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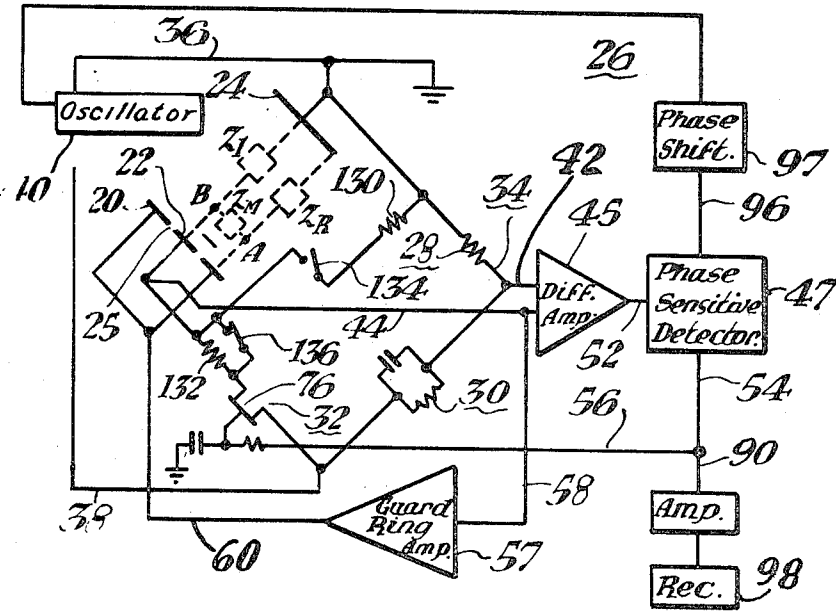
[54] **PHYSIOLOGICAL IMPEDANCE-MEASURING
APPARATUS**
15 Claims, 11 Drawing Figs.

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128/DIG. 4
[51] Int. Cl. A61b 5/05
[50] Field of Search 128/2.1,
2.06, DIG. 4

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ABSTRACT: In a physiological impedance-measuring apparatus, the combination comprising a pair of electrodes for spaced application to predetermined locations of the anatomy, an additional electrode spacedly juxtaposed to one of said first-mentioned electrodes, circuit means for applying a measuring voltage across said first-mentioned electrodes and a portion of said anatomy therebetween and for applying a potential to said additional electrode similar to that on the associated one of said first-mentioned electrodes, and an impedance-measuring circuit coupled to said electrodes for measuring anatomical impedance only between a predetermined pair of said electrodes. In certain applications wherein said circuit means include a unity-gain amplifier for applying a substantially identical potential at lower impedance to said additional electrode for driving the same.



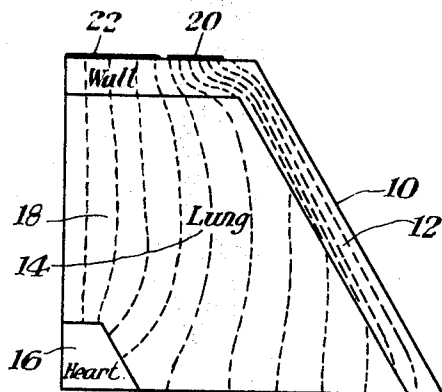


Fig. 1.

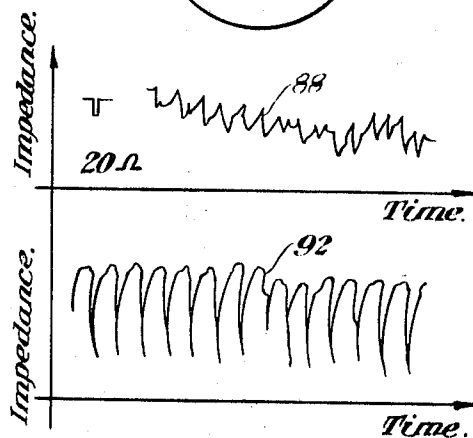
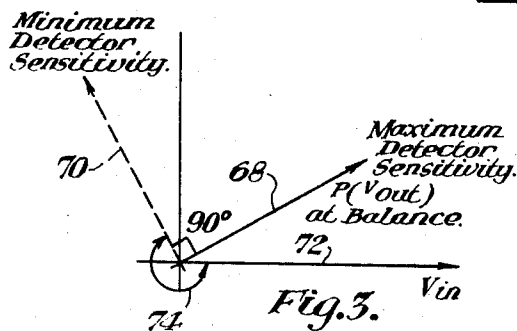
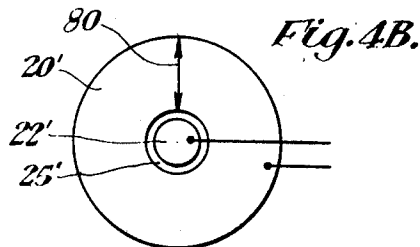
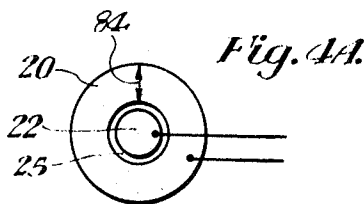
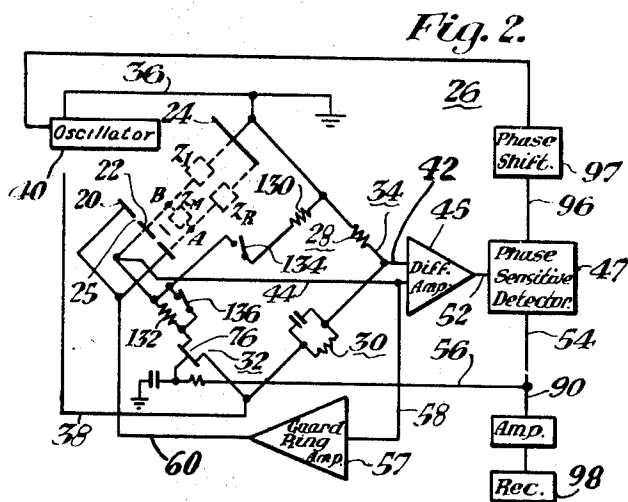
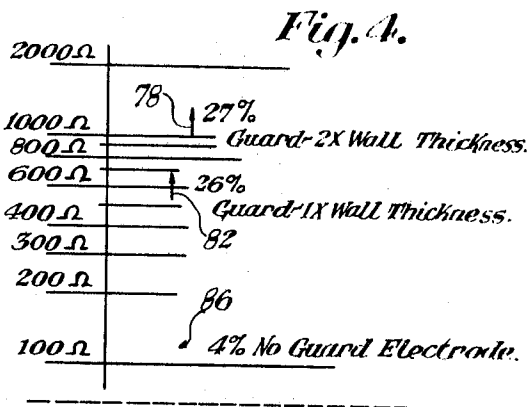
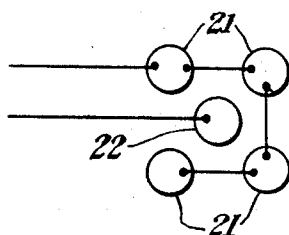


Fig. 5.



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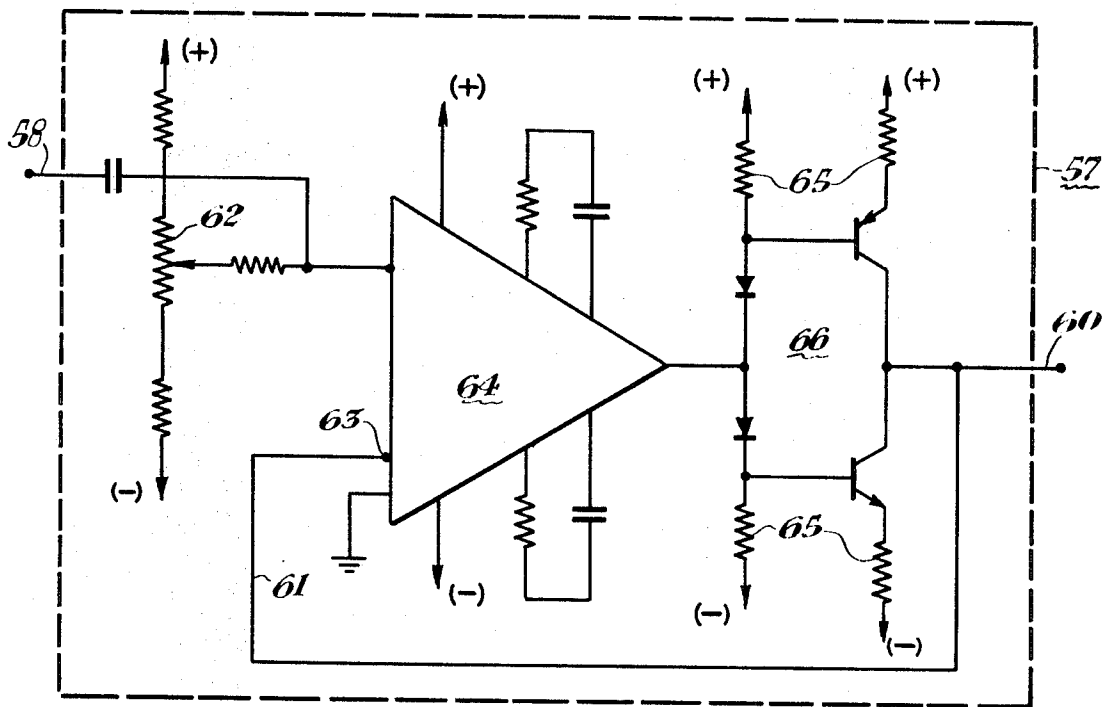


Fig. 6.

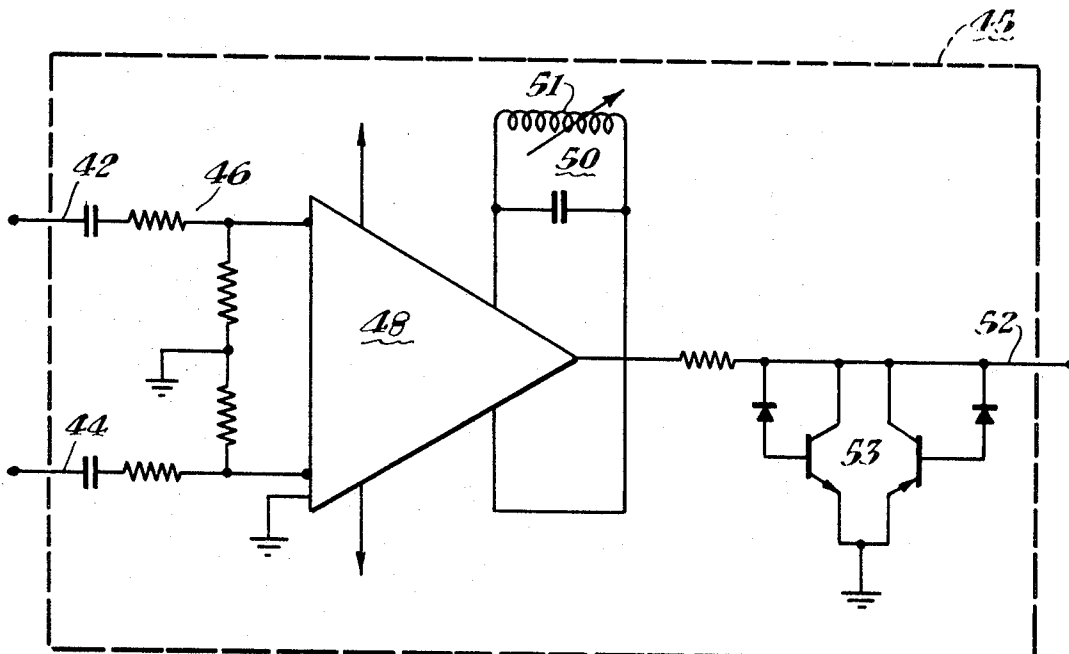


Fig. 7.

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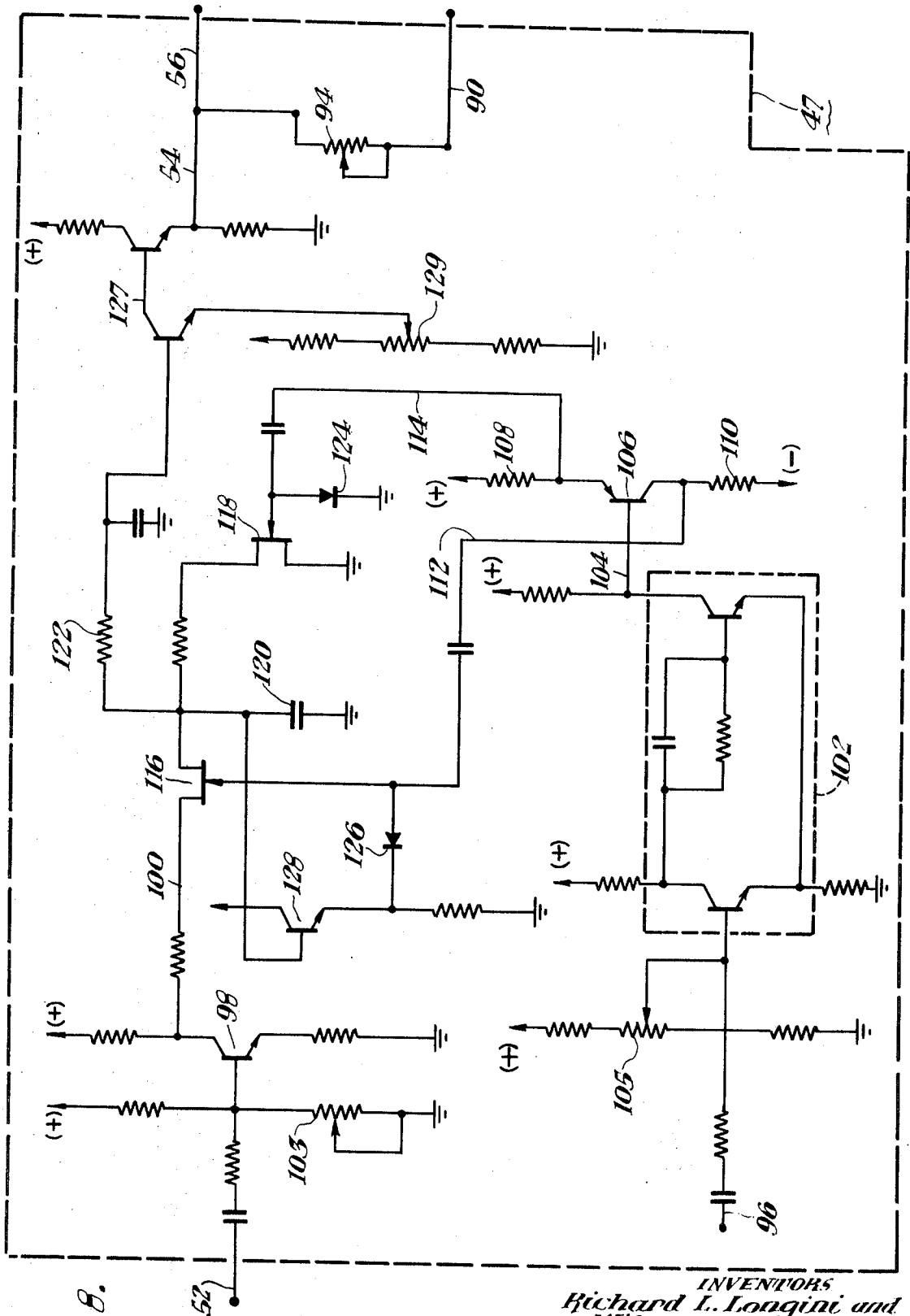


Fig. 8.

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PHYSIOLOGICAL IMPEDANCE-MEASURING APPARATUS

The present invention relates to apparatus for making physiological measurements and more particularly to apparatus of the character described for measuring electrical impedances of the thoracic regions.

Although our invention is adaptable for measuring electrical impedances of various parts of the anatomy, it will be described primarily in connection with the thoracic regions of the body, as the measurement of respiration is desirable in a large number of diagnostic situations. Useful measurements extend from simple detection of apnea to accurate, volumetric determinations for physiological studies of respiration. A good indication of cardiac output can also be obtained.

A reliable impedance pneumograph for measuring the electrical impedance of the human lung has eluded researchers for many years. It is well known that lung impedance changes by about 50 percent during inspiration and expiration. As this impedance is a strong function of the physiological condition of a subject, a reliable measurement of lung impedance, therefore, would be extremely useful in diagnostic and physiological studies. Previous impedance pneumographs have been unreliable both as diagnostic instruments or long term monitors owing to the presence of other significant nonpulmonary, transthoracic impedances. Previously proposed pneumographs have been unable to discriminate accurately among the several transthoracic impedances for the purpose of yielding an accurate measurement of lung impedance.

Certain prior pneumographs use small bilateral electrodes and detect impedance changes of only about 2 to 10 percent of the resting impedance for a maximum breath. During normal respiration, the breathing signal is very small and is easily contaminated by spurious signals resulting from other transthoracic impedance variations.

Owing to these difficulties, known impedance pneumographs are not amenable to situations requiring accurate pneumographic measurements. Such measurements have been made on the other hand by accepted volumetric or flow techniques. Because of the discomfort and inconvenience to the patient, the number of routine measurements which can be made by such means is very limited.

We overcome these difficulties of the prior art with the provision of physiological electrical impedance-measuring apparatus capable of distinguishing among several bodily impedances. Specifically, our measuring apparatus is capable of measuring changes in the impedance of lung tissue, which fills most of the thoracic cavity, against a background of much lower and hitherto interfering impedances of the thoracic wall structures. When used as an impedance pneumograph our measuring apparatus, therefore, is capable of sensitive and accurate measurements of lung tissue ventilation with very little discomfort and inconvenience. Most importantly, interference or disturbance of rhythmic breathing patterns is considerably reduced.

We accomplish these desirable results by providing in a physiological electrical impedance-measuring apparatus, the combination comprising a pair of electrodes for spaced application to predetermined locations of the anatomy, an additional electrode spacedly juxtaposed to one of said first-mentioned electrodes, circuit means for applying a measuring voltage across said first-mentioned electrodes and a portion of said anatomy therebetween, an active means for applying a potential to said additional electrode similar to that on the associated one of said first-mentioned electrodes, and an impedance-measuring circuit coupled to said electrodes for measuring anatomical impedance only between a predetermined pair of said electrodes.

We also provide similar measuring apparatus wherein said circuit means include a driving circuit for said additional electrode for driving said additional electrode at a potential substantially identical to that on said one electrode.

We also provide similar measuring apparatus wherein said circuit means include a substantial unity gain amplifier for applying a substantially identical potential at lower impedance to said additional electrode for driving the same.

We also provide similar measuring apparatus wherein said impedance-measuring circuit includes a phase-discriminating circuit sensitive only to a predetermined phase of impedance changes between said electrodes.

FIG. 1 is a schematic view of current flow in one quadrant of a simplified model of a human thorax;

FIG. 2 is a structural and schematic diagram of electrode and circuit components of our measuring apparatus;

FIG. 3 is a phase diagram illustrating the monitoring of changes in the resistive component of transthoracic impedance;

FIG. 4 is a logarithmic chart of test results with corresponding electrode structures;

FIGS. 4A and 4B and 4C are plan views of electrode structures forming part of our measuring apparatus;

FIG. 5 is a graphical representation of pneumographic impedance changes for differing conditions;

FIG. 6 is a schematic circuit diagram of one form of guard ring amplifier suitable for use in the circuit of FIG. 2;

FIG. 7 is a schematic circuit diagram of one form of differential amplifier suitable for use in the circuit of FIG. 2; and

FIG. 8 is a circuit schematic of one form of phase sensitive detector suitable for use in the circuit of FIG. 2.

Referring now to FIG. 1 of the drawings, the problem and a partial solution involved in measuring thoracic impedances are illustrated. A simplified model of the thoracic region, used for computational purposes, is denoted by reference character 10. The model illustrates a thoracic peripheral structure 12, lung tissue 14, and heart structure 16. For diagnostic and other physiological measurements, it is desirable to measure exclusively, if possible, the thoracic core impedance Z_1 in the region denoted by the reference character 18. Unless some means are provided for eliminating substantial inclusion of the much lower thoracic wall impedance Z_R spurious signals or measurements are obtained. Known pneumographic-measuring apparatus applies a pair of spaced electrodes to the thoracic region with the result that the output signal reflects primarily the much lower impedance combination, $Z_R Z_1 (Z_R + Z_1)^{-1}$, the combined effect of Z_R and Z_1 . As the undesired peripheral impedance Z_R predominates, respiratory changes in Z_1 do not effect a significant change in output in conventional measurements. See FIG. 4, arrow 86 and subsequent description.

We obviate this condition by providing a third electrode 20 spacedly juxtaposed to and desirably surrounding one of primary electrodes 22, 24 with the result that the measuring signal passes centrally through the thoracic region between the primary electrodes 22, 24 when applied thereto (FIGS. 1 and 2). The guard ring electrode 20 in this example surrounds the associated primary electrode 22 (FIGS. 4A, 4B, 4C) and is maintained at a potential similar to that of the associated primary electrode 22. Measuring currents between the primary electrodes 22, 24 thus are restricted so that they cannot bypass the lung tissue by flowing circumferentially through the peripheral thoracic structure. Accordingly, primarily the thoracic core impedance Z_1 is measured rather than a composite impedance reflecting primarily the much lower impedance Z_R of the thoracic periphery. In making such measurements it is necessary to obviate, as explained below, the effect of the skin and electrode paste impedance Z_M across space 25 between the primary electrode 22 and its guard electrode 20.

An exemplary bridge circuit for performing this measurement is illustrated in FIG. 2 of the drawings. The bridge and detecting circuit 26 includes three known impedances 28, 30 and 32 in three branches of the bridge network 34, with the thoracic core impedance Z_1 in the fourth branch when the electrodes 20-24 are applied to the thoracic region. Analogous impedances can be similarly measured in other areas of the anatomy. Desirably, high frequency potential is applied to the impedance bridge 34 on conductors 36, 38 from a conventional oscillator 40 capable of supplying a relatively high frequency alternating voltage. Suitable voltage sources (not shown) are provided for the oscillator 40 and other cir-

cuit components. In this arrangement the frequency of the source 40 is in the neighborhood of 100 kHz., which is sufficiently high to avoid stimulation of the body tissues, electrode polarization, and excessive skin impedance. The power level is well below that which would cause any perceptible heating of the tissues.

The bridge output is applied along conductors 42, 44 to differential amplifying circuit 45. The amplifying circuit 45 (FIG. 7) includes a differential input circuit 46, a differential, high-gain amplifier 48, and a gain control circuit denoted generally by reference character 50. A suitable integrated circuit differential amplifier 48 is available from Radio Corporation of America, catalog number CA3030. Variable inductor 51 is adjusted for maximum gain at the frequency of the bridge input voltage on conductors 36, 38. A clipper circuit 53 limits the peak-to-peak output signal (by clipping both positive and negative peaks) preparatory to shaping the signal to a square wave as described below relative to FIG. 8.

The output from the differential amplifying circuit 45 is supplied on conductor 52 to a phase-sensitive detector 47 (FIG. 8). The detector circuit 47 compares the bridge unbalance signal supplied to it on conductor 52 with a reference input signal on conductor 96, supplied through a phase shifting network 97 of conventional construction from supply oscillator 40, to produce an output on conductor 54. The detector output is fed back on conductor 56 to the voltage-controlled impedance 32 (FIG. 2) to maintain the bridge circuit 34 in dynamic balance. Because changes occur in both amplitude and phase of the transthoracic impedance with breathing, a complete dynamic rebalance of a bridge ordinarily requires adjustment of both the resistive and reactive components of the balancing impedance. The principles on which balancing of the bridge 34 relative to a single preselected phase (in this case that of the real or resistive component R_1 of Z_1) are elaborated upon below in connection with FIGS. 3 and 8 of the drawings.

In order to exclude the thoracic peripheral impedance Z_R and the skin (epidermic) impedance Z_M across the electrode gap 25 and thereby to limit measured impedance to the thoracic core impedance Z_1 , the potential of electrode 22 is sensed and coupled to the guard electrode 20 through guard ring amplifier 57. Desirably the guard electrode is driven in a manner that maintains its equipotential status (to the center electrode 22) although the guard electrode is in a considerably lower impedance path. A suitable guard ring amplifying or driving circuit is shown in FIG. 6 and is provided with input and output conductors 58, 60 connected to the electrode 22 (conductors 44, 58) and to the guard ring electrode 20 (conductor 60) respectively.

The guard ring amplifying circuit 57 includes a DC level adjustment 62, a conventional, high-gain differential amplifier 64 which can be similar to the amplifier 48 of F6 (RCA-CA3030) amplification and an output stage 66. Resistances 65 are for current-limiting purposes. The amplifier output on conductor 60 is fed back on conductor 61 to one of the differential input terminals 63 of the amplifier 64. Because the gain of the amplifier is very high and because the output stage 66 is in both the output and feedback circuits, the net gain of the amplifying circuit 57 is very nearly unity. The circuit 57, then, becomes an impedance transformer to drive the guard electrode 20 (which is in the considerably lower impedance circuit including the peripheral thorax structure) with a voltage which is maintained approximately identical with that on the associated primary electrode 22. Electrode 22 potential thus applied to the guard ring 20 ensures that the comparatively low impedance Z_R representing the peripheral thoracic impedance is confined between the guard ring 20 and the opposite primary electrode 24 (FIG. 2). On the other hand, the comparatively high impedance Z_1 representing the impedance of the thoracic core only, is confined to the primary electrode 22, 24 for sensing by the bridge 34 and related circuitry.

The unity-gain amplifying or driving circuit 57 thus applies a potential to the guard ring 20 which is substantially identical

to that of the associated primary electrode 22. In this example, the voltage on the guard electrode 20 is made as identical as possible with the voltage on electrode 22, as aforesaid. The driving circuit is essentially an impedance transformer which ensures substantial bypassing current through the peripheral thoracic structure. In consequence, the thoracic core and peripheral impedances (Z_1 , Z_R) are isolated between electrodes 22, 24 and 20, 24 respectively.

As shown in FIG. 3, vector 70 representing the reactive changes in core impedance is 90° out of phase with vector 68 representing the resistive changes in thoracic core impedance. This phase relationship is exact only at the condition of perfect bridge balance, i.e., of amplitude null. When the phase-sensitive detecting circuit 47 (FIG. 8) is properly aligned or tuned, it is completely insensitive to the orthogonal component represented by vector 70 (FIG. 3), so that $2V/2x_1=0$, where V denotes detector output signal voltage. Thus, the detector 47 is completely insensitive to the change in the reactive or imaginary component X_1 of the core impedance Z_1 . On the other hand, the detector 47 has maximum sensitivity to the change in thoracic resistive component R_1 represented by the vector 68, i.e. $2V/2R_1$ is a maximum. Vector 72 denotes the input voltage phase while angle 74 relates the phase lag of the reference input 96 through conventional phase shifting network 97 of the detector 47 to the signal voltage on conductor 52.

The bridge circuit 34 is maintained in dynamic balance, as aforesaid, by varying the impedance 32 (FIG. 2). The variable impedance 32 in this example is a field-effect transistor 76, the control electrode of which is coupled to feedback conductor 56 of the detecting circuit 47. Thus, the effective impedance of the variable impedance 32 is varied automatically in response to the bridge output. An operator of our measuring apparatus, therefore, is not required to add compensating capacitors across the bridge 34 to balance the capacitance of individual patients. Our measuring apparatus eliminates the need for a highly skilled operator and it is much less subject to operational error.

The phase-related detector 47 includes an input clipper stage 98 which cooperates with the output clipper stage 53 of the differential amplifying circuit 45 to shape the differential output signal into a substantially square waveform on conductor 100. At the same time, the reference input voltage on conductor 96 is similarly shaped by a Schmidt trigger circuit 102 to provide likewise a square waveform on conductor 104. The symmetries of the square waveforms on conductors 100, 104 are controlled by potentiometers 103, 105 respectively.

The reference square wave is applied to the base electrode of transistor 106, the emitter and collector circuits of which are supplied respectively with positive and negative supply voltages through equal load resistances 108, 110. As a result, a synchronous square wave is developed on conductor 112 connected to the collector circuit of the transistor 106 while a square waveform which is similar but 180° out of phase is developed on conductor 114 connected to the emitter circuit of the transistor 106.

These waveforms (conductors 112, 114) are applied respectively to the gate electrodes of a pair of field-effect transistors 116, 118. The transistors 116, 118 form a series-shunt chopper, with the transistors 116, 118 acting as electronic switches which open and close in alternation. As noted previously, the reference input voltage on conductor 96 is 90° out of phase (or in some predetermined phase relationship) with the input signal on conductor 52 by operation of conventional phase-shifting network 97.

The alternate switching action of the series-shunt chopper alternately charges capacitance 120 through transistor 116 and discharges capacitance 120 through transistor 118 to ground. In this process the average DC voltage across the capacitance 120 becomes an indication of the phase relationship between the voltages on conductors 52, 96.

The switching of transistor 118 is controlled by a fixed voltage clamp consisting of diode 124. On the other hand

transistor 116 is controlled by a variable voltage clamp composed of diode 126 and transistor 128. A variable voltage clamp is necessary owing to the presence of the varying output signal through transistor 116 and across capacitance 120 which is directly connected thereto.

The average DC voltage across capacitance 120 is applied to an amplifying and level-shifting circuit 127 to supply a usable output signal on conductor 54. In addition, any tendency of the capacitance voltage to deviate as a result of a similar deviation in phase of the input signal on conductor 54 is translated into changes in feedback voltage on conductor 56 which oppose such phase deviations and maintain the bridge 34 in dynamic balance. The feedback voltage is connected to the control electrode (FIG. 2) of the transistor 76 which constitutes the impedance 32 of the bridge 34. The transistor 76, therefore, acts as a variable impedance to maintain the bridge 34 in balance.

Thus, the phase relationship between the voltages on conductors 52 and 96 is substantially maintained so that the small phase deviations and related deviations of capacitance 120 voltage are a measure of the changes in impedance Z_i . By maintaining the phase relationship between the voltage on conductors 52 and 96 substantially at 90°, the sensitivity of the detecting circuit 47 is maximized at a phase relationship corresponding, for example to the phase of the real R_i component of thoracic core impedance Z_i and insensitive to the component of impedance Z_i at right angles to it, the reactive component X_i in this case, as explained above in the description of FIG. 3.

A pair of standard resistances 130, 132 are connected in the bridge 34 (FIG. 2) momentarily by closing switch 134 and opening switch 136 respectively. Resistance 132 is very low compared to impedance 130. Resistances 130, 132 can be individually switched to parallel and series respectively with the subject. As the measured impedance is relative only changes therein are normally indicated. Momentary interjection of the known resistances 130, 132 into the impedance-measuring circuit therefore yields absolute bases of comparing subject impedance and conductivity respectively at any given point along the output curve (such as the curves 88, 92 of FIG. 5) of the impedance-measuring circuit. Alternatively, the operation of the switches 134, 136 can be reversed with the resistances 130, 132 being then parallel in the circuit, so that reference values can be established by momentarily shorting resistances out of the measuring circuit. Thus, the resistances 130, 132 relate lung conductivity, in the specific application of our apparatus described herein, to lung resistance for the subject being studied. Once the feedback from the amplifying and matching circuit 126 is thus adjusted, the bridge 34 is subsequently maintained in dynamic balance by the phase-related detector 47 and the variable bridge impedance 32.

The improvements in pneumographic measurements made possible by our apparatus are illustrated in FIG. 4. Arrow 78 represents a 27 percent change in a measured thoracic impedance level of about 1,000 ohms utilizing the electrode assembly shown in FIG. 4B. In the latter arrangement, the guard ring electrode 20' has a width (dimensional arrow 80) equal to approximately twice the peripheral wall thickness of the thoracic region. A similar percentage change (arrow 82 FIG. 4) at a lower measured impedance level of about 600 ohms was obtained with the electrode structure of FIG. 4a. In the latter structure the width 84 of the guard ring 20 is about equal to the thickness of the thoracic wall 12 (FIG. 1). In contrast, arrow 86 (FIG. 4) denotes the usual and unreliable change of about 4 percent measured by conventional, two-electrode impedance pneumographs during the respirational cycle, where no guard ring is used.

The electrodes 20 or 20' and 22 of FIGS. 4A and 4B are made from flexible metallic foil or sheet. Tin is suitable for this purpose and the electrodes are usually applied with an interposed layer of conductive grease. The guard electrode 20 need not be circular or even continuous. As can be seen in FIG. 4C, the guard electrode can be segmented or provided as

discrete but electrically interconnected electrodes 21. Although four such electrodes are shown in equally spaced array, obviously a different number or arrangement can be used, depending on the application of the invention.

In FIG. 5 curve 88 is indicative of variations in phase detector output on conductor 90 (FIGS. 2, 8), which in turn indicates respiratory thoracic impedance changes of a walking subject. Curve 92 of FIG. 5 similarly represents thoracic impedance changes in a subject who is nearly asleep. The sensitivity of an output signal recorder 98 can be adjusted by trimmer-resistance 94 (FIG. 8).

It will be understood that the bridge and detecting circuit 26 can be connected to indicate impedance between another pair of the electrodes 20-24. Thus, the impedance Z_R between the guard ring and the primary electrode 24 or 22, respectively can be similarly and exclusively measured or monitored.

From the foregoing it will be apparent that novel and efficient forms of physiological impedance-measuring apparatus have been disclosed herein. Although our invention is adaptable for measuring various parts of the anatomy, it will be described primarily in connection with the thoracic regions of the body, as the lungs are most sensitive in their electrical conduction processes and are most symptomatic of body conditions. Thus, such measurements will yield information which is highly desirable in a large number of diagnostic situations. While we have shown and described certain presently preferred embodiments of the invention and have illustrated certain presently preferred methods of practicing the same, it is to be distinctly understood that the invention is not limited thereto but may be otherwise variously embodied and practiced within the scope of the following claims.

We claim:

1. In a physiological measuring apparatus, the combination comprising a pair of electrodes adapted for spaced application to predetermined locations of the anatomy, an additional electrode spacedly juxtaposed to one of said first-mentioned electrodes, circuit means for applying an impedance-measuring voltage only across said first-mentioned electrodes and a portion of said anatomy when juxtaposed therebetween and for applying a potential to said additional electrode similar to that on the associated one of said first-mentioned electrodes, and an impedance-measuring circuit coupled to said electrodes and to said circuit means for measuring anatomical impedance only between a predetermined pair of said electrodes, said circuit comprising a substantially unity gain amplifier for applying a substantially identical potential at lower impedance to said additional electrode for driving the same.
2. The combination according to claim 1 wherein said additional electrode is a guard ring spacedly surrounding said one electrode.
3. The combination according to claim 1 wherein said additional electrode includes a plurality of electrically interconnected electrode segments.
4. The combination according to claim 1 wherein said impedance-measuring circuit includes a phase-discriminating circuit sensitive only to a predetermined phase of impedance changes measured between said electrodes.
5. The combination according to claim 1 wherein said amplifier is coupled to said additional electrode and to said one electrode.
6. The combination according to claim 1 wherein said impedance-measuring circuit includes a phase-sensitive detector circuit including circuit means for relating the maximum sensitivity of said detector circuit to the phase of a predetermined component of impedance changes between said first-mentioned electrodes.
7. The combination according to claim 6 wherein said impedance-measuring circuit includes an impedance bridge having a fixed resistive and capacitive impedance and a variable voltage-sensitive impedance, said last-mentioned impedance being coupled in feedback relation to the output of said detector circuit.

8. The combination according to claim 7 wherein said detector circuit includes an amplifying and matching circuit for relating said feedback to said impedance change component to maintain said bridge in dynamic balance.

9. The combination according to claim 1 wherein dual impedance-measuring paths are coupled to said circuit means and to said impedance-measuring circuit, known impedance means are in one of said paths, said anatomical impedance is in the other of said paths, and means form part of said impedance-measuring circuit for alternately connecting said paths to provide relative anatomical impedance measurements and an absolute impedance value for comparison purposes.

10. The combination according to claim 9 wherein said paths include known impedances in series and parallel respectively with said anatomical impedance, and said connecting means include shorting circuit means around each of said impedances for selectively coupling and decoupling said impedances so that series and parallel value bases can be alternately and selectively indicated at any point along an output curve of said impedance-measuring circuit.

11. The combination according to claim 10 wherein said series impedances are relatively small and said parallel impedance is relatively large so that absolute value bases of impedance and conductances can be indicated.

12. The combination according to claim 1 wherein said amplifier is provided with a power supply which is independent of

said impedance-measuring voltage.

13. The combination according to claim 1 wherein said additional electrode is a guard ring spacedly surrounding said one electrode.

14. In a physiological measuring apparatus, the combination comprising a pair of electrodes adapted for spaced application to predetermined locations of the anatomy, an additional electrode spacedly juxtaposed to one of said first-mentioned electrodes, circuit means for applying an impedance-measuring voltage only across said first-mentioned electrodes and a portion of said anatomy when juxtaposed therebetween and for applying a potential to said additional electrode similar to that on the associated one of said first-mentioned electrodes, and an impedance-measuring circuit coupled to said electrodes and to said circuit means for measuring anatomical impedance only between a predetermined pair of said electrodes, said impedance-measuring circuit including an impedance bridge having fixed and variable impedances coupled to said first-mentioned electrodes, said variable impedance being coupled in feedback relation to an output circuit of said bridge for dynamically balancing said bridge.

15. The combination according to claim 14 wherein said additional electrode is a guard ring spacedly surrounding said one electrode and having a width at least equal to about the thickness of the thoracic wall structure being measured.

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