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[54] **IMPEDANCE PLETHYSMOGRAPH HAVING BLOCKING SYSTEM**

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[52] **U.S. Cl.** ..... **128/2.1 Z,** 128/2.05 V

[51] **Int. Cl.** **A61b 5/02**

[58] **Field of Search** ..... 128/2.06 A, 2.06 B, 2.06 G, 128/2.06 R, 2.1 RB, 2.1 Z, 2.05 R

[56] **References Cited**

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Kubicek et al., "NASA Tech. Brief," No. 68–10220, June 1968, 2 pp.

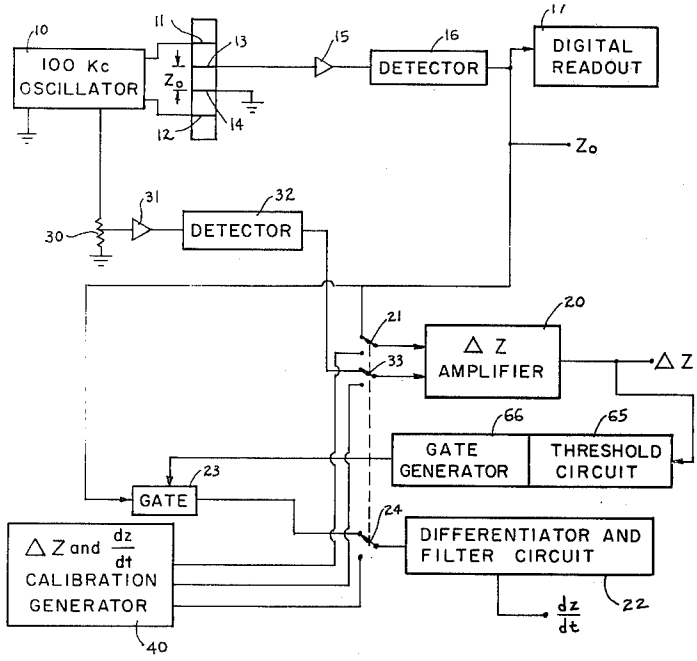
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[57] **ABSTRACT**

An impedance plethysmograph has an oscillator for applying constant current to the thorax of a patient. A detector circuit connected to the patient provides an output corresponding to thoracic impedance. Circuits responsive to detected impedance provide signals corresponding to change of impedance and the first time differential of impedance. Means are provided to block the differential signal when it is meaningless, for example, due to respiration.

2 Claims, 4 Drawing Figures



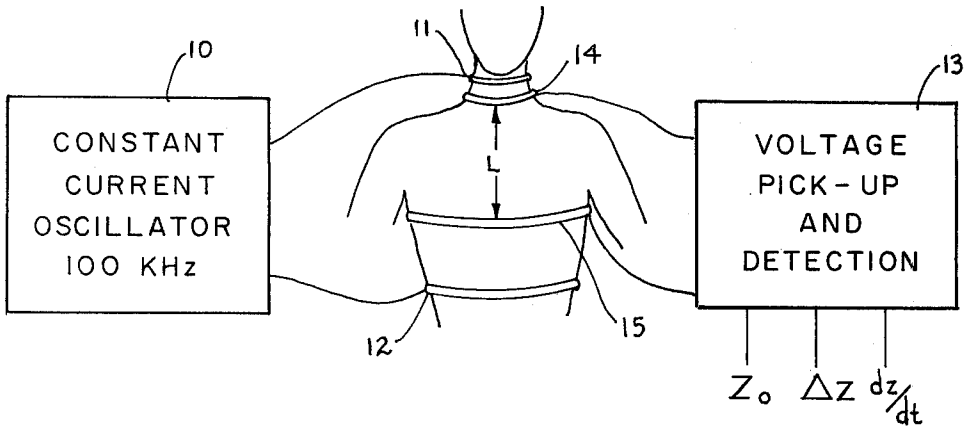

10
CONSTANT CURRENT OSCILLATOR 100 KHz
11
14
L
12
15
13
VOLTAGE PICK-UP AND DETECTION
Z_o ΔZ $\frac{dz}{dt}$

Fig. 1

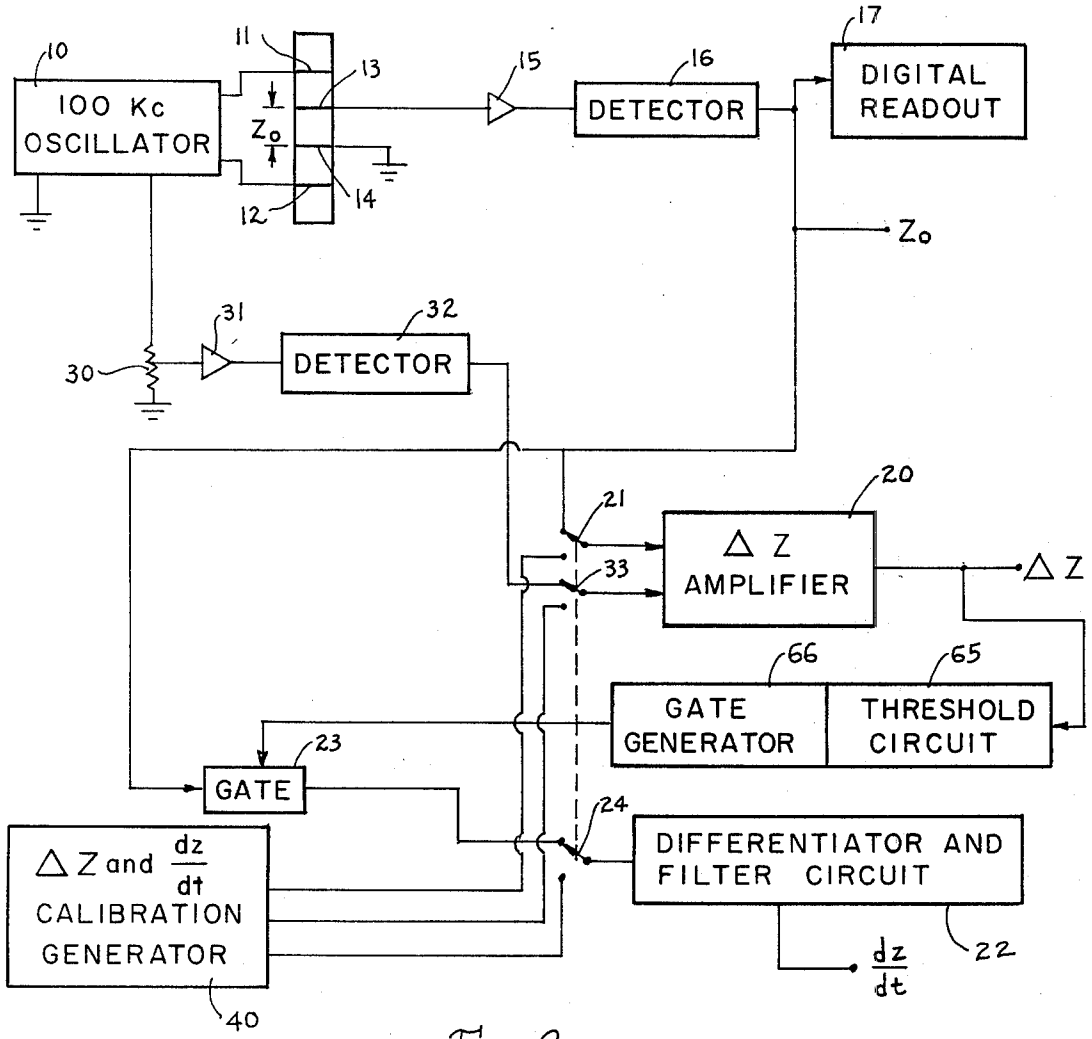

10
100 Kc OSCILLATOR
11
13
Z_o
12
14
15
16
DETECTOR
17
DIGITAL READOUT
Z_o
30
31
32
DETECTOR
20
21
33
Δ Z AMPLIFIER
Δ Z
66
65
GATE GENERATOR
THRESHOLD CIRCUIT
23
GATE
24
DIFFERENTIATOR AND FILTER CIRCUIT
ΔZ and $\frac{dz}{dt}$ CALIBRATION GENERATOR
40
$\frac{dz}{dt}$
22

Fig. 2

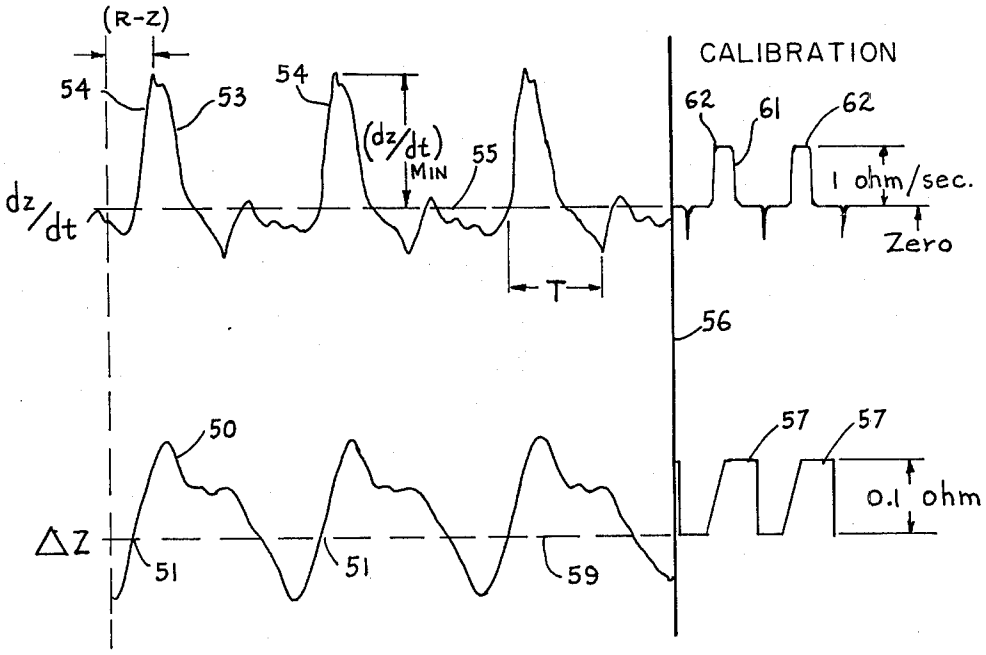

(R-Z)
54
53
54
(dz/dt) MIN
55
CALIBRATION
62
61
62
1 ohm/sec.
Zero
dz/dt
T
56
50
57
57
0.1 ohm
ΔZ
51
51
59

Fig. 3

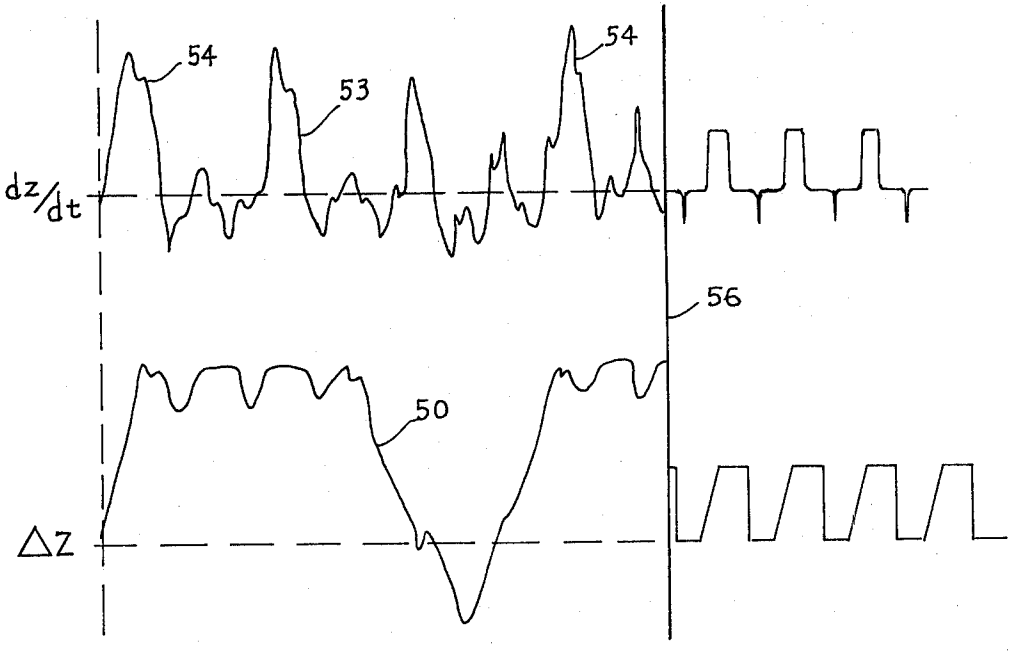

54
53
54
dz/dt
56
50
ΔZ

Fig. 4

IMPEDANCE PLETHYSMOGRAPH HAVING BLOCKING SYSTEM

The present invention relates to plethysmogragphs and particularly to an impedance plethysmograph and process of using the same. The invention is particularly useful in determining cardiac output.

U.S. Pat. No. 3,340,867 discloses a plethysmograph of the type to which the present invention is directed. In accordance with this patent, a current flux is distributed in the mammalian thorax by the placement of electrodes at the neck and lower thorax. The electrodes have the form of bands positioned to encircle the neck and thorax. To these electrodes is applied a fluctuating excitation current. The impedance of the thorax is then measured with either the same electrodes or with different electrodes to obtain information concerning cardiac activity and particularly cardiac output. In accordance with this patent, the major portion of excitation current flux is passed through the lung tissues, rather than through the lower resistivity volume of the major thoracic arteries veins, and the heart. Accordingly, it is possible to measure the blood volume changes of the lungs and derive cardiac output from the impedance changes. Impedance plethysmographic waveforms obtained between the electrodes monitor the pulmonary flow as reflected by impedance changes in the pulmonary vascular bed.

In the system of this patent, the measurement of electrical impedance changes in the thorax during application of a fluctuating current (such as 100kc. current having RMS value of 5.0 ma.) between electrodes placed on the surface of the thorax. This procedure provides the advantage of minimum subject preparation and constraint.

To verify the determination of cardiac output from impedance measurements, an understanding of the physiological phenomena responsible for the measured variations must be obtained. In particular the excitation current flux paths in the thorax between the excitation electrodes must be established.

It has been found that the only tenable theoretical flux distribution path is that in which the majority of the current flux passes from the band electrodes into the lung volume and tends to avoid the heart blood volume regions. Consequently, this current flux distribution implies that the measured impedance is an indication of the total movement of blood in the pulmonary vascular bed.

The impedance waveforms obtained with two band electrodes, one positioned at the neck and the other about 2 cm. below the xiphisternal joint reflect the pulmonary blood pulsations in the lungs rather than the direct ventricular volume change. Equipotential surfaces sketched from thorax potential measurements indicated a flow of current from the blood volume regions. The largest density of the current leaving a band electrode positioned at the midsection of the thorax is found at the base of the lungs on the posterior thorax. These results indicate that in an electrode configuration including one band electrode about the neck and another about the thorax, the majority of the current flux is passed through the lungs, so that the major observed impedance characteristics are controlled by the pulsating pulmonary volume changes. The relatively accurate cardiac output determination made by these impedance measurements are based on an indirect indication of right ventricular stroke volume as reflected by the pulmonary vascular bed.

In order to minimize the influence of respiratory and other effects in the use of equipment of the above type, the first derivative of the change in impedance of the thorax during the cardiac cycle is generally used for computation. Thus, the normal respiration of a patient produces periodic variation in the impedance, and the computation on the basis of the first derivative of the impedance can minimize the effect of such variation. Even when the first time derivative of the impedance is employed, however, it has been found that the best results are obtained when the subject is instructed to hold his breath during the time that the measurements are taken. For example, the patient may be required to hold his breath for about six heart beats in order to avoid drift in the measurement of the change of impedance and the first time differential of the impedance recording. This technique of course requires that the patient be conscious and otherwise capable of following instructions to hold his breath during the measuring period. If the patient is not capable of holding his breath, then great skill must be employed in the interpretation of the recordings in order to provide a reasonable determination of the meaning of the recordings. In the event of erratic variation in the base impedance, the interpretation of the recordings may be thus exceedingly difficult if not impossible. It may be very difficult to determine which portions of a recording produce valid results, and which portions must be disregarded as meaningless.

In accordance with the present invention, this problem is overcome by providing a method and apparatus for blocking out the reading of an instrument of this type when the reading would be rendered meaningless, for example, as a result of respiration.

The invention will now be more fully described with reference to the accompanying drawings, in which:

FIG. **1** is a block diagram, illustrating the basic elements of an impedance plethysmograph showing the interconnections thereof to a human subject in use;

FIG. **2** is a block diagram of an impedance plethysmograph in accordance with the invention;

FIG. **3** is a diagram showing a portion of a record of impedance variation and the first time differential of the impedance for a normal subject, in the use of an impedance plethysmograph; and

FIG. **4** is a diagram representing portion of a strip chart record showing the impedance change and first time differential of the impedance of a patient in which the variation of the base impedance interferes with the interpretation of the recording.

Referring now to the drawings, and more in particular to FIG. **1**, a basis system or an impedance plethysmograph is comprised of a constant current oscillator **10** providing, for example, a constant sinusoidal alternating current of 4 milliamperes RMS, 100 KHz. The output of the constant current generator **10** is connected between an electrode **11** at the upper portion of the neck of the patient, and an electrode **12** at the lower abdomen. The circuit also includes a voltage pick up and detection circuit **13**, the input of which is connected between an electrode **14** at the base of the neck of the patient, and an electrode **15** slightly below the xiphisternal joint of the patient. The electrodes, for example, may encircle the patient's body.

The constant current from the oscillator **10** is passed longitudinally through the thorax of the patient between the electrodes **11** and **12.** The product of this current multiplied by the thoracic impedance generates a voltage $E=IZ_o$ between the electrodes **14** and **15.** This voltage is picked up by the voltage pick up and detection circuit **13,** which employs a high input impedance linear amplifier in its input stage. The circuit **13** also includes detection, balancing, and calibration circuits, and provides the outputs Z_o, ΔZ, dz/dt. z_o is the base impedance of the thorax, and provides a direct numerical measurement of fluid changes in the chest. ΔZ represents the gross impedance change during the cardiac cycle, and is a parameter which may be employed in the measurement of peripheral circulation. The time derivative dz/dt is the first derivative of ΔZ, and is the parameter which is employed in the measurement of stroke volume and the cardiac output.

Referring now to FIG. **2,** therein is illustrated a block diagram of a circuit in accordance with the invention. The circuit is comprised of a constant current sinusoidal oscillator **10,** of conventional nature, which is connected to the electrodes **11** and **12** as above-described. The base impedance Z_o of the thorax appears between the electrodes **13** and **14,** as above-described. The electrode **14** may be connected to a ground reference, and the electrode **13** is connected a high input impedance AC amplifier **15.** The output of the amplifier **15** is connected to a detector **16.** The output of detector **16** is connected to a digital read-out circuit **17** which, after suitable filtering provides a digital read-out corresponding to the base impedance Z_o. The circuit **17,** for example, may comprise filtering circuits for filtering the DC output of the detector **16,** followed by an analog to digital converter and a digital indicator. In a normal subject, the base level indicated by the digital read-out circuit **17** will be about 25 ohms. The output of the detector **16** may also be applied to a terminal Z_o for external use.

The output of the detector **16** is also applied to an amplifier **20** by way of a switch **21,** and to a differentiator and filter circuit **22** by way of a gate circuit **23** and switch **24.** The output of the amplifier **20** is applied to a terminal ΔZ, to produce an output corresponding to the change in the impedance Z_o from its base level.

As one example, the amplifier **20** may be comprised of a differential amplifier, the second input of which is derived from the oscillator **10** by way of a potentiometer **30,** an amplifier **31,** a detector **32,** and a switch **33.** With this arrangement, the potentiometer **30** may be adjusted to an impedance corresponding to the impedance of the thorax at the base level, so that the output of the amplifier **20** reflects changes in the thoracic impedance from this level. A system of this general type is disclosed in U.S. Pat. No. 3,340,867, for example. Alternatively, the amplifier **20** may include a sample and hold circuit having an automatic re-set, for providing the desired output corresponding to the difference impedance.

The differentiator and filter circuit **22** is of conventional nature, and provides an output at the terminal dz/dt corresponding to the first time differential of the base impedance.

In addition, the circuit of FIG. **2** includes a calibration generator **40** for generating calibration signals for the ΔZ and dz/dt outputs. The nature of the calibration signal will be more fully disclosed in the following paragraphs. One output of the calibration generator **40** is applied to the amplifier **20** by way of the switches **21** and **33,** and another output of the calibration generator is applied to the differentiator and filter circuit **22** by way of the switch **24.**

The above-described circuit, with the exception of the gate **23,** is conventional, and has been employed in the past for the measurement of thoracic impedance.

FIG. **3** illustrates a portion of a strip chart recording of the outputs dz/dt and ΔZ from the arrangement of FIG. **2** with a normal subject. Trace **50,** corresponding to the change of impedance Z_o from its base level, is shown with a constant base line, i.e., without the influence of respiratory or other effects. The vertically extending portions **51** of this curve correspond to the maximum decreasing impedance slope during systole. The pulmonary flow rate is a function of the rate of change of impedance, and hence of the slope of the portions **51** of the curve **50.** This slope is determined by differentiating the impedance Z_o, for example in the circuit **22** of FIG. **2,** to provide the wave form **53.** The amplitude of the peaks **54** from a calibration base line **55** thus corresponds to the pulmonary flow rate.

The curves **50** and **53** are typical curves obtained in operation of the instrument, and these curves are shown to the left of a vertical line **56,** representing an arbitrary time at which the switches **21, 24,** and **33** were switched from their operating position, as shown in FIG. **2,** to the output of the calibration generator **40.** As illustrated to the right of the line **56,** the calibration signal for the ΔZ amplifier, which will be generally unchanged in shape from the output thereof, consists of a series of pulses **57** having leading edges with calibrated slopes. Since the normal variation of impedance is around 0.1 ohm, the pulses **57** have amplitudes corresponding to such value. The calibration signal establishes a base line **59** corresponding to the base impedance. Pulses similar to the pulses **57** are applied to the differentiator and filter circuit **42,** whereby the output of this circuit has a wave form **61** with pulses **62** having amplitudes corresponding to the slope of the leading edges of the pulses **57.** For example, the amplitudes of the pulses **62** may correspond to an impedance change rate of one ohm per second. The curve **61** also establishes the base line **55** for the curve **53.**

As above stated, in the normal process of taking measurements with an impedance plethysmograph, the patient is instructed to hold his breath, for example for six heart beats, in order that variations of the base impedance, for example, due to respiration, do not interfere with the output of the device. In the presence of respiration, the curve **50** corresponding to the change of impedance is superimposed on a cyclically varying base which may correspond to impedance changes greater than those resulting from pulmonary flow. As a consequence, the slope of the impedance change during each systole will not correspond exactly to the pulmonary flow rate, and the amplitudes of the pulses **54** will consequently not accurately correspond to the flow rate. While this effect may be minimized by having the patient hold his breath, some interpretation may still be necessary by the operator to ascertain which of the pulses **54** provide an accurate indication of the flow rate. While this technique is of course possible when a patient is conscious and able to understand and follow instructions to hold his breath, it is of course impossible when the patient is not capable of holding his breath in

this manner. In such cases, in the past, it has been necessary to employ great skill in interpretation of the recording in order to ascertain the flow rate. In some cases, particularly when the variations in the base line were erratic, it has not been possible to ascertain the pulmonary flow rate with any degree of accuracy. A portion of a strip chart recording of an abnormal patient is illustrated in FIG. **4.** In this figure, the curve **50** corresponding to the change of impedance has been superimposed upon a base line of unknown or erratic variation, so that it is difficult to visually pick out the portions of the curve corresponding to a systole. The curve **53,** due to the variations in slopes of the curve **50** resulting from the variation in the base line, therefore exhibits peaks **54** of widely differing amplitudes. The selection of the peaks **54** corresponding to the true pulmonary flow rate is consequently difficult.

In order to overcome this problem, in accordance with the invention, means are provided for blocking the presentation of a recording corresponding to the *dz/dt* signal whenever this signal has a meaningless value. Specifically, the signal is difficult to interpret whenever the base line upon which the difference impedance ΔZ is superimposed exceeds a given value. Therefore, in accordance with the invention, referring to FIG. **2,** a threshold circuit **65** is provided connected to the output of the amplifier **20,** for providing an output signal whenever the output thereof exceeds a given value. The circuit **65,** may for example, be Schmitt trigger circuit. Since it is desirable to block out the *dz/dt* signal for the entire stroke, the output of the threshold circuit **65** may be applied to a gate generator circuit **66,** for example monostable multivibrator. The output of the gate generator **66** is applied as a control signal to the gate **23,** to thereby block the gate **23** upon the occurrence of a signal input to the threshold circuit exceeding the determined value. As a consequence, whenever the base line has fluctuated to an extent such that the output of the differentiator and filter circuit **22** would be meaningless or difficult to interpret, this output is blocked by the gate **23,** and hence the output appears merely as a straight line. The operator is then presented only with a recording that is sufficiently accurate to be subject to correct interpretation without difficulty.

While the invention has been disclosed and described with reference to a single embodiment, it will be apparent that many variations and modifications may be made therein within the scope of the invention, and it is therefore intended in the following claims to cover each such variation and modification as falls within the true spirit and scope of the invention.

What is claimed is:

1. In a plethysmograph of the type including means for measuring an impedance in a mammalian subject, means providing a first output corresponding to impedance changes of said measured impedance and means providing a second output corresponding to a time differential of said measured impedance, the improvement comprising a means responsive to said first output connected to inhibit said second output when said first output exceeds a determined magnitude.

2. In a plethysmograph for measuring cardiac output including a source of alternating constant current, electrode means for applying said constant current to a mammalian subject, amplifier and detector means, further electrode means for connecting said amplifier and detector means to the thorax region of said subject for producing an impedance signal corresponding to thoracic impedance, means responsive to said impedance signal for producing a first signal corresponding to the change of said thoracic impedance, and means responsive to said impedance signal for producing a second signal corresponding to the first time differential of said impedance; the improvement comprising means responsive to the amplitude of said first signal for blocking said second signal.

* * * * *