CARDIAC DIAGNOSTIC METHOD

Filed Jan. 7, 1959 8 Sheets-Sheet 1

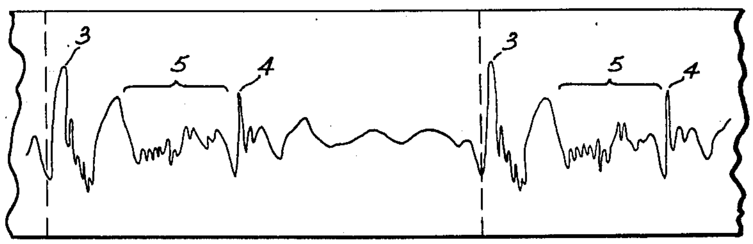

Fig. 1.

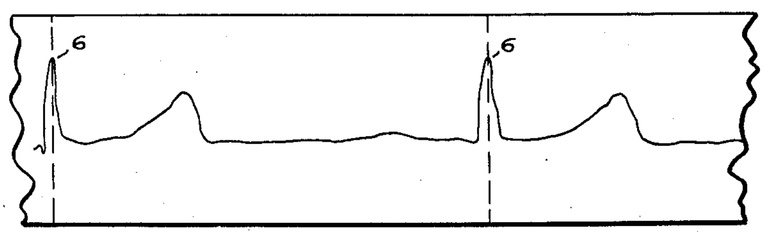

Fig. 2.

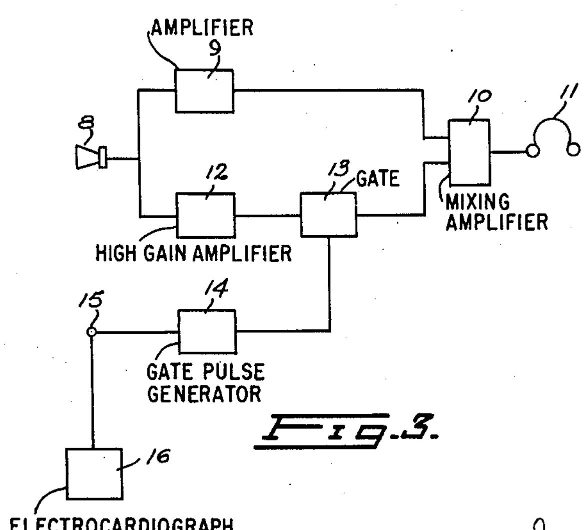

Fig. 3.

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
His attorneys

April 24, 1962 R. S. RICHARDS 3,030,946

CARDIAC DIAGNOSTIC METHOD

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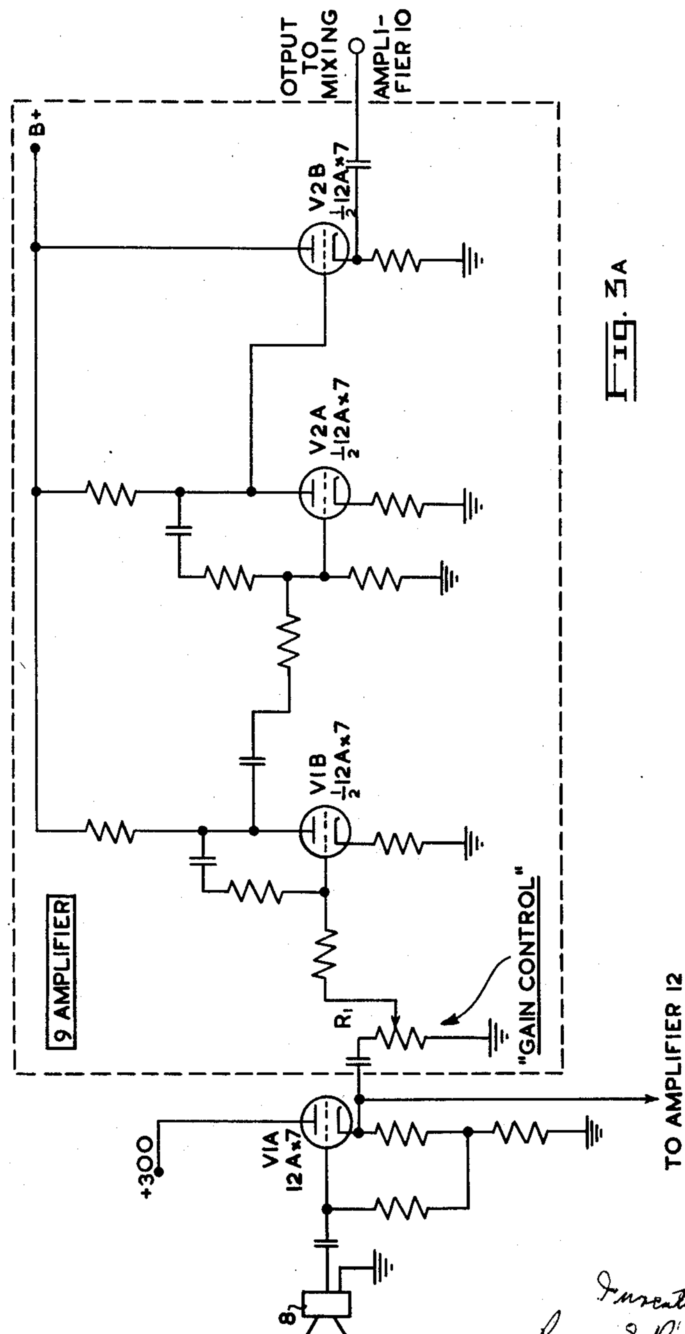

Fig. 3A

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
his attorneys

April 24, 1962 R. S. RICHARDS 3,030,946

CARDIAC DIAGNOSTIC METHOD

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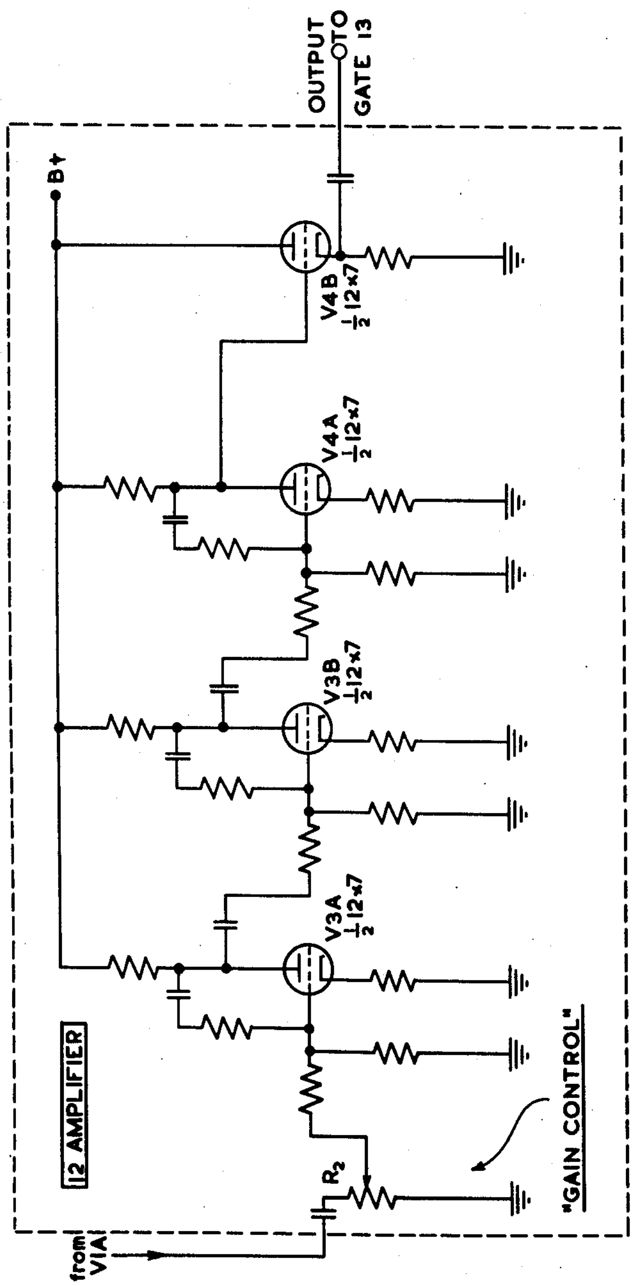

Fig. 3B

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
his attorneys

CARDIAC DIAGNOSTIC METHOD

Filed Jan. 7, 1959 8 Sheets-Sheet 4

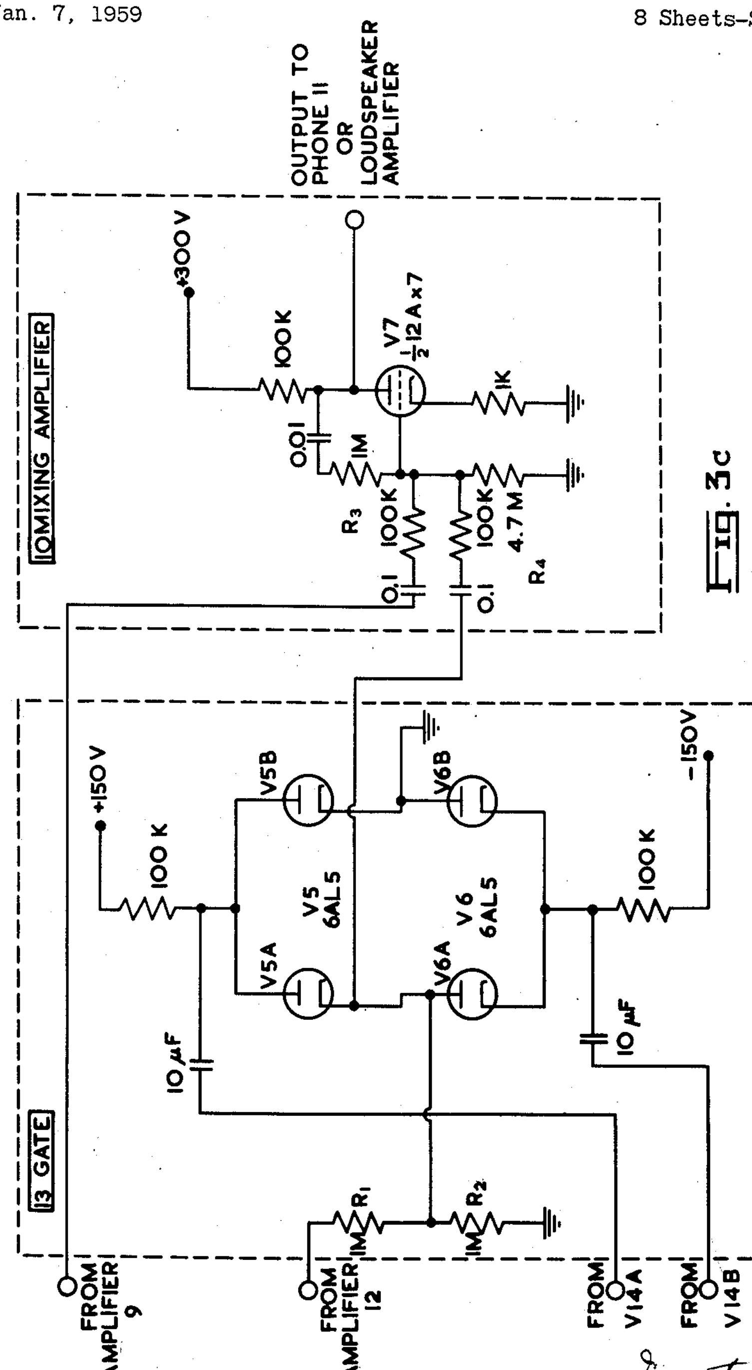

Fig. 3c

Inventor
Roger S. Richards
by Melvin W. Sandmeyer
his attorney

April 24, 1962 R. S. RICHARDS 3,030,946

CARDIAC DIAGNOSTIC METHOD

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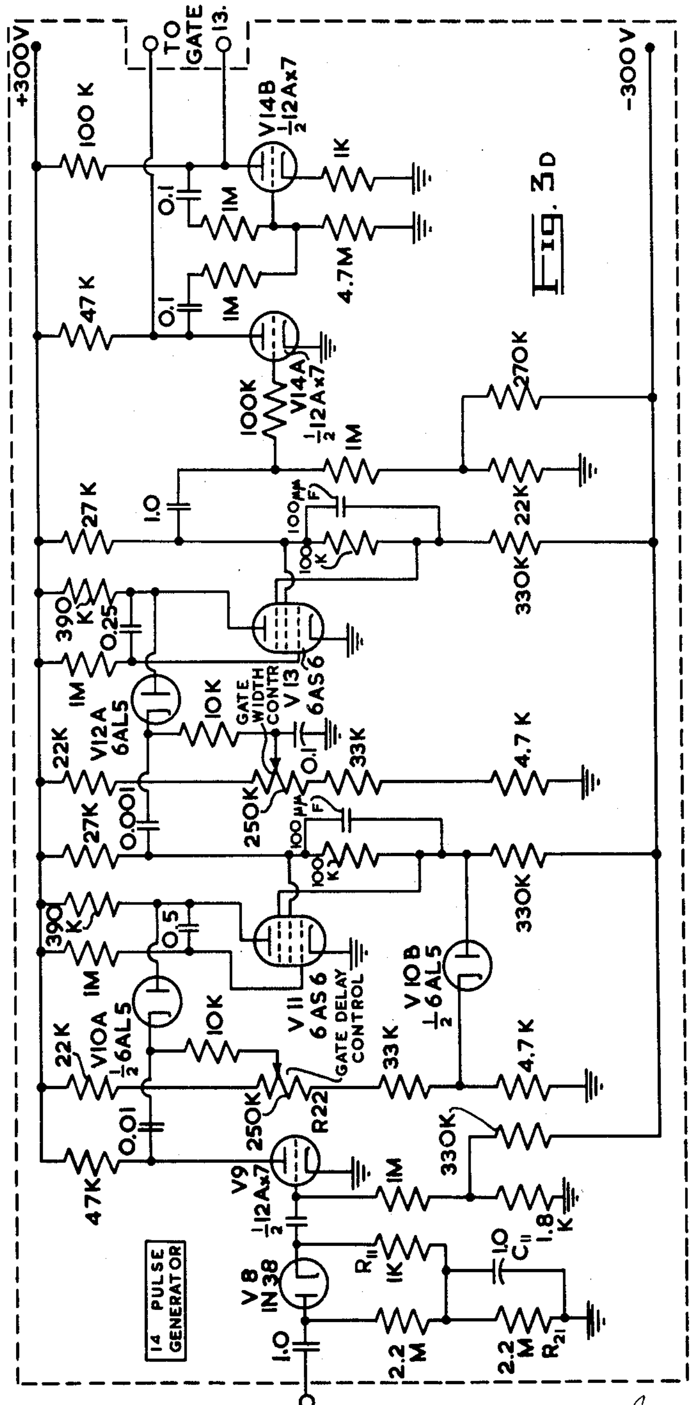

Fig. 3D

Inventor
Roger S. Richards
By Stevens, Davis, Miller & Mosher
his attorneys

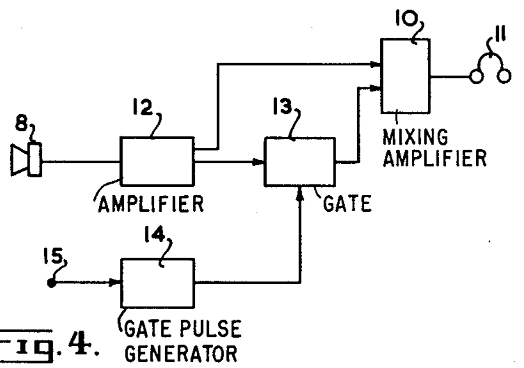

Fig. 4.

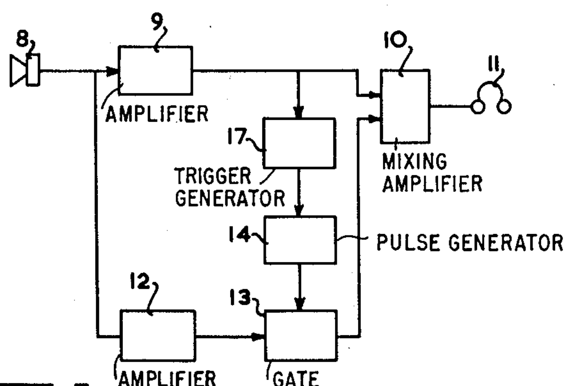

Fig. 5.

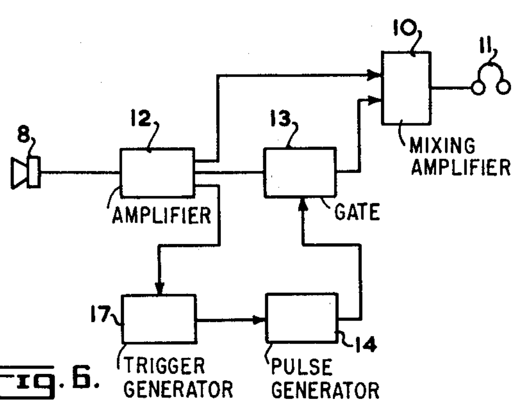

Fig. 6.

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
his attorneys

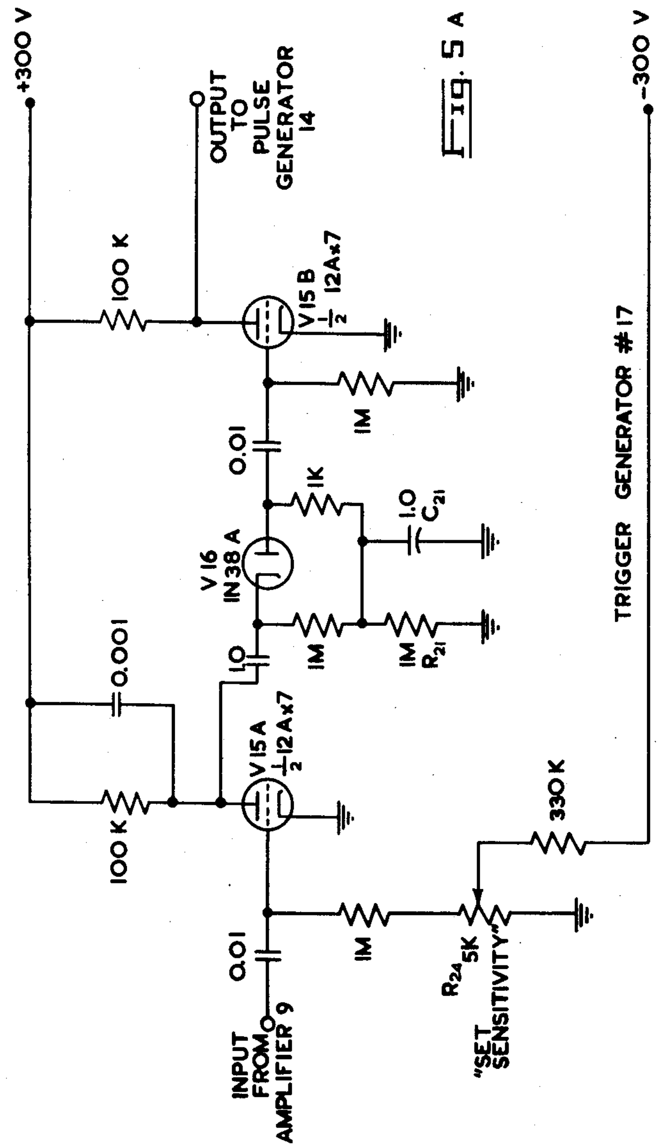

Fig. 5 A

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
his attorneys

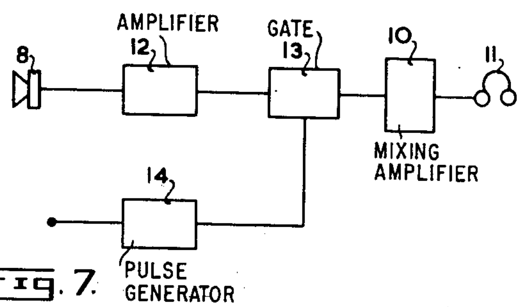

Fig. 7.

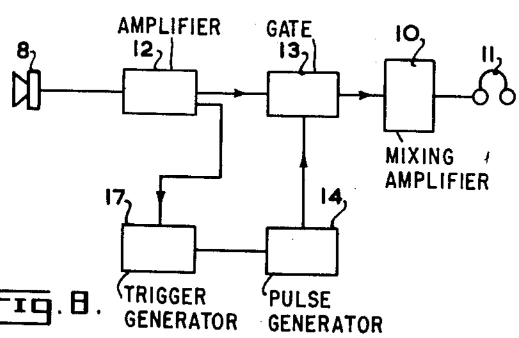

Fig. 8.

Inventor
Roger S. Richards
by Stevens, Davis, Miller & Mosher
his attorneys

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Patented Apr. 24, 1962

3,030,946
CARDIAC DIAGNOSTIC METHOD

Roger S. Richards, Cumberland, Ontario, Canada, assignor to National Research Council, Ottawa, Ontario, Canada, a body corporate of Canada

Filed Jan. 7, 1959, Ser. No. 785,434
2 Claims. (Cl. 128—2.06)

This invention relates to a cardiac diagnostic method for the detection and study of low intensity heart sounds such as murmurs.

The detection of heart murmurs is difficult due to the presence of high intensity normal heart sounds which tend to obscure the weaker higher frequency murmurs. Diagnosis with the common stethoscope therefore, requires a high degree of skill on the part of the physician. It has been proposed to use frequency discriminating devices to isolate the murmurs, but these devices have been unsatisfactory. It is the aim of the present invention to provide a method for selectively amplifying the murmur sounds so that they may be more easily audible or otherwise detectable above normal heart sounds.

The normal heart sounds are rhythmic and are of greater intensity than many of the murmur sounds in which the physician is interested. The murmur sounds have the same rhythm but they usually occur during the interval between intensity peaks of the normal sounds. In the present invention this non-coincidence in time allows the murmurs to be amplified selectively with respect to the normal heart sounds.

This invention may be defined as a method of detecting heart sounds wherein the degree of amplification of said heart sounds is automatically periodically switched from a first degree to a second degree higher than said first degree, for intervals having a predetermined length and time relationship with respect to certain characteristic high intensity heart sounds.

For carrying out this method, the invention employs apparatus for detecting heart sounds, comprising a microphone sensitive to said heart sounds, a variable gain amplifier connected to said microphone, detecting means sensitive to the output of said amplifier, means sensitive to heart action to generate a synchronizing voltage having a predetermined time relationship with said heart sounds, and control means sensitive to said synchronizing voltage and operative to increase the gain of said amplifier for an interval in response to said synchronizing voltage. Said variable gain amplifier may consist of a single amplifier or, as in the preferred embodiment, may take the form of two amplifiers in parallel, one of which is switched by said control means.

Reference will now be made to the accompanying drawings in which,

FIG. 1 is a graphic representation of heart sound intensity variations with time, for a particular heart having a murmur,

FIG. 2 is an electrocardiograph trace made simultaneously with the heart sound trace of FIGURE 1,

FIG. 3 is a block diagram of a circuit illustrating a particular embodiment of apparatus for practicing the present invention,

FIGS. 3*a*, *b*, *c* and *d* together illustrate one form of a schematic circuit diagram of the circuit shown in block form in FIG. 3,

FIGS. 4, 5, 6, 7 and 8 are block diagrams illustrating alternative embodiments of the apparatus, and

FIG. 5*a* is a schematic circuit diagram of a trigger generator shown in block form in the embodiments of FIGS. 5, 6 and 8.

The trace of FIGURE 1 illustrates the relative magnitudes and time relationship of the normal, and murmur sounds, occurring in a particular heart. Numeral **3** refers to the peak accompanying a first normal heart sound while numeral **4** refers to a peak accompanying a second normal heart sound. For this particular heart, the murmur, numeral **5**, lies completely within the interval between normal heart sounds **3** and **4**. For different heart conditions, the time relationship between the murmur **5** and the normal sounds **3** and **4** will vary, but in general, the murmur does not coincide completely with either of the normal sounds. This figure points out the difference in intensities between the murmur and the normal sounds, which again will vary with different heart conditions, but in general is such as to make the murmur difficult to hear.

The circuit of FIGURE 3 provides, according to the present invention, a greater amplification of the murmur than of the normal heart sound. A crystal microphone **8** placed on the chest close to the heart detects all heart sounds, and the resulting signal is passed through an amplifier **9** and a mixing amplifier **10** whose output is connected to earphones **11**. This branch of the circuit amplifies all heart sounds by an equal factor, subsequently referred to as "a first degree." In parallel with the amplifier **9** is a higher gain amplifier **12** which is switched by a gate **13**. The gate **13** connects the high gain amplifier **12** to mixing amplifier **10** only for the duration of an actuating signal fed from a gate pulse generator **14**. Since the outputs of both amplifiers **9** and **12** are fed through the mixing amplifier **10**, the final output of the circuit includes all the heart sounds, but those occurring during the interval when amplifier **12** is connected are amplified to a greater extent ("a second degree" greater than said "first degree") than the remainder of the heart sounds.

By means of a synchronizing voltage at terminal **15** and suitable delay and pulse width adjusting circuits associated with the gate generator **14**, the amplifier **12** is caused to operate during the interval between sounds **3** and **4** of FIGURE 1 and only long enough to amplify the murmur sounds **5**. In practice the ratio of the gain of amplifier **12** to the gain of amplifier **9**, is about 100 to 1 so that the detection of the low intensity murmurs **5** is greatly improved.

The synchronizing voltage at terminal **15** may be derived from any periodic voltage which bears a constant time relationship with the normal heart sounds **3** and **4**. For example the peak of the QRS complex of the electrocardiogram, which is shown as the peak **6** in FIGURE 2, immediately precedes the first peak **3** in the heart sound trace of FIGURE 1. The peak **6**, since it has a constant time relationship with the peak **3** in the heart sound intensity trace, is therefore suitable as a synchronizing voltage for the gate pulse generator **14**. A conventional electrocardiograph is shown at **16**.

FIG. 3 has been drawn in block diagram form and it will be obvious to those skilled in the art that each block may represent various well known circuits. However, one specific circuit for each block will now be described. The details of the input stage and amplifier **9** are shown in FIG. 3*a*, amplifier **12** in FIG. 3*b*, the gate **13** and the mixing amplifier **10** in FIG. 3*c* and the pulse generator **14** in FIG. 3*d*.

Referring now to FIG. 3*a*, the microphone **8** is connected to amplifiers **9** and **12** via a cathode follower V1A. Cathode follower V1A is RC coupled to amplifier **9**, and a variable resistor R_1 provides gain control. The tubes V1B and V2A of amplifier **9** are resistance-capacitance coupled, each having its own plate to grid feedback loop. The last stage of amplifier **9** is a cathode follower V2B, and its output is fed to mixing amplifier **10**.

Referring to FIG. 3*b*, it will be seen that the output of cathode follower V1A is applied to amplifier **12** by an R.C. coupling, with the variable resistor R_2 providing gain control. Amplifier **12** is identical to amplifier **9** except that an additional stage of amplification is used. The cathode follower output of amplifier **12** is fed to gate **13**.

Gate **13**, as shown in FIG. 3*c*, may take the form of a well known four diode switch. The normally conducting diode bridge consisting of diodes V5A—B and V6A—B has a low resistance (about 100 ohms), and short circuits R_2 which is of the order of 1 megohm. R_1 is approximately 1 megohm also, and accordingly the output of the gate is reduced to one ten thousandths of the input. On the other hand, if the diodes are caused to be non-conducting, the output of the gate **13** will be one-half of the input. The diodes are rendered non-conducting by applying a large positive pulse to the cathodes of V6 and a simultaneous large negative pulse to the anodes of V5. These gate pulses are received from pulse generator **14**.

Mixer **10**, to which the outputs of gate **13** and amplifier **9** are applied, consists of a normal inverse feedback linear adder utilizing a single tube V7. Plate to grid feedback in V7 serves to reduce to a small value (e.g. 1,000 ohms) the input impedance of the tube seen at the grid. Thus the signals fed to the grid via isolating resistors R_3 and R_4 will add in linear fashion without interaction one upon the other.

In the pulse generator **14** shown in FIG. 3*d*, tubes V10 V11 form a normal screen coupled monostable phantastron which is triggered from the output of a standard electrocardiograph amplifier (EKG) applied to terminal **15**. Input tube V8 of pulse generator **14** is a self-biased diode limited. Positive pulses from the EKG cause V8 to conduct and charge capacitor C_{11}. This charge will partially leak away between pulses, but by choosing the time constant C_{11} R_{21} to be larged than the interval between pulse and the charging time constant $C_{11}R_{11}$ to be of the same order as the duration of the pulse, a fairly steady charge is soon built up by C_{11} which holds a steady bias on the diode V8. Unless the voltage of an input pulse exceeds this bias value, there will be no output. The circuit can in this way be designed to respond only to the tips of the largest pulses to arrive, provided such pulses arrive in regular sequence, and after an equilibrium has been established.

The pulses passed by V8 are amplified and inverted by the tube V9 so as to be of a suitable size to trigger the phantastron V10—V11. The output of this phantastron, taken at the screen of V11, is a positive going rectangular pulse whose width is linearly proportional to the plate catching voltages determined by the setting of "Gate Delay Control" R_{22}. This pulse is differentiated and passed to a second phantastron comprising tubes V12 and V13. This second phantastron, since it is sensitive only to negative trigger pulses, forms a pulse which starts when a pulse from the first phantastron is finishing. The pulse width of the second phantastron is controlled by the caught anode potential as determined by the setting of the "Gate Width Control" R_{23}. It will be seen that the time lapse between an EKG pulse and a pulse from the second phantastron will depend on the pulse length of the first phantastron, and this is adjustable. This circuit therefore generates a pulse of predetermined length and of predetermined delay with respect to an EKG pulse.

Tube V14A serves to improve the pulse shape of the second phantastron while V14B is a unity gain amplifier which merely inverts the pulse. The two anodes of V14A and V14B therefore produce a pulse pair in push pull, for turning off the diodes V5 and V6 of the gate **13**.

The detailed circuits which have just been described should be considered only as examples of the physical embodiment of the invention, and it will be obvious to those skilled in the art that other circuits could be used. In the drawings, preferred tube types for the circuits disclosed have been indicated.

If it is desired to detect only the murmur sounds, then amplifier **9** may be dispensed with. The system is then as shown in FIG. 7.

A further alternative to the circuit shown in FIG. 3 is shown in FIG. 4. Here the amplifier **9** is dispensed with and the output of amplifier **12** is fed to mixer **10** as well as to gate **13**. Summation of the outputs of amplifier **12** and gate **13** is performed by mixer **10** and thus a higher degree of amplification is obtained when gate **13** is switched on. The output of this system contains all the heart sounds. The detailed circuits already described may be readily adapted to suit this alternative embodiment.

FIG. 5 illustrates an an alternative method of triggering the pulse generator controlling the gate **13**. Instead of using the output of an electrocardiograph for this triggering, as in the embodiments thus far discussed, this circuit derives a trigger signal from heart sound amplifier **9**. It will be seen that the circuit of FIG. 5 is similar to that of FIG. 3 except for the addition of trigger generator **17** whose output feeds pulse generator **14**. The detailed construction of amplifiers, mixer and pulse generator have already been described with regard to FIG. 3. One form of the trigger generator **17** is shown in FIG. 5*a*.

In FIG. 5*a*, the output of amplifier **9** is applied to tube V15A of the generator **17**, this tube having a negative bias adjustable by means of "set sensitivity" resistor R_{24}, and having a smoothing capacitor in the plate circuit. The output of amplifier **9** will always contain a pair of pulses relatively closely spaced, followed by a longer interval, corresponding to the heart sounds as described with reference to FIG. 1. These pairs of pulses derived from the first and second heart sounds will almost always be the largest pulses present. It will also be observed that the "pulses" just referred to may in fact be the envelope of a rapidly oscillating voltage. As a result of applying these pulses to the tube V15A, the tube will be alternately conducting and non-conducting and its output will be a pair of simple, negative going pulses. The anode of V15A feeds a self-biased diode limiter **V16**, similar to V8 of pulse generator **14**, except that the diode is inverted and the time constants are a little shorter. Assuming equilibrium conditions to exist, the diode will be so biased as to accept only the tips of the large pulses corresponding to the first and second heart sounds. The input voltage to trigger generator **17** will be adjusted so as to be large enough to cause **V15A** to be saturated, thus causing **V15A** to act as a limiter: both output pulses applied to **V16** will therefore be of the same amplitude. However, because the pulses come in rapid sequence, followed by a relatively longer quiet period, the circuit can, by suitable choice of the time constant C_{21} R_{21}, be made to respond only to the pulse corresponding to the first heart sound. The output from diode V16 is fed to the grid of amplifier triode V15B and the trigger pulses thereby generated in the plate of this tube are fed to pulse generator **14** which has already been described. By means of the circuit just described, the triggering pulses from the electrocardiograph have been replaced by trigger pulses derived from the output of amplifier **9**. Suitable tube types have again been indicated on the drawings.

It will be obvious from the last paragraph and from the previous description of FIG. 4, that a circuit such as that shown in FIG. 6 may be used. This embodiment is similar to that of FIG. 4 but the triggering pulses are derived from the output of amplifier **12** as described with relation to FIG. 5.

FIG. 8 shows a circuit similar to that of FIG. 7 except that the triggering pulses are derived from the output of amplifier **12** by means of the trigger generator **17** already described.

This application is a continuation-in-part of application Serial No. 561,571 filed January 26, 1956 (now abandoned).

I claim:

1. A method of detecting heart sounds characterized by periodic high intensity peaks and low intensity murmurs occurring in an interval between said peaks, com-

prising electrically detecting and continuously amplifying said sounds by a first degree, generating an actuating signal in synchronism with said interval, amplifying said sounds by a second degree in response to said signal and only during said interval, said second degree being substantially higher than said first degree, and detecting the combined output of said amplification steps.

2. A method of detecting heart sounds characterized by periodic high intensity peaks and low intensity murmurs occurring in an interval between said peaks, comprising continuously electrically detecting said sounds,

generating an actuating signal in synchronism with said interval, amplifying said sounds only in response to said signal and only during said interval, and detecting said amplified sounds.

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