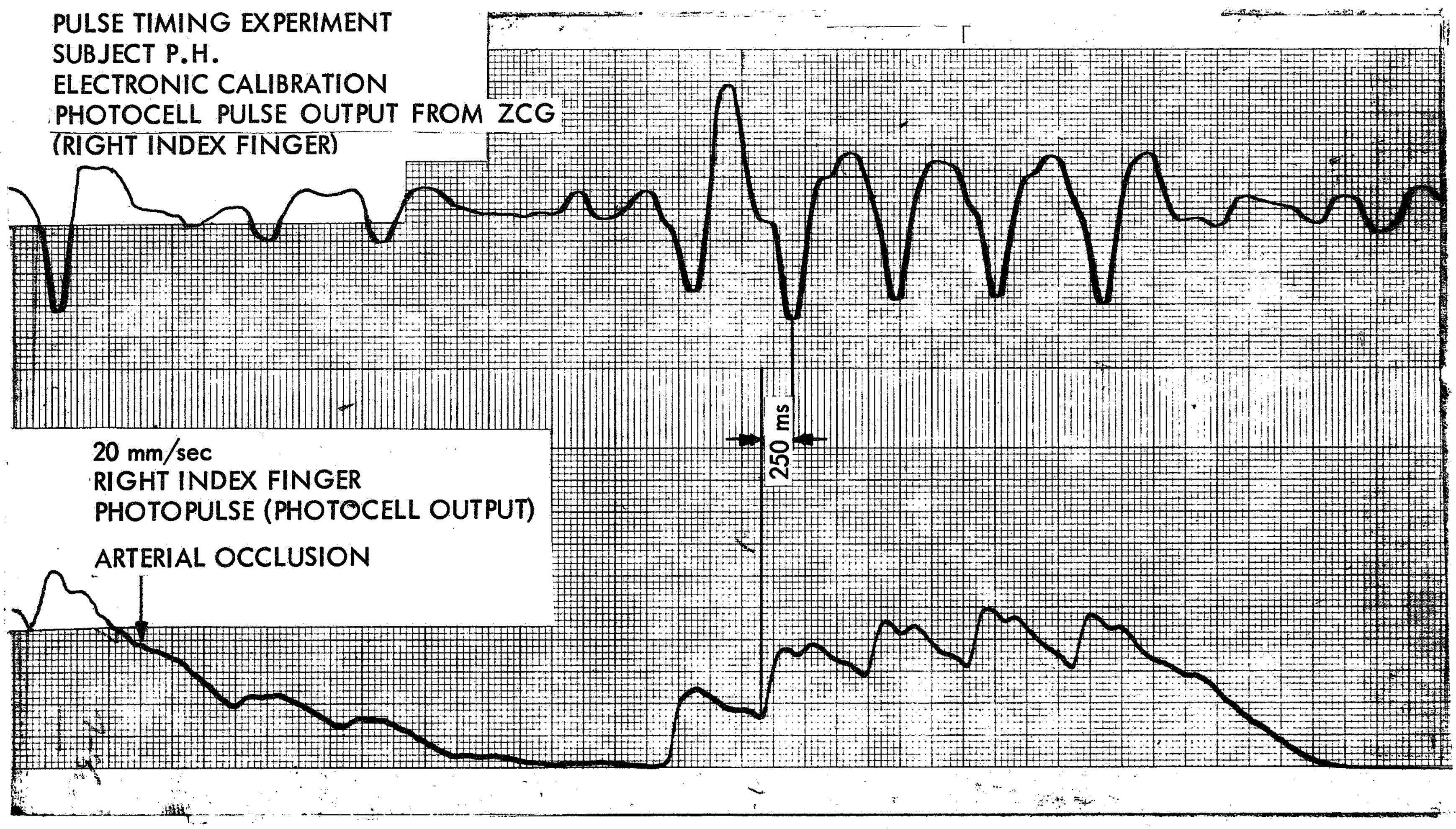


Figure 3-16 Illustration of Impedance Pulse Electronic Phase Lag

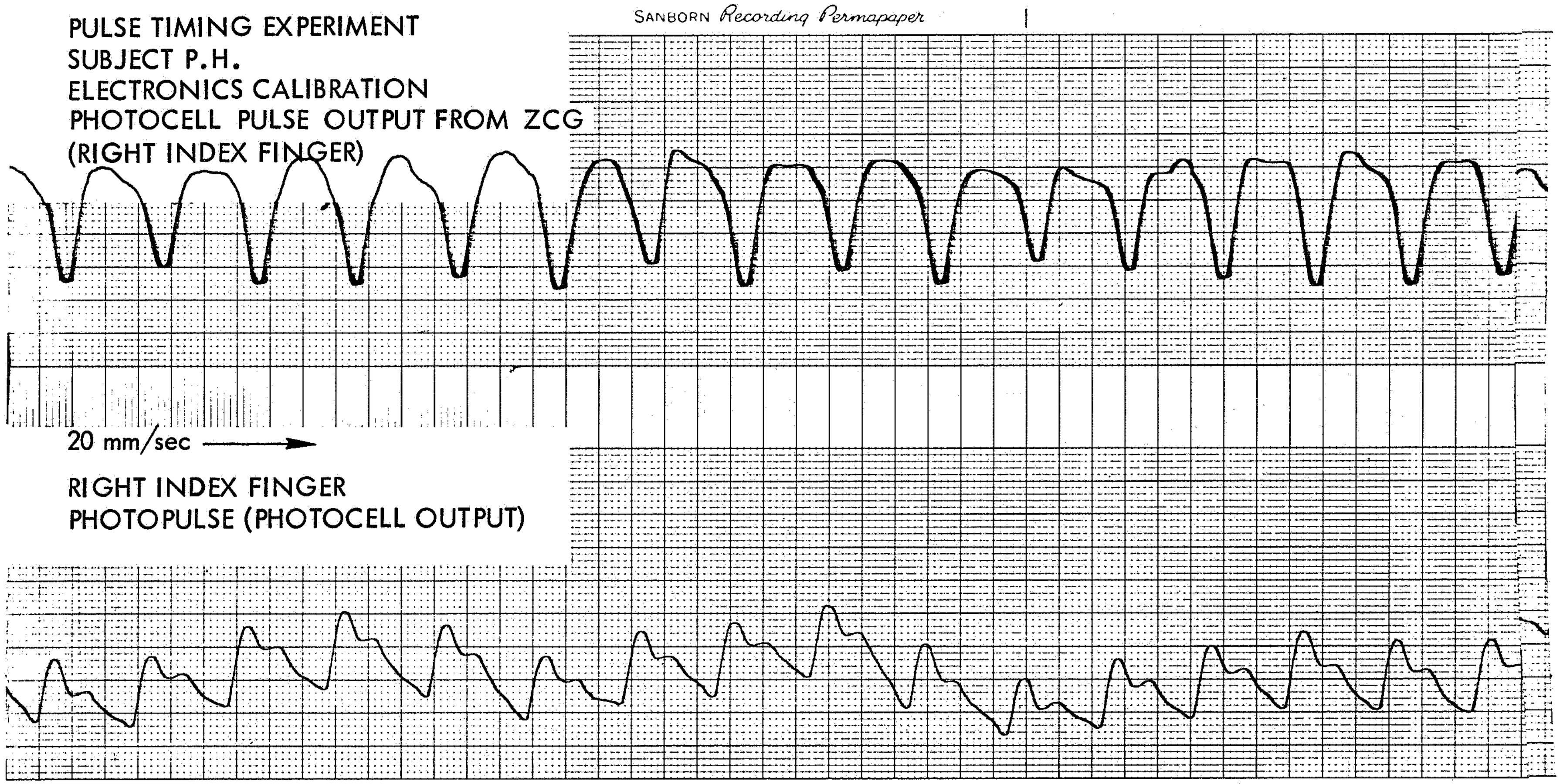


Figure 3-17 Illustration of Respective Waveform Characteristics when Photo Cells were Reversed in the Phase Lag Study (Compare with Figure 3-16)

The dynamic characteristics of the strip chart recorder were slightly different on each channel so that the recorder inputs were reversed. Figure 3-18 and 3-19 show the respective phase lag under recording conditions of 20 mm/sec on the left shifted to 100 mm/sec on the right. The expanded recording rate clearly illustrates both the impedance electronic lag and, upon recorder channel reversal, the differential lag of the recorder channels. These figures also illustrate the pulse waveform distortion introduced by the impedance conditions.

Palm-to-palm impedance pulse compared with finger photo-pulse. - Expanded strip chart recordings of subject pulse waveforms are illustrated in Figures 3-20 and 3-21 to counterbalance differential recorder channel lags.

When corrected for phase lag in the impedance circuitry and the differential recording channel phase lag, the onset of the impedance pulse was found to actually lead the onset of the finger photo pulse by approximately 70 msec.

Since the impedance pulse waveform leads the finger pulse temporally, it is more central and probably thoracic in its source of body impedance change. Further study is required to ascertain the source of the impedance pulse and its monitoring and diagnostic value.

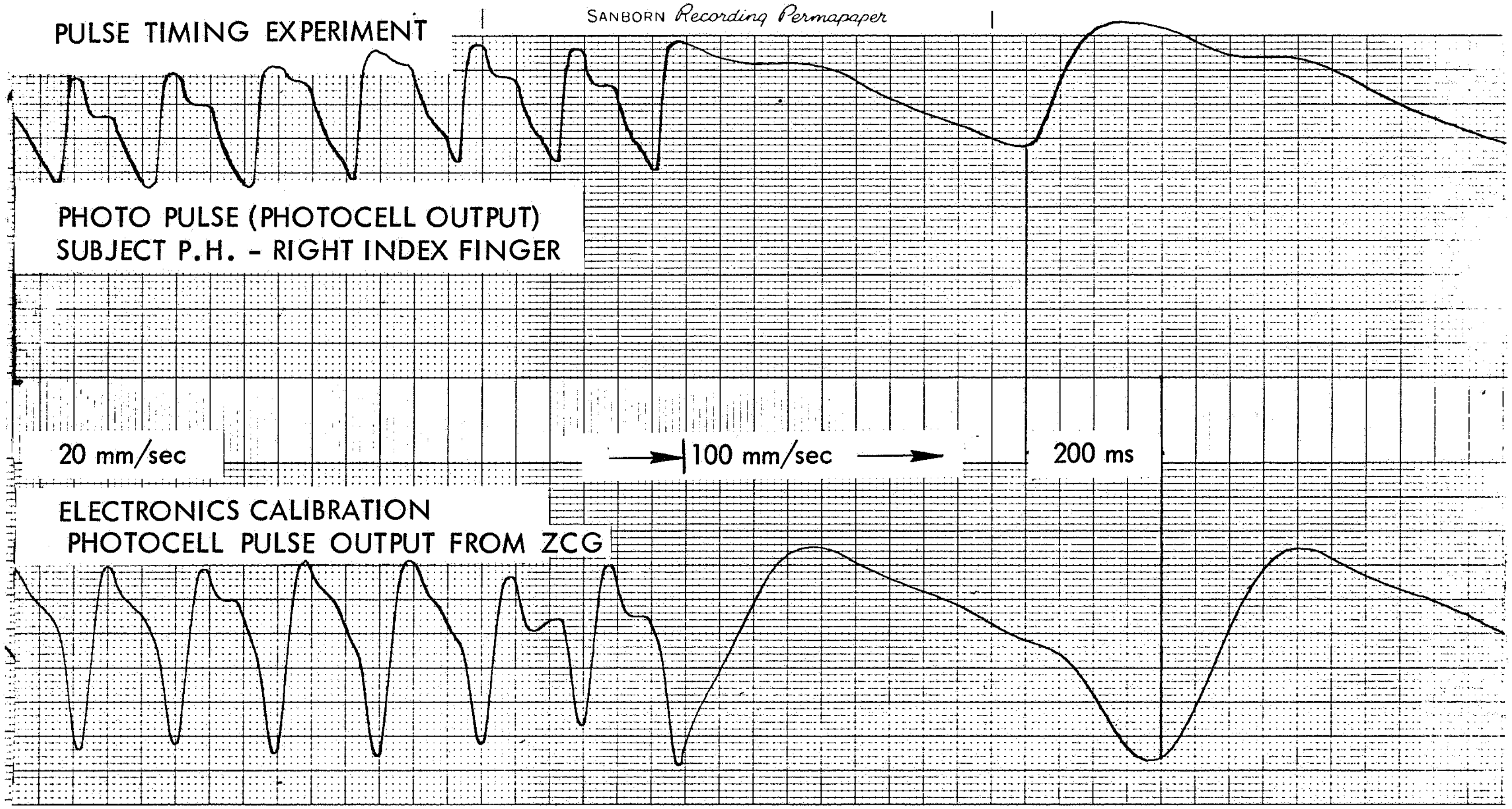


Figure 3-18 Expanded Illustration of Phase Lag (Impedance Channel at the Bottom)

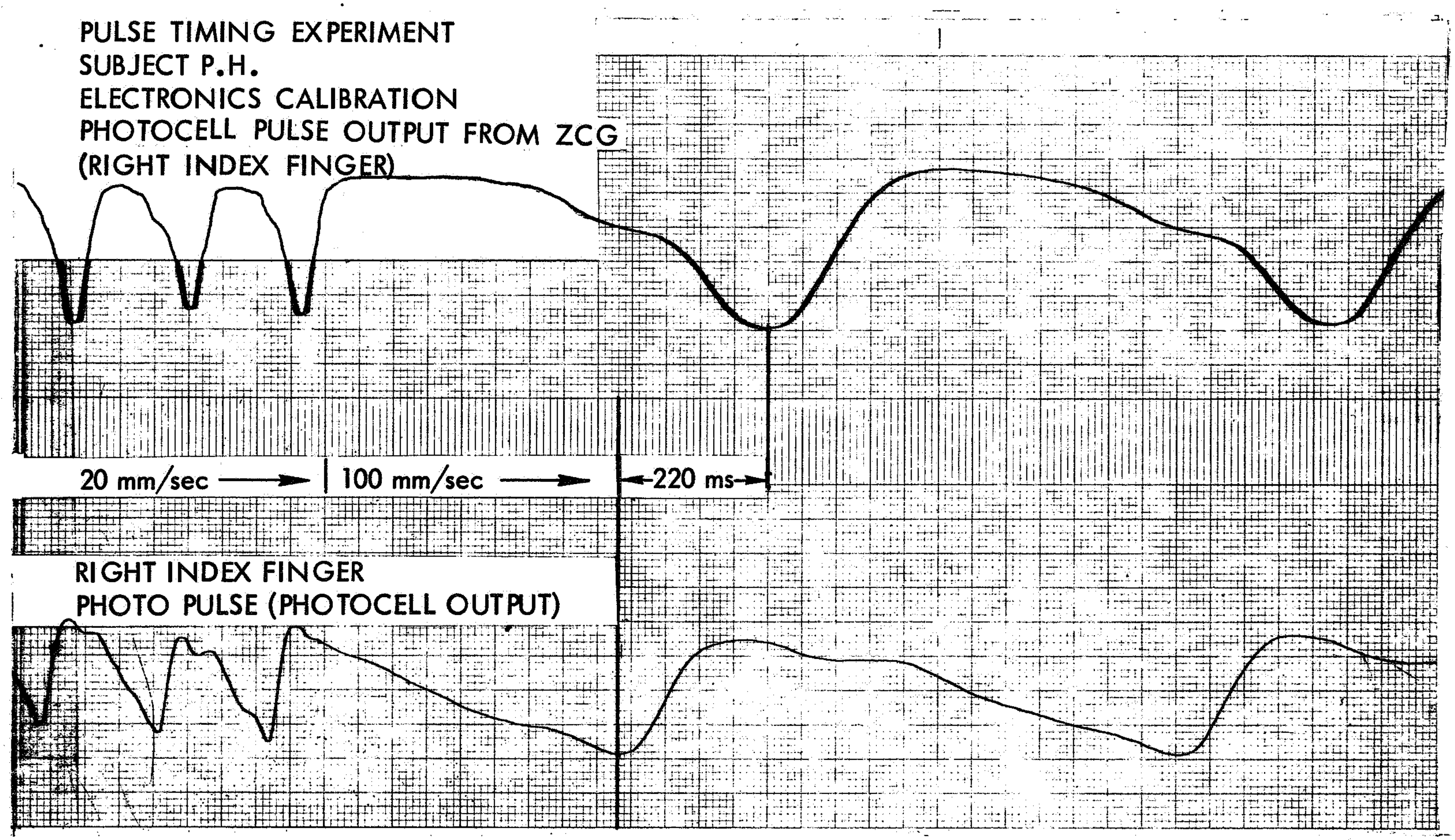


Figure 3-19 Expanded Illustration of Phase Lag (Impedance Channel at the Top)

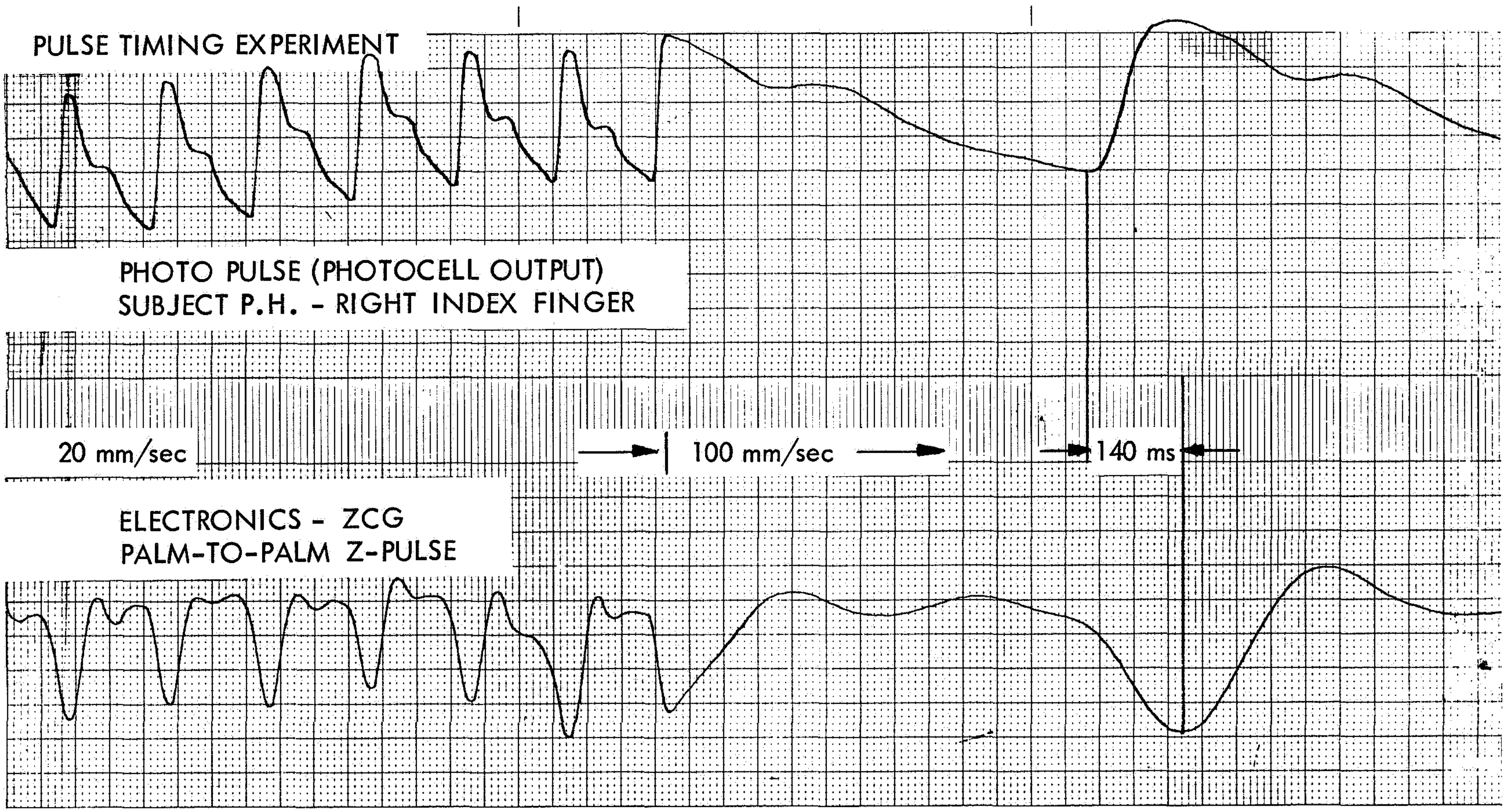


Figure 3-20 Expanded Illustration of Finger Pulse and Palmar Impedance Pulse (Impedance Channel at the Bottom)

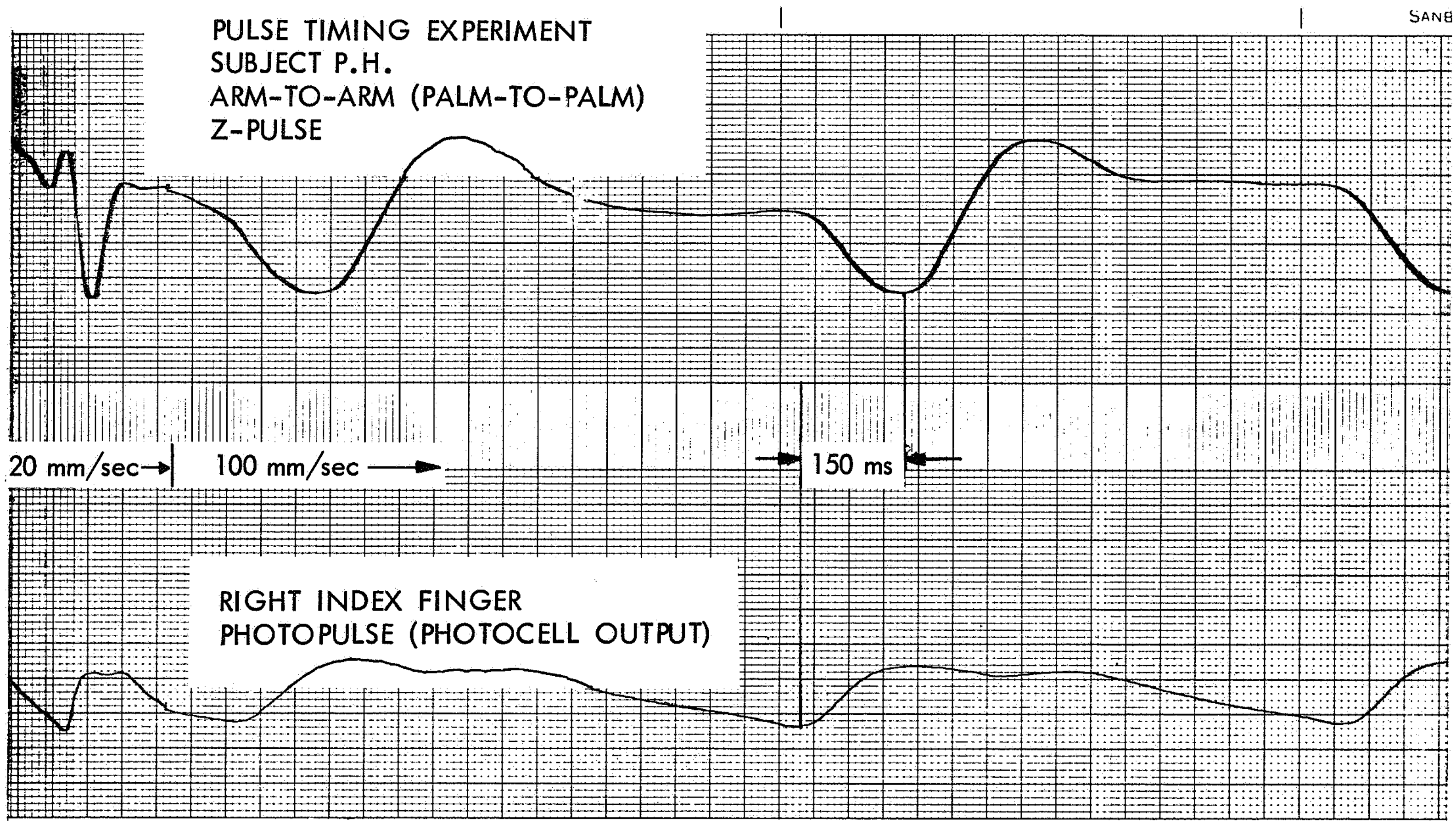


Figure 3-21 Expanded Illustration of Finger Pulse and Palmar Impedance Pulse (Impedance Channel at the Top)

SECTION 4
CARDIAC CLINIC COMPARISON DATA

This section presents data recorded by Dr. Herbert Semler, Cardiology Department, St. Vincent Hospital, Portland, Oregon. A prototype of the commercial version of the unattached monitoring system was used to obtain the data presented here. Comparison data were recorded on a Sanborn Model VISO-100.

The acquisition of data presented in this section represents progress toward the following goals:

1. Establishment of the unattached monitoring concept as a useful technique in the clinical environment.

2. Expansion of applications for which the unattached monitoring system is suitable.

<u>Results of Clinic Study:</u>

1. The data were easily acquired since minimal preparation of the patient was required.

2. Patient attitude was generally positive toward these new instruments; although the novelty of an "instrumented chair" undoubtedly contributed to their enthusiasm, patients were genuinely pleased with the simplicity of the measurements.

3. The unattached monitoring system was used before, and immediately following, exercise. This new application appears to have a promising future in exercise-cardiology studies. Because data can be sensed immediately following strenuous exercise, the system is useful in examining exercise recovery phenomena.

4. The instrument successfully detected depressed S-T segments of the ECG, following exercise during which the patient reported chest pain. (See Figures 4-12, 4-13, and 4-14.)

5. The system used at St. Vincent Hospital appeared to be susceptible to EMI, since the electrocardiographic data at times had 60 Hz noise superimposed on it. Circuit changes designed to suppress EMI are planned, since relatively intense electromagnetic fields are common in major hospitals and clinics.

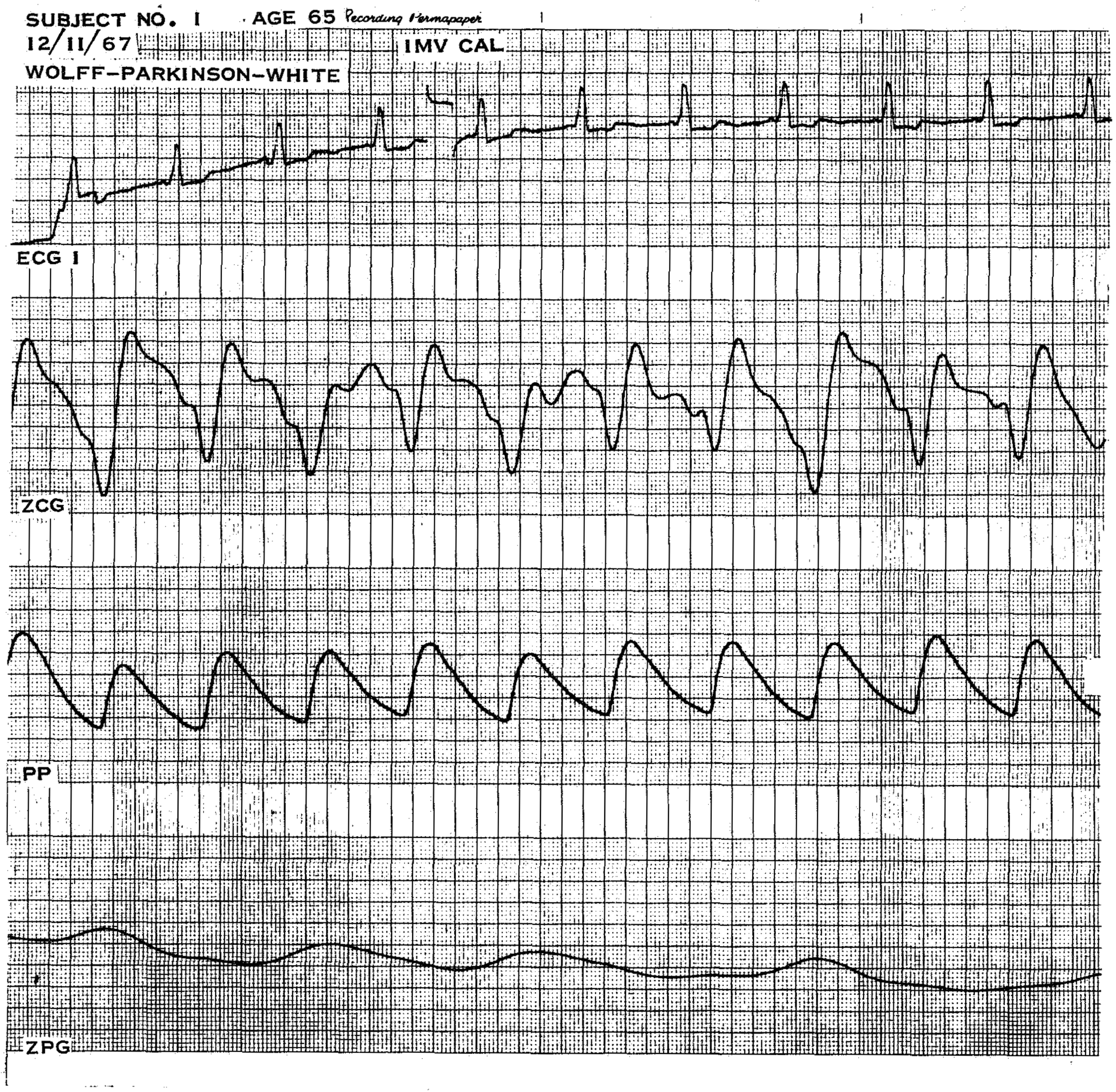


Figure 4-1 Subject No. 1, Wolff-Parkinson-White Syndrome

Subject No. 1. Figure 4-1 presents data for subject No. 1, a female, age 65, with the Wolff-Parkinson-White Syndrome. The record clearly shows ST segment depression and the ZCG (central pulse), PP (peripheral pulse), and ZPG, respiration.

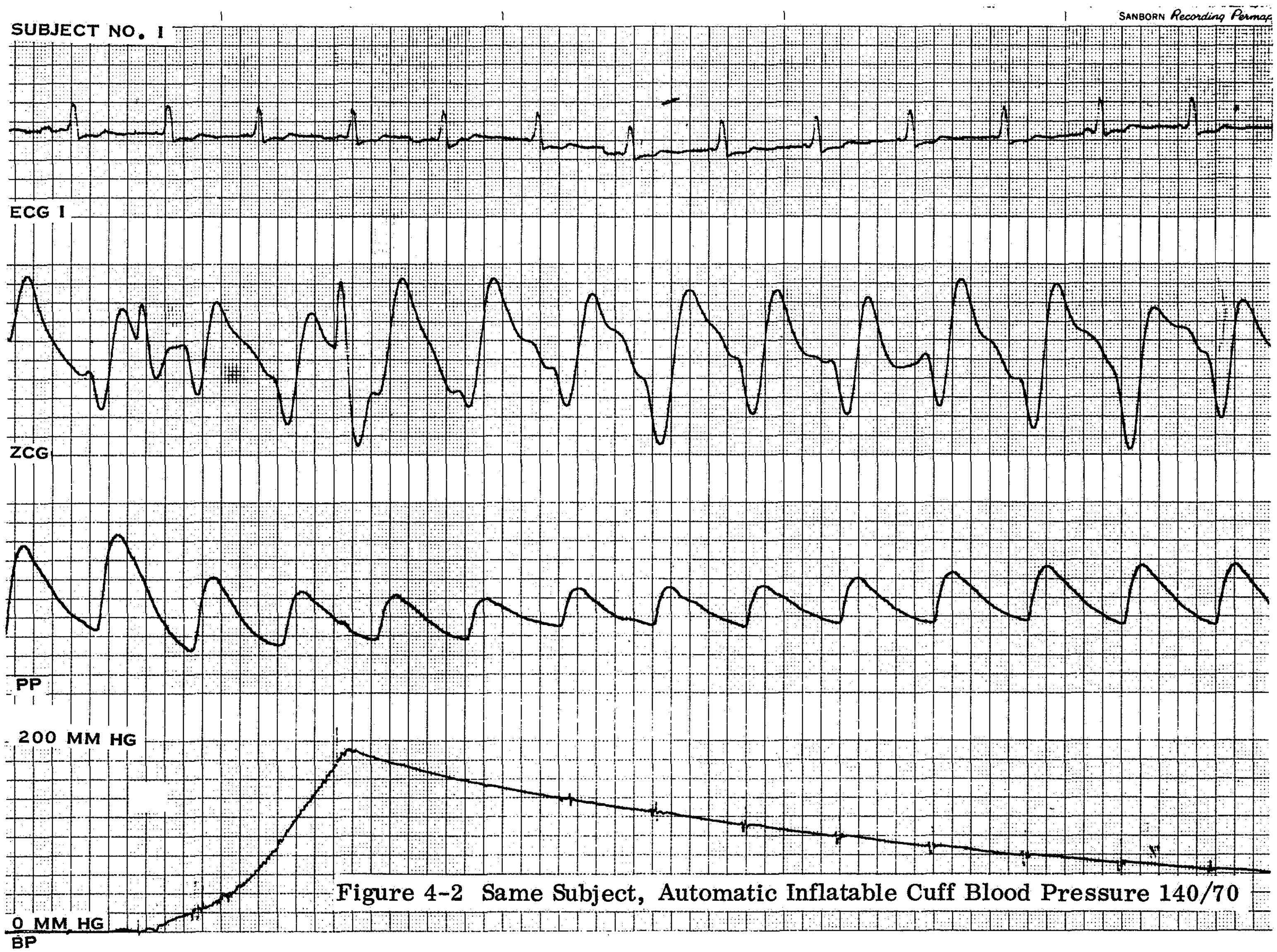


Figure 4-2 Same Subject, Automatic Inflatable Cuff Blood Pressure 140/70

This record (subject No. 1) shows the blood pressure measured with an automatic inflatable cuff system, which reduces the peripheral pulse amplitude.

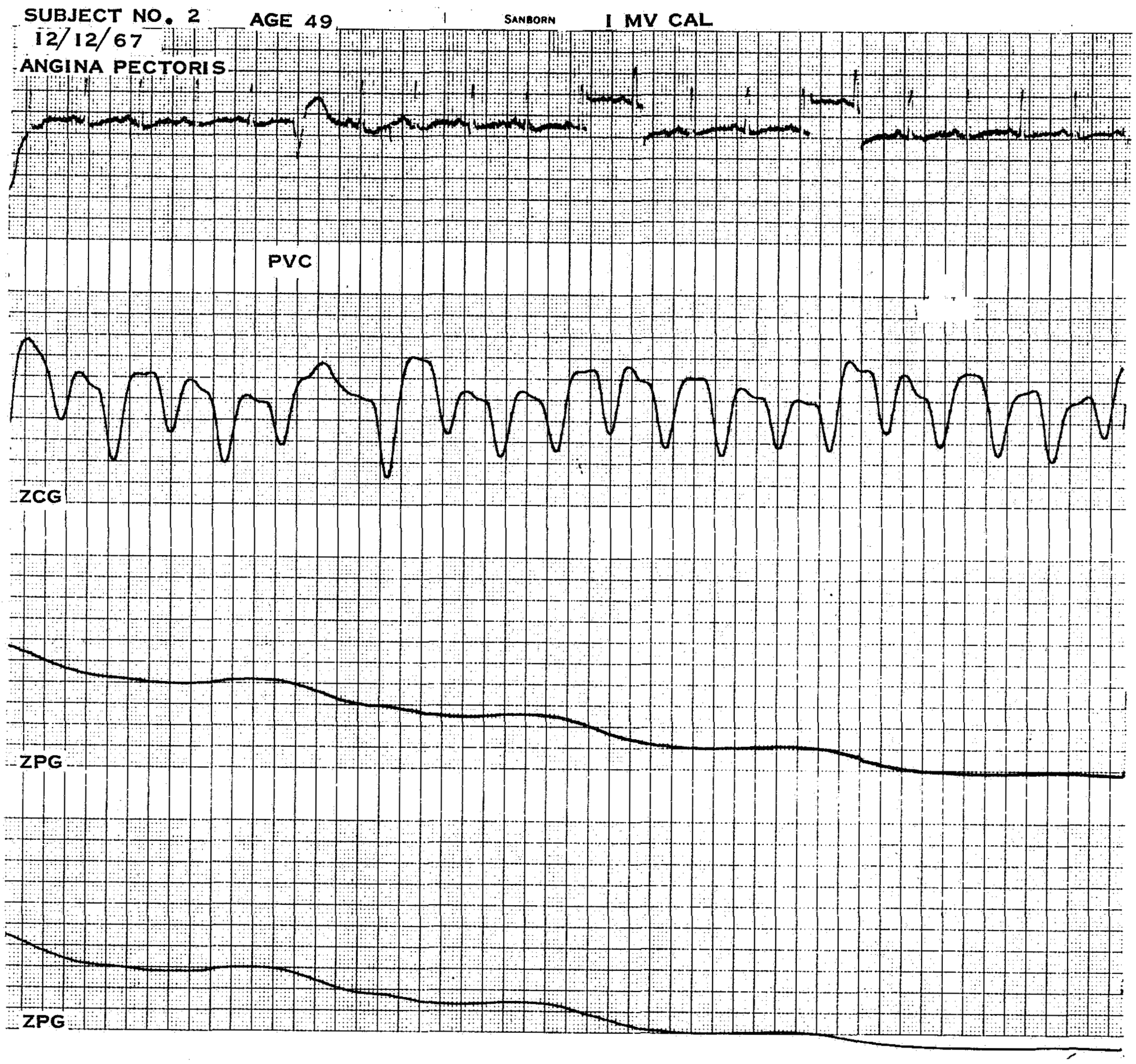


Figure 4-3 A recording of a patient during Angina Pectoris. Since there is no myocardial infarction, there are no changes in the QRS complex; however, there are ST segment T wave changes which are consistent with myocardial ischemia. A premature ventricular contraction is seen and is so marked.

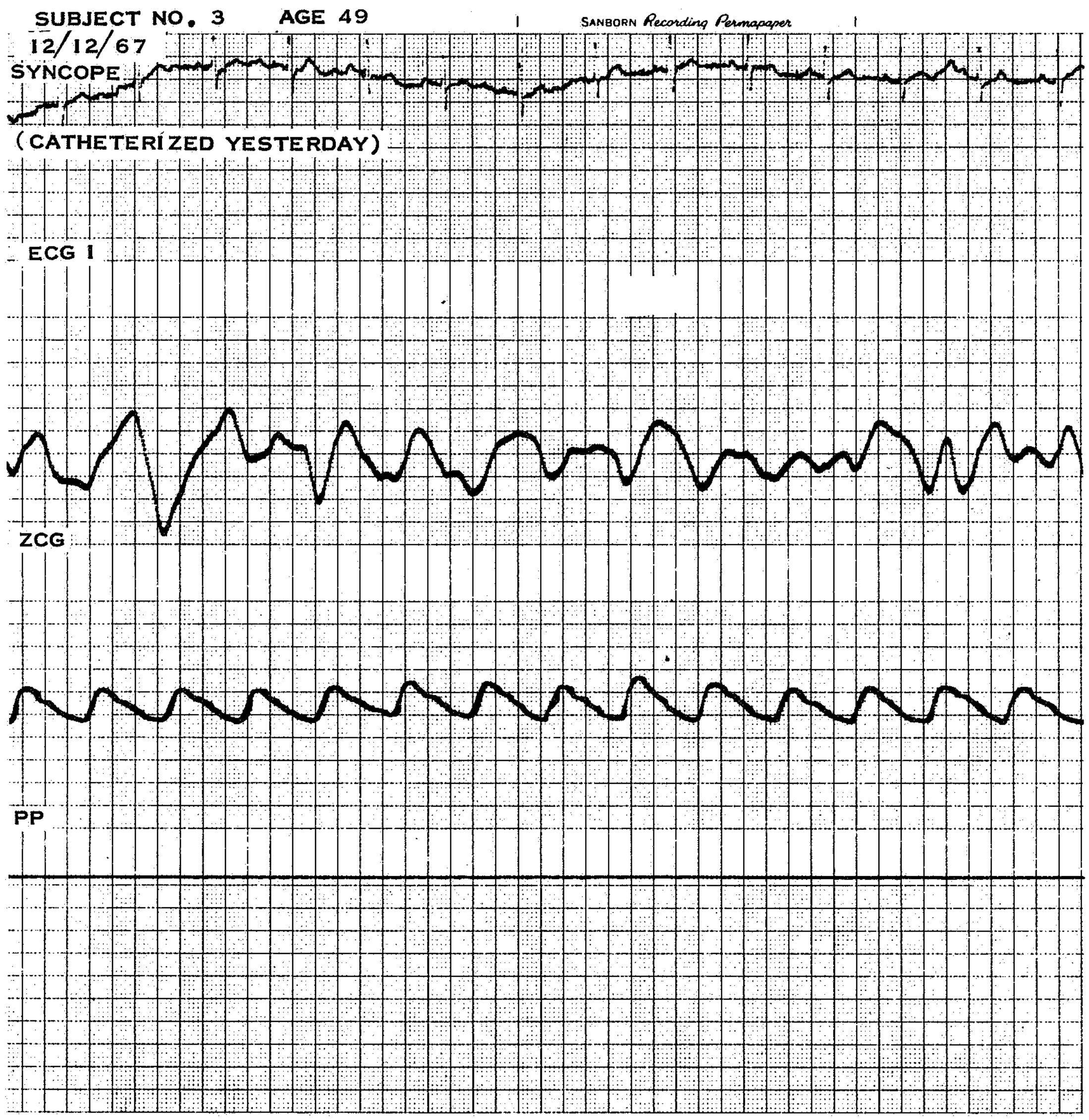


Figure 4-4 Record of Patient Who Complained of Syncope (catheterized on the previous day)

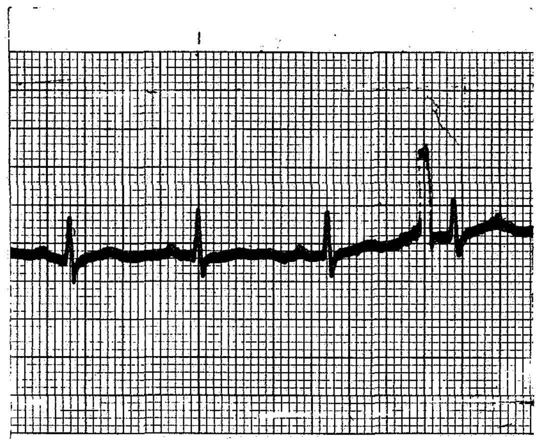


Figure 4-5 Subject No. 3 ECG Record. Recording made on conventional Lead I EGG instrumentation: normal tracing

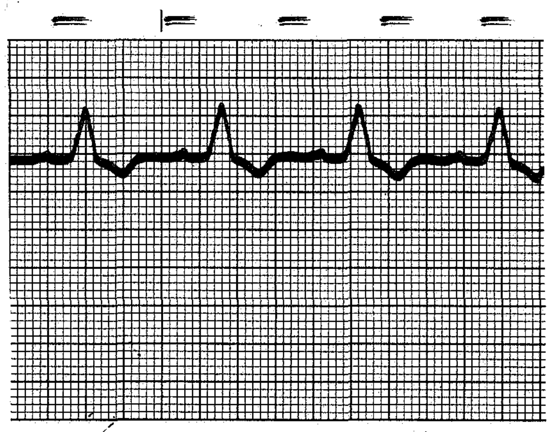


Figure 4-6 Subject No. 4 ECG Record. Recording uses conventional Lead I EGG instrumentation: Left Bundle Branch Block is present.

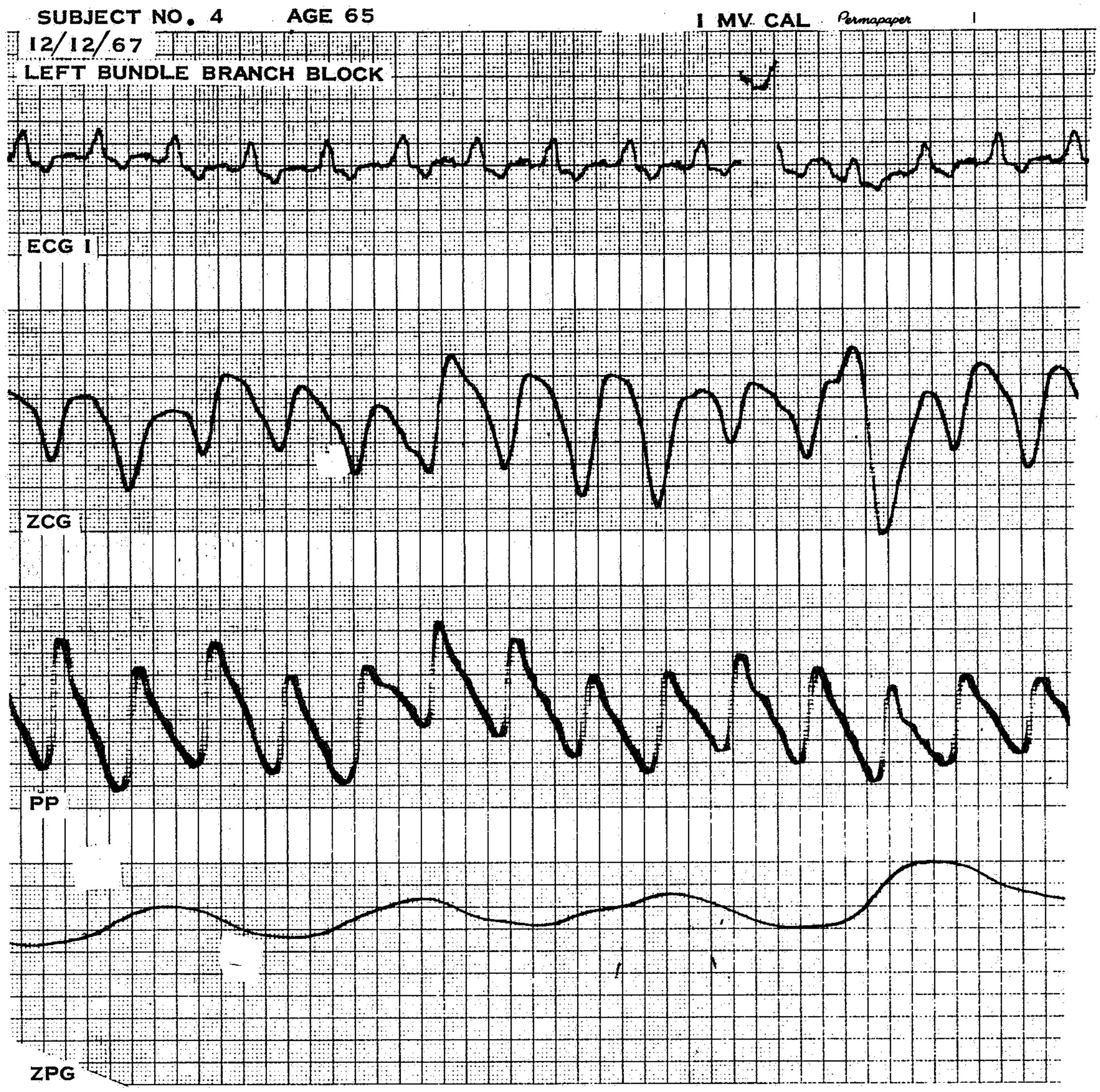


Figure 4-7 Subject No. 4 provides an example of typical complete Left Bundle Branch Block. Note the notched, irregular, and broad QRS complex. The T wave deflection is opposite to the main deflection of QRS as found in a left BBB.

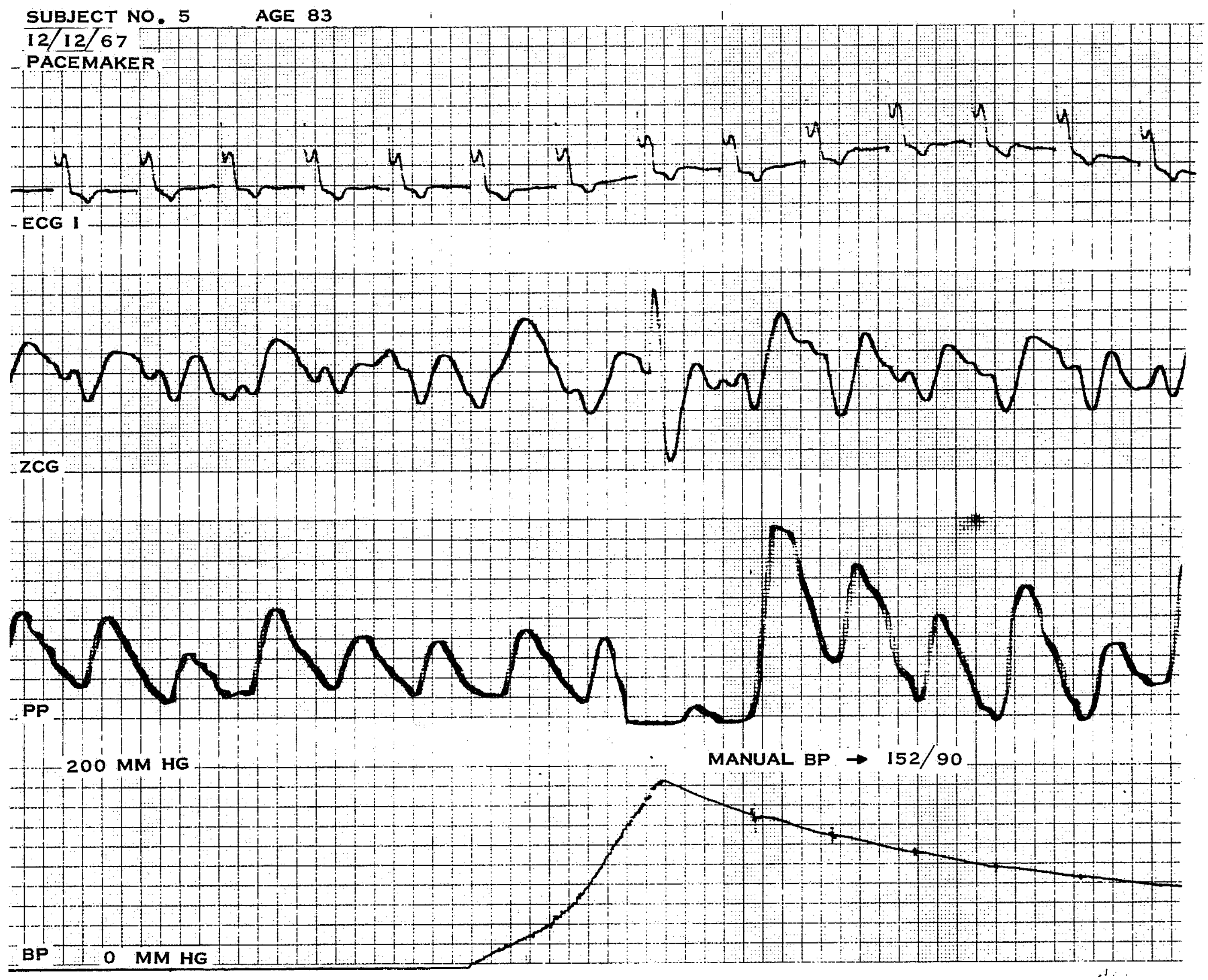


Figure 4-8 Effect of Pacemaker on ECG Lead I and Other Parameters.

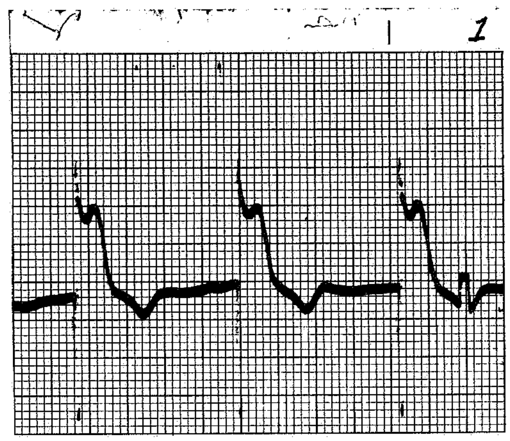


Figure 4-9 (Subject No. 5) Pacemaker Effect On ECG Lead I, using Conventional Technic.

Figure 4-9 is a record of subject No. 5 of ECG Lead I taken with Dr. Semler's conventional electrocardiograph. Note the similarity between this waveform and the previous figure's ECG waveform.

These records for Subject No. 5 were taken on an 83-year cardiac patient showing the effects of an implanted cardiac pacemaker on ECG Lead I. the ZCG (central pulse), the PP, the peripheral pulse. The blood pressure is also shown. This was taken with the automatic inflatable cuff, which is a part of this MediScreen system.

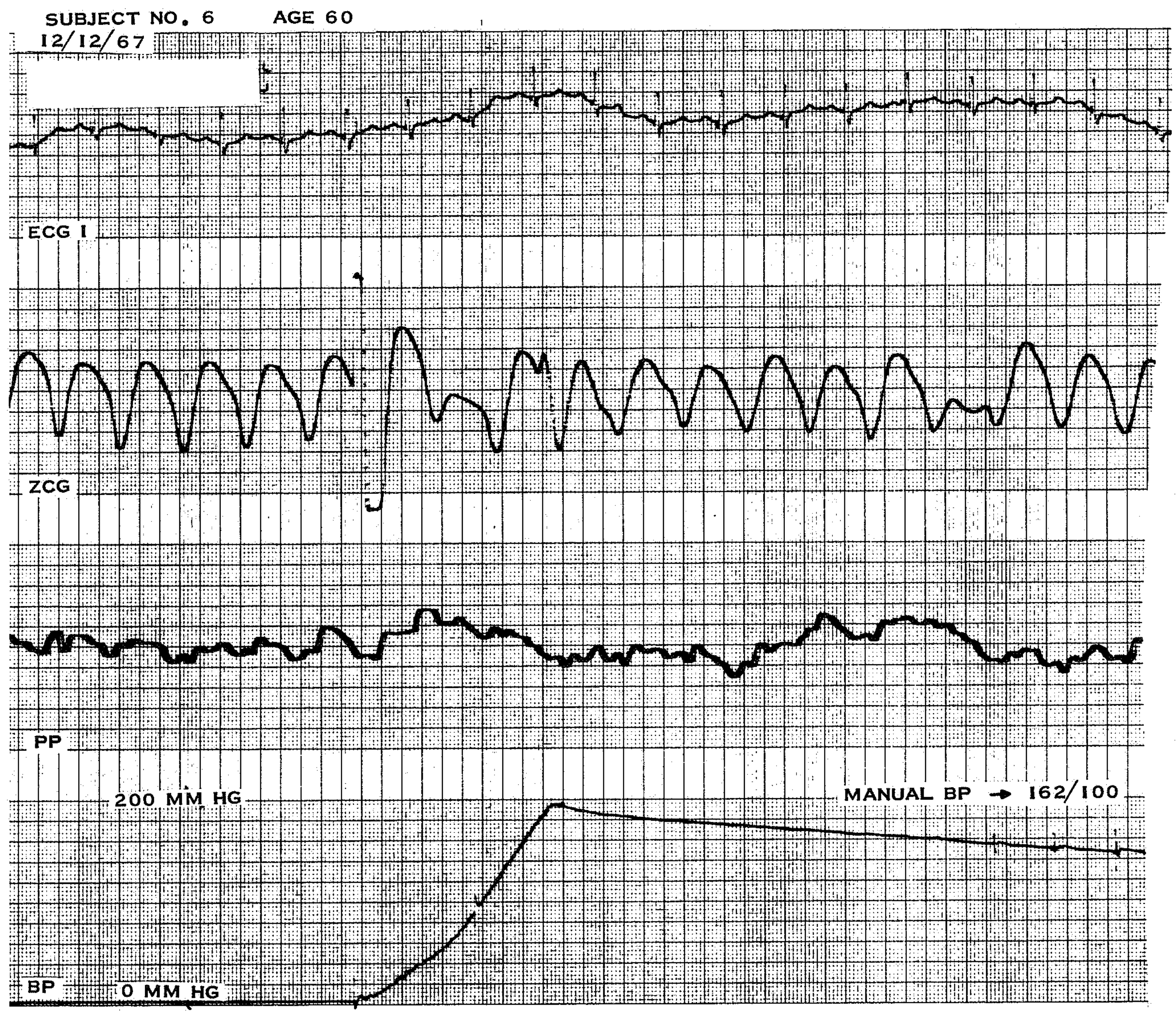


Figure 4-10 Coronary Arteriosclerotic Heart Disease

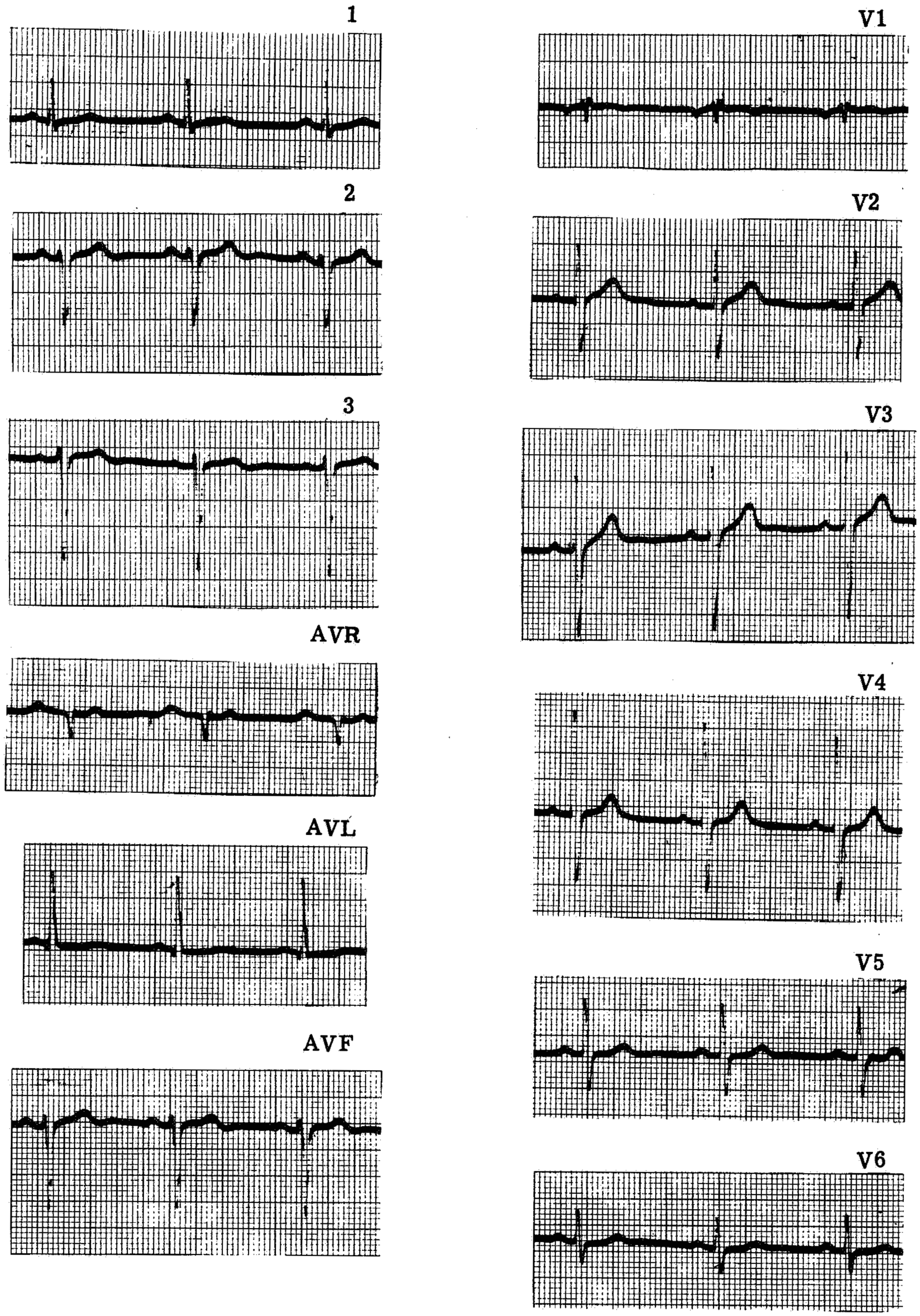


Figure 4-11 (Subject No. 6 with 12 Lead ECG) History of Coronary Arterio-sclerotic Heart Disease. Conventional Instrumentation.

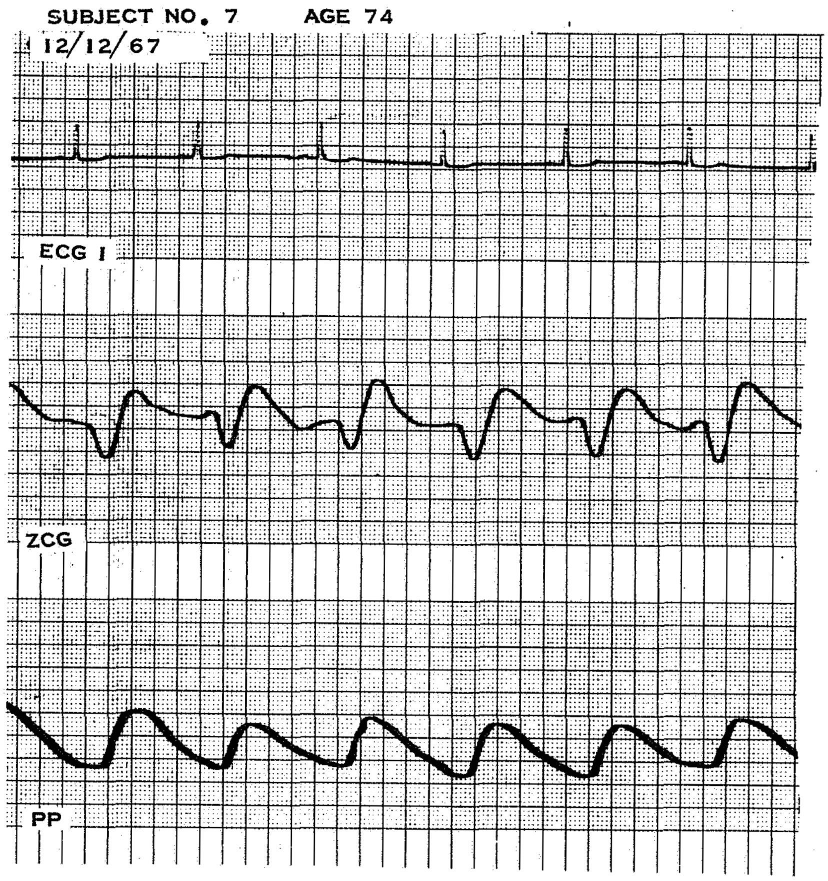
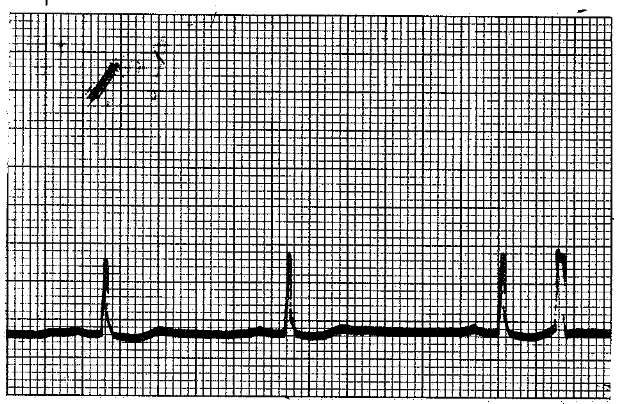
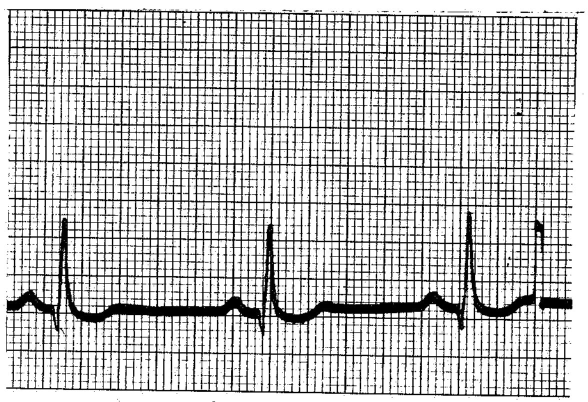


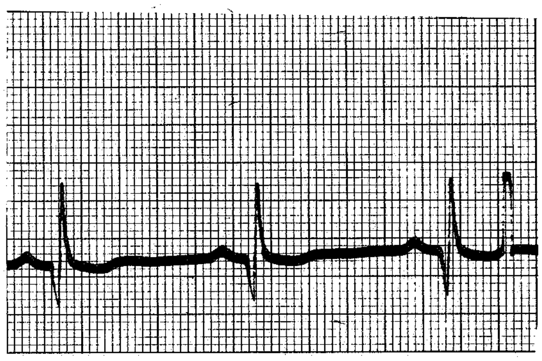
Figure 4-12 Subject 7 - Age 74 - Patient with History of Angina Pectoris.



Lead I



Lead II



Lead III

Figure 4-13 Subject 7 - Age 74 - Angina Pectoris. Note ST segment and T wave changes in Leads I, II and III - conventional ECG instrumentation.

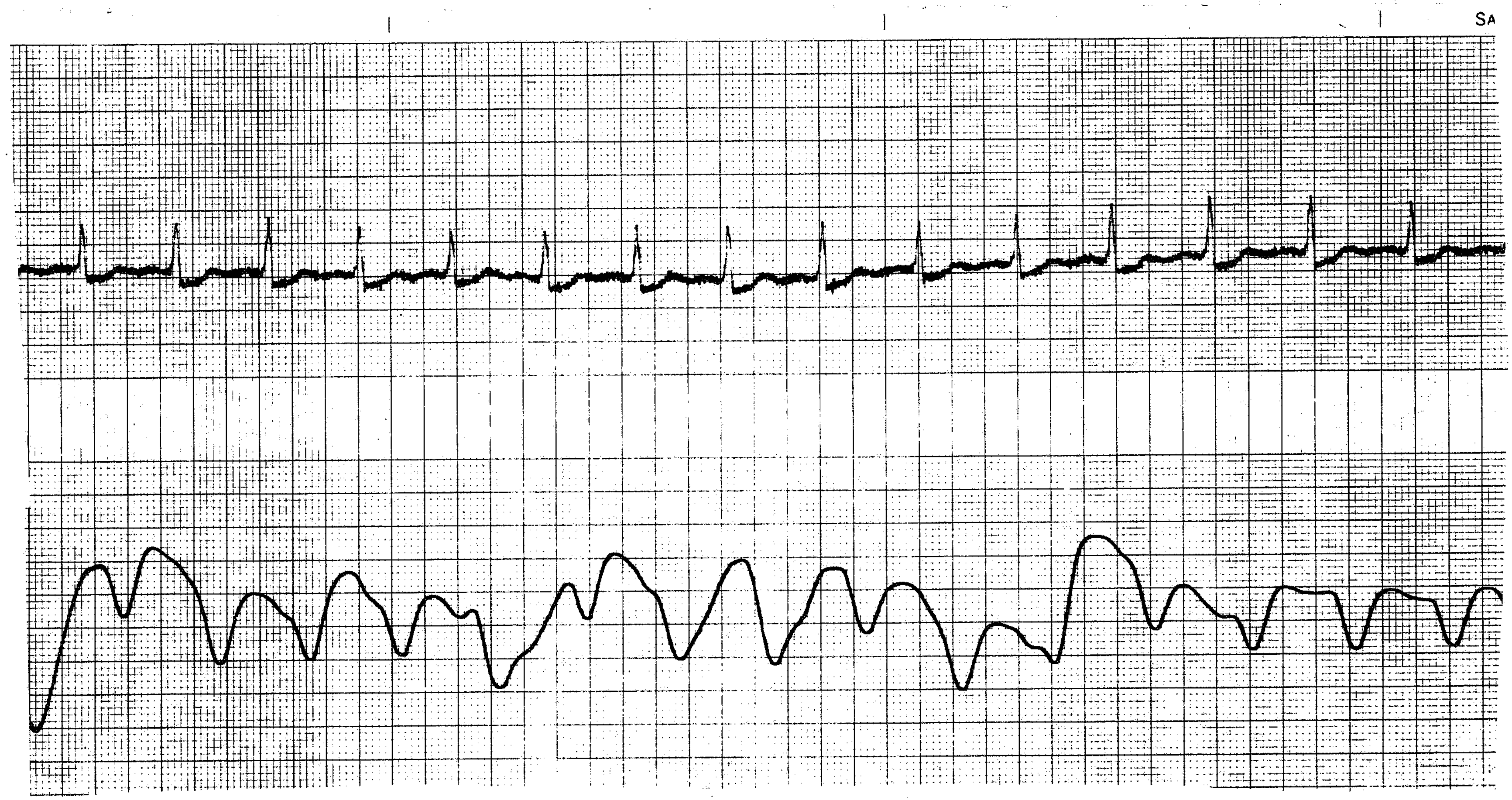


Figure 4-14 Subject No. 8) Recording of ECG and Central Pulse of Patient After Two Minutes of a Double Master's Two-step Exercise Test. Patient had chest pains following exercise. Note S-T changes.

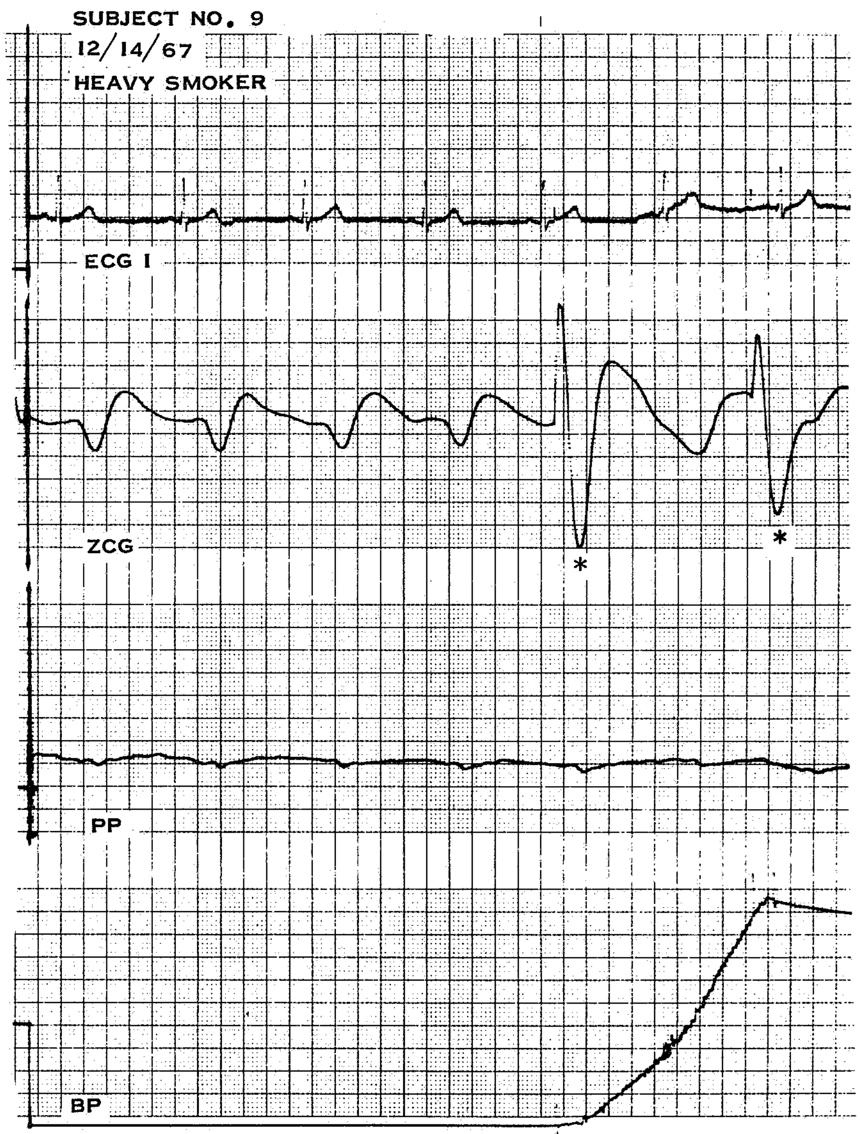


Figure 4-15 Subject 9 (Recorded 12/14/67). Lead I ECG and ZCG (central pulse) artifacts are related to sudden inflation of blood pressure cuff. Reduced PP (peripheral pulse) can be found in persons who smoke heavily.

*Due to cuff inflation artifact

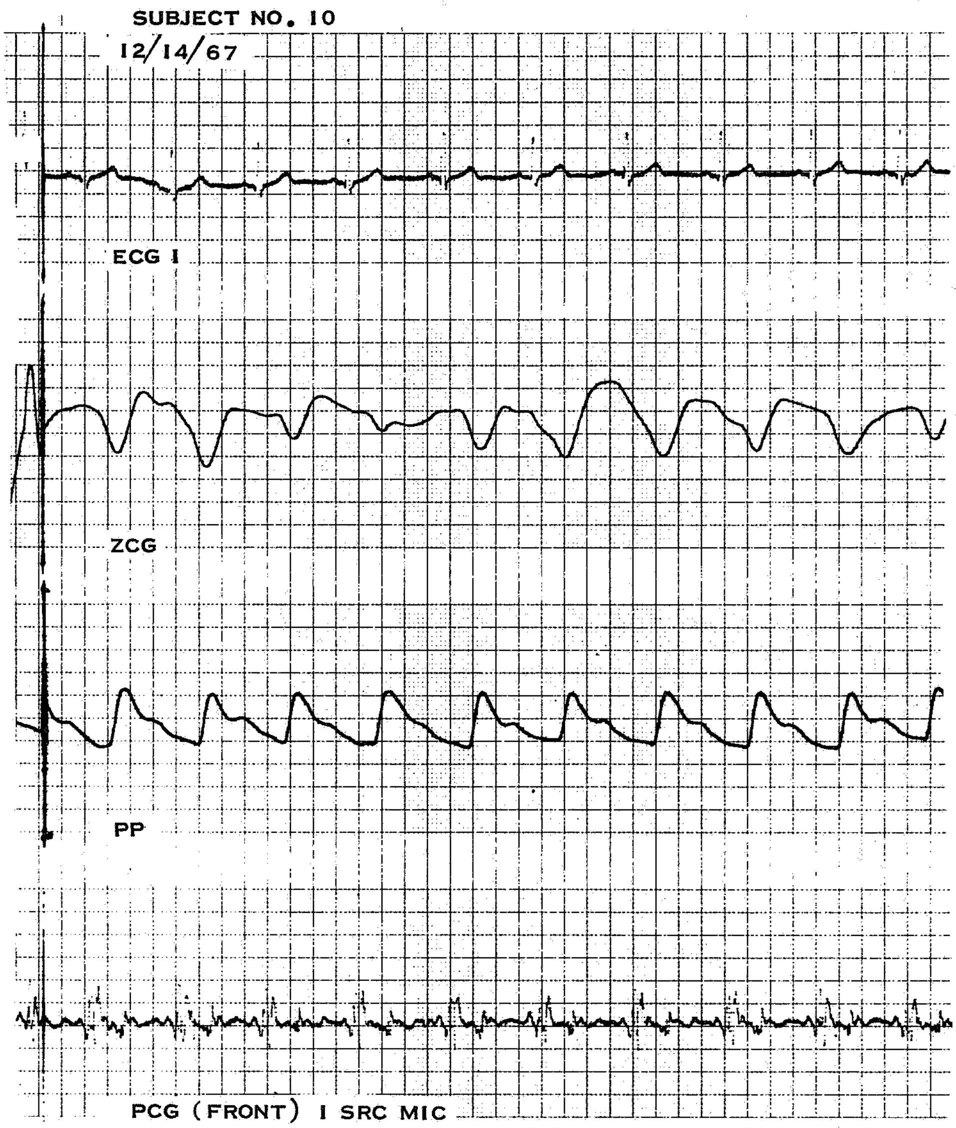


Figure 4-16 (Subject No. 10) Shows a recording of an apparent normal individual using the Unattached Monitoring System. In this instance a microphone was placed on the front of the chest; note the correlation among ECG Lead I, ZCG (central pulse), PP (peripheral pulse) and PCG (phonocardiographic waveform).

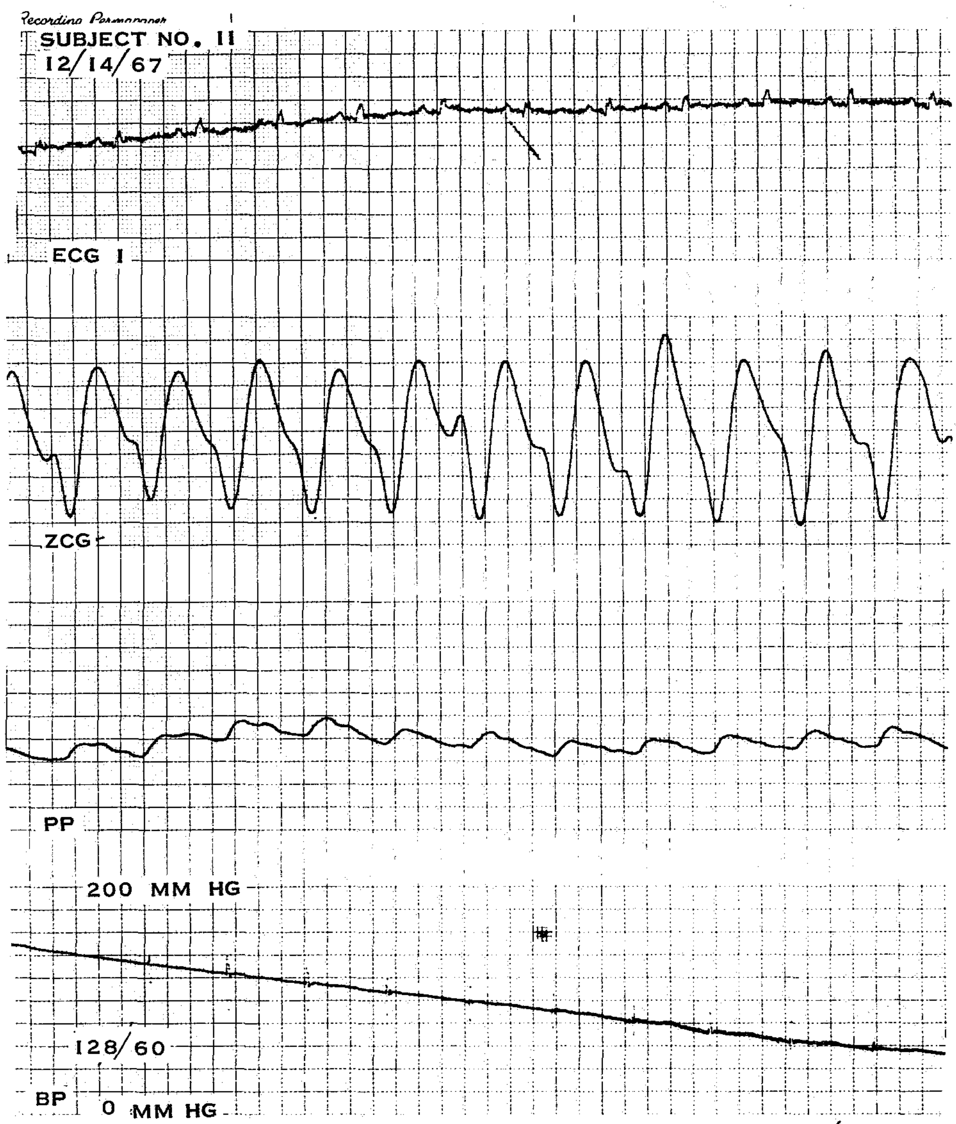


Figure 4-17 (Subject No. 11) Unattached Monitoring System Recording on 12/14/67. This patient has sustained a second myocardial infarction. This recording was made two weeks after second infarction - B. P., central pulse and finger pulse are recorded with ECG Lead I.

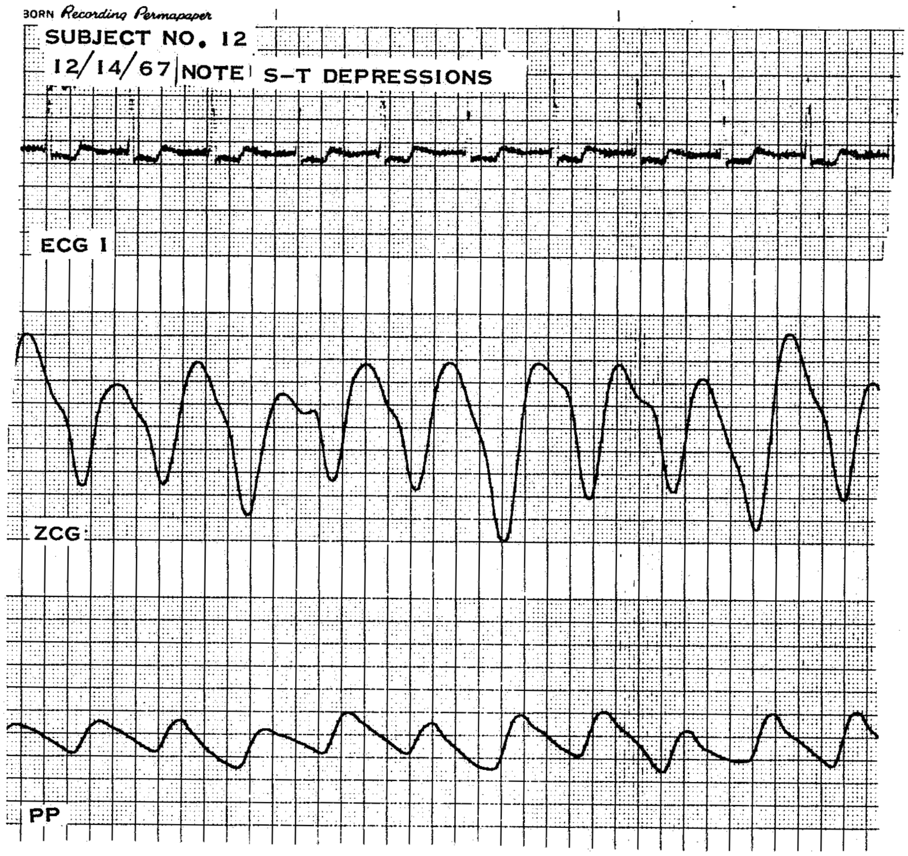


Figure 4-18 (Subject No. 12) Unattached Monitoring System Recording on 12/14/67. This patient, age 69, has aortic stenosis and signs of left ventricular hypertrophy, and is showing early cardiac decompensation.

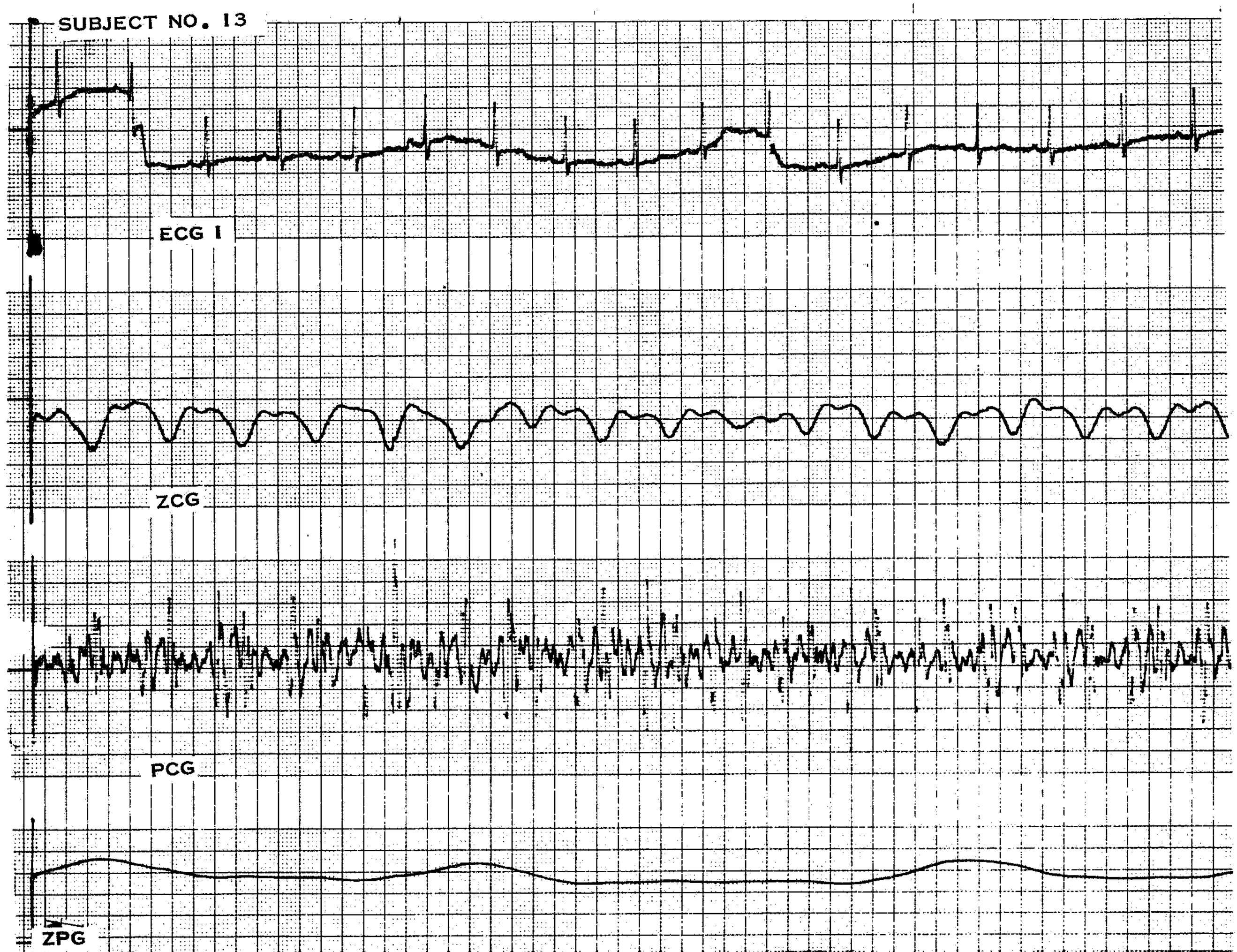


Figure 4-19 (Subject No. 13) Normal Waveform Pattern of ECG Lead I, ZCG (Central Pulse), PCG Phonocardiographic Waveform and ZPG (Respiration Waveform).

SECTION 5
CONCLUSIONS

The technique described in this report for acquiring six channels of physiological data from only two sets of casual contact, unattached sensors, appears to be feasible for intermittent or periodic monitoring. The technique is particularly applicable where rapid screening is required. During the course of this study, a system employing the described technique was used to rapidly acquire data immediately following strenuous exercise to study exercise recovery phenomena. During these tests the system successfully detected depressed ST segments of the ECG Lead I Waveform. Enthusiasm for the concept was shared by patients whose attitudes were very positive toward use of the technique.

Medical technicians operating the equipment were also pleased with the simplicity of operation. Because autoranging circuits are incorporated in the signal conditioners, the only panel controls are:

1. Off-on power switch,
2. ECG-GSR select switch, and
3. Three calibration pushbutton

All other adjustments are made by use of the standard controls available on any strip chart recorder.

Because the data were recorded simultaneously (with the exception of ECG or GSR), important timing relationships could be established between data channels. One should note however, that study of the ZCG electronics phase shift (at data frequency) showed a mean time lag of 210 milliseconds. The time delay in the ZCG channel can be compensated by:

1. Direct mathematical subtraction, or preferably by
2. Electronic circuit compensation.

Electrode/skin impedance studies showed the coupling to be significantly capacitive, even at near dc frequencies. Although the internal frequency response of the ECG amplifier used in this study was flat over the 0.15 to 100 Hz band, distortion was detectable on the S-wave, due to skin/electrode capacitance. To eliminate the effect of this capacitance, the ECG signal conditioner is being redesigned, as described in Section 2.3.1.

The respiration waveform, detected by measuring palm-to-palm impedance, appeared to be identical to that taken by attaching electrodes to the areas just below the axillae, and signal conditioning by a standard impedance pneumograph.

The central pulse, also measured as an impedance change from palm-to-palm (in the 0.8 to 50 Hz data band) may merit further investigation. Although little validation data were gathered on this measurement during the course of the study, crude correlations and evaluations indicate that further processing of the data might indicate level of stroke volume or cardiac output.

The thoracic sound measurements, taken from the back of the chair, appear to be the weakest part of the system. Although this measurement is being investigated further, it is doubtful that the quality of the measurement will be such as to yield anything more than first and second heart sound timing information.

The research and development leading to the unattached monitoring system described in this report have stimulated research into adjacent areas of instrumentation. As noted above, the central impedance waveform may be useful (with further refinements) in determining cardiac output similar to technique developed by Nyboer and others (Reference 4).

With the increasing emphasis on more preventive medical care, the investigators in this study are hopeful that equipment such as that described will be of value to the future practice of rapid screening.

APPENDIX A

LITERATURE SURVEY

The conducted literature search, listed in this appendix, and the sustained literature review maintained during the contract period have revealed only the few similar developments and applications of techniques in combination referenced herein. However, the literature review shows that considerable work has been accomplished on each item and technique independently and that, in some instances, products have been developed. A summary of findings is given below by physiological measure followed by the annotated bibliography.

Electrocardiograph. Unattached, dry, contact electrodes are used to obtain ECG's from patients in the Kaiser-Permanente Multiphasic Health Screening Clinics in both San Francisco and Oakland, California. The approach is considered practical for screening and diagnostic work and has been in use for several years. Chair-mounted electrodes have also been used in earlier screening studies as reported in Reference 2 and, just recently, by the Missouri Regional Medical Program as reported in Reference 3.

Galvanic Skin Response. For the past decade it has been common practice in polygraphic work to use dry finger electrodes. Products such as the Keeler polygraph and the Stoelting polygraph illustrate practical applications. The Stoelting system includes a chair with arm trough electrodes for GSR data acquisition, however, the remainder of the sensors are pneumatic.

Impedance Pneumograph (Respiration). No work has been found on acquiring impedance respiration from the hands. Respiration has been acquired simultaneously with ECG from trans-thoracic paste electrodes. Equipment was developed for and utilized in Project Gemini (Reference 1).

Impedance Pulse. No work has been found on acquiring impedance pulse from the hands.

Thoracic Sounds. Electronic studies of heart and thoracic sounds are manifold. However, no reference indicated the systematic study of dorsal thoracic sounds. Older medical literature illustrates the common practice of dorsal stethoscopic observations.

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ANNOTATED LITERATURE SURVEY

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