A PORTABLE
TAPE RECORDER
FOR MONITORING AMBULATORY SUBJECTS

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by

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Abstract

This thesis describes the design and construction of a portable 4 channel magnetic tape recorder with an endless loop tape storage of 65 minutes. Although primarily designed for recording electrocardiograms the recorder is capable of recording signals with frequencies up to 200 cycles/second. The total recorder weight is 3.0 lbs plus batteries weighing .1 lbs per hour of recording. Although the recorder uses frequency modulation, delta modulation is discussed as an alternative to frequencey modulation.

A system that has been built for transmitting electrocardiograms via telephone is also described.

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1. INTRODUCTION

1.1 General

1.1.1 History of Cardiography*

In 1843 Matteucci proved for the first time that there is an electrometer effect in cardiac muscle. The interpretation of this phenomenon was later given by Englemann in 1877. Koelliker and Müller in 1856 showed the existence of action currents by means of a current sensitive frog muscle. Their results established that an electrical impulse precedes the mechanical expression of the cardiac cycle.

Following this major discovery, activities were concentrated on developing more sensitive galvonometers. This led to the tracing of the electrical phenomena during the cardiac cycle and to a human electrocardiogram obtained from leads placed externally on the body.

In 1920, after Herrick had prepared the way clinically, and Smith experimentally, Pardee described the typical electrocardiogram of infraction of the heart and its end effects. This was the starting point of a new way of regarding electrocardiography, and after a latent period of a few years it was concerning itself with damage to, and alteration of, the heart muscle. The great diagnostic importance of electrocardiography in disease of the coronary vessels aroused increased interest and drew even more attention to this method of investigation.

* Summarized from reference (1).

In the mid-twenties very promising electrocardiographs, based on the principle of the vacuum tube amplifier, were manufactured in Germany and America. They amplified the changes in the voltage arising from the heart's action currents and translated them into current changes which could be easily recorded by means of an oscillograph, without requiring a fragile galvonometer. This change from recording the action current to recording the action potential led to a great technical simplification of the method and made it less expensive. It was then possible to use electrocardiography as a method of general medical practice, whereas formerly it had only been feasible in research laboratories and the larger hospitals.

Recently the construction of light weight transportable amplifiers has advanced its effective range to almost every doctor's office and made it indispensable to a specialist in cardiography.

1.1.2 Present Problems in Electrocardiography

One of the problems that faces a cardiologist of today when he is diagnosing a heart condition is being able to observe the activity of the heart when the patient is experiencing a heart attack. The electrocardiogram may show no abnormality prior to and after the attack. Faced with this situation the diagnosing physician is left with very little information as to the probable cause of the heart pain that the patient has experienced. The solution to a situation such as this is to continuously record the electrocardiogram and examine that portion of the record where the patient had felt that an attack was occurring. However, the people that experience this type of symptom are quite capable of carrying out a normal day's activity, and it is impractical to immobilize the person and continuously record his ECG by conventional methods. Also, in order that the results obtained are not affected by the methods used to record the ECG, it is imperative that the links used to connect the patient to the monitoring device do not impair normal activity.

The freedom of movement afforded by a radio frequency link has led to the development of radio telemetry systems (Buckley 2, Spacelabs Co., 3)* that are used to monitor ECG's. One use of such a system is to monitor the ECG of a patient that is suffering from a known cardiac ailment. In this application, where the patient is confined to one or two rooms, the radio telemetry system is a very effective tool, as the monitored ECG serves to warn of an impending attack. However, for those patients that are not hospitalized but are suspected of having a cardiac ailment because of periodic heart pains, radio telemetry is not applicable as their range of movement is too great for small radio transmitters.

The patient could, however, carry a radio transmitter with him that would send the signal to a nearby magnetic tape recorder. If the patient is to carry two packages (one on his

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* Bracketed references in section 5.

body and the other which must remain within the range of the transmitter) then his activity might be impaired. In order to give the patient a maximum of freedom with a minimum of effort, it is evident that a magnetic tape recorder that could easily be carried would be the most desirable.

A commercial magnetic tape system (The Avionics Dynamic Electrocardiographic System) has been developed (Holter, 4) which will record an ECG for a 10 hour period. After 10 hours the supply reel must be changed. This system, which makes very efficient use of the magnetic tape, does so at the expense of an elaborate playback arrangement. The information is written on the tape at a speed of $1\frac{7}{8}$ inches/minute. To get the ECG back from the magnetic tape, it is run at a speed of $1\frac{7}{8}$ inches/second and displayed on an oscilloscope. The information is thus viewed sixty times faster than it is recorded. While this is an excellent method of quickly spotting changes in an ECG, the time scale must be changed to its original value in order to record the output from the magnetic tape on a conventional ECG pen recorder. Consequently, two magnetic tape decks are used in the playback system when a permanent pen recording is desired.

As well as the complexity of the playback arrangement, the recorder is also limited to one channel, but to do an accurate diagnosis a physician requires more than one ECG.

The vectorcardiogram which requires 3 independent ECG's is a method of obtaining the greatest amount of information with the minimum number of channels.

The literature also describes another portable magnetic tape ECG recorder that has been developed in Norway (Hirschborg, 22). The paper describing this recorder gives no details on the tape deck other than to say that its tape storage is limited to ½ hour. The recorder does, however, play back the information in real time, but it also is limited to one channel.

Small recorders intended for use in space vehicles are also reported but these recorders which are tailor-made for a specific job are not commercially available.

Since a small recorder with the required number of channels was not available, an investigation was started which resulted in the design and construction of a lightweight magnetic tape recorder that would continuously record an ECG.

To cope with the unmanageable record length that would result if an ECG were recorded for days, an endless loop of magnetic tape was used to store the pertinent ECG signals. This method was used as the data that has been recorded some time prior to a cardiac event is of little value, and it is thus advantageous to erase the oldest information just prior to recording the newest. In this manner, using one hour of tape and stopping the unit one-half hour after the event, a record consisting of one-half hour prior to the event and one-half hour following the event will be preserved. The reel of tape may then be removed and examined whenever convenient.

Using this basic approach to the problem of recording the cardiac event, a recorder was designed that would inconvenience the user as little as possible.

Of prime concern in the design and construction of the recorder was the size and weight of the instrument that the patient must carry with him. Also, of almost equal importance was the ruggedness of the recorder as it must withstand all the forces present on the body during normal activity. Since it was desirable to be able to record a vectorcardiogram which requires three independent ECG's, the instrument was also designed with the view of adding additional channels once the recorder had been proven feasible.

This thesis summarizes the criteria that were used in the design of the recorder and lists the goals that were achieved when a working model had become a reality. After a description of the system that was finally used, the thesis is divided into two main parts; The first part deals with the mechanical consideration and the second deals with the electronic portion of the recorder. When dealing with the electronics a number of techniques for writing on magnetic tape are considered. Delta modulation, because of its apparent advantages is considered in detail.

In Appendix A of this thesis there is a description of a system developed by the author for transmitting ECG's over normal telephone circuits. This system enables a physician in a remote area to quickly consult a specialist on the interpretation of an ECG.

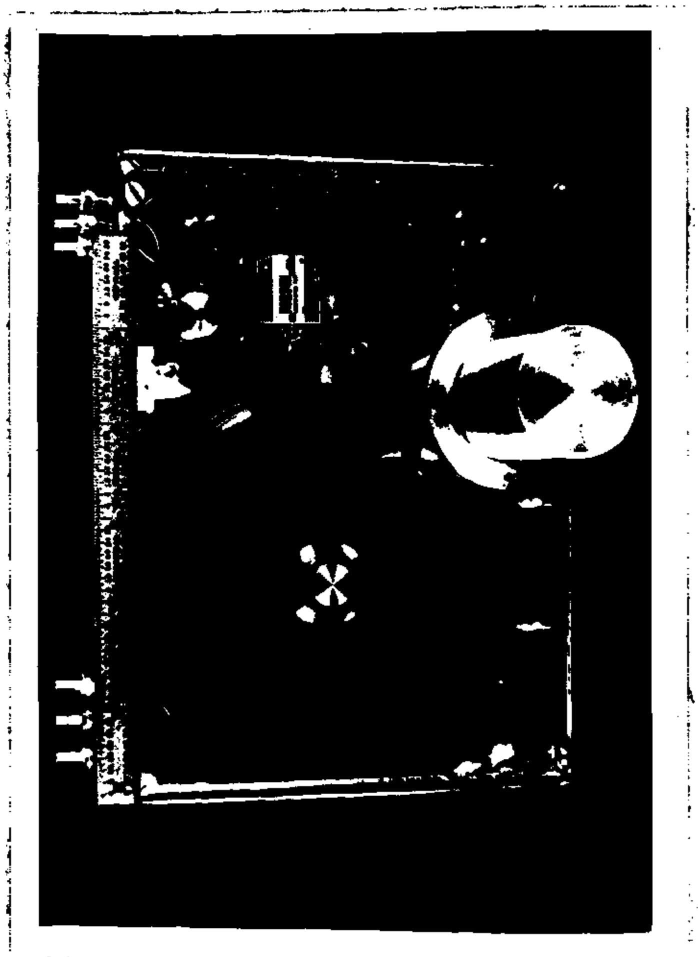

Fig. 1 Magnetic Tape Recorder

1.2 System Description

The recorder that has been developed for monitoring ECG's is contained in the package that is shown in Fig. 1. The total weight including batteries for 5 hours of operating is 3.5 lbs. Contained in the package is the electronic circuitry that is used to convert the ECG signal to a form that is suitable for recording, the tape transport which moves the magnetic tape past the recording station and the magnetic tape that stores the ECG. The reproduce electronics which are used to read the signal from the tape and convert it to a signal that is suitable for examination is contained in a separate package.

The tape transport which must move the tape past the magnetic head at a predetermined speed may be divided into three parts, as shown in Fig. 2.

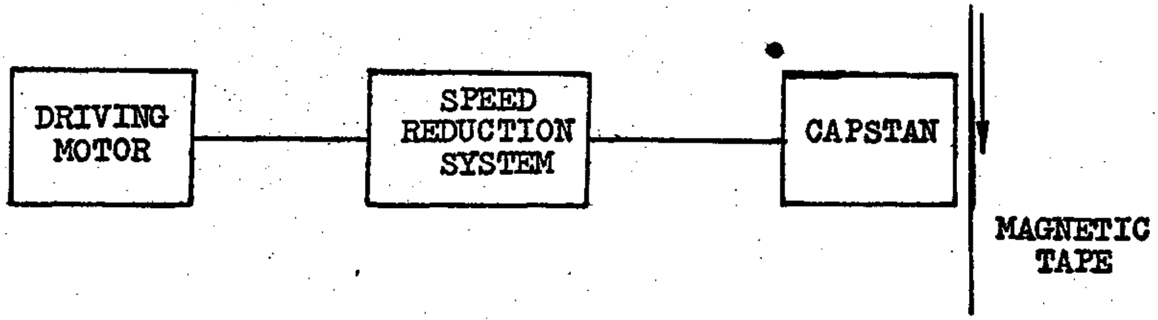

Fig. 2 Tape Transport

The first part of the transport is the motor which must provide the necessary power at the required stability to drive the transport. The next portion of the transport is the speed reduction system which was necessary, as a motor with the required low speed and desired stability was not available. The third portion of the transport is the capstan that imparts a linear motion to the magnetic tape from the rotational motion of the speed reduction system.

The electronics required to convert the ECG to a form suitable for recording take the form of an amplifier, a voltage controlled oscillator and a head drive circuit.

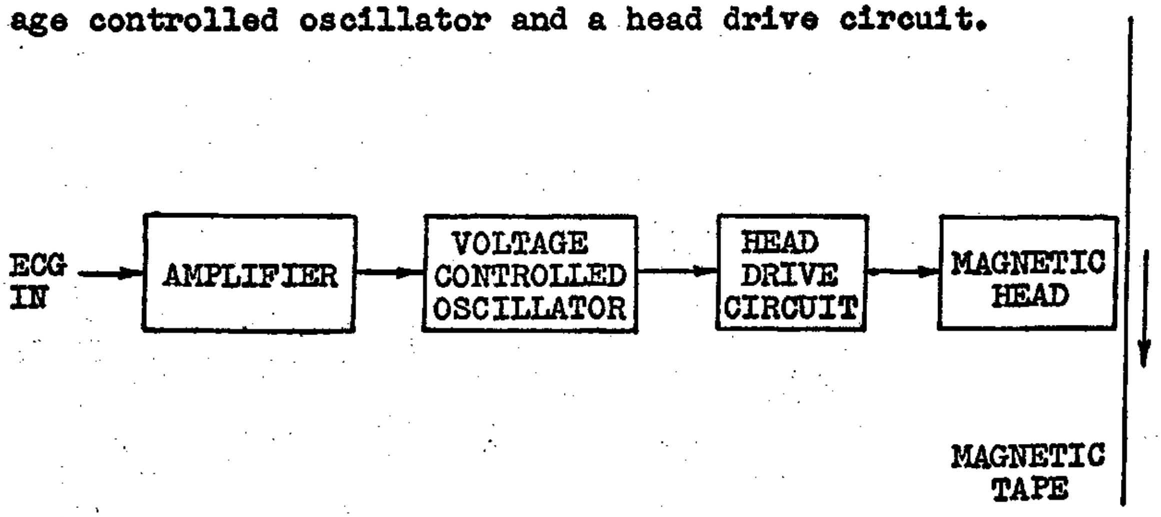

Fig. 3 Write Electronics

The ECG is fed into the amplifier which raises the signal to the level that is required to operate the voltage controlled oscillator. From the voltage controlled oscillator the signal passes to the head drive circuit which provides the necessary current to the magnetic head that in turn inscribes the desired flux pattern on the moving magnetic tape.

When the ECG is to be read from the magnetic tape, the write electronics are disconnected from the magnetic head and the read electronics are connected.

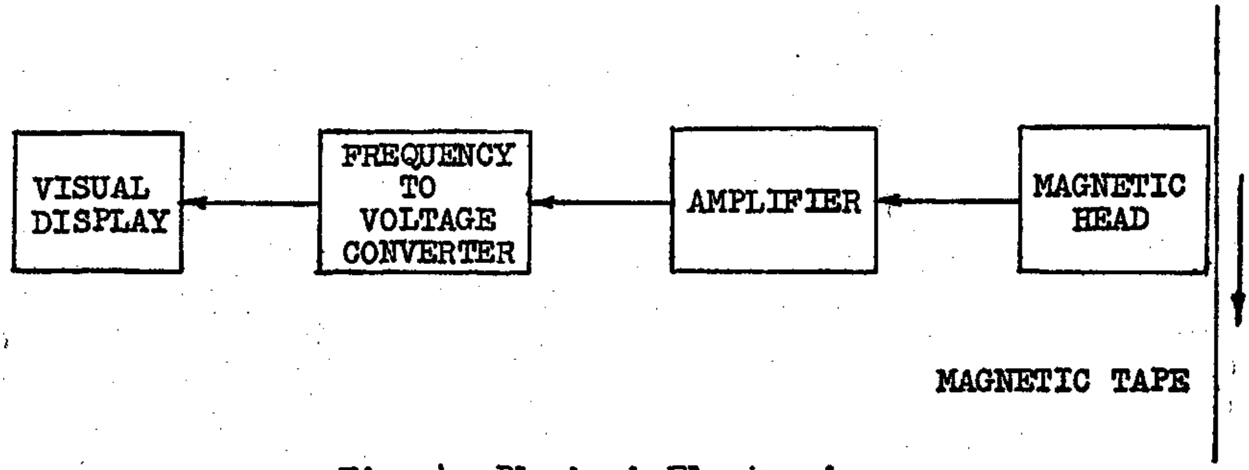

Fig. 4 Playback Electronics

During playback the magnetic head produces a voltage which is proportional to the rate of change of flux passing in front of the head and passes this signal to the amplifier. The amplifier raises the signal level to a value that will allow the converter to work properly. The converter changes the frequency of the signal from the tape to a proportional voltage. The voltage from the discriminator is then displayed in some visual form such as an oscilloscope or pen recorder.

2. THE MECHANICAL SYSTEM

2.1 Consideration in the Design of a Portable Tape Transport

The magnetic tape transport must move the tape past the head at a predetermined speed. If the velocity of the tape during the playback process is not precisely the same as during the record process, the data that are played back will contain distortion. The distortion introduced by speed change may be of two classes.

The first class which is known as flutter may be defined as "The deviation in the reproduced frequency from the recorded frequency which is a result of non-uniform speed of the recording medium during recording and reproduction" (Pear, 5). The flutter introduced into the record which may occur either during the record or playback process may be caused by:

1. Eccentricities of capstan, pulleys, pressure rollers, rotating guides, reels, and bearings.
2. Irregularities in the surface of the magnetic tape and belts with consequent variations in friction.
3. Irregularities in the width of tape or belts causing changes when the edges bear on a guide.
4. Non-uniform compliance in belts or tape.
5. Non-uniform hardness in rubber pressure rollers.
6. Variations in bearing torque.
7. Cohesion between successive tape layers.
8. Variations in drive motor torque.
9. Variation in reel torque.

Of these, variations in motor torque, bearing torque, and eccentricities of rotating elements could produce a cyclic variation in speed. The others are likely to cause a random variation in the speed of the tape past the head.

When size, weight, and environment do not permit an elaborate tape drive so that flutter can be reduced to an acceptable value, it is possible to improve the data signal by introducing a signal to modify the output. This type of flutter compensation is used in conjunction with F.M. telemetry techniques and is usually limited to correcting the noise or amplitude effects in the data.

The second class of error which is the time displacement error may be defined as "The error in the time between two events recorded on a single track when played back from a tape recording" (Pear, 5). The real difference from flutter is that high frequency components of flutter are not considered in this measurement. This is done by making the time intervals used in the measurement much greater than the period of the carriers used for flutter measurements, and this reduces the effect of high frequency components. This type of error may be caused by:

1. Slow changes in the motor speed.
2. Stretching of the magnetic tape.
3. Slippage in the speed reduction mechanism.

This type of error which is of low frequency may be corrected in the playback process by a servo controlled playback system.

A discussion of this method of correction is included in section 2.3.

Another consideration in the overall layout of the tape deck is the amount of magnetic tape that must be stored in order to obtain one complete record. In a stationary machine not limited by the available space, this is relatively unimportant. However, in a portable recorder such as the one described in this thesis, this is important as it does place a lower limit on the size of the instrument. When all the variables were considered, it was found that a reel of $\frac{1}{4}$ inch tape with an outside diameter of $4\frac{1}{2}$ inches would give a stored record of 1 hour.

Another important factor in the layout of the tape deck is the access time that is required to effectively use the data that were recorded. Although it is always desirable from the point of view of convenience to have a fast access time, the complex high speed rewind mechanism needed to give a fast access time is not always warranted. This is particularly true for the type of unit under discussion, since selected portions of the data may be transferred to a paper pen recording when a close examination is required.

2.2 The Difficulties in the Design of a Portable Tape Transport

In the design of a portable tape transport one must deal with the problems mentioned in the previous section, but the factor which finally limits the performance of the tape deck is the power available to drive the capstan drive motor. For example, if the recorder is completely portable and one chooses to use rechargeable nickel cadmium batteries as a source of power, the weight of the battery pack will be approximately 1/10 lb/watt hour* of power that is used. Since the recorder is intended for use during normal activity and if one assumes that a 10 hour running time is sufficient between battery changes, then the weight of the battery pack is 1 lb. per watt. Thus, to design a recorder that will not inconvenience the subject, the total power consumption must be kept down to the order of a watt.

The power required depends to a large extent upon the efficiency of the motor and the speed reduction system. The simplest way of obtaining a constant speed drive is to use a synchronous motor driven from a very stable reference oscillator. However, very small, synchronous motors are notorious for their very poor efficiency (as low as a few tenths of a percent (Fitzgerald and Kingsley, 6)). One hysteresis synchronous motor that was tested had a no load power input of 4 watts with a maximum power output of 5 watts with 12 watts input.

*This figure is based on Union Carbide specifications for nickel cadmium batteries that will deliver less than 100 milliamps for 10 hours.

Another aspect in the tape transport that presents difficulty is the speed reduction system. Traditionally, for a speed reduction system that occupies minimum space, a gear system is used. However, small gears do not produce a speed reduction that has a constant instantaneous speed.

The type of speed reduction that has been adopted by tape recorder manufacturers is a rubber belt system with large inertia flywheels to damp out any oscillations that may persist in the output. This type of speed reduction system has the disadvantage that rubber belts cannot be driven over a small diameter pulley necessary for a large speed reduction.

2.3 A Servo Controlled Playback as an Alternative to a Precision Recorder

If the best design, when size, weight, and complexity are considered, cannot reduce the flutter to a sufficiently low value, then there is a method by which the low frequency flutter may be reduced during playback (Young, 7; Schulze,,8).. This method uses a servo controlled motor to compensate for the variations in speed that occurred during the recording process. This may be done by recording a known frequency on one channel during the recording cycle. Then during playback the motor is driven to produce the same frequency from the tape as originally recorded. Shown in Figs. 5 and 6 is the block diagram of a system that may be used for playback speed correction.

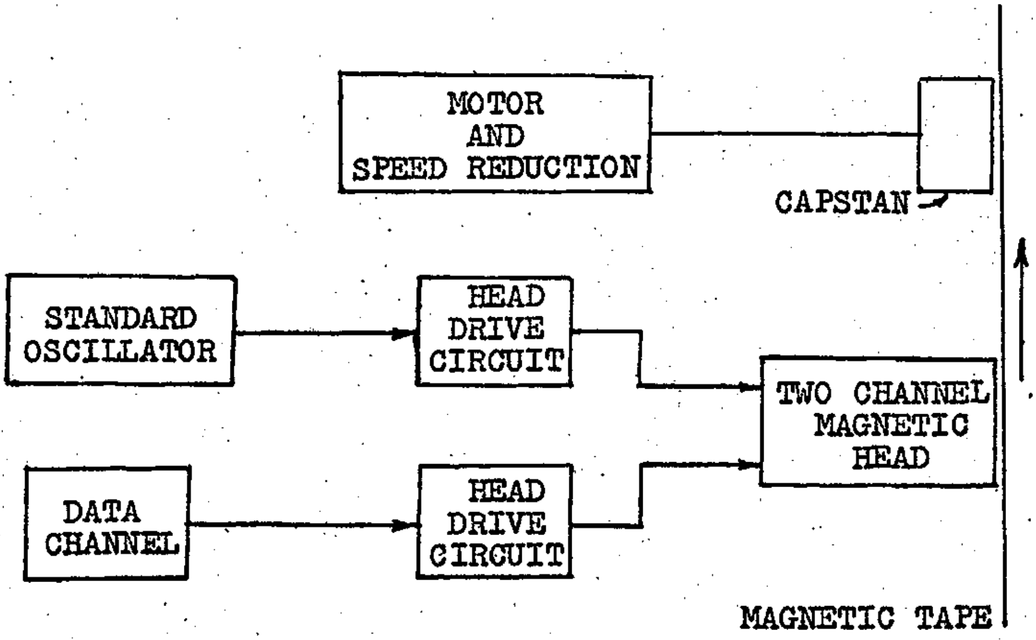

Fig. 5 Recording System for Servo Controlled Playback

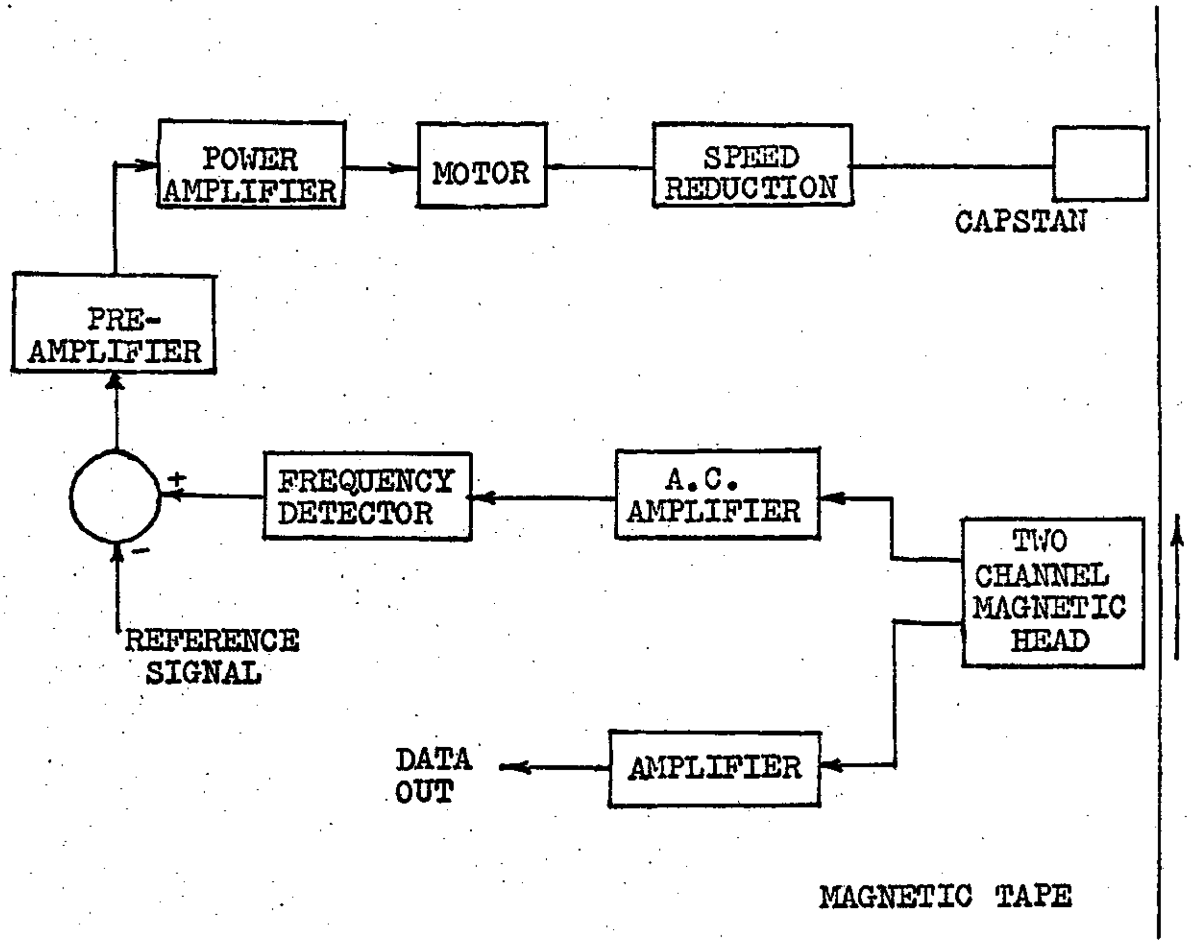

Fig. 6 Servo Controlled Playback System

Some of the distinct advantages of using this type of system are:

1. A very stable drive motor is not required in the recorder.
2. Real time error is reduced to a minimum.
3. Low frequency flutter components are compensated for.
4. Possibly less portable power will be required than for a highly stable motor.

When using this system one must be prepared to accept its shortcomings. Some of these are:

1. One channel of the tape is used for the reference signal, thereby reducing the amount of data that may be recorded on the tape.
2. If the recorder is to be portable and as light and as small as possible, then none of the servo system components may be located in the recorder and, therefore, a second complete tape deck must be used.
3. The servo system will only correct for time errors that are below the cutoff frequency of the servo system, which is in the order of 10 cycles/second.
4. The recorder must still maintain the speed within, say, 10% of the desired speed if precise control is to be obtained during playback.

An examination of the advantages of a servo system when considering a recorder design shows that the only relaxation in the recorder requirements is the precision required in the speed reduction system and the motor speed control. Since the data of interest are of such a nature that a 2% real time error would not lead to misinterpretation of the data, this method was discarded.

A compromise between the synchronous motor drive (which is very inefficient) and an elaborate playback system was arrived at by using an efficient d.c. motor with a speed regulator. The speed control in the motor selected is a mechanically operated governor. This controls speed by opening or closing a contact in series with the armature when corrective action is required. Under operating conditions it was found that the motor speed remained constant within ± 2%.

The requirements of the speed reduction system were met by using mylar* belts. This drive belt has been developed in recent years (Light and White, 9) as the need for a precision drive in instruments has increased.

Fatigue resistance, high strength, uniformity, and ability to drive extremely small pulleys are some of the properties which make the mylar belts attractive for mechanical instrumentation. These belts have, essentially, a flat homogeneous cross section and are fabricated from sheet mylar of from ½ mil to 10 mils in thickness.

The coefficient of friction for these belts over turned metal shafts is reported to be .15 for speeds below 4000 rpm and .2 for speeds up to the point where centrifugal effects begin to appear. This rather low coefficient of friction necessitates quite large initial tensions which require that ball bearings be used in the speed reduction system to minimize power loss due to friction. An initial tension of 2000 lbs/square inch is quoted as a safe design figure for initial tension in the belt.

* DuPont trademark.

2.4 Description of the Tape Transport Design

The length of record required to observe significant changes in the ECG was estimated to be 30 minutes prior to an event and 30 minutes following an event. This figure, together with the required minimum tape speed of $1\frac{7}{8}$ inches/second (section 3.3) fixed the total length of tape at approximately 560 ft. . To store this length of tape, a reel with an outside diameter of 4.5 inches was used.

The requirement of a continuous recording, retaining only a selected portion was met by using the endless loop tape storage reel shown in Fig. 7. The tape is drawn from the inside of the reel, passes over the heads and capstan, and is drawn in on the outside. Since the adjacent layers of tape within the reel must rub against one another, some form of lubrication must be provided to keep friction to a minimum. A graphite coating on one side of the tape performs this function very well. To further prevent the ragged edges of the tape from adhering to one another, the floor of the reel is cone shaped as shown in Fig. 8.

The tape drive system is shown in Fig. 9. This system was developed by Anderson (10) for applications where very high acceleration forces and vibration are present such as on a rocket sled. Although the recorder he developed has many of the characteristics we desire, the weight and power requirements

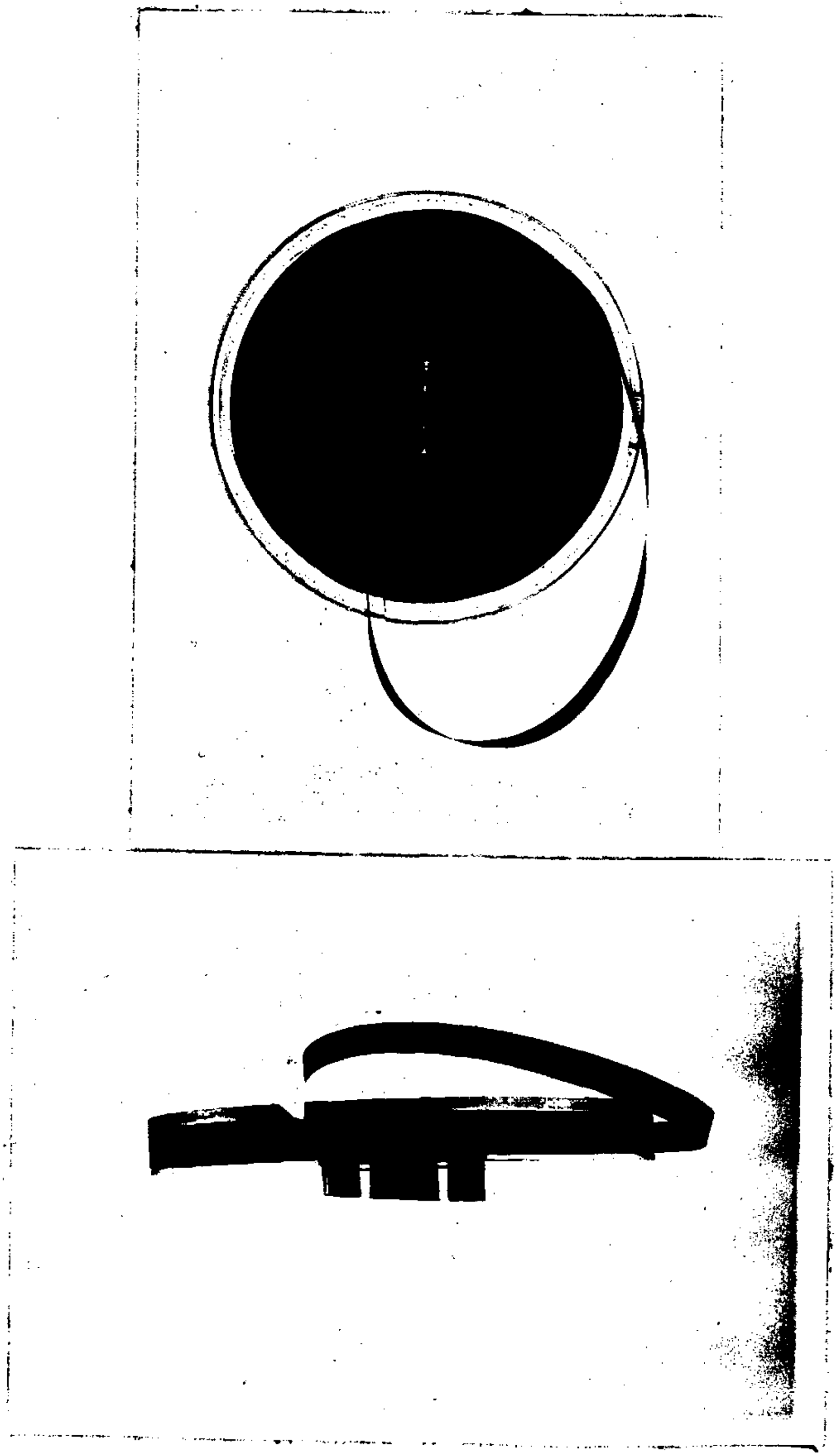

Fig. 7 Endless Loop Storage Reel

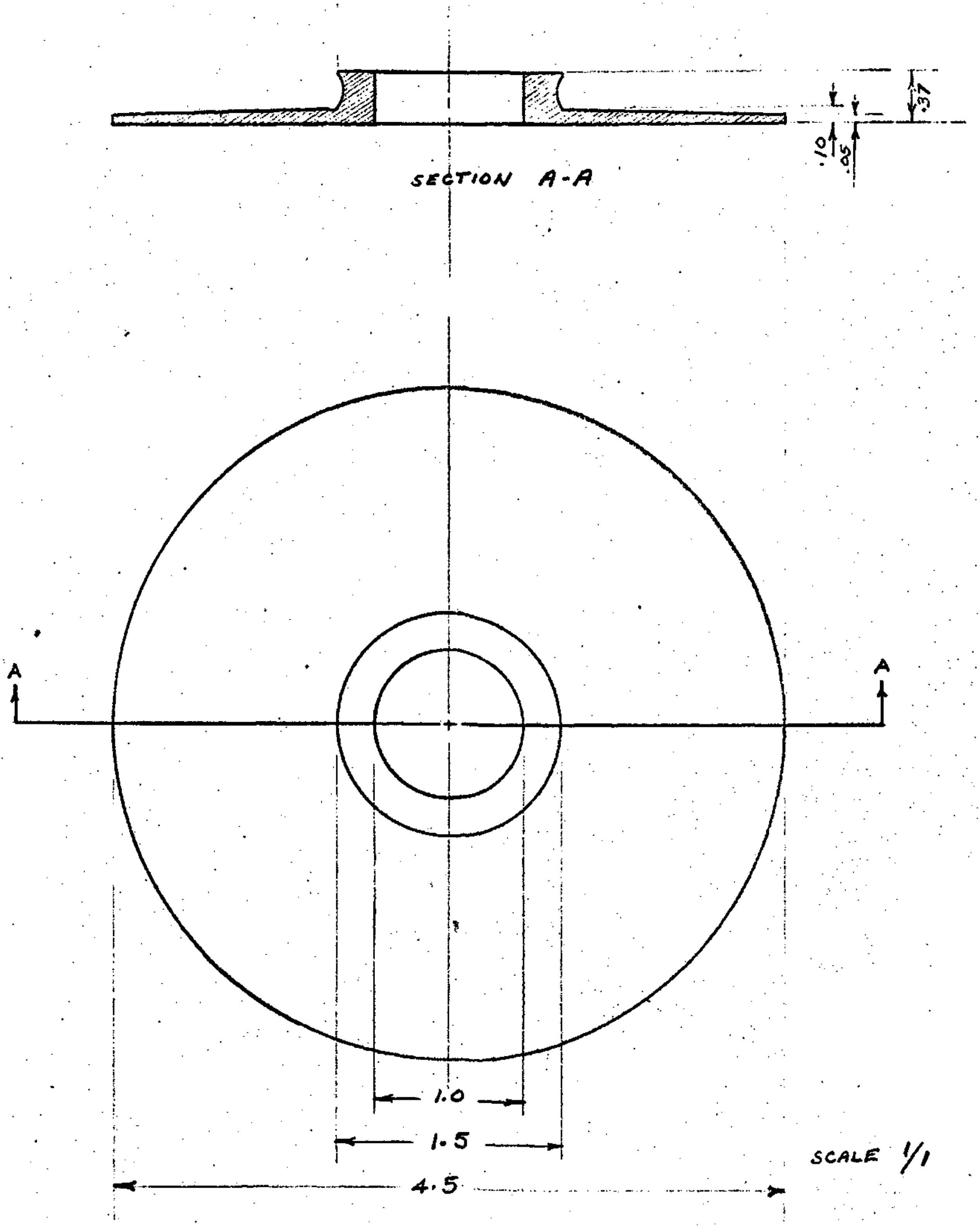

Fig. 8 Cross section of tape storage reel.

prohibit its use in our application.

The drive system consists of a continuous, seamless mylar belt that transmits the driving force from a capstan roller to the magnetic tape itself.

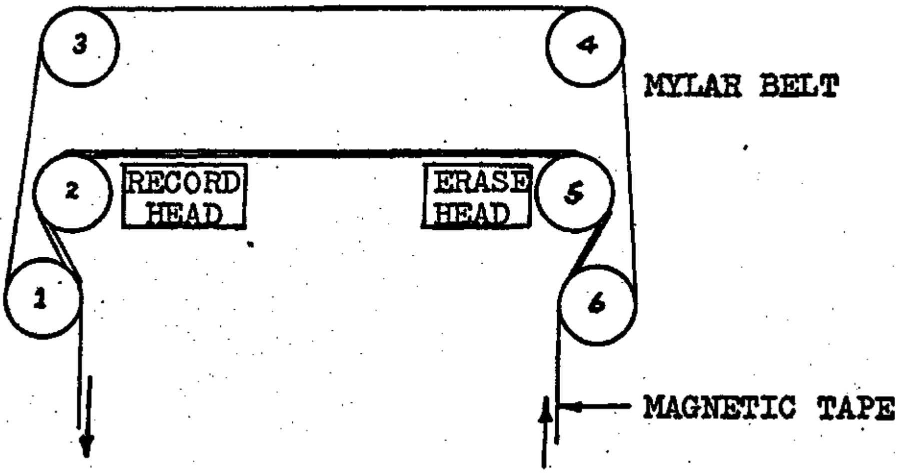

Fig. 9 Tape Drive System

Stretched tightly around pulleys 1 to 6 is an endless 2 mil mylar belt that is slightly less than ¼ inch wide. Pulley 1 is driven but all the rest are idler pulleys. The magnetic tape passes between the mylar belt and roller 5, between the belt and erase and record heads, then between roller 2 and the mylar belt and back to the supply reel. The edges on rollers 1 and 6 are used as guides to feed the vertical position of the magnetic tape. The mylar belt is held in the centre of the pulleys by putting a slight crown on all the pulleys except #1. The initial tension on the

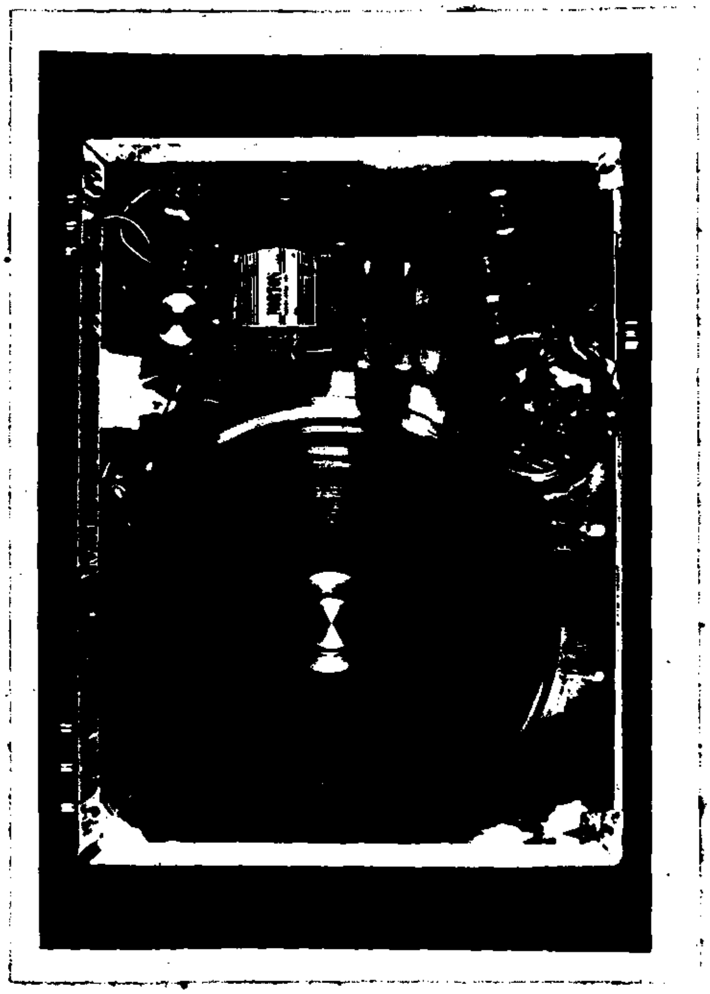

Fig. 10 Tape Drive and Storage Reel

mylar belt is applied by mounting the shaft of pulley 4 on an eccentric. Once the initial tension has been applied to the mylar belt, there is no need for readjustment except when the magnetic tape is removed and a new reel inserted. Shown in Fig. 10 is a photograph of the complete tape drive and storage reel.

Some of the advantages of using this type of drive over the conventional pinch roller and capstan are:

1. The need for pressure pads is eliminated.
2. Rollers 2 and 5 provide isolation points. These two rollers isolate that portion of the tape that is stretched over the heads from variations in supply and take up tension.
3. There is a minimum of sliding contact. One side of the tape (the side with the graphite coating) does not pass over a fixed member. This prevents the graphite from being scraped off.
4. There is a minimum of unsupported tape between capstan and head, thereby preventing horizontal oscillations in the tape.

One disadvantage to using this drive system is the inconvenience in replacing the storage reel. For simplicity the roller that is used to apply the tension to the mylar belt is held in place by screws and not by spring tension. Therefore, to change a tape the screws must be loosened and the belt tension relaxed. The new magnetic tape must be inserted between rollers and belt and then the belt tension must be re-applied.

Fig. 11 is a photograph of the underside of the tape deck showing the speed reduction system. The ¼ inch motor pulley on the extreme right of the picture rotates at a speed of 3000 rpm. This is coupled by a 1/8 inch mylar belt to a 2 inch pulley on the idler assembly. The small pulley on the idler assembly is coupled to the drive capstan by a 3/16 inch mylar belt. The total speed reduction of the assembly is 31.4:1 to give a capstan speed of 95.5 rpm. The 3/8 inch capstan on the top side of the tape deck gives a linear velocity of 1⅞ inches/second to the magnetic tape.

2.5 Tape Deck Test Results

The figures of merit used to judge the quality of a portable tape deck are:

1. Total weight (including batteries).
2. Power consumption.
3. Peak to peak flutter.
4. Real time error.

The total weight of the four channel recorder including batteries for 5 hours of recording is 3.5 lbs. The total power consumption is less than 1 watt.

The peak to peak flutter (from d.c. to 160 cycles/second) for this tape deck is 1%. This was measured by recording a 2000 cycle/second signal on the tape and then playing this

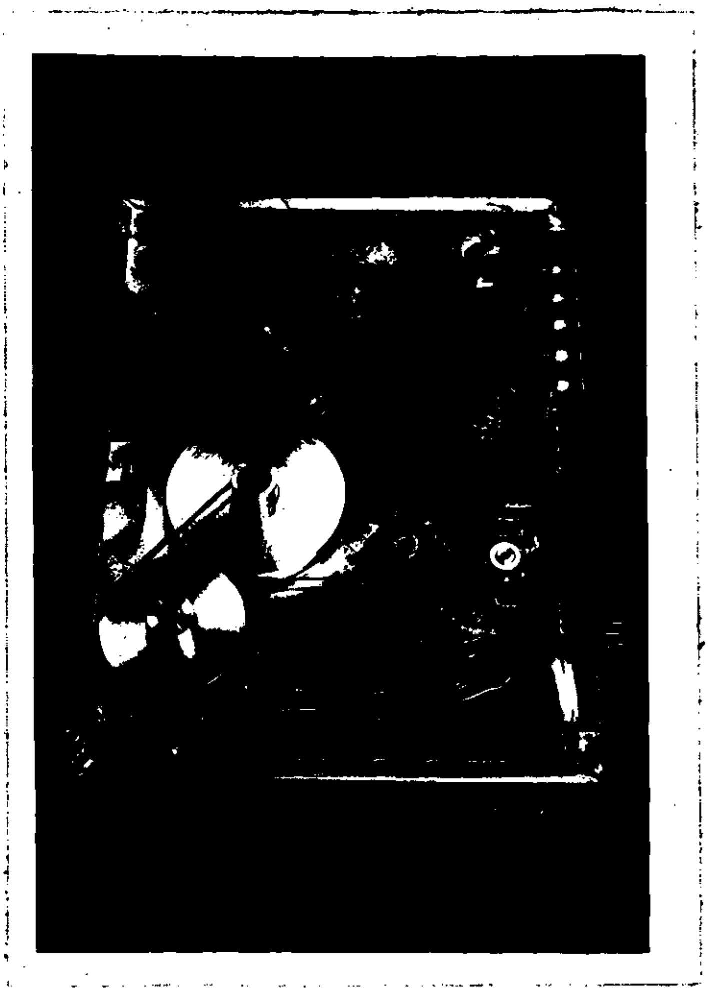

Fig. 11 Speed Reduction System

back through an F.M. discriminator (section 3.4.2). The output of the discriminator was then recorded using a paper pen recorder and the peak to peak deflections noted.

The real time error is less than 2% for ± 10% supply variations. This was measured by frequency modulating a 2000 cycle/second carrier at 100 cycles/second and recording this for a period of 1 hour. This was then played back and demodulated to obtain the 100 cycle/second. The zero crossings of the sine wave were then counted for 10 seconds at 10 second intervals and the totals for the 10 second intervals recorded. This was continued for 30 minutes at the same motor supply voltage as during recording. The motor supply voltage was then varied +10% and then -10% and the zero crossing counted for each case as before.

3. THE ELECTRICAL SYSTEM

3.1 Nature of the Data to be Recorded

Shown in Fig. 12 is a typical ECG waveform.

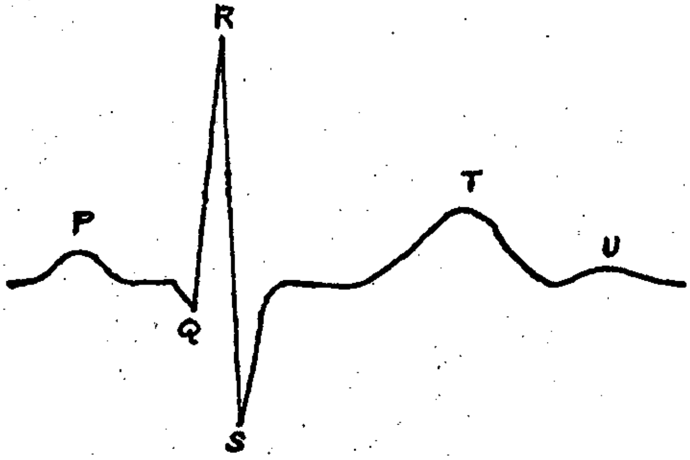

Fig. 12 A typical ECG waveform

This waveform is the activity of the heart as measured between two electrodes placed on standardized positions on the body. To the diagnosing physician the absolute amplitude, relative amplitudes, relative polarities of various segments, the time duration of segments, and the time intervals between segments are important parameters in diagnosing a heart condition.

Before a physician examines a series of recordings he may mentally note the quality of the ECG recording. One of the things he may note is whether or not the recording has a steady unwavering noise free baseline. This is important as the polarities of the waveforms (Wilson, et al., 11) are referred to the baseline as this is the only unalterable segment of the signature. The baseline is also used as a reference for measuring the amplitudes of the various segments.

While it is relatively easy to assess the quality of a particular recording, more information must be known about the amplitude versus frequency spectrum before the minimum requirements of a data channel used to transmit ECG's can be established. There have been several papers published (Scher and Young, 12; Thompson, 13; Langner and Giselowitz, 14) that have reported experiments to establish the minimum upper cutoff frequency that is required for recording ECG's, but none of these have defined the frequency spectrum that must be used to transmit a complete ECG. In general, however, the results of the analysis that have been published indicate that a maximum harmonic content occurs at the fundamental component of the QRS complex. Beyond the fundamental component of the QRS complex, the published literature indicates that the harmonic content decreases at a rate greater than 6 db per octave.

Another publication (Rikli, et al., 15) which was concerned with the computer extraction of ECG variables indicated that the QRS duration is not likely to be less than .05 seconds.

For this reason 20 cycles/second was chosen as the frequency at which the maximum frequency amplitude product is most likely to occur.

As well as the wide range of frequencies contained in an ECG, there is also a wide range of peak amplitudes. The clinical standard that has been adopted when recording ECG's is to adjust the sensitivity of the recorder to produce a deflection of 1 cm/millivolt (Wilson, et al., 11) and record the ECG on 5 cm. paper. An examination of some typical ECG records has also shown that a peak to peak amplitude of 5 mv is very rarely exceeded. For this reason a peak to peak amplitude of 5 mv was chosen as the maximum amplitude. This figure as well as the maximum harmonic content at 20 cps and the harmonic content decreasing at a rate greater than 6 dbs/octave defines the maximum slope that is likely to occur in an ECG at 314 millivolts/second by the following reasoning.

If $$A = 2\tfrac{1}{2} \sin \omega t$$

then $$\left.\frac{dA}{dt}\right|_{max} = 2\tfrac{1}{2}\, \omega \cos \omega t$$

and $$\left.\frac{dA}{dt}\right|_{max} = 314 \text{ mv/second}$$

On the other end of the spectrum we wish to know the lowest frequency that must be transmitted. This is of course fixed by the lowest frequency of the ECG which will be the fundamental component of a fourier analysis. The lower limit is usually set in the range between .05 to .5 cps (Thompson, 13). The reason for the extremely low frequency requirement is that too much phase shift in the lower frequencies cannot be allowed. Even though it is desirable to maintain a very good low frequency response and a very good high frequency response, there are situations where a wide bandwidth may not be desirable. One of these cases is an athlete under strenuous activity. In this case it is desirable to limit the bandwidth to as small as possible to reduce the amount of muscle noise in a recording. It is just as important to limit the low frequency response as it is to lower the high frequency response. This is important since under the most adverse conditions, deep breathing will result in a change in chest impedance. If the electrodes are slightly polarized, then there will be large voltage changes at the input of the amplifier which could drive the amplifier out of its dynamic range. Since the breathing rate is comparable to the lower frequency cutoff point generally used, it is essential that the lower cutoff point be no lower than 1 cycle/second if used on a very active subject. The phase shift produced by an amplifier with this cutoff frequency will become less serious when using it on a subject under strenuous activity as the heart rate will generally be higher than normal.

3.2 Capabilities of a Magnetic Tape Channel

Shown in Fig. 13 is the schematic representation of the direct magnetic tape recording and reproduce process.

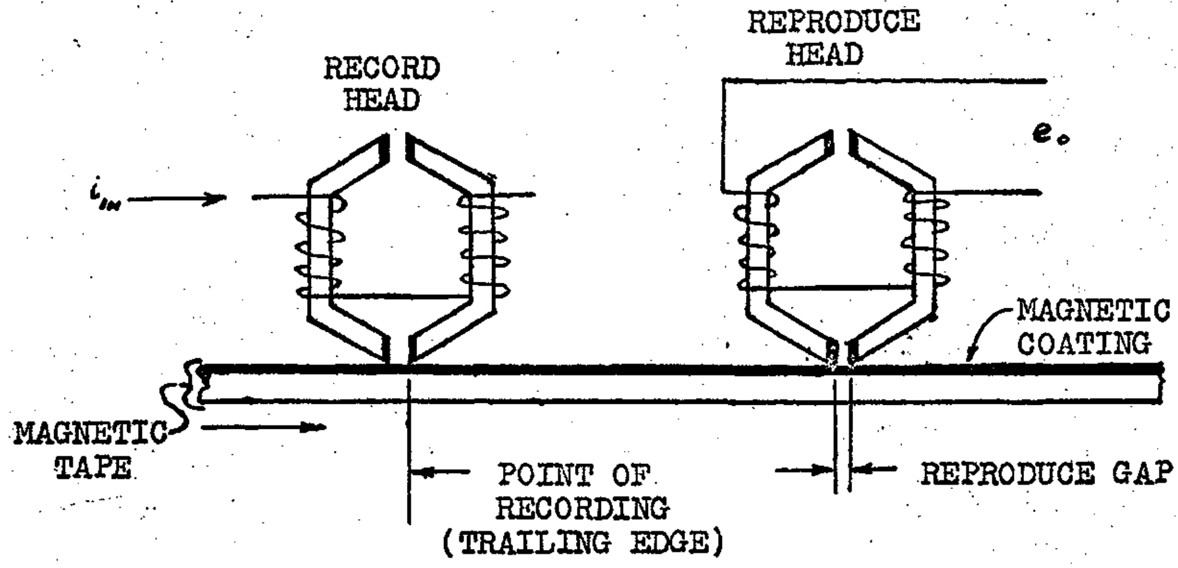

Fig. 13 Magnetic tape recording and reproduce process.

In the direct record process the signal to be recorded is mixed with a high frequency bias and presented to the head as a varying current i. The recording head consists of a magnetic core in the form of a closed ring having a short non-magnetic path in series with the magnetic path of the core. The magnetic tape consists of a plastic base such as acetate or mylar* on the surface of which fine particles of magnetic materials are dispersed uniformly. The magnetic surface of the tape contacts the magnetic head at the gap, in effect shunting the gap and completing the magnetic path of the head

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* DuPont trademark

core. The signal current flows through a winding which links the magnetic core and produces magnetic flux, ϕ, whose magnitude is proportional to the recording current.

$$\phi = K\,i \qquad (1)$$

Where K is a constant of proportionality

The tape is moving across the heads at a linear velocity of v. Any particle in the magnetic medium crossing the gap remains in a permanent state of magnetization which is proportional to the flux flowing through the head at the instant that particle passes out of the gap. (This proportionality will be discussed further when the high frequency bias is discussed.) Thus the actual recording takes place at the trailing edge of the record gap.

If the signal to be recorded is sinusoidal, the intensity of magnetization on the tape will vary sinusoidally along the length of the tape. A wavelength of recorded signal along the tape will occur for each complete alternation of the input electrical signal. This wavelength is directly proportional to tape speed and inversely proportional to the frequency of the recorded signal. If

λ = wavelength on tape (inches)

v = tape speed (inches/second)

f = frequency (cycles/sec) of signal

then

$$\lambda = \frac{v}{f} \qquad (2)$$

During playback the magnetized surface of the tape passes the gap of a reproduce head which is similar in construction to the record head. The portion of the tape in contact with the gap is bridged by the magnetic core of the reproduce head which causes the magnetic lines of flux to flow through the core. The magnitude of the flux is a function of the average state of magnetization on that portion of the tape actually spanned by the gap at any given instant. As the tape passes by the reproduce gap, the amount of flux through the core varies with the varying state of magnetization on the tape and causes a voltage to be generated in the winding linking the core. It is important to note that the voltage generated is proportional, not to the magnitude of the flux but to its rate of change. Thus, if flux is proportional to recording current,

$$\phi = K\,i = K \sin(2\pi ft) \qquad (3)$$

the playback voltage from the reproduce head is

$$e = K'\frac{d\phi}{dt} = K''\frac{d}{dt}\sin(2\pi ft)$$

$$e = 2\pi K'' f \cos(2\pi ft) \qquad (4)$$

From this it can be seen that the playback voltage is dependent upon the frequency and for constant current recording the output voltage will vary in direct proportion to frequency.

Fig. 14 illustrates the recording and reproduce characteristics as a function of frequency.

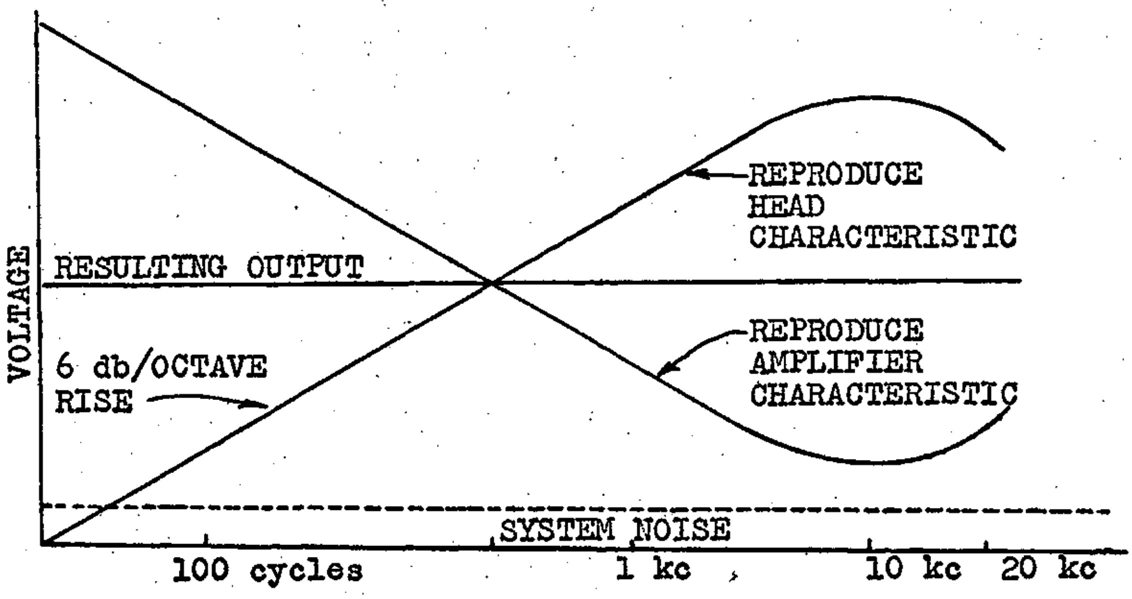


Fig. 14 Recording and reproduce characteristic.

This figure illustrates that the reproduce amplifier must have a frequency response characteristic which is the inverse of the reproduce head characteristic in order that an overall flat frequency response can be obtained. This is referred to as playback equalization. This figure also illustrates the upper and lower frequency limit of the tape.

As we go lower and lower in frequency, the output voltage decreases until it approaches the inherent noise level of the system. At this point it is impossible to recover the signal by equalization. This condition leads to the principal limitation of the direct recording process. There is a lower frequency limit, below which it is impossible to record and playback successfully. Typically, this frequency is 50 cycles/second.

At the high frequency end of the recording spectrum there is also a limit. When the tape is being played back the average value of tape magnetization spanned by the gap is continually changing, resulting in an output voltage from the head. If we have a very short wavelength recorded on the tape which is equal to the dimension of the gap itself, then under this condition the average magnetization under the head is zero and does not change as the tape moves by. The output of the head is, therefore, zero. This is also illustrated in Fig. 14. As the frequency approaches that value at which the wavelength equals the gap width, the output from the reproduce head falls off rapidly.

Although the limitation on high frequency response resulting from the "gap" effect cannot be eliminated, it can be improved in two ways.

Since $e \rightarrow 0$ as $\lambda \rightarrow d$ and $\lambda = \frac{v}{f}$, we can record and play back higher frequencies by either reducing the size of the reproduce gap, or by increasing the tape speed. If the gap size is reduced to get better resolution at high frequencies the voltage output of the reproduce head will fall. The result is a deterioration in the dynamic range or signal to noise ratio. On the other hand, the tape speed may be increased which increases the amount of tape required for a given record length and also results in increased head wear.

Another aspect of the direct record process is the addition of high frequency bias. In a.c. biased recording a large high frequency alternating signal is added to the signal to be recorded and both are impressed upon the record head simultaneously. Thus, as the tape passes in front of the record head, it is subjected to a field consisting of the intense high frequency bias and the information signal to be recorded. The resulting transfer characteristic is shown in Fig. 15. For comparison, the transfer characteristic obtained in the absence of high frequency a.c. bias is also shown.

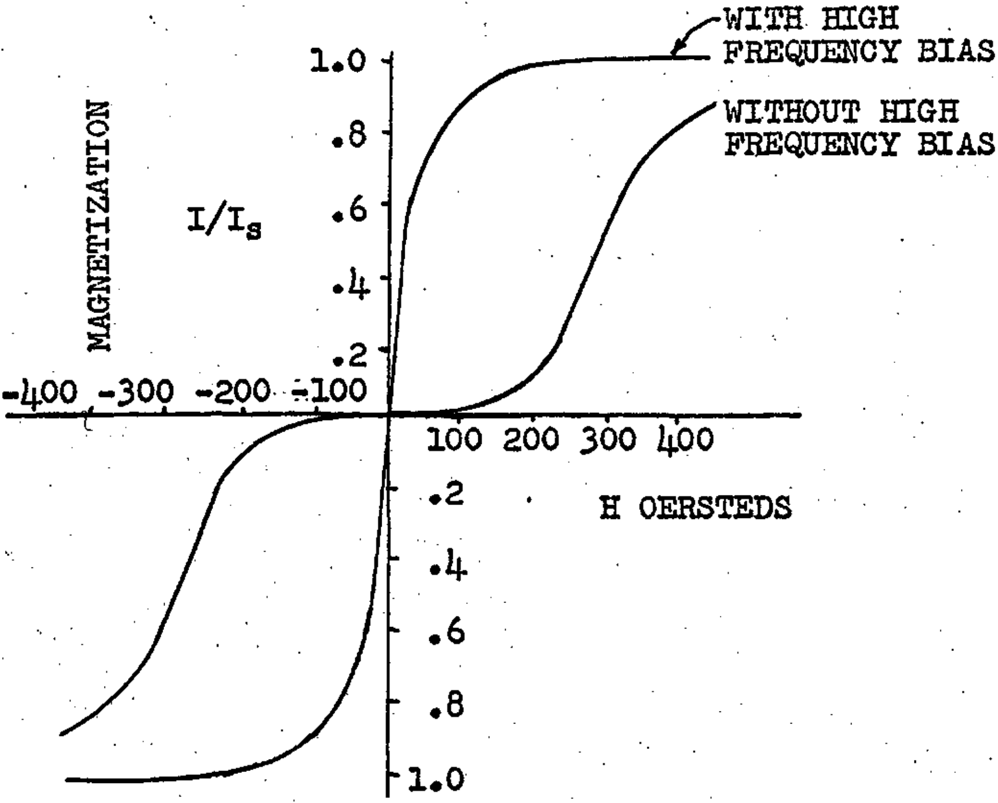


Fig. 15 Magnetization Characteristic*

* From reference 16.

The wavelengths which would correspond to the high frequency bias are too small to be resolved by the playback head. The actual bias frequency is not too critical but it is usually selected to be at least 3.5 times the highest frequency to be recorded to minimize any interaction which may occur between bias frequency and higher order harmonics of the signal frequency. Other than linearizing the transfer characteristics, the high frequency bias does not enter into the recording or subsequent playback process.

3.3 Selection of the Carrier System

3.3.1 Analogue systems

As shown in a previous section, a channel that is to be used to record an ECG must be capable of recording frequencies as low as .5 cycle/second. It has also been shown that this low a frequency response is not possible when using a magnetic tape channel in the direct record mode. Bearing these two things in mind, the following analogue recording techniques are suggested.

1. Amplitude modulation of a carrier.
2. Frequency modulation of a carrier.
3. Pulse width modulation.

Amplitude modulation which is the simplest form of modulation cannot be used for data recording because of the amplitude instability of the record playback process. The amplitude instability of the process is brought about by the surface condition of the magnetic tape medium. The effect manifests itself by instantaneous signal drop out or reduction in signal level. Signal dropout or reduction is primarily caused by small "nodules" or clusters of oxide particles which form along the surface of the tape. As the tape passes across the head, the nodules cause the tape to be withdrawn from the head and results in a drop in signal level. A similar drop in signal level may occur if a foreign particle of dust finds its way onto the surface of an otherwise perfect tape.

A second method to overcome the inability to record very low frequency phenomena is to use a carrier which is frequency modulated by the information signal. The advantage of using this method over amplitude modulation is an improved signal to noise ratio in the presence of noise and a demodulated signal level that is independent of the received carrier level. The greatest drawback in using frequency modulation is that any change in tape speed either during the record or playback process results in the carrier frequency being shifted and, upon demodulation, these changes in speed appear as noise in the output signal. This limitation caused by flutter is particularly acute if a frequency multiplexing scheme is to be used. In order to record a multiplicity of carriers in the available spectrum, it is necessary to restrict the maximum frequency deviation of any one carrier. The frequency deviation commonly used has been ± 7½% of centre frequency. Thus, if a 7½% frequency deviation corresponds to a 100% input signal, a 1% deviation in frequency resulting from flutter in the transport would appear as a $\frac{100}{7.5} = 13.3\%$ noise signal. The noise introduced by flutter is most commonly reduced to an acceptable level by recording only one carrier per channel and using a very large frequency deviation. The most common wide deviation systems use a deviation of ± 40% of centre frequency. In this case a 1% speed error will only cause a $\frac{100}{40} = 2.5\%$ noise signal.

A second drawback when using frequency modulation is a less efficient use of magnetic tape than when using the direct record method. To ensure an adequate sampling rate when using F.M., the carrier centre frequency is made at least five times the highest frequency to be recorded. This, of course, means that the tape speed must be proportionately increased.

The third method that is used in magnetic recording to obtain low frequency response is to use pulse width modulation. In this method the information signal is used to modulate the width of a constant frequency pulse train. Using this method it is possible to record several channels of information on the same magnetic tape channel. This is done by sampling each channel in sequence and converting an amplitude into a pulse duration. This process known as time division multiplexing is most commonly used when a large number of channels of information are to be recorded simultaneously. The main drawback in using pulse duration recording is a less efficient use of the tape (Weber, 17) than frequency modulation and, consequently, this method finds use only when many channels of low frequency information are being recorded.

3.3.2 Digital systems

Since it is quite often desirable to have information in digital form so that data reduction may be done using digital computers, the possibility of recording an ECG in digital form was explored. Pulse code modulation was the first to be examined as this is almost the universal language of computers.

As in pulse width modulation, a sampling technique is used to measure a varying signal. The sampled readings are thus converted into a code consisting of a series or group of binary digits. In contrast to the decimal system, the binary system employs only two digits. Digital recording is accomplished by magnetizing the tape to saturation in either of its two possible directions (+ or -) at discrete points along its length. Thus, there is only one of two states of magnetization at any point on the recorded tape. Any of several basic techniques and variants may be used for recording binary digits (Weber, 17).

Although the electronics required for writing the digitized information on the tape is quite simple, the electronics required to convert the analogue information to digital form is not. The circuitry is quite complex and bulky. For our application both the volume occupied by a digital to analogue converter and the power required is prohibitive to the use of pulse code modulation. The utilization of magnetic tape is also quite low as compared to frequency modulation (Weber, 17).

A second type of system for transmitting information in digital form which is known as delta modulation does have some advantages when compared to pulse code modulation. The chief advantage of this system is an extremely simple conversion circuit. This method which is rather novel and has not as yet found widespread use will now be dealt with in detail.

3.3.3 Delta Modulation

In the other types of modulation that have been discussed, an amplitude, a frequency, or a pulse width has been made proportional to a voltage. In the pulse code modulation the instantaneous value of a voltage is coded into a binary word consisting of ones and zeros. Delta modulation (deJager, 18) differs from these systems in that the value of the instantaneous received signal is made up of all previously transmitted values. The received signal is constructed by adding or subtracting a fixed quantity (Δh) from the received signal at a fixed rate. The signal is transmitted by sending a positive pulse at each upward step and a negative pulse at each downward step. The type of coding that results is a 1 digit code. A received signal using this type of code will appear as in Fig.17.

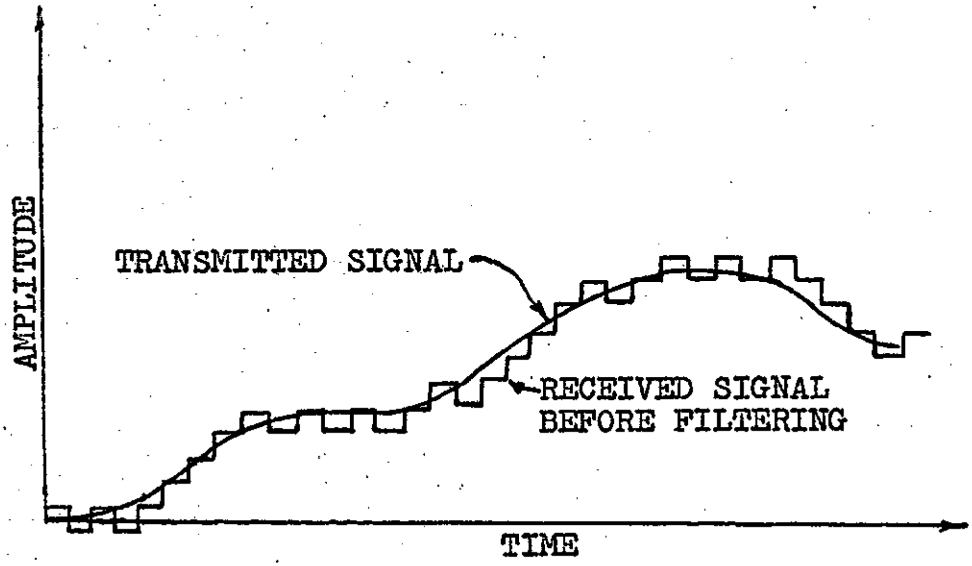


Fig. 17 Transmission by delta modulation.

In this system of pulse modulation the receiver consists only of a linear network and a pulse regenerator for elimination of the transmission noise. The main problem is ensuring the right sequence of pulses at the sending end. A system that will generate the pulse train that contains the information must have a network that remembers the reconstructed signal, a network that compares the reconstructed signal with the signal being presented to the input, and a network that generates a pulse of the proper polarity at a fixed rate. Shown in Fig. 18 is the block diagram of the elements of the transmitter and receiver.

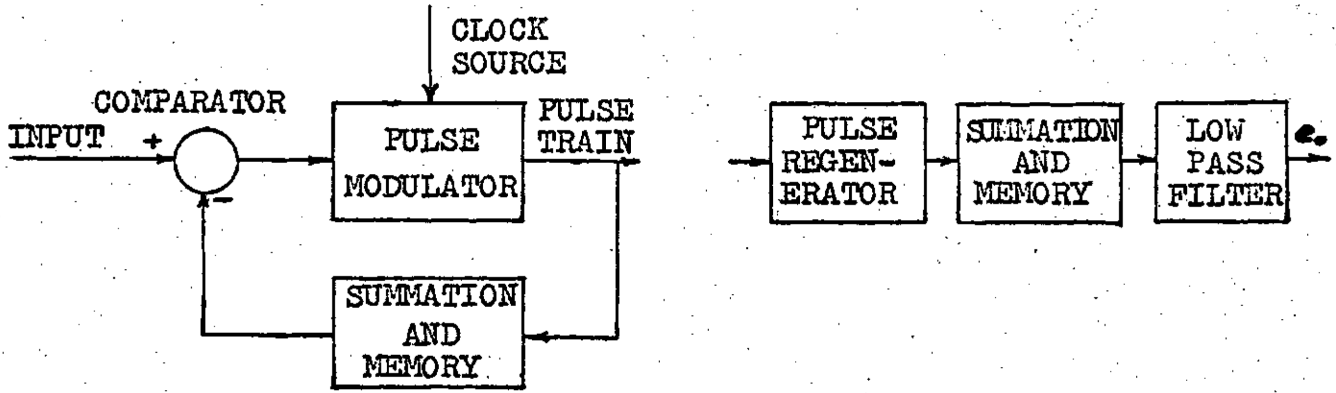


Fig. 18 Delta modulation transmitter and receiver.

The input signal is fed to the comparator where the value of the signal in memory is subtracted from it. The sign of the difference is then fed from the comparator to the pulse modulator. The pulse modulator then generates a pulse of known width and amplitude with the same polarity as the sign of the output from the comparator. This pulse is then sent along the transmission path and to the memory where it is added to the existing signal. At the receiver the pulse is reshaped and added to the memory. The low pass filter on the output filters the unwanted high frequency components from the received signal.

In the simplest form of delta modulation the summation and memory element in the feedback loop of the transmitter is an integrator. Variations in the type of delta modulation coding may result if the element in the feedback loop is changed. One variation is to use a pure integration followed by a partial second integration. In this case the type of coding is referred to as double integration delta modulation. Before discussing the practical circuit of a delta modulator we shall investigate the capability of a delta modulation channel.

3.3.4 Capability of Single Integration Delta Modulation

The first point of consideration in single integration delta modulation is the output of the modulator under conditions of zero input. In this case, if the integrator also has a zero level stored in it, then the output of the comparator is also zero. The output of the pulse modulator in this situation is indeterminate. However, the pulse modulator must be constructed in such a way that either a positive or negative pulse is transmitted in this situation. Let us assume that a positive pulse is generated in an indeterminate situation. This pulse is then integrated and the sign of the integrated value is changed. The input to the pulse modulator when the time comes for the next pulse to be generated will be negative and, consequently, a negative pulse will be generated. Thus we can see that under the condition of zero signal input to the comparator, the output of the pulse modulator will be a pulse train that is alternately positive and negative.

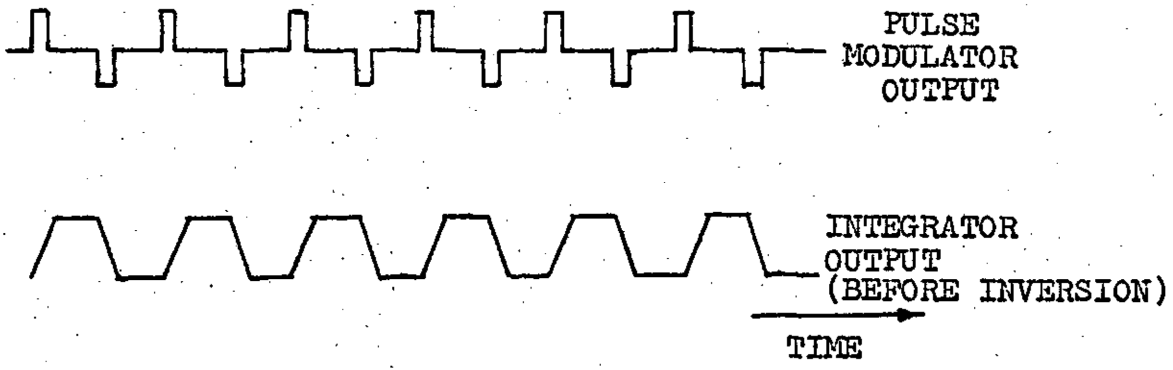


Fig. 19 Modulator signals with 0 input.

Another type of input to the modulator that is of interest is a step input. With a positive step applied to the input, the output of the comparator and, hence, the pulse modulator must be positive until the voltage on the integrator is equal to the input voltage. This is illustrated in Fig. 20.

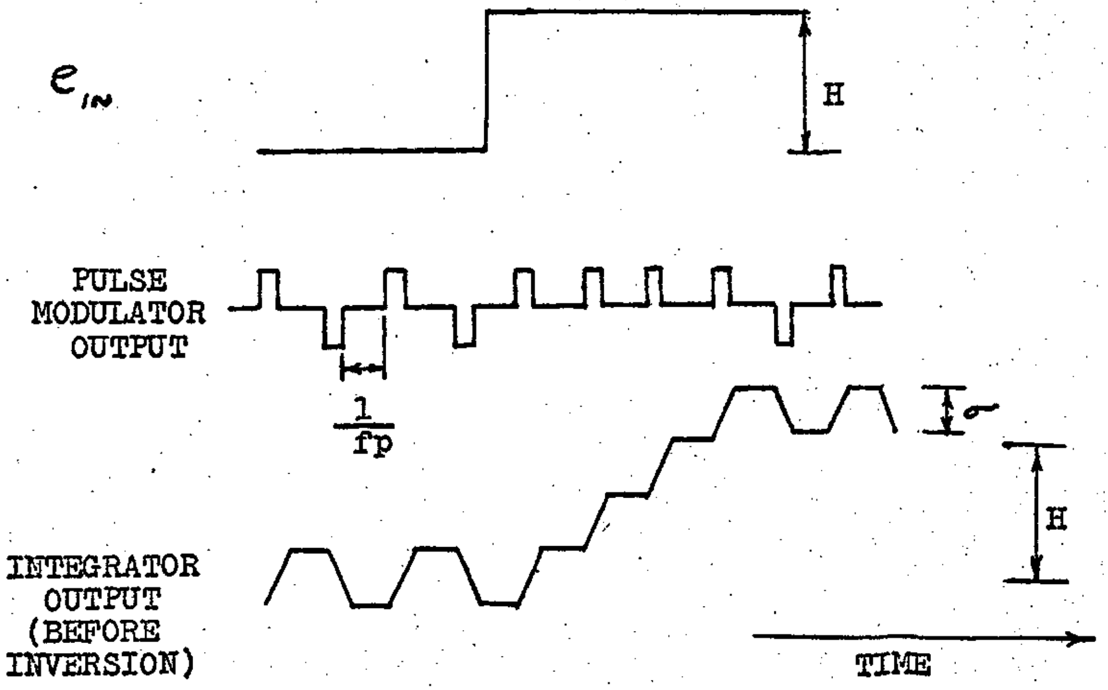


Fig. 20 Modulator signals with a step input.

If the step size at the integrator due to one pulse is σ and the pulse repetition frequency is fp, then the time T taken for the output of the integrator to reach the amplitude H of the step function is:

$$T = \frac{H}{fp\,\sigma} \qquad (5)$$

By rearranging this equation and differentiating with respect to time, we see that:

$$\frac{dH}{dT} = fp\,\sigma \qquad (6)$$

This equation shows that the slope that may be transmitted without overloading is the product of the pulse repetition frequency and step size. This relationship fixes the step size for a given pulse repetition frequency and the maximum slope contained in a signal. Conversely, the pulse repetition frequency is fixed once the step size is chosen. If the system is to transmit a sine wave of amplitude "a" and frequency f, then the maximum slope in this signal is $2\pi fa$. Equating this to equation 6,

$$2\pi fa = fp\,\sigma$$

or

$$a = \frac{fp\,\sigma}{2\pi f} \qquad (7)$$

We see that the maximum amplitude of a sine wave that may be transmitted is a function of pulse repetition frequency, step size, and signal frequency in accordance with the foregoing equation.

Fig. 20 also illustrates another limitation of delta modulation. This limitation is the accuracy of a waveform when transmitted by delta modulation. The difference between the original signal and the received signal gives rise to quantizing noise. De Jager (18) has shown that when using single integration delta modulation the signal to noise ratio due to quantizing noise is given by:

$$\frac{S}{N} = C_1 \frac{(fp)^{\frac{3}{2}}}{f \cdot (f_o)^{\frac{1}{2}}} \tag{8}$$

where $f_o = \omega_o/2\pi$ is the cutoff frequency of the low pass filter at the receiver, fp is the pulse repetition frequency and f is the frequency of a sine wave of maximum amplitude as per equation (7). De Jager (18) has also evaluated the constant C_1 to be .20.

3.3.5 Double Integration Delta Modulation

When using the delta modulation system to transmit information there exists the possibility of altering the feedback network to increase the coding accuracy. De Jager (18) has considered the addition of partial second integration in the feedback network and has shown that it results in an improved signal to noise ratio when compared to single integration delta modulation under the same conditions. When using a double integrator type of element in the feedback loop, it has been shown (de Jager, 18) that the feedback loop should have a frequency response as shown in Fig. 21.

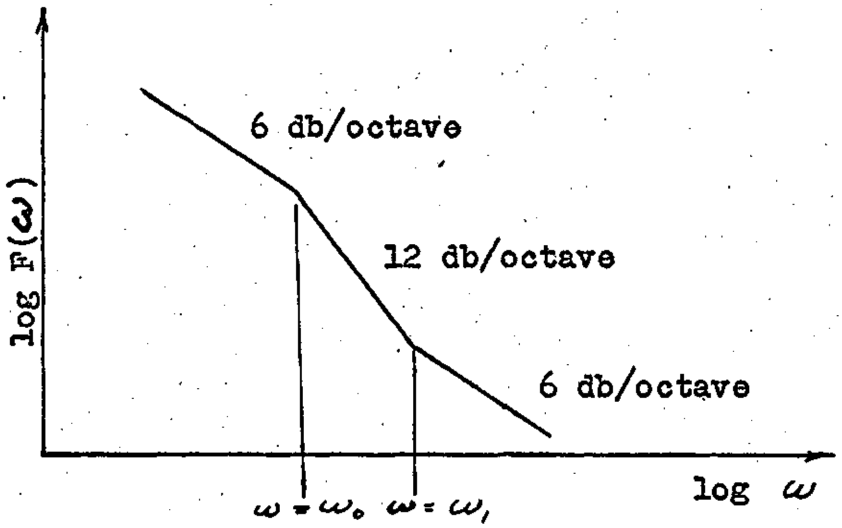


Fig. 21 Frequency response of the feedback loop for double integration.

In Fig. 21, $\omega_0 = 2\pi f_0$ where f_0 is the highest frequency to be transmitted and $\omega_1 \tau = 1$ where τ is about equal to the interval between two pulses. It has also been shown that when using this type of network in the transmitter, a single integration network may be used at the receiver.

When using double integration delta modulation, the signal to noise ratio is given by:

$$\frac{S}{N} = C_2 \frac{(fp)^{\frac{5}{2}}}{f \cdot (f_0)^{\frac{3}{2}}} \qquad (9)$$

The constant C_2 has been evaluated to be .026.

3.3.6 An Estimate of the Required Pulse Repetition Frequency to Transmit an ECG

If we assume a required signal to noise ratio of 100 (40 db) and a cutoff frequency of 200 cycles/second in the low pass filter at the receiver, then using equation (8) we may estimate the required pulse repetition frequency. In doing this calculation we shall also use f = 20 cycles/second as a representative signal frequency. Rearranging equation (8):

$$fp = \left(\frac{\frac{S}{N} \cdot f \cdot f_0^{\frac{1}{2}}}{.2} \right)^{\frac{2}{3}} \qquad (10)$$

Substituting the appropriate values and solving, fp = 2.7 kilocycles/second.

If we then use this pulse repetition frequency and find the signal to noise ratio for double integration delta modulation by using equation (9) the signal to noise ratio is 175 or 45 db.

When using a pulse repetition frequency of 2.7 kilocycles/second, the step size may be found by using equation (6) and the maximum slope criteria as established in section 3.1. The step size turns out to be .116 millivolts. The accuracy of representation when using a 5 millivolt full scale deflection is then 2.32%.

These calculations would indicate that a pulse repetition frequency of 2700 pulses/second would give reasonable fidelity when transmitting an ECG. However, the final test that determines whether a record is acceptable or not is a comparison of the transmitted and received record. This is particularly true with an ECG if a diagnosis is to be made on the basis of pattern recognition. In order to compare a record before and after transmission, a delta modulator and receiver were built.

3.3.7 The Delta Modulator Circuit

The published articles on delta modulation describe many methods of obtaining delta modulation. The majority of the modulators have been built for transmitting speech signals, but the most novel and simplest modulator which has been described by Balder and Kramer (19) was used to transmit video signals. Their delta modulator circuit using single integration has only 3 resistors, 2 tunnel diodes, 1 inductor, 1 capacitor, and a clock source. This circuit using an inductor as an integrator, although very simple, is not capable of performing double integration. A second circuit which they describe also uses the two tunnel diodes as a comparator and pulse modulator, but by using capacitors and transistors in the feedback loop, they obtain double integration. However, this network is not capable of integrating at the very low frequencies that we are interested in, and hence this portion has been redesigned. Before discussing this feedback circuit which has been designed for low frequency application, we shall examine the operation of the two tunnel diodes, which is the combined comparator and pulse modulator.

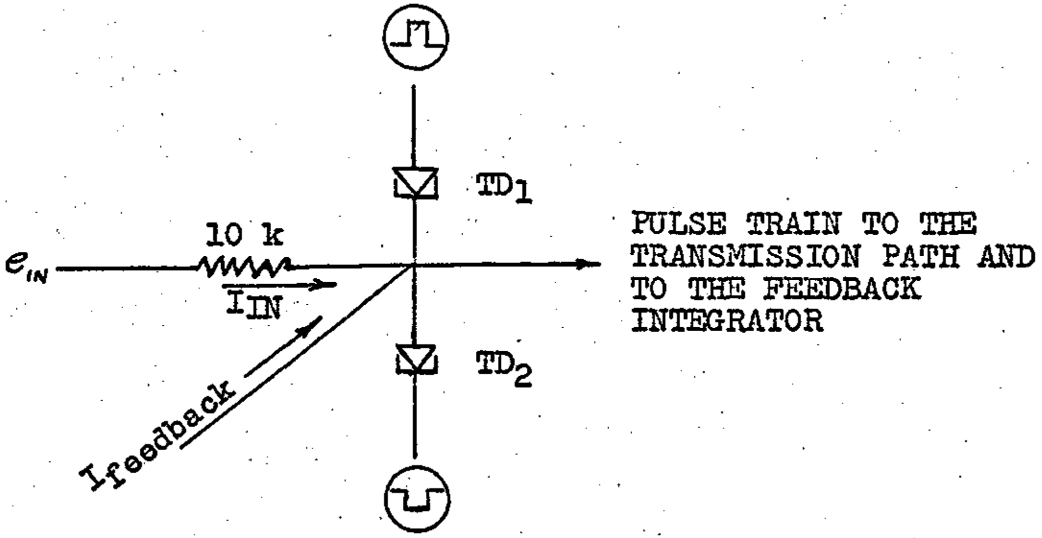


Fig. 22 Comparator and pulse generator.

The circuit shown in Fig. 22 consists of two tunnel diodes in series fed by two synchronous pulse sources of opposite polarity. At the start of the supply pulse, the circuit is biased by a signal (the sum of the input and feedback current) at the connecting point of the two diodes. The polarity of this bias determines the polarity of the output pulse.

The operating principle of the tunnel diode pair will be explained with the aid of Fig. 23.

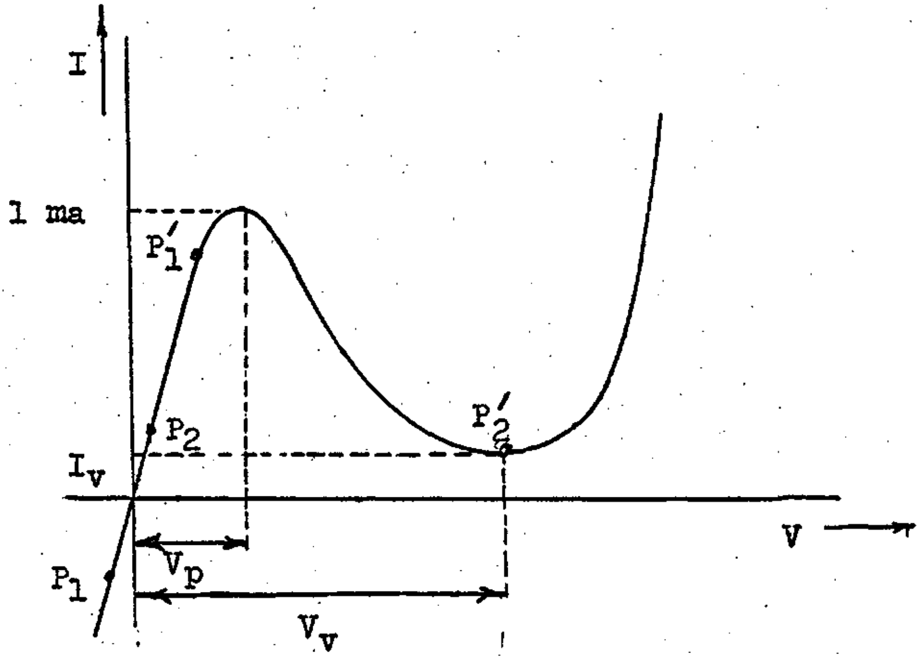


Fig. 23* Operation of the comparator and pulse generator.

If the sum of the input currents I_{in} and $I_{feedback}$ is positive before the application of the clock pulse, then TD2 will be biased at point P_2 and TD1 will be biased at point P_1. When the clock pulse is applied the current through TD2 increases past the peak point and comes to rest at P_2' on the IV characteristic. The current through TD1 increases but cannot pass the peak point and, therefore, comes to rest at point P_1' on the characteristic. To ensure this operation, each clock supply voltage must be $\frac{V_v + V_p}{2}$. Since V_p is much less than V_v the output voltage is very nearly equal to the supply voltage. The output voltage is thus the same polarity as the input current and the sensitivity is determined by the minimum bias current required to overcome hysteresis.

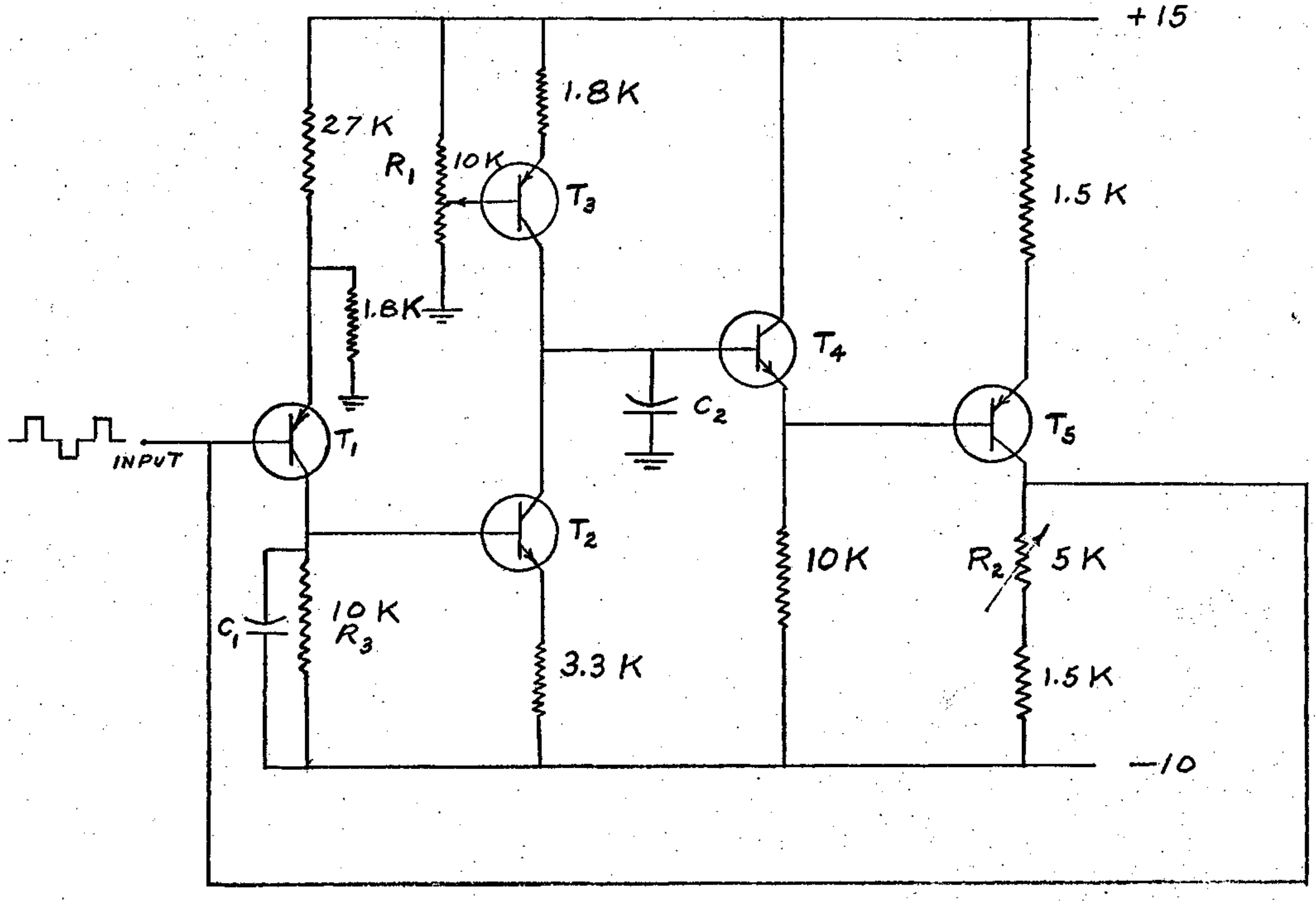
* As in reference 19.

From the midpoint of the tunnel diodes, the pulses are sent along the transmission path and to the feedback loop. The feedback loop which is an integrator and amplifier is shown in Fig. 24.

The pulses from the tunnel diode pair are applied to the base of T_1. With C_1 removed from the collector of T_1 as in the case of single integration delta modulation, this transistor acts as an amplifier. With C_1 in place, it also acts as an integrator with time constant C_1R_1. From the collector of T_1 the signal is fed to the base of T_2. T_2 and T_3 are both current sources that draw and feed equal d.c. current from capacitor C_2.

If a negative pulse is applied to the base of T_2, the current through T_2 decreases and charge is added to C_2 via T_3. If a positive pulse is applied to T_2, then the current increases through T_2 and charge is drawn from C_2 via T_2. The output from the integrating capacitor C_2 is taken to the base of T_4 which is in the emitter follower configuration. The last stage is T_5, which amplifies the feedback current. The feedback current which is fed back to the centre of the tunnel diodes is taken from the collector of T_5.

Potentiometer R_1 is used to make the current through T_3 equal to the current through T_2 in the quiescent state. Variable resistor R_2 is used to adjust the d.c. feedback current to the tunnel diodes so that under condition of zero input, the output of the tunnel diodes is alternately positive and



T_1, T_3, T_5 - 2N1307 C_2 - .39 μf
T_2, T_4 - 2N1306

Fig. 24 Feedback integrator and amplifier.

negative.

This circuit, although it does not have an infinite memory, does perform the required function. Ideally the circuit should integrate a d.c. quantity, but problems with drift make this impractical. In order to increase the d.c. stability of the circuit, it does not integrate a signal with a frequency less than 2.7 cycles/second. Using a circuit with this characteristic does not adversely affect the performance of the transmitter, but it does mean that the receiver must have an integrator with the same characteristic.

When second order integration is desired in this circuit, capacitor C_1 is added. This capacitor is chosen so that $\tau = R_1C_1$ is equal to the time interval between two pulses. The amplitude versus frequency characteristic of the circuit with C_1 in place is shown in Fig. 25, where $f_2 = \frac{1}{2\pi\tau}$

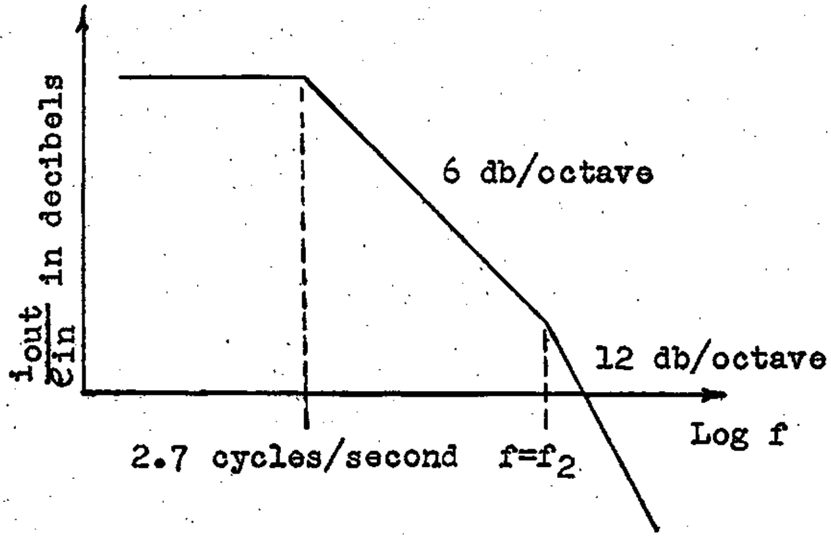


Fig. 25 Frequency response of the integrator and feedback amplifier.

This characteristic is not the same as de Jager (16) has given as the most desirable (Fig. 19) but gives essentially the same performance. The characteristic given by de Jager has the 12 db/octave slope starting at ω_0 which is the highest signal frequency and then returning to a 6 db/octave slope at f_2. However, for a maximum signal frequency of 200 cycles/second and a pulse repetition frequency of 2000 P/second, the break points ω_0 and ω_1 in Fig. 21 are so close together that the characteristic has almost a constant slope of 6 db/octave, which is the characteristic of single integration. In order to give second order integration and still maintain stability, the characteristic shown in Fig. 23 has been used.

3.3.8 Method of Transmission and Receiving

Although most analysis on delta modulation has been done assuming both polarity of pulses are present at the receiver, it has been pointed out that it is necessary to transmit only one polarity of pulse. To implement this method of transmission the receiver must assume that the opposite polarity of pulse is present whenever the pulse being sent does not arrive. In practice, this is done by placing a charge Q in the receiving integrator between pulses and removing 2Q when a pulse arrives.

In order to simulate as closely as possible the conditions of transmission found in actual use, transmission using only one polarity of pulses was used during the tests that were carried out. To extract one polarity of pulse from the delta modulation circuit that has been described, the circuit shown in Fig. 26 was used.

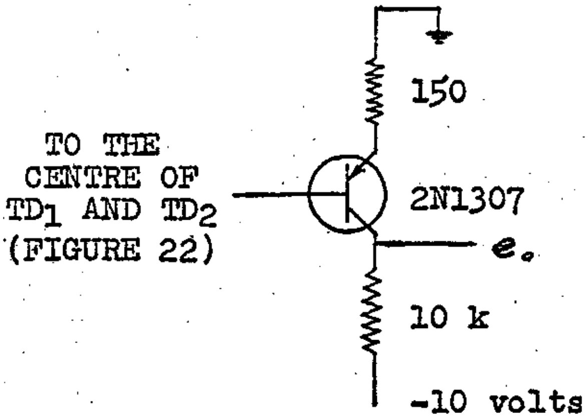


Fig. 26 Negative Pulse Extractor.

With zero or a positive input to the circuit in Fig. 26, the output remains at -10 volts. When a negative pulse input of 200 mv arrives at the input, the transistor turns on and the output rises to almost zero volts. The pulses are taken from the collector and fed to a pulse generator which regenerates the received pulse and feeds them to the receiving integrator. The receiving integrator is shown in Fig. 27.

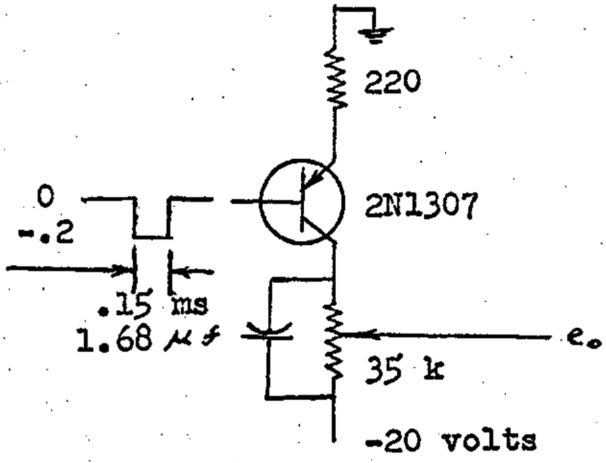


Fig. 27 Receiving integrator.

When a pulse arrives from the pulse generator, the transistor is turned on for .15 milliseconds producing a voltage change at the collector (100 mv) which is twice the voltage change at the transmitting integrator. Between pulses current is drawn from the capacitor via the 35 K resistor. The RC decay time constant of this integrator has been chosen to equal that of the transmitting integrator. The output voltage from the receiver is taken to a pen recorder which also serves as the low pass filter with a cutoff frequency of 160 cycles/second at the receiver.

3.3.9 Test Results

In order to have a close look at the rather complex waveforms of an ECG before and after transmission, some type of a standard ECG signal was desired so that comparisons of the received signal could be made under different conditions of coding. To generate a standard ECG a photoelectric waveform generator was used. The arrangement of the elements of the generator is shown in Fig. 28.

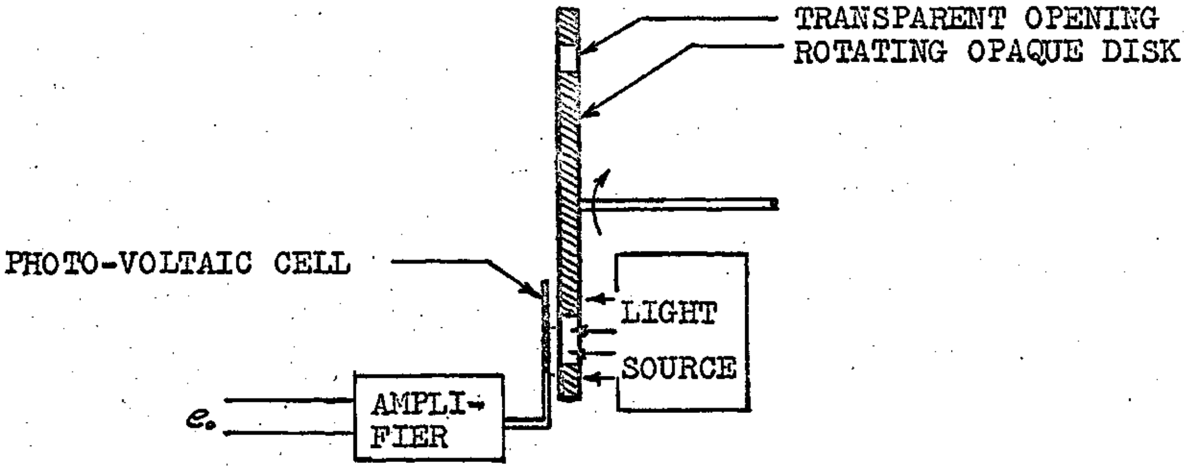


Fig. 28 Photo-electric waveform generator.

With the elements arranged as shown above, the width of the opening in the rotating disk is made proportional to the amplitude of the desired electrical signal output. The photo voltaic cell is masked except for a narrow slit that is parallel to the radius of the rotating disk which prevents erroneous signals from light sources within the room. The disk is rotated by a variable speed motor which enables one to set the rate at which one complete ECG signature is generated. The ECG signal that was generated by this method and used during the delta modulation tests as a standard input is shown in Fig. 29.

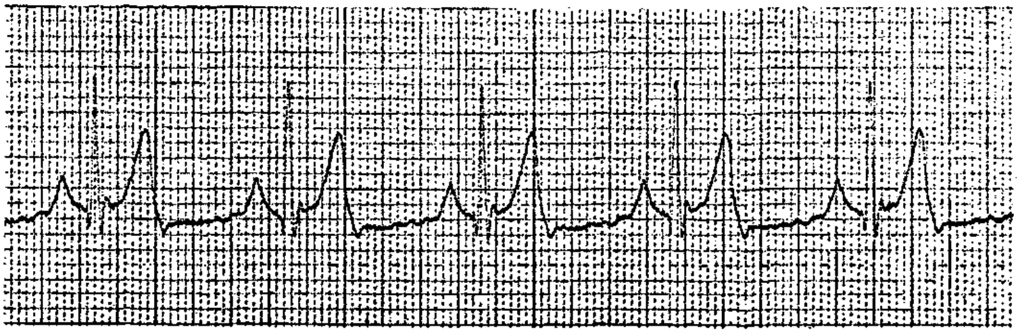


Fig. 29 The ECG generated by the photo-electric waveform generator.

The calculations done previously indicated that a pulse repetition frequency of about 2700 pulses/second would be adequate for the transmission of ECG's. For this reason tests were run using pulse repetition frequencies above and below this figure to note the changes in the waveform fidelity. Tests were run using pulse repetition frequencies of 1500, 2000, 2500, 3000, 3500 and 4000 pulses per second.

In the test results shown in Figs. 30 to 35, the step size at the integrator was held constant at 50 mv. To vary the number of steps in the encoded signal, the input to the modulator was varied accordingly. The scale factor used in these tests is that the ECG before amplification was 2 mv peak to peak. The allowable number of steps peak to peak was then found by using this figure along with equation (6) and a maximum slope as established in section 3.1 of 314 mv/second. The number of steps from peak to peak and the pulse repetition frequency are indicated on the following examples. In all the examples shown, the uppermost trace is a 1 second time marker; below that is the received signal and the bottom trace is the ECG before encoding.

A close examination of the records shown in Figs. 30 to 35 shows that below a pulse frequency of 3000 pulses/second, minor distortions in the received waveform can be detected. At a pulse frequency of 2000 pulses/second and below, there is quite noticeable distortion and noise. At 3000 pulses/second and above, there is no distortion apparent to the naked eye. This indicates that for visual diagnosis of an ECG, a pulse frequency of at least 3000 pulses/second is required. To send this pulse frequency then requires an upper cutoff frequency greater than 3000 cycles/second.

When using frequency modulation an upper frequency of 2900 cycles/second is required when using a carrier with a centre frequency of 2000 cycles/second and ± 40% deviation. Although the upper frequency for delta modulation and frequency modulation are comparable, frequency modulation does make more efficient use of the magnetic tape. For this reason, a frequency modulated carrier with a centre frequency of 2000 cycles/second has been used for writing the ECG on the magnetic tape.

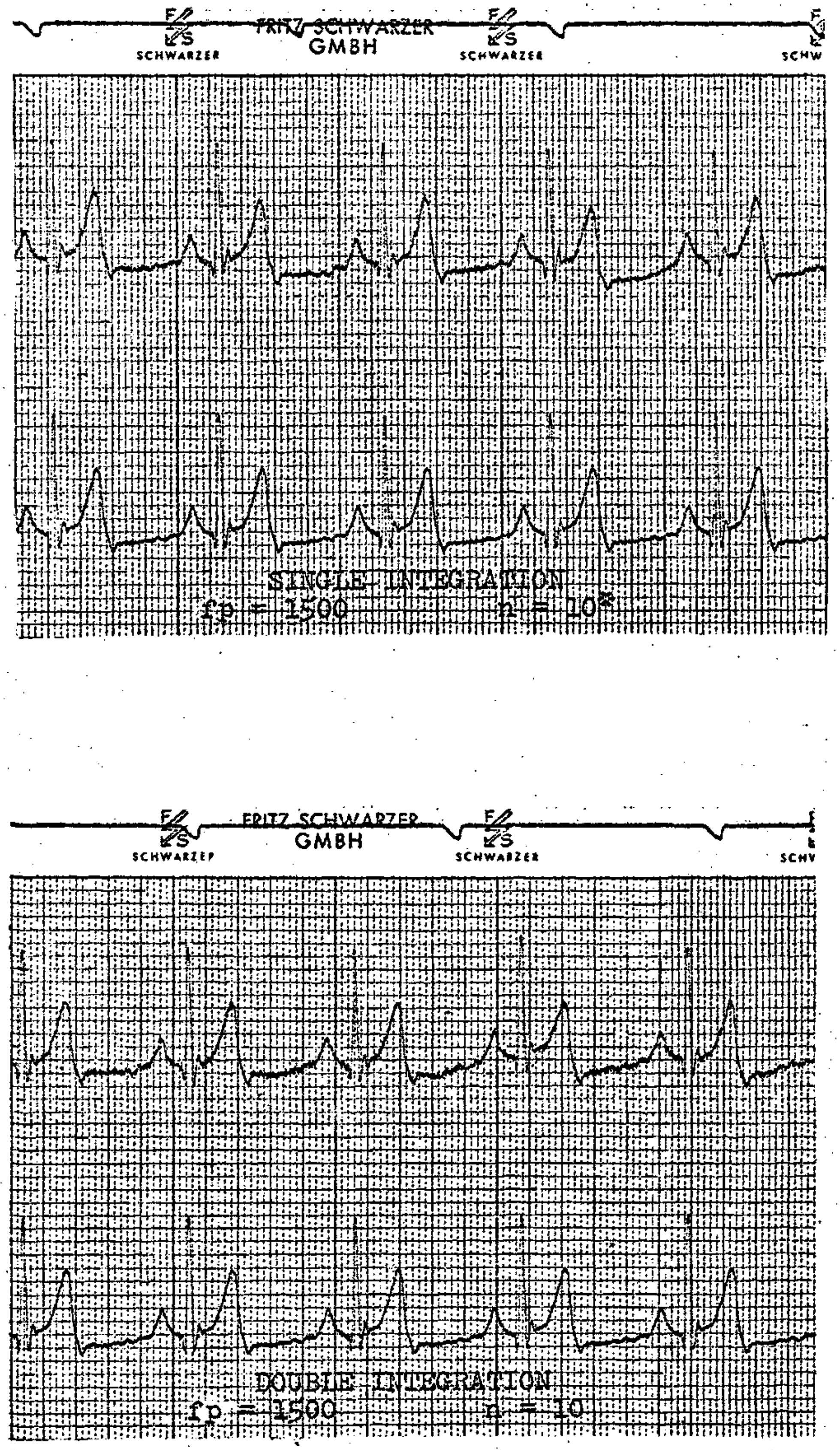
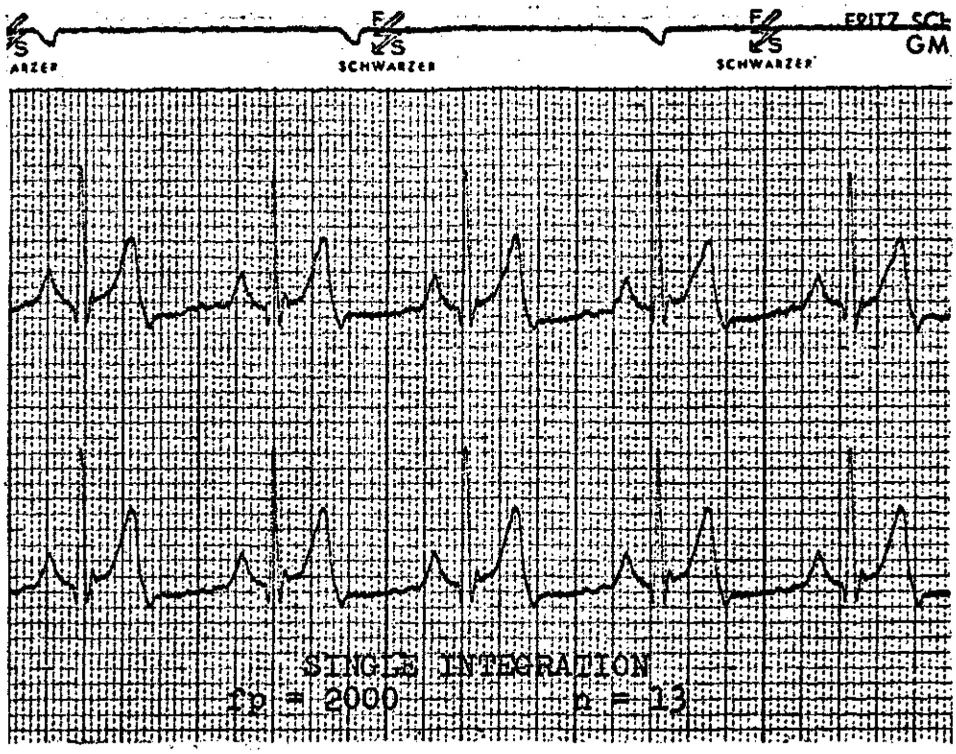


Figure 30

* n is the number of steps from peak to peak.


FRITZ SCHWARZER GMBH
SINGLE INTEGRATION
fp = 2000
n = 13

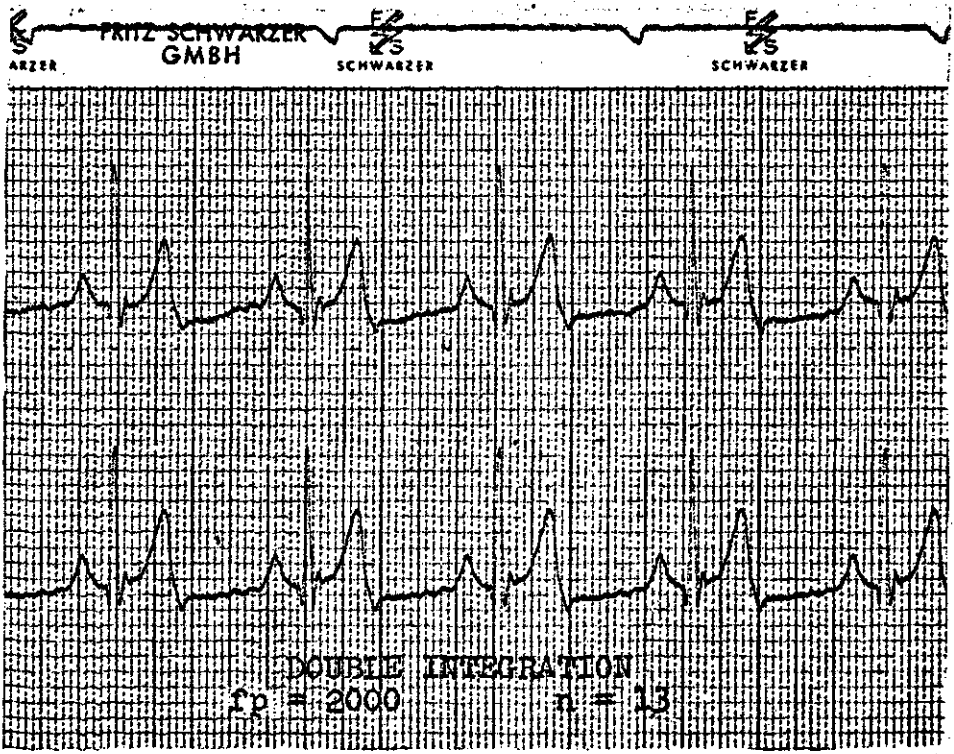
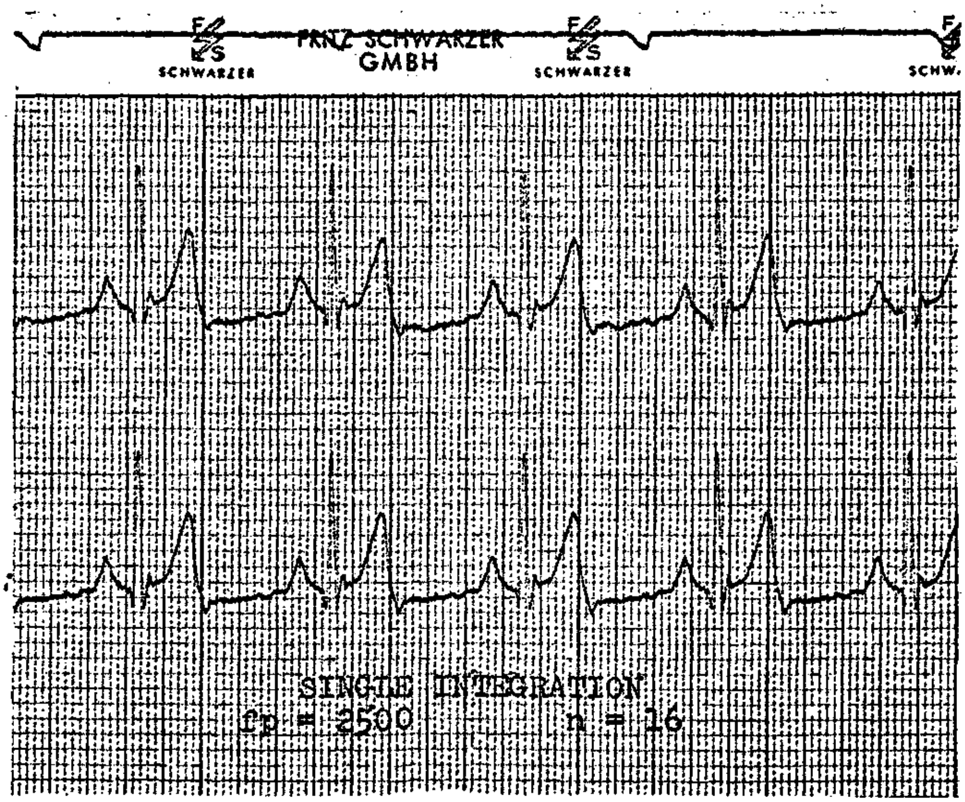
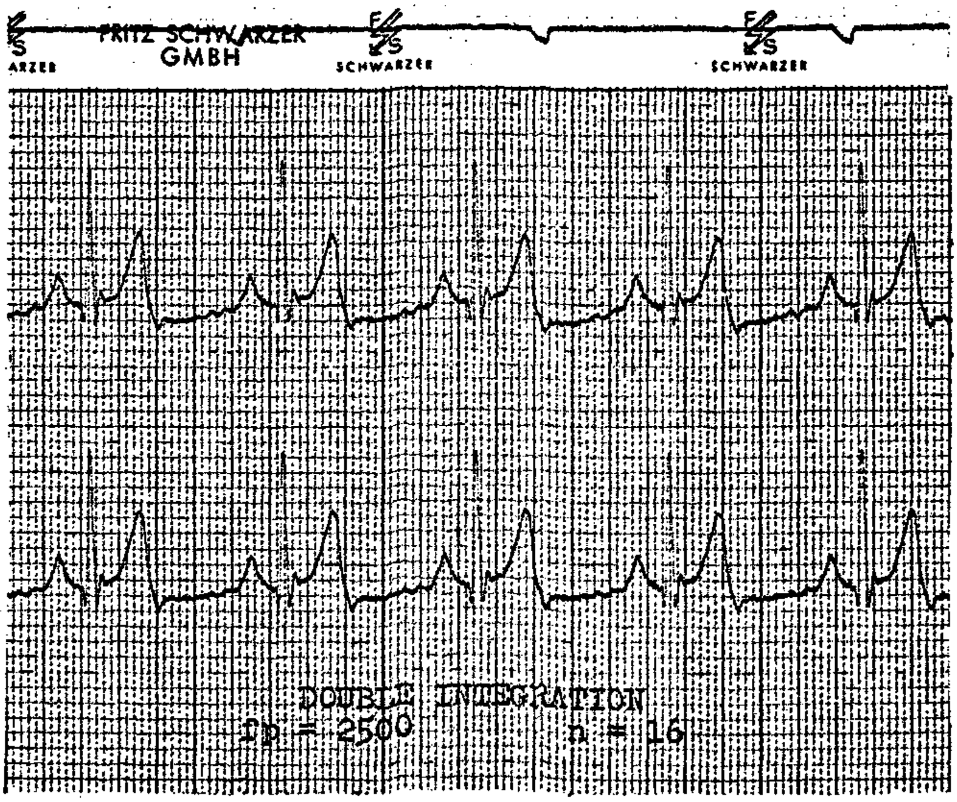

FRITZ SCHWARZER GMBH
DOUBLE INTEGRATION
fp = 2000
n = 13

Figure 31



SINGLE INTEGRATION
fp = 2500 n = 16



DOUBLE INTEGRATION
fp = 2500 n = 16

Figure 32

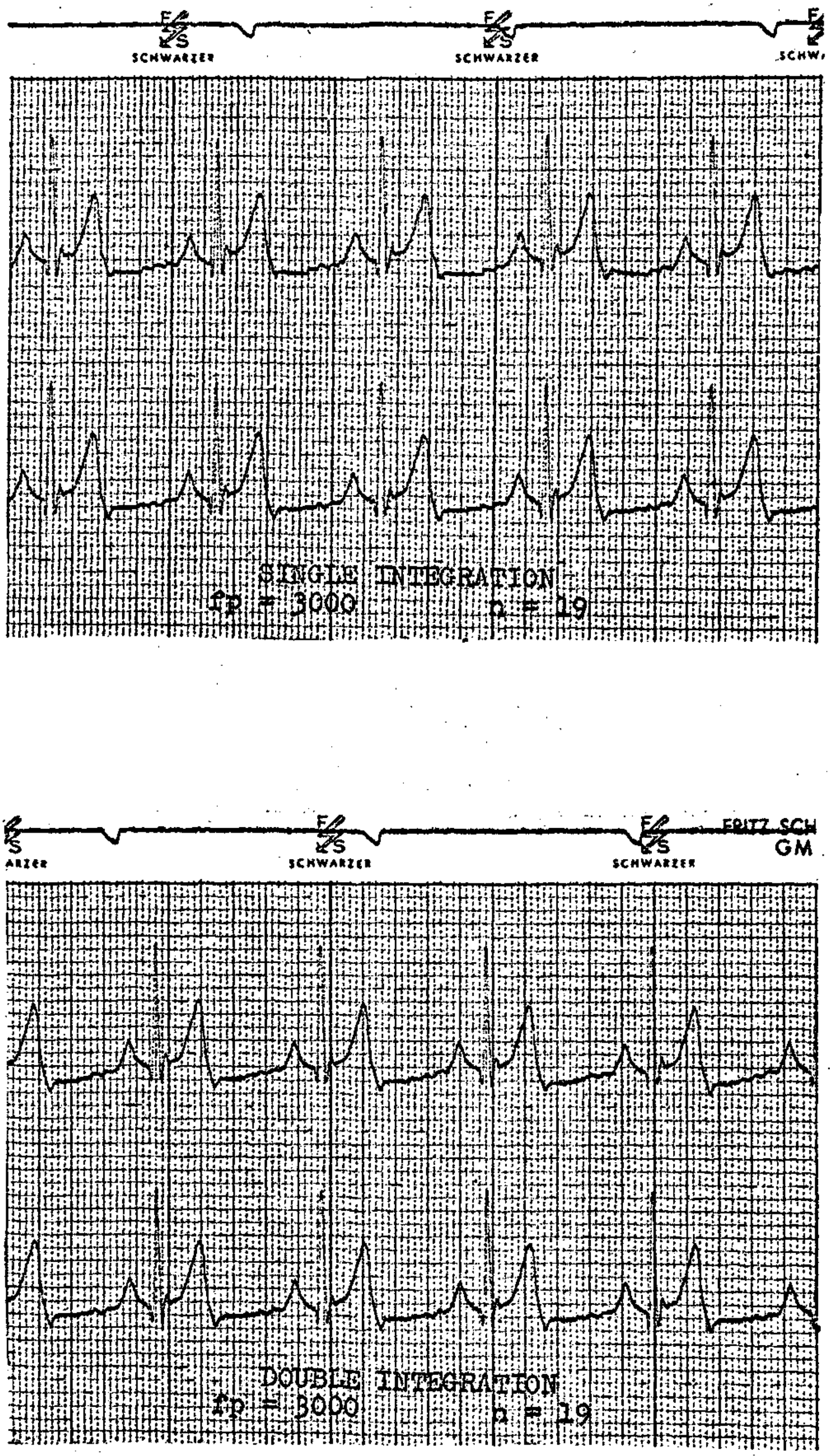
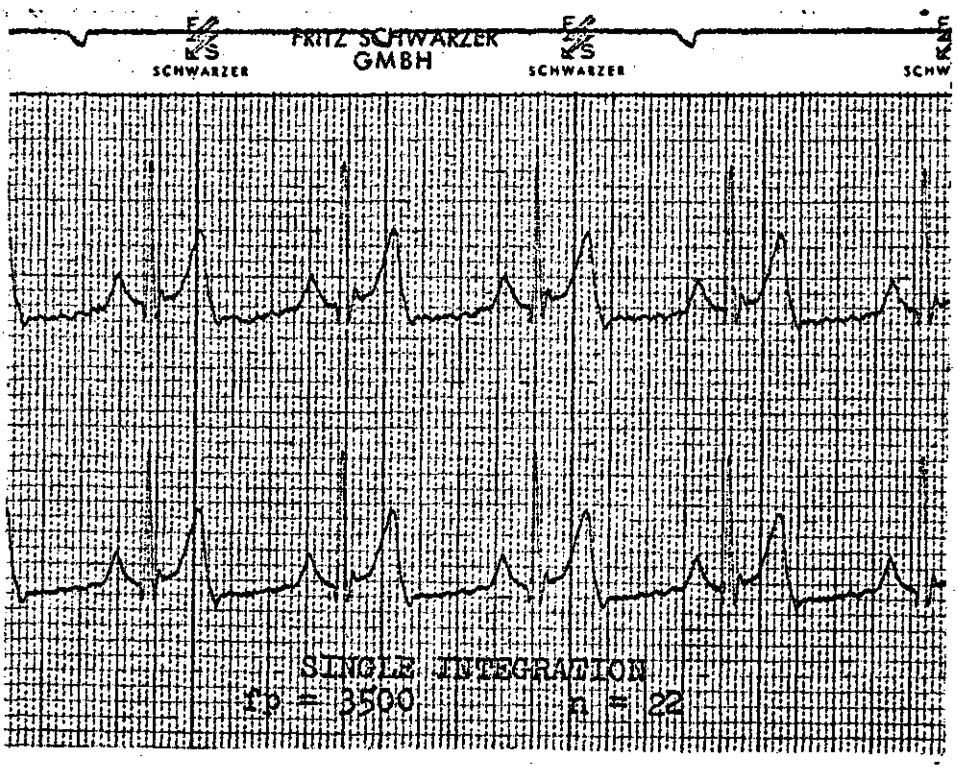

SCHWARZER
SCHWARZER
SCHWARZER
SINGLE INTEGRATION
fp = 3000
n = 19
FRITZ SCH
GM
SCHWARZER
SCHWARZER
SCHWARZER
DOUBLE INTEGRATION
fp = 3000
n = 19

Figure 33


FRITZ SCHWARZER
GMBH
SCHWARZER
SCHWARZER
SCHW
SINGLE INTEGRATION
fp = 3500
n = 22

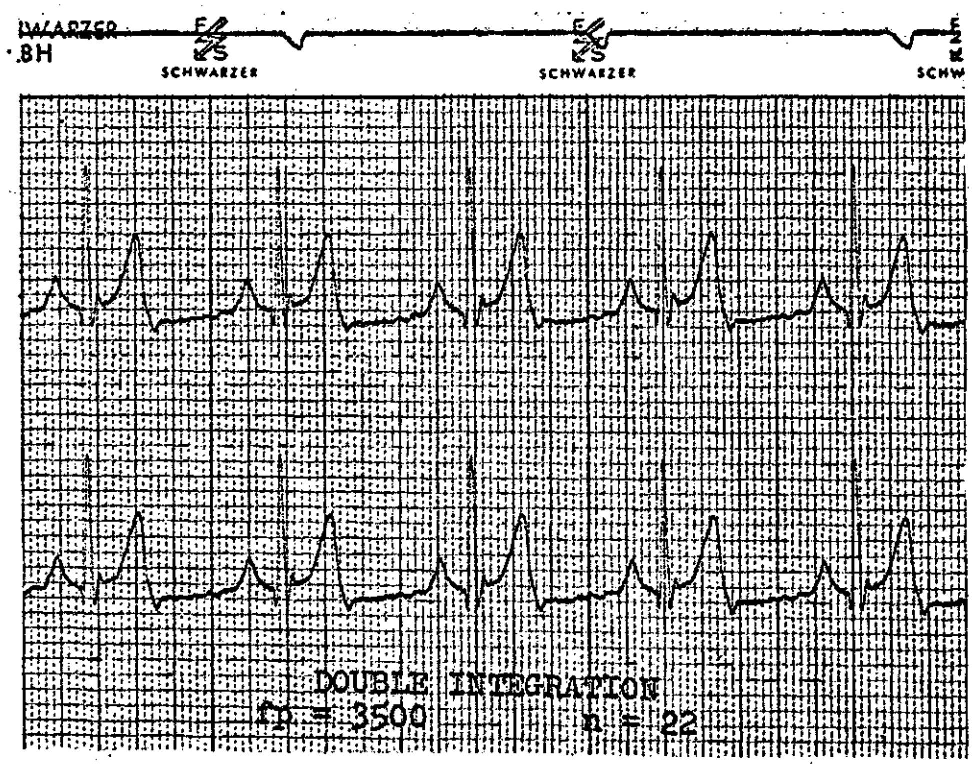

SCHWARZER
.BH
SCHWARZER
SCHWARZER
SCHW
DOUBLE INTEGRATION
fp = 3500
n = 22

Figure 34

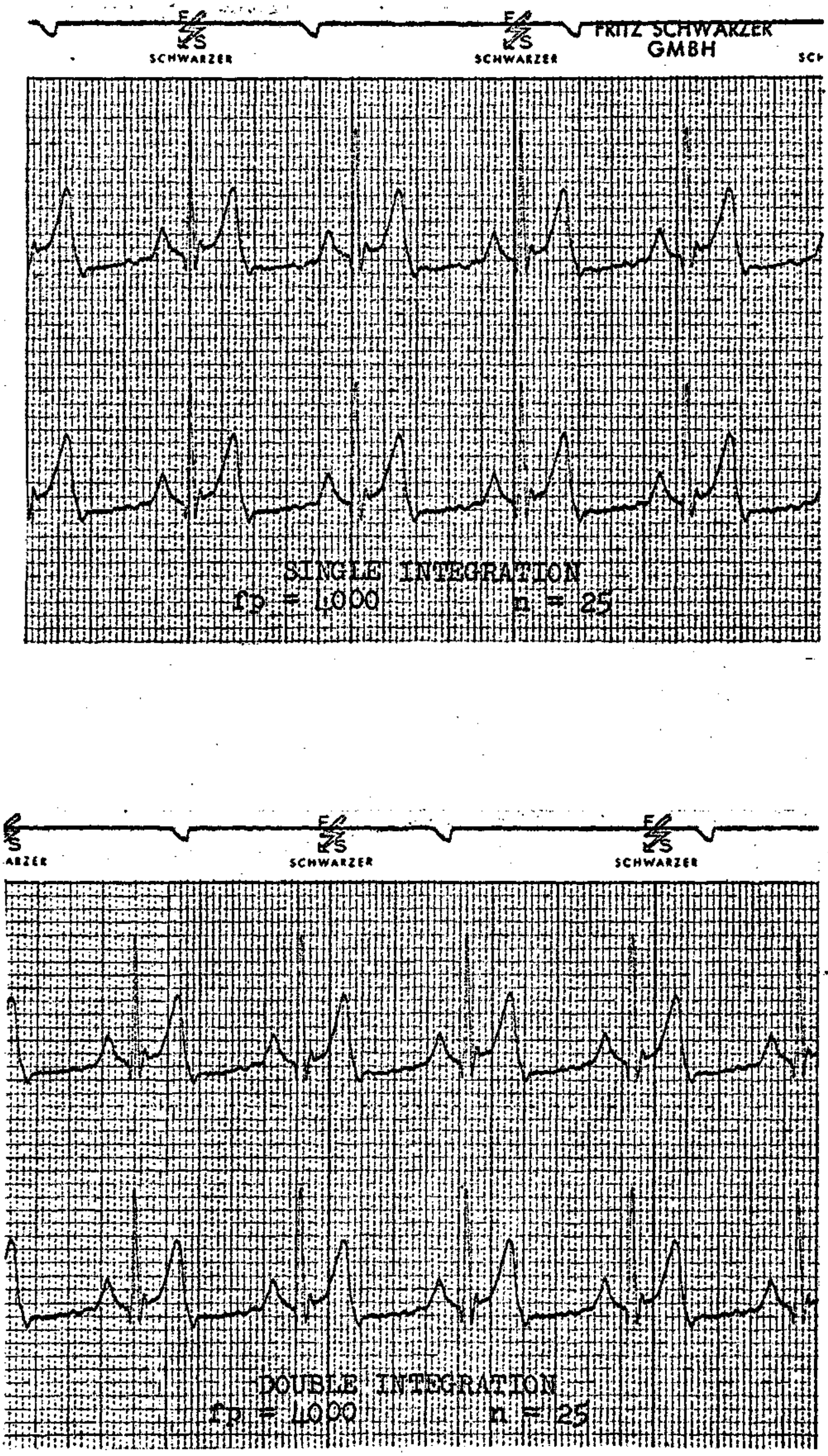

SCHWARZER
FRITZ SCHWARZER GMBH
SINGLE INTEGRATION
fp = 1000
n = 25
SCHWARZER
DOUBLE INTEGRATION
fp = 1000
n = 25

Figure 35

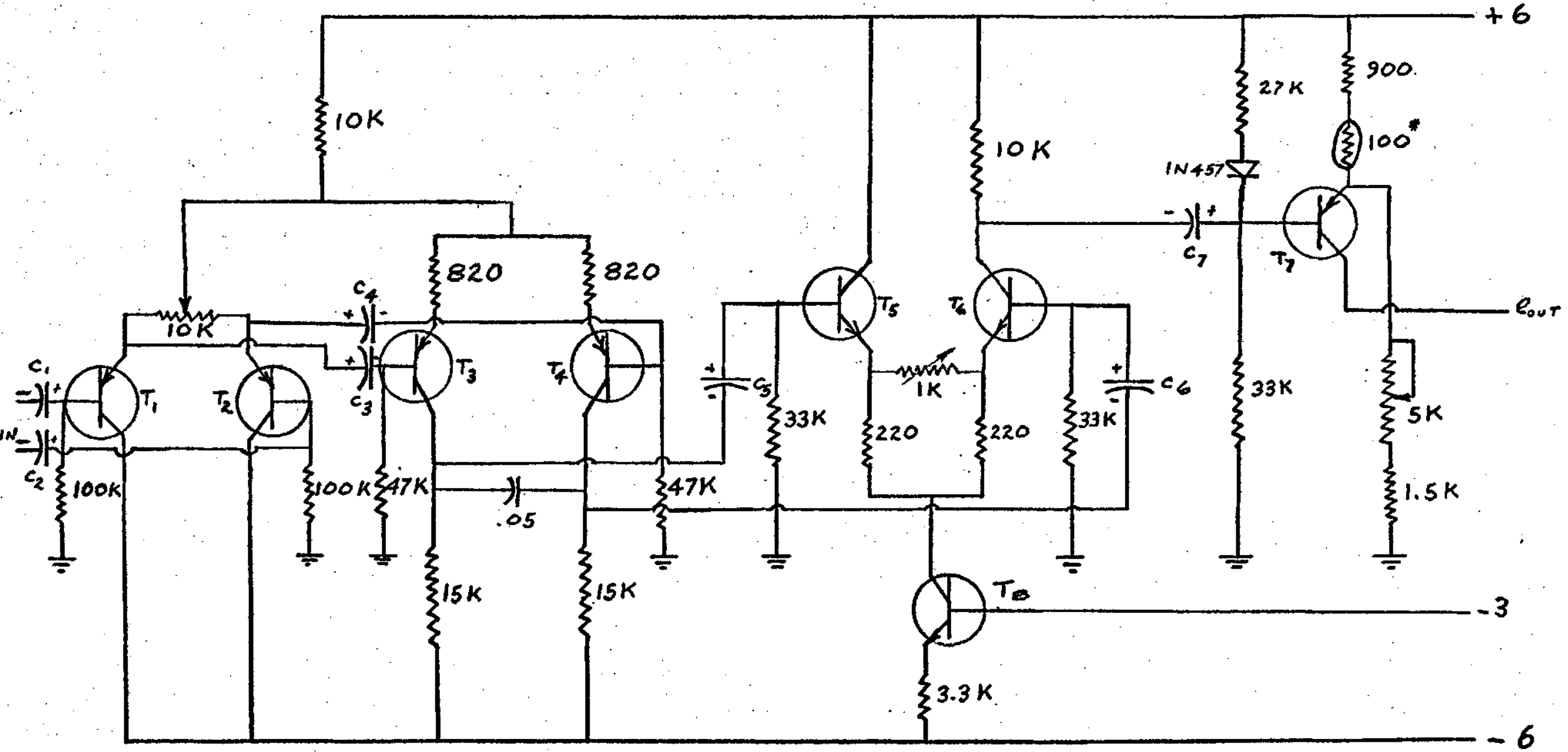
3.4 The Electronic Circuits

3.4.1 The Record Electronics

The ECG as taken from skin electrodes is a voltage with a peak to peak amplitude of from 1 to 5 millivolts from a source impedance of typically 5 kilohms. As well as the ECG signal from the electrodes, there will also be noise due to muscle movement. Along with the noise there will also be a change in impedance between two electrodes if placed on the chest due to expansion and contraction of the chest during the breathing cycle. To minimize the variations in impedance and the muscle noise, a differential amplifier with a large input impedance and a high common mode rejection is desirable. Also, because there are d.c. potentials on the surface of the skin, the amplifier must be a.c. coupled to the electrodes. The voltage gain required in the amplifier to give the required frequency shift in the carrier oscillator is approximately 1000.

The circuit diagram of the amplifier used to meet the above requirements is shown in Fig. 36. This circuit is similar to the one described by Buckley (2). Bias resistors have been changed to operate from the supply levels shown, and an additional stage (T_7) has been added to give the required gain. As well, a capacitor between the collectors of T_3 and T_4 has been added to limit the bandwidth of the amplifier.

Transistors T_1 and T_2 make up the differential emitter follower that is required to give a high input impedance. From the emitters of T_1 and T_2, the signal is a.c. coupled to



T_1, T_2, T_3, T_4 - PHILLIPS BCZ-14
T_5, T_6, T_7, T_B PHILLIPS BC-109
C_1 TO C_7 = 10 μF
*TEXAS INSTRUMENT SENSISTOR

Fig. 36 ECG Amplifier.

the next differential stage made up of transistors T_3 and T_4. This stage which is in the common emitter configuration gives a gain of approximately 6. This stage as well as the first stage is biased from a common emitter resistor which results in improved common mode rejection.

Further common mode rejection is obtained in the final differential stage by feeding the emitters of T_5 and T_6 from the collector of T_8, which acts as a constant current source. It is in this final differential stage that the differential to single ended signal conversion is made. Gain adjustment is also provided for in this stage in the form of the 1 kilohm variable resistor between the emitters of T_5 and T_6. From the collector of T_6 the signal in single ended form is a.c. coupled to the final stage of amplification. The signal from the collector of T_7 is then directly coupled to the carrier oscillator. The quiescent current in this final stage and, therefore, the centre frequency of the carrier oscillator is adjusted by means of the 5 kilohm variable resistor from the emitter of T_7 to ground. The adjustment to obtain maximum common mode rejection in this amplifier is the potentiometer between the emitters of T_1 and T_2.

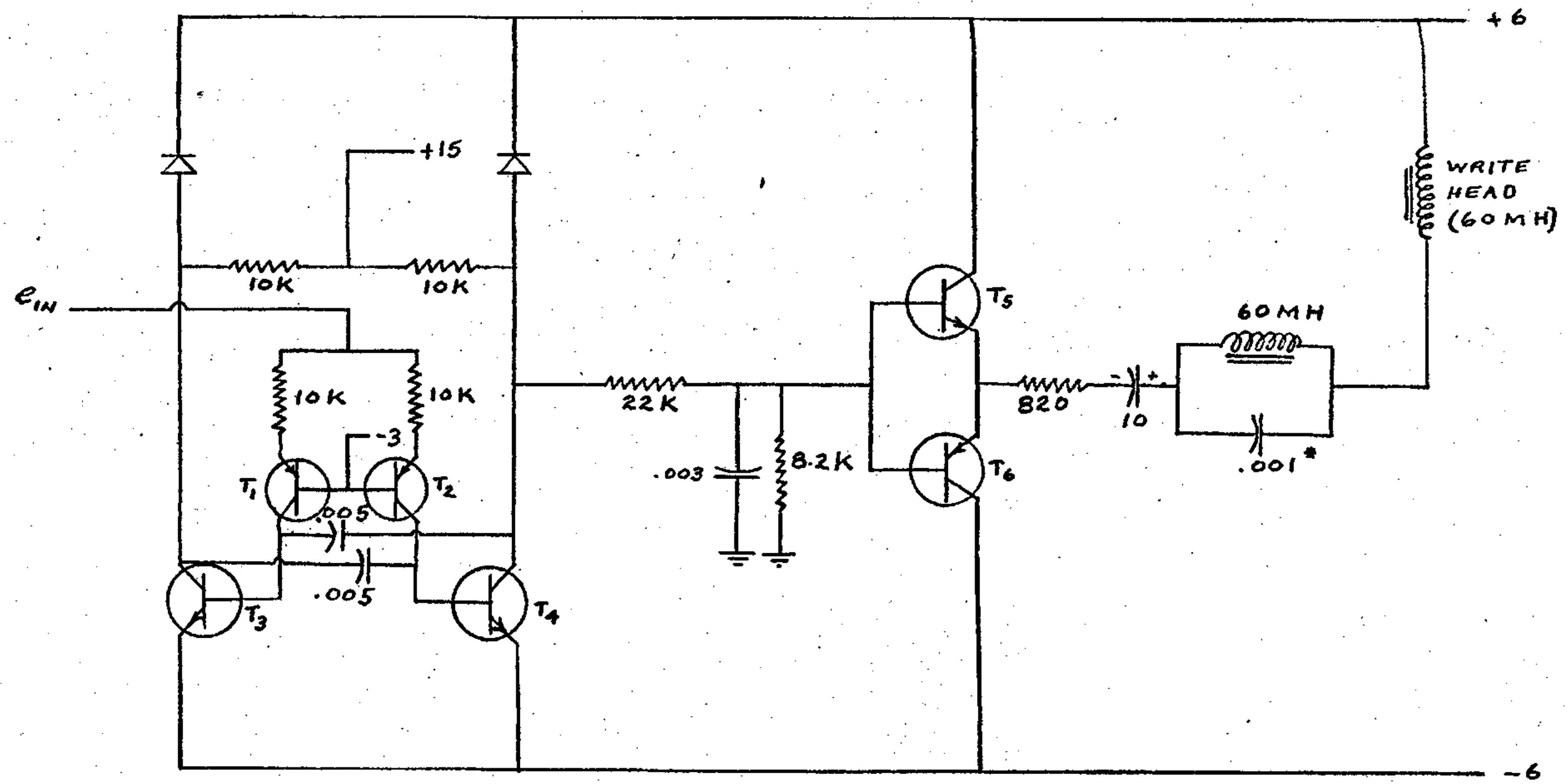
The characteristics of this amplifier may be summarized as follows:

Gain	435 to 925
Common Mode Rejection	greater than 5×10^3
Input Impedance	150 K
Bandwidth (3 db points)	0.45 cycles/second and 200 cycles/second

From the ECG amplifier the signal then passes on to the carrier oscillator. The function of this circuit is to convert the incoming voltage to a proportional frequency so that the information may be recorded on the magnetic tape. The voltage controlled oscillator that is used for this purpose is shown in Fig. 37. The circuit has the same configuration as the one described by Biddlecomb (20). It has, however, been redesigned to give the required centre frequency of 2000 cycles/second when operating from the voltage supplies as indicated.

In this circuit the input voltage is applied to the two 10 kilohm resistors that are connected to the emitters of T_1 and T_2. The bases of T_1 and T_2 are biased by a low impedance voltage source. In this way the input voltage is divided into two equal currents at the collectors of T_1 and T_2. The remaining part of the circuit is a conventional cross coupled multivibrator that oscillates at a frequency that is primarily determined by the current supplied from the collectors of T_1 and T_2 and the size of the timing capacitors.

The circuit originally described (Biddlecomb, 20) used zener diode bias supply regulation thus making the circuit insensitive to supply voltage changes. However, since we were interested in



T_1, T_2 - PHILLIPS BCZ-14
T_3, T_4 - PHILLIPS BC-109
T_5 - 2N1306
T_6 - 2N1307

ALL CAPACITIES IN μf
* PADDED TO GIVE RESONANCE WITH 60 MH COIL AT 20 KC/S

Fig. 37 Carrier oscillator and head drive circuit.

using several of these oscillators in the same package, it was more economical powerwise to provide well regulated supplies and run all the circuits from the same supplies and delete the zener diodes.

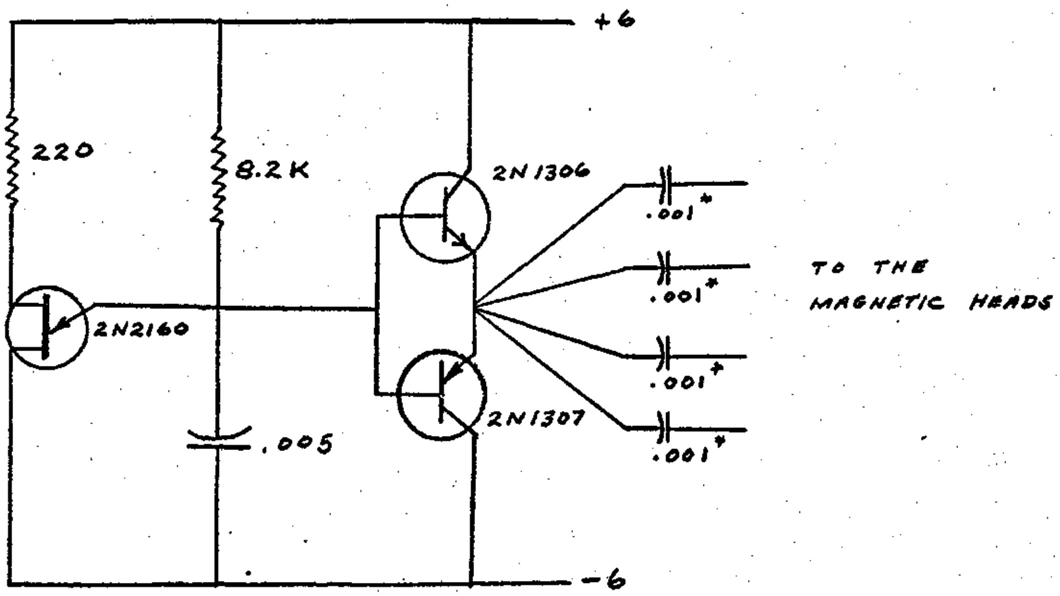
The centre frequency stability of this oscillator is determined by the stability of the quiescent current supplied by the preceding direct coupled amplifier stage (T_7 in Fig. 36), the inherent changes in the circuit parameters due to temperature change and the stability of the supply levels. To minimize the drift due to changes in battery voltage, voltage regulators are used on the supply levels that are relatively independent of battery voltage and temperature.

To assess the change in centre frequency due to quiescent current and circuit parameters changing with temperature, the circuit was tested in a temperature controlled chamber. As the temperature of the chamber was raised from 0°C to 50°C, the centre frequency also increased in a very nearly linear fashion. To compensate for this increase in frequency with temperature, a temperature sensitive resistor (sensitor) was placed in series with the fixed emitter resistor of T_7 (Fig. 36). This sensistor which has a positive temperature coefficient regulates the quiescent current to maintain a stable centre frequency. Final tests showed that the frequency drift due to a temperature change from 0°C to 50°C was less than 2% of full scale deviation, which is ± 40% of centre frequency.

From the collector of T_4 (Fig. 37) the square wave generated by the carrier oscillator passes on to the head drive circuit. This circuit provides the required power to drive the magnetic recording head. The required power is gained by using T_5 and T_6 in the complementary pair emitter follower configuration. The head impedance at the centre frequency of 2000 cycles/second is approximately 750 ohms inductive and 50 ohms resistive, and the recommended recording current for this head for saturation recording is 2 milliamps peak to peak. At carrier frequencies above 2000 cycles/sec. it was noted that there was an optimum recording current less than 2 milliamps peak to peak. This optimum recording current was set using the attenuator network preceding T_5 and T_6. To limit the current at the lowest carrier frequency, a resistor was inserted in series with the head and parallel tuned circuit.

The parallel tuned circuit is used to isolate the 20 kc high frequency bias from the carrier signal source. This is necessary as the 20 kc high frequency bias is fed to the head through a series tuned circuit and the carrier source must present a high impedance to the 20 kc bias if the Q of the head is to be maintained.

The 20 kilocycle/second high frequency bias is supplied by the oscillator shown in Fig. 38. This oscillator is a unijunction relaxation oscillator followed by a complementary pair emitter follower to provide the necessary power. A high



* PADDED TO GIVE RESONANCE WITH
THE MAGNETIC HEAD AT 20 KC/S

Fig. 38 High frequency bias oscillator.

frequency bias current of 2 milliamps peak to peak is supplied to the head by forming a series tuned circuit with the head.

The electronics just described is sufficient to record one ECG. The ECG amplifier, the carrier oscillator and the head drive circuit must be duplicated for each electrocardiogram required. To record signals other than an ECG (below 200 cps) a suitable signal conditioner must be used in place of the ECG amplifier.

A completed circuit board containing an ECG amplifier, carrier oscillator and head drive circuit is shown in Fig. 39. To obtain a high component density on this board, the components are mounted vertically on the printed circuit board. In order to make the circuit rugged as well as insensitive to small temperature differences, the circuit has been encapsulated in Sylgard 184.* This material is transparent and relatively easy to cut when cured, which makes the circuit easy to service even though it is encapsulated.

The power supply voltage regulator board shown in Fig. 40 was made using this same type of construction. The circuit diagram of the power supply voltage regulators is shown in Figs. 41 and 42. These voltage regulators were required as the nickel cadmium rechargeable batteries that provide the power to the electronics do not deliver a constant voltage during their entire discharge.

* Dow Corning Product

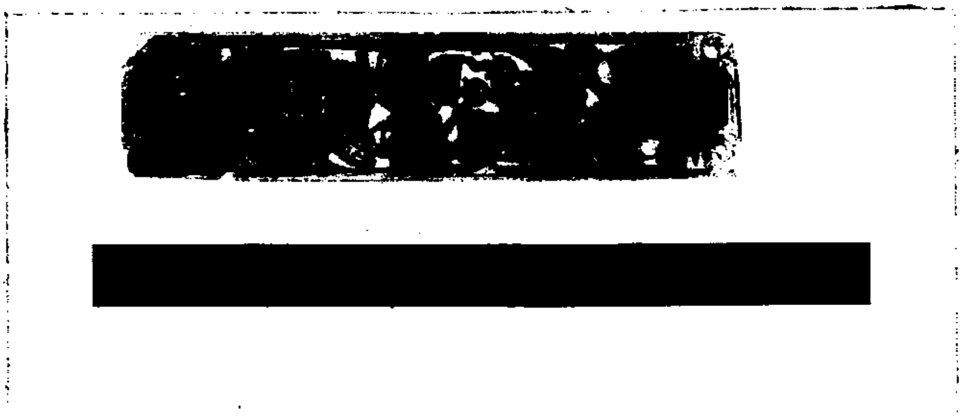


Fig. 39 The Record Electronics

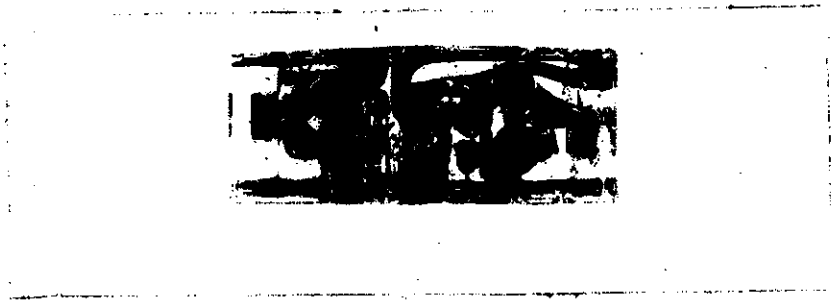


Fig. 40 The Power Supply Regulators

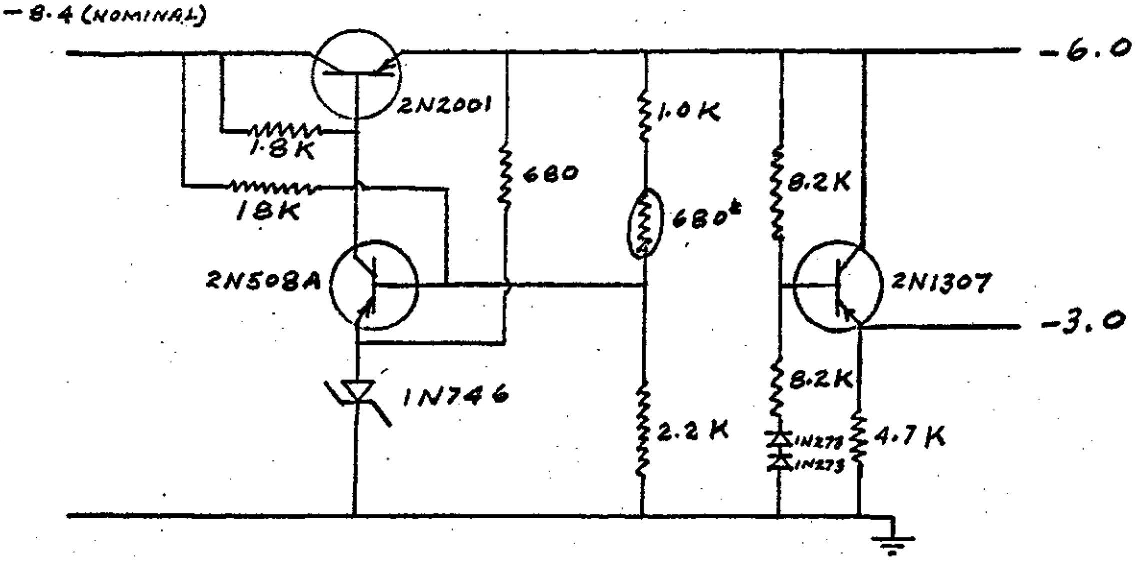
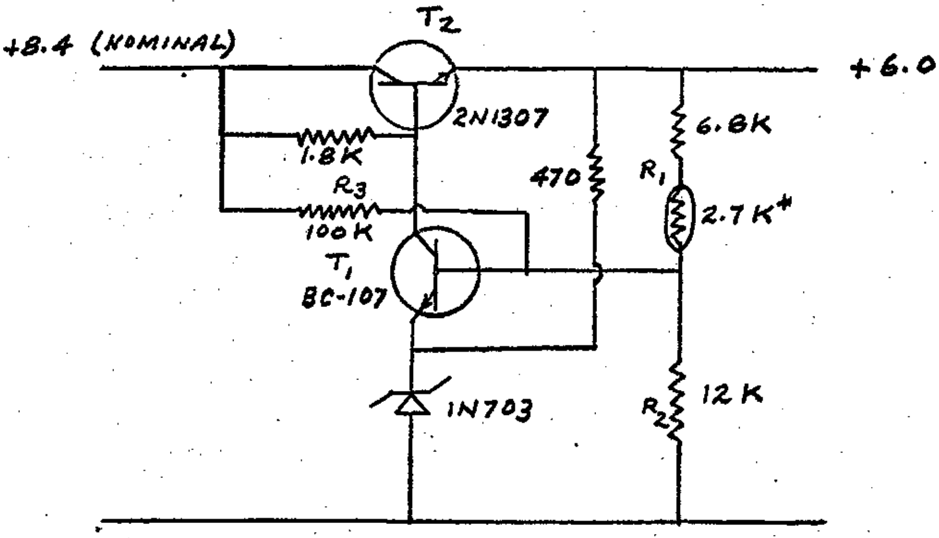


Fig. 41 -6 and -3 volt regulator.



* TEXAS INSTRUMENT SENSISTOR

Fig. 42 +6 volt regulator.

The voltage regulator is a series controlled regulator with a zener diode as a reference element and a 1 transistor feedback amplifier. With reference to Fig. 42, we see that the sampled output voltage from the centre of R_1 and R_2 is compared to the voltage on the zener diode. The difference is amplified by T_1 and fed to the base of T_2, the series control element. If the output voltage is too high, the current through T_1 increases and the voltage at the collector of T_1 decreases. This signal, applied directly to the base of T_2, the series control element, then decreases the output voltage to its proper value. If the output voltage tries to decrease, the opposite happens.

The feed forward resistor R_3 is used to cancel the effects of the imperfect reference element. As the input voltage increases, the current through T_1 and hence the zener diode reference must increase. Because the zener diode has a finite resistance, the reference voltage therefore increases resulting in a higher output voltage. To overcome this a small current is fed forward to the base of T_1. This further increases the current through T_1 as voltage increases which lowers the voltage at the collector of T_1 and, hence, lowers the output voltage to its proper value.

When the effects of temperature on this circuit were checked, it was found that the output voltage decreased as temperature increased. To compensate for this, part of R_1 was made temperature sensitive by using a sensistor. This decreased the sampled voltage as temperature increased and forced the output to remain constant.

The operation of the -6 volt regulator shown in Fig. 41 is the same as the +6 volt regulator just described. Also shown on the circuit diagram in Fig. 41 is the -3 volt regulator. This voltage level is created by using a voltage divider followed by an emitter follower to give a low output impedance.

Final tests on the plus and minus 6 volt regulators showed that under constant load and over a temperature range of from 0°C to 50°C and with the respective input voltages between 7.5 and 10.0 volts, the output voltage will vary less than 1% in the worst case combination. Under the same conditions, the -3 volt level varies less than 2%.

3.4.2 Time Delay Circuit

To ensure that the tape recorder is stopped 30 minutes after an event, thereby preserving 30 minutes prior to and following an event, a 30 minute time delay is used. This enables the patient to push a button when the event is occurring and the recorder will then automatically stop 30 minutes later.

The timing circuit and the switching network used to disconnect the batteries from the recorder is shown in Fig. 43. The time delay is initiated by removing the short from capacitor C_1. C_1 then charges towards +15 volts via resistor R_1. When C_1 charges to a voltage that will permit 1 milliamp to flow through FET* T_1 then TD1 changes to its high state and turns transistor T_2 on. T_3 is then turned on and a pulse is delivered to the coil of latching relay R1. This changes the state of the latched relay and the power is disconnected from the recorder.

Potentiometer R_2 is used to adjust the initial voltage on C_1 and, hence, the length of the time delay.

3.4.3 Playback Electronics

To reproduce the ECG that has been written on the magnetic tape the frequency modulated signal reproduced by the magnetic head must be amplified, limited, and then demodulated. The signal from the reproduce head has a peak to peak amplitude of approximately 1 millivolt at the lowest carrier frequency of 1200 cycles/second, and decreases to ½ millivolt at the upper frequency of 2800 cycles/second. The amplifier and limiter used to bring this signal up to a level that will operate the frequency demodulator is shown in Fig. 44.

Transistors T_1, T_2, and T_3 make up the direct coupled preamplifier. The gain of this amplifier is approximately 1000 with correction for the decrease in input amplitude as the frequency increases. This correction is accomplished with the emitter bypass capacitors of T_2 and T_3. To provide for variations in capacitance and resistance values, the bypass capacitor on the emitter resistor of T_3 is connected to the

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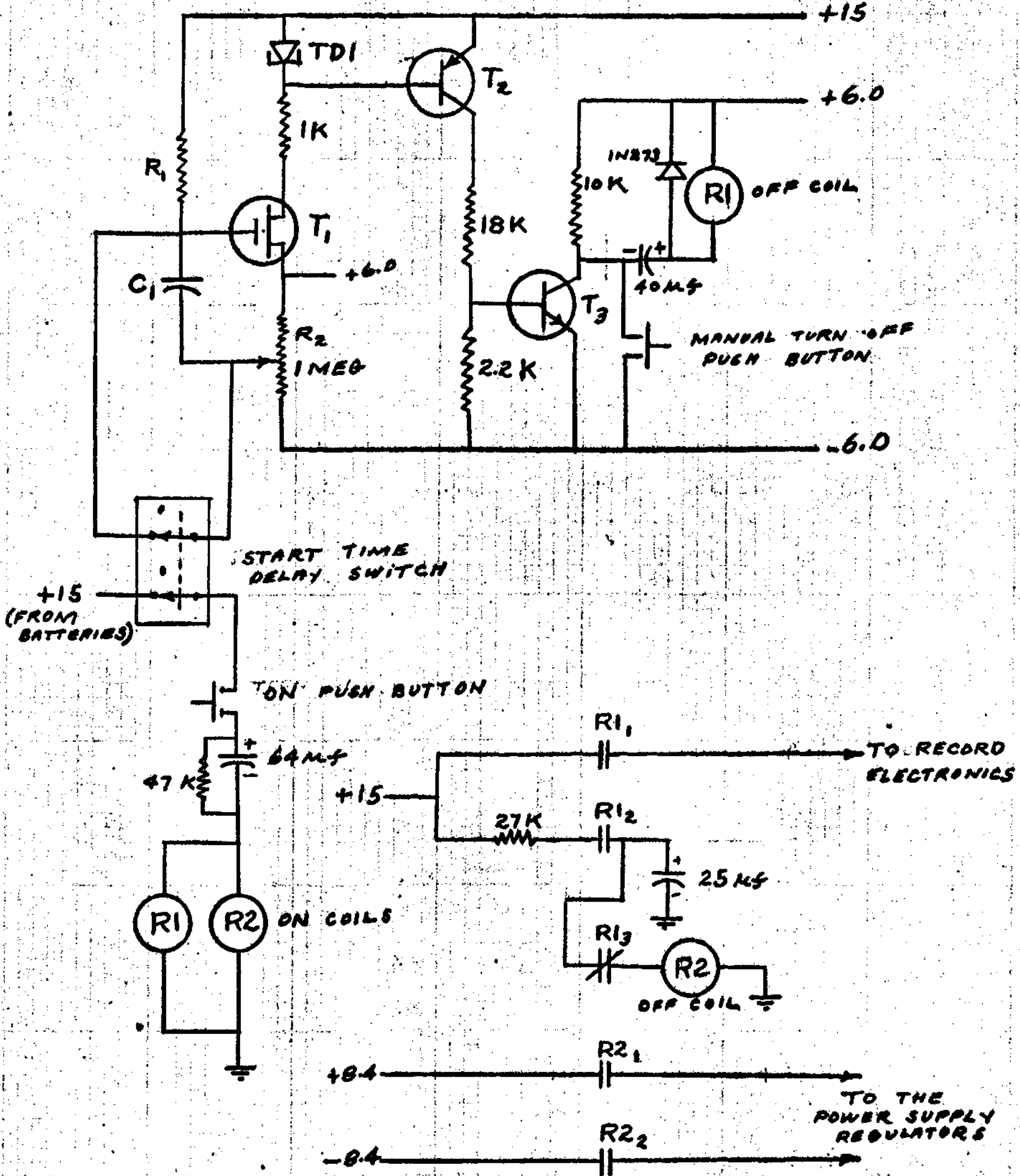
* Field effect transistor



$R_1 = 10^{11}$ OHMS

$C_1 = .04$ μf (MYLAR)

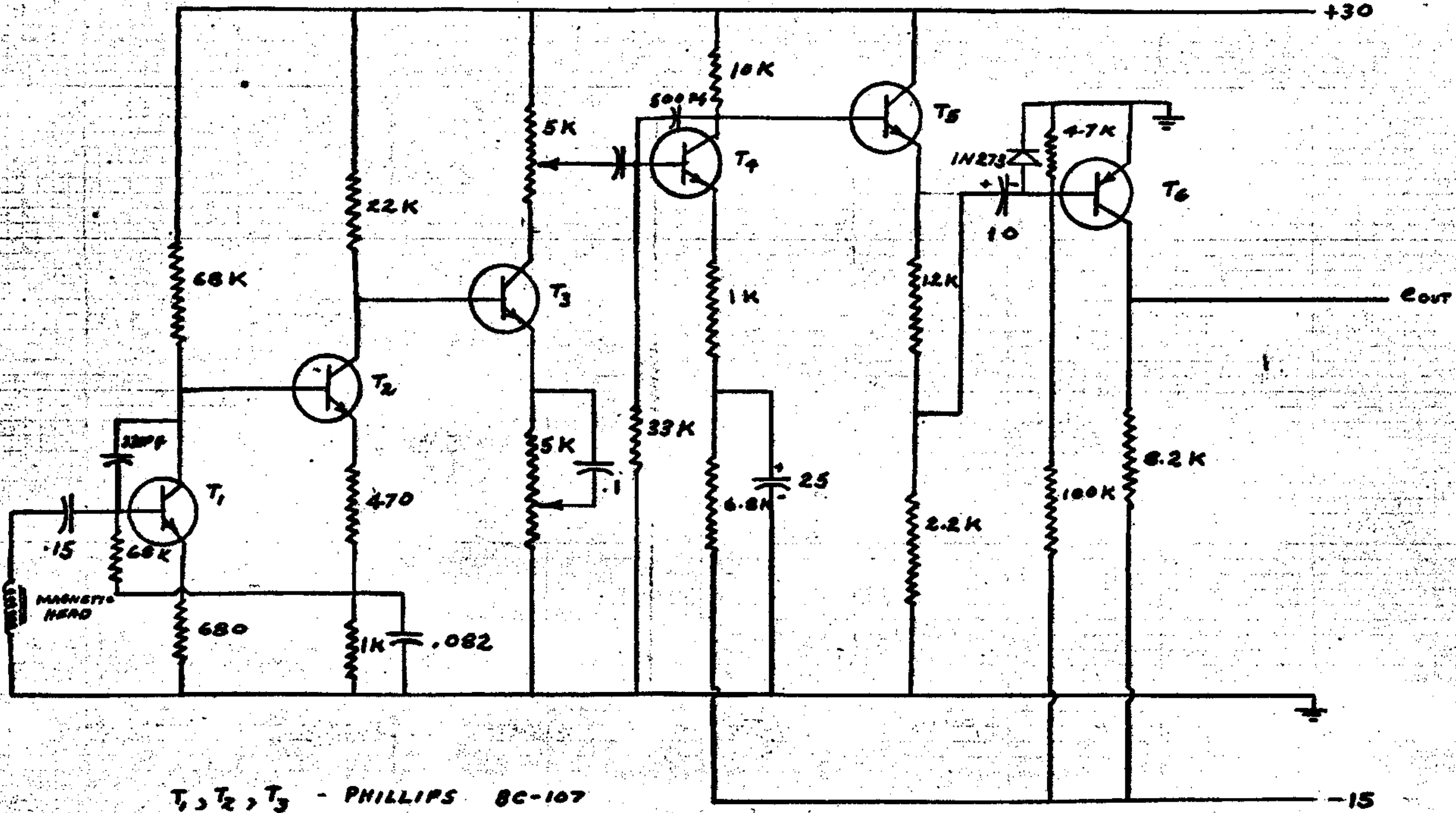
R1 AND R2 = TYPE FL LATCHING RELAY (POTTER AND BRUMFIELD)

T_1 = 2N3631

T_2 = 2N1307

T_3 = 2N1306

TD1 = 1N2939



T1, T2, T3 - PHILLIPS BC-107
T4, T5 2N1306
T6 2N1307
ALL CAPACITIES IN μf UNLESS STATED

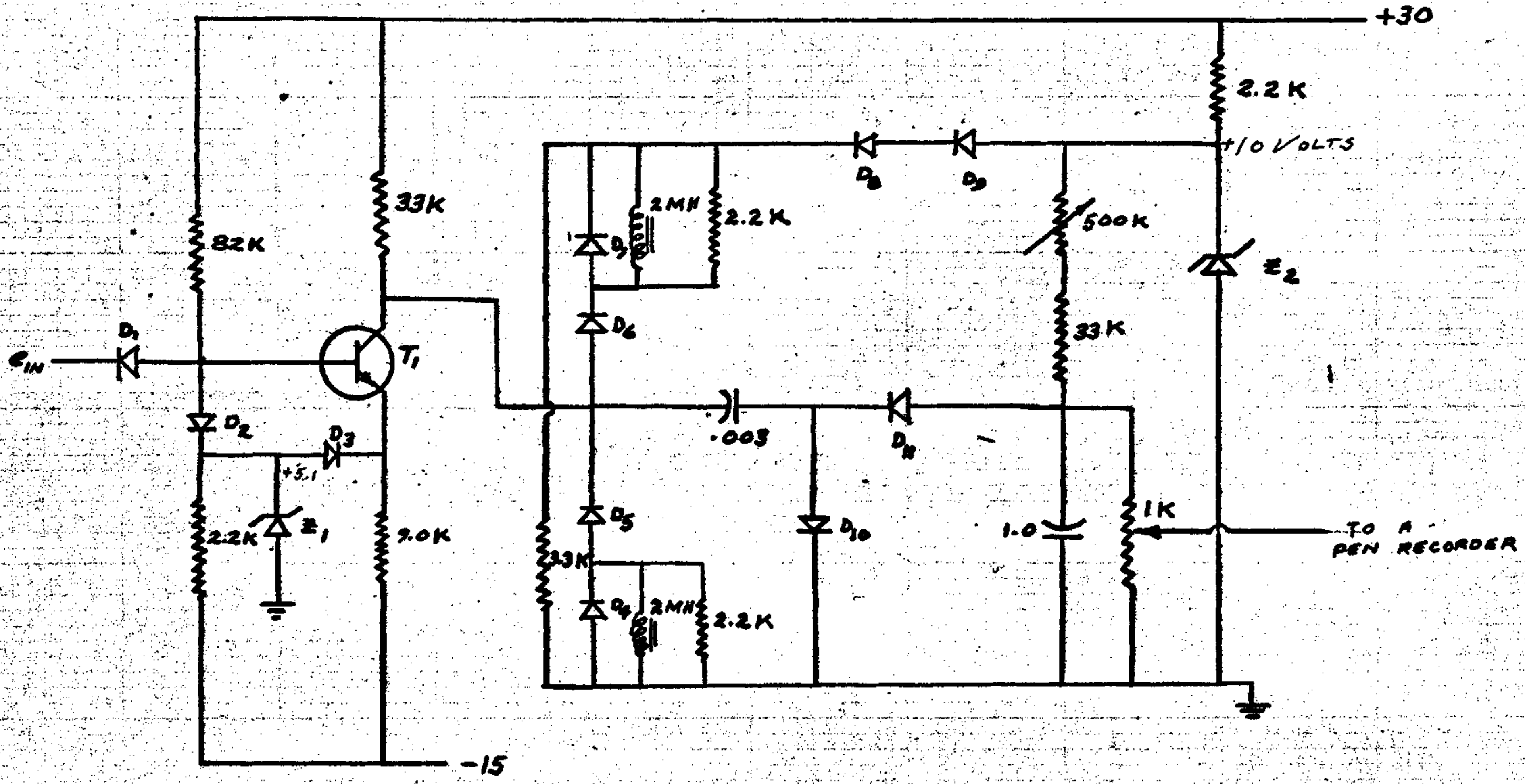
Fig. 44 Playback amplifier and limiter.

wiper of the resistance potentiometer. This permits the 3 db breakpoint of this network to be adjusted. To limit the bandwidth of the amplifier and thus decrease the noise in the output, feedback capacitors are used from the collectors of T_1 and T_2 to their respective bases.

The output from this amplifier is taken from the wiper on the collector load resistor of T_3. This permits an adjustment of the output level of the amplifier.

From the collector of T_3 the signal is a.c. coupled to the fourth stage providing a gain of 10. From the collector of this stage, the signal passes onto T_5 in an emitter follower configuration. This stage provides the necessary power to drive the limiting stage. The limiting circuit is a transistor in the common emitter configuration with a v ery high gain. This transistor is biased so that it switches on at the negative going zero crossing point and switches off at the positive going zero crossing providing good limiting action.

From the collector of T_6 the signal is then fed to the frequency demodulator shown in Fig. 45. This circuit is a pulse averaging type of demodulator based on a "diode pump" arrangement. The configuration which has been analyzed by Mitchel (21) and Buckley (2) shows very good linearity over a wide frequency range. The basic portion of this circuit is shown in Fig. 46.



T_1 - 2N1306

Z_1 - 1N751

Z_2 - 1N758

D_1 to D_{11} - 1N273

ALL CAPACITIES IN μf

Fig. 45 Diode pump demodulator.

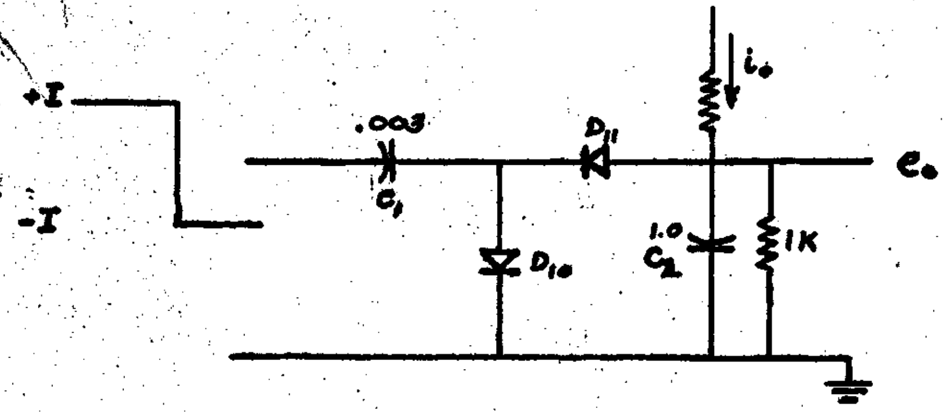


Fig. 46 Basic portion of the diode pump demodulator.

Capacitor C_1 is fed from a current source that alternates between +I and −I with each zero crossing of the input waveform. When current is flowing into C_1, diode D_{10} is forward biased and diode D_{11} is reverse biased. When the input current changes polarity and current is drawn from C_1, D_{10} is reverse biased and D_{11} forward biased, drawing charge from C_2. Thus for each period of the input waveform, a given charge is removed from C_2. If a constant current i_o is fed into C_2 that is equal to the average current withdrawn via D_{11}, then at a centre frequency f the voltage e_o is equal to 0. If the frequency increases, then more charge per unit time is withdrawn from C_2 than is added and, hence, the output voltage becomes negative. If the frequency decreases the output voltage becomes positive.

In the complete circuit shown in Fig. 45, transistor T_1 acts as the alternating current source. Diodes D_4 to D_9 and the 2 millihenry coils are required to improve the linearity over the basic circuit and, as well, temperature compensate the circuit.

A printed circuit board with the amplifier, limiter and frequency demodulator is shown in Fig. 47.

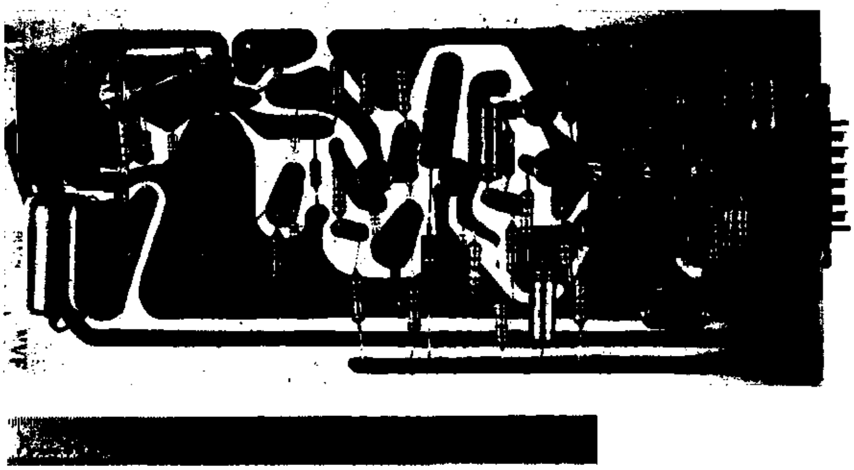


Fig. 47 Playback Electronics

4. RESULTS

4.1 System Test Results

Shown in Fig. 49 is the frequency response of the system from the ECG amplifier to the output of the frequency demodulator. This response was obtained by using 1 write head followed by a reading head, thus including the complete system in the test. Although the system is capable of recording signals with a frequency from 0 to 200 cycles/second, the bandwidth of this channel has been limited to decrease the noise produced by the muscle activity of a subject.

A check on the linearity of the carrier oscillator showed that the linearity of the conversion from voltage to frequency was approximately 1%. The carrier demodulator was also checked and found to convert the frequency back to voltage with a linearity of better than 1%.

To test the performance of the recorder under the forces of activity likely to be encountered, the ECG of a subject was recorded with the recorder mounted as in Fig. 50, while he underwent a series of exercises. The reproduced signal showed that activities such as running upstairs, jogging, cross country running, and sprinting do not affect the performance of the recorder. A typical reproduced signal which was recorded while the subject was jogging is shown in Fig. 51.

4.2 Discussion

This thesis has described the design and construction of a tape recorder suitable for a subject to carry during his

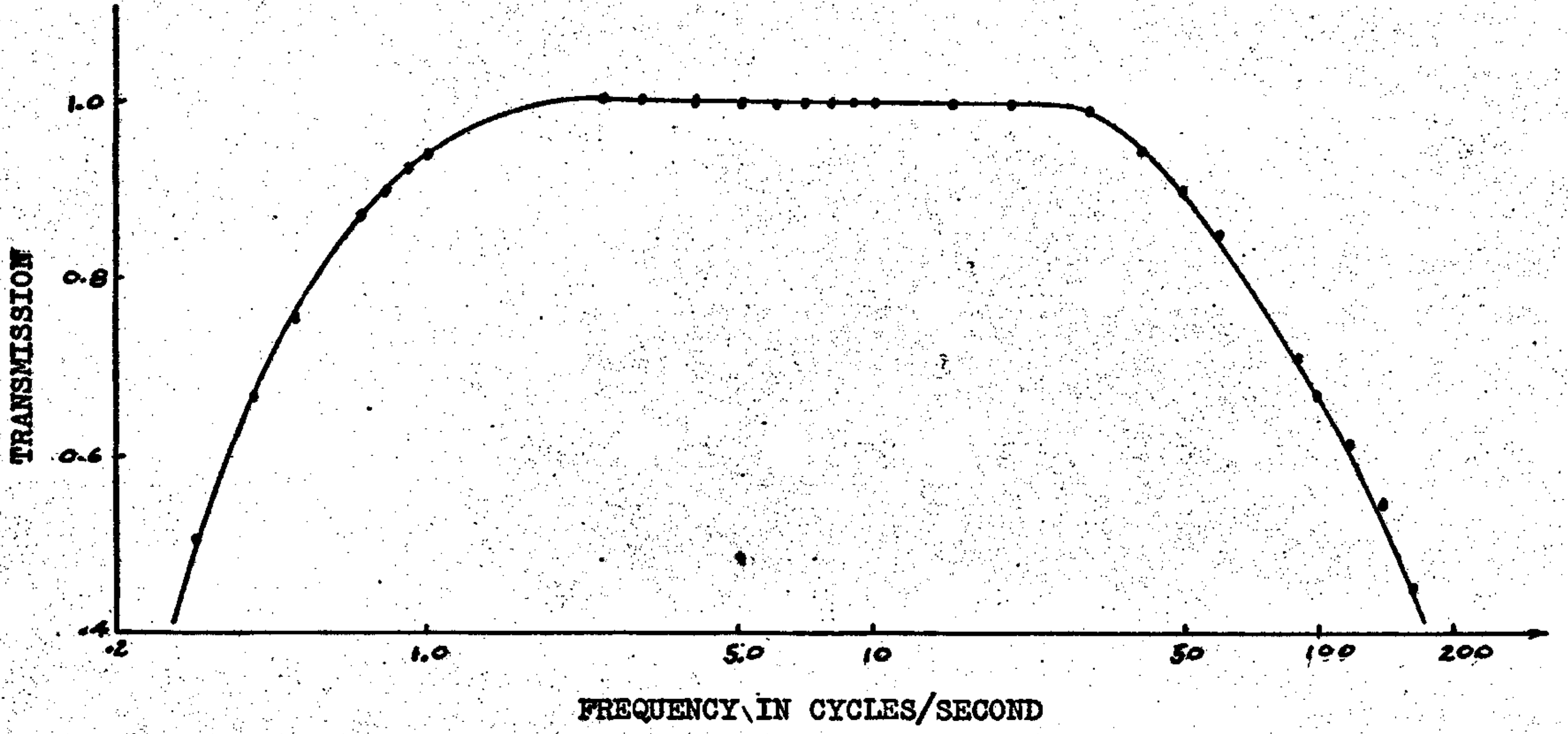


Fig. 49 Complete system frequency response

Fig. 50 Photograph showing the method of mounting for testing the recorder

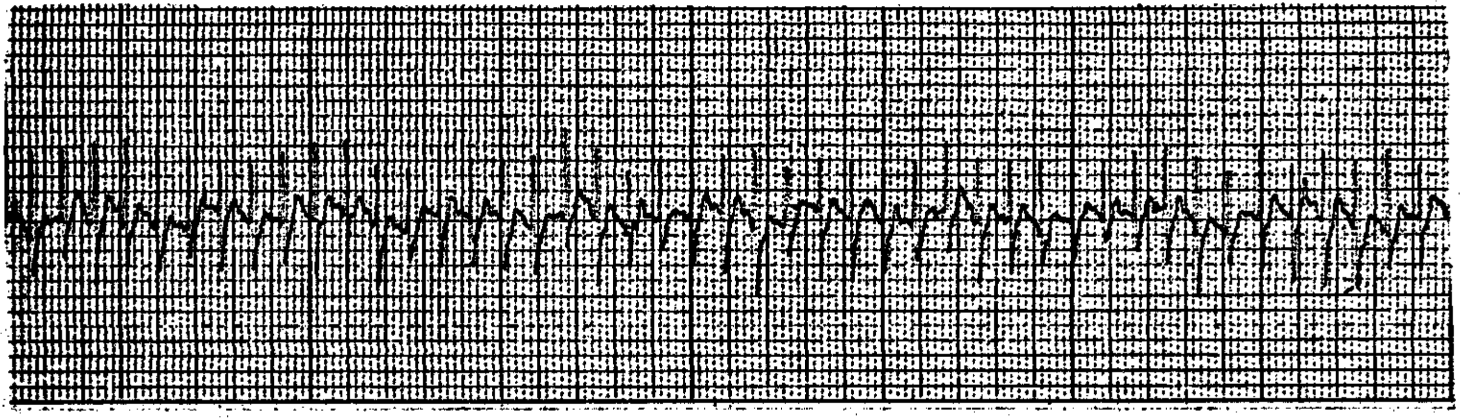


Fig. 51 An ECG recorded while the subject was jogging.

normal activity and capable of recording up to 4 channels of information. The complete electronic system required for recording 1 ECG has been described and tests have been run on this channel to show that this method of monitoring an ECG is a practical solution to the problem outlined in the introduction.

While this recorder was designed for recording 3 ECG's recent work has shown that it is also suitable for recording such other physiological data as body temperature and respiration rate. A record from the tape recorder showing 2 ECG's, respiration rate and body temperature taken from a subject at rest is shown in Fig. 52.

The characteristics and capabilities of the recorder may be summarized as follows:

Number of channels	4
Frequency range (F.M.)	D.C. to 200 cycles/second
(Using direct write the frequency range would be approximately from 100 cycles/second to 3 kilocycles/second)	
Linearity (F.M.)	approximately 1%
Stored record length	1 hour 5 min.
Total weight without batteries (4 channels of electronics)	3.0 lbs.
Power consumption	1 watt
Weight of batteries	.1 lbs/hour of recording
Real time error	2%
Peak to peak flutter	1%

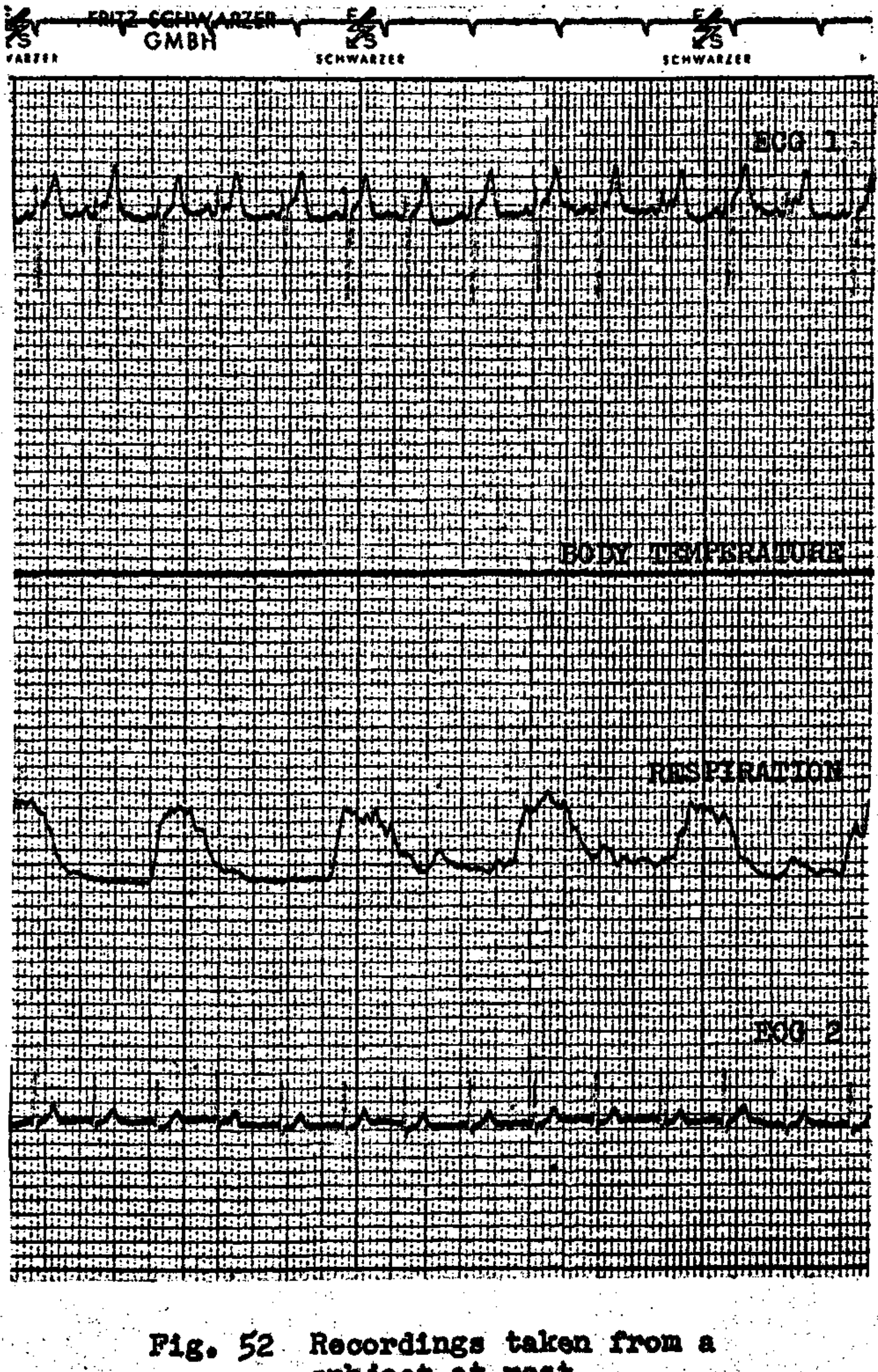


Fig. 52 Recordings taken from a subject at rest.

It is interesting to note how the weights of the various parts of the recorder compare. If one assumes a battery weight of ½ lb. (5 hours recording) the distribution of weight is as shown in Fig. 53.

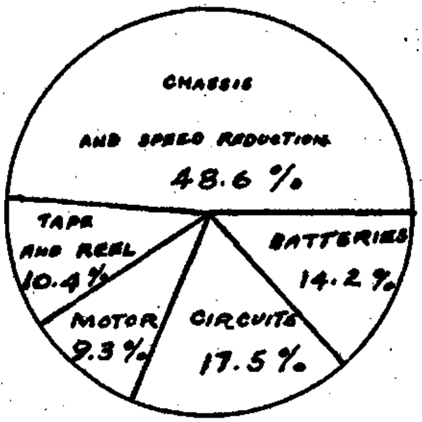


Fig. 53 Weight distribution of the recorder.

This indicates that the areas requiring further development on a recorder of this type are; miniaturization and minimizing total power consumption. By reducing the physical size of the circuits and minimizing total power consumption, both the size of the recorder and the weight of the recorder could be reduced. As well as reducing the power consumed by the circuits, it would also be desirable to reduce the power required to drive the tape transport.

Research could also be done to determine the best ECG electrode and amplifier combination to give an ECG with the least baseline waver

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APPENDIX A

ECG VIA TELEPHONE

A.1 Introduction

An electrocardiogram is the electrical activity of the heart as measured between two electrodes placed on more or less standardized positions on the subject. To the diagnosing physician the amplitude and waveform are important parameters for the diagnosing of heart disease. The methods of obtaining the electrocardiogram and the equipment used have become more or less standardized and are found in almost every hospital and doctor's office. While most electrocardiograms may be interpreted by the general practitioner, not every doctor has had the training nor experience to interpret all the irregularities that may be observed. In situations where the general practitioner has obtained an unusual electrocardiogram, it is a common practice to mail this record to a specialist in electrocardiography to obtain his interpretation.

In Saskatchewan there are a relatively large number of small hospitals, some distance from the larger centres, which cannot afford a specialist in electrocardiography. The mailing in of electrocardiograms for interpretation is a slow process and completely inadequate for emergency cases. To provide a faster method of consultation, a number of systems have been developed for sending electrocardiograms over a standard telephone circuit while the electrocardiogram is being recorded from the patients. The systems have the advantage that both the recording physician and the specialist have identical records and may discuss any irregularities immediately.

A number of such systems were evaluated and, although found satisfactory from a technical point of view, were considered unnecessarily expensive. A project was therefore initiated to develop a simple inexpensive method of transmitting electrocardiograms by telephone.

A.2 System Consideration

A system for the transmission of an electrocardiogram by a telephone must be simple, and adapt readily to standard electrocardiogram recorders. It must provide the recording physician and the specialist with identical records in a form they are familiar with, with a minimum of equipment, adjustments or controls. The system must be reliable and inexpensive.

The problem is essentially one of designing an inexpensive system of coupling standard electrocardiograph recorders to the standard telephone system. Since telephone companies do not look with favour on direct electrical connection to their system, coupling of the recorders must be indirectly made via the telephone handsets. Since the information contained in the electrocardiogram is within the bandwidth of one to one hundred cycles per second, a system of carrier modulation must be selected, taking into account the characteristics of the overall telephone communication circuit.

An examination of the amplitude response curves of a typical telephone circuit (which includes the microphone and speaker) indicated that no portion of the curve was sufficiently flat to allow an amplitude modulation system to be used. In addition, an amplitude modulated carrier is subject to interference due to extraneous line noise and interference. Using frequency modulation, the effects of nonlinear amplitude response may be eliminated by amplifying and limiting before demodulation. However, frequency modulation is subject to another form of distortion caused by a nonlinear phase response of the carrier media. If this form of modulation is to be used, a centre carrier frequency and deviation must be selected to keep distortion to a minimum. Although the response of telephone handsets varies considerably, an examination of a number of units indicated that a centre frequency of 1800 cycles per second would achieve the best results. The total bandwidth of an FM modulation system is given by:

$$\text{B.W.} = 2(m_f + 2)f_m$$

$$\text{where} \quad m_f = \frac{\text{deviation}}{\text{modulating frequency}}$$

$$\text{and} \quad f_m = \text{modulating frequency}$$

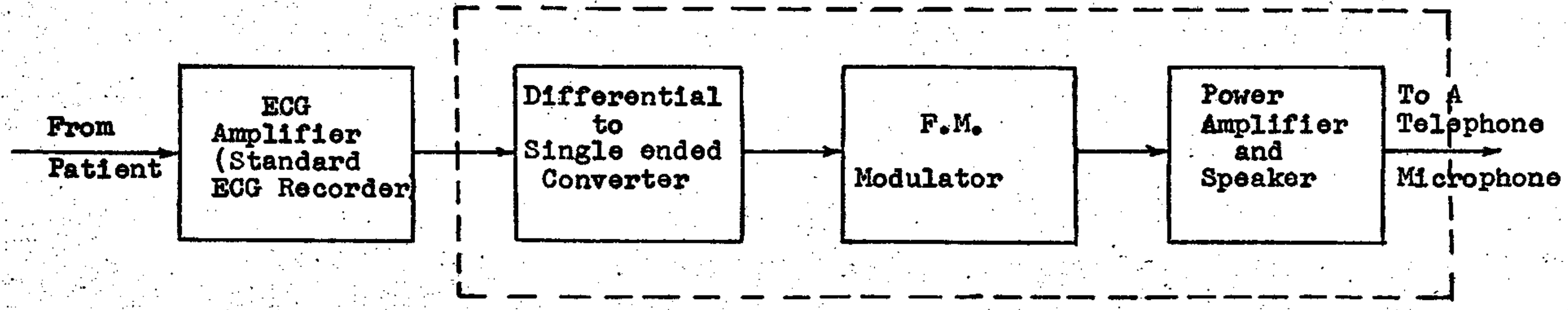
Thus, using a deviation of 200 cycles/second and assuming all necessary information of the electrocardiogram is contained in the bandwidth indicated above, the total transmission bandwidth is in the order of 800 cycles/second. This falls in the region of almost flat amplitude response for the telephone system and, therefore, one may assume that this is also the region of linear phase response.

Since indirect coupling must be used, one is limited to either acoustic or electromagnetic form of coupling. Both forms were investigated and found to place limitations on the system. A block diagram of the overall system is shown in Fig. A.1. The system consists of a simple transmitter which receives its input from a standard ECG recorder, converts this signal to a frequency modulated audio signal which is coupled acoustically to a standard telephone handset microphone via a small speaker. The receiver picks up the audible audio signal from the speaker of a standard telephone handset via a microphone, demodulates this signal and inputs the demodulated signal into a standard ECG recorder. Specifically, the system consists of two units identical to the one shown in Fig. A.2. The system is simple to use, requiring only that the operator "plug" the unit into the recorder and place the telephone handset in the required position. The overall cost of the system is roughly $300.

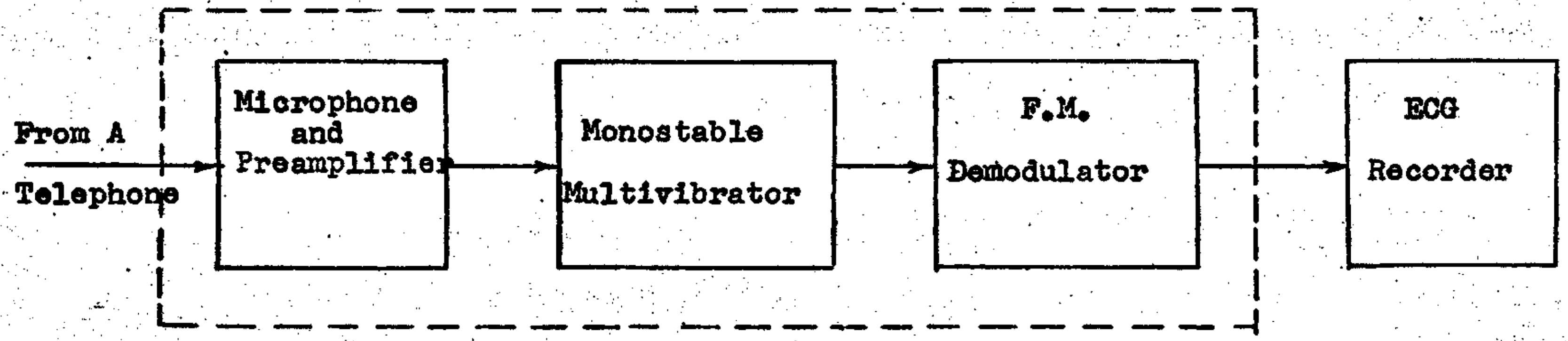
A.3 System Details

A.3.1 Transmitter

The circuit diagram of the transmitter is shown in Fig. A.3. The output of the transmitting recorder is fed to the base of transistors T_1 and T_2. These transistors make up an amplifier which converts the differential signal from the



TRANSMITTING SYSTEM



RECEIVING SYSTEM

Fig. A.1 ECG VIA TELEPHONE SYSTEM

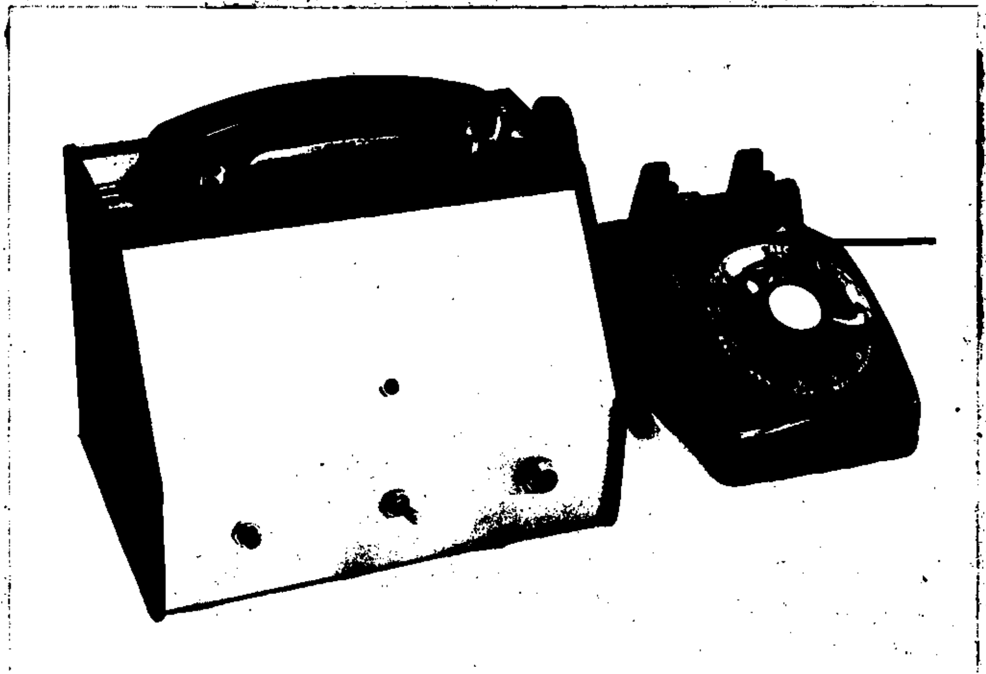
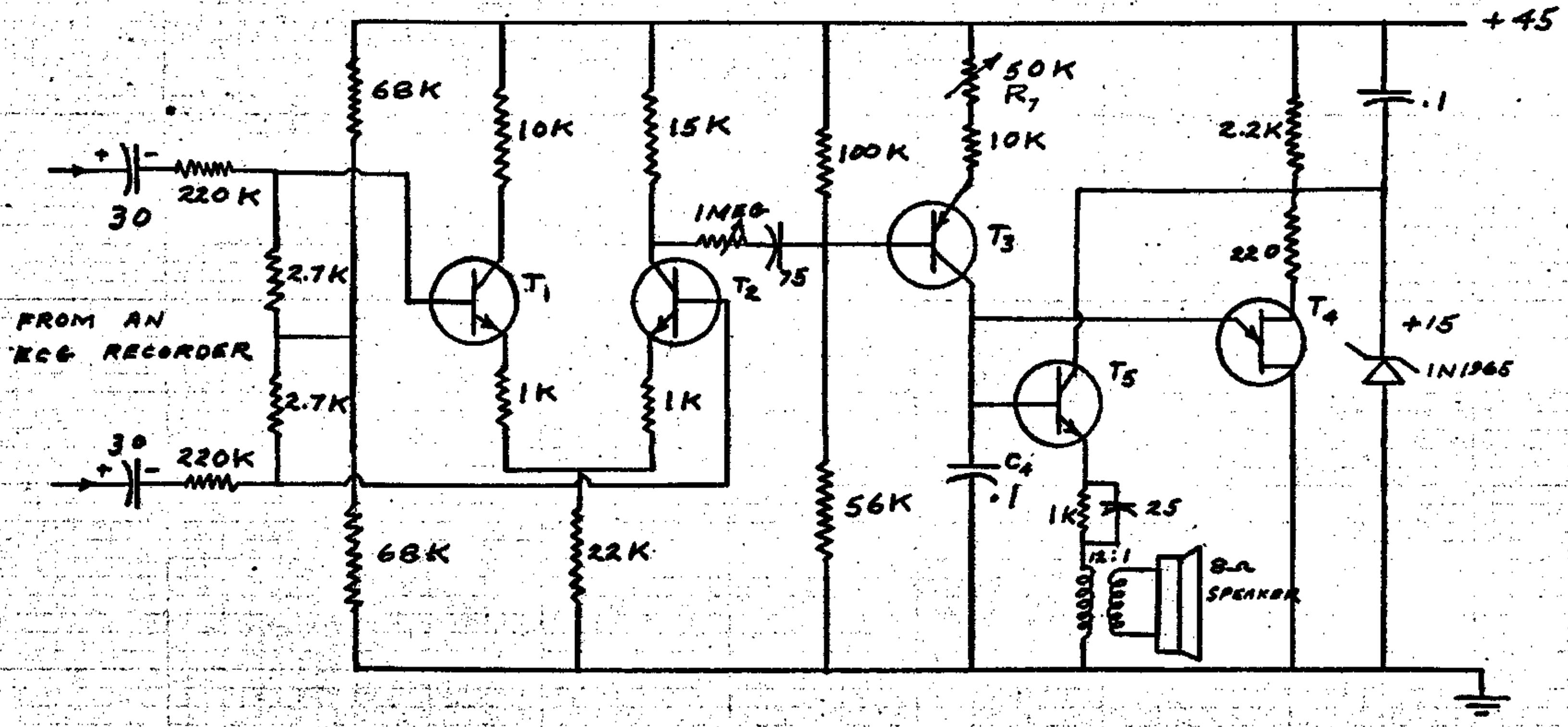


Fig. A.2 ECG via telephone transmitter.



T_1, T_2, T_5 - 2N1306

T_3 - 2N1306

T_5 - 2N2160

ALL CAPACITIES IN μf

Fig. A.3 Transmitter circuit diagram

recorder to the single ended signal that is fed to the frequency modulating circuit. The differential to single ended converter was required to reject the common mode signal coming from the recorder before it was fed to the modulator. The improvement in the ratio of signal to common mode signal as given by the analysis in A.8 is 20.

The output of the converter is fed directly to the frequency modulation circuit. Frequency modulation in this circuit is obtained by varying the current supplied to timing capacitor C_4 of a "free running" unijunction multivibrator. The active elements of the multivibrator are T_3 and T_4. Under quiescent conditions, T_3 appears as a constant current source supplying current to C_4. C_4 is discharged by the unijunction transistor T_4 when C_4 has charged to T_4's firing potential. The "free running" or centre frequency of the multivibrator is altered by adjusting the current that T_4 delivers to C_4. This is accomplished by adjusting the variable resistor R_7. When a modulating voltage is applied to the base of T_3, the current supplied by T_3 to C_4 is varied proportional to the modulating voltage, thereby producing a deviation from centre frequency proportional to the input voltage.

The power amplification needed to drive the speaker that couples the circuit to the telephone circuit is supplied by T_5 in an emitter follower configuration. The transformer is used to match the impedance of the power amplifier to the impedance of the speaker. Coupling the transmitter to the telephone circuit is done by placing the microphone of the telephone handset on top of the speaker of the transmitter.

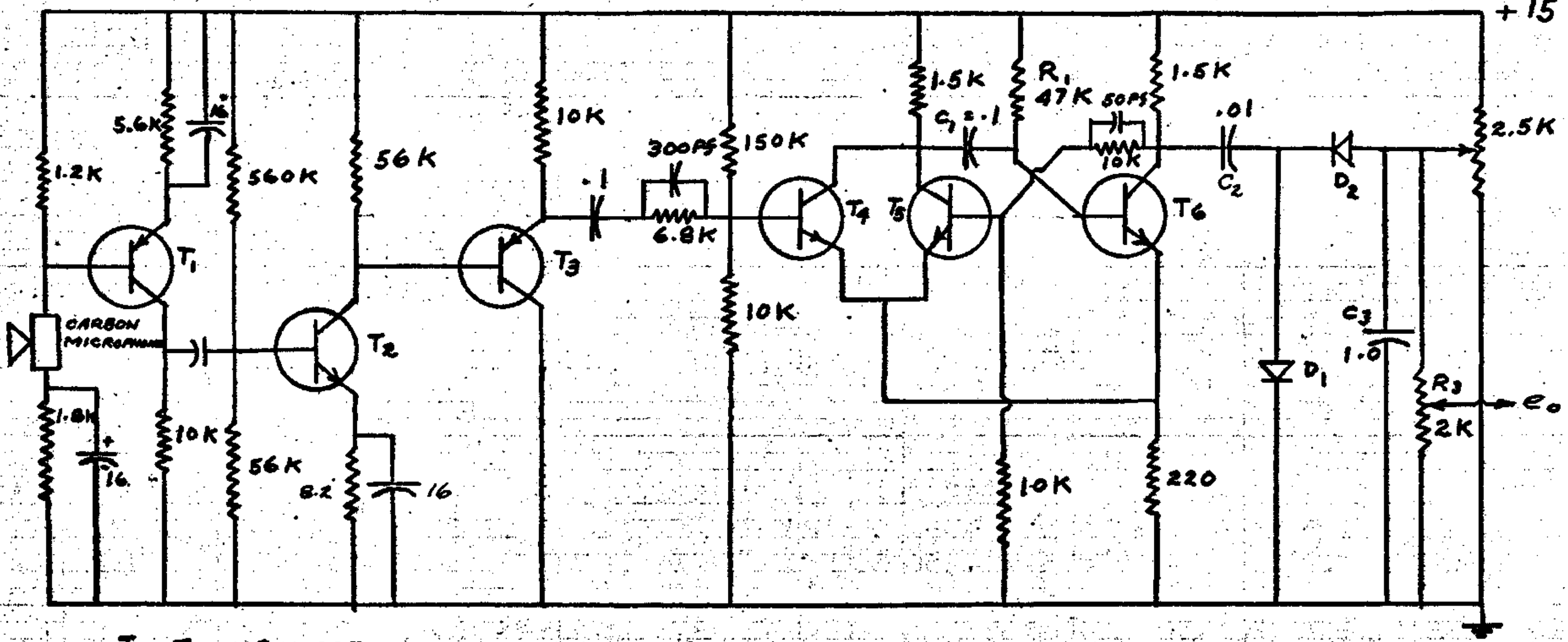
A.3.2 Receiver

The requirements of the receiving unit are to pick up the audible signal delivered by the speaker of the telephone handset and demodulate the frequency modulated signal. The first requirement of the pickup may be done in one of two ways. The most obvious method is to place a microphone very near the speaker of the telephone handset and produce a voltage proportional to the audible signal. A second method is to pick up the stray magnetic field which surrounds the speaker that produces the audible signal. Both of these methods have been tried with the former chosen as the most satisfactory.

Pickup using a magnetic pickup coil was tried using a coil of a thousand turns of wire wound on a plastic spool that fitted snugly over the speaker end of the telephone handset. The voltages induced in this coil by the stray field of the speaker were in the order of one to ten millivolts, depending of course on the signal strength. The major disadvantage found in using this type of coupling was that the voltages induced in the coil by stray electrostatic or electromagnetic fields within the near vicinity of the receiver were in the same order of magnitude as the received signal. One source of the interference was found to be the starting of fluorescent lights. Shielding the pickup coil did not significantly reduce the noise pickup in the system. For the reason just mentioned, magnetic pickup was abandoned and acoustic pickup was used.

Shown in Fig. A.4 is the final form of the receiver circuit with a carbon microphone as the pickup element. Using a microphone for pick up makes the receiver subject to interference that is generated in the form of audible noise within the receiving room. However, because this type of noise is more easily controlled than electrical noise, this type of pickup has a definite advantage over electromagnetic pickup. In an effort to reduce the effects of audible noise within the room, acoustic shielding has been placed around the pickup microphone. The acoustic shield is a cup made of lucite with a foam rubber lining and a foam rubber cushion at the bottom for the microphone to sit on. The foam rubber lining and cushion prevent a direct mechanical connection between microphone and handset. The precaution of suspending the microphone in foam rubber was necessary as the carbon microphone is quite sensitive to mechanical vibrations.

Shown in Fig. A.4 is THE carbon microphone as it appears with the rest of the circuit elements that make up the receiver. The receiver circuit may be broken down into three parts. The first portion of the circuit which amplifies and clips the incoming signal is made up of transistors T_1, T_2, and T_3. T_1 is saturated under quiescent conditions, but when an incoming signal increases the resistance of the carbon microphone, T_1 approaches cutoff and a negative half sine wave is produced



T_1, T_3 - 2N1307
T_2, T_4, T_5, T_6 - 2N1306
D_1, D_2 - HR10211
ALL CAPACITIES IN μf UNLESS STATED

Fig. A.4 Receiver circuit diagram

at the collector. From the collector of T_1, the signal is fed to the base of T_2. Transistor T_2 is biased in saturation and when the negative half sine wave appears at the base, T_2 is switched off producing a positive square wave at the collector. This signal is then passed to the emitter follower T_3, which provides the driving power to operate the next section of the receiver.

The next section of the receiver which is made up of T_4, T_5, and T_6 is a one shot multivibrator. Under quiescent conditions T_4 and T_5 are cut off and T_6 is saturated. When a positive signal is applied to the base of T_4, it is turned on. When T_4 is turned on, T_6 is turned off and T_5 is in turn turned on. Then, after a time determined by C_1 and R_1, T_6 turns on again and T_5 is shut off. The bias on T_4 has been adjusted so that it will turn on at the zero crossing point of a sine wave that is applied at the carbon microphone. The time constant R_1C_1 has been adjusted to make the period of the one shot equal to about three quarters of the period of the carrier centre frequency. Thus, if the input to the first section of the receiver is a sine wave of varying amplitude and frequency, then only the variations in frequency will be seen at the output of the second section. Variations in frequency will be seen as variations in the time between pulses of constant height and width at the output.

The output of the one shot is fed to the last section of the receiver which is a "diode pump demodulator". When the output of the one shot is positive, current flows through C_2 reverse biasing D_2 and through D_1 to ground. When the one shot returns to its more negative state, C_2 discharges, reverse biasing D_1 and forward biasing D_2 drawing current from C_3. For the balance condition a current equal to the current drawn from C_3 is injected into C_3 via R_2. It can be seen that if the frequency of the sine wave applied to the input of the receiver increases the current drawn from C_3 increases, and the voltage across R_3 will become negative; if the frequency at the input decreases the current drawn from C_3 decreases, and the voltage across R_3 becomes positive. A more complete discussion of this basic circuit as well as its many refinements has been made by Buckley (2).

A.4 System Cost

Below is an estimate of the cost of building a transmitter and receiver.

		TRANSMITTER		RECEIVER
Components		35.00		15.00
Power Supply	(Acopian Model #45A10)	50.00	(Acopian Model #15A10)	50.00
Chassis and Labour		65.00		65.00
Total		$150.00		$130.00

A.5 Test Results

Shown in Fig. A.5 is a typical transmitted and received electrocardiogram. The electrocardiogram was transmitted from the Canora Union Hospital at Canora, Saskatchewan, and received at the University Hospital at Saskatoon. The total map distance of transmission is about two hundred miles. It may be noted that the higher frequency components of the records show up more clearly on the received record than on the transmitted record. The reason for this is that the mechanical portion of the pen recorder at Canora has a lower cutoff frequency than the mechanical portion of the pen recorder in Saskatoon.

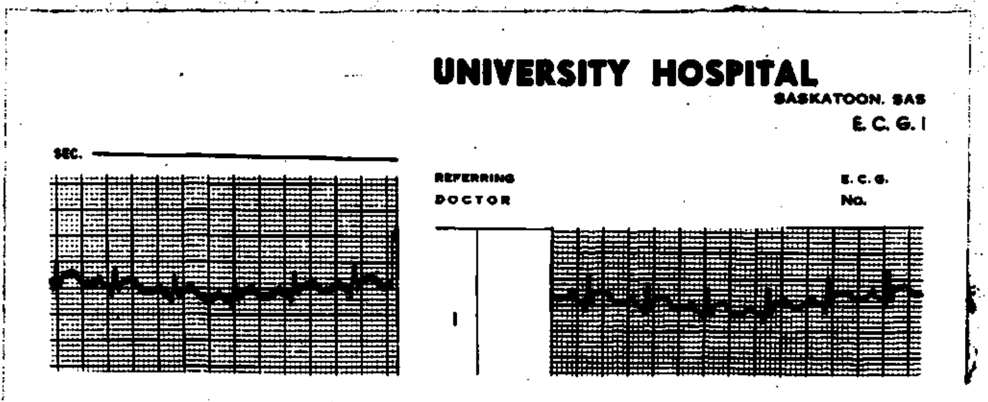


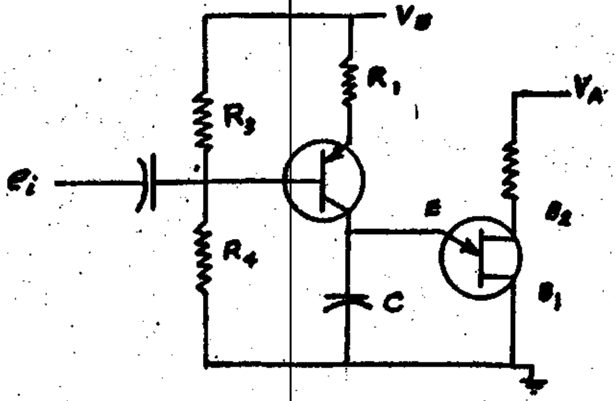
Fig. A.5 Transmitted and Received Record
(Transmitted to the left, Received on right)

A.6 Discussion

The system that has been developed demonstrates that ECG's may be transmitted over normal voice channels using inexpensive and simple solid state circuits. The system is quite simple to operate and requires very little extra effort by the sender other than to place a long distance telephone call.

A check on the calibration of the entire system during transmission may be carried out by sending the 1 millivolt pulses that are derived from a standard cell at the input of the transmitting ECG amplifier. If adjustment is required the output level of the receiver is adjusted until the proper pen deflection is noted.

A.7 Modulator Analysis



When capacitor C charges to a potential V_a, the unijunction will fire and discharge C very rapidly. If the collector of the transistor is assumed a current source and the leakage of the unijunction is assumed negligible, then the free running frequency f is given by

$$it_o = CV$$

$$f = \frac{i}{CV}$$

if V_E = voltage from the emitter to the unijunction to B_1
during conduction

and ηV_a = voltage required to fire the unijunction.
The current supplied to C is given by

$$i = (V_B - V_1)/R_1$$

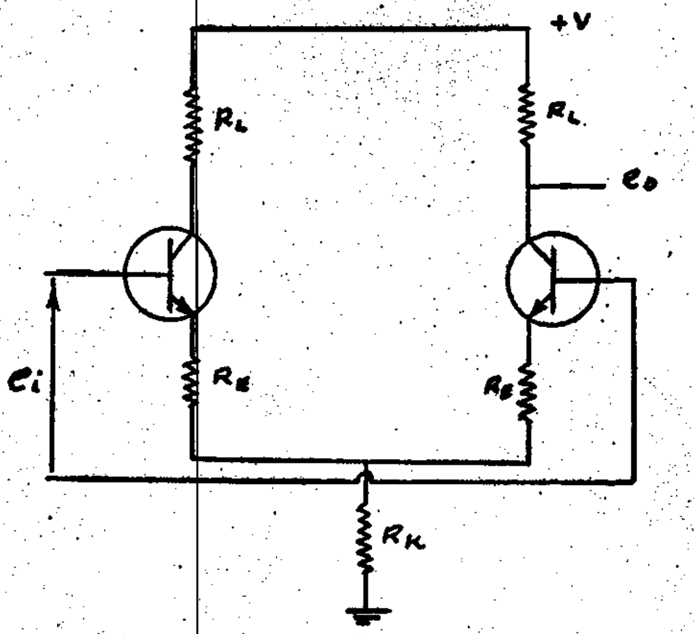
and $\therefore$

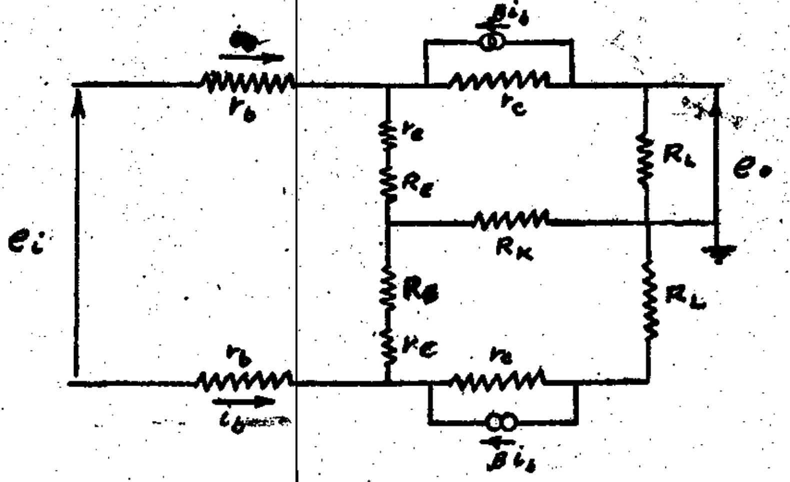
$$f = \frac{(V_B - V_1)}{R_1 C(\eta V_a - V_E)}$$

This expression for f also assumes that discharge time is very short.

It can also be seen from the expression for f that a change in V_1 will result in a proportionate change in frequency. Since change in V_1 is equal to the input signal, the change in output frequency is directly proportional to input voltage.

A.8 Differential to Single Ended Converter Analysis





Equivalent Circuit of Differential Amplifier

If r_c is assumed to be very large, then the differential gain is given by

$$e_o = -\beta\, i_b\, R_L$$

$$e_i = r_b i_b + (\beta+1)(R_E + r_e) i_b$$

$$A_{DIFF} = \frac{-\beta\, R_L}{r_b + (B+1)(R_E + r_e)}$$

The gain, if the output is taken single endedly, is:

$$A_{S.E.} = \frac{1}{2}\,\frac{-\beta\, R_L}{r_b + (B+1)(R_E + r_e)}$$

For a common mode signal with single ended output:

$$i_{b1} = i_{b2}$$

$$e_o = -\beta\, i_{b1}\, R_L$$

$$e_i = i_{b1}\, r_{b1} + i_{b1}(\beta+1)(R_E + r_e) + 2\beta\, i_{b1}\, R_K$$

$$A_{C.M.} = \frac{-\beta\, R_L}{r_{b1} + (\beta+1)(R_E + r_e) + 2\beta R_K}$$

The improvement in the ratio of differential signal to common mode signal is given by:

$$\frac{A_{S.E.}}{A_{C.M.}} = \frac{r_b + (\beta+1)(R_E + r_e) + 2\beta R_K}{2\, r_b + (\beta+1)(R_E + r_e)}$$

For the value of components used and $\beta = 100$, $\frac{A_{S.E.}}{A_{C.M.}} = 20$.