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A study and design of a cardiac interbeat interval timer system

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A STUDY AND DESIGN OF A CARDIAC INTERBEAT
INTERVAL TIMER SYSTEM

BY

RICHARD WILLIAM MOONEY, 1943-

A THESIS

Presented to the Faculty of the Graduate School of the

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(Advisor)
ABSTRACT

This thesis reports the design of a system which can measure and record the cardiac interbeat times of human subjects in a continuous beat to beat manner, using the bioelectrical signal associated with the heart as an input. The design is in response to the needs of clinical researchers.

The means of obtaining the bioelectrical signal and previous designs of electrocardiographic equipment, especially electrocardiotachometers, are reviewed. A design is proposed which consists of a preamplifier to amplify the bioelectrical signal, a 60 Hz rejection filter, and a beat detector which uses a differentiation technique to obtain a reliable single output pulse for each cardiac cycle. The detector output pulse is used to control the input of a clock pulse train to a five decade counter. The pulses within the train are repeated at a rate of $10^4$ pulses per second. The output of the counter is stored in memory circuits at the end of each cardiac excitation interval, and the counter is reset for the next interval. The output of the memory circuits is recorded with a digital printer, yielding a printed record of interbeat times.
ACKNOWLEDGEMENT

The author acknowledges the help and guidance of his graduate advisor, Dr. Herbert Crosby, in preparing this thesis. Also the author expresses his thanks to Dr. Michael Kaplan, a Clinical Psychologist and presently Principle Scientist of the U. S. Army Behavior and Systems Research Laboratories, for his time, explanations, and encouragement in pursuing this study.
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I INTRODUCTION

In 1903, Dutch psychologist William Einthoven devised a string galvanometer which for the first time enabled measurement and recording of the electrical signals arising within the human body. The signal which Einthoven was able to record was the bioelectrical signal associated with the heart and normally initiating the frequency of the heart's mechanical pumping. Since that time, advances in both equipment design and medical knowledge have allowed recording and usage of this signal, commonly referred to as the ECG (electrocardiogram), in valuable medical research and diagnosis.

The repetition rate of the ECG signal and the inverse of this rate, the interbeat times, have been found to vary with time, condition, stress, breathing, and other factors. This variance with stress and human condition has led psychological research laboratories to further investigate the effects of external stimuli upon the interbeat times.

Dr. Michael Kaplan, a Clinical Psychologist and presently Principle Scientist of the U.S. Army Behavior and Systems Research Laboratories, is one researcher interested in this particular area. The major obstacle he has encountered in this research is the lack of proper measuring equipment to provide beat intervals on a continuous beat to beat basis. For meaningful research, each measured interbeat interval must be accurate to within ± 1 millisecond.
Several previous attempts by research laboratories to obtain the required interbeat data have proved inadequate. One method consisted of using a graduated scale to measure the interbeat intervals from an analog ECG recording. This method proved unacceptable because of the obvious inadequacy of using eyesight and a graduated scale for precision measurement, and because of uncertainty about the characteristics of the chart drive mechanism of the recorder. A second method consisted of using a commercially available cardiotachometer which used a voltage peak detector circuit and a function generator to provide a voltage output which was supposedly proportional to the cardiac rate. This method proved unacceptable because slight variations in normal ECG signals would sometimes present several voltage peaks per beat and sometimes no peak per beat which the cardiotachometer would detect. Also, the function generator's output voltage was not truly proportional to beat frequency.\(^{(2),(3)}\)

In response to the need and problems of these researchers, the purpose of this thesis is to report the design of a cardiac interbeat timer system capable of providing interbeat interval data continually on a beat to beat basis with an accuracy of at least \(\pm\) 1 millisecond per interval.
II BACKGROUND ON THE CARDIAC BIOELECTRICAL SIGNAL

A. Detection of ECG Signal

The bioelectrical signal associated with the heart can be obtained from the body tissues through an electrode interface to provide the required input signals to amplifying circuits. The research projects which need cardiac interbeat time data usually obtain the input ECG signal using a standard, body surface, bipolar electrode placement, consisting of two active electrodes and a reference electrode. The two active electrodes furnish the inputs to a differential amplifier. The electrode which provides the non-inverting input to the amplifier is commonly identified as the positive electrode, and the electrode which provides the inverting input to the amplifier is identified as the negative electrode. A commonly used electrode placement consists of the positive electrode on the subject's left arm, the negative electrode on the right arm, and the reference electrode on the right leg. Another electrode placement consists of the positive electrode positioned between the ninth and tenth ribs (ninth intercostal space) of the subject's left side, the negative electrode on the upper center chest (top of sternum), and the reference electrode at a convenient spot between the ribs of the subject's lower right side. The latter electrode placement was used throughout the design testing reported in this thesis. The elec-
trodes, which were used during testing, were Beckman Instruments Incorporated silver and silver chloride electrodes (Type 650437), electrically attached to the body surface by means of Beckman electrode paste (Type 201210).

B. Description of ECG Signal

The ECG signals of normal, healthy subjects have been studied and analyzed for many years. A sketch of a typical normal ECG signal obtained by using the electrode placement described in the previous paragraph is presented in Figure 1.

The letters used in Figure 1 to identify the various portions of the waveform (P, Q, R, S, T, and U) have become standardized symbols for referring to the waveform. Some of the acceptable normal amplitudes and time intervals between the various peaks are also summarized in Figure 1. The time interval of most interest in this study, the cardiac interbeat interval, may vary normally from 0.2 seconds to 2 seconds.\(^{(4),(5),(6)}\)

Along with the desired bioelectrical signal, an undesired differential DC signal as great as 100 millivolts may be present between the two active electrodes. Also, there may be an undesired DC signal common to both active electrodes as referenced to the isolated ground of the input amplifier, and other common signals due to electromagnetic or electrostatic inductions usually at a frequency of 60 Hz.\(^{(7)}\) These undesired signals must be effectively
Notes on ECG Amplitudes and Durations:

1) P Wave - Average amplitude of 0.1 mv (0.02 mv minimum to 0.3 mv maximum). Average duration of 90 ms (70 ms minimum to 120 ms maximum).

2) Q Wave - Average amplitude of -0.06 mv (-0.01 mv minimum to -0.3 mv maximum).

3) R Wave - Average amplitude of 1.0 mv (0.5 mv minimum to 2.6 mv maximum). Average duration of 40 ms (rise time of 20 ms and fall time of 20 ms).

4) S Wave - Average amplitude of -0.18 mv (-0.01 mv minimum to -0.9 mv maximum). First portion of wave is continuation of negative slope of R wave.

5) U Wave - Amplitude less than 0.03 mv.

Figure 1. Normal Electrocardiograph (ECG) Signal Obtained Using Bipolar Leads.
eliminated by the input amplifier to obtain a useful ECG signal.
III REVIEW OF LITERATURE

A review of technical literature published during the past thirty-five years did not reveal any reports on a cardiac interbeat timer. However, the problem of obtaining the heart rate, the inverse of the interbeat times, was the subject of a number of studies resulting in designs of heart rate devices known as cardiotachometers. Because of the similarity of functions of an interbeat timer and a cardiotachometer, the literature on the latter was carefully reviewed.

The earliest electronic cardiotachometer noted in this review was reported in 1938 by J. Warren Horton. His design used vacuum tube technology and required five stages of amplification to obtain an ECG signal. The peak amplitude of the signal was used as a trigger to alternately activate two relays. Capacitors in series with the relay contacts were alternately charged through the relay contacts and then discharged through a discharge circuit. The current in the discharge circuit was measured by means of a meter calibrated in beats per minute (BPM) with a range from 30 to 240 BPM. A critical evaluation by Horton of his own design recognized problems with 60 Hz noise and unsteadiness in meter readout as well as the requirement of critical adjustment to prevent self oscillation.

The designs of cardiotachometers during the twenty years following the publication of Horton's article were
not far different from his concept. A. W. Melville and J. B. Cornwall reported in 1958 on a different concept for obtaining the rate output.\(^9\) Their design used the amplified ECG as the input to an amplitude-triggered circuit which initiated a sequence of relay closures. During the interval between trigger pulses, a capacitor was charged through relay contacts so that at the next trigger pulse, the charge was transferred to a second, much smaller capacitor, and the first capacitor was then discharged in preparation for the next interval. The voltage across the second capacitor was measured using a high resistance voltmeter with a readout scale calibrated in BPM. The large time constant set by the meter resistance and the output capacitor averaged the output rate over several beats.

In 1961, L. J. Ryan reported on an instantaneous pulse rate monitor, which was quite similar to the design by Melville and Cornwall except that his design used the output capacitor to control the plate current of a tube.\(^{10}\) The plate current was then measured by a meter calibrated in BPM. The design provided a discrete output at the end of each cardiac cycle.

Also in 1961, several other cardiotachometers which deviated from previous methods were reported. S. R. Gilford and M. Marten designed an instantaneous reading cardiotachometer which used the amplitude of the ECG to trigger a function generator, and an on-line computer to obtain heart rate data from the function generator's output voltage.\(^{11}\)
D. W. Hoare and J. M. Ivison reported on their experience and innovations in measuring the heart rates of active subjects. (12) Their design included a dual system of electrodes which required an amplitude-triggered signal in both systems before triggering a monostable multivibrator. The multivibrator prevented false triggering of the device during a preset length of time following each trigger. The output consisted of the pulses recorded on a constant speed pen recorder.

In 1963, M. McDonald and W. J. Perrin reported on a cardiotachometer which used semiconductor design instead of vacuum tubes and relays. (13) Their cardiotachometer was capable of providing instantaneous heart rates at the end of each beat. The design used the amplified ECG signal to trigger a voltage function generator which produced a voltage that increased linearly with time and reset to zero at each trigger. The output was recorded on non-linearly graduated recording paper with the peaks of the output graph yielding the beat to beat heart rate.

In 1965, J. Czekajewski and P. A. Tove reported on their cardiotachometer design which used the amplitude of the ECG signal to trigger control circuits which would alternately charge two capacitors. (14) The charged capacitor was used as an output while the other capacitor was charging, providing an instantaneous readout at the end of each beat on a meter calibrated in BPM. Their design was quite similar in concept to the earlier design of Melville and Cornwall.
except that the active components were semiconductor devices instead of vacuum tubes and relays.

In 1969, David A. Winter and Brian G. Trenholm recognized the problem of obtaining reliable triggering from an ECG. They proposed a system to obtain a reliable output pulse from the QRS complex of the ECG by using an active filter to reduce noise, a non-linear amplifier to amplify just the QRS complex, and a Schmitt-trigger to provide an output pulse. The device appears to be a reliable source for a single output for each beat, but the repeatability of the time of trigger during each cardiac cycle is not assured.

Other published works on cardiotachometers during the previous decade emphasized different aspects of readout mechanisms. All of these used amplitude triggering from the ECG signal with a gradual change from the capacitor charge readout system to a more complicated function generator system to provide readouts which could be calibrated in BPM. Repeatedly, the problems of false triggering were mentioned, and some attempts to reduce its occurrence were examined. The accuracy of these systems was generally stated as ± 1 or ± 2 BPM, and most used an analog meter readout which assured further human error in the measurement.
A. General Design

A cardiac interbeat timer system which uses the bio-electrical ECG signal as an input and provides continual discrete cardiac interval data is required. The design for this system must provide amplification of the ECG to a useful level, so that a specific portion of the input can be used to obtain a single, reliable output pulse for each cardiac cycle. The time between these pulses can then be measured and printed. All of these functions can be performed by a timer system such as the one outlined in the block diagram of Figure 2.

The interbeat timer system of Figure 2 accepts the ECG signal from electrodes attached to the body surface of the subject and amplifies this signal to a useful level with a preamplifier. The amplified ECG signal is filtered through a 60 Hz rejection filter to reduce any excessive 60 Hz signal which might be present after amplification and interfere with the proper operation of the subsequent stages. The ECG output from the filter becomes the input to the beat detector circuit. This circuit uses a specific portion or slope of this ECG signal to trigger a single output pulse for each cardiac cycle in a manner which eliminates the problems of false or skipped triggering described in Sections I and II that have plagued electrocardiotachometers.
Figure 2. Block Diagram of Cardiac Interbeat Timing System
The output pulses from the beat detector are used to control timing circuitry which must accurately measure the time between beats in a discrete and direct manner. Each beat detector output pulse triggers a sequence of control circuits which operate the timer. A train of clock pulses is the input to a counting circuit and is controlled so that the counter counts the number of clock pulses between beat detector outputs. At each beat detection, the output of the counter is stored in memory circuits, and the counter is reset for the next interval. The count stored in the memory is then printed by a digital printer, yielding a printed record of interbeat times in a continual, beat to beat manner.

B. Differential Input Preamplifier

Although a commercially available differential-input ECG preamplifier could be used to obtain the required bioelectrical signal, a self-contained preamplifier can be inexpensively included as a part of the timer system. A differential-input preamplifier with high common mode rejection, at least 100 db, is required to reject the undesired common signal, usually a 60 Hz signal plus a DC component. The preamplifier must use direct coupled inputs because of the adverse effect on the preamplifier common mode rejection of any imbalance between capacitor used at the inputs, and yet be capable of handling a differential DC level possibly as great as 100 millivolts across the input.
A high differential input impedance is required to prevent distortion of the input signal. A high common mode input impedance is required to minimize the effect of slight resistance imbalance at the inputs. The preamplifier must have a bandwidth between 0.1 Hz and 100 Hz and have a gain of 5000.

1. Preamplifier Circuit

The circuit presented in Figure 3 meets all of the requirements for the ECG preamplifier. This differential-input preamplifier uses integrated circuit operational amplifiers as components and is a modified version of a biological preamplifier reported in the Fairchild Semiconductor Linear Integrated Circuits Applications Handbook. Three Fairchild 741C operational amplifiers were used as a compromise among cost, performance, and availability. The preamplifier directly couples each active electrode to the non-inverting input of a Fairchild 741C operational amplifier to provide the high differential and common mode input resistances.

The circuit of Figure 3 can be analyzed to describe its behavior by using standard operational analysis techniques. The desired biological signal can be defined as $V_S$, and the common signal as $V_{CM}$. The inputs $E_1$ and $E_2$, as labeled in Figure 3, can be determined in terms of $V_S$ and $V_{CM}$ as
Figure 3. Electrocardiograph (ECG) Preamplifier
Because of the symmetry of the amplifier circuit, the inverting inputs of $A_1$ and $A_2$ appear to have a voltage $V_{CM}$ applied through a resistance $R_5/2$. Therefore, the outputs of the two operational amplifiers are

$$E_1 = V_{CM} - \frac{V_S}{2}$$

and

$$E_2 = V_{CM} + \frac{V_S}{2}$$

The first component of equations (3) and (4) resulted from the non-inverting input, and the second component resulted from the inverting input. Substituting for $E_1$ to obtain $e_1$ in terms of the desired signal $V_S$ yields

$$e_1 = \frac{R_3 + \frac{R_5}{2}}{R_5} \left( V_{CM} - \frac{V_S}{2} \right) - \frac{R_3}{R_5} V_{CM}$$

$$e_1 = \left( \frac{2R_3}{R_5} + 1 \right) \left( V_{CM} - \frac{V_S}{2} \right) - \frac{2R_3}{R_5} V_{CM}$$
Similarly,

\[ e_1 = V_{CM} - V_S \left( \frac{R_3}{R_5} + \frac{1}{2} \right) \] (5)

Similarly,

\[ e_2 = V_{CM} + V_S \left( \frac{R_4}{R_5} + \frac{1}{2} \right) \] (6)

The capacitor components of the second stage of the preamplifier determine the pass bandwidth of the amplifier and can be ignored at this point so that the output of the preamplifier is

\[ E_A = \frac{R_9}{R_7} e_2 - \frac{R_8}{R_6} e_1 \] (7)

Substituting equations (5) and (6) into equation (7) yields

\[ E_A = \frac{R_9}{R_7} (V_{CM} + V_S \left( \frac{R_4}{R_5} + \frac{1}{2} \right)) - \frac{R_8}{R_6} (V_{CM} - V_S \left( \frac{R_3}{R_5} + \frac{1}{2} \right)) \] (8)

The symmetry requirements of the circuit demand that \( R_1 = R_2, R_3 = R_4, R_6 = R_7, \) and \( R_8 = R_9. \) Therefore,

\[ E_A = \frac{R_8}{R_6} (V_{CM} + V_S \left( \frac{R_3}{R_5} + \frac{1}{2} \right)) - \frac{R_8}{R_6} (V_{CM} - V_S \left( \frac{R_3}{R_5} + \frac{1}{2} \right)) \]

\[ E_A = V_S \left( \frac{2R_8R_3}{R_6R_5} + \frac{R_8}{R_6} \right) \]

\[ E_A = \frac{2R_8R_3 + R_8R_5}{R_6R_5} V_S \] (9)
Substituting the component values indicated in Figure 3, \( R_3 = 3 \times 10^5 \) ohms, \( R_5 = 1.2 \times 10^4 \) ohms, \( R_6 = 10^4 \) ohms, and \( R_8 = 10^6 \) ohms, yields

\[
E_A = \frac{2(3 \times 10^5)(10^6) + 1.2 \times 10^4 (10^6)}{(1.2 \times 10^4)(10^4)} V_S
\]

\[
E_A = 5100 V_S
\]

Therefore, the preamplifier has an expected gain of 5100.

The bandwidth of the signal \( V_S \) which is passed by the preamplifier is determined in the second stage, as indicated previously. The low frequency cutoff, \( f_L \), is determined by capacitors \( C_6 \) and \( C_7 \) and their respective series resistors \( R_6 \) and \( R_7 \) so that

\[
f_L = \frac{1}{2\pi R_6 C_6}
\]

Substituting the actual component values of Figure 3,

\[
f_L = \frac{1}{2\pi (10^4)(100 \times 10^{-6})}
\]

\[
f_L = 0.16 \text{ Hz}
\]

The high frequency cutoff, \( f_H \), is determined by the parallel resistor and capacitor circuits in the feedback loops of the second stage, so that

\[
f_H = \frac{1}{2\pi R_8 C_8}
\]

Substituting the actual values,
\[ f_H = \frac{1}{2\pi (10^6)(0.001 \times 10^{-6})} \]
\[ f_H = 160 \text{ Hz} \quad (14) \]

2. Tests of Preamplifier

The preamplifier of Figure 3, as discussed in the previous section, was constructed and tested. The gain was experimentally determined by impressing a 12 Hz triangular signal across the inputs, and measuring the output. The 12 Hz signal was used because of the similarity between the triangular signal and the QRS component of the ECG signal. The input and output of the preamplifier were monitored with a dual trace oscilloscope, and a photograph of the actual oscilloscope traces is presented in Figure 4. The amplitude of the input signal was set at 0.0045 volts peak to peak, and the triangular output signal was measured at 22.5 volts peak to peak. Therefore, the gain of the preamplifier found by measurement was

\[ G = \frac{V_{\text{OUT}}}{V_{\text{IN}}} = \frac{22.5}{0.0045} \]
\[ G = 5000 \quad (15) \]

This measured gain differed only slightly from the theoretical gain calculated at 5100.

The common mode rejection of the preamplifier was determined by placing a common 50 Hz, 100 mv peak to peak sine wave signal from each input to ground, and measuring
Figure 4. Preamplifier Output with a 12 Hz 0.0045 Volt P-P Triangular Input

Figure 5. Preamplifier Output with Actual ECG Input
the output. The 50 Hz signal was used to allow distinction of the input signal from possible 60 Hz noise. The output was measured and found to be 4 millivolts peak to peak. The common mode rejection ratio (CMRR) is defined as the ratio of the true common mode input signal to the differential input that would produce the same output signal. Therefore, with the gain of 5000.

\[ CMRR = \frac{V_{IN}}{V_{OUT}} \frac{V_{OUT}}{5000} \]  \hspace{1cm} (17)

\[ CMRR = 100 \times \frac{4}{5000} \]

\[ CMRR = 125,000 : 1 \]  \hspace{1cm} (18)

The common mode rejection (CMR) in decibels is

\[ CMR = 20 \log_{10} CMRR \]  \hspace{1cm} (19)

\[ CMR = 20 \log_{10} 125,000 \]

\[ CMR = 102 \text{ db} \]  \hspace{1cm} (20)

The CMR was better than the required value of 100 db and could probably be improved further by closely matching the components of the preamplifier.

Since the gain and common mode rejection of the preamplifier met or exceeded the previously stated requirements, an actual bioelectrical signal, obtained as described in Section II, was used as an input to the preamplifier. The resulting output ECG signal was entirely satisfactory, and
a photograph of an oscilloscope trace of the actual output is presented in Figure 5.

C. Sixty Hz Rejection Filter

A filter to reduce any 60 Hz noise present at the output of the preamplifier is desirable because of the sensitivity of the subsequent stage. The outputs obtained from the preamplifier described in Section IV-B indicate that the filter would be optional for this particular preamplifier. However, since a different preamplifier or different electrodes and lead configurations which would produce a significant noise level might be used, the filter is included as a part of the basic interbeat timer.

1. Description of the Filter

A narrow band rejection filter of the type reported by Jerome Lyman in "400 Ideas for Design, Volume 2" (1971), is presented in the circuit diagram of Figure 6.\(^{(22)}\) The circuit is an active filter requiring two operational amplifiers and a twin-T circuit. According to Lyman's report, the twin-T has a transfer function of

\[
H_T(s) = \frac{s^2 + \omega_0^2}{s^2 + B_w s + \omega_0^2} \quad (21)
\]

where \(\omega_0\) is the null frequency in radians and \(B_w\) is the bandwidth of the twin-T filter's rejection. This passive
FOR 60 Hz REJECTION

\[ C_A = 0.1 \mu \text{fd} \]
\[ R_A = 26.7 \, \text{kΩ} \]

Figure 6. 60 Hz Rejection Filter
circuit is not useful because the bandwidth is four times the center frequency. The transfer function of the entire active filter is

\[ H_p(s) = \frac{K}{K + 1} \frac{s^2 + \omega_0^2}{s^2 + \frac{B_\text{w}s}{K + 1} + \omega_0^2} \]  \hspace{1cm} (22)

where \( K \) is the gain of amplifier \( A_5 \) determined as

\[ K = \frac{R_{15}}{R_{14}} \]  \hspace{1cm} (23)

The frequency response of the filter to a sinusoidal input is expressed by

\[ H_p(j\omega) = \frac{K}{K + 1} \frac{-\omega^2 + \omega_0^2}{-\omega^2 + \frac{j\omega B_\text{w}}{K + 1} + \omega_0^2} \]  \hspace{1cm} (24)

The effect of the active filter is to reduce the bandwidth of the twin-T by a factor of \((K + 1)\). The bandwidth of the filter, \( B_\text{wF} \), therefore becomes

\[ B_\text{wF} = \frac{B_\text{w}}{K + 1} \]  \hspace{1cm} (25)

The overall gain of the filter circuit, \( G_\text{F} \), outside of the narrow rejection band is

\[ G_\text{F} = \frac{K}{K + 1} \]  \hspace{1cm} (26)
For the filter required in the interbeat timer system the null frequency, \( f_0 \), is 60 Hz. Therefore,

\[
\omega_0 = 2\pi f_0 = 2\pi(60)
\]

\[
\omega_0 = 120\pi \text{ radians} \quad (27)
\]

For the twin-T filter, the null frequency is determined as (21)

\[
\omega_0 = \frac{1}{R_A C_A} \quad (28)
\]

\( R_A \) and \( C_A \) can be chosen from standard components to produce the required \( \omega_0 \). These were chosen as \( R_A = 26.7 \text{k ohms} \) and \( C_A = 0.1 \text{ microfarads} \). The gain of the second stage amplifier can be calculated from equation (23)

\[
K = \frac{R_{15}}{R_{14}} = \frac{10^4}{10^2}
\]

\[
K = 100 \quad (29)
\]

The gain of the filter from equation (26) is

\[
G_F = \frac{K}{K + 1} = \frac{100}{100 + 1}
\]

\[
G_F = 0.99 \quad (30)
\]

Since the bandwidth of the twin-T is four times the null
frequency, the bandwidth of the filter can be found using equation (25)

\[ B_{WF} = \frac{B_w}{K + 1} = \frac{4f_0}{K + 1} = \frac{4}{101} \]

\[ B_{WF} = 2.4 \text{ Hz} \] (31)

2. Testing of the Filter

The filter shown in Figure 6 was constructed using two Fairchild 741C operational amplifiers as A4 and A5. The rejection of a 60 Hz signal was measured by placing a 4 volt peak to peak 60 Hz sinusoidal signal across the filter input and measuring the filter output. A photograph of an actual oscillograph trace showing the input and output of the filter during this measurement is presented in Figure 7. The attenuation at 60 Hz can be calculated in decibels as

\[ \text{Attenuation} = 20 \log_{10} \frac{V_{IN}}{V_{OUT}} = 20 \log_{10} 20 \]

\[ \text{Attenuation} = 26 \text{ db} \] (32)

The twin-T filter was constructed of standard one percent resistors and ten percent capacitors without attempting to match components. The attenuation could undoubtably be improved by matching the components.

The bandwidth of the filter was measured by using a constant amplitude, variable frequency oscillator to present a signal to the filter input. The filter input and output
Figure 7. Input and Output of 60 Hz Rejection Filter with 4 Volt P-P 60 Hz Input

Figure 8. Input and Output of 60 Hz Rejection Filter with Actual Amplified ECG Input
were monitored simultaneously while ranging the input frequency from 0 to 100 Hz. The three decibels-down points were noted, and the rejection bandwidth was approximately 2.5 Hz centered about 60 Hz.

The filter was checked using an actual bioelectrical signal from the preamplifier as the input to the filter. A photograph of an oscilloscope trace showing the input and output of the filter is presented in Figure 8. The ECG output of the filter is satisfactory, indicating that rejection of the 60 Hz component does not adversely affect the ECG signal.

D. Beat Detector

In order to measure the interbeat times, it is necessary to obtain a reliable indication once each cardiac cycle which can be used for timing. The QRS component of the ECG is generally the most prominent part of the signal and cardiotachometers use the amplitude of this component to trigger their circuits. However, other parts of the ECG sometimes trigger the circuit, and sometimes a small-amplitude QRS signal might not trigger the circuit. Studying the ECG signal reveals the fact that the slopes of the normal QRS component are unique in the signal and either the rise or fall of this component could provide a reliable trigger. Differentiation of the ECG signal would produce a signal which would have amplitude corresponding to the slopes of
the ECG. If the rise of the QRS component is used as the trigger, the differentiated ECG signal can be used as an input to a comparator which would yield an output only for the peak corresponding to the rise of the QRS. The output of the beat detector must be a five volt signal to meet the requirements of the circuits which follow.

1. Beat Detector Circuit

The diagram of the beat detector circuit which contains the required differentiator and comparator is presented in Figure 9. The operational amplifier \( A_6 \) and associated circuitry perform the differentiation on the amplified ECG signal input \( E_F \), and operational amplifier \( A_7 \) acts as a comparator, triggering only once each cardiac cycle with the transistor circuit setting the voltage level of the output.

a. Differentiator

The true operational amplifier differentiator would have the input applied through a capacitor component to the inverting input and a feedback resistor between this input and the output of the amplifier. The output of this differentiator from operational amplifier theory is

\[
E_{OUT} = -R_F C_{IN} \frac{dE_{IN}}{dt}
\]  

(33)

where \( R_F \) is the feedback resistor and \( C_{IN} \) is the input capacitor. This differentiator circuit is not practical because of its extreme sensitivity to high frequency noise.
Figure 9. Beat Detector Circuit

NOTE: A6 AND A7 ARE FAIRCHILD 741C OPERATIONAL AMPLIFIERS
However, the differentiator circuit of Figure 9 has resistor $R_{16}$ added in series with the input capacitor. This limits the gain of the circuit at frequencies above the transition frequency. The transition frequency is defined as

$$f_t = \frac{1}{2\pi R_{16} C_{16}}$$  \hspace{1cm} (34)$$

For the circuit of Figure 9, $R_{16}$ and $C_{16}$ were chosen to produce a transition frequency of about 20 Hz.

$$f_t = \frac{1}{2\pi (5 \times 10^3) (1.5 \times 10^{-6})}$$

$$f_t = 20 \text{ Hz}$$ \hspace{1cm} (35)

Since the effect of $R_{16}$ in the low frequency range is negligible, the output of the differentiator, from equation (33) is

$$E_{\text{OUT}} = -R_{16} C_{16} \frac{dE_{\text{IN}}}{dt}$$ \hspace{1cm} (36)

Using an ECG signal as $E_{\text{IN}}$, the output should have a negative amplitude pulse corresponding to the slope of the rise of the QRS complex. Based on a rise time of 20 milliseconds and a normal amplitude from 0.5 to 2.6 millivolts multiplied by a gain of 5000, the range of instantaneous values, in volts per second, of this slope would be

$$\frac{0.0005 \times 5000}{0.02} \leq \frac{\Delta E_{\text{IN}}}{\Delta t} \leq \frac{0.0026 \times 5000}{0.02}$$
$125 \text{ V/SEC} \leq \frac{\Delta E_{IN}}{\Delta t} \leq 650 \text{ V/SEC}$ (37)

The instantaneous value of the output of the differentiator during the rise of the QRS complex is then determined by the variable feedback resistor $R_{17}$ using equation (36). $R_{17}$ can be adjusted so that the output is driven to its -15 volt limit during the rise of the QRS, yielding a very distinctive output pulse.

The differentiator was tested using a 12 Hz, 10 volt peak to peak triangular signal as an input. A photograph of an oscilloscope trace showing the input and output of the differentiator is presented in Figure 10. The differentiator was then tested with an actual amplified ECG signal as the input. A photograph of an oscilloscope trace showing the input and output of the differentiator using the ECG input is presented in Figure 11. During the testing, a value of about 30 kilohms for $R_{17}$ was found to provide a reliable and useful pulse.

b. Comparator

The output of the differentiator is used as the input to the comparator stage. The open circuit gain of the comparator amplifier, $A_7$, is high, so that a slightly negative signal at the inverting input results in a +15 volt output, and a slightly positive signal at the inverting input results in a -15 volt output. As in Figure 9, the polarity of the signal at the inverting input is
Figure 10. Input and Output of Differentiator Stage with 10 Volt P-P 12 Hz Triangular Input

Figure 11. Input and Output of Differentiator Stage with Actual ECG Input
determined by the differentiator output signal and the voltage divider circuit of $R_{19}$ and $R_{20}$. The output of the comparator amplifier is -15 volts unless the differentiator output is more negative than -10 volts, in which case the comparator output is +15 volts. A photograph showing the input and output of the comparator amplifier, using a test signal as an input, is presented in Figure 12. A photograph of an oscilloscope trace showing the output of the comparator resulting from an actual amplified ECG signal applied to the input of the differentiator stage is present in Figure 13. This photograph shows distinctly the single +15 volt output signal during the rise of the QRS complex.

c. Detector Output Voltage Circuit

The output of the comparator amplifier is then applied through a voltage divider to the base of transistor $T_1$. This transistor acts as a switch, conducting only when the output of the amplifier is high. The output of the beat detector is taken at the emitter of $T_1$. This output can be adjusted with $R_{26}$ so that when $T_1$ is conducting, the output of the detector is +5 volts, and when $T_1$ is off, the output is zero. A photograph showing an actual ECG input to the detector and the detector output pulse is presented in Figure 14.

2. Testing of the Beat Detector

The beat detector circuit was tested for several hours
Figure 12. Input and Output of Comparator Stage with Detector Input of 10 Volt P-P Triangle

Figure 13. Output of Comparator with Actual Amplified ECG Input to the Detector
Figure 14. Detector Output with Actual ECG Input

Figure 15. Detector Output with Input of Actual ECG from an Active Subject
using an actual ECG input and continued to provide a reliable single output pulse once each cardiac cycle. During this testing, the worst case ECG signal conditions of an active human subject performing many types of activity were presented to the beat detector without affecting its performance. A photograph of an oscilloscope trace showing the ECG of the human subject running in place and the corresponding beat detector output is presented in Figure 15.

E. Timer Control Circuit

A control circuit is needed which can accept the output of the beat detector as its input and provide proper control for the clock pulse input to the counter, for the memory circuit, for the counter reset, and for the print command to a digital printer. The circuit of Figure 16 provides this control. The components of this circuit are Texas Instruments TTL integrated circuits. The logic level for this TTL logic is

LOGICAL 0 ≤ 0.8 VOLTS
LOGICAL 1 ≥ 2.4 VOLTS

Three SN74121 monostable multivibrators with external timing components are used. The multivibrators are triggered by the Schmitt-trigger C input going from logical 0 to logical 1 with either or both of the A and B inputs at logical 0, or by either or both of the negative-edge-trig-
SN74121 MONOSTABLE MULTIVIBRATOR Q1

SN74121 MONOSTABLE MULTIVIBRATOR Q2

SN74121 MONOSTABLE MULTIVIBRATOR Q3

Figure 16. Timer Control Circuit
gered inputs A and B going to logical 0 with input C at logical 1. The width of the output pulse is determined by the external timing components by

\[ t = 0.693 R_T C_T \] (38)

The output Q and its inverse \( \bar{Q} \) are available from the multivibrator. (23)

In the circuit of Figure 16, the beat detector output \( E_D \) is applied to the Schmitt-trigger input \( C_1 \) of monostable multivibrator \( Q_1 \) so that each time \( E_D \) goes to logical 1, indicating the end of a cardiac cycle, the multivibrator is triggered. The width of the output pulse \( Q_1 \) is determined from equation (38) to be

\[ t_1 = 0.693 \frac{R_{T1A} (R_{T1B} + R_{T1C})}{R_{T1A} + R_{T1B} + R_{T1C}} C_{T1} \] (39)

Selecting components as labeled in Figure 16, the variable resistor \( R_{T1C} \) can be used to adjust the width of \( t_1 \) to 0.90 milliseconds. The output \( Q_1 \) and the clock signal \( C_P \) can be used as inputs to a two input, positive NOR gate. The output of the NOR gate, \( Y_1 \), can then be used directly as the count input to the counter, disabling the input during pulse \( Q_1 \).

\[ Y_1 = \overline{Q_1} + C_P \]

\[ Y_1 = \overline{Q_1} \cdot C_P \]

Since the counter must have time to complete the count
process after the input is disabled, a delay is needed between the beginning of \( Q_1 \) and the control signal which holds the count in memory. A delay of one microsecond would provide sufficient time for the counter to complete its count and the memory circuits to follow the counter outputs. This delay can be accomplished by using \( Q_1 \) as the input to the negative-edge-triggered inputs \( A_2 \) and \( B_2 \) of monostable multivibrator \( Q_2 \). This multivibrator is then triggered when \( Q_1 \) goes from logical 1 to logical 0, which occurs with each input pulse \( E_D \). The width of the output pulse \( Q_2 \) is

\[
t_2 = 0.693 R_{T2} C_{T2} \tag{41}
\]

For the components of Figure 16, \( t_2 \) has a width of about one microsecond. Then the output \( Q_2 \) is applied to the negative-edge-triggered inputs of monostable multivibrator \( Q_3 \). Therefore, \( Q_3 \) will be triggered one microsecond after \( Q_1 \) is triggered. The width of the output pulse \( Q_3 \) is

\[
t_3 = 0.693 R_{T3} C_{T3} \tag{42}
\]

For the components shown in Figure 16, \( t_3 \) has a width of about ten milliseconds. Then the output \( \overline{Q}_3 \) can be used directly as control \( Y_3 \) to hold the count in memory for ten milliseconds.

\[
Y_3 = \overline{Q}_3 \tag{43}
\]

The counter can then be reset and the printer commanded to
print the output by a control signal $Y_2$ obtained by using $\overline{Q}_3$ and $\overline{Q}_1$ as inputs to a two input positive NOR gate.

$$Y_2 = \overline{Q}_1 + \overline{Q}_3$$

$$Y_2 = Q_1 \cdot Q_3$$

The control circuit input, monostable multivibrator outputs, and control circuit outputs are summarized in Figure 17.

F. CLOCK PULSE GENERATOR

The accuracy of a timing system based on the concepts of the block diagram of Figure 2 depends upon the accuracy of the pulse repetition rate of the clock pulse train which is the input to the counter. A highly accurate pulse or square wave generator, available commercially, can provide the required signal, but a generator constructed specifically for the timer system is also possible. A crystal oscillator could be used to provide a dependable repetition rate for the clock pulses or an astable multivibrator could be used. The output of the clock generator must have an amplitude of 5 volts and a pulse width of at least 50 nanoseconds.

The circuit diagram of an inexpensive clock pulse generator is presented in Figure 18. This circuit uses two Texas Instruments integrated circuit monostable multivibrators, each having external timing components. The multivibrators are connected so that one multivibrator triggers
Figure 17. Input and Outputs of Control Circuit
Figure 18. 10k Hz Clock Pulse Generator
the other. The circuit is described in "Bulletin CA-128 - A Texas Instruments Application Report". (24) The width of the clock pulse is determined by the timing circuit of Q4 as

\[ t_4 = 0.693 R_{T4} C_{T4} \]  \hspace{1cm} (45)

and the time between pulses is determined by the timing circuit of Q5 as

\[ t_5 = 0.693 \frac{R_{T5A} (R_{T5B} + R_{T5C})}{R_{T5A} + R_{T5B} + R_{T5C}} C_{T5} \]  \hspace{1cm} (46)

With the components chosen in Figure 18, \( t_4 \) is about 0.05 milliseconds, and \( t_5 \) can be set using the variable resistor \( R_{T5C} \) so that

\[ t_4 + t_5 = 0.10 \text{ milliseconds} \]

This yields a clock pulse train with a repetition rate of \( 10^4 \) pulses per second. The circuit was tested and calibrated with a Hewlett Packard 2401C integrating digital voltmeter and counter. The repetition rate was found to be stable. However, tests were not conducted to determine the effects of temperature changes or prolonged usage on the generator output.

G. COUNTER

The circuit diagram of a five decade counter consisting of five Texas Instruments SN7490 integrated circuit decade
counters is presented in Figure 19. The counter inputs are $Y_1$, the controlled clock pulse train, and $Y_2$, the counter reset control. The outputs are five decades of 8421 BCD information. A contact identification diagram for the SN7490 and the truth tables of the BCD count sequence and the reset inputs are presented in Figure 20. According to the reset truth table of Figure 20 and the circuit diagram of Figure 19, the first decade of the counter will reset to 9, 1001 in the 8421 BCD, and each of the remaining four decades will reset to 0, 0000 in the 8421 BCD, when the $Y_2$ reset input is at logical 1. This resets the decimal output of the counter to 00009, which corresponds to the 0.9 milliseconds that the counter is disabled during the memory-hold and counter-reset sequences. When $Y_2$ returns to logical 0, the counter begins counting the $Y_1$ input signal. The range of the counter is therefore from 00009 to 99999, corresponding with times from 0.9 milliseconds to 9.9999 seconds.

H. Memory

The memory circuit of Figure 21 consists of three Texas Instruments SN74100 integrated circuit eight bit bistable latches. Each of the 20 BCD outputs of the counter was applied to an input of a latch. The inputs of the latches are directly transferred to the outputs when the control input $Y_3$ is at logical 1. When $Y_3$ is at logical 0, the outputs are latched and held until $Y_3$ returns to a logical 1.
Figure 19. Five Decade Counter Circuit with BCD Output
NOTE: External connection of BD to A is required for BCD decade count output.

<table>
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<th>D</th>
<th>C</th>
<th>B</th>
<th>A</th>
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<td>0</td>
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</tr>
</tbody>
</table>

Figure 20. Inputs and Outputs for SN7490 Dual-In-Line Package Decade Counters
Figure 21. Five Decade Memory Latch
Since \( Y_3 \) is at logical 0 for ten milliseconds at the end of each cardiac cycle, the counter output at the end of the cardiac cycle is held for ten milliseconds, long enough to allow a printer to record the counts.

I. Printer

The outputs of the memory are applied to the data inputs of a digital printer which can accept at least five decades of BCD inputs from five volt TTL circuits and print this information in the decimal system. A number of digital printers are available in the commercial market which meet these requirements. The Hewlett Packard Models 5050B and 5055A, the Franklin Electronics Series 1200 and 1600, and the Addmaster Model 55 are examples of acceptable printers. The signal \( Y_2 \), which resets the counter, is also used as the print command to the printer. This signal is slightly less than 0.9 milliseconds in duration and has an amplitude of about 4 volts. Some printers require a higher voltage print command signal and this can be provided, if necessary, by using \( Y_2 \) to control a simple transistor switch which can provide the necessary amplitude. The inputs are held by the memory for about 10 milliseconds after the beginning of the print command signal, allowing sufficient time for the printer to complete its required transfer functions.
The interbeat timer system described in Section IV was fabricated and tested. A photograph of the breadboard unit is presented in Figure 22. A connector for the bioelectrical input signal, a connector for the circuit power input requirements, and an output connector which mates to the digital printer are shown in Figure 22. The bioelectrical signal inputs are fused with two milliamp fuses as a safety precaution in addition to the normal procedure of isolating the circuit ground from the AC power line ground. \(^{(7),(25)}\) The breadboard unit was tested as completely as possible without using an output printer. All of the output points from each of the circuits which compose the system, including all of the counter and all of the memory outputs, were monitored with an oscilloscope using an actual ECG signal as the input to the system. The system required +15 volts, -15 volts and +5 volts power inputs, which were obtained for testing purposes from Power Design Corporation 0 to 50 volt, 1.5 ampere power supplies. These power supplies can readily be constructed and included as part of the timer system, so that only 115 volt AC line power would be required to operate the system. During the testing, it was found necessary to isolate the +5 volt power for the monostable multivibrators from the +5 volt power for the counters to prevent a false count. After these separate 5 volt power supplies were provided, the system functioned
Figure 22. Photograph of Interbeat Timer System Fabricated for Testing
properly, and provided the required interbeat interval data in a continual beat to beat manner with an accuracy much better than the required ± 1 millisecond per interval.

A. System Accuracy

The accuracy of the timer system depends upon the accuracy of the clock pulse train. The repetition rate of the pulses was set to 10^4 pulses per second with an accuracy of ± 1 pulse per second, corresponding to ± 1 count per second at the timer output. The rest of the system can have a maximum error of ± 1 count per interval due to the uncertainty of the voltage level of the clock pulse train when the counter input is enabled and disabled. Therefore, the accuracy of the timer is within ± 1 count per second plus ± 1 count per interval. For a heart rate of 60 beats per minute, the accuracy would be ± 0.2 milliseconds per interval measured. This accuracy exceeds the required accuracy of ± 1 millisecond per interval. The accuracy could be improved further by increasing the repetition rate of the clock pulses and increasing the number of decades in the counter, memory, and printer components.

B. System Timing Range

The range of the timer system is from 1.0 milliseconds to 9.9999 seconds. This adequately covers the normal range
of interbeat intervals.

C. System Cost

The cost of fabricating the interbeat timer system is quite reasonable, which is advantageous to the small research laboratories needing this type of equipment. The cost of the circuit components, less the digital printer, at single-unit retail prices is slightly less than two hundred dollars. The cost of the digital printer can vary from about four hundred dollars to several thousand dollars depending on the sophistication and the number of print columns of the printer.

D. System Upkeep

A distinct advantage of this interbeat timer system is the use of integrated circuit components. This allows for convenient plug-in replacement of components should repair of the unit become necessary. The number of different types of integrated circuits was kept at a minimum so that repair supplies could be maintained simply and inexpensively.

E. System Outputs

The timer system as described in this report will provide output data on the cardiac interbeat intervals in a
readily usable manner, printed directly in time units on the paper tape output of the digital printer. Additional channels on the printer can be used to record real time data or to record such things as conditions and stimuli to which a human subject might be exposed during testing. The system can also be readily expanded to provide the output data on punched paper tape or on magnetic tape for direct computer reduction.
BIBLIOGRAPHY


Bibliography (continued)

Richard William Mooney was born on December 24, 1943, in Evansville, Indiana. He received his primary education and part of his secondary education in Evansville, Indiana, and completed his secondary education in Kirkwood, Missouri. He studied for four years at Washington University, St. Louis, Missouri, and received a Bachelor of Science degree in Electrical Engineering in June 1965.

He has been enrolled in the Graduate School of the University of Missouri-Rolla since September 1965, except for a period from January 1969 to September 1971, during which he served in the U. S. Army as a Laboratory Engineer for the U. S. Army Behavior and Systems Research Laboratories in Arlington, Virginia. During the periods of his graduate study, he has been employed as a Laboratory Engineer for the Instrumentation and Standards Laboratories of McDonnell Douglas Corporation in St. Louis, Missouri.