DESIGN OF AN INEXPENSIVE TAKE-HOME RECORDER FOR CARDIAC MONITORING

by

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UNIVERSITY OF UTAH GRADUATE SCHOOL

SUPERVISORY COMMITTEE APPROVAL

of a thesis submitted by

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ABSTRACT

Circuitry has been designed to be used in conjunction with a conventional battery powered cassette recorder for inexpensive take-home cardiac monitoring. Patients experiencing cardiac disturbances at home can make immediate ECG recordings that can later be monitored by the physician for rate and rhythm changes.

Since the required bandwidth for ECG recording is .05 Hz to 100 Hz, and since typical battery powered cassette systems display a bandpass response that rolls off on the low end at about 200 Hz and on the high end at about 5K Hz, it has been necessary to devise a frequency modulation-demodulation scheme to faithfully recover the ECG waveform. A second signal (reference frequency) is also necessary to compensate for the variable tape speed (flutter and wow), of the recording system. Also a third signal is necessary to be actuated by the patient to indicate a particularly noticeable episode during his ECG recording.

The recording packet which is battery powered contains an ECG amplifier plus three voltage controlled oscillators (Signetics 566) to frequency modulate 1.) the ECG signal; 2.) the reference signal; and 3.) the episode signal. These three FM signals are multiplexed (summed) and recorded on a conventional cassette recorder.

The playback system to be used by the physician at his convenience for monitoring, utilizes phase locked loop (Signetics 565) integrated circuits for bandwidth separation and demodulation of the three signals. The reference signal is subtracted from the ECG waveform utilizing operational amplifier circuitry to compensate for tape speed variations. The episode signal is amplified with an operational amplifier such that when actuated a light emmitting diode is turned on.

The goal of the project was to keep the total component cost less than $100.00, yet retain quality signal reproduction; which was accomplished using a Concord model F-26 cassette recorder.

I. INTRODUCTION

The work reported in this thesis derived its original impetus from the desire of G. Landon Beales, M.D., a cardiologist, to have available an inexpensive, convenient, take-home electrocardiogram (ECG) recording system. With such a device a patient experiencing cardiac disturbances at home could make an immediate electrocardiogram recording of the event. It was also desirable to record simultaneously with the ECG a second signal that could be actuated by the patient to indicate a particularly noticeable episode. This valuable diagnostic information would then be stored, and could be played back for monitoring by the doctor at his convenience.

Of the current recording mediums available, a battery powered cassette system offered attractive advantages. Particularly it is inexpensive, compact, portable, and easily manipulated. There are, however, two basic limitations of such a system that must be overcome. First, the required bandwidth for ECG reproduction is .05 to 100 Hz. Typical battery powered cassette recorders display a band-pass response that rolls off on the low end at about 200 Hz and on the high end at about 5K Hz. It was therefore necessary to devise a modulation-demodulation scheme in order to faithfully record and reproduce the ECG signal.

The second problem was distortion derived from variable tape speed (flutter and wow)[1], which can be removed by recording simultaneously with the frequency modulated ECG signal a reference frequency signal, which upon playback, is subtracted from the ECG waveform.

Since inexpensive cassette recorders utilize only one recording track, and considering the above requirements, frequency modulation with multiplexing, similar to that used in modern day telephone telemetry[2], lends itself well to the situation.

The challenge then at the onset was, using state of the art components, to design, construct, and test inexpensive circuitry to provide a clinically satisfactory electrocardiogram cassette recording system.

II. REVIEW OF LITERATURE

Einthoven, a Dutchman, developed the first practical ECG instrument in 1903.[3] The device consisted of a very thin gold plated quartz fiber suspended between the poles of a permanent magnet. When signals from the surface of the body were connected to the ends of the fiber, current from the heart voltages induced electromagnetic fields which interacted with the permanent field of the magnet and caused the fiber to move (Faraday's Law). This motion was magnified by an optical system and subsequently reproduced on a moving strip of photographic paper. This "string galvonometer ECG" was delicate and difficult to adjust and has been superseded by modern day differential amplifier circuitry.

In 1906 Einthoven successfully transmitted ECG signals over telephone lines.[4] Medical telephone telemetry later became popular after World War II and in recent years has gained widespread use.[5,6] Bell Telephone developed the first commonly used telephone telemetry single channel data set transmission system (Bell Telephone 603A and 603B), consisting of a frequency modulated transmitter and receiver. Later in 1967 Bell Telephone set 604A and 604B was introduced, capable of transmitting and receiving three channels of analog data over one telephone line. This system is currently in use by the Latter-day Saint Hospital in Salt Lake City for transmission of ECG's and other physiological data.[7]

Hewlett-Packard's H-P Model 1515A ECG phone terminal is also capable of three channel transmission. Twelve standard leads are recorded in four sets of three leads each with automatic lead switching. Marker identification of lead sets and standardization are also automatic. Within ten seconds after connection of electrodes a complete 12-lead electrocardiogram can be produced.[3]

In the area of continuous cardiac monitoring, Norman Holter made significant contributions. In 194[illegible] Holter developed "radioelectrocardiography" by which an ECG was transmitted via radio transmission from an 80-pound backpack unit.[8] In the ensuing years the electrocardiographic units were miniaturized and used clinically in the area of exercise tolerance tests.[9] With further technological developments, Holter designed a battery operated, miniaturized portable tape recorder in 1961, for continuous recording of the patient's electrocardiogram. This eliminated the need for radiotelemetry and radio receiving equipment, and allowed the patient a greater range of activity. Holter also developed specially designed play-back equipment for high-speed analysis of the voluminous data obtained from each recording.[10] Continouous recordings are made for a period of 10 to 24 hours and are correlated with the patients record of his activities during the monitoring period for clinical evaluations. This procedure of continuous portable recording of electrocardiograms has become known as "Holter monitoring".[11]

The Holter-Avionics system is widely used for detecting transient disturbances in heart rate and rhythm during long-term recording of the ECG under dynamic conditions. The system consists of three main part : 1.) a portable tape recorder which records the ECG of the patient; 2.) an oscilloscope on which the ECG signals recorded by the tape are displayed by rapid superposition; and 3.) a unit for printout of the ECG complexes from the tape onto conventional ECG paper.[12]

Survival Technology Inc., of Bethesda, Md., has developed a portable heart monitoring device about the size of a pocket calculator called the Cardio Beeper that detects arrhythmias. This device contains integrated circuitry to measure the heart rate and determine if the rate monitored by the device exceeds that of a preset oscillator, at which time a "beep" alarm is activated. The patient can then call his physician and transmit his ECG over the telephone line via a voltage controlled oscillator within the Cardio Beeper that sends modulated audio tones which are decoded at the doctor's office.[13]

Recent developments in solid state integrated circuits have made the development of the take-home ECG cassette recorder economically feasible. While most transmission and monitoring systems cost several thousands of dollars, it is our goal to develop a take-home recorder for more general use at a component cost of less than \$100.00.

III. PRELIMINARY FEASIBILITY STUDIES

In order to determine if the proposed theory is feasible and practical for ECG recording and reproduction using a cassette system, it was necessary to do some preliminary studies using comparatively expensive FM modulation-demodulation equipment, currently in use for telephone telemetry at the LDS Hospital in Salt Lake City.

The circuitry utilizes three frequency modulated, multiplexed channels with carrier frequencies at 1075±100 Hz, 1935±100 Hz, and 2365±100 Hz (Figure 1).

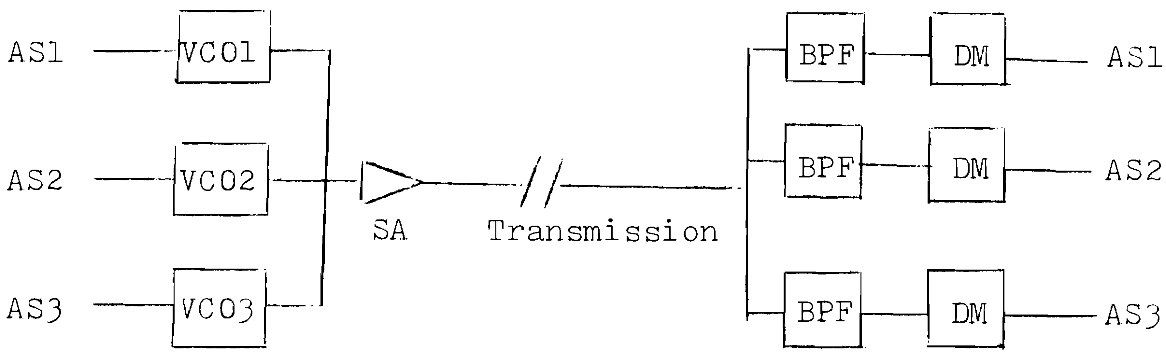


Figure 1. Three channel FM telemetry. (AS-analog signal, VCO-voltage controllable oscillator, SA-summing amplifier, BPF-bandpass filter, DM-demodulator)

Two of the channels were used in this study: one to carry the ECG signal and the other to carry a reference signal for tape speed compensation. The two waveforms were frequency modulated, multiplexed, then recorded. Upon playback after discrimination, the reference signal was inverted and added to the ECG waveform, thereby compensating for the variable tape speed. Figure 2a is a playback from the recorder of an ECG waveform coming directly from demodulator two, and Figure 2b is the tracing of the synchronous compensating signal also taken directly from demodulator three output. The noise added by variable tape speed is apparent on both tracings.

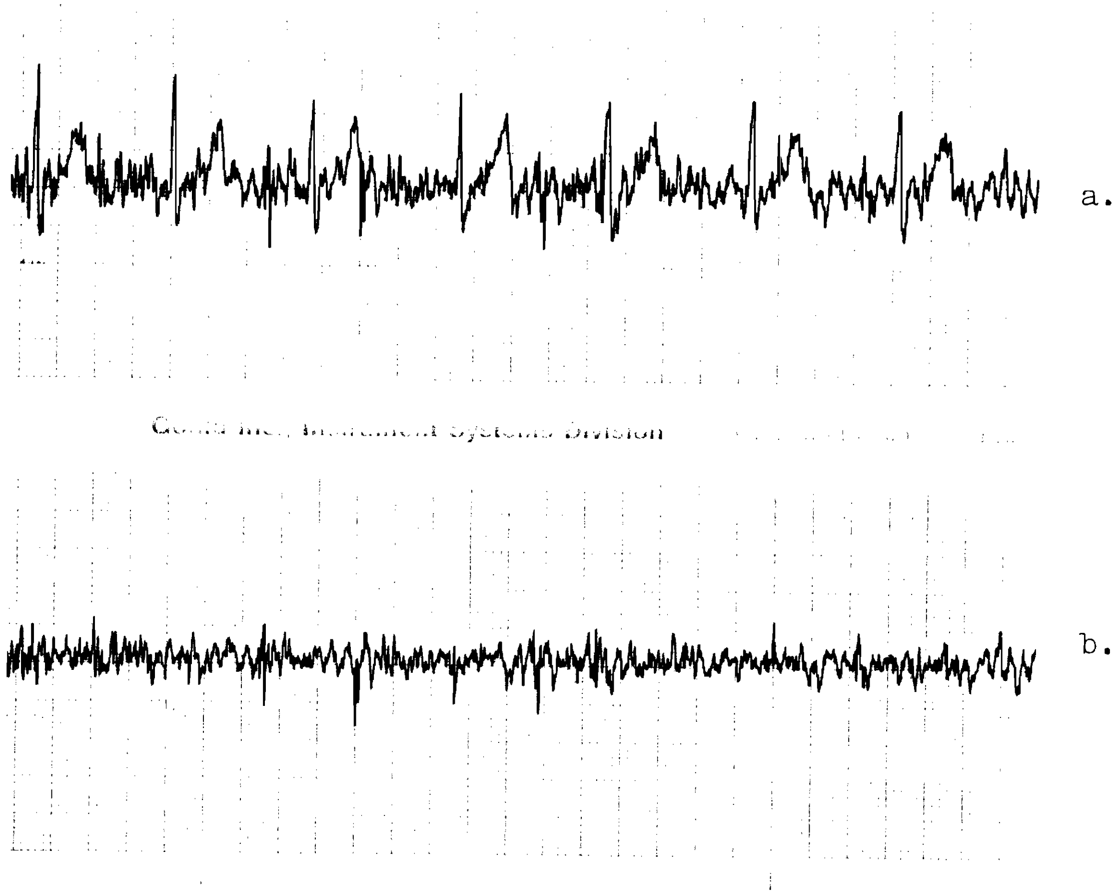


Figure 2. a. ECG Uncompensated tracing; b. Compensating reference signal. (25mm/sec, 1mv/1.8cm)

Figure 3a is a tracing of an original ECG waveform prior to modulation and recording. Figure 3b is the recovered ECG coming out of the circuitry that subtracts the compensating signal (Fig. 2b) from the ECG signal (Fig. 2a).

a.

b.

Figure 3. a. Original ECG signal; b. Compensated ECG playback waveform. (25mm/sec., a. 1mv/cm, b. 1mv/1.3cm)

More distortion was encountered than expected in the final output (Fig. 3b). It was subsequently determined that this was due to a difference in amplification of the demodulated ECG signal as compared to that of the compensating signal. Once this was corrected, the tracings illustrated in Figures 4 and 5 were obtained.

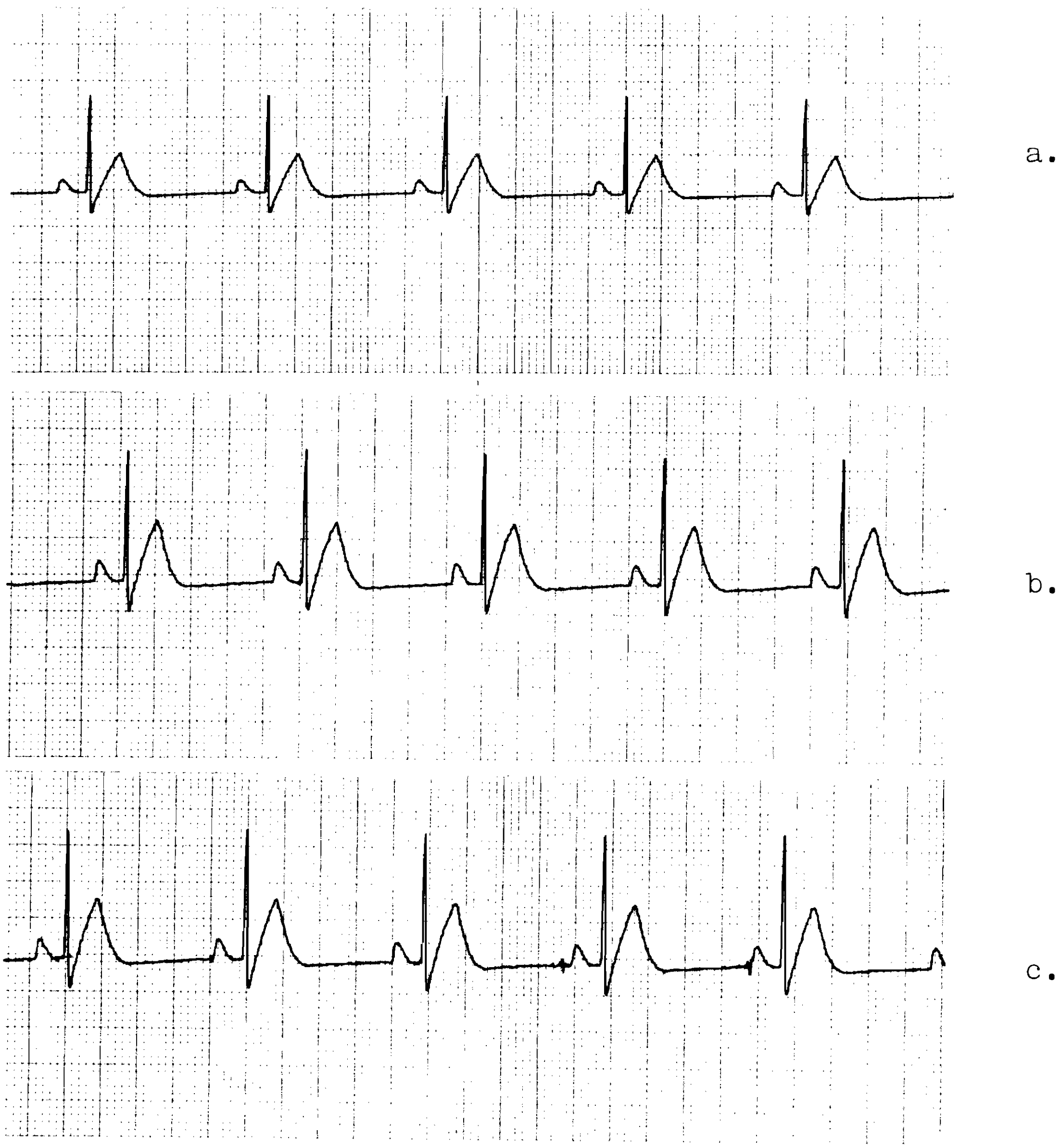
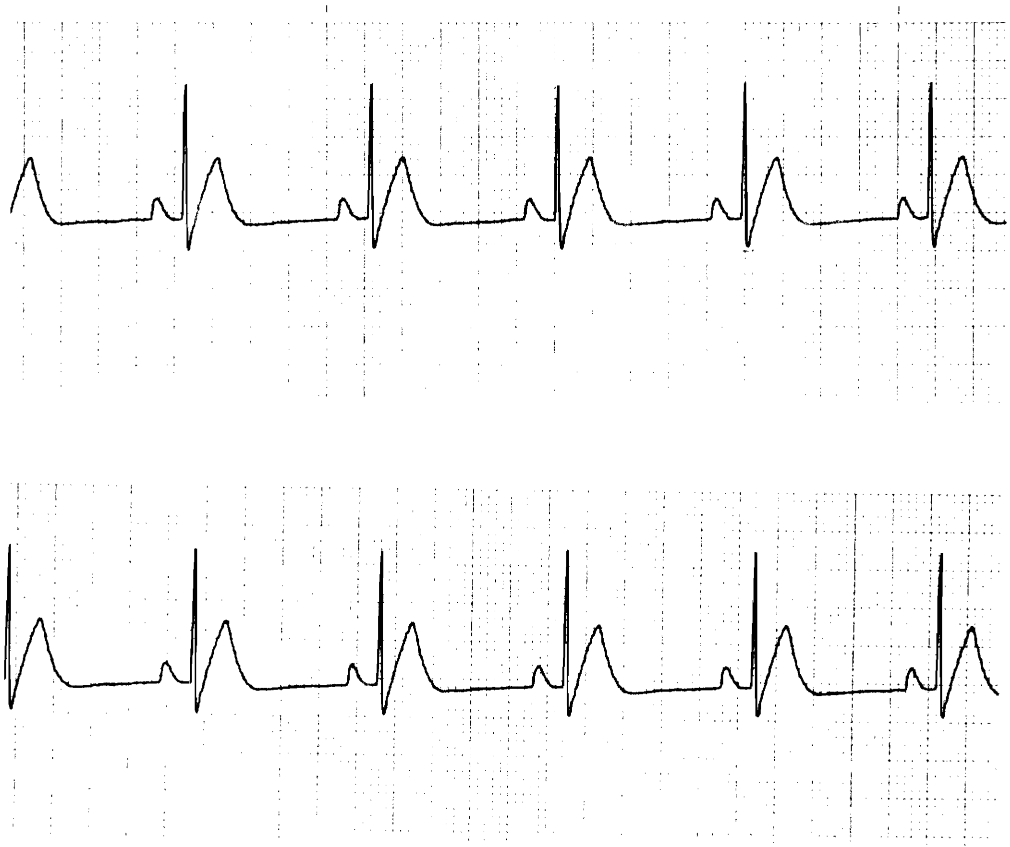


Figure 4. a. Original simulated ECG signal; b. Compensated playback with Concord recorder and inexpensive tape; c. Compensated playback with Concord Recorder and 3M tape. (25mm/sec., a. 1mv/2.2cm, b. and c. 1mv/3.0cm)



a.

b.

Figure 5. a. Compensated playback with Sears recorder and inexpensive tape, b. Compensated playback with Sears recorder and 3M tape. (25mm/sec., a. and b. 1mv/3.0cm)

Figures 4 and 5 also illustrate differences between two different cassette recorders (Sears, model 174.34940000; and Concord, model F-26); as well as the differences between an inexpensive recording tape and a more expensive 3M tape. All playback outputs were passed through a 48 Hz low pass filter. These tracings illustrate that the more expensive recorder (Concord) as well as the more expensive tape (3M) are capable of recording higher frequencies which appear as noise in the final demodulated, compensated ECG tracings.

IV. MATERIALS AND METHODS

For FM demodulation phaselock loop (PLL) can be used with superior performance to that of a conventional discriminator.[14] The basic phaselock loop structure is shown below.

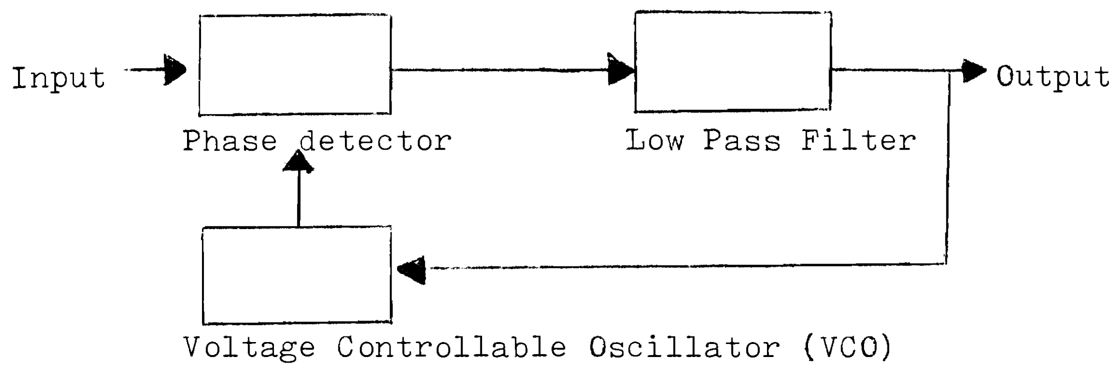


Figure 6. Phaselock loop structure.

The VCO operates at a set free-running frequency. When an input signal is applied to the system, the phase comparator compares the phase and the frequency of the input with the VCO frequency and generates an error voltage that is filtered, amplified, and applied to the control terminal of the VCO. This control voltage forces the VCO to shift its frequency to match that of the input.

If the frequency of the incoming signal is sufficiently close to that of the free-running VCO value, the VCO synchronizes, or locks, with the incoming signal. The PLL then tracks the frequency changes of the input signal, with the error voltage generated from the phase detector becoming the actual FM demodulated signal. The "lock range" is the range of frequencies over which the PLL can maintain lock with the input signal, and the "capture range" is the frequency with which the PLL can acquire lock with an incoming signal (the capture range is smaller than the lock range).[14]

The PLL has the distinct advantage of functioning as a self contained receiver by combining the functions of frequency selectivity (bandpass filter), and demodulation[15], thereby eliminating the need for bandpass filters preceding the PLL demodulators, as in conventional discriminators. Phase lock loop receivers are used to recover a signal deeply embedded in noise.[14]

The PLL receiver also has a distinct disadvantage. It will lock to both harmonic and subharmonic signals of the fundamental frequency; therefore, care must be taken in choosing carrier frequencies for the three analog signals.[15]

The VCO of the phaselocked loop can be used separately as a precision voltage controllable waveform generator (FM modulator). Because of its similarity to the PLL, it lends itself well to be used in conjunction with PLL as a FM modulation-demodulation system.[15]

Early studies used Signetics linear integrated circuits: function generator (566), for frequency modulation; and phase locked loop (565) for demodulation. The 565 phase locked loop integrated circuit is advertised to have extreme stability of center frequency, a wide range of operating voltage (±5 to ±12 volts), very high linearity of demodulated output, and center frequency programming by means of a resistor, capacitor, voltage or current.[16] Figure 7 illustrates a typical connection diagram for FM demodulation. The free running center frequency is determined by the values of

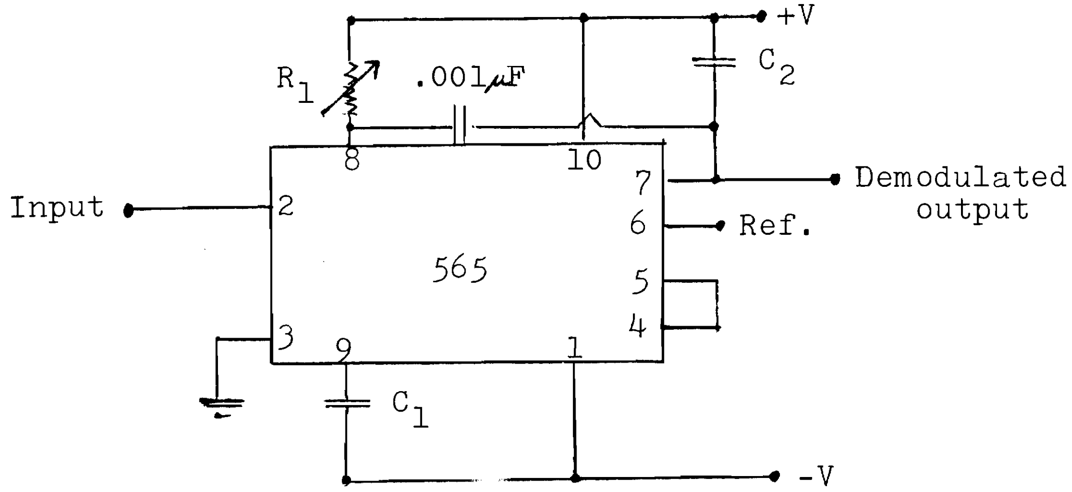


Figure 7. Signetics 565 PLL demodulation diagram.

R_1 and C_1, and is approximately equal to $1.2/4R_1C_1$. Capacitor C_2 forms part of a low-pass filter which determines the capture characteristics of the phase locked loop. The .001μF capacitor connected between pins 7 and 8 acts to eliminate oscillation in the control current source. The jumper wire between pins 4 and 5 connects the VCO to the phase comparator. When in lock, the average dc level of the phase comparator output signal, pin 7, is directly proportional to the frequency of the input signal.

The 566 integrated circuit is the voltage controlled oscillator portion of the 565.[9] This VCO features a wide range of supply voltage (10 to 24 volts), excellent linearity of modulation (0.2%), and excellent frequency stability.[16] Figure 8 illustrates a typical connection diagram for FM modulation.[16] The input control terminal

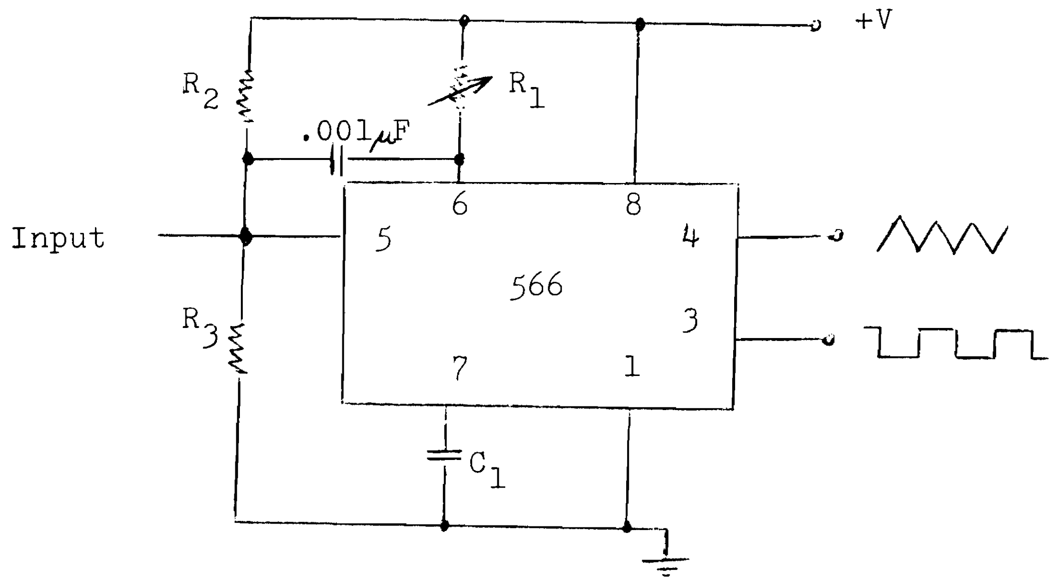


Figure 8. Signetics 566 VCO connection diagram.

(pin 5) must be biased with resistors R_2 and R_3 to a voltage (Vc) in the range $3/4 \leq Vc \leq V+$. The output frequency is then approximately $f_o=2(V+ - Vc)/R_1C_1V+$. A triangular wave output is available at pin 4 and a simultaneous square wave output appears at pin 3.

The third linear integrated circuit to be used is the operational amplifier (741 frequency-compensated). This component, valuable for its diverse configurations and applications, is utilized for amplification, filtering, and multiplexing.

The overall scheme calls for the design of two separate component packets. The first to be used by the patient, along with the cassette recorder, to make the appropriate ECG recordings. This package contains an ECG amplifier, three VCO's, and a multiplexer; and is dc powered by rechargable batteries. The amplified ECG signal is frequency modulated by the first VCO. The second VCO modulates the reference signal, and the third VCO modulates an episode signal (negative step). The three FM signals are then multiplexed (added) and fed to the input of the recorder.

Care was observed in choosing components for this packet that have low power dissipation, since it is battery powered. The 566 VCO has a power dissipation of 300 mW,[16] and the operational amplifiers have a typical power dissipation of 50 mW. The system therefore requires 1.05 watts of power. Small nine volt nickel-cadmium rechargable batteries have a capacity of 70 ma hours giving a useful period of at least 1 hour continuous power supply.

The second component is utilized by the physician upon playback for discrimination, and contains three PLL demodulators. The demodulated reference signal is subtracted from the demodulated ECG signal as well as from the demodulated episode signal, utilizing operational amplifier circuitry. This component packet is ac powered.

The center carrier frequencies were chosen to reduce harmonic interaction while being far enough apart so that capture by a PLL of adjacent center frequency does not occur. The carrier frequency for ECG transmission is 2.5 KHz, for the reference signal is 4KHz, and for the episode signal is 6KHz.

In general the goals for circuitry design are to use state of the art technique; to use a minimum of components; to use as inexpensive components as possible, yet retain quality signal reproduction; and to design the circuitry such that it can be used with most common cassette recording systems.

V. RESULTS AND TESTS

The first step in design, construction, and testing was to design an ECG amplifier and the corresponding VCO FM modulator, and PLL demodulator (at 2.5 KHz center frequency). It was hoped that a single operational amplifier (one stage) differential, bandpass, ECG amplifier could be constructed. This however, proved difficult because of large, ac coupling capacitors needed and problems obtaining a satisfactory common-mode rejection ratio. A two stage amplifier proved to be more practical.

The Signetics 566 VCO was easily constructed with a variable resistor being the means of adjusting the center frequency. The triangular wave output of pin 4 was viewed on an oscilloscope and the potentiometer adjusted until the triangular wave period was .4ms (2.5 KHz). The VCO was tested with a variable voltage source at the input and was shown to have a voltage to frequency characteristic of approximately 1 KHz per volt of input. Therefore, if an ECG amplifier with a gain of 500 was coupled to this VCO, an initial input of ± 2 mv (the desirable deviation range) to the amplifier would result in a frequency variation of ± 1 KHz at the VCO output.

The Signetics 565 PLL demodulator was also easily constructed and the center frequency adjusted in a similar manner as that of the VCO while observing the triangular wave output of pin 9. After several trials it was determined that the PLL would demodulate an unbiased sinusoidal or square wave with a peak to peak amplitude in the range of .1 to 1 volt. This PLL was shown to have a capture range of 1.9KHz to 3.0KHz and a lock range of 1.0KHz to 3.9KHz. Figure 9a illustrates the output from a two stage ECG amplifier (gain of 1000) which is the VCO input. Figure 9b is the corresponding demodulated PLL output (the VCO square wave output being the PLL input) which has been passed through a 100 Hz low-pass filter and amplified 45X.

a. b.

Figure 9. a. Original simulated ECG signal.
b. Same signal after modulation-demodulation processing.
(25mm/sec., a. 1mv/.5cm, b. 1mv/.6cm)

Excellent FM modulation-demodulation linearity was observed.

The next step was to construct VCO's and PLL's for the two other signals at 4KHz and 6KHz carrier frequencies. The three FM multiplexed signals were applied to the three PLL inputs to see if the original analog signals could be recovered upon demodulation. It was at this point that difficulties were encountered. Figure 10 illustrates the most notable problem occurring in the demodulated ECG waveform: the QRS complex is distorted.

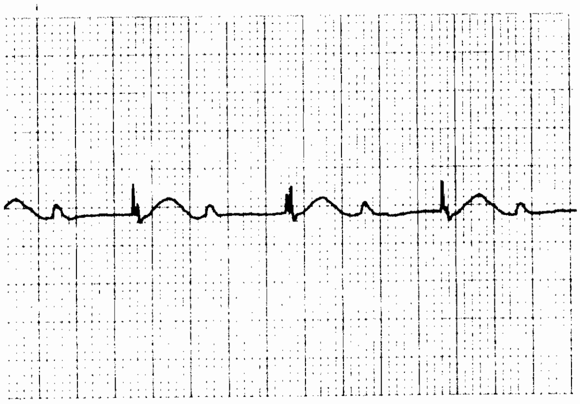


Figure 10. Distorted ECG waveform. (25mm/sec., 1mv/.8cm)

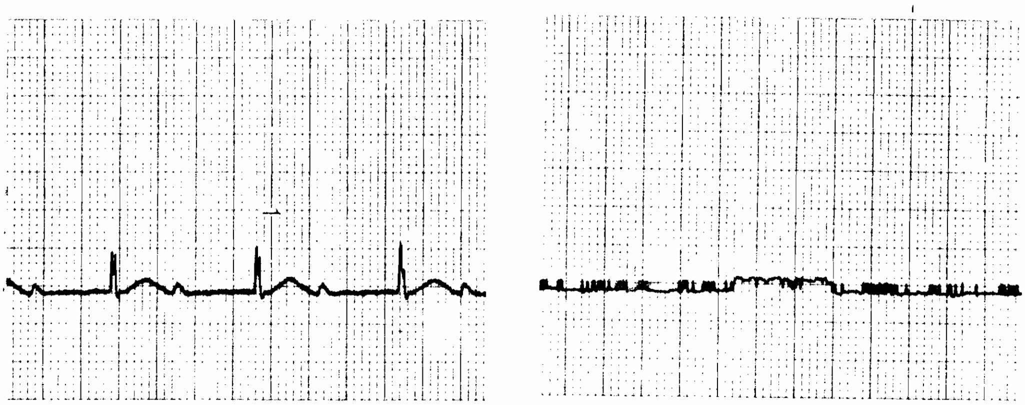


Figure 11. Distorted ECG and Episode Signals upon playback from Sears recorder. (25mm/sec., 1mv/.8cm)

The same multiplexed signal was recorded on the Sears recorder and the playback applied at the PLL inputs, giving the tracings illustrated in Figure 11. Significant distortion occurs in the ECG as well as the episode signal.
It appears that the 6 KHz PLL receiver has difficulty locking to the episode FM modulated signal.

A more thorough study of the bandpass characteristics of the Sears and Concord cassette recorders was made and is illustrated in Figure 12. The input level is a 6 volt peak to peak sinusoidal wave and the playback peak to peak level at various frequencies of interest is graphed on the chart. The Sears recorder exhibits very poor frequency characteristics with a continuous rolloff and noticeable attentuation at 6 KHz. This problem could possibly be solved by emphasizing the 6 KHz and 4 KHz carrier frequencies relative to the 2.5 KHz carrier frequency prior to recording. However, since the Concord recorder bandpass characteristics are much better, it was decided to design the recording circuitry to be used in conjunction with the Concord cassette system, with little sacrifice in price difference.

The distortion of the QRS complex was not satisfactorily alleviated by the changing of the ECG carrier frequency to different positions and proved to be a more serious problem. The problem finally however, was isolated. The multiplexed signal which serves to add noise to the signal input of the PLL's, acted also to degrade the performance of the 566 PLL's by reducing the lock range of the PLL receiver to

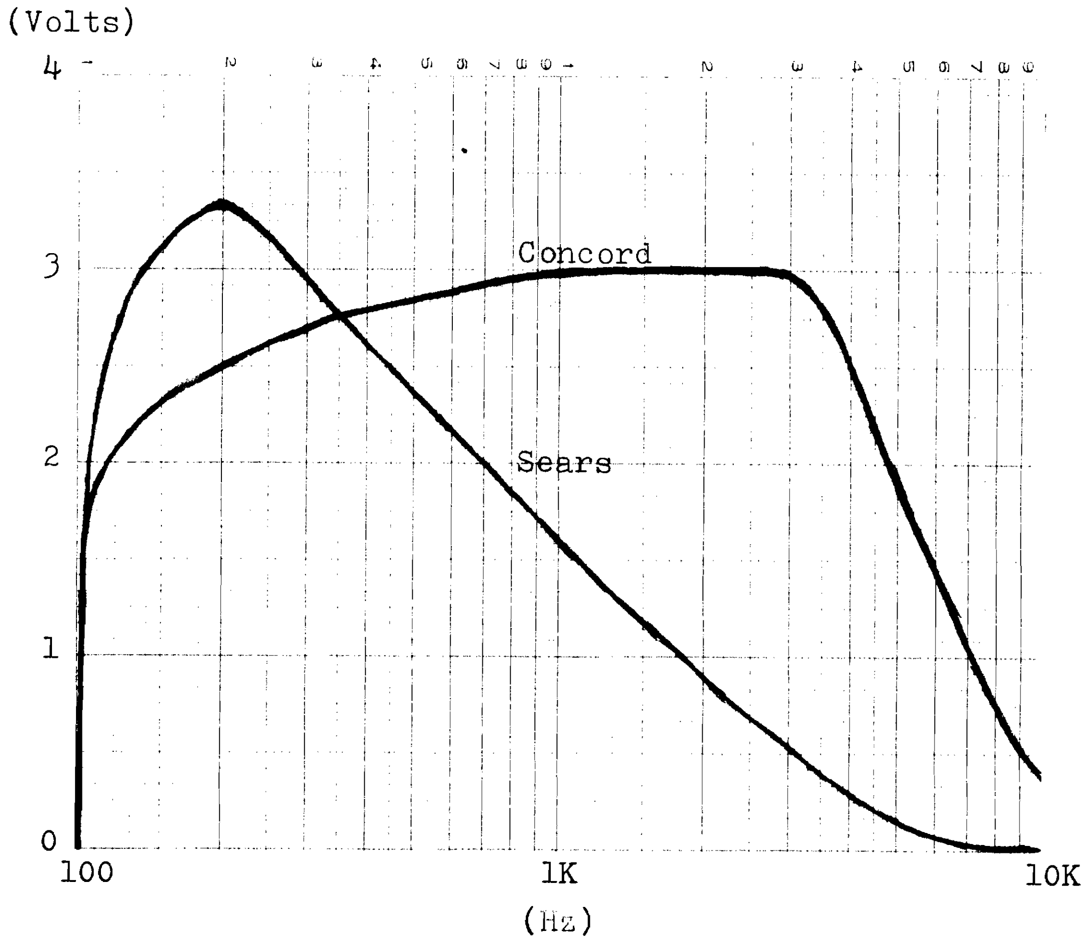
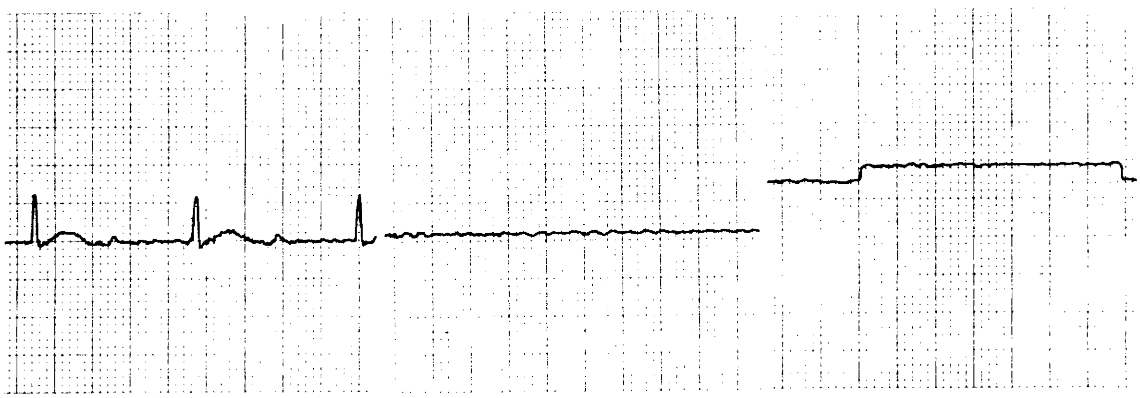


Figure 12. Bandpass characteristics of Sears and Concord cassette recorders.

that approximately equal to the capture range. Thus the PLL demodulator would loose lock with the greater frequency variation of the QRS complex signal when the frequency variation exceeded that of the capture range rather than that of the lock range as anticipated. The solution to this problem was to decrease the frequency variation of the ECG modulated signal such that the PLL receiver would remain in lock with the ECG QRS complex FM signal variation. The problem solution therefore meant decreasing the gain of the ECG amplifier.

Since the design criteria requires that the FM system monitor a voltage variation of ±2 mv at the input to the ECG amplifier; and since the lock and capture range of the PLL that demodulates the ECG signal has been reduced to a total range of 1 KHz receiving a multiplexed signal; and since the VCO voltage to frequency transfer characteristic is 1 KHz per volt input; the gain of the ECG amplifier must necessarily be reduced to 250 (4 mv times 250 equals 1 volt). The demodulated ECG signal must then be amplified to a suitable level for output monitoring.

Figure 13 illustrates recorded playbacks of the three signals (ECG, compensation, episode) using the Concord recorder and an ECG amplifier with a gain of 250.



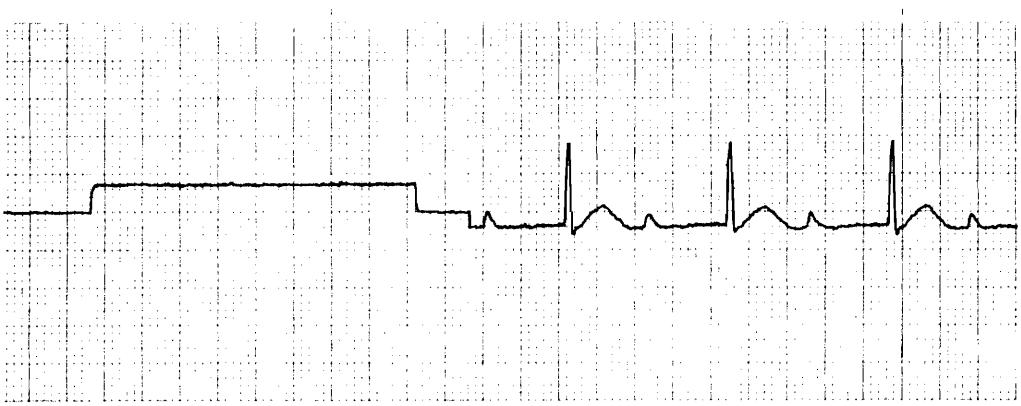
a. b. c.

Figure 13. a. Playback of a simulated uncompensated ECG signal; b. Playback of the compensation signal; c. Playback of the episode signal. (25mm/sec., 1mv/.5cm)

It will be noted that the distortion due to variable tape speed is greatly reduced over that of the tracings of Figure 2. This is due to the fact that this system has a greater voltage to frequency transfer characteristic than that used in the feasibility study. Therefore, the sensitivity to flutter and wow is greatly decreased with this FM system.

Figure 14. illustrates a tracing in which, upon playback, the compensation signal has been subtracted from the episode and ECG signals. The amplification of the demodulated signals is 27X. This tracing is clinically acceptable with a signal to noise ratio of approximately 80/1.

The episode and ECG signals of Fig. 14 were amplified after demodulation by the same factor (27X); in the actual playback circuitry (Fig. 16) the two signals will have separate amplifiers, with the episode signal amplified to a voltage greater than 2 volts in order to actuate a light emitting diode and marker on the ECG recorder.



a. b.

Figure 14. a. Compensated episode signal; b. Compensation of a simulated ECG signal. (25mm/sec., 1mv/cm)

VI. CIRCUITRY

The elements of the component packet for ECG recording are illustrated in Figure 15. Operational amplifiers A1 and A2 comprise the ECG amplifier with a gain of 250. The first stage (A1) is a differential amplifier with a gain of 5. Resistors R1 and R2 comprise the input impedance of 200K ohms for the ECG amplifier. Variable resistor R5 is for common-mode rejection adjustment. The common mode rejection ratio of this two stage amplifier is 86 db.

Stage A2 is a bandpass amplifier with a gain of 50. Capacitor C1 and Resistor R6 are a high pass filter with the 3 db point at .5 Hz. Capacitor C1 also provides ac coupling. Capacitor C2 and resistor R7 are a low pass filter with a 3db point at 100 Hz. Resistor R9 is adjusted to bias the output of stage A2 to match the input bias of VCO1.

Center frequency programming of VCO1 is determined by the component values of capacitor C4 and resistors R12 and R13. Fine tuning for a carrier frequency of 2.5 KHz (.4 ms period) is accomplished by adjusting potentiometer R12 (as previously described). Resistors R14 and R15 form a voltage divider reducing the square wave output amplitude from pin 3 to about 100 mv peak to peak. This output can be fed directly to input pin 2 of the corresponding PLL for a demodulation check.

Voltage controlled oscillator 2 provides the reference frequency signal. Finetuning of the center frequency of 4 KHz (.25ms period) is accomplished by adjusting potentiometer R18. The square wave output from pin 3 is also passed through a voltage divider (R20 and R21) to produce a peak to peak output of about 100 mv.

The episode signal is modulated by VCO3. Closing the episode switch places R23 in parallel with R24 producing an effective resistance of 9.5K ohms. This alters the input bias scheme with a subsequent voltage input change of about 60 mv, and a change in output frequency from 6KHz to 6.3 KHz. Fine tuning is accomplished by adjustment of R25 to achieve a 6 KHz center frequency (.17ms period).

The three modulated signals are fed into a summing amplifier (A3) producing a multiplexed signal of about 300 mv peak to peak amplitude. This signal goes to the cassette recorder input for recording.

Figure 16 illustrates the components of the playback circuitry. Because of the high frequency rolloff characteristics of the recorder, the square waves generated by the VCO's are converted to sinusoids. The Fourier components of the square waves are the fundamental and odd harmonics. The third and higher harmonics are eliminated by the recorder characteristics. This signal is fed simultaneously to the input of all three phase locked loop receivers. Receiver PLL1 is tuned to a center frequency of 2.5 KHz (.4ms period) by adjustment of resistor R34 while viewing the triangular wave output of pin 9 on an oscilloscope. Capacitor C9 effects the capture range of the PLL; the smaller the capacitor value the larger the capture range. The capture and lock range of PLL1 receiving a multiplexed signal is 2.0KHz to 3.0KHz. The demodulated output from pin 7 is the ECG waveform (Fig. 13a).

Receiver PLL2 is tuned to a center frequency of 4KHz (.25ms period) similar to that of PLL1 by adjustment of resistor R35. The larger capacitor C12, as compared to C9 of PLL1, and the resistor between pin 6 and 7 have an effect of reducing the lock and capture range of PLL2. The capture and lock range of PLL2 receiving a multiplexed signal is 3.6KHz to 4.3KHz. The output from pin 7 is the compensation signal (Fig. 13b).

Receiver PLL3 is tuned to a center frequency of 6KHz (.17ms period) similar to the previously mentioned method by adjustment of resistor R37. This PLL also has the reduced lock and capture configuration with capacitor C15 and the resistor between pin 6 and 7. The capture and lock range of PLL3 receiving a multiplexed signal is 5.7KHz to 6.4KHz. The demodulated output from pin 7 is the episode signal (Fig. 13c).

The compensation output from PLL2 is fed into both the

ECG (A4) and the episode signal (A5) differential amplifiers. This compensates for the output bias of about +4 volts from pin 7 of the PLLs. Amplifier A4 with a differential configuration subtracts the compensating signal from the ECG waveform. Resistor R40 acts to reduce slightly the amplitude of the compensating signal as compared to that of the ECG signal. Capacitors C17 and C18 plus resistors R41 and R42 form a low pass filter with a 3 db point of 50 Hz. The gain of amplifier A4 is 30 which provides an output of 1 mv/cm (4 volts per cm) relative to the original input to the ECG amplifier of the recording packet. As illustrated, amplifier A4 will likely have a biased output which can be compensated for by placing a resistor from the appropriate voltage supply to the inverting input of A4.

Differential amplifier A5 amplifies the episode signal with a gain of 267 to amplifier saturation, when actuated, which in turn lights the light emitting diode.

The original prototype design as supplied with ±6 volts power supply which works well for the recording stage. The playback component with a power supply of ±6 volts limits the output of the ECG signal (amplifier A4) resulting in clipping due to amplifier saturation. A power supply of ±9 volts is recommended.

The bandpass characteristics of .5 Hz (A2) high pass filter and the 50 Hz (A4) low pass filter have a tendency to distort the high and low frequency-components of the ECG waveform over ECG recordings using the conventional bandpass requirement of .05 Hz to 100 Hz. However, with the presently designed cassette ECG recording system, the primary concern is to monitor the rate and rhythm changes which are basically uneffected by the reduced bandwidth of .5 Hz to 50 Hz. This reduced bandwidth is routinely used by the Latter-day Saint Hospital in Salt Lake City for ECG recordings since it provides the advantage of minimizing low frequency drift, attenuates 60 Hz line interference, and minimizes skeletal muscle artifact.

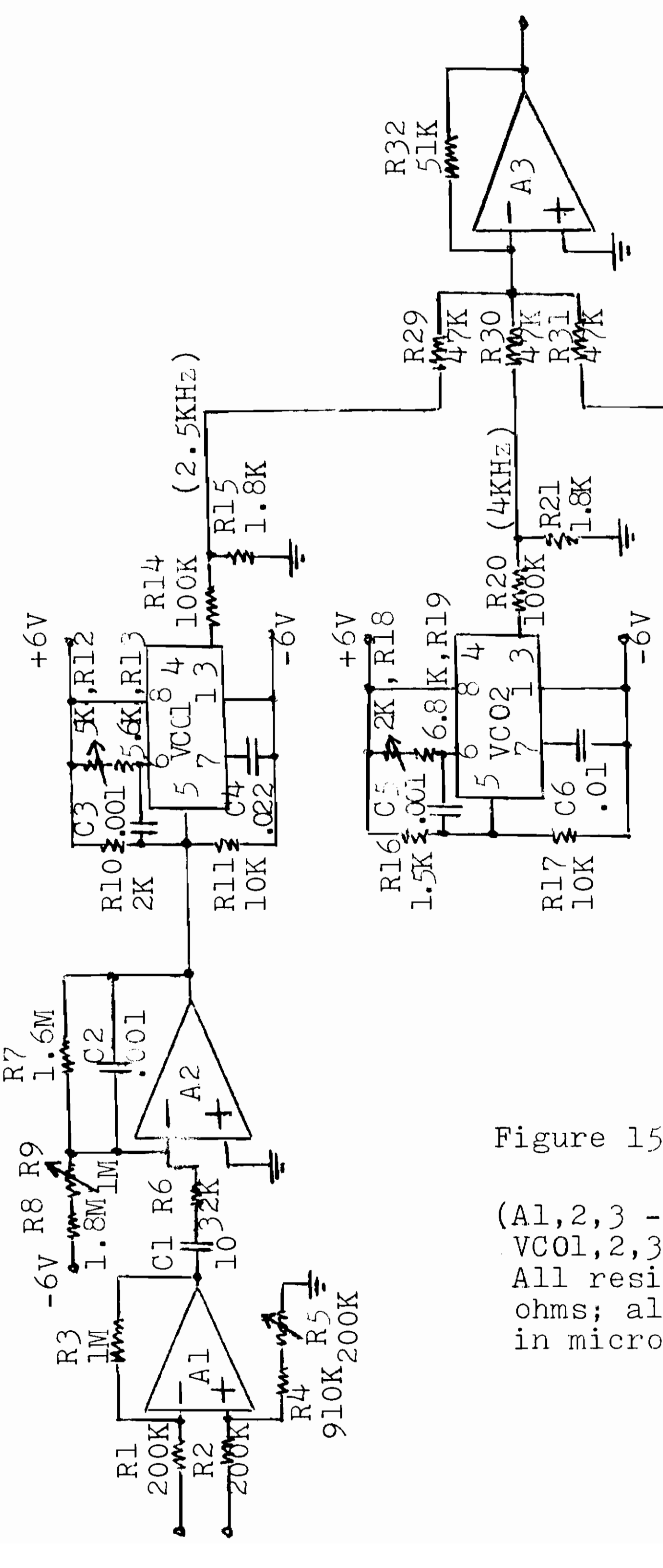


Figure 15. Recording component schematic.

(A1,2,3 - Fairchild OP. Amp. VCO1,2,3 - Signetics VCO 566; All resistance values are in ohms; all capacitor values are in microfarads.)

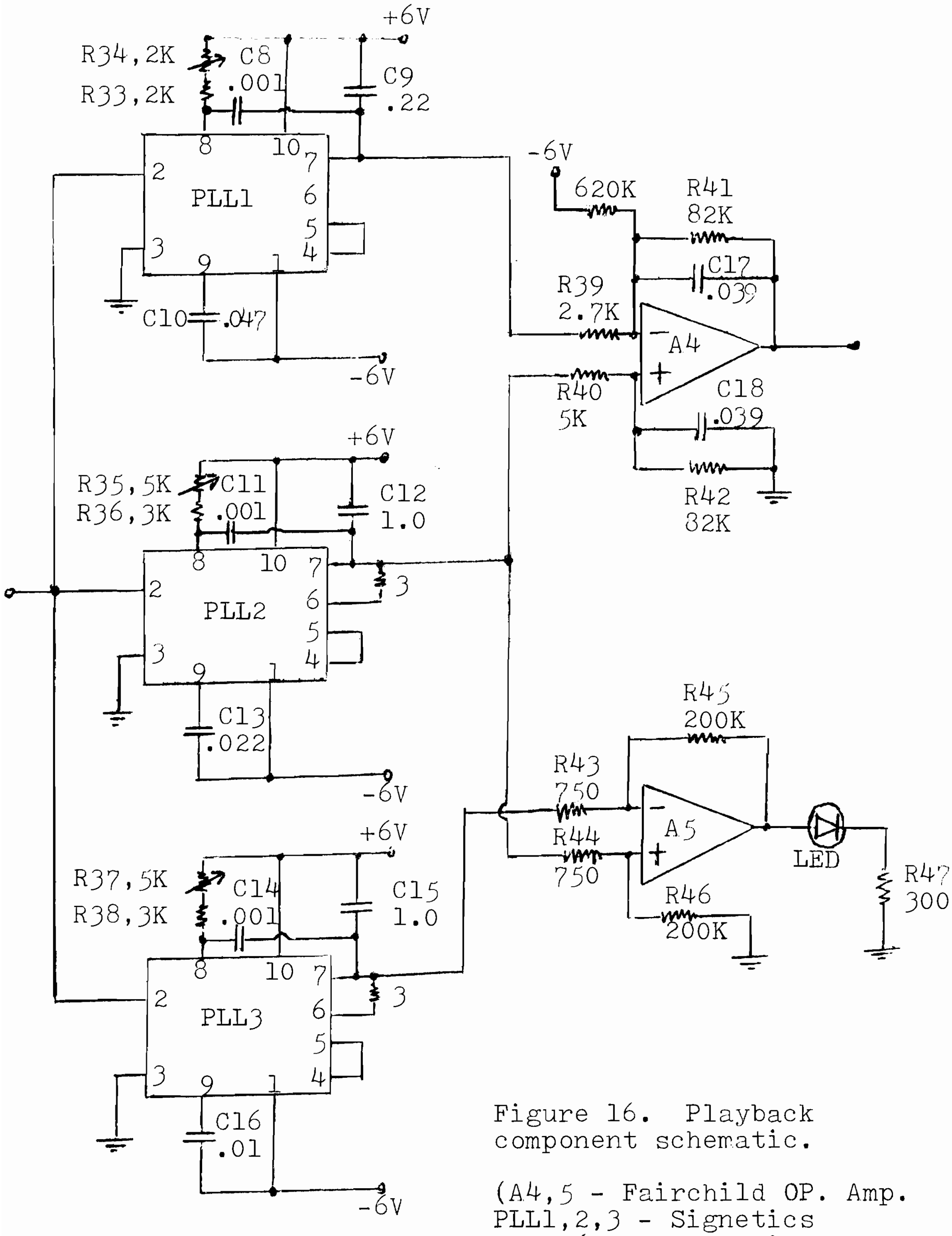


Figure 16. Playback component schematic.

(A4,5 - Fairchild OP. Amp. PLL1,2,3 - Signetics PLL 565. All resistance values are in ohms; all capacitor values are in microfarads.)

VII. SUMMARY AND CONCLUSIONS

Circuitry utilizing integrated circuits has been designed as peripheral equipment to a cassette recorder for inexpensive take-home cardiac monitoring. The circuitry is for frequency modulation for recording and demodulation upon playback of three synchronous multiplexed analog signals: the ECG signal, a reference signal which after demodulation is subtracted from the ECG signal to compensate for variable tape speed, and an episode signal actuated by the patient during ECG recording to indicate a particularly noticeable episode.

The original goals were to keep the total component cost under $100.00, to produce clinically acceptable ECG tracings, and to design the components such that they could be used with most common battery powered cassette recording systems. The components, excluding the cassette recorder, can be purchased for less than $60.00. An expensive item is the cassette recorder which has considerable variations in price ($40.00 - $100.00). The circuitry described in this thesis has been designed to work with the Concord Model F-26 cassette recorder and other cassette recorders of similar quality.

The quality (signal to noise ratio) of the ECG reproduction, although acceptable, could probably have been improved were it not for the degradation of the phase lock loop FM demodulator with an input of a multiplexed signal, which reduced the lock range; and therefore required a reduction in the gain of the ECG amplifier.

The design goal of the project was achieved since the recorder and components can be purchased for less than $100.00 (in single quantities) and the system provided acceptable ECG's for clinical use.

The knowledge and experience gained from this thesis project has been of considerable value. Hopefully this work will lead to further experiments, improvements, and other applications of physiological monitoring with an inexpensive cassette recording system.

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