

- [54] **CARDIAC ARRHYTHMIA DETECTOR**
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 128/2.06 A, 2.06 E, 2.06 F, 2.06 R, 2.1 A,
 2.1 R

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[57] **ABSTRACT**

A selective circuit is provided for passing certain parts of, and information contained in, a complex input waveform containing both desired signals and undesired noise components. The circuit is particularly suited for use in a heart arrhythmia detector. The circuit operates on the first time derivative of the complex input waveform. The circuit provides an averaging means wherein the peak amplitude as well as the duration of the peak amplitude of a number of beats is averaged for the particular individual to establish a floating standard and the alarm is sounded when a signal is received which deviates by a predetermined amount from the thus established standard. The circuit is essentially self-adjusting to the norm for a particular individual both as to peak and duration rather than comparing an individual heartbeat with an arbitrary standard.

8 Claims, 11 Drawing Figures

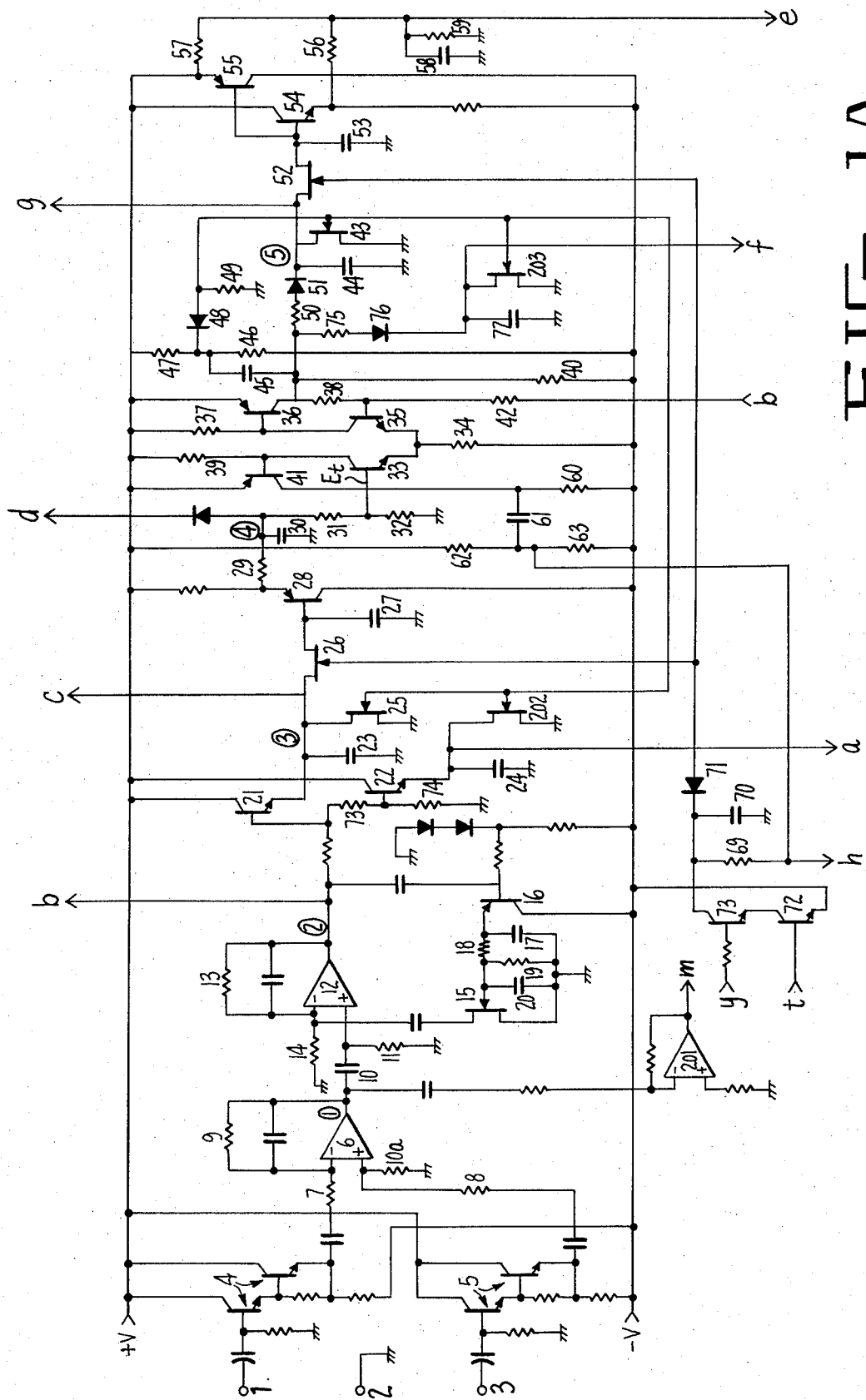


FIG. 1A.

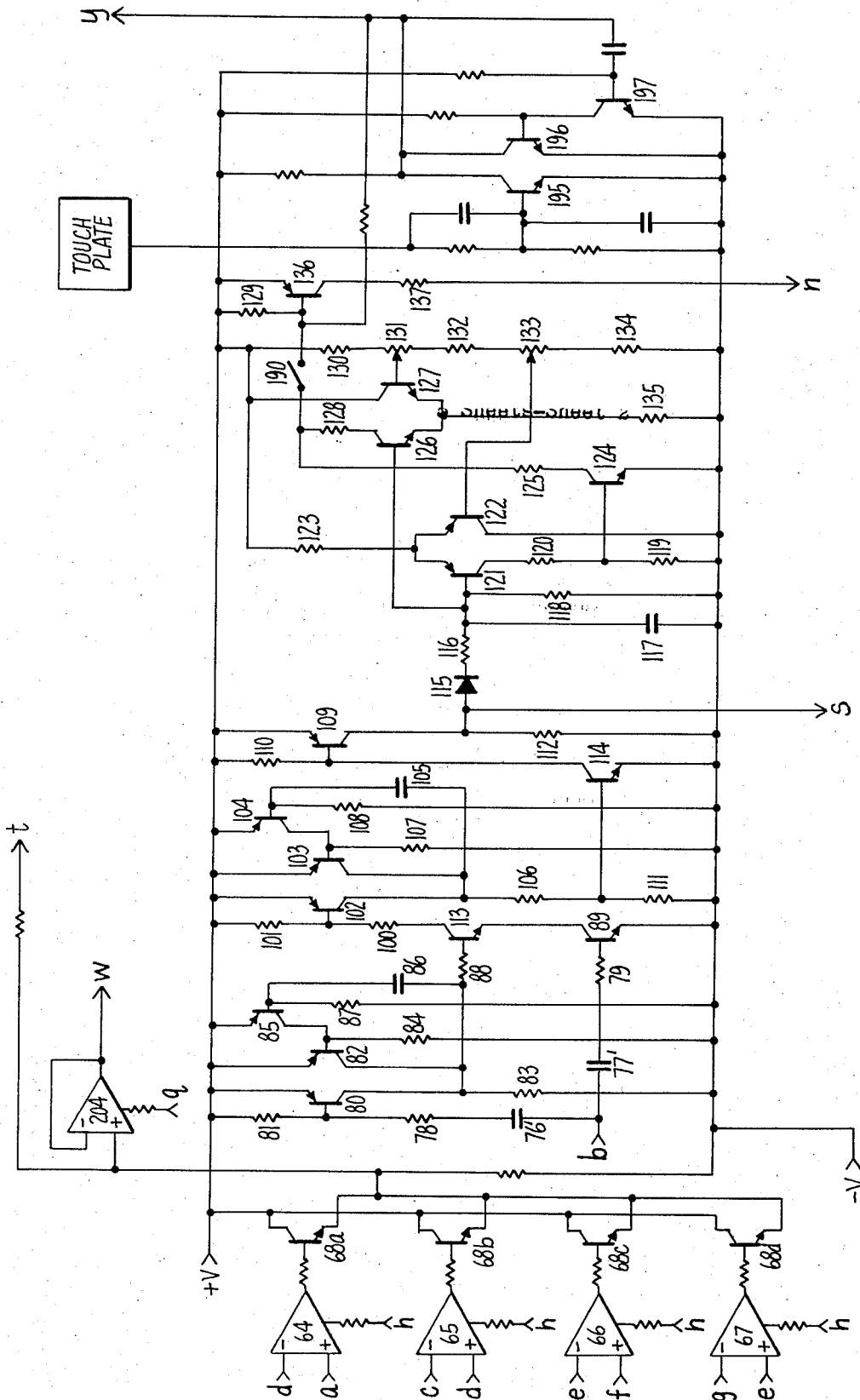


FIG. 1B.

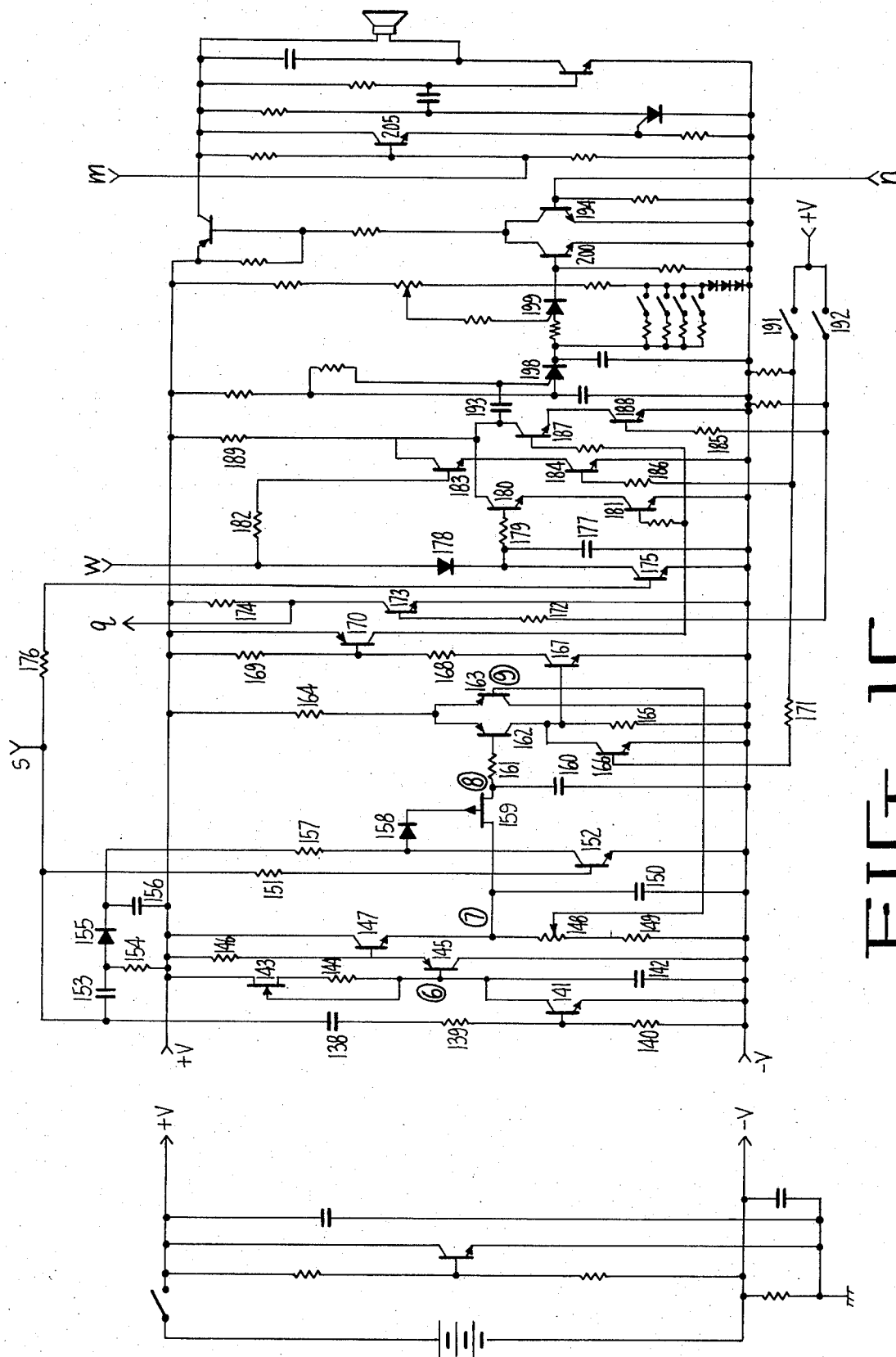
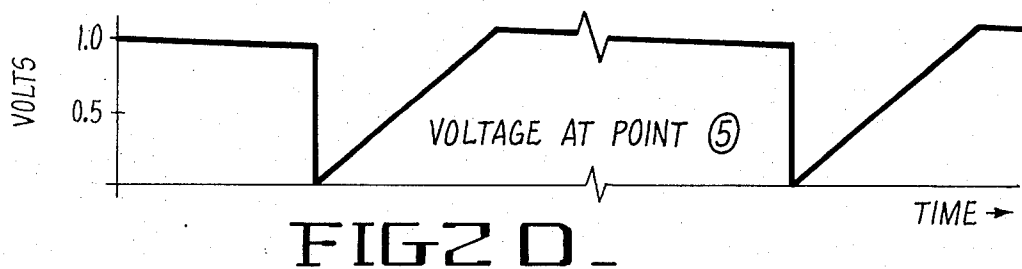
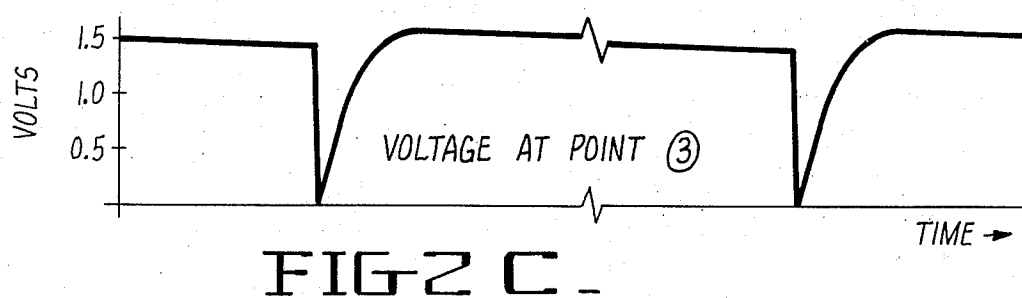
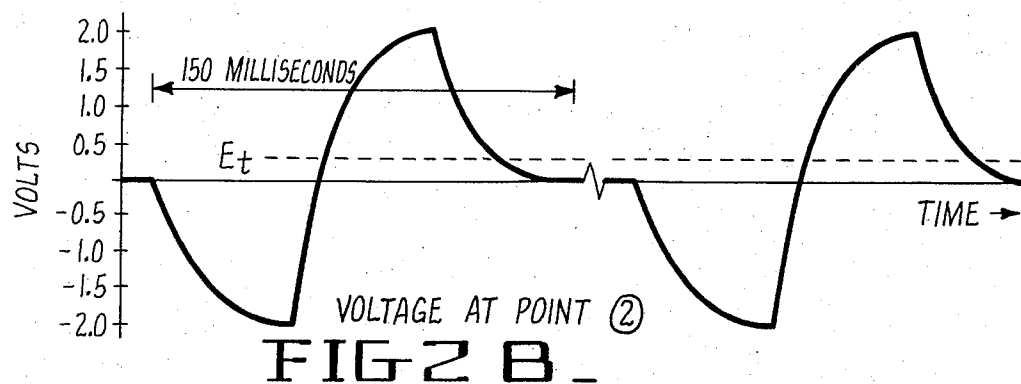
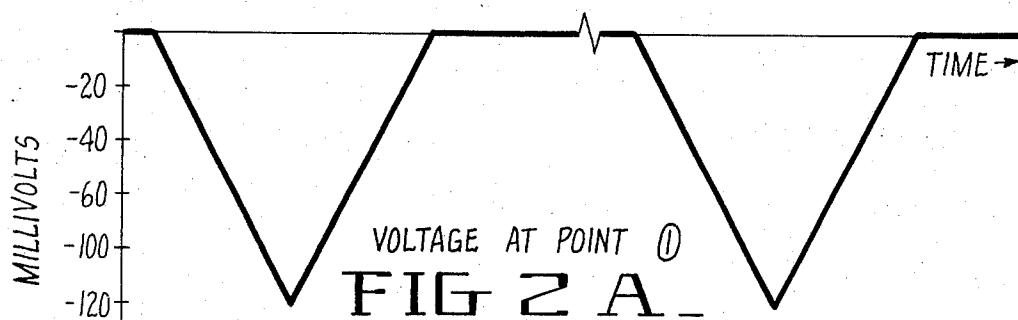
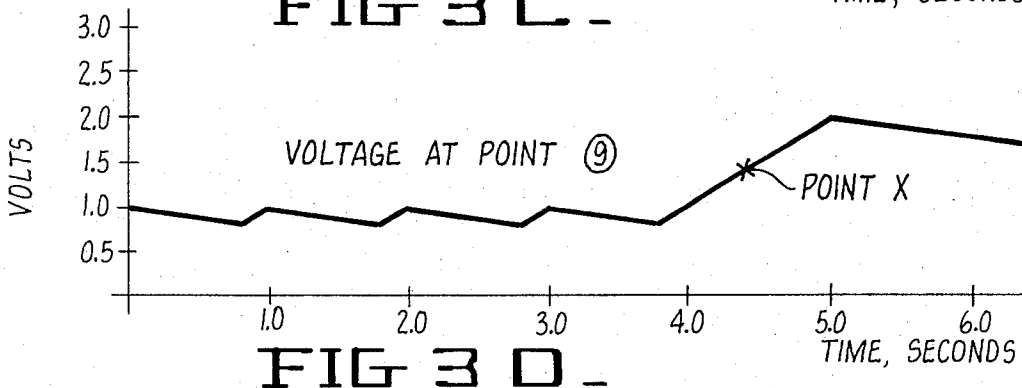
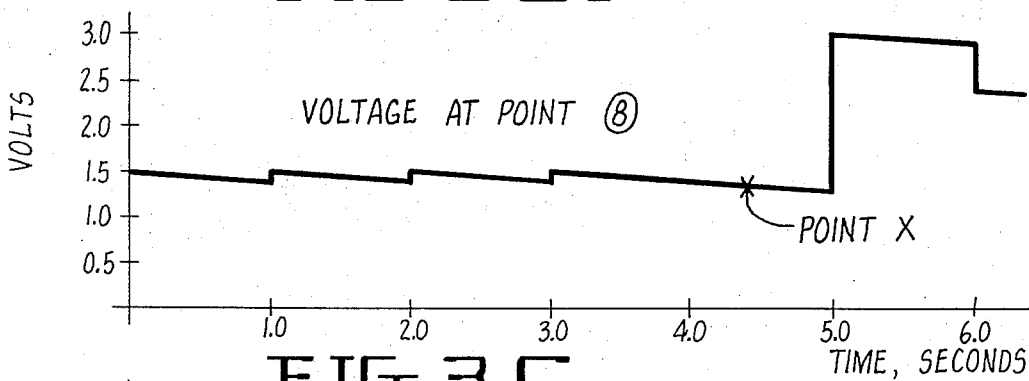
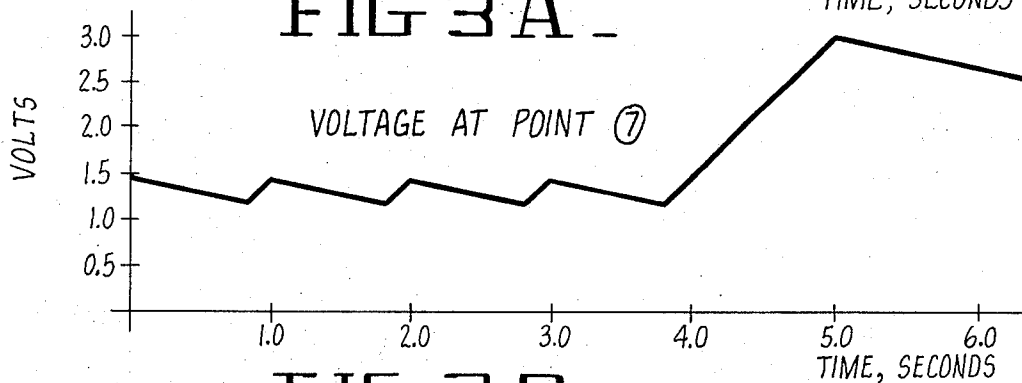
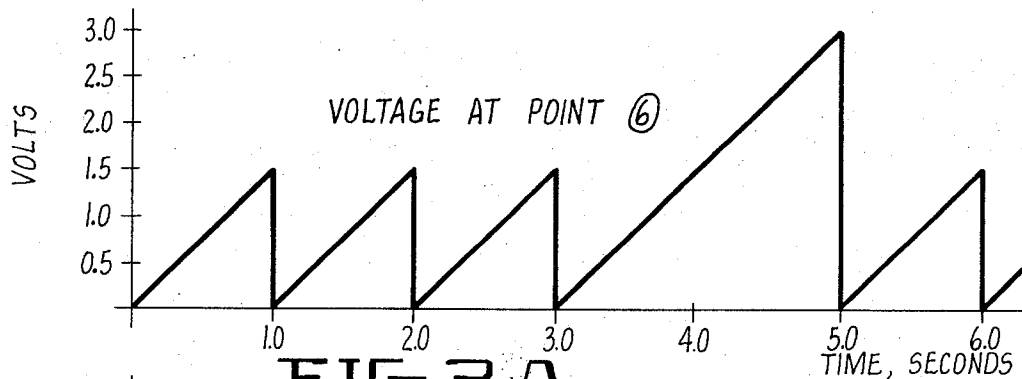


FIG. 1C





CARDIAC ARRHYTHMIA DETECTOR

SUMMARY OF THE INVENTION

In many applications of electronics it is desired that only certain shapes of electrical waveforms be allowed to pass onto subsequent circuitry or that waveforms of certain characteristics be prevented from passing. When dealing with relatively simple waves, active or passive filters will selectively remove or pass selected frequency components. However, in the case of a complex waveform which has characteristics of time, amplitude and frequency which simultaneously occur with variable relationships, a much more difficult problem is presented. In the past, waveform pattern recognition has required a large amount of expensive equipment and resort is frequently had to digital computer systems for such waveform recognition.

In many applications such as in a heart arrhythmia detector which may be mounted on the body of an individual, it is desired to provide extremely compact equipment which has a low power drain. The present invention provides a compact circuit of low power drain of the analog type which can be adjusted to selectively pass or reject selected types of complex electrical waveforms. The circuitry of the present invention can be incorporated in a small self-contained unit which can be worn on the person of the user such as that shown in U.S. Pat. Nos. 3,138,151 and 3,547,107.

The present invention was specifically developed for the use in such a self-contained heart arrhythmia detector which can be mounted on the chest, carried in a connecting or placed on a table.

One problem in such devices is that electrical noise signals may be generated in various manners, such as by imperfect contact between the input electrodes and the skin of a wearer as well as in other manners, such as muscle tremor and, since the desired signal is extremely minute, these extraneous signals can mask the desired signal, either preventing an alarm when it should be given, or triggering a false alarm. This is undesirable, not only as a nuisance, but also because it causes misdirection of efforts in emergency situations and leads to erroneous diagnosis which can be serious to the extent of being fatal. Such electrical noise can be generated in various fashions, both within and without the body of the individual being monitored. The circuitry of the present invention makes it unlikely for the alarm to be triggered by electrical noise.

The present invention discloses circuitry which analyzes the heart waveform by processing the first time derivative of the waveform, rather than the waveform itself. This unique treatment results in greatly improved rejection of noise-type waveforms such as may stem from electrode movement or muscle actions of the patient whose heart is being monitored. The basic circuitry detects abnormal ventricular activity of the heart. Also disclosed is circuitry for a time-derivative-actuated R-wave detector and a missing-R-wave detector. These circuits may be employed either individually or in coordination to provide the optimum processor for a given patient's heart. A further improvement allows for addition of a heart bradycardia or tachycardia detector and means for transmission of the patient ECG over the telephone.

BRIEF DESCRIPTION OF THE DRAWINGS

In the drawings forming a part of this invention:

FIGS. 1A, 1B and 1C are a schematic diagram of an electrical circuit embodying the present invention.

FIGS. 2A, B, C and D and 3A, B, C and D are diagrams of approximate waveforms in various parts of the circuit.

DESCRIPTION OF THE PREFERRED EMBODIMENT

The input electrodes are connected to the circuit at the terminals labeled 1, 2 and 3. The differential signal of interest is sensed by the electrodes connected to terminals 1 and 3. The electrode connected to terminal 2 provides a ground return to the signal source to provide a common-mode signal return. The electrodes may be of any conventional design, such as those shown in U.S. Pat. Nos. 3,138,151 and 3,547,107. The signals from the terminals 1 and 3 are applied to compound emitter followers 4 and 5 where the signal impedance level is reduced for delivery to the differential amplifier 6. The latter amplifier in conjunction with components 7, 8 and 9 provides a voltage gain of about 120, providing a typical output signal of about 120 millivolts, peak. Components 10 and 11 form a high-pass filter which differentiates the signal from amplifier 6. The 3-dB cutoff frequency of this filter is about 35 Hz. The differentiated signal is applied to the automatic-gain-controlled amplifier 12 for further increase in signal voltage level. Components 13 and 14 provide for a minimum voltage gain of about 20. The voltage gain is increased as necessary by the shunting effect of the control element, field-effect-transistor 15. Controlling gate voltage for the field-effect-transistor is derived from the negative peaks of the output signal from amplifier 12. These peaks cause conduction pulses through the PNP transistor 16, which produce a negative charge on capacitor 17. This charge is filtered by components 18, 19 and 20 and applied to the gate of transistor 15. Thus, as the negative peaks at the output of amplifier 12 tend to grow in magnitude, the voltage at the gate of transistor 15 becomes more negative, increasing the effective resistance of this control element, decreasing the gain of amplifier 12, and decreasing the output at point 2. The electrodes are placed on the patient's body so that the voltage waveforms at points 1 and 2 are polarized as shown in FIG. 2. Note that a positive first time derivative of the waveform at point 1 produces a positive waveform at point 2, and vice versa.

Components 21-77 form the abnormal-ventricular-activity (AV) detector. Significant waveforms in this circuitry are shown in FIG. 2. Transistors 21 and 22 and capacitors 23 and 24 form two positive peak detectors, one with an output (at point 3) equal to the full positive peak (less one base-emitter voltage drop) of the output of amplifier 12, and one with output equal to approximately six-tenths of this value (less one base-emitter voltage drop). Field-effect-transistors 25 and 26 provide means for discharging the capacitor 23, and transferring its charge to capacitor 27, respectively. Gate control signals for these transistors are developed from circuitry described below. Transistor 28 is used as an emitter follower to drive the integrating circuit formed by components 29, 30, 31 and 32. The latter network produces, at point 4, an output voltage equal to approximately three-fourths of the running mean of the positive peaks stored on capacitor 27 taken over approximately a 22 period. Resistors 31 and 32 divide the point 4 voltage by about six. This voltage E_t is ap-

plied to transistor 33, which is part of the voltage comparator circuit comprised of components 33-41. The other input to this comparator circuit comes from point 2 through current-limiting resistor 42. Thus, whenever the voltage at point 2 exceeds the voltage at the junction of resistors 31 and 32 (i.e., exceeds one-sixth of the average of the positive peaks of the voltage at point 2,) the voltage at the collector of transistor 36 is caused to approach the positive supply voltage. This action is reinforced by the positive feedback furnished by the resistor 38 between transistor 36 collector and transistor 35 base. This positive-going excursion of the transistor 36 collector voltage declares a significant ventricular action has occurred and begins a sequence of electronic events which evaluate whether or not the ventricular action should be classified as normal or abnormal.

The positive-going voltage transition at the transistor 36 collector causes a positive-going pulse to be transmitted to the gates of field-effect-transistors 25, 202, 43 and 203, in turn causing their effective resistance to be sharply reduced and thus discharging capacitors 23, 24, 44 and 77, respectively. The discharging pulse is transmitted to the gates of the field-effect-transistors through the pulse forming network of elements 45 through 49. This pulse is approximately 1 millisecond wide; the discharge of elements 23, 24, 44 and 77 is very rapid. The discharge of these capacitors prepares them to receive new information in the form of a measured amount of charge from their driving elements. At this time, the voltage at the output of amplifier 12 (point 2) has passed a threshold of approximately one-eighths (i.e., three-fourths divided by six) of its previous positive peaks and presumably is approaching another positive peak. Since capacitor 23 has been quickly discharged and released, it will follow the positive excursion at point 2 and hold the peak value attained (less a base-emitter voltage drop). Capacitor 44 will charge through resistor 50 and diode 51 so long as the collector voltage of transistor 36 is near the positive supply voltage (i.e., as long as the voltage at point 2 stays above the threshold of one-eighth of the running mean of its previous positive peak). The time constant of resistor 50 and capacitor 44 is long compared to the normal amount of time over which the latter charges. Thus the voltage on capacitor 44 is an approximately linear function of how long the collector voltage on transistor 36 stays near the positive supply voltage. In summary, then, capacitor 23 gives a voltage proportional to the peak of the positive first derivative of the significant ventricular action appearing at point 1 and capacitor 44 gives a voltage proportional to the time duration of the this positive first derivative.

Field-effect-transistor 52 acts as a switch to transfer the charge from capacitor 44 to capacitor 53 at the appropriate time. Gate-control voltage for this transistor is derived from the same source as that for transistor 26; the circuitry is described below. Transistors 54 and 55 are emitter followers which couple the voltage on capacitor 53 to the running-mean network 56-59 without significant offset voltage. The running-mean network 56-59 produces a voltage equal to approximately three-fourths of the average of the voltage on capacitor 53 taken over approximately a 20-second period.

When the voltage at point 2 passes its peak positive value and decreases below the threshold value described above, the voltage at transistor 36 collector is

caused to switch to a highly negative value, approaching the negative supply voltage. This negative voltage transition is sensed by the pulse-forming network of components 60-63, which in turn transmits a positive pulse (hereafter sometimes referred to as an interrogation pulse) approximately one millisecond long to the bias terminals h on operational amplifiers 64-67 which are configured as voltage comparators. If, during the 1 millisecond period, the voltage at the positive input terminal of any comparator is more positive than that at its negative input terminal, the output terminal of that comparator will produce a positive voltage which will turn on one of the emitter followers 68A through D and produce a positive voltage at the emitter of that transistor. Thus, the transistors 68 form an "OR" function; an emitter will go high if any of the voltage comparator 64-67 outputs produces a positive voltage (i.e., the OR function will produce a positive output if the particular significant ventricular action just completed is detected to deviate more than a prescribed amount from either the mean of its normal positive peak values, or from the mean of its normal duration).

The positive pulse formed by the network of elements 60-63 is also transmitted to the gates of transistors 26 and 52 through the delay network of resistor 69 and capacitor 70 and current-limiting diode 71, so that the charge on capacitors 23 and 44 is made to transfer to capacitors 27 and 53, respectively. However, if any of the comparator (64-67) outputs goes positive during the interrogation period, transistor 72 is made to turn on by the positive voltage appearing at the emitter of transistor 68, thus clamping out the charge-transfer pulse to the gates of elements 26 and 52 preventing the capacitors 27 and 53 from receiving voltage information from capacitors 23 and 44 respectively. This inhibiting function only occurs for ventricular actions which are classed as abnormal, so that the averaging networks 29-32 and 56-59 do not receive input information due to abnormal heart beats.

During approximately the first minute of operation it is necessary to allow all ventricular activity to load the running-average networks made up of elements 29-32 and 56-59. Thus, it is necessary to lock out the charge-transfer inhibiting capability of transistor 72. This is accomplished by the action of transistor 73. When the entire circuit is first actuated with power, the touch-plate monostable multivibrator (discussed more completely below) is caused to produce a negative voltage at the base of transistor 73, turning this transistor off and defeating the charge-transfer inhibit function of transistor 72.

The voltage on the positive input of voltage comparator 64 is equal to approximately six-tenths of the positive peak value achieved by the output of amplifier 12 during a given ventricular action less one base-emitter voltage drop. The voltage on the negative input is equal to approximately three-fourths of the running mean of the positive peaks achieved by amplifier 12 output during heart beats which are classed as normal, less one base-emitter voltage drop. Thus, the output of the comparator 64 will become positive during the interrogation pulse received at its bias terminal if the positive peak at amplifier 12 output exceeds 125 percent of the running mean of the normal positive peaks.

The voltage on the positive input of comparator 65 is equal to that on the negative input of comparator 64. The voltage on the negative input is equal to the full

positive peak achieved by the amplifier 12 output, less one base-emitter voltage drop. Thus, the output of comparator 65 will become positive during the period of the interrogation pulse received at its bias terminals if the positive peak of the amplifier 12 output is less than 75 percent of the running mean of the normal positive peaks.

The positive input of voltage comparator 66 is derived from resistance-capacitance network 75-77. This network charges the capacitor 77 during the same time period that capacitor 44 is being charged through resistor 50, except that the RC time-constant is chosen to be somewhat longer than that of elements 50 and 44, so that capacitor 77 will charge to only about six-tenths of the charge on capacitor 44. The voltage at the negative input of comparator 66 is equal to approximately three-fourths of the running means of the voltage achieved by capacitor 44 during normal heart beats. Thus, comparator 66 will produce a positive voltage at its output during the period when the interrogation pulse is applied to its bias terminal if the time period that the amplifier 12 output spends above the threshold voltage is longer than 125 percent of the running mean of the time spent above the threshold during normal heart beats.

The positive input of comparator 67 is equal to the negative input of comparator 66. The negative input is equal to the voltage on capacitor 44. Thus, the output of comparator 67 will produce a positive voltage during the period of the interrogation pulse if the period spent by the amplifier 12 output above the threshold voltage is less than 75 percent of the running mean of the time spent above the threshold during normal heart beats.

Thus, the emitter of any one of transistors 68a, 68b, 68c or 68d will produce a positive voltage if during a significant ventricular action, the peak of the positive first derivative of the voltage at point 1 deviates in magnitude by more than 25 percent from the running mean of the peaks during normal heart beats, or if the time period spent by the positive first derivative above a threshold value equal to one-eighth of the running mean of its normal peaks deviates more than 25 percent from the running mean of the same time periods for normal heart beats. Anytime any one of the emitters of transistors 68a, 68b, 68c or 68d becomes positive, an abnormal ventricular action is declared. The positive emitter voltage is transferred to the alarm criteria circuitry, described below, through operational-amplifier follower 204.

Components 76'-114 form the R-wave detector. The detector operates by determining whether or not the elapsed time between a strong negative voltage deflection and a strong positive voltage deflection at point 2 is within a certain allowable limit set by an internal monostable multivibrator. The R wave is identified by requiring a first time derivative which is positive in polarity, indicating a slope on the heart waveform of about 0.02 volts per second and having a time duration of about 50 milliseconds, followed immediately by a first time derivative which is negative in polarity, indicating a slope on the heart waveform of about -0.02 volts per second and having a time duration of about 50 milliseconds and includes means for indicating each time this criterion is met. If properly sequenced negative and positive voltage excursions do occur, the collector of transistor 109 is caused to become strongly positive for a period of time set by a second monostable

multivibrator, normally about 65 milliseconds. For voltage excursions which are not of sufficient magnitude, or for which the strong negative excursions are not followed within the allowable time limit by a sufficiently strong positive excursion, transistor 109 will not produce a strongly positive collector voltage.

Components 76' and 78 couple the signal from amplifier 12 output into the base of transistor 80. The latter element and elements 81-87 form a monostable multivibrator with a time-out period of approximately 100 milliseconds. When a strong negative voltage excursion is coupled into the base of transistor 80, the collector of this transistor becomes strongly positive, furnishing base current to transistor 113 and turning this element on for about 100 milliseconds. If a strong positive voltage excursion is coupled from point 2 through elements 77' and 79 into the base of the transistor 89 during this time period, sufficient current will be drawn through resistors 100 and 101 to fire the monostable multivibrator formed by elements 102-108. This causes the collector of transistor 102 to become strongly positive, causing sufficient current flow through resistor 111 to activate transistor 114. Current drawn by transistor 114 through resistor 110 causes transistor 109 to be turned on, thereby causing its collector voltage to become strongly positive. This condition continues for about 65 milliseconds, the time-out period of the monostable multivibrator. The output of the R-detector circuitry (i.e., the collector of transistor 109) is made available to the missed-R-detector and the abnormal-rate circuitry discussed below.

Elements 115-134 form the abnormal-heart-rate detector. Each time an R-wave is detected by the R-wave detector, a precise (both in amplitude and duration) positive pulse is applied by transistor 109 to diode 115. Current then flows through diode 115 and resistor 116 to charge capacitor 117. The time constant of resistor 116 and capacitor 117 is chosen such that the capacitor is only partially charged during an individual input pulse. However, discharge is at a relatively constant rate through resistor 118. Thus, if the heart rate increases, the voltage on capacitor 117 increases, while if the rate decreases the voltage on the capacitor decreases. The transistor pair 121-122 and associated resistors form a differential voltage comparator which detects abnormally low heart rates; if the voltage on capacitor 117 becomes less than the threshold voltage on the base of transistor 122 (set by the adjustable voltage divider formed by resistors 130-134) the transistor 121 will conduct. This will cause sufficient current flow in resistor 119 to turn on transistor 124. The resultant current drawn through resistor 129 is sufficient to cause a voltage drop across this element which will turn on transistor 136. The latter element then delivers current to the tone oscillator circuit described below. Similarly, transistors 126 and 127 form a differential voltage comparator which detects abnormally high heart rates. When the voltage on capacitor 117 increases above the level set at the base of transistor 127 by the voltage divider network of elements 130-134, transistor 126 draws sufficient current through resistor 129 to turn on transistor 136. The latter element then delivers current to the tone oscillator circuit described below.

Components 138-171 form the missed-R-wave (MR) detector. The purpose of this circuit is to detect a sudden elongation in the period between R-waves. Such

elongation often accompanies heart arrhythmias, and is manifest typically as a missing QRS wave complex due to the heart's substitution of an ectopic beat for normal activity. FIGS. 3A, B, C and D show typical approximate waveforms at four significant points in the MR detector circuitry. These points are labeled on the schematic diagram. Field-effect-transistor 143 and resistor 144 form a constant current source which charges capacitor 142 at a rate of approximately 1.5 volts per second. The resultant voltage ramp is shown in FIG. 3A. Capacitor 142 is suddenly discharged each time an R-wave is declared by the R-wave detector. The discharge is accomplished by transistor 141 which receives a strong forward bias from transistor 109 through the resistance-capacitance network of components 138-140 each time an R-wave is detected. If an R-wave is skipped (i.e., the period between R-waves suddenly elongates), the discharge fails to take place and the voltage ramp on capacitor 142 continues upward, as shown between the 3- and 5-second points on FIG. 3A. Transistors 145 and 147 in association with resistors 146, 148 and 149, form a compound emitter follower without significant offset voltage between input and output. Thus, when the voltage at point (6) becomes greater than that at point (7), point (7) will follow point (6), but when the voltage at point (6) decreases below that at point (7), point (7) will not track point (6), but instead capacitor 150 will slowly discharge through resistors 148 and 149. The approximate voltage waveform at point (7) is shown in FIG. 3A. Components 153-156 form a simple dc-supply which holds the cathode of diode 158 about 6 volts more positive than the drain of transistor 143. This produces a strong reverse bias on the junction of field-effect-transistor 159, thus inducing a high resistance between source and drain in this element and isolating point (8) from point (7). However, when the collector of transistor 109 becomes strongly positive, transistor 152 is forward biased, pulling the gate of transistor 159 to a low voltage and greatly reducing the source-drain resistance of transistor 159, thereby connection capacitor 150 to capacitor 160. Since the latter capacitor is much smaller in value than the former, the voltage on capacitor 160 becomes very nearly equal to the voltage on capacitor 150 at the instant it was connected to capacitor 160. When the collector of transistor 109 becomes strongly negative once again, transistor 159 again goes to the high-resistance state and capacitor 160 is isolated from capacitor 150. Thus, the voltage waveform at point (8) is shown in FIG. 3C. Transistors 162 and 163 form a differential voltage comparator; whenever the voltage at the base of transistor 163 (point 9) becomes less than that at the base of transistor 162, the latter element will draw sufficient current through resistor 165 to forward bias transistor 167, in turn forward biasing transistor 170 and causing its collector to become strongly positive. The voltage at the base of transistor 162 is established by that at point (8) through resistor 161. The voltage waveform at point (9) is about two-thirds of that at point (7). When a missing R-wave is encountered, capacitor 142 will not be discharged at the normal point in time, and will continue to charge, as shown at the 4-second point in FIG. 3A. Thus, the voltage at point (9) will also continue to increase as shown in the corresponding point on FIG. 3D. Meanwhile, point (8) has remained isolated from point (7) since transistor 109 has not produced a positive pulse

to turn on transistor 159. Consequently the voltage at point (8) will remain at a nearly constant level. Finally, at the point labeled X on FIGS. 3C and 3D, the voltage at point (9) will exceed that at point (8), forward biasing transistor 170 by the process described above. A strongly positive voltage at the collector of transistor 170 declares a missing R-wave.

It is very important to note that this circuitry will detect a missing R-wave regardless of the rate of arrival of R-waves from the heart. Faster rates will reduce the peak height of the waveform in FIGS. 3A and 3B and decrease the period of this sawtooth waveform. The other waveforms in FIG. 3 will be scaled proportionately since they are all dependent on the waveform at point (7). But at the occurrence of a missing R-wave, a crossover point for the voltages at points (8) and (9) will be caused by the same train of events as detailed above for a heart rate of one beat per second. Slower heart rates will cause the waveform in FIG. 3A to grow in magnitude and lengthen in period, but once again, the crossover between the voltages at points (8) and (9) will be established as above.

Components 172-192 form a logical network with which any of several different modes of operation of the overall circuitry may be selected by manipulation of switches 190-192. Description of the circuitry will not be given in great detail, since similar logic has been used by many others in the past and the circuitry employed here is not unique in principle. Switch 190 controls the operation of the abnormal-heart-rate detector circuitry: when the switch is closed the circuit is active, when it is open, the circuit is an abnormal. When the circuit is active, an abnormal heart rate will turn on the tone alarm 205 by forward-biasing transistor 194. When switches 191 and 192 are both closed, both the MR detector and the AV detector are deactivated. When switch 191 is open and switch 192 is closed, the MR detector is activated and the AV detector is deactivated. Transistor 188 is forward-biased so if a missed R-wave is detected, the collector of transistor 170 will go strongly positive, forward-biasing transistor 187 and pulling its collector strongly negative. Each missed R-wave will deliver a negative pulse to the alarm criteria circuitry, discussed below, through capacitor 193. When switch 191 is closed and switch 192 is open, the AV detector is activated and the MR detector is deactivated. Transistor 184 is forward-biased so if an abnormal ventricular action is detected, transistor 183 will be forward-biased and its collector pulled strongly negative. Each AV will deliver a negative pulse to the alarm criteria circuitry through capacitor 193. When both switches 191 and 192 are open, an AV will produce a positive charge on capacitor 177 and forward-bias transistor 180. A strong positive voltage on the collector of transistor 109, caused by a detected R-wave, will forward-bias transistor 175 and discharge capacitor 177, turning off transistor 180. However, if a positive pulse is received from the MR detector before a positive pulse from the R-wave detector, transistor 181 will be forward biased and the collector of transistor 180 will be pulled strongly negative, delivering a negative pulse to the alarm criteria circuitry through capacitor 193.

Transistors 195-197, and their associated passive components form a touch-plate-activated monostable multivibrator. This circuit provides a means by which the patient himself can turn on the tone oscillator discussed below. The basic principles of this multivibrator

have been previously described in the patent application of one of us, Ser. No. 260,775, filed June 8, 1972. Output from the multivibrator causes transistor 136 to be strongly forward-biased which in turn causes transistor 194 to be turned on, activating the tone.

Unijunction transistors 198 and 199, and associated passive components, form the alarm criteria circuitry. Through the use of this circuitry it is possible to require that more than one occurrence of an abnormal heart activity take place in a certain programmable time period before the alarm itself will be activated. The basic circuitry for this alarm criteria operation has been previously described.

The actual alarm is manifest through the operation of a tone oscillator which drives a loudspeaker or sounder. The oscillator may be activated through the forward-biasing of either transistor 200 or transistor 194. The former is activated by the alarm criteria circuitry, and the latter by the abnormal-rate circuit or touch-plate circuit. The tone oscillator itself is of conventional design. When the oscillator is keyed on, its frequency is modulated by the voltage waveform appearing at the output of amplifier 6. This waveform is amplified by operational amplifier 201 and delivered to the base of transistor 205 which controls the period of oscillation of the tone oscillator. The advantage of modulating the tone oscillator with the patient's ECG is that the ECG may thereby be transmitted over the telephone for recording and analysis in real-time at the doctor's office.

With reference to the abnormal-ventricular-activity detector, it was stated that the peak amplitudes of the positive first derivative is averaged over approximately a 20-second period. In actuality the optimum averaging period varies between about 10 and about 40 seconds, depending on the ECG characteristics of a given patient. The same considerations apply to the averaging period for the amount of time spent by the positive first derivative above the adaptive threshold; the optimum period may vary from about 10 to about 40 seconds. Similarly, the adaptive threshold described as one-eighth of the average peak amplitude of the positive first derivative may be optimized anywhere between about one-sixteenth and one-fourth of the average peak amplitude.

We claim:

1. A cardiac arrhythmia detector comprising in combination:

- a. a set of electrodes adapted to be applied to the body,
- b. means connected to said electrodes for generating a signal proportional to the first time derivative of the input signal applied to said electrodes,
- c. means for measuring the peak amplitude of said first time derivative of each waveform due to either normal or abnormal ventricular pumping action,
- d. means for establishing the running average of the last 10 to 40 of said peak amplitudes,
- e. means for establishing an amplitude threshold defined as $1/16$ to $1/4$ of said running average,
- f. means for measuring the time interval spent above said amplitude threshold by each first time derivative waveform due to either normal or abnormal ventricular pumping action,
- g. means for establishing the running average of the last 10 to 40 of said time intervals,

h. means for comparing each measured peak according to (c) above, to the running average according to (d) above, to identify whether or not the individual peak differs from said running average by more than plus or minus 25 percent, and classifying as normal, on this basis, those for which the difference is less than plus or minus 25 percent, and abnormal those for which the difference exceeds plus or minus 25 percent,

i. means for comparing each time interval according to (f) above, to the running average according to (g) above, to identify whether or not the individual interval differs from said running average by more than plus or minus 25 percent, and classifying as normal, those for which the difference is less than plus or minus 25 percent, and abnormal those for which the difference exceeds plus or minus 25 percent,

j. means for eliminating from consideration, for purposes of establishing the running averages of (d) and (g) above, those derivative waveforms found to be abnormal according to (h) or (i) above, and

k. alarm means and means for delivering a positive output to said alarm means whenever a derivative waveform is classified as abnormal according to either (h) or (i) above.

2. The cardiac arrhythmia detector of claim 1, having in addition thereto a modulator, sounder and touch plate means and means for actuating same from said generated signal whereby a wearer can touch said plate and means whereby said sounder will issue an audible ECG signal which can be sent over a telephone or the like.

3. The cardiac arrhythmia detector of claim 1, having means inhibiting said alarm until a plurality of abnormal indications are detected within a time period and means for varying said period.

4. A detector for missing R waves comprising in combination:

- a. a set of electrodes adapted to be applied to the body,
- b. circuit means connected to said electrodes, said circuit means including means to identify a missing R wave, including means for measuring the interval between each successive pair of normal R waveforms, and means for declaring a missing R wave when the current interval is greater than about 1.5 times the immediately previous interval, alarm means and means for actuating said alarm means upon said declaration.

5. A combined cardiac arrhythmia detector and missing R wave detector comprising in combination:

- a. a set of electrodes adapted to be applied to the body,
- b. first circuit means connected to said electrodes, said first circuit means including means whereby the output of the said means is proportional to the first time derivative of the input signal applied to said electrodes,
- c. means for measuring the peak amplitude achieved by the first time derivative of each waveform due to either normal or abnormal ventricular pumping action, any time said derivative waveform exceeds an amplitude threshold defined as $1/16$ to $1/4$ of a running average,
- d. means for measuring the time interval spent above said amplitude threshold by each first time deriva-

- tive waveform due to either normal or abnormal ventricular pumping action,
- e. means for establishing the running average of the last 10 to 40 of said time intervals,
 - f. means for comparing each so measured peak according to said running average to identify whether or not the individual peak differs from said running average by more than plus or minus 25 percent, and classifying as normal, on this basis, those for which the difference is less than plus or minus 25 percent, and abnormal those for which the difference exceeds plus or minus 25 percent,
 - g. means for comparing each said time interval to the said running average to identify whether or not the individual interval differs from said running average by more than plus or minus 25 percent, and classifying as normal, those for which the difference is less than plus or minus 25 percent and abnormal those for which the difference exceeds plus or minus 25 percent,
 - h. means for eliminating from consideration, for purposes of establishing said running averages, those derivative waveforms thus found to be abnormal,
 - i. means for producing a first output signal whenever a derivative waveform is classed as abnormal,
 - j. a second circuit means connected to said electrodes, said means, including means to identify an asynchronous R wave, wherein said means comprises circuitry for measuring the interval between each successive pair of R waveforms,, and means for declaring an asynchronous R wave when the current said interval is greater than about 1.5 times the immediately previous interval, and means for

- producing a second output signal upon said declaration, and
- k. alarm means and means for actuating said alarm means by the occurrence of said first output signal followed immediately by said second output signal.
6. A cardiac R wave detector comprising in combination:
- a. a set of electrodes adapted to be applied to the body,
 - b. circuit means connected to said electrodes for generating a signal proportional to the first time derivative of the input signal applied to said electrodes,
 - c. said circuit means including means for detecting a first time derivative which is positive in polarity and has a slope on the heart waveform of about 0.02 volts per second and having a time duration of about 50 milliseconds, and means for detecting immediately following said first time derivative, a second first derivative which is negative in polarity, has a slope on the heart waveform of about minus 0.02 volts per second and having a time duration of about 50 milliseconds, and
 - d. means for producing an output signal each time that both criteria are met.
7. The cardiac R wave detector of claim 6, having a means whereby an alarm is sounded when heartbeats are slower than a normal standard and means for selecting said standard.
8. The cardiac R wave detector of claim 6, having means associated therewith whereby an alarm is sounded when heartbeats are faster than a normal standard and means for selecting said standard.

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