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AN INSTANTANEOUS RESPIRATORY TACHOMETER

by

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A THESIS SUBMITTED
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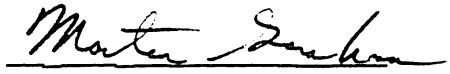
REQUIREMENTS FOR THE DEGREE OF

MASTER OF SCIENCE

Thesis Director's signature:



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ABSTRACT

The development of an instrument which measures the rate of respiration from breath to breath is discussed. The device uses as its input the impedance change of the chest cavity associated with respiration. It is designed to be part of the Physiograph system. The tachometer contains a trigger which produces a set of pulses corresponding to the points of maximum inspiration. The pulses control circuitry which measures time between the pulses. The time measurement is converted to a measure of rate that is linear within 5% from 5 to 100 breaths per minute.

Included is a discussion of the effect of component drift in the switching circuitry. Also described is a method of obtaining linearization of the rate with an RC integrator using a non-linear resistor. Finally, the trigger, which undergoes transition when the impedance signal moves a set amount from the relative maxima or minima, is discussed. Records of the tachometer operation in the Physiograph system are included.

ACKNOWLEDGEMENTS

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INTRODUCTION

Tachometers used to count the heart rate have been valuable for many years. These cardiotachometers generally fall into two categories. One consists of those which are average reading tachometers, such as the one developed by Roy.[1] In this device the d-c component of a pulse train is sampled to get a measure of the rate. The second type consists of the instantaneous tachometers, which give the exact rate from beat to beat. The rate measure is produced by integrating a constant voltage with respect to the time between beats. Appropriate non-linearity may be added to give an output which is linearly proportional to the rate. Such a device was constructed by Manzotti in 1956.[2] Though a respiratory tachometer would often be a time saving convenience, it was not until 1960 that the impetus for this project was provided by the work of Clynes[3]. He showed the relation between instantaneous heart and respiratory rates, and he also demonstrated the existence of a respiratory control mechanism of the heart rate. An instantaneous respiratory tachometer is needed both to study this relationship further and to exploit the possible diagnostic information in it. The difficulty of obtaining a signal which mirrored respiration inhibited the development of such a device. The tachometer proposed here is designed to be part of the Physiograph system. It relies upon the recently developed impedance pneumograph to provide a signal proportional to the volume change of the chest cavity associated with respiration. However, this signal is

not as convenient to use as the electrocardiograph signal is for tachometry. Whereas the QRS segment of the electrocardiogram provides a clean reference pulse, the impedance pneumograph signal is slowly moving and confused by artifacts from the heart's impedance change and from extraneous body movement.

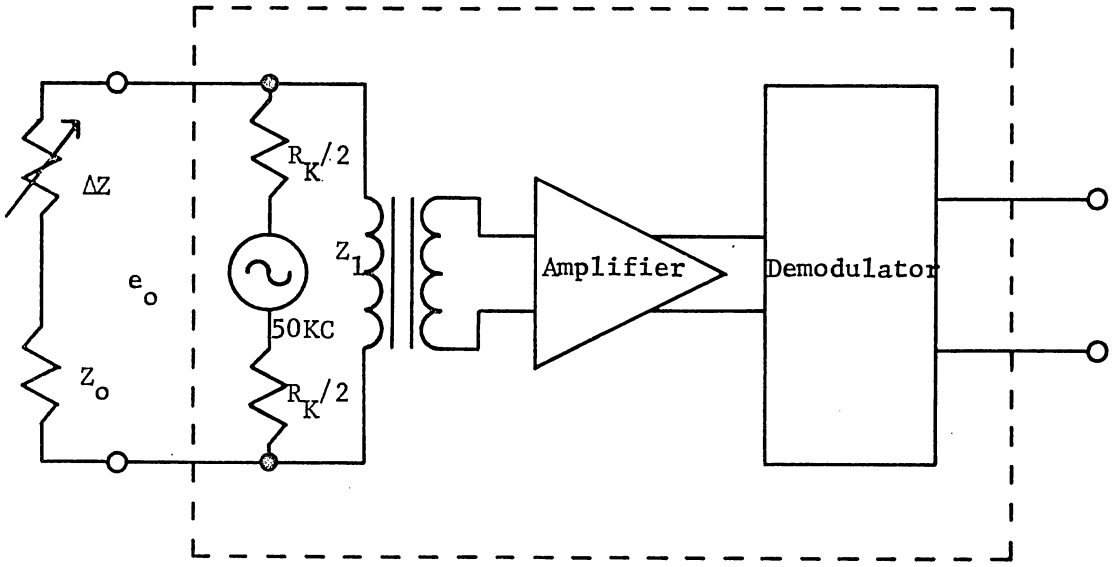
Thus it is evident that the most significant problem for a tachometer using this signal is the detection of the fundamental frequency reflecting the respiration rate. The method of detection must also be consistent from one breath to another in order for the meaning of the output of the device -- in terms of rate -- to remain constant. The other differences which arise are not so fundamental. First, since the respiratory rates are generally slower than cardiac rates, it is desirable to maintain an output between breaths, not just at the end of a cycle. To do this easily a double set if integrating circuits was used. Second, the range of respiratory rate change is greater than that of the heart rate. It is more difficult to provide a linear output for a given error over a range of 20 to 1 than for the common cardiotachometer range of 4 to 1. As mentioned previously, the link between the subject and the tachometer is the impedance pneumograph. Therefore, it is proper to describe this device briefly and to assess its potential as an indicator of respiration.

The Impedance Pneumograph

Measurement of respiration has been done in many ways. The pneumotachograph and the spirometer are the instruments used to give

accurate measurements of air velocity and volume. Both have the disadvantage that their input source is the air stream of the subject. The pneumograph, an extendable belt placed around the chest, alleviates this difficulty, but is leaky and impossible to calibrate. The impedance method is convenient. It is possible to use electrocardiograph electrodes to make the measurement. It is also possible to quantitatively relate the impedance change to the volume change and to maintain that relation so that calibration is possible. The first step in the development of the impedance pneumograph was the choice of the frequency at which the impedance is to be measured. This frequency must be high enough so that no internal tissue is excited and no sensation is noticed under the electrodes. Thus the circuit required an osciliator, an amplifier, and a demodulator. A block diagram of the circuit developed under Dr. Geddes at Baylor University is shown in figure 1[4].

The low level of the oscillator and its high output impedance insure complete safety for the subject. Since the impedance to be measured is usually a variation of a few ohms about a constant impedance of several hundred ohms, it is desirable to capacitively couple the output to be used with the tachometer. Only the variation is important for the rate measurement. With the resting level removed, the gain may be increased to make the variation cover the full range of output voltages. Placement of the electrodes is also important in determining the maximum signal obtainable. For both man and dog,



BLOCK DIAGRAM OF THE IMPEDANCE PNEUMOGRAPH

FIGURE 1

the position of the sixth rib appears to be optimum. Thus it is possible to get a large signal from which the tachometer must determine the rate. However, the question of whether or not this signal truly reflects respiration must be considered.

Impedance Spirometry

Work on impedance spirometry is currently being conducted at Baylor University by L. E. Baker[5]. For test purposes the impedance pneumograph is direct coupled. Thus the transthoracic impedance is characterized by a constant impedance, Z_o, measured at the resting respiratory level plus the change in impedance, ΔZ, during respiration. R_K and Z_1 are much larger than $Z_o + \Delta Z$, thereby maintaining a constant current in the subject over a very wide range of electrode impedance. Therefore, the changes in e_o are directly proportional to the changes in ΔZ, if Z_o is between 100 and 1000 ohms. The voltage developed across ΔZ is linear over a much greater range than is encountered in practice. Typical resting impedance for a 150 pound, 5' 10" man is 159 - j80 ohms. With an increase in frequency there is a continuous decrease in impedance, primarily in the reactance. However, the change in impedance from the resting expiratory level to full inspiration is approximately 20 ohms for all frequencies, since the change is essentially resistive. The typical record shown in figure 2a also indicates the time correlation of the two events. For analysis, the curves showing the relation between impedance and volume are

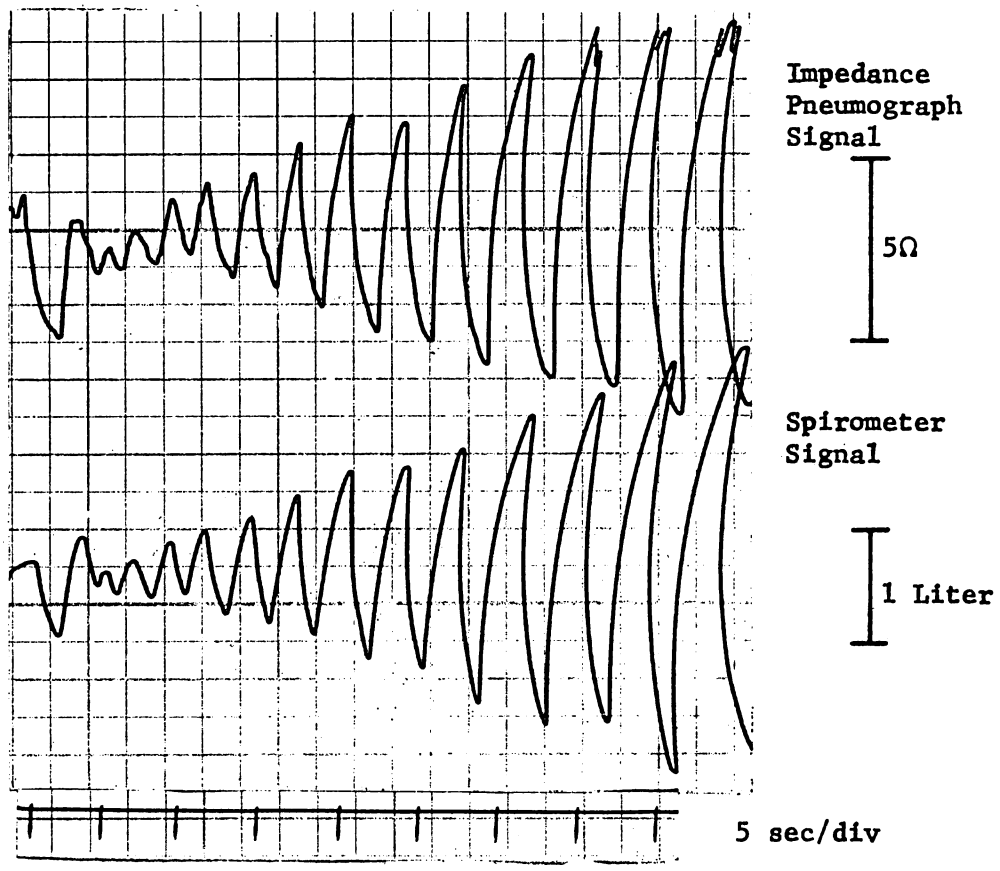


Figure 2a

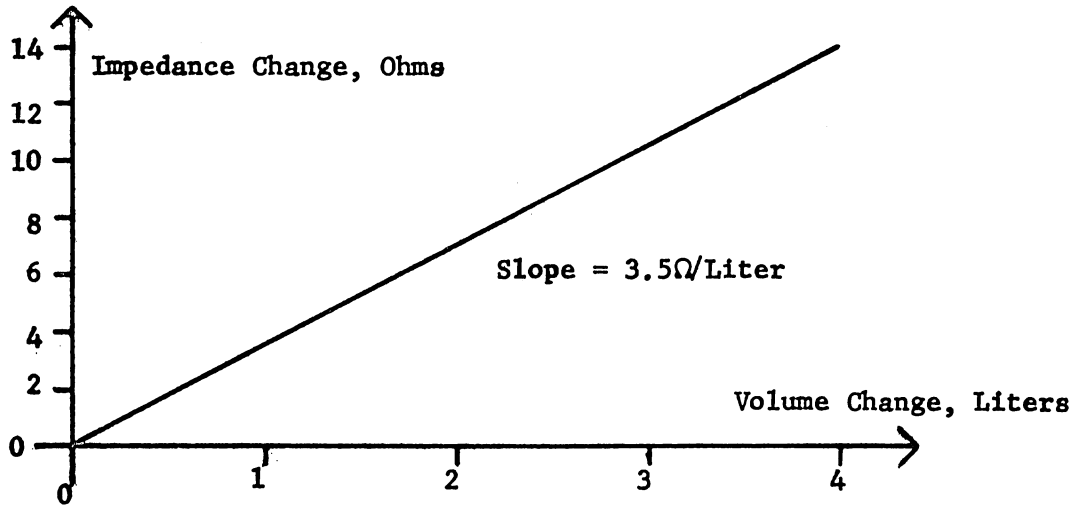


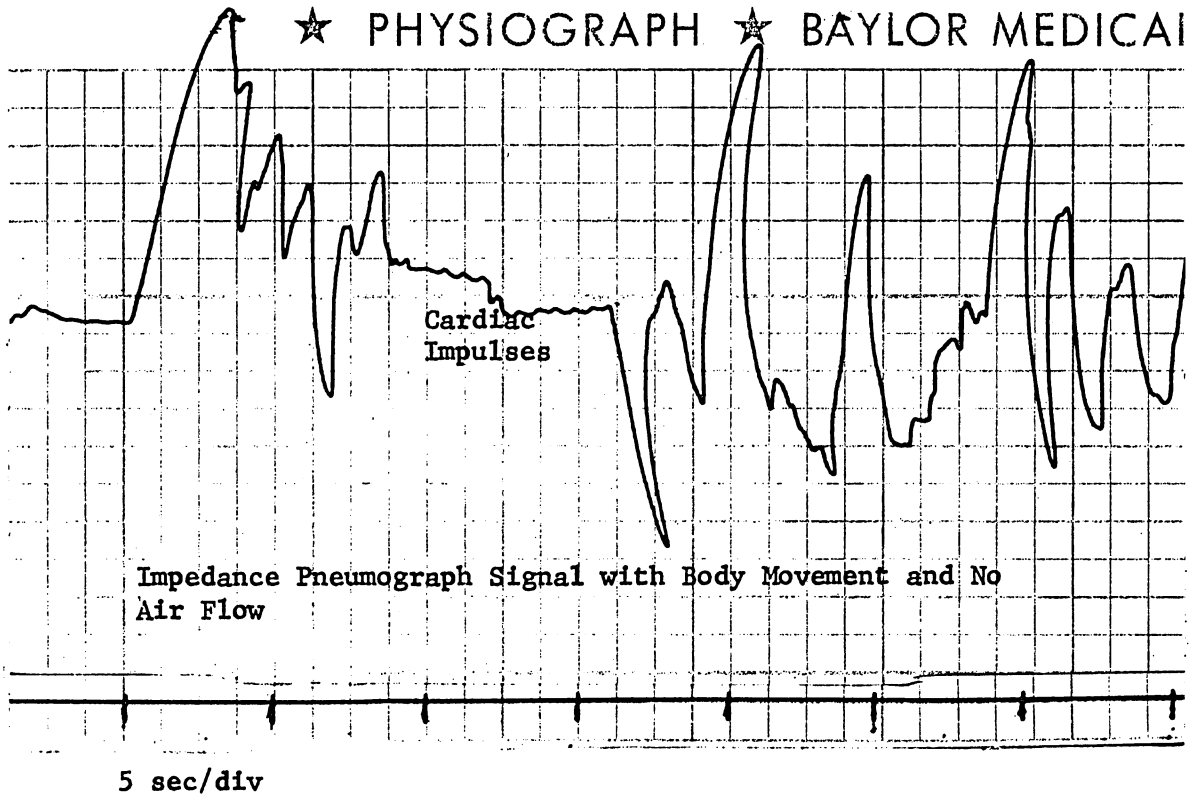
Figure 2b

THE IMPEDANCE - VOLUME RELATIONSHIP

FIGURE 2

usually divided into three areas. Whithin any one of these regions -- normal, heavy, or deep breathing -- the error is less than 16%. The curve in figure 2b is a plot of the data shown in figure 2a. It indicates an overall relation of about 3. 5 ohms per liter, typical of individuals with medium builds.

It should be noted, however, that it is possible for the impedance pneumograph to generate a signal which is similar to the signal associated with respiration, but which does not involve air exchange. Figure 3 shows such a signal, generated by body movement alone with the electrodes in the standard position, on a midaxillary line at the xiphoid level. Change in electrode position may prevent the generation of such spurious signals in some cases. In general the conditions of each test must be examined for suitability before this method is used. Now that a signal which is a measure of respiration has been obtained and evaluated, a device must be designed that can use this information to determine the rate of respiration.



ARTIFACTS IN THE IMPEDANCE SIGNAL

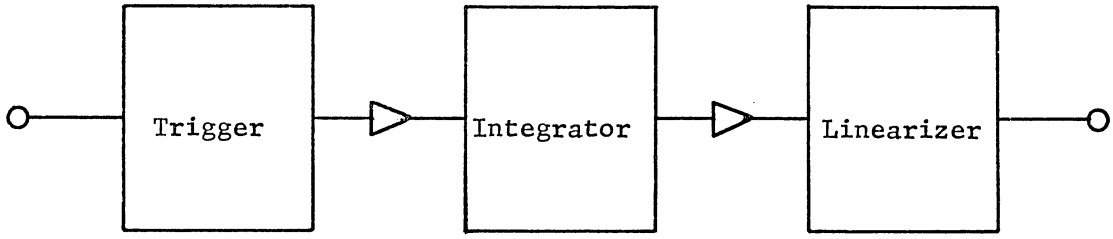
FIGURE 3

DEVELOPMENT

In order to extract the rate from the impedance pneumograph signal, the tachometer integrates a voltage with respect to the time between breaths. The integrator then has a voltage on it which is proportional to the time between breaths. To obtain a rate display this voltage is stored and inverted, since rate is inversely proportional to time, and then linearized. A trigger circuit is required to detect the interval between breaths. Thus the basic modules in the tachometer are the trigger section, the integrator section, and the linearizer section. The signal flows through these modules sequentially as shown in the block diagram of figure 4. Since the nucleus of the tachometer is the integrating circuitry, it was designed first to insure that the necessary logical operations could be carried out.

The Integrator

In most cardiotachometers only one integrating element is used. The integrator is sampled at the end of each cycle, the voltage inverted and displayed briefly, and then the element is reset to begin integrating over the next cycle. Since the respiratory rate is rather slow, this method gives no output over most of the cycle. To provide a continuous output two integrators can be used. While one is integrating, the other holds the value it had reached at the end of the previous cycle. In the first tests, Philbrick K2X operational amplifiers, controlled by a three pole, four position switch were used to do the integrating.



BLOCK DIAGRAM OF THE TACHOMETER

FIGURE 4

Each operational amplifier was controlled by a wafer and the output was controlled by the third as shown in figure 5a. In any position one integrator charged, the other held its output. In switching between hold and charge the integrator was reset by connecting the output to ground momentarily. The circuit worked as expected. The next step was to implement the switching electronically.

It was decided to use relays controlled by one bistable and two monostable multivibrators as shown in figure 5b. The difference here is that instead of resetting before beginning the charging cycle, the resetting and charging begin simultaneously. This difference merely provides an effective delay in starting the integrating. However, it does cause a minor problem. If the switching done by the bistable multivibrator is not completed by the time the monostable multivibrator begins to reset, a spike appears at the output. It can easily be removed with shunt capacity. The control circuitry was designed for use with the operational amplifiers so that its voltages are different from those of the rest of the circuit. This difference is neither essential nor desirable. However, the circuit existed and it worked. To redesign for Physiograph voltages would add nothing worthwhile to the project.

The pulse used to trigger the bistable multivibrator in the original tests was provided by the trigger stage of the Mark V Respiratory tachometer developed at Baylor. It differs from the circuit described here in that it uses a different method of triggering and the

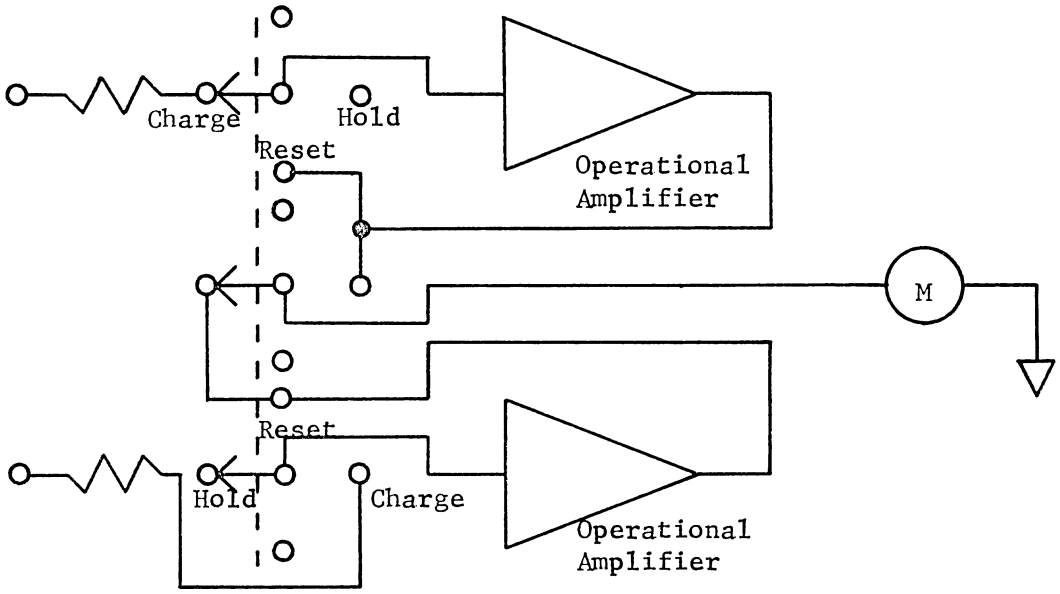


Figure 5a. Switched Integrator Circuit

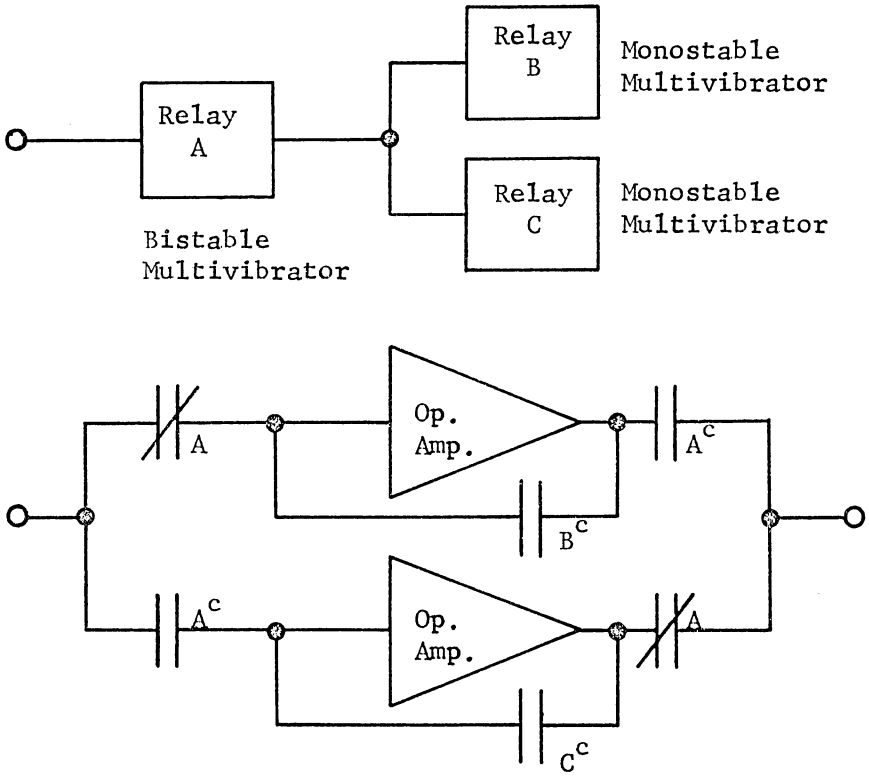


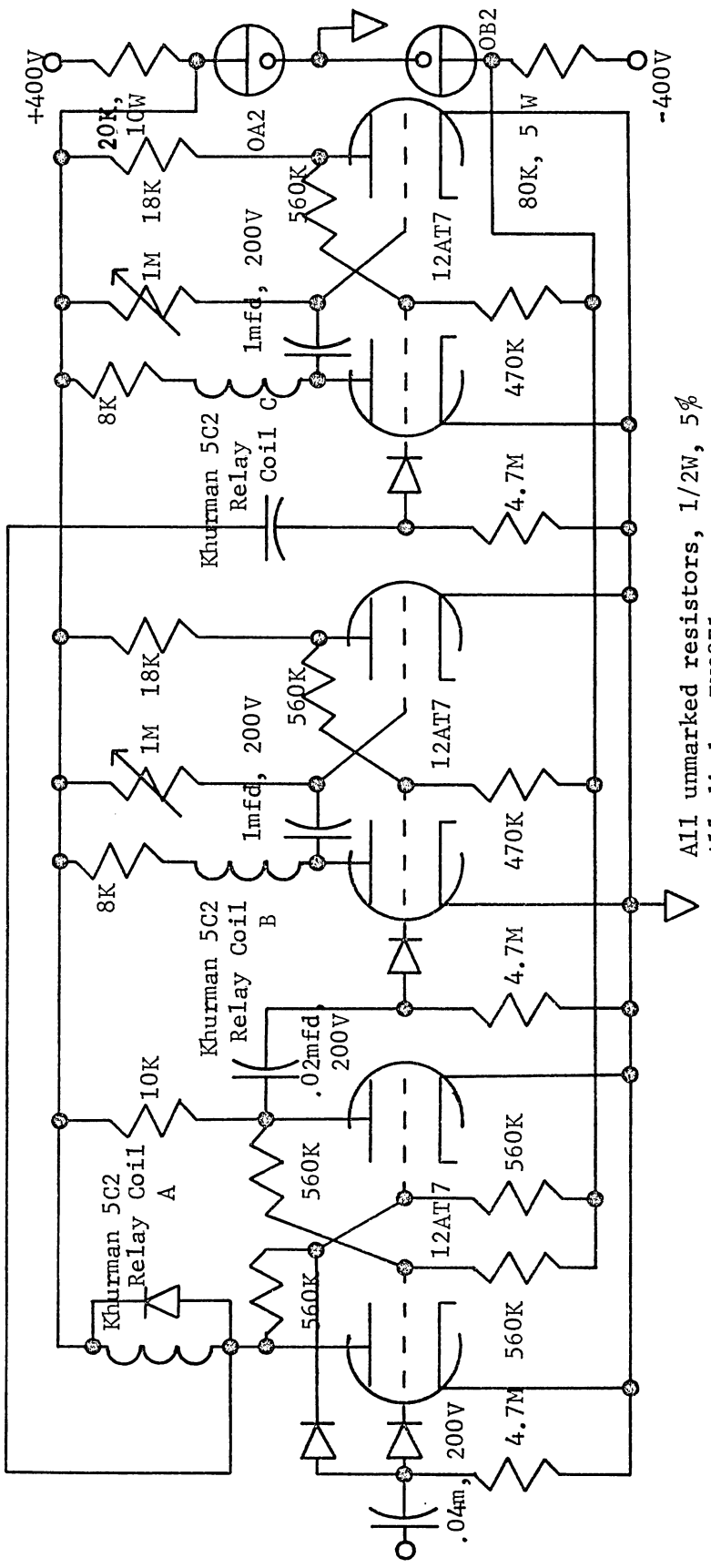
Figure 5b. Relay Controlled Integrator Circuit

BLOCK DIAGRAM OF THE OPERATIONAL AMPLIFIER INTEGRATOR

FIGURE 5

display is not instantaneous. It counts trigger pulses for one minute and holds the count for one minute. This pulse is a 10 volt, 10 millisecond pulse. Tests were made putting the impedance pneumograph, the Mark V tachometer, and the integrating circuitry into the Physiograph system. The output, inverted for display, was then calibrated in terms of rate. With the success of these tests, the integrator control circuitry was established as that shown in figure 6. The next task was to simplify the integrating circuitry from an operational amplifier to an RC integrator.

The first capacitors tried were very large capacity electrolytics. These large capacitors were used in the hope that they would be able to store enough energy to drive the meter, a 0--100 d-c microampere ammeter, during the hold phase of the operation. It was soon learned that the time constant of the electrolytics themselves, independent of the external circuit, was too short to maintain the output voltage with sufficient accuracy. The final choice for the integrating capacitor was a 10 microfarad ± 10%, 400 volt mylar capacitor. With this choice the design of the integrating circuitry was complete. The RC integrator circuit is shown in figure 7. Intimately connected with the integrating circuit is the linearizing network used to conver the capacitor voltage to a linear measure of rate.



SCHEMATIC OF THE INTEGRATOR CONTROL CIRCUIT

FIGURE 6

The Linearizer

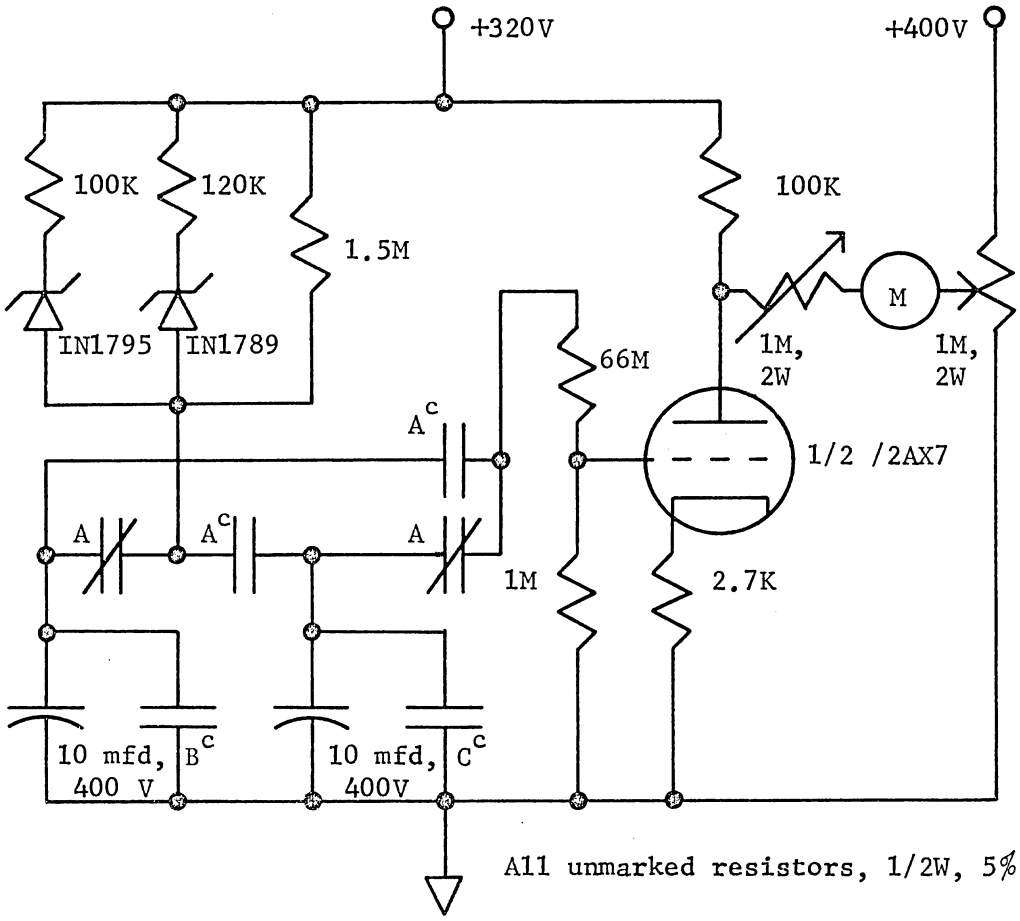
In principle this part of the circuit is the simplest to build, and indeed the final network is simple, yet all the possibilities which had to be explored made this task one of the most time consuming of the whole project. The close connection between the integrator and the linearizer first became a factor in the design when the original linearizer, a diode network between the capacitors and the meter, had to be abandoned because it was impractical to use mylar capacitors large enough to supply the required energy. Thus an amplifier had to be incorporated into the linearizer design. If an amplifier with phase reversal is used, an added advantage is gained in that the capacitor voltage is inverted. The first amplifier used a 2N217 transistor. It seemed natural to use a transistor since, besides the convenience, the capacitor in series with a large resistance drove the transistor as a current source. Also for a PNP transistor the slowest rate can drive the output as near to cut off as desired, so that there is no constant voltage component of the output to be eliminated. With the 2N217 the largest practical series resistance was ten megohms. This gave a discharge time constant of 100 seconds, which allowed a 10% change in the output voltage at the slowest rate to be measured. Though the time constant is too short, a more sensitive transistor would make this approach practicable, if it proved worthwhile with the 2N217. Since the input is a current source, the gain of the amplifier is just β times the ratio of the resistance in the collector to that in

the base. The problem then was to calculate the output voltage of the transistor as a function of rate. Since the output had to be linear with respect to rate, the transfer function of the diode network could be found. It would then be a simple task to construct such a network. The construction of this network is simplified by the appropriate choice of the charging time constant. If the shape of the capacitor voltage curve with respect to time were hyperbolic, no diode network would be necessary. Consequently, the diode transfer function was calculated for numerous time constants until by trial and error a suitable combination was found.

Either a series or a shunt diode network could be used. In order to use diodes in a series network, the supplies must be floating. Such supplies are not available in the Physiograph. It is very difficult to use zener diodes since extremely close control on the zener voltages is required. The total collector voltage variation is only four volts. The shunt network is also difficult to manipulate, and without adding another stage the transistor is heavily loaded in maintaining a linear output. To solve the problem, it was decided to use a tube for amplification. The tube gave increased input impedance so that the discharge time constant could be increased to 700 seconds. It also provided a much greater swing in the output voltage. However, in order to operate the diodes and still maintain some degree of independence from the meter controls, the tube, a 12AT7, could not be operated linearly over the full range of input voltages for any charging time

constant which utilized a significant portion of the available capacitor voltage. Another problem associated with the diode network in the plate is that drifts in the tube and power supply would disturb the calibration.

In order to reduce the number of components which affect the calibration and in order to provide independence of the linearing network and the meter controls, the diodes were placed in the charging circuit. Thus the voltage variation on the capacitor with time is made to approximate an inverted hyperbola. All that is required of the amplifier is that it invert the capacitor voltage linearly. A shunt diode network was tested here with the same results as obtained previously in that the loading was excessive and the interrelations between the shunt paths were too complex to deal with easily. Thus the final choice for the network was a series zener diode array. The output voltage was then plotted as a function of the charging time constant and the reset time. When the combination which gave the longest linear portion was found, it was extrapolated and set as the relation to be obtained. The next step was to build up the network starting with the lowest rate, since here the zener network is essentially one resistor. To find this resistor value, the capacitors were not reset to zero, but to the voltage associated with a given rate. The slower rates were then recalculated in terms of this rate. The fastest such rate to which the extrapolated line could be fitted then became a break point of the diode network. In this way the relation between rate and voltage was fitted to the desired



SCHEMATIC OF THE LINEARIZING CIRCUIT

FIGURE 7

line. For the fastest rates the effective delay, controlled by the time the monostable multivibrators are in the quasi-stable state can be used to fit the relation to the line. For the linearizing circuit shown in figure 7, the effective delay time is 400 milliseconds. Given a set of pulses it was then possible to measure the instantaneous rate. The problem that remained was to generate the pulse to control the display circuitry.

The Trigger

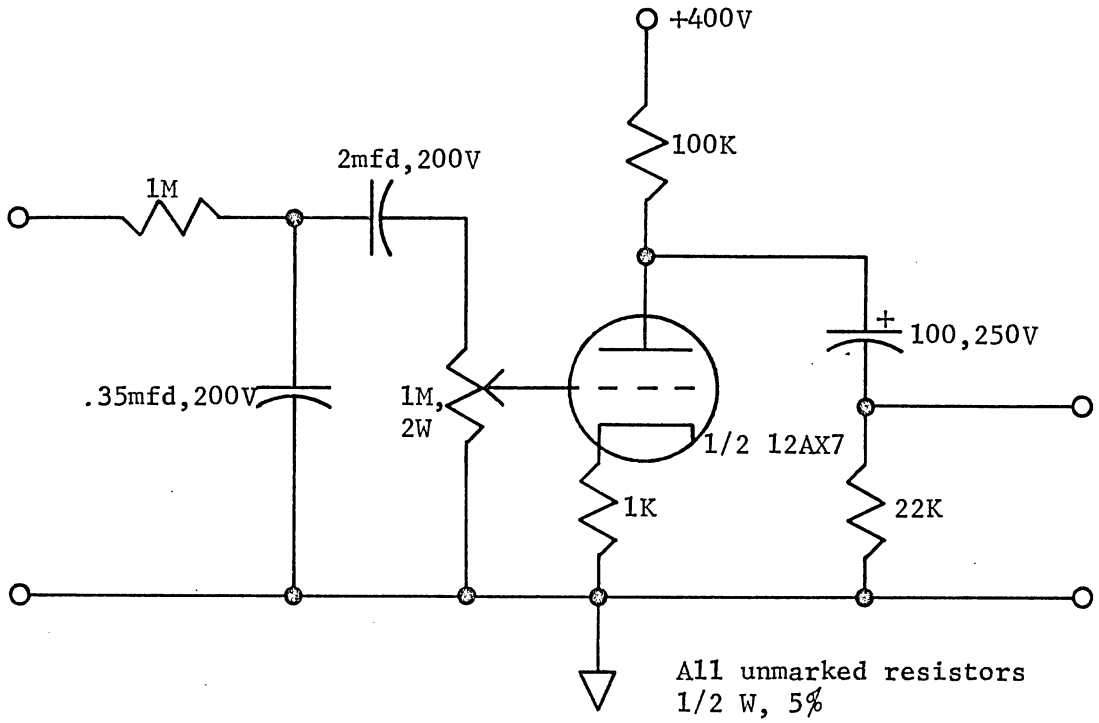
The records of the impedance pheumograph signals in figures 2 and 3 show that they are far from regular. The signal does not vary about any constant voltage, the variation is not constant from breath to breath, and unwanted signals from cardiac impedance change and body movement are included. The nature of the signal eliminates a threshold trigger circuit as a possible triggering scheme. Perhaps the simplest appropriate trigger circuit is that employed in the Mark V tachometer developed at Baylor University. Here the input signal is squared and clipped. The trigger pulse is obtained by differentiating this squared pulse. This circuit has several disadvantages. It depends upon the value of the input voltage with respect to ground. Noise occurring during the transition segment of the squared pulse may produce a spurious output pulse. The size of the change which produces a pulse depends upon its frequency components. It was felt that a different type of triggering circuitry should be explored. The principle to be investigated involved a modification of the threshold technique.

The modification was to set a new reference for the threshold each half cycle. The reference is either the maximum positive or negative excursion of the signal. A pulse is generated whenever the signal varies a set amount, the threshold level, from the reference point. A technique of this type offered the advantages that the actual value of the input signal with respect to ground was not important, no noise could generate a spurious signal unless its magnitude exceeded the threshold level, and the size of the signal which produced a pulse did not depend upon its frequency components.

However, this method of triggering does possess several disadvantages. Setting the threshold level is complicated in that two must be set -- one for the maximum positive excursion, the other for the maximum negative excursion. Another complication comes from consideration of what is meant by rate. Since the signal is nonrepetitive, a measure of the time between similar points on the curve is meaningless. Becuase the inspiration peak is sharper, it is usually chosen as the reference for rate measurement. However, depending upon how fast the signal moves from peak to threshold, the pulses may not have the same interval between them as the peaks have. The problem is obviously minimized by using the smallest threshold level which will protect against spurious pulse generation. With these reservations in mind, the development of the circuitry capable of implementing this mode of triggering can be described.

In the Physiograph system the signal from the impedance pneumograph is amplified for display. This amplified version, which is usually a 20 volt peak to peak signal, was used as the input to the tachometer. The only unfortunate aspect of this signal is that for certain applications, the interference from the caridac impedance changes is comparable in amplitude with the smallest breaths to be counted. However, the cardiac impulses are of much higher frequency than the breaths. Therefore, they can be attenuated by filtering. A simple RC band pass network with a gain control was used. The low frequencies were attenuated to remove some of the base line drift. At the frequencies involved, the output impedance of the network necessarily had to be made large in order to keep the size of the capacitors within reason. This output impedance had to be reduced to drive the trigger. The output of the filter network was amplified by half of a 12AX7. The other half amplifies the integrator output. The output impedance of the 12AX7 was reduced by decoupling the output through a 100 microfarad electrolytic capacitor to a 22K ohm resistor. The gain of the circuit shown in figure 8 is about unity in the mid-frequency range for the gain control approximately 25% of maximum. Thus the circuit gives a large signal, properly filtered, from a relatively low output impedance for the trigger circuit.

The first circuit designed to perform the function described above was built in two stages. The first stage was a differential amplifier, and the second was a Schmitt trigger. Setting the reference



SCHEMATIC OF THE FILTER CIRCUIT

FIGURE 8

point was accomplished by placing the coil of a relay in the final stage with its contacts in the first stage. Assuming the relay coil was deenergized, a capacitor at one grid of the differential amplifier was charged positively through a diode to the maximum input voltage. As the input decreased the diode became back biased, so that the capacitor retained the maximum voltage. The other grid was made to follow the input with a d-c offset. Thus when the input had moved from the maximum by the amount of the offset, the change in the output of the differential amplifier caused the Schmitt trigger to change states. This event energized the relay, which then connected the capacitor in the first stage to the input through a reversed diode to charge it negatively; the other grid was connected to the input through an offset of the opposite polarity. The process was then repeated and the relay was again deenergized. The output pulse was obtained by differentiating and rectifying the output voltage of the Schmitt trigger. The pulse that was chosen to control the bistable multivibrator corresponded to the inspiratory phase of respiration.

A difficulty that arose when the circuit was tested was the occurrence of oscillations in the Schmitt trigger output. The oscillations were associated with the inductance of the relay coil, so that series resistance was added to the plate circuit, in order to limit the peak current and to damp the oscillations. The extra resistance also increased the size pulse. Series resistance did not eliminate the problem entirely. The largest crossover capacitor which could be

used without creating oscillations made the action of the Schmitt trigger extremely sluggish. The obvious solution was to speed up the action of the first stage. Consequently, sufficient regenerative feedback was added to get the desired pulse width at the output for the slowest frequencies to be counted.

All that remained was to place a controllable offset between the input and the second grid of the first stage to obtain the circuit of figure 9a. Using transistor power supplies in series with the input at zero caused the circuit to operate in a free running mode. The frequency of the spontaneous action depended upon the output impedance of the filter, and the values of the filter, and the values of the reference and crossover capacitors. When a resistor divider network was used to get the offset, the values also affected the spontaneous action. Due to this interaction, the value of the offset voltage was frequency dependent. In order to achieve satisfactory operation, a proper balance between the filter output impedance, the two first stage capacitances, and the offset divider resistances had to be attained. Rather than go through such a complicated procedure to implement the trigger circuit, the circuit of figure 9b was proposed.

The circuit consisted of an amplifier with a small regeneration region followed by a variation of Schmitt trigger. The left hand grid of the first stage was connected to the input through a capacitor. The grid was then returned to a fixed voltage through a pair of diodes. Thus, if the relay in the second stage is eenergized, the grid can go

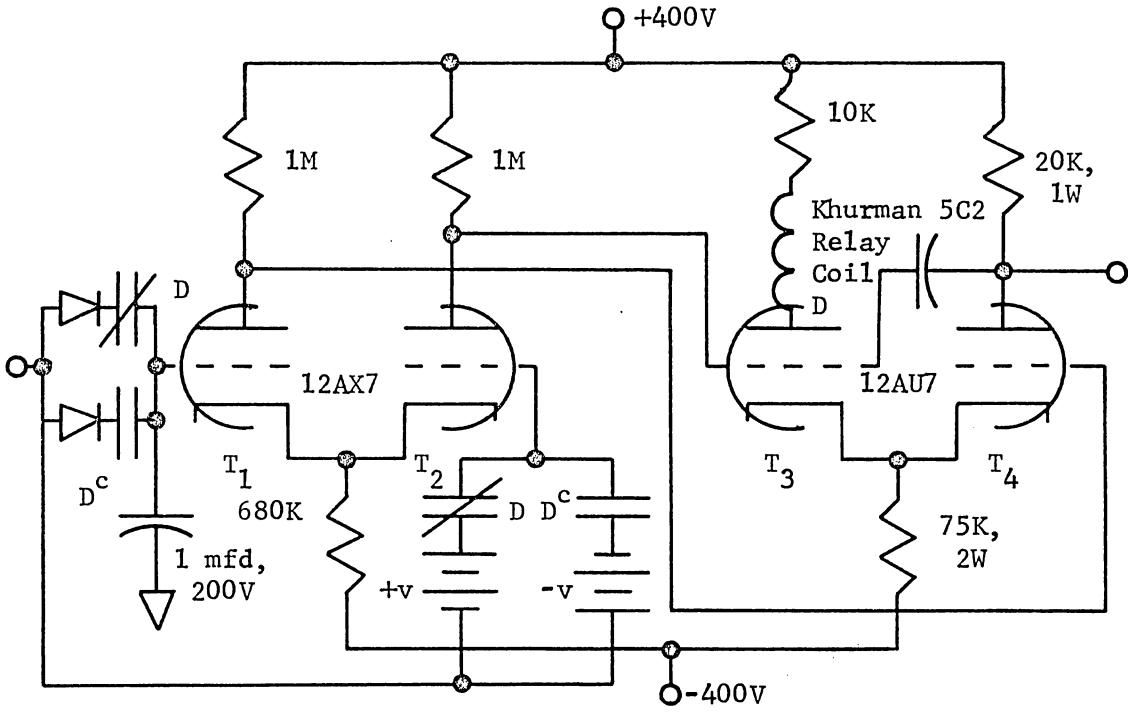
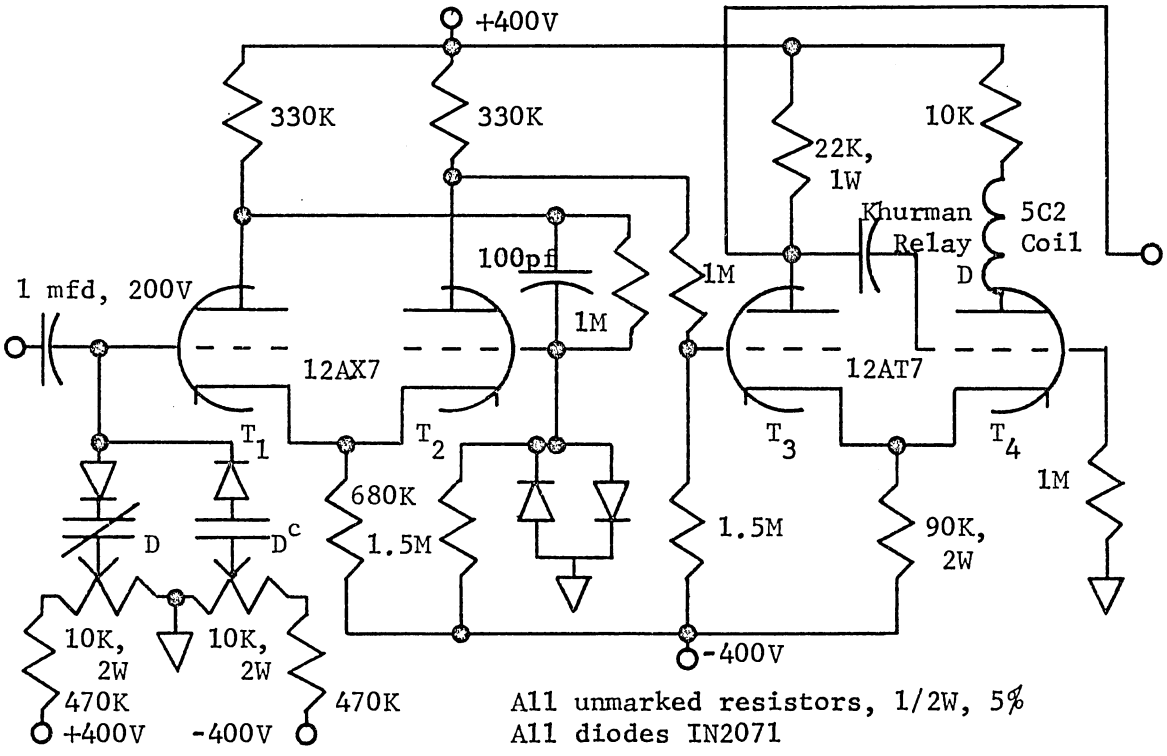


Figure 9a. First Circuit



All unmarked resistors, 1/2W, 5%
All diodes IN2071

Figure 9b. Final Circuit

SCHEMATIC OF THE TRIGGER CIRCUIT

FIGURE 9

positive as far as the fixed positive voltage permits. Then as the input becomes more positive it merely charges the capacitor. However, as the input voltage falls from the peak value, the diode is cut off and the capacitor cannot discharge. Thus the grid voltage moves down with the input. When the voltage on the left hand grid goes though zero regeneration begins. The change in the output is enough to cause the Schmitt trigger to change states. The point of regeneration is near zero because the right hand grid is returned to ground through a pair of diodes. The small diode drops give sufficient hysteresis to prevent spurious triggering. The voltage change at the plate is coupled through a compensated attenuator to the Schmitt trigger. The attenuator was designed so that the steady state voltages at the grid of the trigger were symmetrical about zero. In order to make the neighborhood of zero the region of transition, the right hand grid was returned to ground through a resistor. Regeneration in the trigger is provided by capacitively coupling the left hand plate to the right hand grid. Differentiating the plate voltage yields a fifty volt pulse 2 milliseconds in width. The pulse is used to trigger the bistable multivibrator in the integrator control circuit discussed above. Thus the description of the circuit is complete. To get a better understanding of the operation of the final circuit, it is analyzed in detail in the nest section.

ANALYSIS

The operation of the various portions of the circuit is discussed in general in the preceding section. In this section the operation of the final circuit is discussed in detail for aspects of the operation not generally known and for aspects of the operation of the well understood circuitry which are vital to the proper functioning of the device. The modules are discussed in the same order in which they were developed. The integrator control circuitry is composed of standard bustable and monostable multivibrators. Exposition of these circuits is text book material[6]. However, certain features are discussed in order to indicate what aspects of the circuit are important in the tachometer. The linearizing circuit is simple enough to permit a full description. The basis of the trigger circuit is the Schmitt trigger. However, its unconventional use in the tachometer warrants a thorough discussion.

The Integrator

The bistable multivibrator must have two stable states, such that in one of them the relay in the plate is energized and in the other the relay is deenergized. The monostable multivibrators must have a stable state in which the relay in the plate is deenergized and a quasi-stable state in which the relay is energized. The second important requirement for the proper operation of both circuits is that the pulse used to trigger them be of sufficient amplitude and duration.

All the relays used in the tachometer are Khurman 5C2, two-pole, two-position relays requiring 100 milliwatts per pole for energization. With a coil impedance of 10K ohms, the maximum current for energization is 4.4 milliamperes. Assuming that T_1 of figure 10a is cut off, the plate voltage will be about 150 volts. With no loading by the grid of T_2, the voltage there would be 150 - 0.5 (150 + 105) = 22 volts. Thus the tube goes into saturation. Assuming the grid impedance is now 1K ohm, the grid voltage becomes 0.3 volts. From the tube characteristics, the corresponding plate voltage is 80 volts. The divider to the grid of T_1 produces a voltage there of about -10 volts. These voltages are confirmed by measurements of the circuit. Since the cut off voltage is approximately -4 volts, the current through the relay coil when T_1 is cut off is determined by the divider to the grid of T_2. In this case the current through the coil is 250 microamperes. Thus if the divider resistor values change with time, this change would first affect the relay in the plate of the cut off tube, by causing the tube to come out of cut off. Assuming that the relay remains energized if the current stays greater than 2 milliamperes, the grid can safely come up to -3 volts since the tube is still sufficiently cut off to deenergize the relay. If the saturation voltage of T_2 increases by 10%, R_1 decreases by 5%, and R_2 increases by 5%, then the grid voltage of T_1 rises to -4 volts, which is still permissible for proper operation.

When T_1 is conducting, it should have at least 5 milliamperes

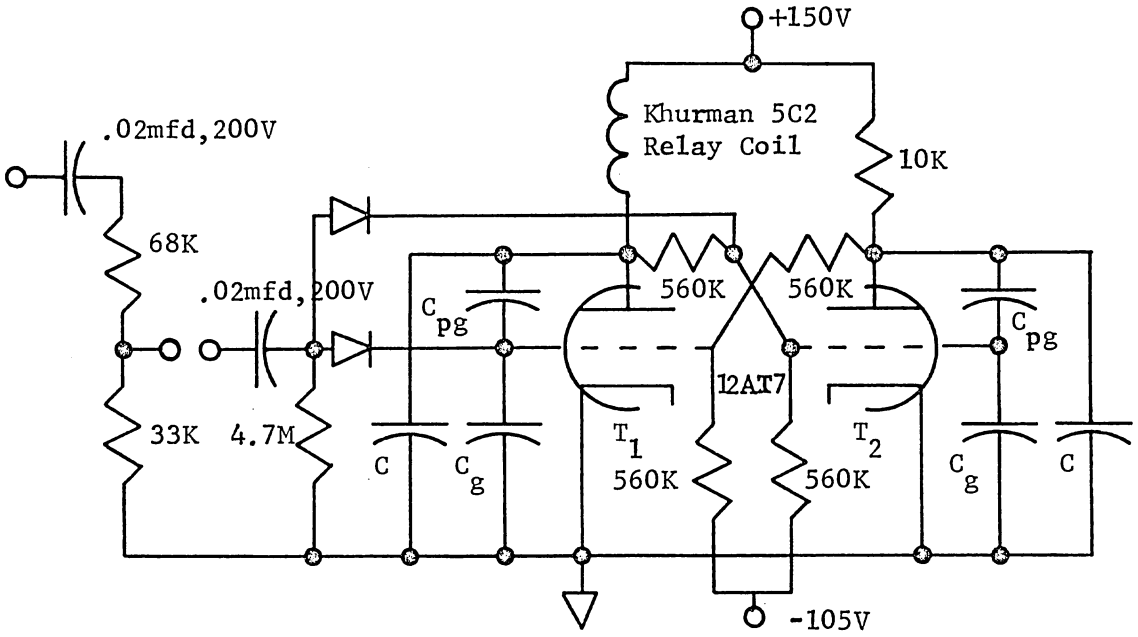


Figure 10a. Schematic of the Bistable Multivibrator

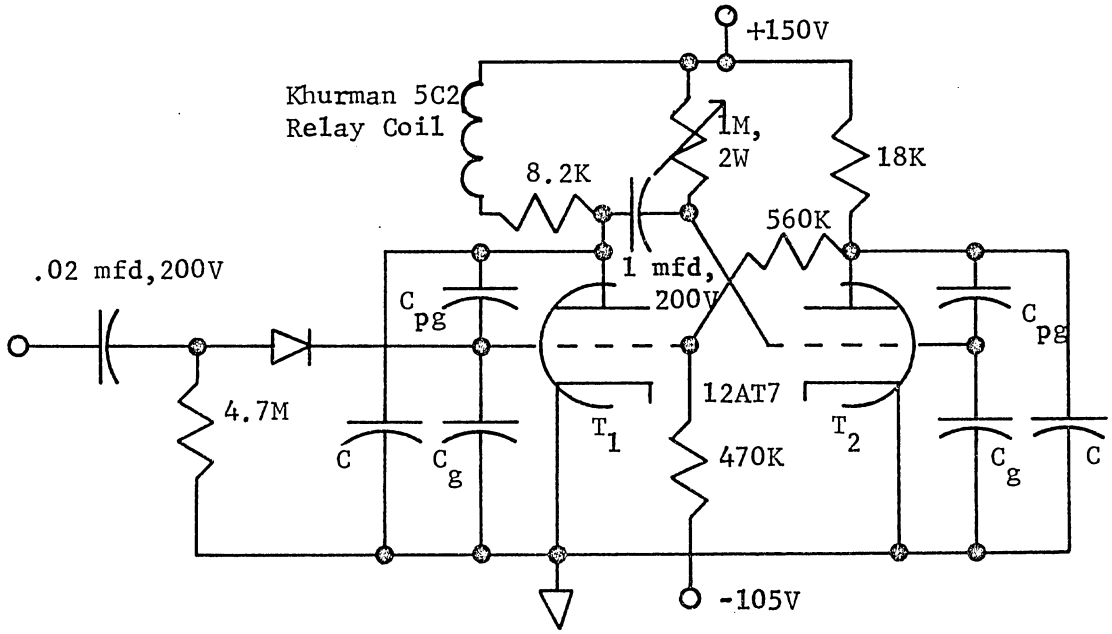


Figure 10b. Schematic of Monostable Multivibrator

All unmarked resistors, 1/2W
All diodes IN2071

SCHEMATICS OF THE CONTROL CIRCUITS ELEMENTS

FIGURE 10

in it to insure energization of the relay. The required grid voltage for this condition must be greater than -0.5 volts. If the voltage on T_2 in cut off drops 10%, R_1 increases 5%, and R_2 decreases by 5%, the voltage on the grid of T_1 with no grid current would be 9 volts. Thus the tube would still be well saturated, so that the current through the relay would change only slightly. The circuit should then be able to maintain the necessary stable states over a wide range of circuit parameters. Table 1 summarizes the above comments on the range of operation.

TABLE 1

± 10% Plate Voltage Change
± 5% Divider Resistor Change

	Off Tube	Off Tube	Off Tube	On Tube
e_p	150 v	80 v	135 v	88 v
e_g	-10 v	0.3 v	-4 v	0.25 v
I_p	250 μa	7 ma	1 ma	7 ma

Variation in the pulse used to trigger the bistable multivibrator also occur. The nature of these variations will determine whether or not the circuit operates successfully. It is well known that the multivibrator is more sensitive to a negative pulse than to positive pulse. Tests showed that a negative pulse of only 1.8 volts was sufficient to produce transition, whereas if a positive pulse was

used the amplitude had to be almost 10 volts. However, the source of the pulse is the differentiated output of the trigger section. The voltage change there is about 100 volts. Also the leading edge of the pulse is much sharper than the trailing edge. Thus it was decided to use the positive pulse in order to protect the circuit from inadvertant triggering by a small negative pulse. The nature of the pulse waveform, as well as diode coupling, prevents triggering by the trailing edge of the trigger pulse.

The positive pulse height must be almost 10 volts because until the grid of the off tube reaches about -1 volt, causing the plate to drop to about 100 volts, the on tube remains in saturation. How long must the pulse last? To get an idea of the duration needed, consider T_1 is cut off when a positive pulse is applied to its grid forcing it into saturation. Assuming that the relay coil is in the plate of T_2 gives the longest transition time. The plate voltage drops from 150 volts to 80 volts with a time constant controlled by the load and plate resistances.

$$\tau_1 = (R_1 // r_p)(C_{pg} + C) = (8k)(15\,pf) = 0.12\mu s$$

Therefore, the time required for the voltage to drop to 80 volts is approximately given by

$$t_1 = 4\tau_1 = 0.50\,\mu s.$$

The grid of T_2 follows the plate through the divider gain until the tube comes out of saturation. The grid voltage then moves toward -10 volts with a time constant governed by the parallel resistance of R_1 and R_2

and the grid capacitance enhanced by the Miller effect.

$$\tau_2 = (R_1 // R_2)(C_m) = (280k)(60\,pf) = 17\,\mu s$$

The tube is cut off when the voltage reaches -5 volts. The time for this to occur is given by solving the appropriate exponential equation.

$$t_2 = 12\,\mu s$$

Assuming the plate begins to rise at this time, it will be charged to 150 volts by the current flowing in the relay coil, which cannot change instantaneously. The change in voltage at the plate is just the saturation current divided by the plate capacitance.

$$\frac{\Delta ep}{\Delta t} = \frac{1}{C}\, i_s$$

Therefore, the time required for the voltage to rise to 150 volts is approximately given by

$$t_3 = \Delta t = \frac{C \Delta e_p}{i_s} = 0.10\,\mu s.$$

Thus the total time required for transition,

$$t_T \leq 12.6\,\mu s.$$

Tests show this time to be an accurate prediction of the pulse duration needed. The width of the pulse used is approximately 2 milliseconds. This time is set by the time constant of the differentiator of the trigger output. Thus there is no danger of the pulse width becoming too narrow, rather the pulse is likely to fail to trigger the multivibrator only if the amplitude falls below 10 volts. The drift of the differentiator component values is not so important here as the speed with which the output of the trigger changes states.

In order to find the slowest speed of transition which will provide a 10 volt output pulse consider a voltage ramp applied to the differentiator in figure 10a. If the ramp is assumed to start at zero and reach E volt at time t_o, the current through the differentiator is given by

$$i(t) = \frac{EC}{t_o}(1 - e^{-t/RC}).$$

Since i(t) is a maximum at t t_o, using an expansion to solve for t_o gives

$$t_o = 2RC\left(\frac{i(t_o)R}{E} + 1\right)$$

Substituting the appropriate values into the above equation yeilds

$$t_o \doteq 5\text{ms}.$$

The conditions which govern the speed of transition are discussed in the section on the trigger.

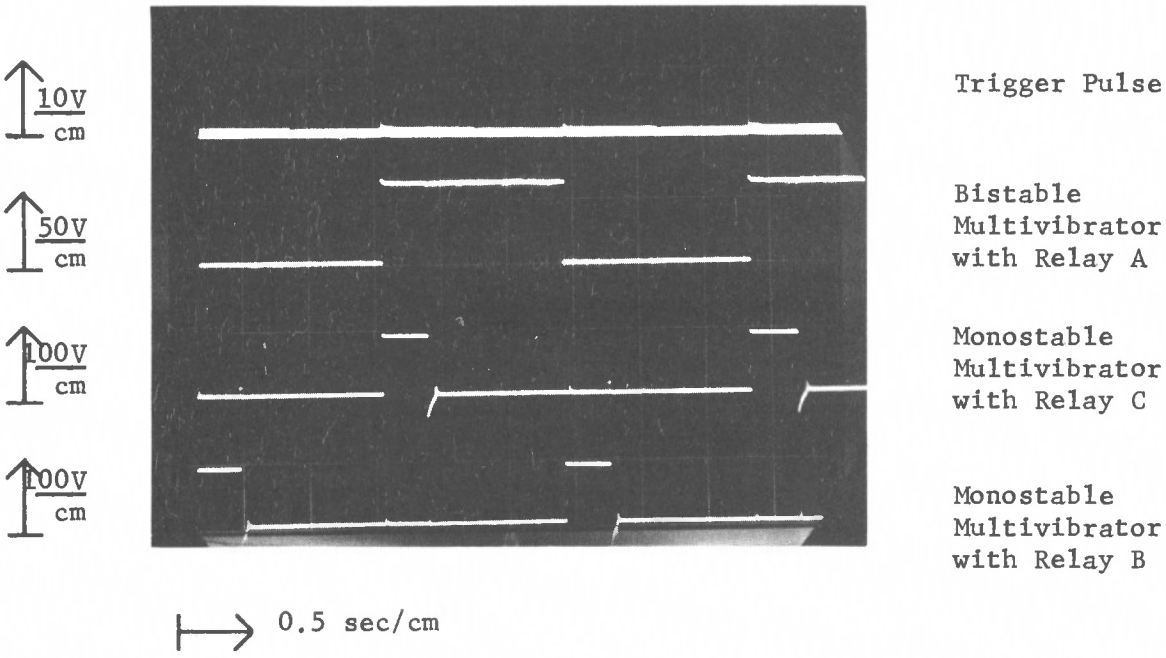
Much of the above discussion can be applied to the similar aspects of the monostable multivibrators. The same tube and relay are used, however, there is an additional 8K ohm resistor in each plate circuit. Also, since the trigger pulse is derived from the differentiation of the bistable multivibrator output, which changes about 70 volts, the tubes are further cut off, so that the trailing edge of the pulse does not cause spurious triggering. As shown in figure 10b, in the stable state T_1 is cut off and T_2 is saturated. The voltage at the grid of T_2 is slightly positive. From the tube characteristics the plate voltage should be about 60 volts. The divider to the grid of T_1 gives a voltage of 60 - 0.54(60 + 105) = -29 volts. In the quasi-stable

state T_2 is momentarily cut off. Its plate voltage is then 150 volts. The divider to the grid of T_1 would make the voltage there 150 - 0.54(150 + 105) = +13 volts. Grid loading makes the voltage 0.3 volts. Again assuming 10% voltage and 5% resistor value changes, the cut off voltage at T_1 becomes -22 volts and the saturation grid voltage would become 1 volt without grid loading. Table 2 contains a summary of the range of operation with drift.

TABLE 2

			± 10% Plate Voltage Change ± 5% Divider Resistor Change	
	T_1 Off	T_1 On	T_1 Off	T_1 On
ep	150 v	60 v	135 v	66 v
eg	-29 v	0.3 v	-22 v	0
lp	250μa	5 ma	250 μa	5 ma

Since the plate of T_1 is capacitor coupled to the grid of T_2, the pulse must only bring T_1 out of cut off. The transition time is comparable to that of the bistable multivibrator, so that the 50 volt, 2 millisecond pulse from the differentiator is quite sufficient to trigger the monostable multivibrators. Calculations of the bistable miltivibrator transition time shows that pulse shape will essentially be controlled by the differentiator. Reasonable variations in it will not alter the pulse sufficiently to prevent triggering. The relation between the important control voltages is shown in figure 11.



VOLTAGE WAVEFORMS IN THE INTEGRATOR CONTROL CIRCUIT

FIGURE 11

The Linearizer

The simplicity of the linearizing circuit is evident from the schematic of figure 7. The heart of the circuit is the charging network of three resistors and two zener diodes. The function of the amplifier is merely to invert the capacitor voltage. To understand the operation of this network, consider the events of the charging cycle. An equivalent circuit to the charging network for the first part of the cycle, assuming zero zener resistance, is shown in figure 12a. Though the largest zener voltage is 100 volts, the equivalent holds only until the capacitor voltage has reached 200 volts, because the zener breakdown is no longer maintained. To find the thevenin equivalent, consider the following equations with respect to figure 12a:

$$i_1 = \frac{V_c - V_{oc}}{R_1}$$

$$i_2 = \frac{V_c - (V_{oc} + V_{z1})}{R_2}$$

$$i_3 = \frac{V_c - (V_{oc} + V_{z2})}{R_3}$$

Since the sum of i_1, i_2, and i_3 must be zero, it is possible to solve for V_{oc_1}.

$$V_{oc_1} = V_c - \frac{V_{z1}}{(1 + \frac{R_2}{R_3} + \frac{R_2}{R_1})} - \frac{V_{z2}}{(1 + \frac{R_3}{R_2} + \frac{R_3}{R_1})}$$

The equivalent resistance of the network is just the parallel combination of the three resistors.

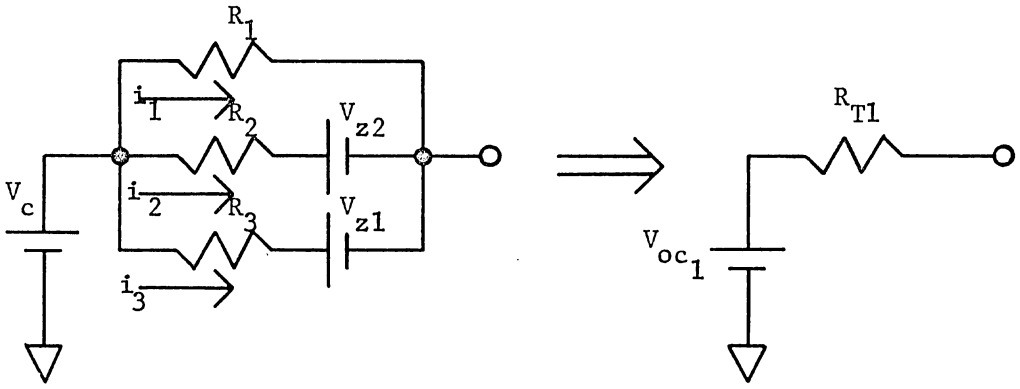


Figure 12a. Charging Equivalent: 100 to 50 Breaths/Minute

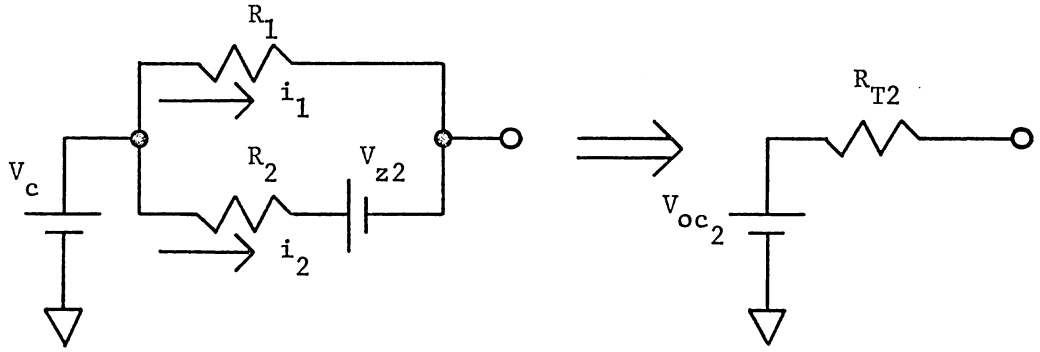


Figure 12b. Charging Equivalent: 50 to 20 Breaths/Minute

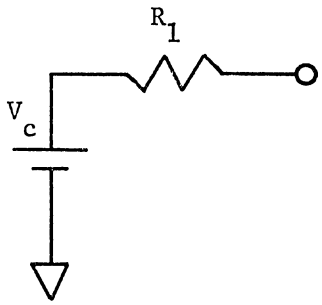


Figure 12c. Charging Equivalent: 20 to 5 Breaths/Minute

EQUIVALENT CIRCUITS OF THE LINEARIZING NETWORK

FIGURE 12

$$R_{T_1} = R_1 // R_2 // R_3$$

$$= \frac{R_1 R_2 R_3}{R_1 R_2 + R_2 R_3 + R_1 R_3}$$

When the capacitor voltage becomes 200 volts, the equivalent circuit is as shown in figure 12b. The thevenin equivalent can be found in the same manner as above. In figure 12b

$$i_1 = \frac{V_c - V_{oc2}}{R_1}$$

and

$$i_2 = \frac{V_c - (V_{oc} + V_{z2})}{R_2} \quad .$$

Setting i_1 equal to $-i_2$ and solving for V_{oc_2} yields

$$V_{oc2} = V_c - \frac{V_{z_2}}{1 + \frac{R_2}{R_1}} \quad .$$

Again the thevenin equivalent resistance is the parallel combination of the two resistances.

$$R_{T_2} = R_1 \quad R_2 = \frac{R_1 R_2}{R_1 + R_2}$$

The equivalent circuit holds until the capacitor reaches 45 volts beyond the previous breakpoint or 245 volts. Here the equivalent circuit is the thevenin circuit of figure 12c. Solving the expressions above gives the values for all the thevenin equivalent circuits.

$V_{oc_1} = 247$ volts $\quad V_{oc_2} = 269$ volts $\quad V_{oc_3} = 320$ volts

$R_{T_1} = 52.5\text{k}\,\Omega \quad R_{T_2} = 111\,\text{k}\Omega \quad R_{T_3} = 1.5\text{m}\,\Omega$

It is now a straight forward task to use these equivalent circuits to

calculate the capacitor voltage as a function of time within the various regions. A composite curve can then be made of the capacitor voltage with time.

In the first interval the charging begins when the reset cycle is complet, after 400 milliseconds. Knowledge of the boundary values permits calculation of the various exponentials from the general equation for the charging voltage.

$$v = A + Be^{-t/\tau}$$

Since the value of the capacitor is 10 microfarads, the first time constant is 0.525 seconds. Evaluating the constants of the above equation yields

$$v_1 = 247\ (1 - e^{-t/.525})$$

The time constant of the second interval is 1.11 seconds. The voltage in this interval is

$$v_2 = 269 - 69e^{-t/1.11}$$

The time constant of the third interval is 15.0 seconds. The voltage in this interval is

$$v_3 = 320 - 75e^{-t/15.0}$$

Calculating the times at which the break points occur completes the process of constructing the charging curve. Table 3 is a compliation of all the data necessary for the construction.

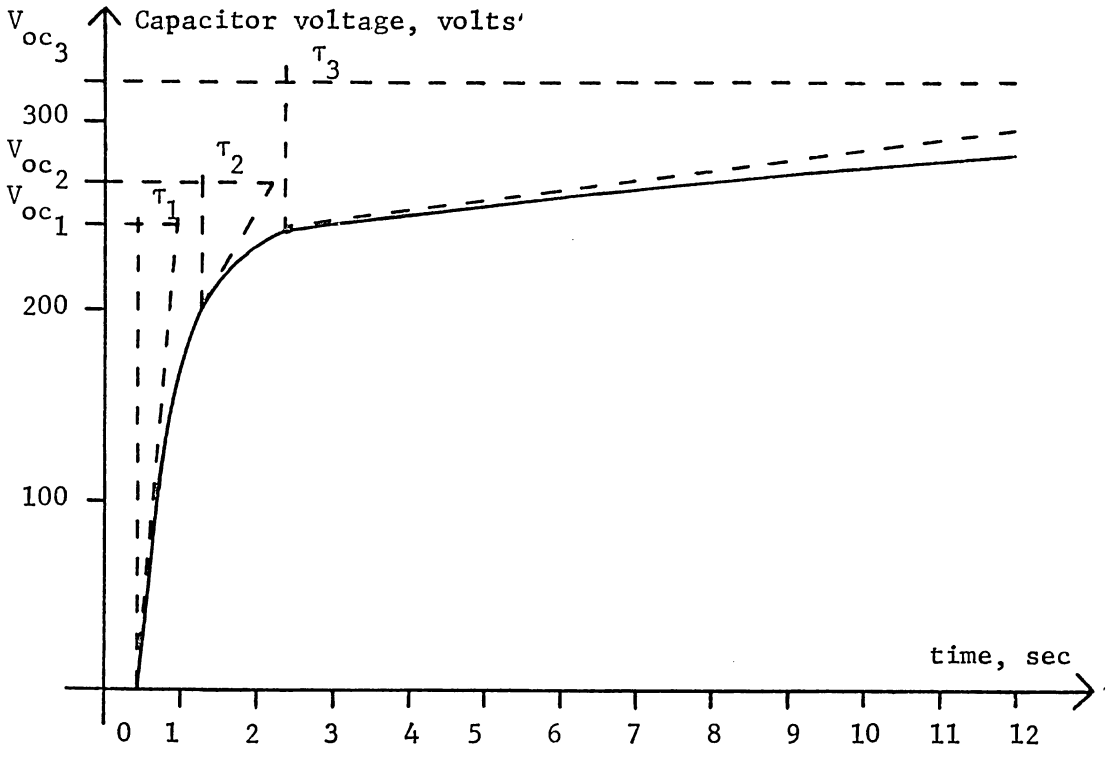


Figure 13a. Capacitor Voltage Calculated from Equivalent Circuits of the Linearizing Network

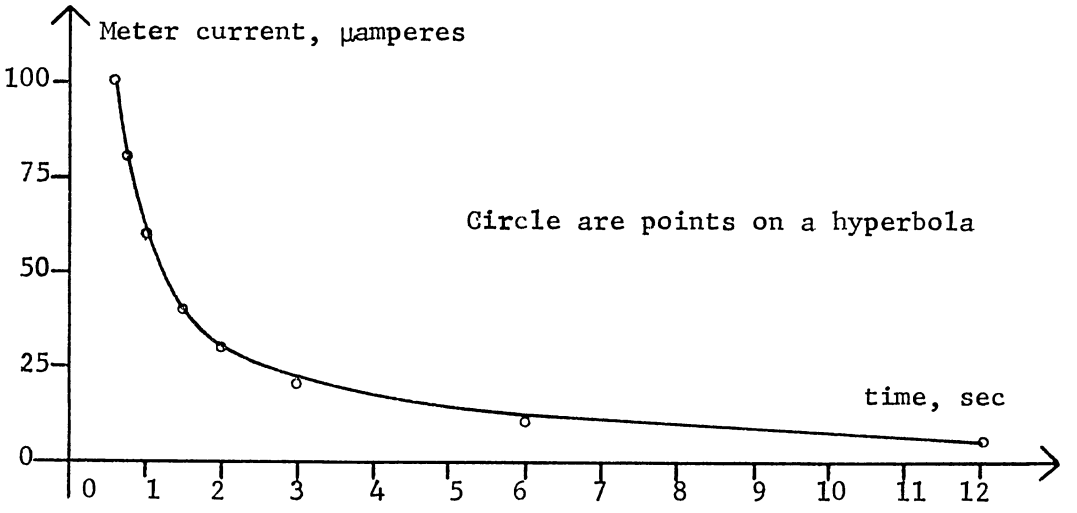


Figure 13b. Comparison of Inverted Capacitor Voltage to a Hyperbola

HYPERBOLIC SHAPE OF THE INTEGRATED VOLTAGE

FIGURE 13

TABLE 3

Voltage Interval Volts	Time Constant Sec.	Charging Voltage Volts	Time Interval Sec.	Voltage Function Volts
0 - 200	0.525	247	0.400 - 1.275	$247(1-e^{-t/.525})$
200 - 245	1.11	269	1.275 - 2.445	$269 - 69e^{-t/1.11}$
245 - 320	15.0	320	2.445 - 12.0	$320 - 75e^{-t/15.0}$

The composite curve is plotted in figure 13a. In figure 13b the curve is inverted and compared to a hyperbola which has a 20 volt change for a 10 breath/minute rate change with 5 breaths/minute as the reference point. It should be noted that most of the possible range of available capacitor voltage is utilized to provide resolution of the rate. Theoretically, the largest possible voltage change of 10 breaths/minute is 30 volts. Finding the network which would implement this utilization factor involves an iterative process since there is no unique solution, and the best solution may be considerably more complicated than the network used here. The effect of drift on the accuracy of the rate measurement is small. The only parameters involved are the charging voltage, set by a regulator, the resistor values, the zener voltages, and the capacitor value. Assuming that the resistor and capacitor values are most likely to change, a calculation was made to test the error created by a 5% increase in all the resistor values. A change in the capacitor value is effectively included since the resistor value change primarily effects the time constants. The maximum error created by this drift is only 2%.

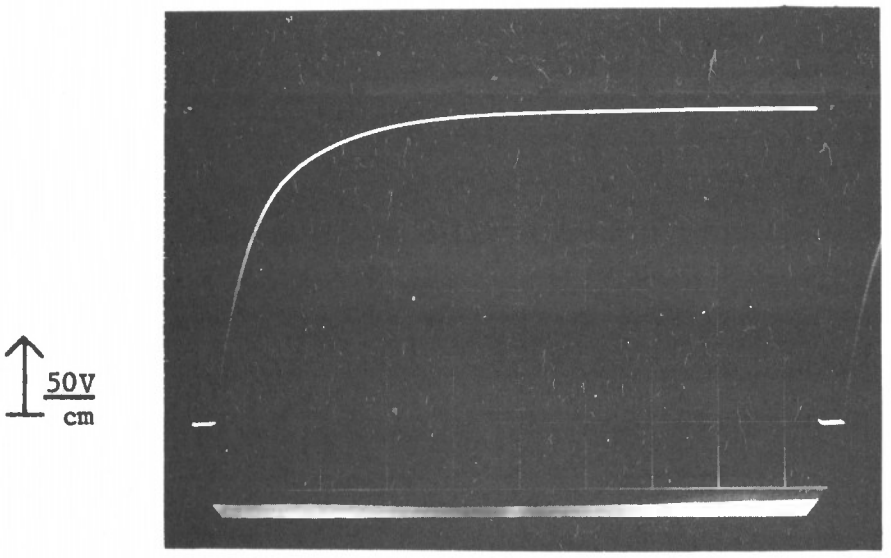


Figure 14a. Voltage on the Integrating Capacitor

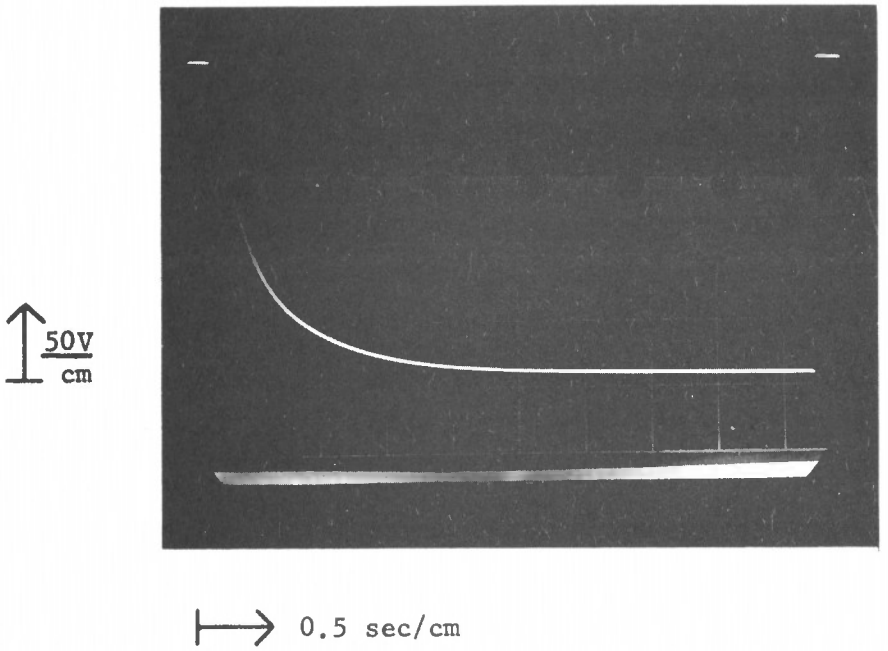


Figure 14b. Inverted Integrating Capacitor Voltage

VOLTAGE WAVEFORMS IN THE LINEARIZING CIRCUIT

FIGURE 14

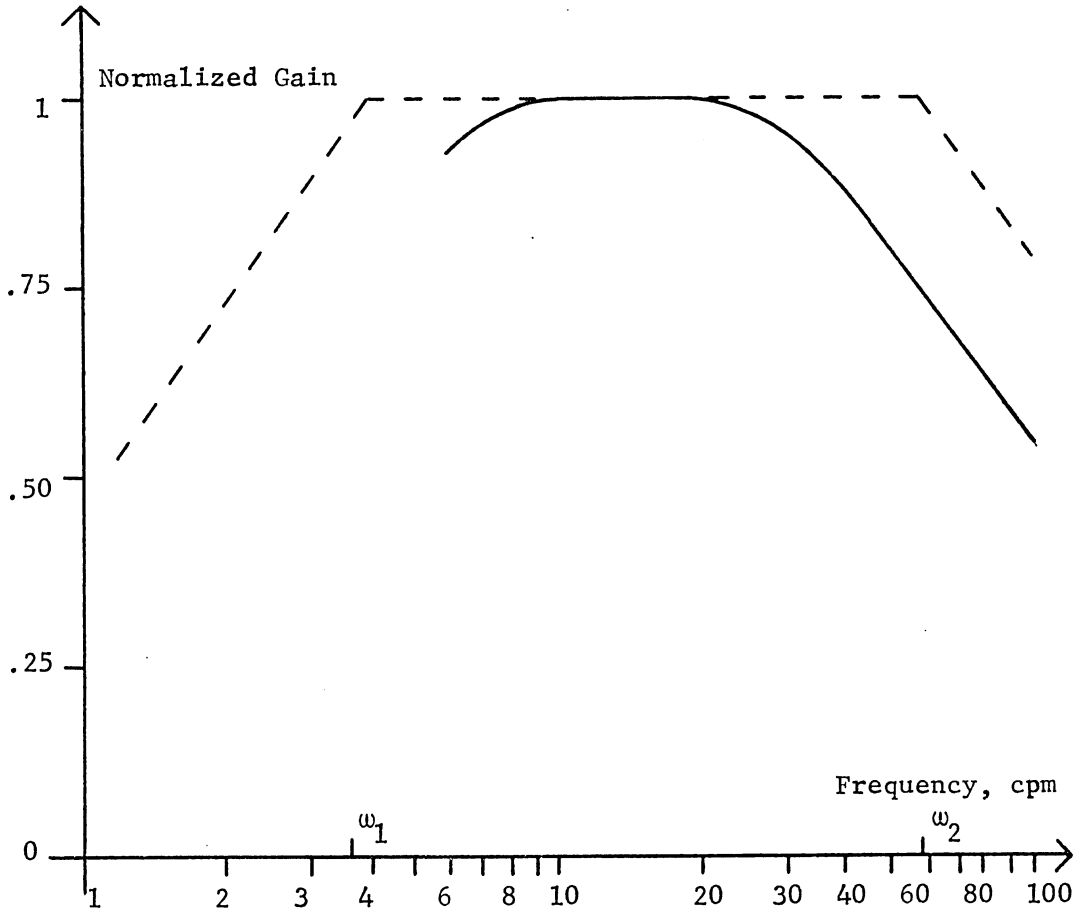
The photograph of figure 14a shows the charging of the capacitor for most of the interval used. The discrepancy in voltage between the calculated value and that shown in the picture is due to the loading of the oscilloscope probe. In reality the actual voltages agree exactly with those calculated. The photograph in figure 14b is merely the curve of figure 14a inverted. It illustrates the function of the amplifier in providing an output which is a hyperbolic function of time. At this point analysis of the display circuitry is complete. The next section deals with the analysis of the circuit which provides the display with a set of pulses.

The Trigger

As described in the development section, the trigger is preceded by a filter used primarily to attenuate the high frequencies associated with cardiac impulses. The schematic of this simple network is shown in figure 8. The transfer function of the network is given by:

$$G(s) = \frac{\frac{1}{\tau_1}\, s}{s^2 + \left(\frac{R_1C_2 + \tau_1 + \tau_2}{\tau_1 \tau_2}\right)s + \frac{1}{\tau_1 \tau_2}} \qquad \begin{aligned} \tau_1 &= R_1 C_1 \\ \tau_2 &= R_2 C_2 \end{aligned}$$

The asymptotic behavior of this function with its break points is shown in figure 15, along with the actual input-output relation for the total filter circuit. As should be obvious τ_1 controls the high frequency cut off, and τ_2 controls the low frequency cut off. The filter provides sufficient attenuation to eliminate the problem of spurious



TRANSFER CHARACTERISTIC OF THE FILTER NETWORK

FIGURE 15

cardiac triggering.

Analysis of the trigger is discussed in three parts. The first part concerns the operation of the first stage, as a function of the grid voltages. The second part concerns the nature of the grid voltage and its relation to the filter output. In the final part the operation of the Schmitt trigger is discussed, including the nature of the output pulse that is generated.

Consider the operation of the first stage of the trigger circuit shown in figure 9b as a function of its input grid voltage. This voltage ranges from approximately +10 to -10 volts. Over this range the cathode voltage remains constant as far as the plate voltage is concerned. Thus a load line for both triodes can be constructed, so that the active region performance can be determined. When the plate current of T_1, i_{p1}, is zero, the plate voltage is 400 - 330K(400/1.33M) = 301. When the plate current of T_2, i_{p2}, is zero, the plate voltage is 400 - 330K(800/2.83M) = 306. When the plate voltage of either tube is zero, assuming the cathode voltage remains zero, the plate current is 400/330K =1.2 milliamperes. Thus the load line was constructed with $e_p = 330$ volts when $i_p = 0$, and $i_p = 1.2$ ma. when $e_p = 0$. The slope implies a load of 250K ohms. Since the cathode voltage does not vary much from zero, the cathode current is 400/680K = 0.60 ma. The sum of the plate current of T_1 and T_2 must equal the cathode current, 0.60 ma.

If the input grid voltage, $e_{g1} = 10$ volts, then the full cathode

current is assumed to flow through T_1. From the plate characteristics, the plate voltage is 160 volts and the grid to cathode voltage is -1.5 volts. Thus the cathode voltage is 11.5 volts, and since the grid of T_2 is clamped to ground, eg_2 is the assumed diode drop -0.5 volts. The grid to cathode voltage is -12 volts and T_2 is indeed cut off. Output conditions remain unchanged until T_2 comes out of cut off. This occurs when the grid to cathode voltage, $e_{c2} = -4.5$ volts. Since the grid voltage is still -0.5 volts, the cathode voltage must be 4.0 volts. The grid to cathode voltage of T_1, e_{c1}, is still -1.5 volts, so that the input grid voltage is 2.5 volts. The first stage operates as a differential amplifier with gain,

$$G = \frac{\Delta e_{p2}}{\Delta e_{g1}} = \frac{\mu R_1}{2(r_p + R_1)} = 38$$

The divider to the grid of T_2 causes the voltage there to come up from -0.5 volts when the plate of T_1 is halfway through its excursion. The plate voltage at this time is approximately 230 volts. The corresponding plate current is 0.30 ma. and the grid to cathode voltage is -2.5 volts. Since i_{p1} is 0.30 ma., i_{p2} is also 0.30 ma. and e_{c2} is -2.5 volts. Thus the cathode voltage is 2.0 volts. At this point, since the grid of T_2 is free to move, regeneration takes place. The regeneration is assumed to be complete before the input voltage changes, and the gain of the cathode follower at T_2 is assumed to be 0.5 since it is loaded by the same follower output impedance due to T_1. The cathode voltage then changes by 0.5 volts while the grid of

T_2 changes by 1 volt from -0.5 to +0.5 volts. The cathode voltage becomes 2.0 volts. Since e_{c1} drops to -3.0 volts, e_{p1} comes up to 250 volts. The conditions before and after regeneration are summarized in table 4.

TABLE 4

	T1		T2
Before Regeneration	$e_{g1} = 0,5v$		$e_{g2} = -0.5v$
	$e_{p1} = 230v$	$e_k = 2.0v$	$e_{p2} = 230v$
	$i_{p1} = 0.30$ ma.		$i_{p2} = 0.30$ ma.
After Regeneration	$e_{g1} = -0.5v$		$e_{g2} = +0.5v$
	$e_{p1} = 250v$	$e_c = 2.5v$	$e_{p2} = 200v$
	$i_{p2} = 0.20$ ma.		$i_{p2} = 0.40$ ma.

Now that the grid of T_2 is clamped the cathode follows the input grid with a gain of about 0.5. The amplifier action continues until T_1 becomes cut off. This occurs when $e_{c1} = -4.5$ volts, but e_{c2} cannot become greater than -1.5 volts, so that $e_k = 2.0$ volts. Thus T_1 is cut off when the input grid voltage, $e_{g1} = -2.5$ volts. The output conditions of the first stage remain unchanged as e_{g1} goes to -10 volts. As e_{g1} goes positive, T_1 comes out of cut off at $e_{g1} = -2.5$ volts, but regeneration does not take place until $e_{g1} = 0.5$ volts. The output plate changes from 230 to 250 volts. T_2 is fully cut off when e_{g1} reaches 2.5 volts. Thus the cycle is complete and the operation described for the full range of input voltages. Of particular interest is the action during the regeneration phase, especially the regeneration time.

The model of the circuit during regeneration is shown in figure 16. Here A is the gain of each triode, R is its output impedance, and C is the series combination of the compensating capacitor and the input capacity of the grid of T_2 accentuated by the Miller effect. Applying the Kirchnoff voltage laws yields

$$Ri_1 + e_1 = -Ae_2$$

$$RC \frac{de_1}{dt} + e_1 = -Ae_2$$

Since the gain from grid to cathode of T_2 is 0.5,

$$e_2 = 0.5e_1$$

Substituting this relation into the above equation yields

$$RC \frac{de_1}{dt} + e_1 = -\frac{A}{2} e_1$$

$$RC \frac{de_1}{dt} = -(\frac{A}{2} + 1) e_1$$

A/2 is much greater than 1, so that

$$RC \frac{de_1}{dt} \doteq -\frac{A}{2} e_1,$$

and

$$\frac{de_1}{e_1} \doteq -\frac{A}{2RC} dt.$$

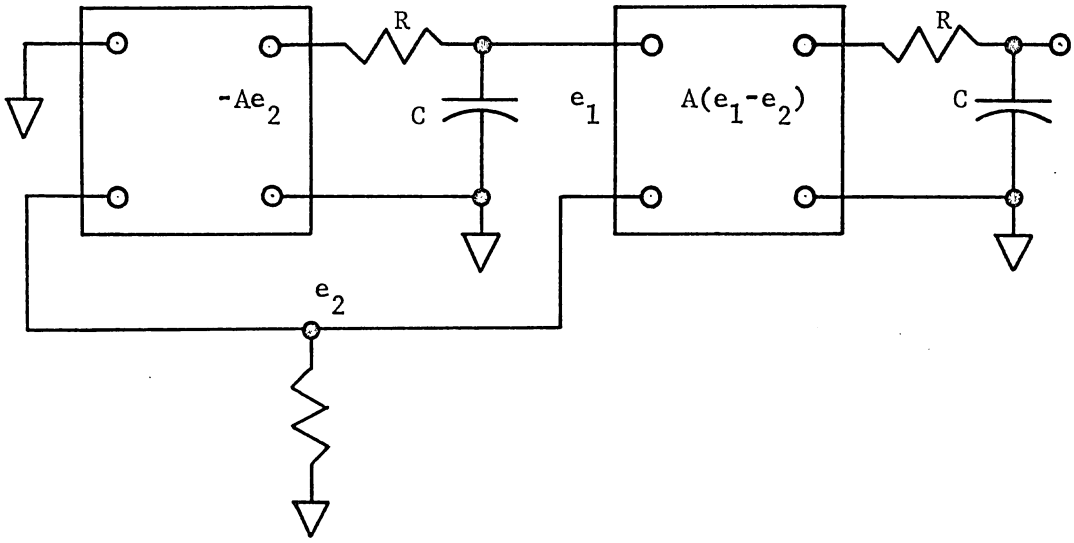
Integrating gives

$$\ln e_1 \doteq \frac{-At}{2RC} + k,$$

or

$$e_1 \doteq E\varepsilon - \frac{A}{2RC} t.$$

The change at the grid of T_2 is limited to 1 volts due to the clamping action of the diodes. To get an idea of the regeneration time, assume



MODEL OF THE SCHMITT TRIGGER DURING REGENERATION

FIGURE 16

the action is initiated by a 0.1 volt change at the plate of T_1. Thus

$$\epsilon - \frac{A}{2RC} t \doteq 10$$

Simplifying gives

$$-\frac{A}{2RC} t \doteq 2.3,$$

or

$$t \doteq \frac{4.6RC}{A}$$

Substituting the appropriate values into the above equation yields

$$t \doteq 1.5 \mu s.$$

Thus the regeneration is completed extremely rapidly so that there is no chance of noise at the input causing a spurious trigger pulse at the output.

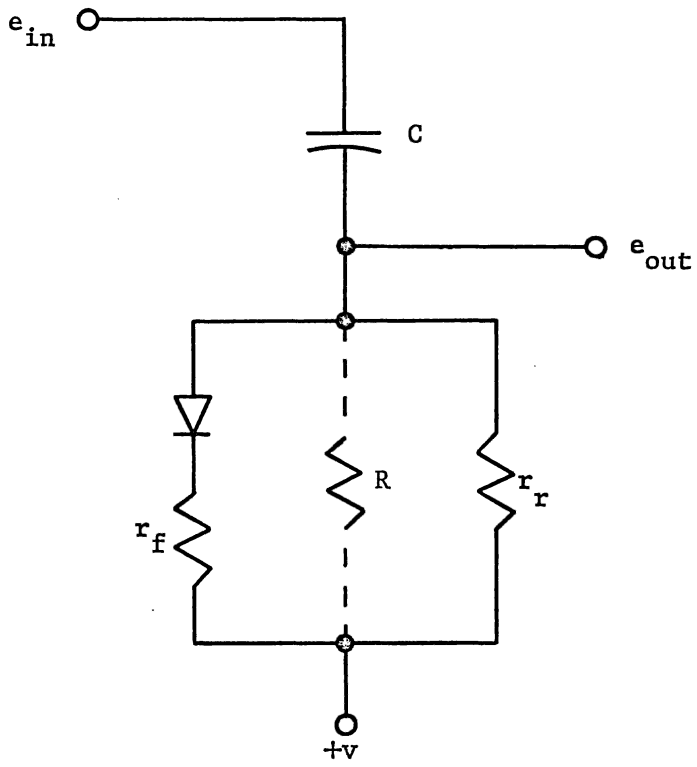
The next phase of the trigger operation which must be investigated is the manner in which the input network sets the threshold at which the triggering is to be done. Figure 17 shows the equivalent for the positive peaks in the input voltage. The diode shown is ideal. Its forward resistance is represented by r_f; its reverse resistance, by r_r. R stands for whichever is appropriate. Assuming a sinusoidal input, $E \sin \omega t$, the magnitude of the grid voltage is

$$eg_1(t) = v + \omega E / \sqrt{\omega^2 + 1/RC} \sin \omega t$$

In the forward biased condition R is r_f, which is very small. Here the grid voltage becomes

$$eg_1(t) = v$$

In the reverse biased condition R is r_r, which is very large. In this case the grid voltage becomes



TRIGGER INPUT NETWORK FOR POSITIVE PEAKS IN THE INPUT NETWORK

FIGURE 17

$$eg_1(t) = v + E \sin \omega t$$

The final point of interest is where these representations of the grid voltage hold. A e_{in} increases from zero with the capacitor originally zero, the grid voltage follows the input until the voltage reaches v volts. Here the diode conducts and the grid voltage remains v volts until the input voltage reaches a peak. At this point the change in voltage across the capacitor is zero, so that the current through it is also zero and the diode turns off. The grid voltage again follows the input. The analysis of the network is similar for the negative peaks in the input voltage. Thus, since it has been shown that regeneration takes place in the neighborhood of a grid voltage of zero, the voltage $\pm$ v is the threshold voltage. It is the amount which the input voltage must move from a maximum or minimum in order for the transition in the output to take place. Gain relations from input to output of the trigger shown in figure 18 demonstrate the relation between input level and threshold.

When regeneration occurs in the first stage, the output moves the input grid of the second stage through the voltages required to incite regeneration. Regeneration occurs with the input grid at $\pm$4 volts, depending upon which way the voltage is moving. The transition changes the state of each triode either from cut off to conducting the full cathode current of 5 ma. or vice versa. The corresponding plate voltages are 400 and 300 volts. The time of transition can be found in a manner similar to that used for the first stage. However, the

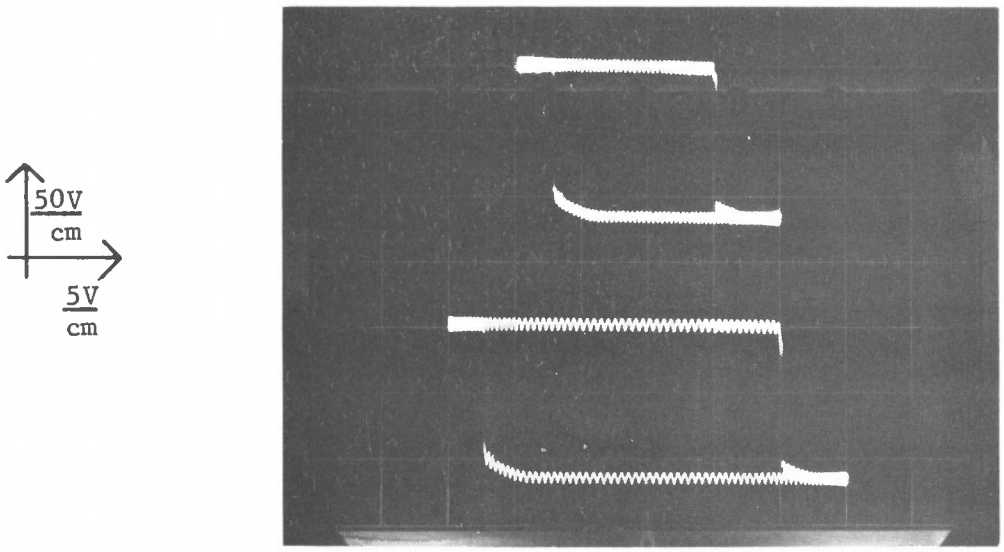


Figure 18a. Threshold Held Constant with a Change in Input Level

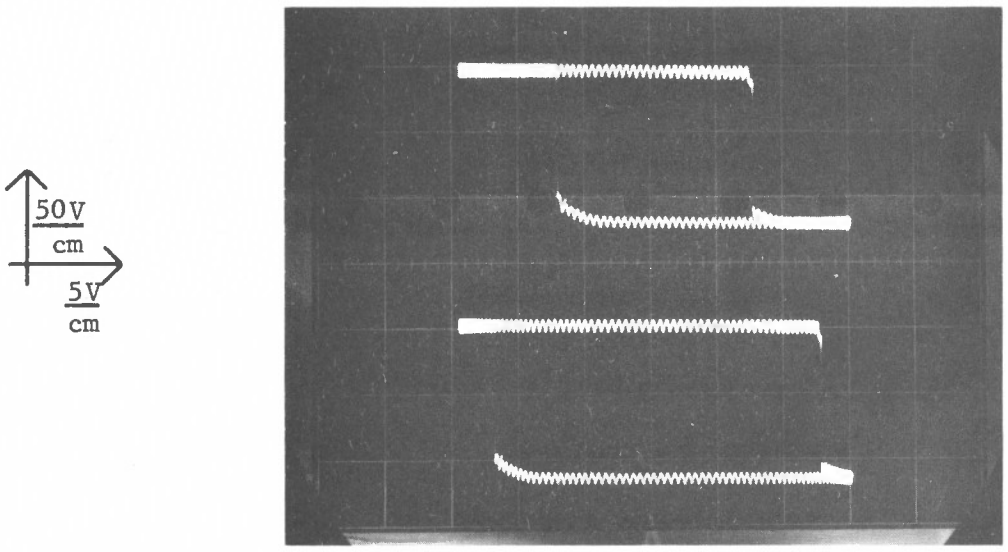


Figure 18b. Input Level Held Constant with a Change in Threshold

GAIN OF THE TRIGGER CIRCUIT

FIGURE 18

gain here is given by

$$A = \frac{-\mu R_1}{r_p + R_1} = -30.$$

The voltage at the grid of T_4 is

$$e_1 = E\,\epsilon^{-\frac{A}{2RC}t}.$$

A 16 volt change at the grid of T_4 causes an 8 volt change at the chatode, which is enough to either turn T_3 on or off. The change can be initiated by a 1 volt change. Thus

$$\epsilon^{\frac{-At}{2RC}} = 16,$$

$$-\frac{At}{2RC} = 2.8,$$

and

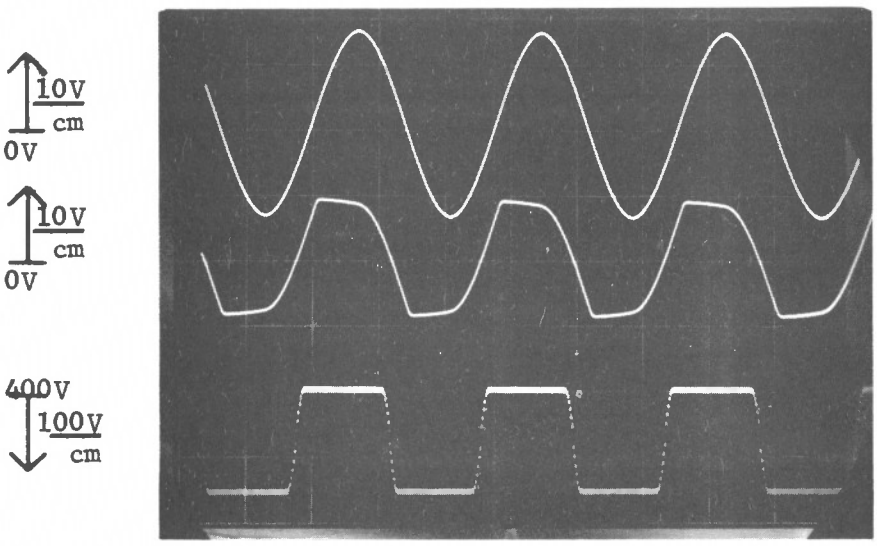
$$t = \frac{-5.6RC}{A}.$$

Substituting the appropriate values into the above equation yields

$$t = 0.4 \text{ ms}$$

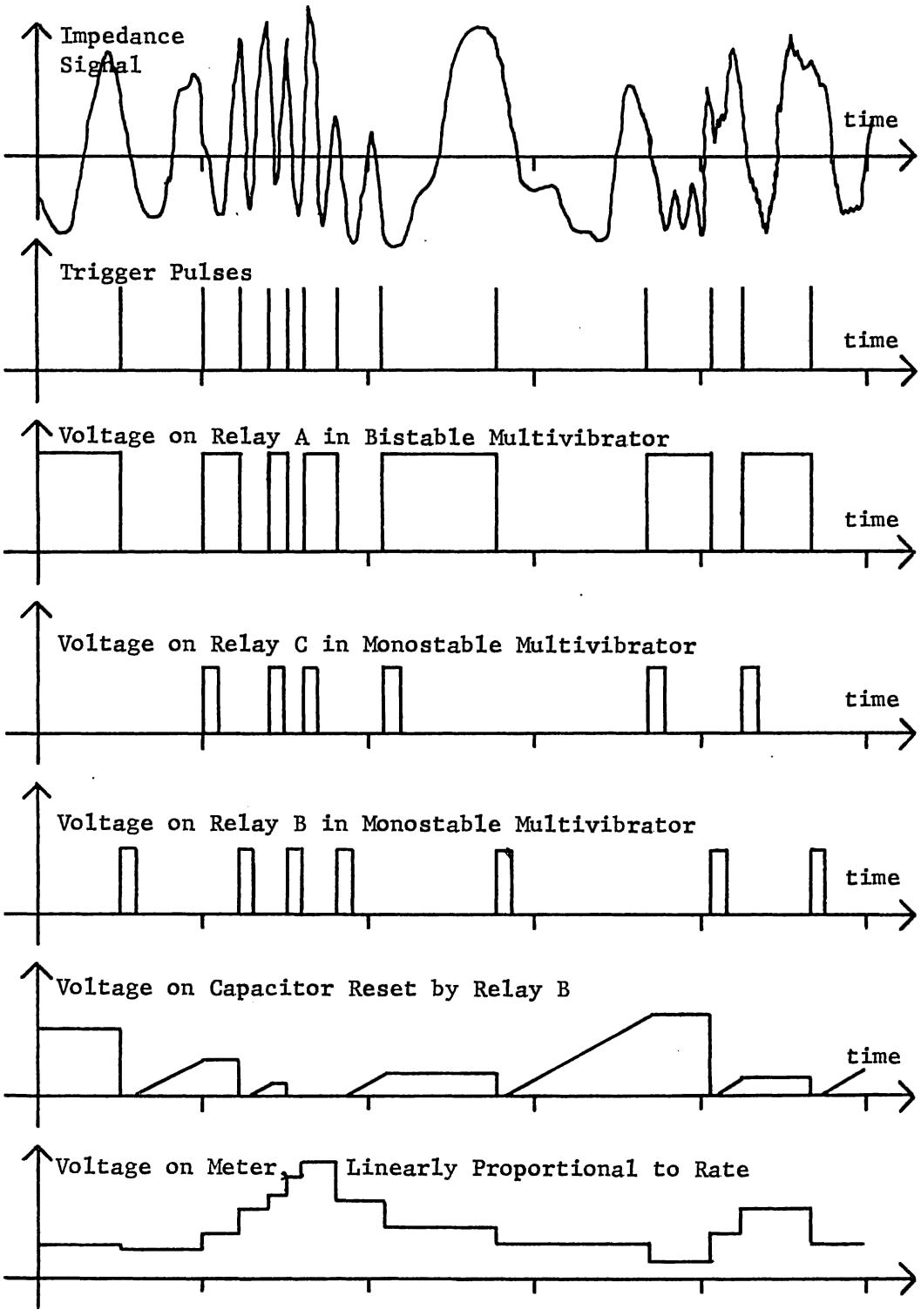
This time agrees well with the observed time of 0.5 milliseconds and is one-tenth of the maximum time permitted for transition, in order for the pulse derived from differentiation of the output to trigger the bistable multivibrator, as calculated in the integrator section. The relation of all the voltages in the trigger, including the transition regions, is shown in figure 19.

As a summary of the circuit operation, the important circuit voltages and their time relationships are shown in figure 20.



VOLTAGE WAVEFORMS IN THE TRIGGER CIRCUIT

FIGURE 19



VOLTAGE WAVEFORMS THOUGHOUT THE TACHOMETER

FIGURE 20

RESULTS

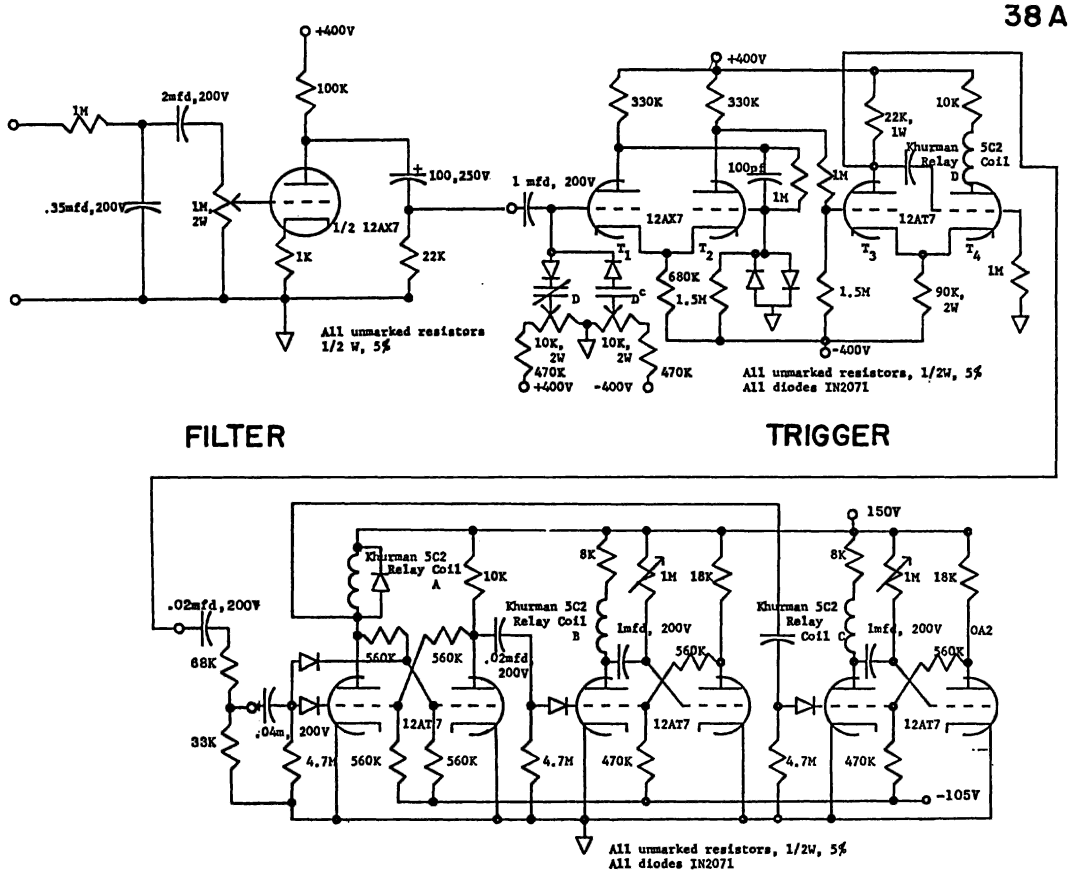
The result of the development work is a device which measures the respiratory rate instantaneously from 5 to 100 breaths per minute. The schematic of this device is depicted in figure 21. The rate is displayed with less than an error of 5% or 1 breath per minute, whichever is larger, both as a meter reading and as a linear output voltage which may be recorded by a Physiograph channel. The output voltage is held with a maximum drift of one breath per minute. Table 1 contains a summary of the pertinent aspects of its operation.

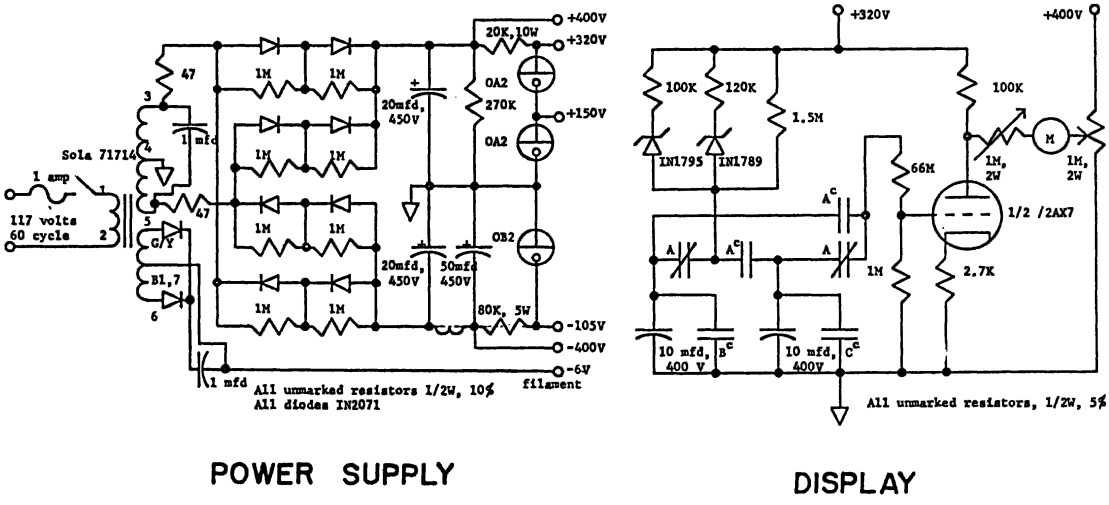
TABLE 1

Range	5-100 Breaths/Minute
Error	Max {5%, 1 Breath/Minute}
Drift	1 Breath/Minute

Test records made on the Physiograph, showing the impedance pneumograph signal, the generated trigger pulse and the calibrated linear rate output were made with the setup pictured in figure 22. The impedance pheumograph signal has the differentiated trigger output added to it for test purposes, so the exact point of triggering can be determined. The records presented in figure 23 show tests for minimum, maximum, and what may be optimum threshold setting. The minimum breath to be counted is usually of the order of 150 milliliters. As described in the introduction, the proportional impedance change depends upon the subject, so that each case is

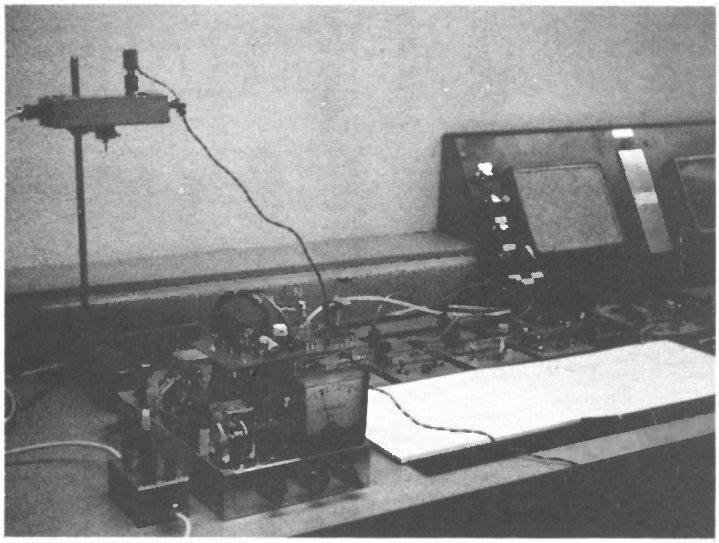
different. However, an appropriate relation the threshold levels may be set and the change in the impedance pneumograph signal which causes triggering then determined by the input level to the trigger.





SCHEMATIC OF THE TACHOMETER

FIGURE 21



PHOTOGRAPH OF THE TEST SETUP

FIGURE 22

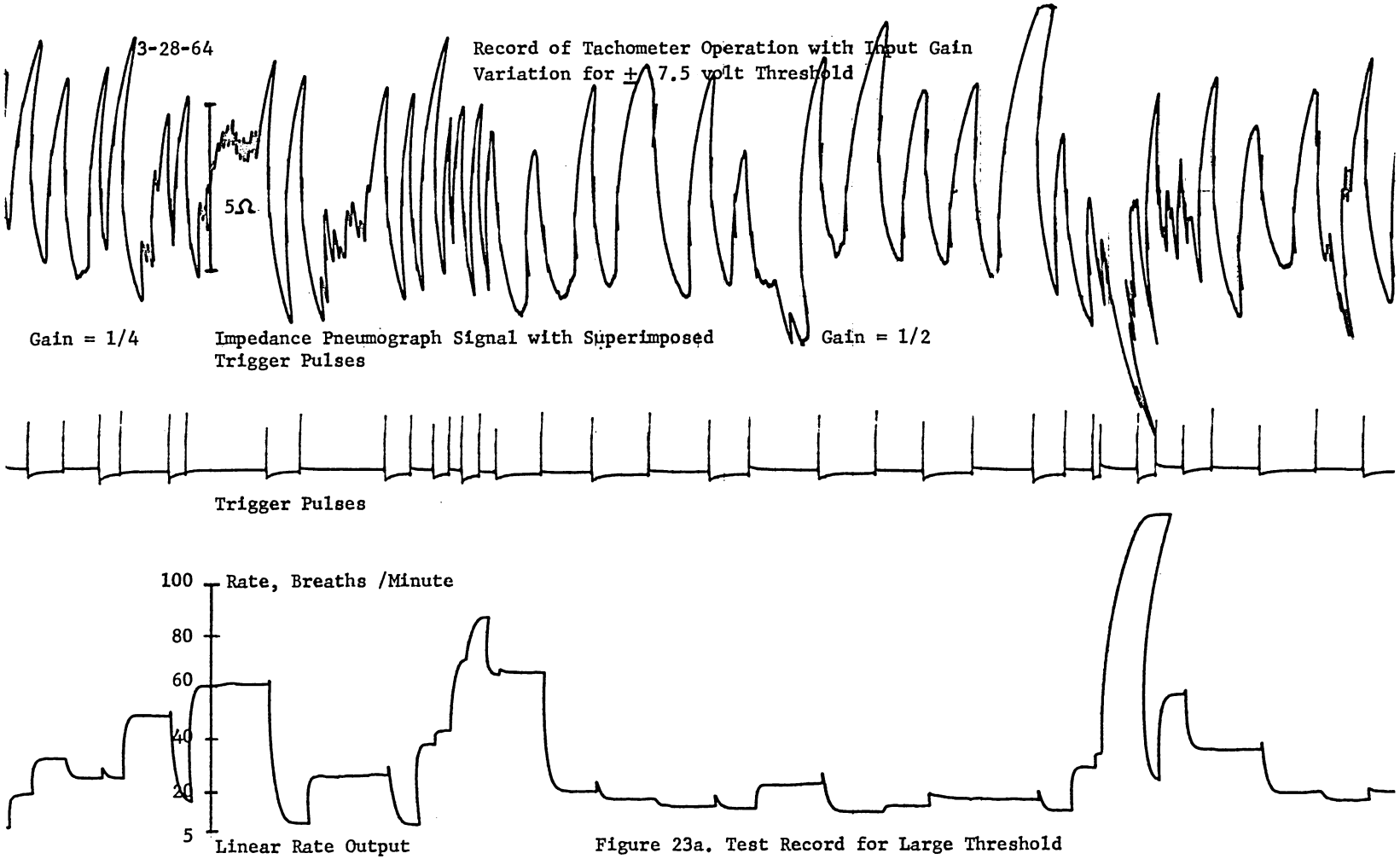


Figure 23a. Test Record for Large Threshold

FIGURE 23

5 sec/div

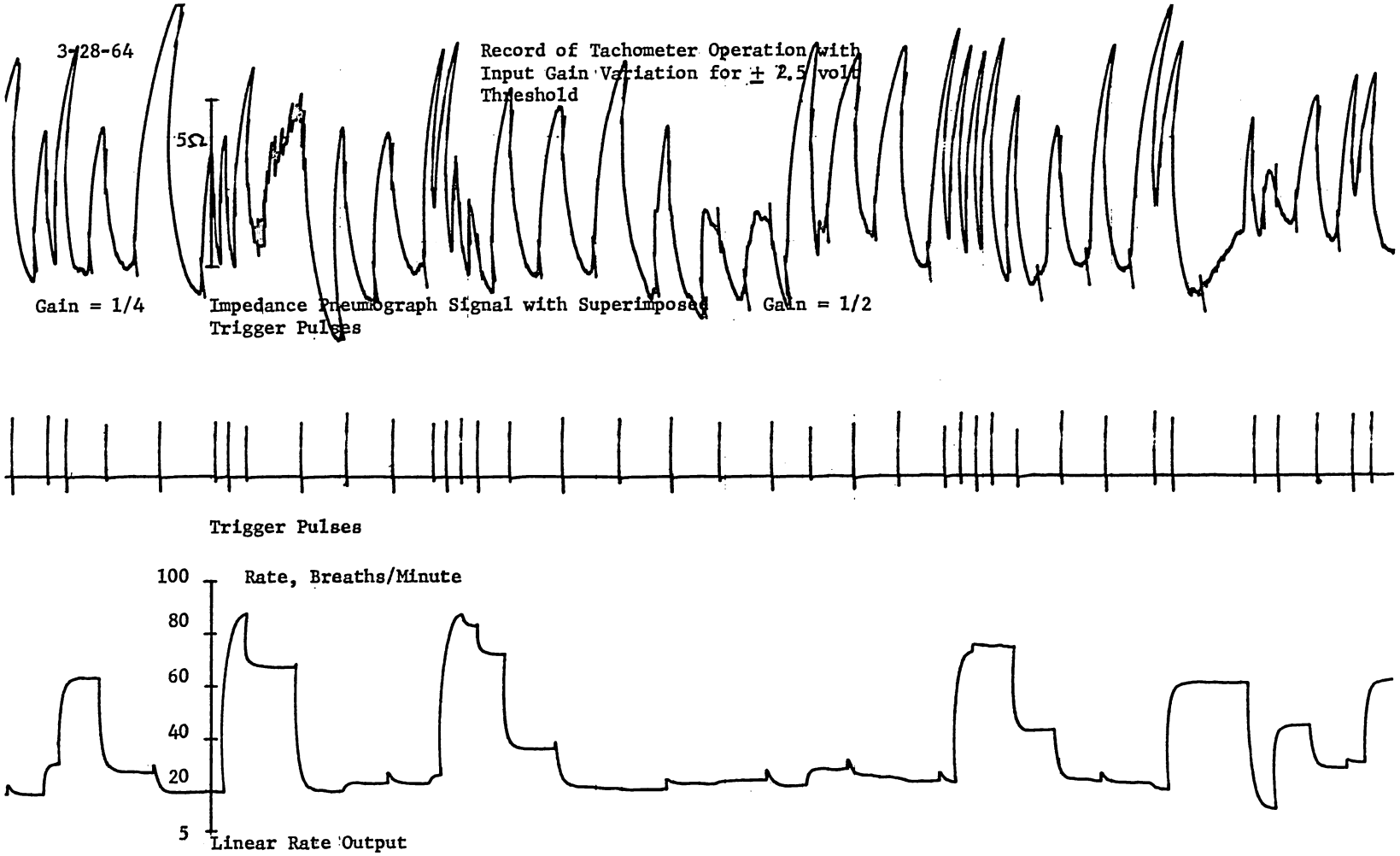
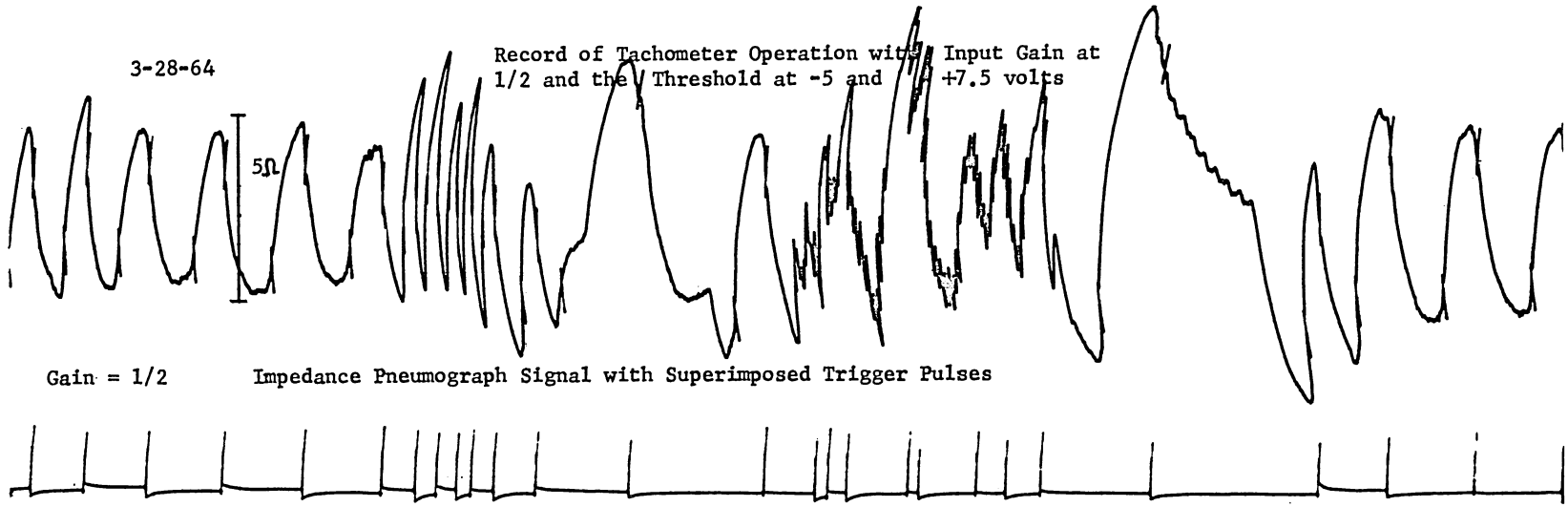


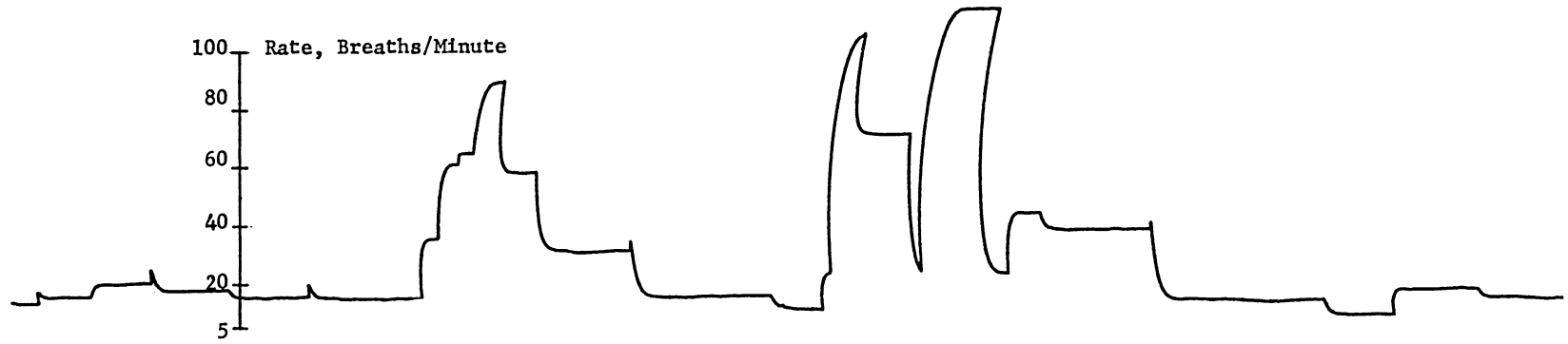
Figure 23b. Test Record for Small Threshold

5 sec/div

FIGURE 23



Trigger Pulses



Linear Rate Output

Figure 23c. Test Record for Asymmetric Threshold

FIGURE 23

5 sec/div

CONCLUSIONS

Rather than use two integrators, it is possible to do the rate measurement using one integrator and a storage device. At the end of the charging cycle, the integrating capacitor is connected to a storage capacitor. The integrating capacitor is disconnected and reset. The stored voltage is then displayed. Though adding complication to the circuit by requiring isolation of the integrating and storage capacitors, this method eliminates the problem of matching two integrators and the problem of the switching transients. At present the rate determined by one breath is displayed for the length of time that the next breath takes. A worthwhile addition to the tachometer may be an additional pair of integrators or one integrator with storage to enable the rate to be displayed for the time that the breath which determined it lasted.

The next step in taking the circuit beyond the bread-board stage will be to place the components on a standard Physiograph plug-in chassis with the meter on the front panel. The experimenter will have the option of displaying the linear output on a data channel. In the plug-in form the device may be used in the patient monitor system. Here it may save the attending personnel the time of taking routine data. This data may be put to further use in the study of the relation of the instantaneous respiration rate to other physiological phenomen, especially the instantaneous heart rate, as mentioned in the introduction. Also, with proper computer handling, the recorded

rate data may be a valuable aid in diagnostic procedures.

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[5] See the unpublished paper (Baylor, 1963) by L. E. Baker et al., "Impedance Spirometry," pp. 1-7.

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