A SPHERICAL POLARCARDIOGRAPH COMPUTER

by

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ABSTRACT

Until recently, the major portion of the study of the electrical activity of the heart has been done with the aid of electrocardiograms and vectorcardiograms. However, such information as the variation of the magnitude and angle of the heart vector with time is not directly discernible from either of these recordings. A polarcardiograph was developed by W.K.R. Park to present the plane projection of the heart vector in magnitude and angle as a continuous function of time. The polarcardiograph proved to be useful but it was not sufficiently stable. An electronic device which would be stable and at the same time present the heart vector in three dimensions, magnitude, frontal angle and polar angle as continuous functions of time, would be useful in electrocardiographic research. The design of such a computer, the "spherical polarcardiograph", is described in this thesis.

The spherical polarcardiograph, which must compute the spherical polar coordinates of points from their respective Cartesian coordinates, has been developed using analog multipliers, subtractors and adders as well as a two-phase sinusoidal voltage source and a device for generating a voltage proportional to the phase difference of two sinusoidal signals.

With the exception of the third coordinate computation and the gated feedback circuitry, the system is similar

to that used by Park. Automatic balancing of the circuit occurs for a short interval during the rest period of the heart.

The spherical polarcardiograph has not been constructed in final form but tests on the individual units indicate that the instrument will be well within the accuracy required for normal electrocardiographic purposes.

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ACKNOWLEDGEMENT

The author expresses his indebtedness to members of the Department of Electrical Engineering at the University of British Columbia, especially to Dr. A. D. Moore for his guidance throughout the research project.

The spherical polarcardiograph was developed with the assistance of Dr. G. E. Dower, following his suggestion that such a device might be useful in electrocardiographic research. The project was supported by the Medical Board Fund of the Vancouver General Hospital.

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A SPHERICAL POLARCARDIOGRAPH COMPUTER

I INTRODUCTION

The instrument to be described, the spherical polarcardiograph, is a computer which transforms electrical voltages proportional to the Cartesian coordinates x, y, and z to voltages proportional to the radius r, the polar angle θ and the azimuthal or frontal angle $\varnothing$ in spherical polar coordinates. The spherical polarcardiograph will be used in the study of the electrical activity of the heart, and it is hoped that it will overcome the shortcomings of the electrocardiograph and the vectorcardiograph which are instruments used at present for this purpose. The relationship between the coordinate systems used in this application is shown in Figure 1.

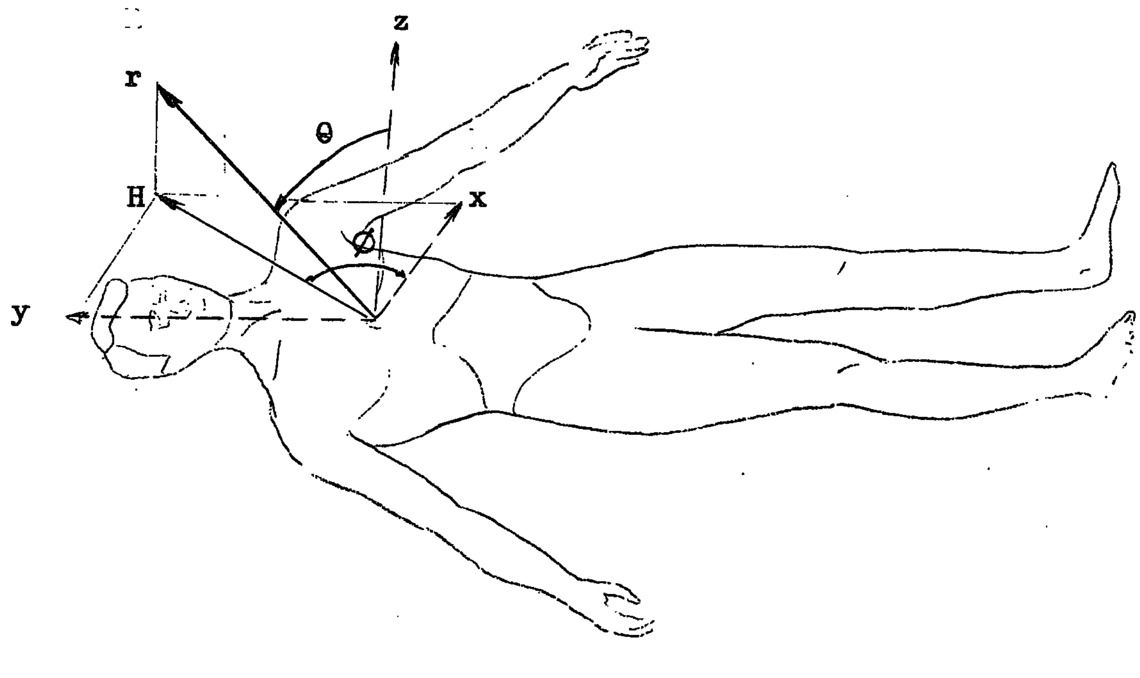

Figure 1. Relationship of Coordinate Systems to the Body.

The potential differences which appear at the surface of the body were first explained by Einthoven[1] using the simplified model of the heart and body appearing in Figure 2. The

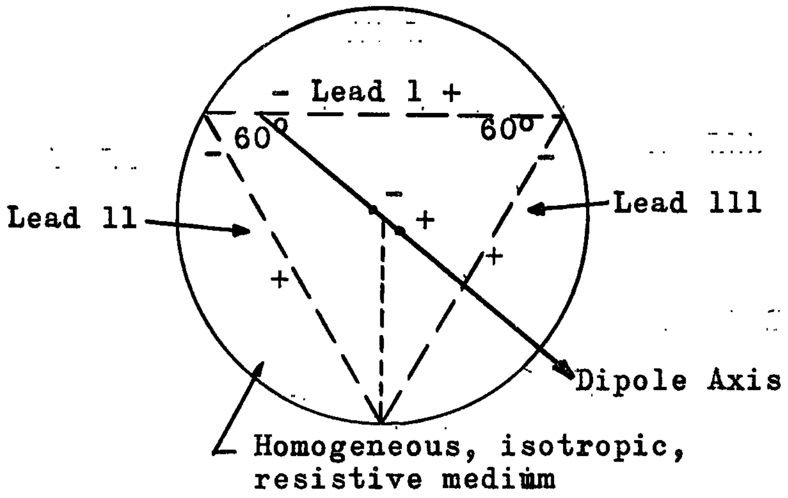

Figure 2. Simplified Model of the Heart and Body

following assumptions, although in error, have been useful in understanding the potential differences.

1. It is assumed that conduction is confined within the body and that the surface of the body can be represented by a sphere.

2. It is assumed that the limb electrodes are located electrically on the sphere at the corners of an equatorial plane of the sphere, this plane being called the frontal plane.

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[1] W. Einthoven, G. Fahr, A. De Waart (in English), "Über die Richtung und die Manifeste Grösse der Potentialschwankungen im Menschlichen Herzen und über den Einfluss der Herzlage auf die Form des Elektrokardiogramme," American Heart Journal, 1950, XL, 163.

3. It is assumed that the body is a homogeneous, isotropic, resistive medium.

4. It is assumed that the electrical sources within the heart can be represented by a single current dipole or heart vector of variable strength and orientation, but fixed in position at the centre of the spherical conducting medium.

The graphic recordings of the voltages between electrodes on the body as functions of time are called electrocardiograms. Graphs or tracings I, II and III, of Figure 3, represent typical voltage variations occuring between limbs as obtained with a standard electrocardiograph. After the pioneering work of Einthoven, it was found that a study of the voltages appearing at the front and back of the chest (sagittal axis) revealed further information about the heart not found in recordings from the frontal plane. To obtain a reference potential for voltages from the chest, Wilson[2], using Einthoven's equilateral-triangle hypothesis, connected three equal resistors from the limbs to a common terminal (Wilson's Central Terminal). With this point as a reference, an exploring electrode is used to obtain the remaining nine tracings of Figure 3. The three graphs obtained by placing the exploring electrode on the right arm, left arm and left leg are the augmented waveforms AVR, AVL and AVF, respectively.

The use of the electrocardiogram in this manner presents certain disadvantages. To illustrate some of the shortcomings,

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[2] F. N. Wilson, F. D. Johnston, A. G. MacLeod, P. S. Barker, "Electrocardiograms that Represent the Potential Variations of a Single Electrode," <u>American Heart Journal</u>, 1934, IX, 447.

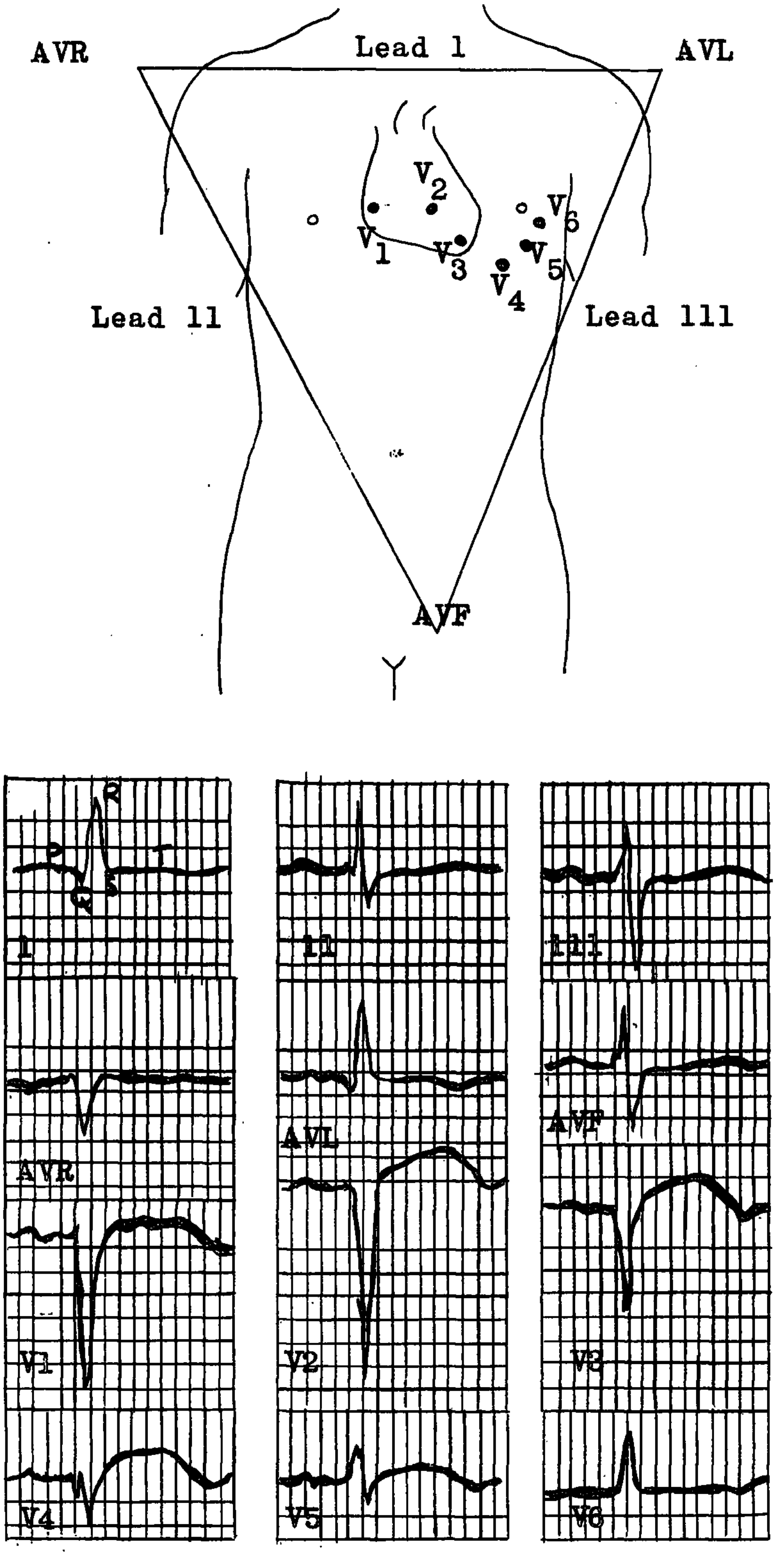

Figure 3. Electrocardiograms and the Associated Electrode Positions.

let us again consider the twelve electrocardiograms taken with the lead system in Figure 3. It is apparent that there is a great difference in wave shapes obtained from the points V1 to V6 although the electrode positions are in close proximity. Hence, we may conclude that the waveforms are sensitive functions of electrode position. When interpreting electrocardiogram changes in consecutive tracings, the cardiologist must bear in mind the fact that some of these changes may be due to error in electrode positioning. When describing the graphs, the magnitude and sign must be listed for each of the segments shown in III of Figure 3 for each of the twelve graphs. The cardiologist is faced with the difficult task of forming an overall picture of the heart activity from these numerous elements of information.

The preceding discussion gives little justification for the assumption that voltages proportional to the Cartesian projections of the heart vector can be found on the surface of the body. It has been shown[3,4] that such voltages do exist by using concept of the image surface. The assumptions upon which the image surface is based are that electrical conduction within the body is a linear phenomenon and that the heart vector is fixed in position at the origin of the

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[3]H. C. Burger and J. B. van Milaan, "Heart Vector and Leads, Part III," British Heart Journal, 1948, X, 229.

[4]Ernest Frank, "The Image Surface of a Homogeneous Torso," American Heart Journal, 1954, XLVII, 757.

coordinate systems; the inaccuracies in the assumptions are minor. The image surface is the locus of the tips of vectors $\vec{c}$ such that the scalar product $\vec{c} \cdot \vec{r}$, where $\vec{r}$ is the heart vector, gives the potential differences between points on the surface of the body and the origin of the coordinate system. There is a one-to-one correspondence between points on the image surface and points on the body. Then for any pair of points on the surface of the body with mappings $\vec{c}_1$ and $\vec{c}_2$ on the image surface, the potential difference is $v_1 - v_2 = (\vec{c}_1 - \vec{c}_2) \cdot \vec{r}$. Hence, it is apparent that by suitably proportioning the potential differences between certain points on the body, signals proportional to the Cartesian projections of $\vec{r}$ may be obtained.

The science of vectorcardiography[5] has been developed to exploit the fact that the orthogonal projections of the heart vector can be observed simultaneously, thereby surmounting some of the difficulties encountered in electrocardiography and obtaining more information from a single observation. In plane vectorcardiography, two orthogonal components are applied simultaneously to the deflection plates of a cathode-ray tube. The resulting figure, with a time scale obtained by periodic modulation of the beam, is known as a vectorcardiogram. An example of vectorcardiograms and electrode positions[6] from which they are derived is shown in Figure 4. Note that, in

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[5] P. W. Duschosal, R. Sulzer, La Vectorcardiographie, S. Karger, New York, N. Y., 1949.

[6] G. E. Dower, J. A. Osborne, "Comments on Vectorcardiographic Lead Systems: A New System Proposed," forthcoming publication.

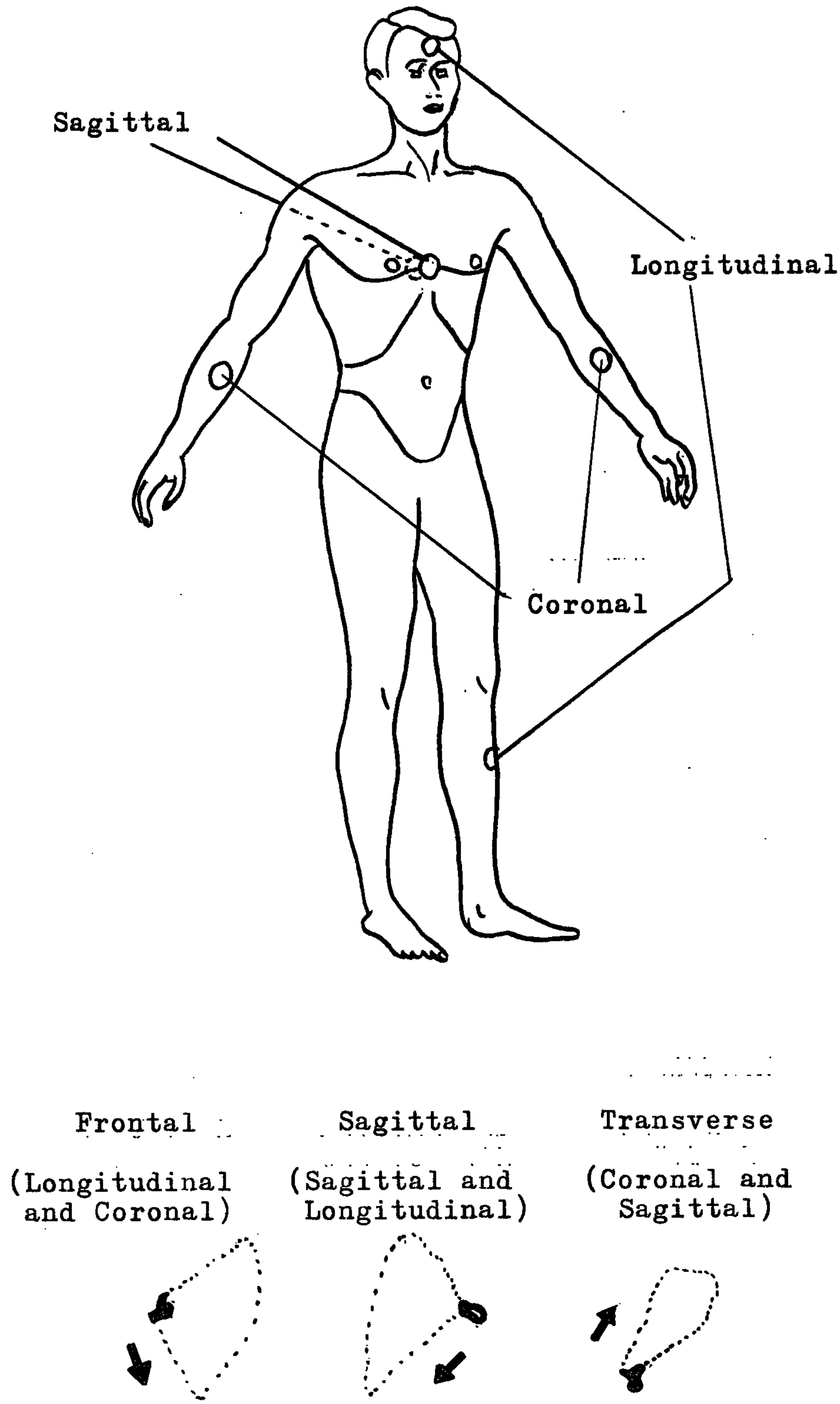

Figure 4. Plane Vectorcardiograms and the Associated Electrode Positions.

the lead system shown, the sagittal electrode is positioned in the centre of the chest so as to make it easier to place correctly.

The vectorcardiograph groups the information of the electrocardiograms into three recordings and at the same time introduces the direction of the plane projection of the heart vector. However, an examination of Figure 4 shows that much of the information regarding the slowly varying waveforms is obliterated around the origin. As an example of the information lost in this manner, consider the deviations of the S-T segment (shown in Figure 3) from the base-line or origin of the electrocardiogram. Such deviations may imply important abnormalities such as myocardial infarctions (death of a region of the heart muscle due to poor blood supply). It is common for heart ailments to alter the slowly varying segments of the waveforms, which are lost in the vectorcardiogram, leaving the rapidly varying segments in their normal forms.

Three-dimensional or spatial vectorcardiography has, in principle at least, advantages over plane vectorcardiography in that all of the information can be viewed at once but it suffers from the same difficulties as plane vectorcardiography.

To preserve the advantages of the electrocardiogram, which gives good definition of the slowly varying waveforms, and of the vectorcardiogram which gives direction, the polarcardiograph computer[7,8] has been developed to transform

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[7] R. McFee, "A Trigonometric Computer with Electrocardiographic Applications," Review of Scientific Instruments, 1950, XXI, 420.

[8] W. K. R. Park, "A Polarcardiograph Computer," M.A.Sc. Thesis, University of British Columbia, 1954.

signals representing the Cartesian projections of the heart vector in two dimensions to an equivalent representation in polar coordinates. The polar coordinates (plane direction and magnitude) are displayed on continuous recordings as functions of time to preserve the detail lost in the vectorcardiogram. The polarcardiograph developed by Park was tested under clinical conditions at the Vancouver General Hospital by Drs. Dower and Osborne. Although the results were difficult to obtain, nevertheless the principle was proved. The main difficulty encountered in the polarcardiograph was its instability and time-consuming balancing procedure. This difficulty proved that if a three-dimensional computer were to be developed stabilizing circuits would be required.

II THE PRINCIPLE OF THE SPHERICAL POLARCARDIOGRAPH COMPUTER

To develop a general mathematical analysis, it will be assumed that there exists within the body a heart vector or dipole with instantaneous magnitude $\underline{r}$, frontal angle $\underline{\emptyset}$, and polar angle $\underline{\theta}$. Further, it will be assumed that the Cartesian components $\underline{x}$, $\underline{y}$, and $\underline{z}$ of this heart vector can be obtained by suitably combining the potential differences from three pairs of electrodes on the surface of the body. Let $\underline{H}$ and the frontal angle $\underline{\emptyset}$ represent the instantaneous projection of $\underline{r}$ on the frontal plane so that the Cartesian coordinates in that plane are given by

$$x = H \cos \emptyset,$$

and

$$y = H \sin \emptyset.$$

The solutions for $\underline{H}$ and $\underline{\emptyset}$ are

$$H = \sqrt{x^2 + y^2},$$

and

$$\emptyset = \tan^{-1} \frac{y}{x}.$$

By combining the solution for $\underline{H}$ with the third Cartesian coordinate $\underline{z}$, the final magnitude $\underline{r}$ and the polar angle $\underline{\theta}$ are found to be

$$r = \sqrt{H^2 + z^2} = \sqrt{x^2 + y^2 + z^2},$$

and

$$\theta = \cot^{-1} \frac{z}{r}.$$

The spherical polarcardiograph computer is a device intended to transform input voltages proportional to $\underline{x}$, $\underline{y}$, and $\underline{z}$ to output voltages proportional to $\underline{r}$, $\underline{\theta}$, and $\underline{\emptyset}$.

In carrying out this transformation, $\underline{H}$ and $\underline{\emptyset}$, the components in the frontal plane, are computed first, and then $\underline{H}$

is combined with z to obtain r and θ according to the above equations. A block diagram of an elementary circuit intended to give these results is shown in Figure 5. At this time it is impossible to use electromechanical resolvers because of the frequency requirements.

The principle used is basically that of Park's polar-cardiograph computer. That is, a pair of quadrature sinusoidal signals of equal amplitude are first generated; these will be represented by

$$e_1 = E \sin (\omega t + \Gamma),$$

and

$$e_2 = E \sin (\omega t + \Gamma + 90^\circ) = E \cos (\omega t + \Gamma),$$

where ω represents the four-kilocycle-per-second carrier frequency. Multiplication of e_1 and e_2 by voltages proportional to x and y respectively, and addition of the products, gives a resulting sum S so that

$$\begin{aligned} S &= xe_1 + ye_2, \\ &= E \sin (\omega t + \Gamma) \cdot H \cos \varnothing + E \cos (\omega t + \Gamma) \cdot H \sin \varnothing, \\ &= HE \sin (\omega t + \Gamma + \varnothing). \end{aligned}$$

The multiplication of the Cartesian coordinates and the carrier-signal component is accomplished by two pentagrid tubes of the normal frequency-conversion type. The signals from the two tubes must be subtracted for proper multiplication, and the subtractor outputs from the x-axis and y-axis channels must be added to obtain S. Simple circuits exist for carrying out the addition and subtraction steps.

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[9] Park, op. cit., p. 11.

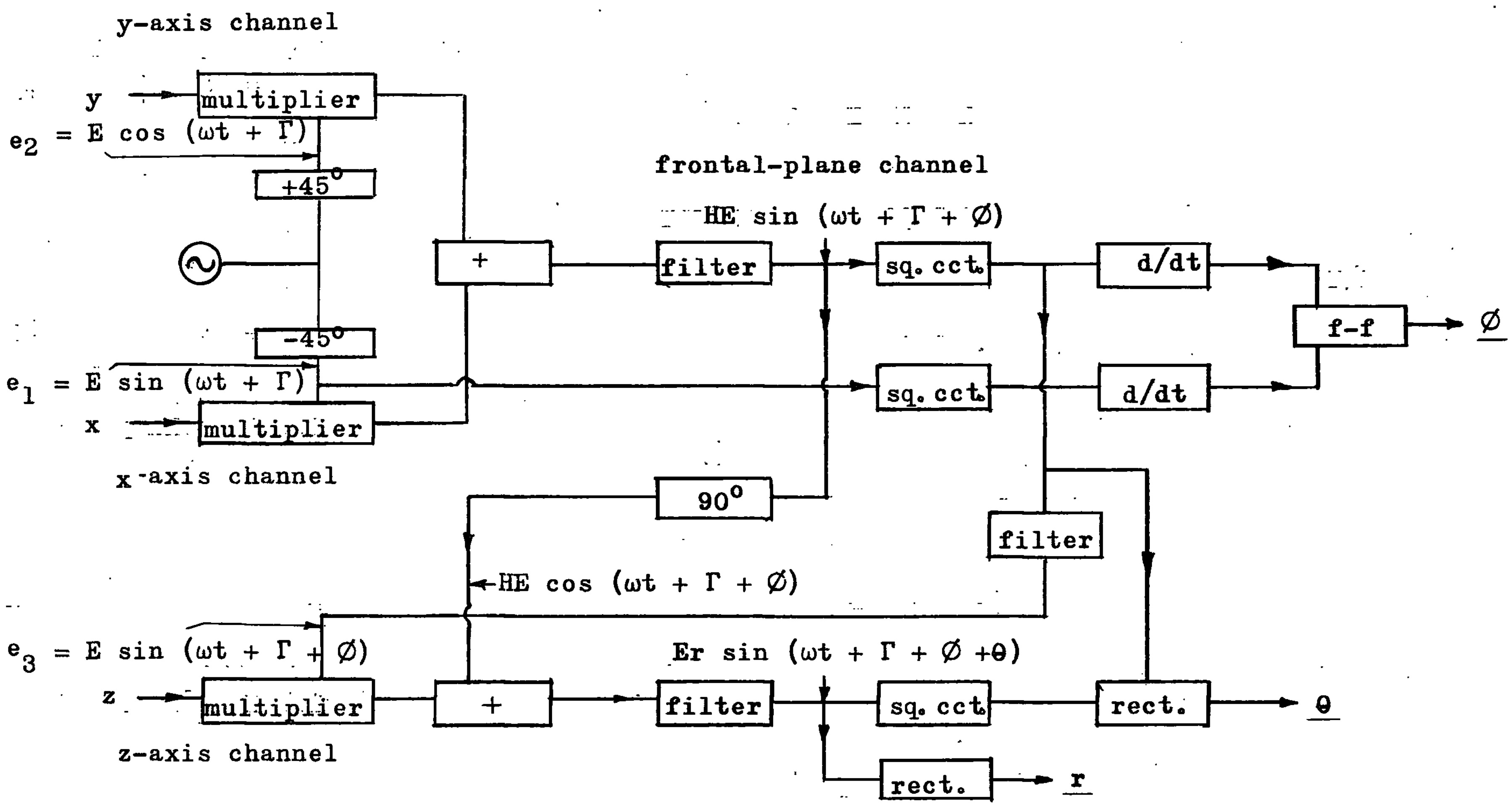


Figure 5. Elementary Block Diagram of the Spherical Polarcardiograph.

At this point the waveform representing $\underline{S}$ contains many unwanted components that were introduced by the multiplier sections. Hence the modulated signal must be filtered before a true representation of $\underline{S}$ is obtained. That $\underline{S}$ is the desired solution in the frontal plane can be seen by noting that the amplitude of $\underline{S}$ is proportional to the magnitude $\underline{H}$ and that the phase of $\underline{S}$ differs from that of $\underline{e_1}$ by the frontal angle $\underline{\emptyset}$.

To obtain the frontal angle $\underline{\emptyset}$, the waveform representing $\underline{S}$ is fed through a cathode-coupled clipper circuit to produce square waves. The square waves are then differentiated and clipped, leaving a negative pulse marking the trailing edge. At the same time, a reference signal proportional to $\sin(\omega t + \Gamma)$ is similarly operated upon. The time delay between the two negative pulses represents the phase difference $\emptyset$ of the two waveforms. The pulses fed to a flip-flop or bistable multivibrator circuit so as to turn a tube alternately off and on. The resulting conduction time of the tube will be proportional to the frontal angle $\underline{\emptyset}$.

The reference signal $\underline{e_3}$ to be fed to the z-channel multiplier is obtained by filtering the clipped wave derived from $\underline{S}$, the frontal-plane signal.

If $\underline{S}$ is of sufficient magnitude to produce a square wave in the clipping circuit, then $\underline{e_3}$ will be of constant magnitude and can be made to have the magnitude E. That is,

$$e_3 = E \sin(\omega t + \Gamma + \emptyset).$$

The waveform representing $\underline{S}$ from the frontal-plane channel is shifted ninety degrees and added to the product of $\underline{e_3}$ and

$\underline{z}$ to form

$$e_3 z + HE \sin(\omega t + \Gamma + 90^\circ)$$

$$= zE \sin(\omega t + \Gamma + \emptyset) + HE \sin(\omega t + \Gamma + \emptyset + 90^\circ),$$

$$= zE \sin(wt + \Gamma + \emptyset) + HE \cos(\omega t + \Gamma + \emptyset),$$

$$= E\sqrt{z^2 + H^2}\left[\frac{z}{\sqrt{z^2 + H^2}} \sin(\omega t + \Gamma + \emptyset) + \frac{H}{\sqrt{z^2 + H^2}} \cos(\omega t + \Gamma + \emptyset)\right],$$

$$= Er \cos\theta \sin(\omega t + \Gamma + \emptyset) + \sin\theta \cos(\omega t + \Gamma + \emptyset),$$

$$= Er \sin(\omega t + \Gamma + \emptyset + \theta),$$

where $\theta = \cot^{-1}\frac{z}{r}$; $r = \sqrt{z^2 + H^2}$. This is the desired result since the amplitude is proportional to the heart-vector magnitude, and the phase of the sinusoid differs from that of $\underline{e_3}$ by the polar angle $\underline{\theta}$. As was the case in the frontal-plane channel, a filter is required to obtain a waveform which is a true representation of the sinusoid $Er \sin(\omega t + \Gamma + \emptyset + \theta)$. To obtain a voltage proportional to the polar angle $\underline{\theta}$, the sinusoid is first fed through a cathode-coupled clipper circuit to obtain a square wave which is added to the square wave from the frontal-plane phase-comparison circuit. The output of the adder circuit is clipped to reject the negative portion of the waveform. The width of the remaining positive pulse is a linear function of $\underline{\theta}$, the phase difference of the two sinusoids being compared. An electromechanical recorder which is sensitive to the quasi-dc components gives an indication of $\underline{\theta}$. The polarcardiograph computer developed by Park operated on the same principle as that described above except that, since the z-axis was not considered, the only components needed were those necessary to produce the frontal-plane outputs $\underline{H}$ and $\underline{\emptyset}$.

Experience with the polarcardiograph showed that instability due to various causes was the most severe limitation on the effectiveness of the equipment. Principal among these sources of instability are amplifier and multiplier drift and fluctuating voltages of polarization on the patient electrodes. Therefore, in designing the present computer, in addition to adding the z-axis input to get a three-dimensional representation, special circuitry was developed to remove or correct such causes of instability. The principal means by which this has been done is a gated feedback circuit which automatically rebalances the system once during each heart beat. It is hoped that, with these additions, the computer will be sufficiently accurate to be of substantial aid in diagnosing heart disorders.

III AUTOMATIC BALANCING WITH GATED FEEDBACK

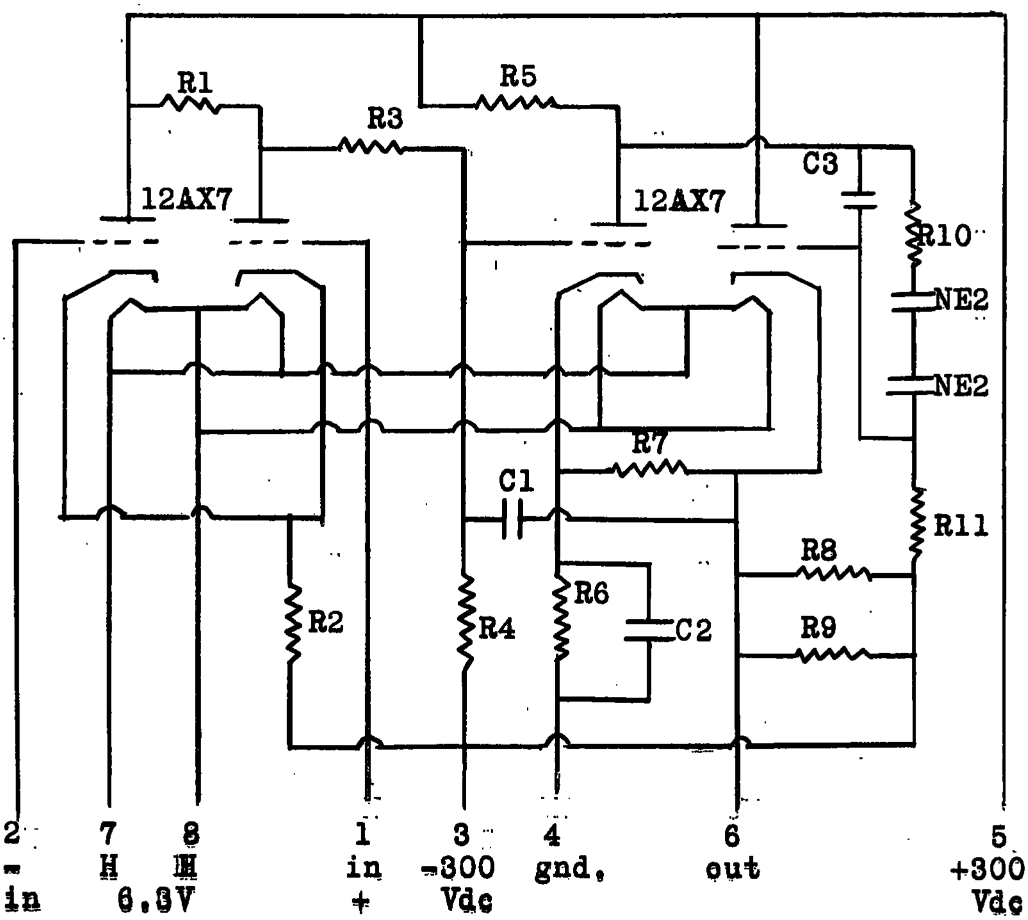
With maximum input signals of the order of one millivolt, the computer must be stable to be of any practical use. A measure of the stability of the computer is the extent to which manual balancing must be carried out each time the instrument is to be used. The procedure requires such adjustments as balancing the two pentagrid tubes of each multiplier section to give zero output for zero input, and equalizing the various channel gains. Although the computer is balanced prior to its use, the instrument will not maintain this condition to the required extent. Hence, if valid results are to be obtained, the procedure must be repeated every few minutes. The time consumed with such operation emphasizes the need for automatic balancing.

The obvious time for automatic balancing is during the isoelectric period marking the rest time between heart beats. The problem now becomes threefold, namely: where should the signal for feedback be obtained; how should the signal be directed to the proper point for correction; how should the signal be reintroduced to the circuit for proper correction?

To begin the investigation, a study was made to determine the necessary loop gain to provide at least a twenty-to-one reduction in the error during feedback. The computer design is such that the input amplifiers to each channel are the equivalent of those in the normal electrocardiograph. It was also found necessary to include a cathode-ray tube so as to present vectorcardiograms of any two desired channels and to aid in the balancing procedure. To obtain sufficient deflection on the face of the cathode-ray tube, a double-ended deflection amplifier with a gain of fifteen is required following the input amplifier. To adjust the signal amplitude to that required for the pentagrid tubes which multiply, for example, $\underline{e_1}$ and $\underline{x}$, a thirty-to-one reduction is necessary. Since the effective gain through the multiplier section is approximately unity, the forward gain from the deflection-amplifier input to the multiplier output is seen to be 15/30 or 0.5. Therefore, if feedback is taken from the multiplier output, the backward path will require a gain of 20/0.5 or forty. To eliminate the need for additional feedback amplifiers, it was attempted to obtain the feedback signal from a point further on in the circuitry.

By doing this, another problem was introduced. Although the gain was sufficient, the modulated component signal had been combined and it was therefore necessary to supply feedback to four inputs from one signal. Various types of phase-sensitive detectors were designed and tested but the difficulty of slight phase shifts in the preceding circuitry caused the idea to be abandoned.

If the feedback signal is to be obtained before the component signals are combined, separate subtractors will be required following the two tubes of each multiplier section. Since the external circuit of the amplifier shown in Figure 6 can be designed to add and subtract in one operation, separate subtractors in the backward path were considered. The required gain of a subtractor amplifier in



R1	.22	M
R2	.22	M
R3	1.0	M
R4	2.2	M
R5	.47	M
R6	10	K
R7	.22	M
R8	.22	M
R9	.27	M
R10	.68	M
R11	4.7	M
C1	7.5	μμf.
C2	500	"
C3	7.5	"

General Specifications

GAIN:
15,000 dc, open loop

POWER REQUIREMENTS:
4.5 ma. at +300Vdc
4.5 ma. at -300Vdc
0.6 amperes at 6.3V

TUBE COMPLEMENT:
2 12AX7

INPUT IMPEDANCE:
Above 100 megohms

OUTPUT IMPEDANCE:
Less than 1 K open-loop
below 1 ohm fully fed back

DRIFT RATE:
5 mv. per day, referred to the input

VOLTAGE RANGE:
-50 Vdc to +50 Vdc, at output and both inputs

INPUT CURRENTS:
less than 0.1 microamp for either input

OUTPUT CURRENT: -1 ma. to +1 ma. over full voltage range.

Figure 6. Philbrick Model K2-W Operational Amplifier

the backward path would be forty plus that needed to overcome the losses in the detection circuit. If this high gain were used in the operational amplifier that was to be employed as a subtractor, the amplifier would be driven beyond its linear range. If the output of the subtractor were not linearly dependent upon its input, the signal fed to the deflection-amplifier inputs would not be proportional to the existing error. Since, in this case, the feedback would not balance the computer properly, the idea of a high-gain subtractor amplifier in the feedback path was discarded.

A subtractor for each multiplier circuit was finally put into the forward path and the operational amplifier that was formerly to be used as both an adder and subtractor circuit is now used only as an adder. So that the subtractor circuit will not be overdriven, its gain is lowered to six. The subtractor using this low gain is inherently so stable that a manual balance is not required. The feedback signal which comes from the subtractor output is given a gain of twenty by a separate amplifier in the feedback path.

Since the feedback signal, for proper correction, must be applied to one of two inputs, a phase-sensitive detector is required. Figure 7 shows the complete automatic balancing circuit. An error in the multiplier section produces a modulated signal output from the subtractor. This signal operates the detector circuit so as to raise the potential of one grid or the other of the deflection amplifier. The reference signal appearing at the cathodes of the vacuum diodes is derived through a step-up transformer from the

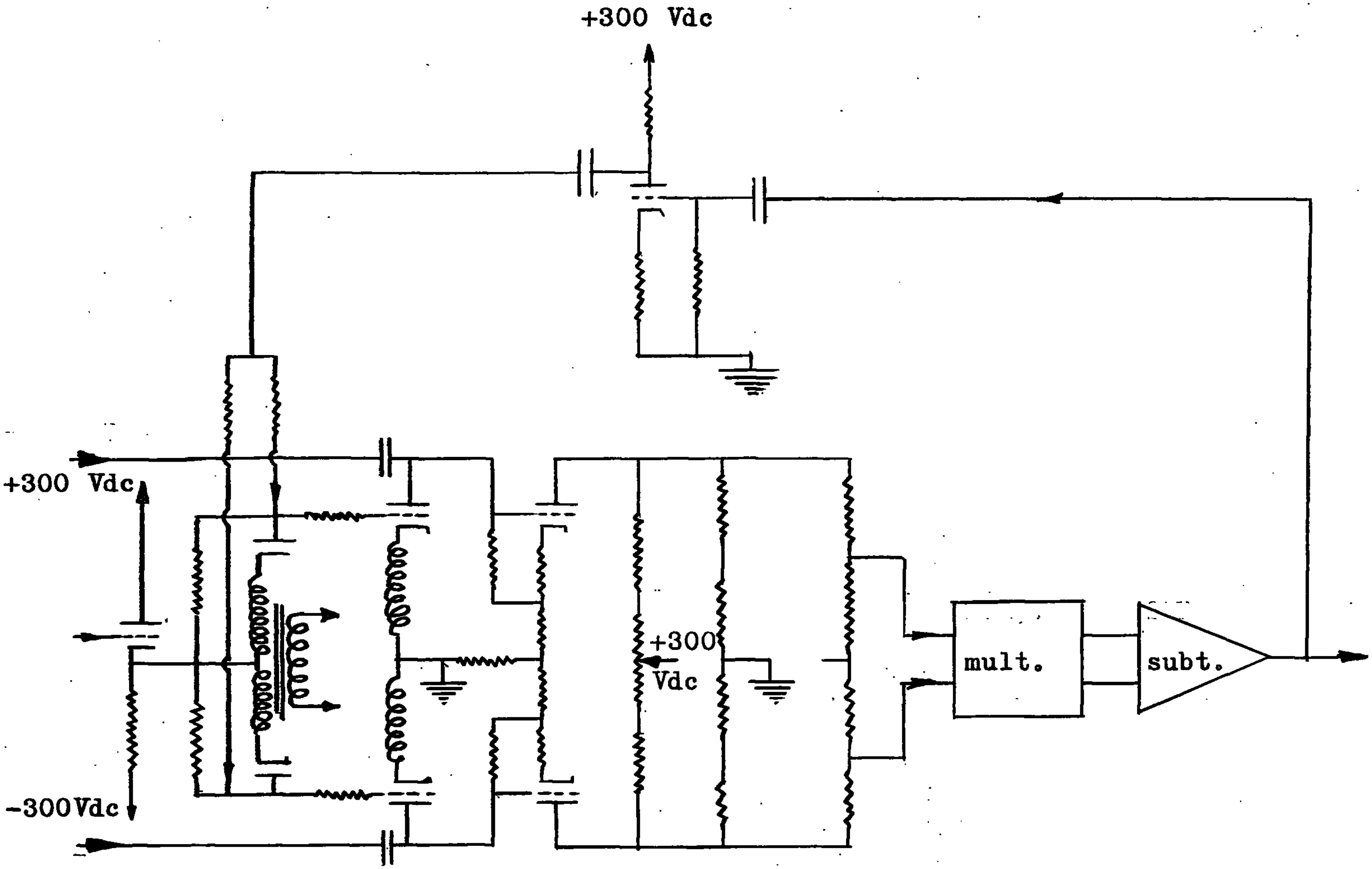


Figure 7. Gated Feedback Circuit.

local oscillator which generates e_1 and e_2 of the multiplier section. The centre-tap of the transformer secondary winding is connected to the plate of the diode through a one-megohm resistor. The error signal is fed from the feedback amplifier, through an isolating resistor, to the diode plate. When the reference signal is of such a phase as to hold the cathode positive with respect to the centre-tap, and hence with respect to the plate, the tube will not conduct. On the other hand, when the cathode is forced negative with respect to the centre-tap and diode plate, the tube conducts and thus lowers the potential of the plate. For complete detection, the effect of the reference signal on the plate circuit must be greater than that of the error signal.

There are two possible conditions which may arise in the plate circuit of the diode. Either the error signal is in phase with the reference signal or it is 180 degrees out of phase. If the latter exists, the diode will conduct whenever the error signal is positive. Hence the plate potential will never rise above a previously set level. However, if the former case prevails, the diode will conduct during the negative half-cycle of the error signal but will be turned off during the positive half-cycle. Here the diode plate potential, which will be used in the feedback, follows the positive half-cycle of the error signal.

The problem of reintroducing the signal to the system was one of charging the input capacitors of the deflection amplifier. Tests showed that the circuit was highly receptive to the carrier frequency of four kilocycles per second which, during feedback, resulted in a line on the cathode-ray tube. To charge an input capacitor and to attenuate the four-kilocycle-per-second signal, a high-μ triode was connected from the deflection-amplifier grid to ground with a choke in the cathode circuit. The choke causes sufficient degeneration to reduce the ratio of carrier to rectified dc component by a factor of better than twenty to one. The error signal is introduced at the gating-tube grid through an isolating resistor from the diode detector.

To control the period in which feedback occurs, a square wave is introduced at the centre-tap of the reference transformer of the detection circuit. This gating pulse originates in a cathode-coupled delay multivibrator with controllable delay, the first stage of which is triggered from a positive-going portion of the input waveform. The trigger pulse, with the aid of a rotary switch, may be taken from any one of the six deflection-amplifier plates. The positive square wave of fixed width, at some variable time after the trigger pulse, is fed to the centre-tap of the reference transformer to control the bias of the gating tube. To control the voltage levels, the square wave is fed through a cathode follower whose cathode is returned, through a resistor, to a potential below that of ground. A diode catcher on the potential divider feeding the cathode follower governs the level to which the cathode may rise. Between pulses the cathode follower output, because of the negative voltage return, becomes highly negative. The gating-tube grid levels are set so that when the pulse is applied the grid potential of the gating tube rises to just below cutoff, permitting feedback if there is signal from the subtractor output. The gating pulse amplitude and width is fixed, but the delay time is adjustable by the operator to ensure that feedback takes place exactly during the rest period of the heart. The proper delay time may be obtained by adjusting the delay control until the clamp pulse occurs during the rest period of the heart, as indicated either by spot-brightening on the cathode-ray tube or by temporarily superimposing the clamp pulse on one of the normal electrocardiograph outputs.

During one stage of the work on feedback, complete balance caused both gating tubes to conduct an equal amount. The in-phase components, if given full deflection-amplifier gain, would have been sufficient to shift the multiplier tubes beyond their linear range. To compensate for this, it was found necessary to redesign the deflection amplifiers to give a high in-phase rejection ratio. The common cathode resistor in the deflection-amplifier circuit of Figure 7 accomplishes this result. With the present design, the in-phase components do not arise to the same extent in the feedback signal; however, the deflection amplifier cathode-circuit design is retained to compensate for the in-phase components which do arise.

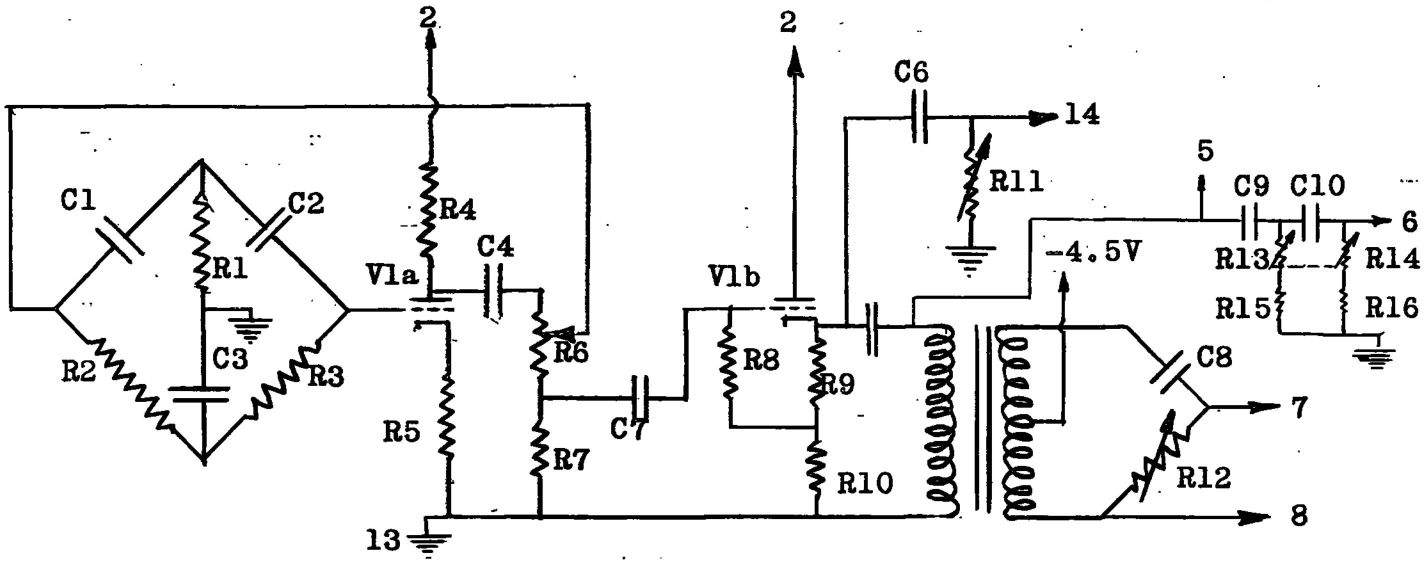
As will be discussed later, the operation of the z-axis channel depends upon there being a component of the heart vector lying in the frontal plane. That is, if the frontal-plane circuitry is completely balanced, feedback on the z-axis channel will be of no use. However, it is hoped that, since the balancing action takes place very rapidly, the z-axis channel can be balanced in a similar manner. If, after final construction is complete, it is found that the z-axis channel does not balance with normal feedback, the following recommendations are made. The feedback time may be divided so that the z-axis channel is balanced first. This may be done by applying a short time delay to the square wave initiating the feedback to the frontal-plane circuitry. If the latter also proves insufficient, a separate signal may be applied temporarily to the multiplier tubes and the feedback reference transformer of the z-axis channel during feedback. The two lines feeding signal from the frontal-plane circuitry to the z-axis channel must be broken during this period. This method will be adequate, but the complexity of the necessary circuitry suggests that it be used only if absolutely required.

IV FREQUENCY CONSIDERATIONS

The spherical polarcardiograph, like the present-day electrocardiographs, is sensitive to input frequencies up to 100 cycles per second. To allow for this frequency range and ±100 cycles per second oscillator drift it is seen that a bandwidth of 400 cycles per second is required. A carrier frequency of four kilocycles per second, as was used by Park, was chosen so that the band-pass amplifier and detection circuits could be easily designed and so that the stray capacitance effect would not be too great.

The oscillator which produces the carrier-frequency signal is a resistive-capacitive twin-T type (Figure 8), which gives good frequency stability and low harmonic content. The cathode follower on the oscillator output feeds signal directly to the reference-phase circuit of the frontal-plane channel and through isolating transformers to the feedback detection circuits and the multiplier sections. The reference-phase circuit contains an R-C phase-shifting network to compensate for any phase shift in the filtering circuits.

To adapt the spherical polarcardiograph for recording vectorcardiograms, it is desirable to apply time markers to the cathode-ray tube presentation by intensity modulation. Although the present design does not incorporate the time-marker circuit, provision has been made to locate the time-marker generator near the oscillator. This has been done so that the oscillator may be used as a source of synchronizing signal.



R1	22 kilohmn
R2, R3, R7, R8	0.47 megohm
R4	0.2 "
R5, R9	2.2 kilohm
R6	0.5 megohm
R10	0.1 "
R11	0-250 kilohm
R12	0-10 "
R13, R14	0-30 "
R15, R16	27 "

C1, C2	100 μμf.
C3	200 "
C4, C8	0.005 μf.
C5, C7	0.01 "
C6	0.001 "
V1	12AX7
C9, C10	0.001 "

Figure 8. The Oscillator and its Output Circuits

The multiplier tubes, as was predicted by Park,[10] introduce many unwanted frequency components which must be rejected. These components are rejected by passing the signal through the high-pass filter and double-tuned amplifier of Figure 9. The filter is necessary to prevent the low-frequency components from overloading the amplifier tube. The T-network of standard slug-tuned chokes is the equivalent circuit of a transformer. By adjusting the shunt resistors and the chokes, a pass-band amplification constant within three per cent is obtained between 3800 and 4200 cycles per second.

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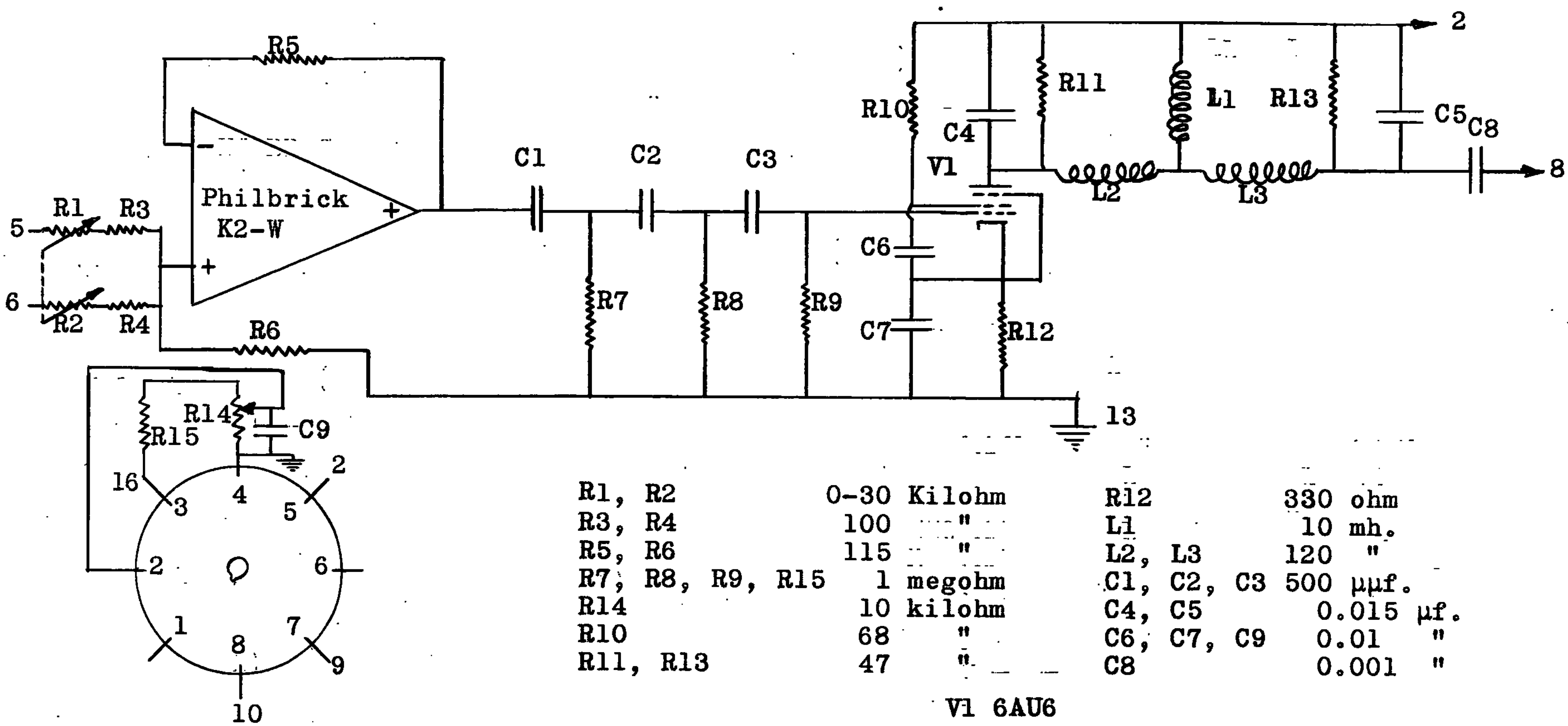
[10] *Loc. cit.*



R1, R2	0-30 Kilohm	R12	330 ohm
R3, R4	100 "	L1	10 mh.
R5, R6	115 "	L2, L3	120 "
R7, R8, R9, R15	1 megohm	C1, C2, C3	500 μμf.
R14	10 kilohm	C4, C5	0.015 μf.
R10	68 "	C6, C7, C9	0.01 "
R11, R13	47 "	C8	0.001 "

V1 6AU6

Figure 9. Adder Circuit, High-Pass Filter, and Band-Pass Amplifier

V COMPUTATION OF THE THIRD COORDINATE

Instruments such as the polarcardiograph compute the angle and magnitude of the plane projection of the heart vector. The principle of computation of the three spherical polar coordinates $\underline{r}$, $\underline{\theta}$ and $\underline{\emptyset}$ of the heart vector has been described in chapter II. The signals used in this extension are the frontal-plane output $HE \sin(\omega t + \Gamma + \emptyset)$ and the third input signal $\underline{z}$. The computation could proceed by rectification of the frontal-plane signal and by combination with $\underline{z}$ in precisely the same way as was done for the $\underline{x}$ and $\underline{y}$ inputs. However, the extra distortion introduced and the additional circuitry required did not seem justifiable since there is a simpler method.

The carrier-frequency signal for the z-axis multiplier section is obtained by filtering the output of the circuit which produces square waves from the frontal-plane signal. The signal is adjusted so that it is proportional to $E \sin(\omega t + \Gamma + \emptyset)$. The frontal-plane signal, $HE \sin(\omega t + \Gamma + \emptyset)$, is shifted ninety degrees and added to the z-axis multiplier output to give $zE \sin(\omega t + \Gamma + \emptyset) + HE \cos(\omega t + \Gamma + \emptyset)$ $= Er \sin(\omega t + \Gamma + \emptyset + \theta)$. The adder-circuit output must be fed through a filter to reject the unwanted components introduced in the multipliers. Half-wave rectification produces a quasi-dc component which may be interpreted (by using an electromechanical recorder), as an indication of the magnitude $\underline{r}$ of the heart vector.

The z-axis channel signal differs in phase from the frontal-plane signal by the polar angle $\underline{\theta}$. Since the polar angle $\underline{\theta}$ varies between zero and 180 degrees, a signal proportional to $\underline{\theta}$ may be obtained by adding the square wave resulting from HE $\sin(\omega t + \Gamma + \emptyset)$ to the square wave resulting from rE $\sin(\omega t + \Gamma + \emptyset + \theta)$. The output of the adder circuit, if clipped so as to retain only the positive portion of the signal, is a square wave of width proportional to the polar angle $\underline{\theta}$. An indication of $\underline{\theta}$ is obtained by feeding the square wave to an electromechanical recorder which is sensitive only to the quasi-dc component of signal.

One apparent disadvantage of this system is that the z-axis channel will be non-operational if the frontal-plane signal is insufficient to produce a square wave at the phase meter. The result is a cylindrical volume in the space in which the heart vector lies within which the output from the z-axis channel will have no meaning. Thinking of the whole volume in which the heart vector may lie as a sphere, there is an "apple-core" surrounding the polar axis; no information can be obtained for heart vector positions within the apple-core. Tests have shown that the ratio of the radius of the cylinder to the maximum magnitude of the heart vector is approximately one to thirty with the equipment that has been built. Thus, for heart vectors of maximum amplitude, accurate information will be obtained for polar angles greater than two degrees. Since it is not thought likely that the heart-vector direction will coincide with that of the z-axis or sagittal axis very frequently, the loss of information through this apple-core effect is not considered serious. If such cases do arise, a change of coordinate axes will supply the missing information.

The indicated direction of the heart vector has little meaning when the magnitude is small. This fact has prompted the development of a threshold control to suppress the frontal-angle output when the frontal-plane magnitude is too low. A threshold control was added to Park's polarcardiograph shortly after it was completed. In the present design of the polar-cardiograph, a triode replaces the vacuum diode which formerly rectified the differentiated square-wave pulses. The square wave is taken from the second stage of the squaring circuit, integrated and applied through an amplifier to the grid of the rectifying triode. The pulses which mark the trailing edge of the square wave are normally passed to the flip-flop because the peak of the triangular wave (integrated square wave) is applied to the grid of the triode at the same time, thus allowing the triode to conduct. However, if the magnitude becomes too small to develop a triangular wave of sufficient height to cause the triode to conduct, the angle reading is suppressed. The threshold control as described here is only partially successful, mainly because it is too critical to adjust.

The threshold control designed for the spherical polar-cardiograph uses the same basic principle with the exception that a sine wave is used in place of the triangular wave. It is thought that the circuit will not be as critically sensitive to a slight difference between the times of arrival of the peak of the sine wave and of the pulse at the rectifying triode as it is in the former control. A sinusoid, obtained from the filter feeding the z-axis multiplier carrier signal, is shifted ninety degrees, given half-wave rectification, and applied to the rectifying-triode grid. The ninety-degree phase shift ensures that the peak of the sine wave will occur simultaneously with the trailing edge of the square wave in the phase-meter circuit. The spherical polarcardiograph completely specifies the heart vector with output voltages proportional to $\underline{r}$, $\underline{\theta}$, and ϕ and at the same time suppresses these results when they have no meaning.

VI VOLTAGE AND CURRENT REQUIREMENTS

The low input signal level and the limited linear range of many of the tubes of the computer imposes a need for voltage regulation. The unbalance resulting in the plane polarcardiograph from slight voltage variations has emphasized this requirement. The circuits that receive regulated voltage are those requiring great stability and drawing a constant current. the input amplifiers use fifty milliamperes at +300 volts dc and forty milliamperes at -300 volts dc, both regulated. The heaters of these units and the multiplier tubes are fed in series by a regulated dc supply. The units containing the deflection amplifier, the multiplier section, the subtractor and the feedback circuit draw ninety milliamperes at +300 volts regulated and 14 milliamperes at -300 volts regulated. The adder, filter, and double-tuned amplifier circuits use a total of forty milliamperes at +300 volts regulated and nine milliamperes at -300 volts regulated. In addition, a regulated +300-volt supply is needed for the oscillator (two milliamperes), and for the unit providing the magnitude output, polar angle output and the filter for the z-axis carrier signal (fourteen milliamperes); the latter unit also uses five milliamperes at -300 volts regulated. The remaining units requiring only an unregulated +300-volt supply are the cathode-coupled clipper or squaring circuit (25 milliamperes), the frontal-channel phase meter (35 milliamperes), and the delay clamp-pulse generator (two milliamperes).

The total current requirements are 196 milliamperes at +300 volts regulated, 62 milliamperes at +300 volts unregulated and 78 milliamperes at -300 volts regulated. The ac heater requirements total 21 amperes at 6.3 volts. To dissipate the heat, the chassis has been designed with side flaps which are to be left open during operation. The regulation of the supply voltage for the more critical circuits will aid the spherical polarcardiograph in maintaining its balanced condition for longer periods than did the plane polarcardiograph. Coupling the regulation with the gated feedback, results in a stable computer.

VII THE OPERATING ADJUSTMENTS

Before taking any tracings with the spherical polar-cardiograph, the computer zero-level must be established, each multiplier section balanced for zero signal output for zero input and the channel gains equalized. The balancing procedure, if the computer is first allowed to become warm, need only be performed once. Any subsequent unbalances will be compensated for by the gated feedback.

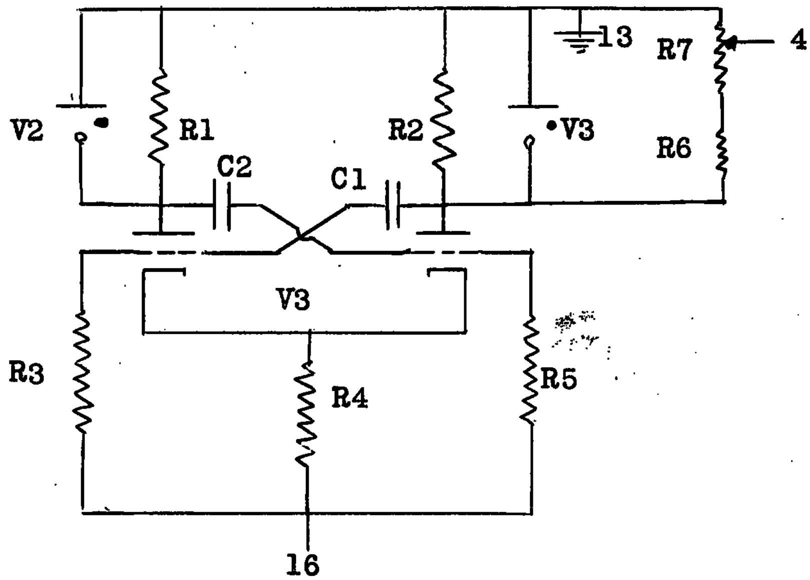
When performing the balancing procedure on, for example, the x-axis channel, the y-axis and z-axis channels must have feedback applied to avoid interference. The zero-level of the channel is set by adjusting the x-axis centring control for zero deflection and then the corresponding multiplier pair is balanced to show zero magnitude output. A similar method is used on the y-axis channel but additional precautions must be taken when balancing the z-axis channel. A carrier signal must be applied to the multipliers and the frontal-plane signal disconnected before the z-axis channel is adjusted. After initial channel adjustments, channel gains are equalized by applying equal signals to the $\underline{x}$ and $\underline{y}$ inputs and then to the $\underline{y}$ and $\underline{z}$ inputs and adjusting channel gains so as to produce a 45-degree deflection on the cathode-ray tube in each case. Then the gains as far as the deflection-amplifier outputs will be equal so that the cathode-ray tube presents a valid vectorcardiogram. The initial balancing must be followed by the calibration of the instruments used to record angle and magnitude outputs.

In greater detail, the balancing procedure is as follows. First ground all inputs by placing the input rotary switch to position nine. Open switch one so that the final tube of the clamp generator is turned off and all feedback loops are closed. By means-of-the cathode-ray tube rotary switch, observe deflections in the frontal plane. Open the feedback loop of the x-axis channel by placing switch two in position two, thereby lowering the grid of the corresponding gating triode below cut-off potential. Return the spot on the cathode-ray tube to the vertical line with the x-centring control, thus effectively tieing the signal grids of the multiplier tubes together. The variable plate load resistor of the multipliers is then adjusted until the magnitude r reads zero. At this point, the x-axis channel balance is complete so that switch two may be returned to its normal position (position one). The procedure is repeated with the y-axis channel using switch three to break the feedback loop and the y-centring control on the cathode-ray tube to set the zero level of the channel.

Since the z-axis channel operation depends upon a carrier signal from the frontal-plane being fed to the multipliers, the input rotary switch must be moved to position eight so that signal is applied to the y-axis input. Switch one is returned to its normal position to remove all feedback. Open switch five to prevent the frontal-plane signal, HE $\sin(\omega t + \Gamma + \emptyset)$, from entering the z-axis channel. The cathode-ray tube picture must be changed to present the x-z plane. The z-centring control is used to return the spot to the x-axis. The multiplier is adjusted to give zero magnitude output. At this stage, the zero level or origin of the computer has been set and the three multiplier sections have been balanced.

Before adjusting the x-, y-, and z-axis channel gains of the instrument, it is necessary to check that switches one to five are in their normal positions. The input rotary switch is placed in position seven so that one millivolt is applied to one side of each input, the other sides remaining grounded. The input amplifier gain controls are adjusted until x-y and x-z presentations show 45-degree deflections on the face of the cathode-ray tube. The gains to the deflection-amplifier outputs are now equal and the multipliers are balanced so that the instrument is ready to be calibrated.

The angle calibration is accomplished by applying a one-millivolt signal at approximately one cycle per second to various inputs. The calibration signal is taken from the free-running multivibrator shown in Figure 10. By applying the calibrating signal to various combinations of the six available inputs, the frontal and polar angles can be calibrated in 45-degree steps. Position six of the rotary switch applies signal to one input of the x-axis channel which results in a frontal-angle output alternating between zero and 180 degrees. Position five maintains the signal input to the x-axis channel and applies the same



R1, R2	68 kilohms	R6	1 megohm
R3, R5	4.7 megohms	R7	0-100 ohm
R4	15 kilohms		
C1, C2	0.05 μf.		
V1, V2	VR 90		
V3	12AX7		

Figure 10. Free-Running Multivibrator
For Calibration.

signal to one input of the y-axis channel to show a frontal angle of 45 and 225 degrees. If the x-axis channel inputs are grounded with the signal remaining on the y-axis (position four), the frontal angle will register ninety and 270 degrees. Position three of the rotary switch retains the signal on the y-axis channel and simultaneously applies the signal to the opposite input of the x-axis channel as was previously used. The reversal of the x-axis channel inputs changes the instantaneous polarity so that the frontal-angle reading will be shifted ninety degrees from that of position five to give 135 and 315 degrees. The frontal-angle output is now calibrated in 45-degree steps for 360 degrees. Non-linearity in the calibration of the frontal-angle output should be corrected by adjusting the adder control for this channel.

Calibration of the polar-angle recordings can be performed during the frontal-angle calibration. With frontal-plane signal and with no z-axis signal, the polar angle will register as ninety degrees. If a signal is applied to either input of the _x_- or _y_-axis and the same signal is applied to one input of the z-axis, the polar-angle output will alternate between 45 and 135 degrees. The rotary switch allows for this calibration in positions five and four respectively. Any non-linearity arising in the polar-angle calibration should be corrected by adjusting the z-axis adder control.

A summary of the balancing and calibrating procedure is as follows:

1. Place the input rotary switch in position nine, open switch one, apply the x-y plane to the cathode-ray tube, place switch two in position two and zero the x-coordinate of the spot on the tube face; adjust the multipliers to show zero magnitude output and return switch two to position one.

2. With the input rotary switch still in position nine, repeat 1. using the y-axis centring control and switch three.

3. Place the input switch in position eight, change switch one to its normal operating position and open switch five; with the cathode-ray tube presenting the x-z plane, centre the spot on the face of the tube and adjust the multiplier control for zero magnitude output.

4. Check that switches one to five are in the normal operating positions.

5. Place the input switch in position seven and adjust the input amplifier gains of the x and y axes to show 45 degrees on the cathode-ray tube.

6. Adjust the z-axis input amplifier gain control until the cathode-ray tube presentation, which shows the x-z plane, reads 45 degrees.

7. Place the input rotary switch in position six and note the zero- and 180-degree frontal angles.

8. Turn the input switch to position five and note the 45- and 225-frontal angles. Also note the ninety-degree polar angle.

9. With the input switch in position four, note the

90- and 270-degree frontal angles and the 45- and 135-degree polar angles.

10. Place the input switch in position three and note the 135- and 315-degree frontal angles.

11. If non-linearity is apparent in the angle indications it is necessary to correct the corresponding adder controls.

With the spherical polarcardiograph completely balanced and calibrated, the input rotary switch is returned to position one, its normal operating condition. Position two of the rotary switch is provided as a frontal-plane quadrant shift. The shift is necessary because of the ambiguity of the phase meter at the 180-degree mark. If the preceding instructions are followed carefully, the computer can be used to obtain accurate information concerning the electrical activity of the heart.

VIII PHYSICAL LAYOUT

The computer, with the exception of the power supply, is designed so as to be housed in an 11" x 14" x 20" Hammond cabinet. When in use, the instrument will rest on a push-cart with the power supply on a lower shelf. With this arrangement, the computer can easily be moved to the patient's bedside.

For an instrument of this size and complexity, servicing is an important factor. The apparatus has been designed for construction in units so as to simplify the technician's task. If trouble is suspected in any portion of the equipment, the appropriate unit can be withdrawn and tested independently of the remainder of the installation. Connections to the units are to be made through sixteen-terminal plug-in connectors. All of the inter-unit wiring will be housed on a back panel (Figure 11) to which the units connect. Figure 12 shows a complete block diagram of the spherical polarcardiograph. Figures 13 to 19 inclusive show the circuit diagrams and element values of the remaining units. For the suggested unit positions consult Figure 20. It will be noted that several of the units in the finished instrument will be identical.

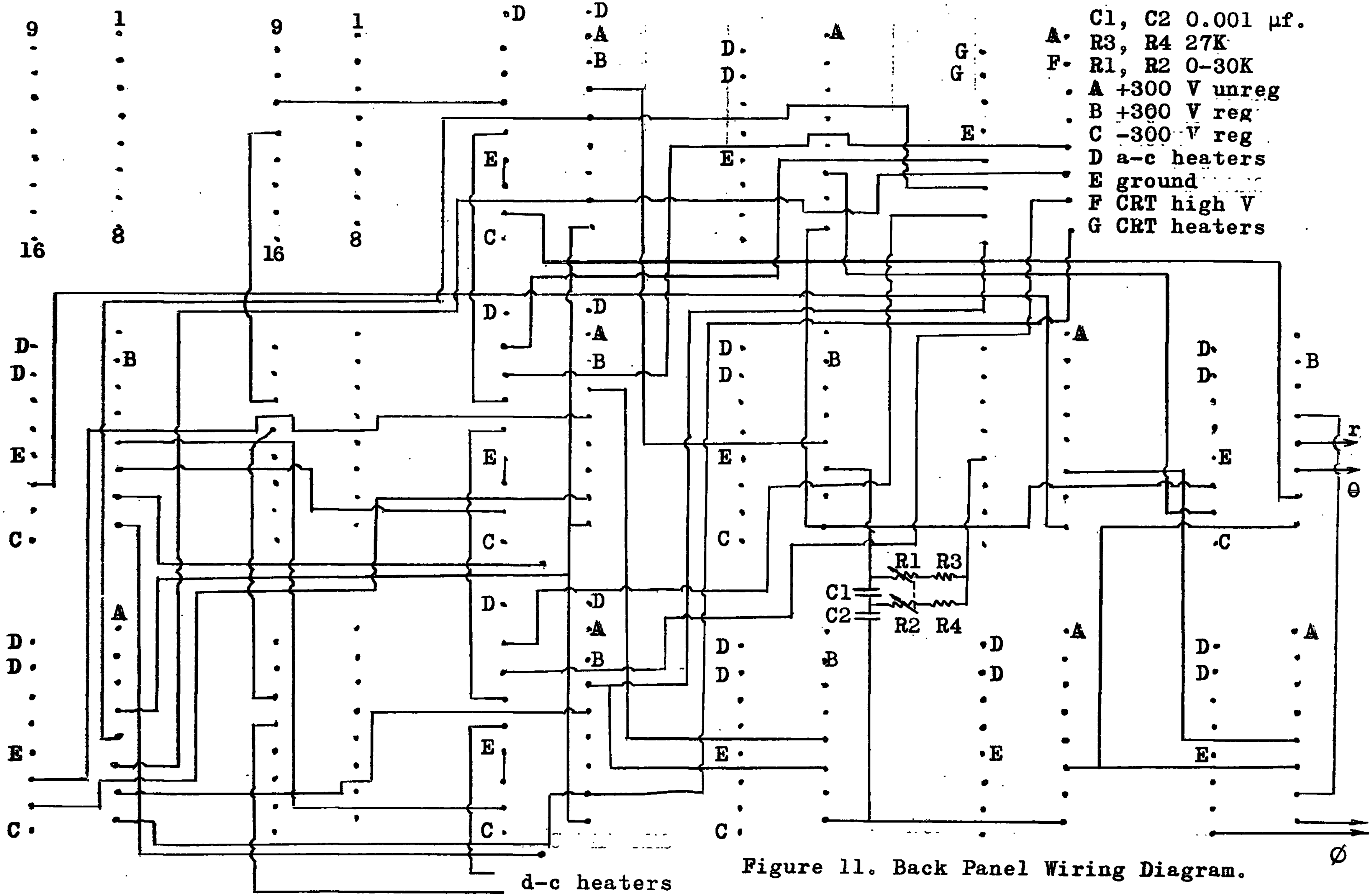


Figure 11. Back Panel Wiring Diagram.

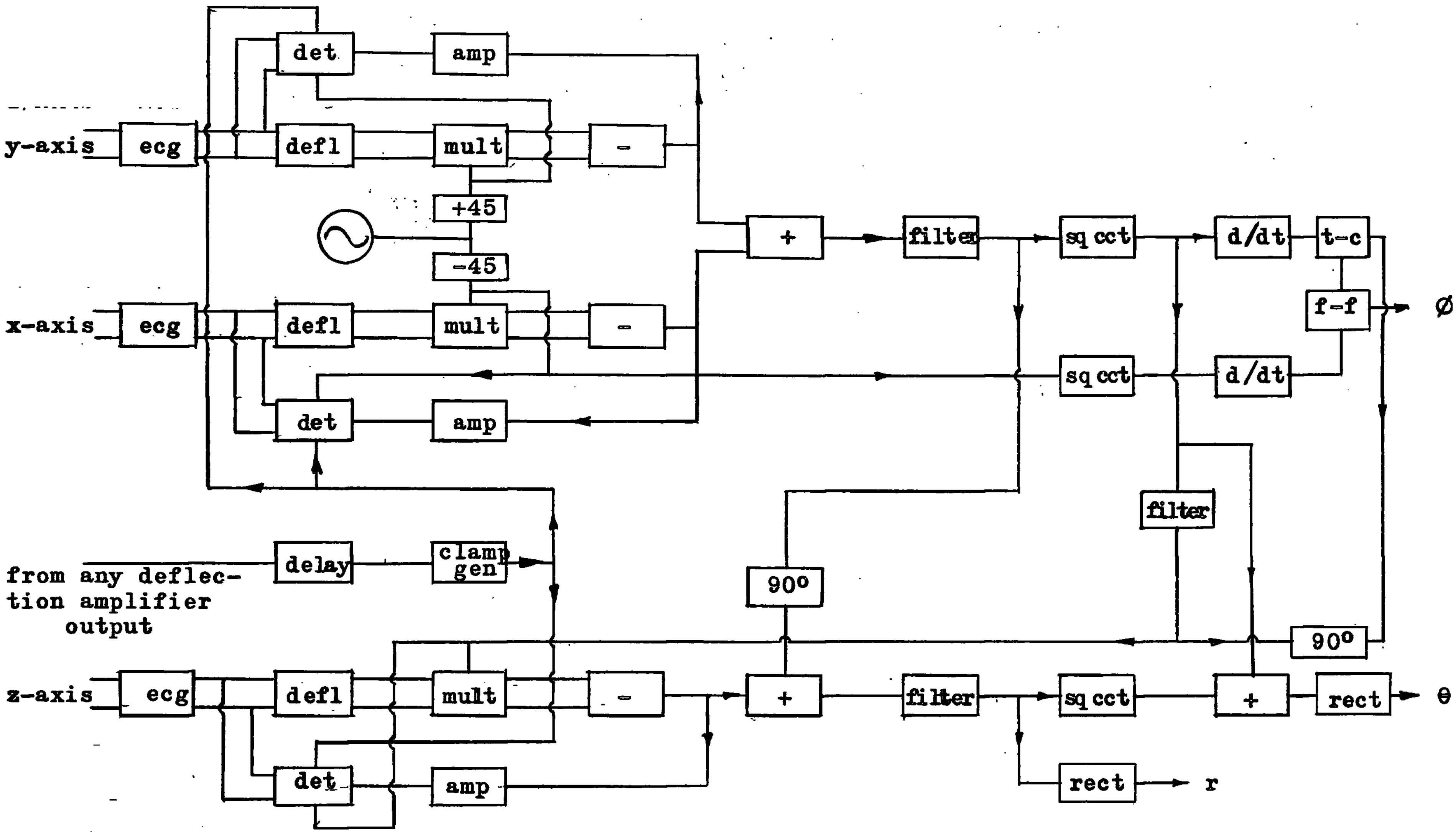


Figure 12. Block Diagram of the Spherical Polarcardiograph.

R1, R2, R34	1 megohm
R3, R4	0.47 "
R5, R7, R16, R17	2.2 "
R8, R10	2.2 kilohm
R9, R25	10 "
R11, R23, R29	0-10 "
R12, R14, R20, R21, R32	0,15 megohm
R13	0-0.25 "
R15, R18	0.5 "

R19, R22	4.7 megohm
R24	0.5 kilohm
R26	20 "
R27, R30	68 "
R28, R31	447 "
R33	3.3 "
C1, C2, C5, C6	0.1 μf.
C3, C4	0.01 "
L1, L2	130 mh.

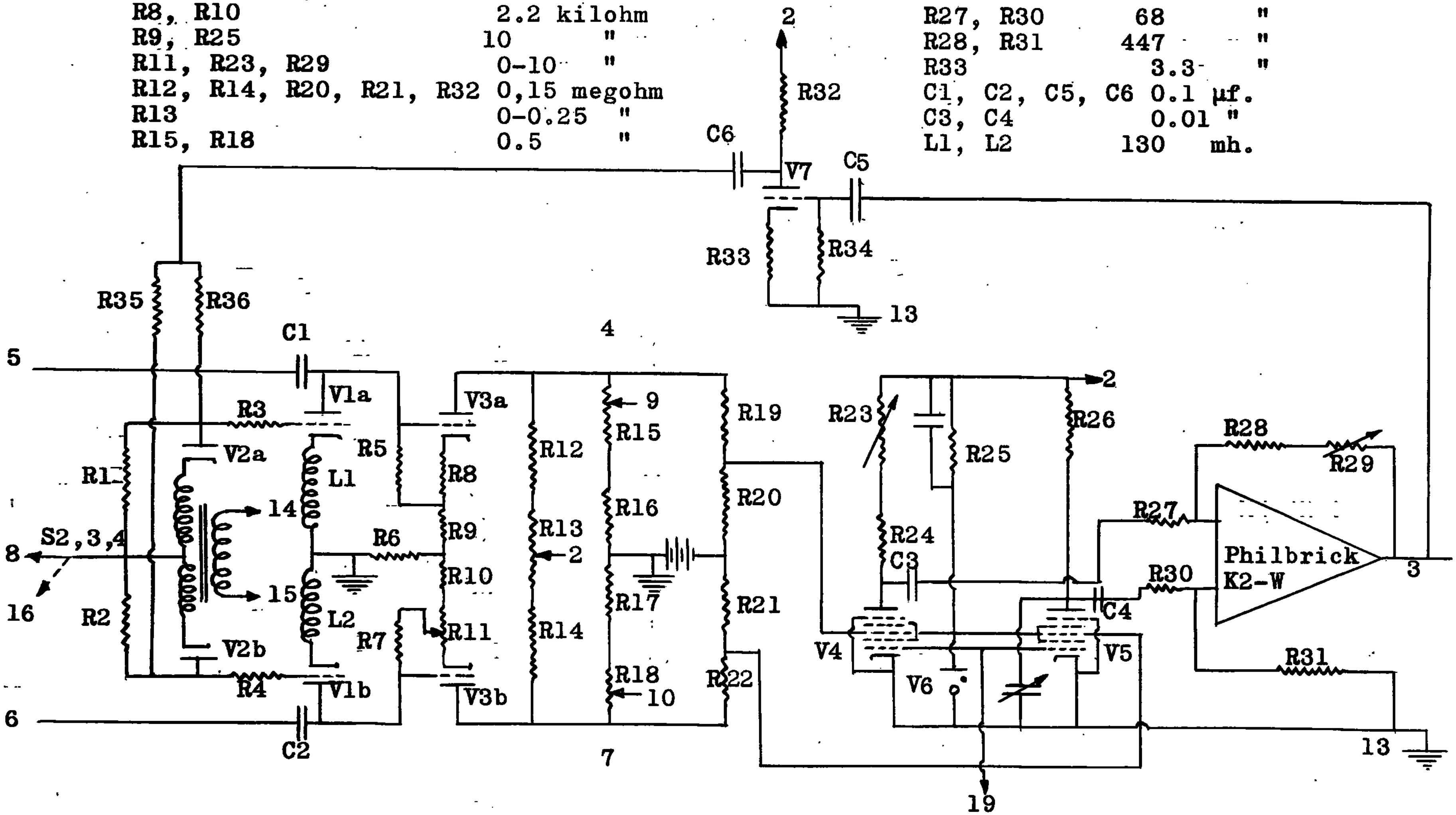
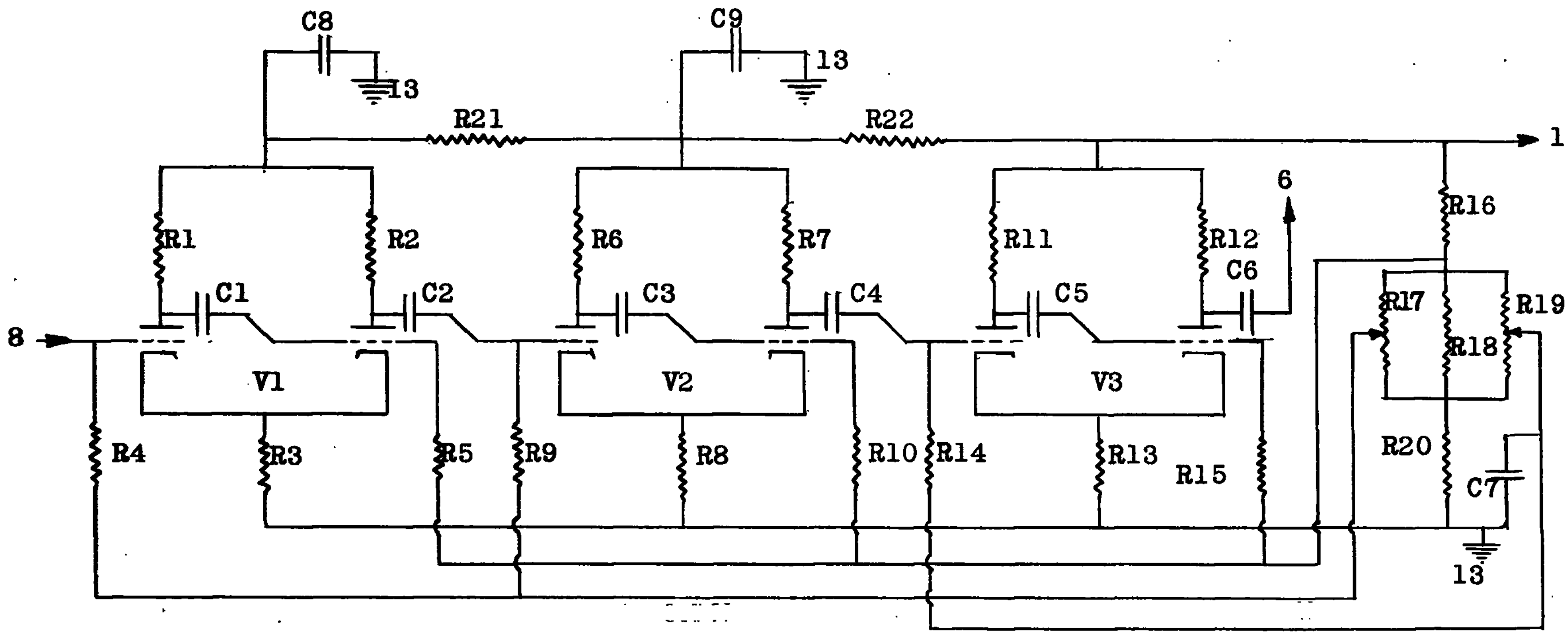


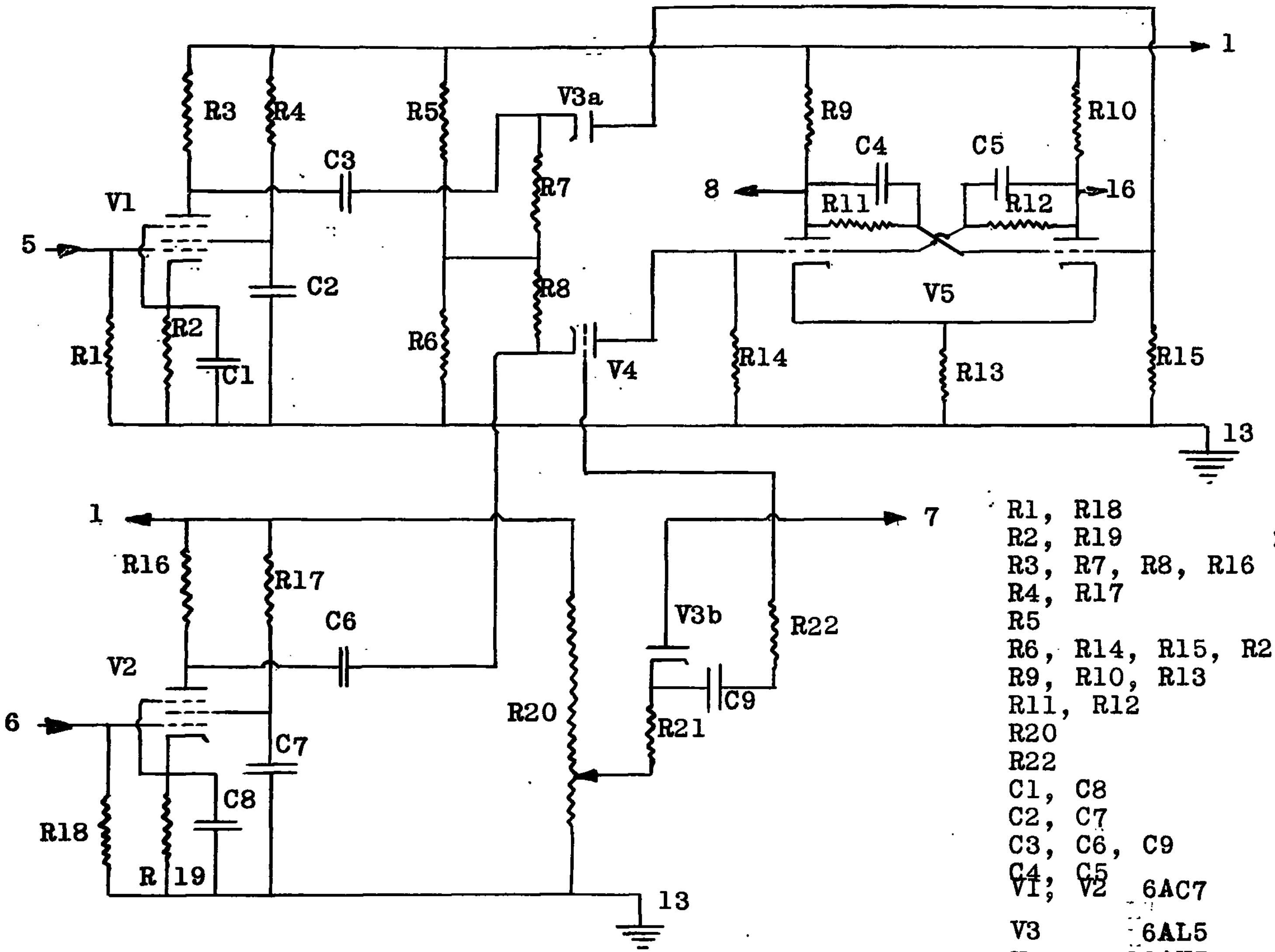
Figure 13. Deflection Amplifier, Multiplier,

Subtractor, and Feedback Circuit.



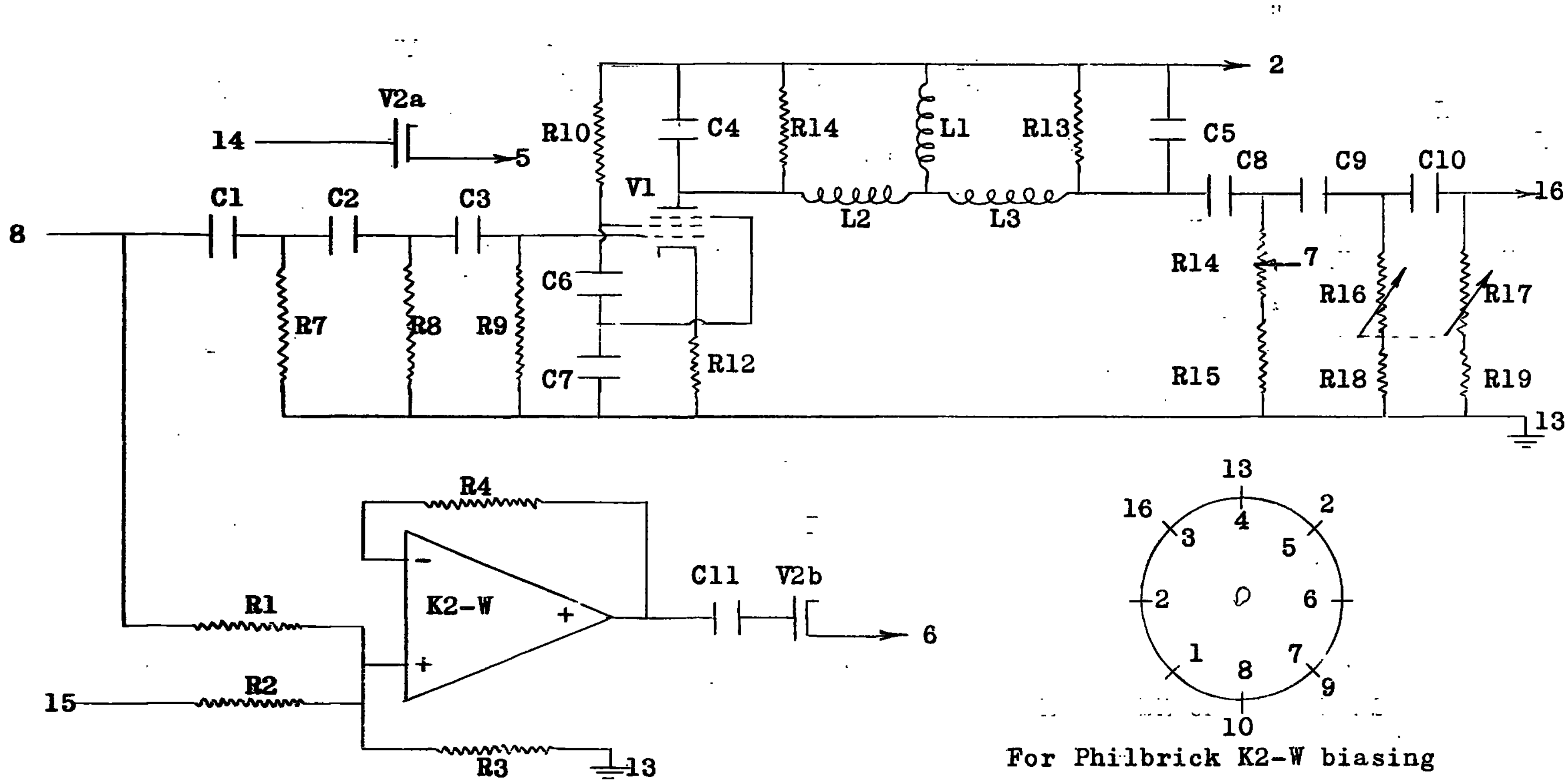
R1, R6, R12	6.8 kilohm	R17, R19	0-250 kilohm
R2, R7	100 "	R18	10 "
R3, R8, R20	68 "	R21, R22	6.7 "
R4, R5, R9, R10, R15	2.2 megohm	C1, C2, C3, C4, C5	0.001 µf.
R11	2.2 kilohm	C6	100 µµf.
R13	33 "	C7, C8, C9	8 µf.
R16	220 "	V1, V2, V3	12AU7

Figure 14. Cathode-Coupled Clipper or Squaring Circuit



R1, R18	1	kilohm
R2, R19	270	ohmn
R3, R7, R8, R16	10	kilohm
R4, R17	68	"
R5	0.2	megohm
R6, R14, R15, R21	0.1	"
R9, R10, R13	22	kilohmn
R11, R12	150	"
R20	0-1	megohm
R22	0.25	"
C1, C8	8	μf.
C2, C7	0.1	"
C3, C6, C9	0.01	"
C4, C5	50	μμf.
V1, V2	6AC7	
V3	6AL5	
V4	12AX7	
V5	12AU7	

Figure 15. Differentiating Amplifier and Flip-Flop



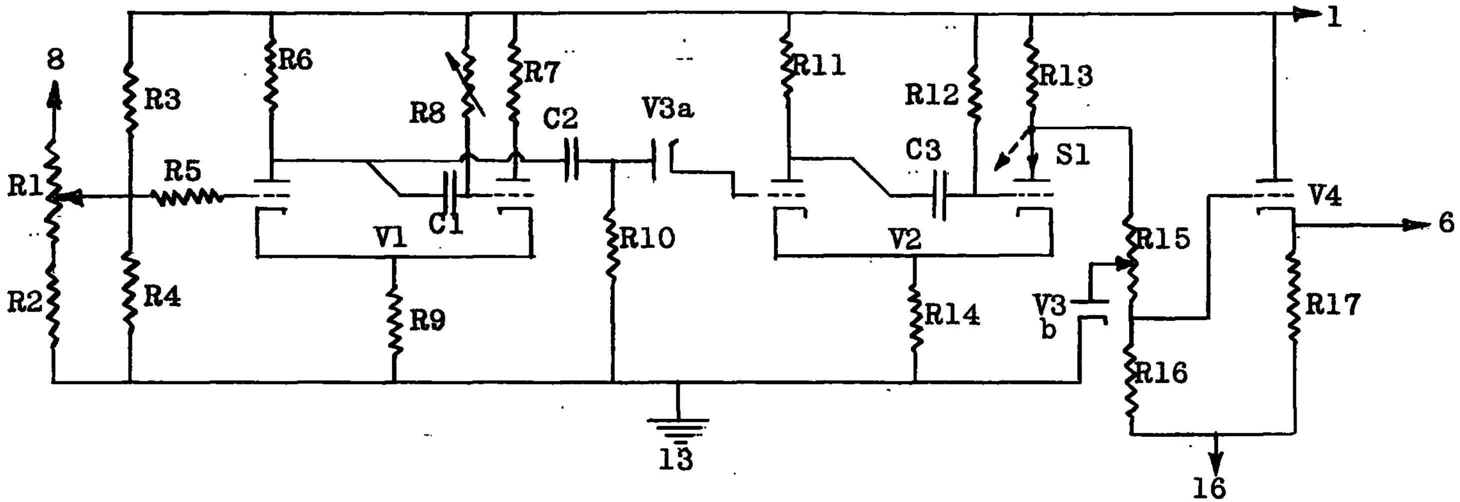
For Philbrick K2-W biasing arrangement see Figure 9

R1, R2, R3, R4	100	kilohms
R7, R8, R9	1	megohm
R10	68	kilohms
R11, R13	47	"
R12	330	ohms
R14	100	kilohms
R15	500	"
R16, R17	0-30	"
R18, R19	27	"

L1	10	mh.
L2, L3	120	"
C1, C2, C3	500	μμf.
C4, C5	0.015	μf.
C6, C7	0.01	"
C8, C9, C10, C11	0.001	"

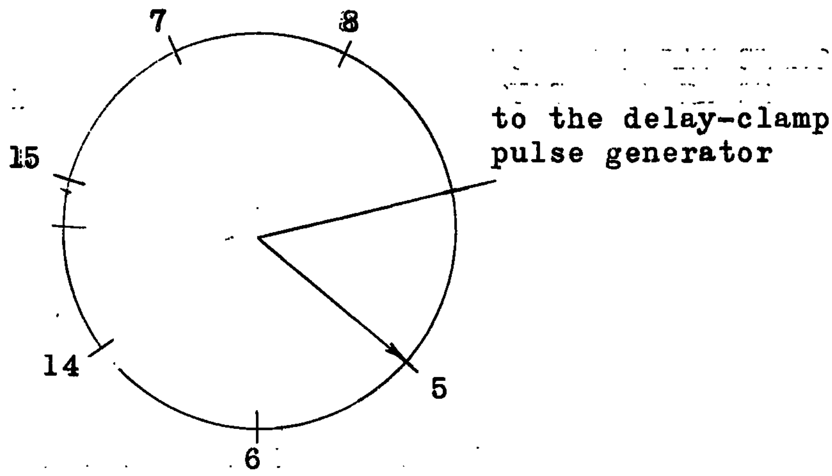
V1	6AU6
V2	6AL5

Figure 16. Z-axis Carrier-Signal Filter, Magnitude Output, and Polar Angle Output.



R1	0-100 kilohm	R15	0-500 kilohm
R2	500 "	R16	0.68 megohm
R3	10 megohm	R17	0.22 "
R5	o.5 "	C1	1 µfd.
R6, R7, R11, R13	0.15 "	C2, C3	0.1 "
R8	0,5-3.0 "	V1, V2	12AX7
R9, R14	10 kilohm	V3	6AL5
R10, R12	1 megohm		

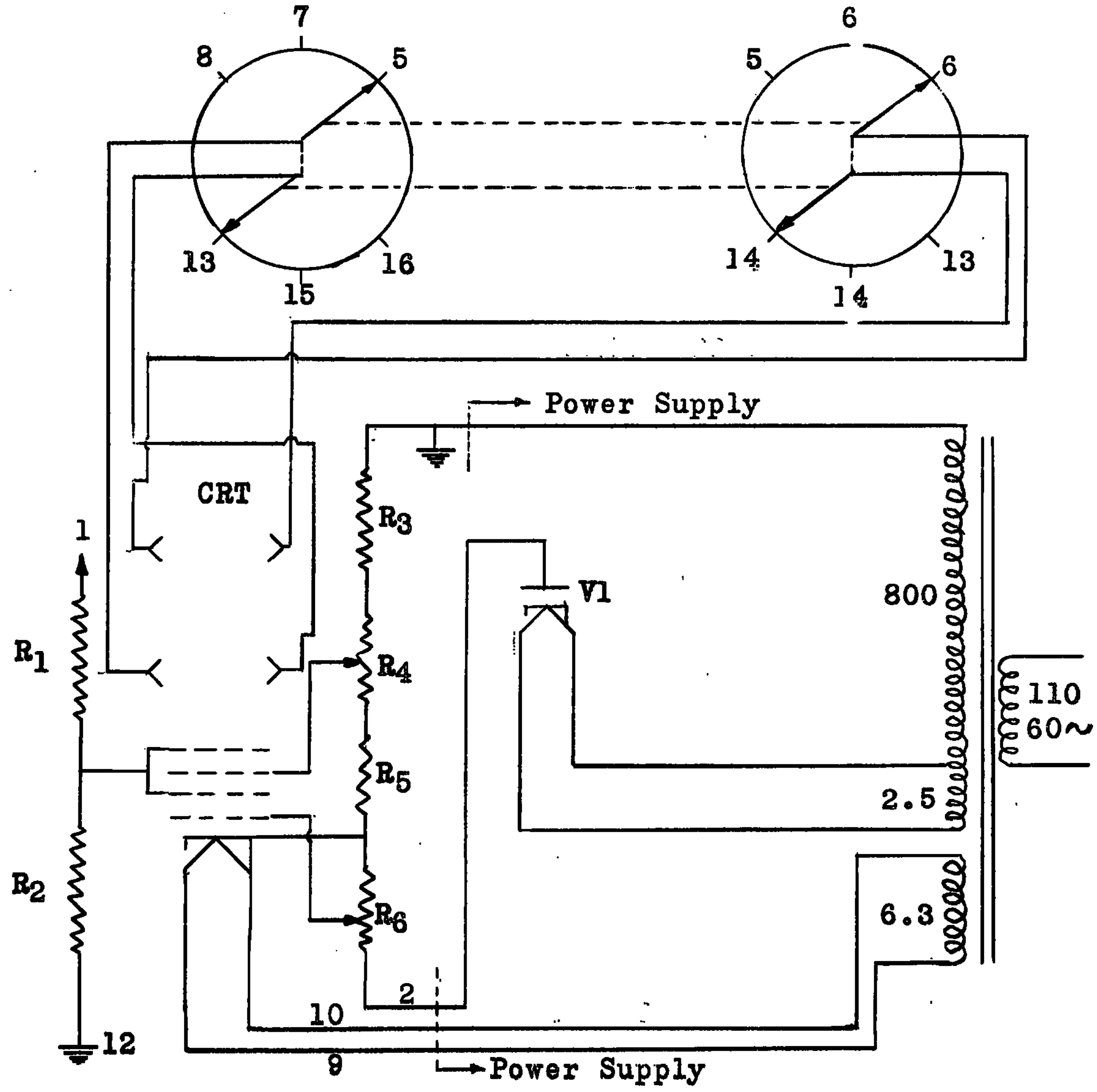
Figure17a. Delay Circuit and Clamp-Pulse Generator



single deck rotary switch

the numbers represent the terminals to which
the deflection-amplifier outputs are connected

Figure 17b. Switch for Selecting Trigger
Pulse for Generator



R1, R2	- 150 Kilohms	Transformer	214-60
R3	- 3.3 Megohms	CRT	RP1
R4	- 2.0 Megohms		
R5	- 1.0 Megohms		
R6	- 0.5 Megohms	V1	- 2X2

Figure 18. Cathode-Ray-Tube Circuit and its View-Selector Switch.

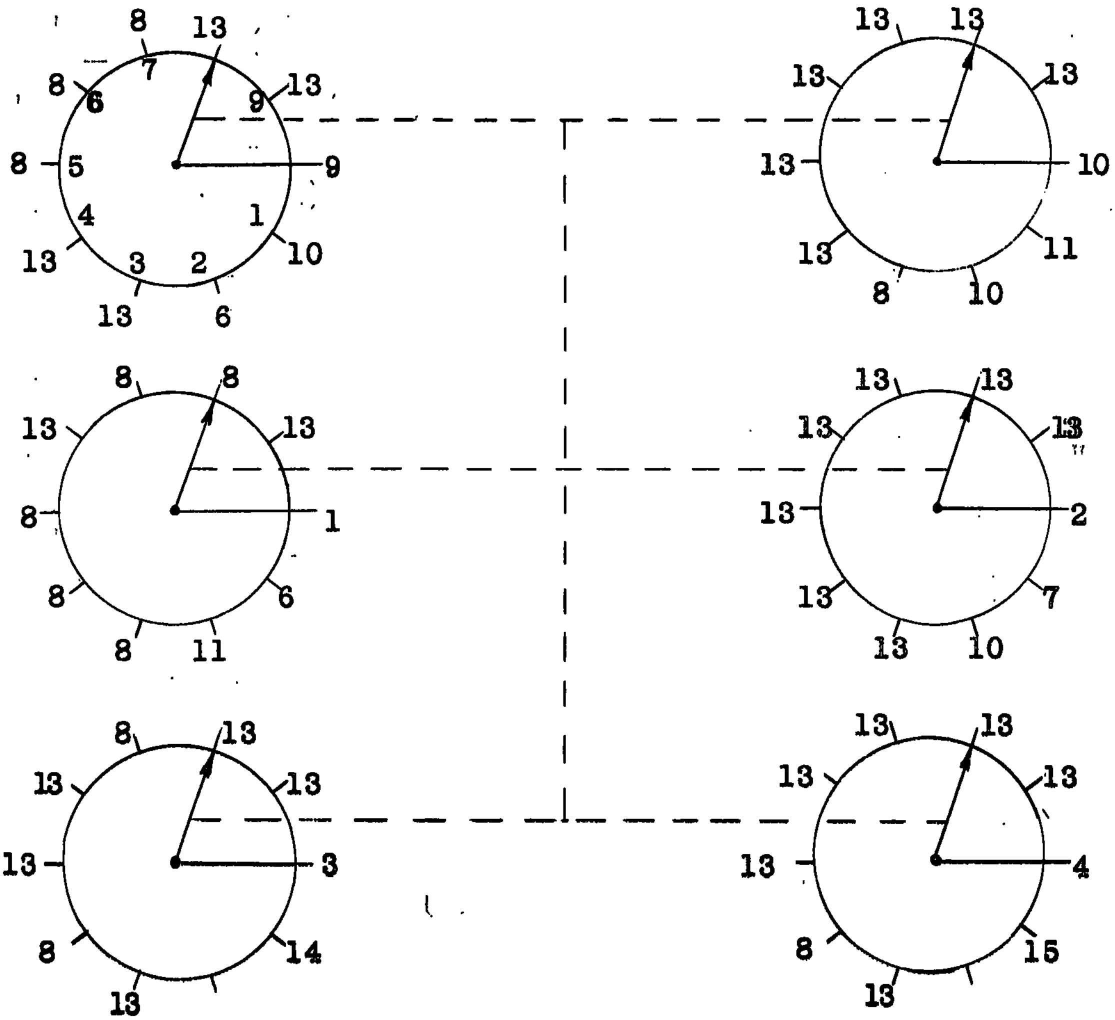


Figure 19. Input Rotary Switch.

clamp generator trigger selector	input amplifier	deflection amplifier multiplier section feedback circuit	add, filter and band-pass amplifier	squaring circuit	phase meter and threshold control
oscillator and time marker	input amplifier	deflection amplifier multiplier section feedback circuit	add, filter and band-pass amplifier	squaring circuit	rectifier, add, filter and band-pass amplifier
input rotary switch calibration multivibrator	input amplifier	deflection amplifier multiplier section feedback circuit	squaring circuit	view selector CRT focus intensity	

Figure 20. Proposed Unit Positions.

IX CONCLUSION

The spherical polarcardiograph is an electronic device which transforms voltages proportional to the Cartesian projections $\underline{x}$, $\underline{y}$, and $\underline{z}$ of the heart vector to voltages proportional to the spherical polar coordinates $\underline{r}$, $\underline{\theta}$, and ϕ.

The gated feedback discussed in chapter III and the voltage regulation of the more critical units discussed in chapter VI result in a stable computer. The feedback loop which is closed between heart beats rebalances the instrument sufficiently so that the manual balance procedure need be carried out only once after the computer is warm.

The method of introducing the third coordinate is simple but it leaves a cylindrical volume surrounding the polar axis in which no information can be obtained. Although very little information will be lost in this region, it can be restored by exchanging the inputs.

Although the computer has not been constructed in final form, tests on the individual units have indicated that the device will be well within accuracy requirements. However, only prolonged clinical tests can determine the ultimate usefulness of the spherical polarcardiograph in the field of medical research.

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