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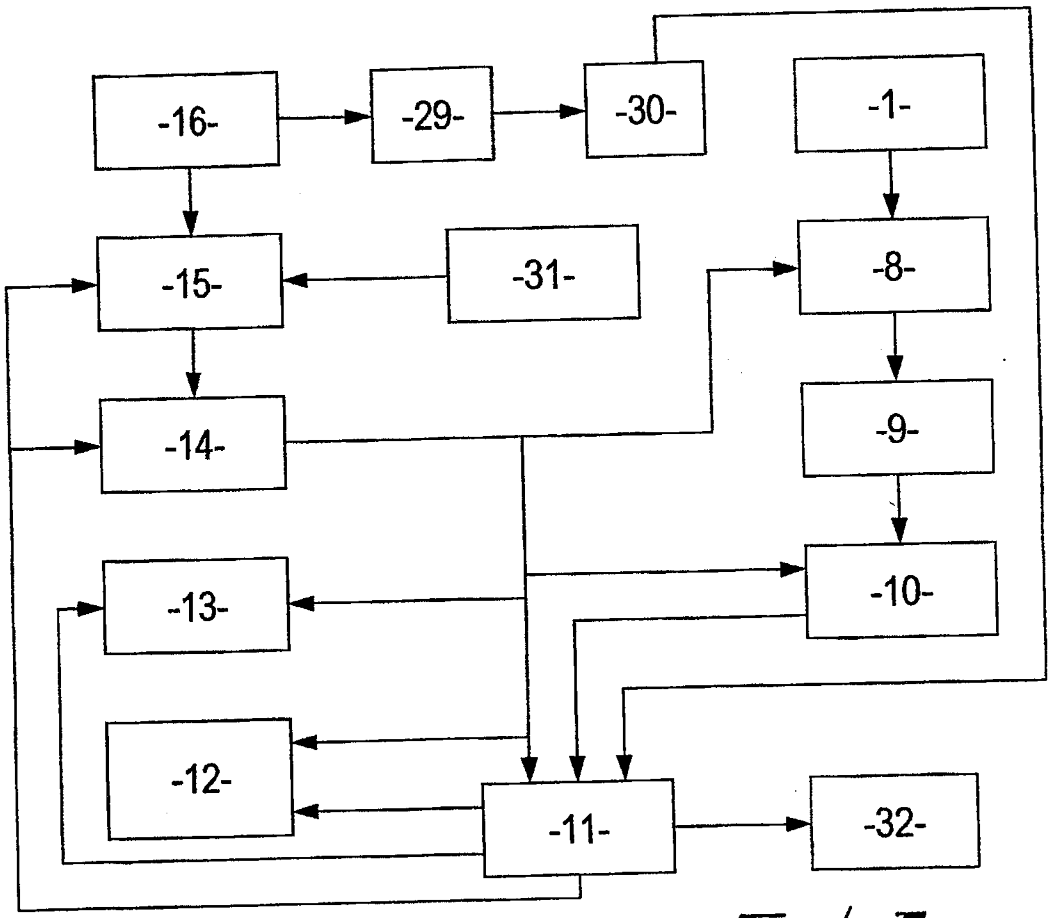

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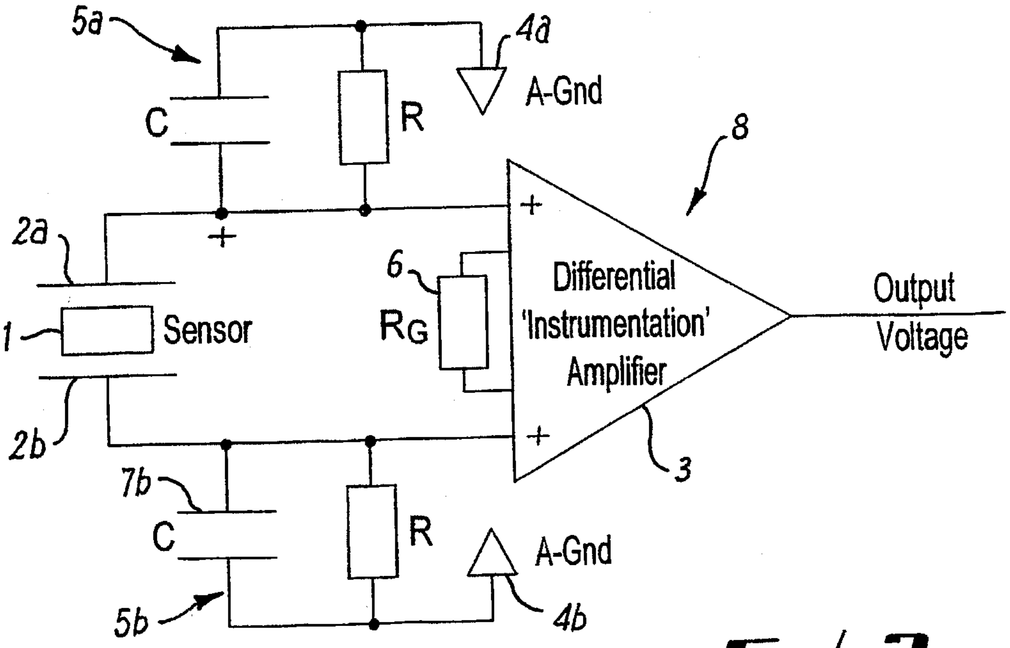
FIG. 1



5a
4a
A-Gnd
C
R
8
2a
+
+
6
Differential
'Instrumentation'
Amplifier
Output
Voltage
1
Sensor
RG
2b
+
3
7b
C
R
A-Gnd
5b
4b

FIG. 2

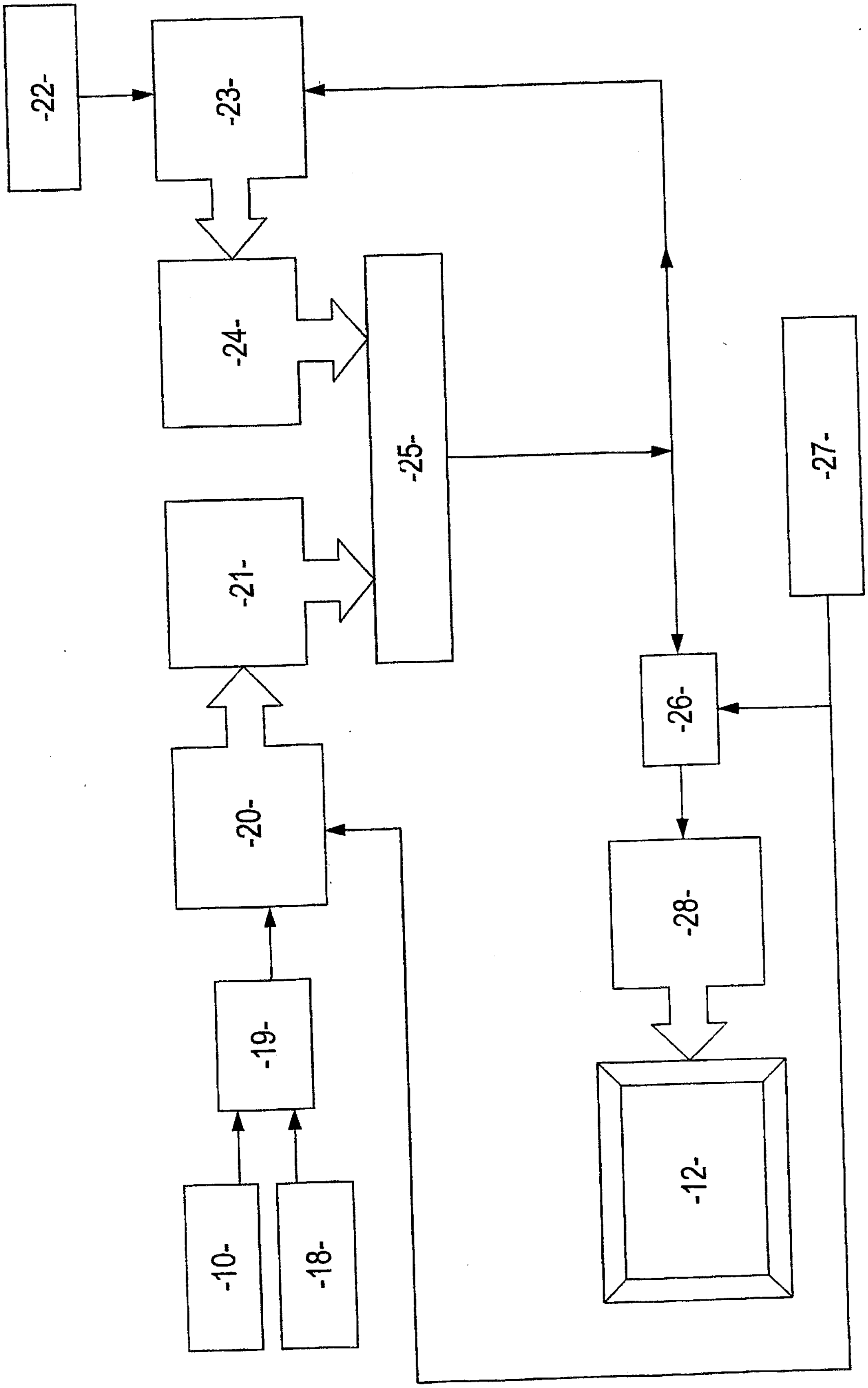


Fig. 3

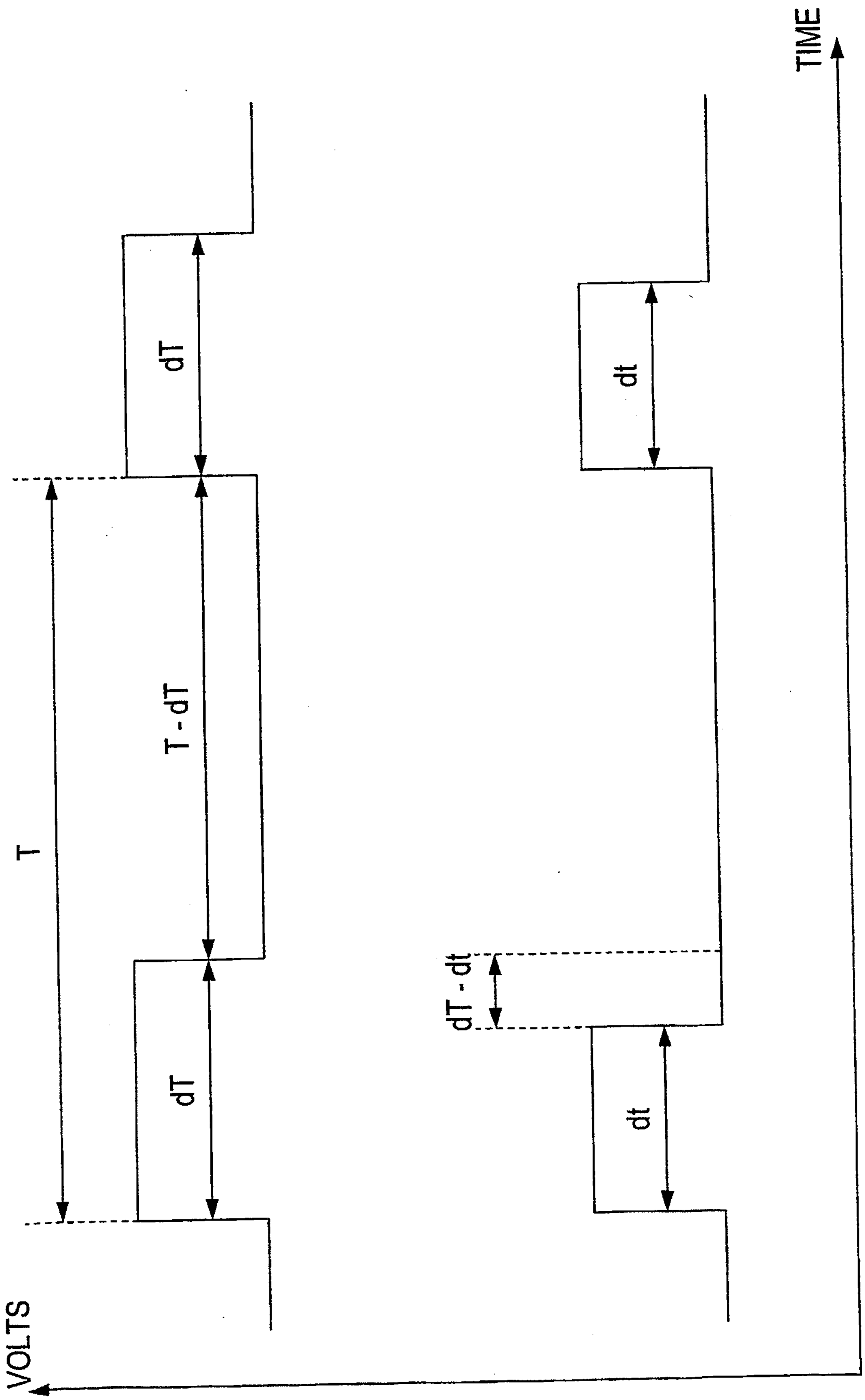


FIG. 4

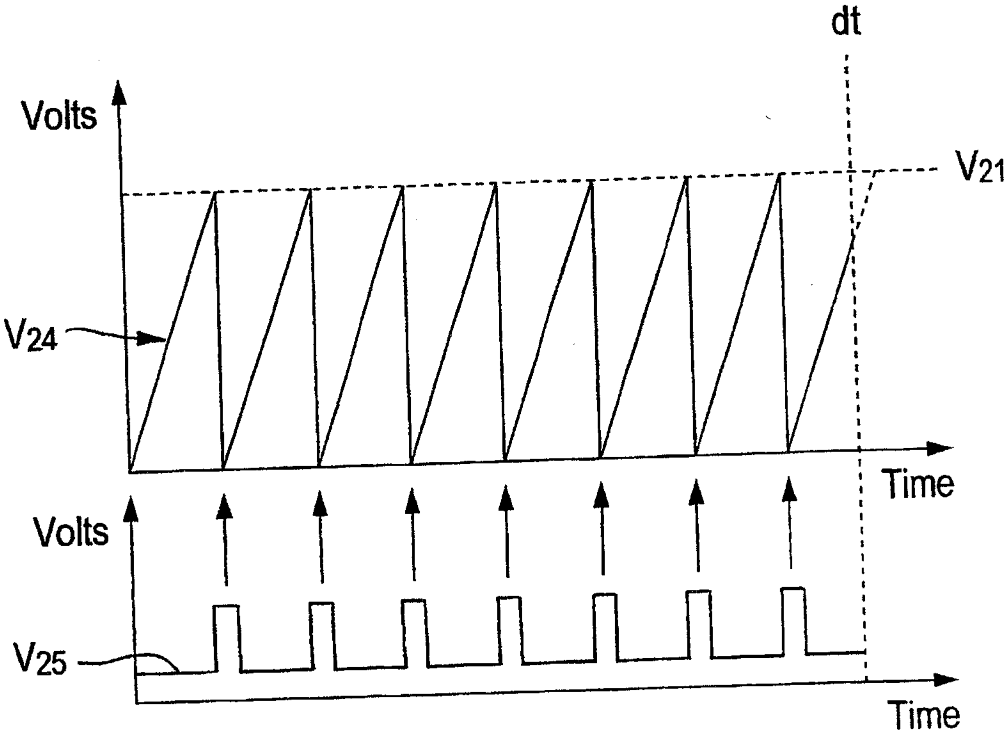


FIG. 5

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A RESPIRATORY MONITOR

This invention relates to a respiratory monitor.

A fundamental physiological parameter is the breathing rate of a subject, whether human or animal which needs to be recorded. A number of devices have been designed to perform this operation.

As many medical treatments, including oxygen therapy, nebuliser treatment and active humidification degrade the performance of the technique used to sense breathing, it is important to have as sensitive a monitor as possible. This not only provides more accurate information on the subject's condition but additionally prevents the monitor generating false alarms.

Previous devices have concentrated mainly on improving the technique used to sense breathing and find more sensitive materials to manufacture sensors from. A favoured method of sensing breathing is to use a temperature sensitive sensor positioned so as to be impinged upon by the subject's respiratory airflow. A sensor of this type manufactured from a particularly sensitive material is described in US Patent No. 5,311, 875. This discloses a sensor formed from a thin film of poly vinylidene fluoride (PVDF) provided with electronics for interpreting the signals produced by the PVDF sensor. However even with sensors such as these more sensitivity is required when dealing with the very ill or with infants, who being smaller have a very much smaller respiratory airflow which is thereby more difficult to detect.

It is therefore an object of the present invention to provide a more sensitive respiratory monitor.

According to the present invention there is provided a respiratory monitor comprising a sensor operative to produce a signal representative of a subjects breathing, a differential amplifier for amplifying the signal, the amplifier being operative to produce an output proportional to the difference between two inputs thereto, means for converting the output to indicate the subjects breathing rate and means for displaying the breathing rate, the inputs of the differential amplifier being connected to analogue circuit common ground by a filter comprising a capacitor and a resistor in parallel.

In a preferred embodiment, the sensor is preferably a pyroelectric sensor, advantageously positioned so as to be impinged upon by the subjects respiratory airflow of a subject and thereby provide information about the regularity and frequency of a subject's breathing. The sensor may be formed from a lamina of poly vinylidene fluoride. An electrical potential difference is generated between the two faces of the lamina corresponding to a temperature difference caused by the respiratory airflow of the subject.

The inputs of the differential amplifier are each connected to opposite faces of the lamina forming the sensor via electrodes provided on the faces of the lamina. Suitable electrodes may be formed by screen printing or by deposition. Means may also be provided for varying the gain of the differential amplifier.

Typically the differential amplifier may be located at a distance from the sensor and the electrodes are then connected to the differential amplifier by any suitable form of electrically conducting cable.

In order to minimise noise in the amplified signal it is preferred that the cables connecting opposite faces of the sensor to the differential amplifier inputs run in close proximity to each other and preferably in a twisted pair arrangement. This means that any noise pick up is induced in both cables and is thereby removed from the output of the differential amplifier which is dependant upon the voltage difference between the inputs only.

A low pass filter may be provided to filter the output of the amplifier as breathing rates, particularly of the sick, are predominantly comprised of low frequency components.

The means for converting preferably acts to convert a bipolar signal input to it, into a unipolar square wave output. The output of the means for converting is typically high when a breathing event is detected and low otherwise. However if desired, this may be reversed.

Preferably, a microcontroller is provided to calculate an instantaneous breath rate or a number of actual breaths per minute. The microcontroller may distinguish between successive breaths by monitoring the unipolar square wave signal output for successive rising edges of the square wave. Most preferably, a liquid crystal display is provided to display the results of the calculations of an instantaneous breath rate or a number of actual breaths per minute. Which of the calculated quantities is displayed may be controlled by a user interface, preferably a keypad.

Preferably an alarm is provided, the alarm preferably being operative if the instantaneous breath rate drops below a first predetermined value or rises above a second predetermined value, or if no breath has been detected for a particular time period.

The microcontroller may be omitted and means for generating a signal of successive pulses, the pulse rate being proportional to the subjects breathing rate may be provided instead.

The means for generating may comprise a voltage comparator provided for comparing a first analogue voltage with a second analogue voltage and outputting a pulse when the second analogue voltage is equal to the first analogue voltage.

The first analogue voltage is preferably proportional to the period of time that the output of the means for covering was low in the previous cycle. The second analogue voltage is preferably a linear ramp voltage.

Preferably the means for generating comprises a first clock, a first counting unit and a first analogue to digital converter, the output of said first clock being input to said first counting unit, the output of said first counting unit being input to said first analogue to digital converter and the output of said first analogue to digital converter being said first analogue voltage.

The first analogue voltage is preferably fixed at a particular value when the output of the means for converting changes from low to high and is preferably reset whenever the output of the means for converting changes from low to high.

Most preferably the means for generating comprises a second clock, a second counting unit and a second analogue to digital converter, the output of said second clock being input to said second counting unit, the output of said second counting unit being input to said second analogue to digital converter and the output of said second analogue to digital converter being said second analogue voltage.

The second analogue voltage is preferably reset whenever a pulse is output by the voltage comparison.

A third counting unit and a timer are preferably provided, said third counting unit counting pulses generated by the voltage comparator and said timer being operative to reset said third counting unit after a preset time interval. In this manner the subjects breathing rate may be calculated. Most preferably means are provided for adjusting the timer, thus varying the time interval. A liquid crystal, or other suitable, display may be provided, for displaying the output of said third counting unit.

In order that the invention may be more clearly understood two embodiments of the invention will now be described, by way of example, with reference to the accompanying drawings, in which:

Fig. 1 is a block circuit diagram of a respiratory monitor;

Fig. 2 is a circuit diagram of the amplification arrangement part of the monitor of Figure 1;

Fig. 3 is a block circuit diagram of an alternative monitor to that of Figure 1.

Fig. 4 is a comparison between the typical output waveforms of the signal conditioner and the timing means in the alternative embodiment of figure 3; and

Fig. 5 shows typical input and output voltage signals for the voltage comparator in the alternative embodiment of figure 3.

Referring to Figures 1 and 2, the respiratory monitor comprises a sensor 1 formed as a lamina of the piezoelectric/pyroelectric polymer PVDF (poly vinylidene fluoride). When subjected to an impulse force or to a temperature gradient, the material experiences charge separation between the parallel faces of the lamina, which establishes an electric potential difference between the two faces of the laminas. The magnitude of this potential difference increases with either the magnitude of the impulse force or the magnitude and gradient of the temperature step. Of course it is to be understood that other materials with these or similar properties may be substituted for PVDF if required.

The sensor 1 is connected to monitoring electronics by electrodes 2a, 2b, formed typically by deposition or screen printing of a thin metallic film on the lamina surface, each electrode acting to collect surface charge induced by the impulse force or the temperature gradient.

In this invention, mainly the pyroelectric properties of the sensor 1 are utilised. The sensor 1 is placed in the vicinity of the respiratory tidal air flow of the subject. The pyroelectric properties of the sensor 1 create a potential difference across the sensor 1 in response to the temperature differential that is created by the impinging respiratory gases during inspiration and expiration. The polarity of the potential difference is indicative of the mode of respiration, enabling a distinction to be made between inhalation and exhalation, and hence an identification of the transition between these respiratory modes.

The potential difference generated by the sensor 1 is often quite small, particularly if the subject for monitoring is very sick. To allow this potential difference to be monitored it is necessary to amplify it. This amplification is performed by the arrangement 8 in Figure 2.

Sensor 1 is floating electrically relative to the input terminals of a differential amplifier 3. A pair of parallel capacitor-resistor combinations 5a, 5b connects the electrodes 2a, 2b and the inputs of a differential amplifier 3 to the analogue circuit common grounds 4a, 4b.

As charge separation occurs in sensor 1 in response to respiratory activity in the subject complementary potentials are established on electrodes 2a, 2b. The respective capacitors 7a, 7b of the combinations 5a, 5b are thereby charged to voltages of opposite polarity. The high impedance inputs to differential amplifier 3 prevent loading of the capacitor-resistor combinations 5a, 5b. The differential amplifier 3 generates a scaled output voltage proportional to the difference between the instantaneous values of the voltages at its inputs, which in this case are equal to the respective capacitor 7a, 7b voltages. The process of inputting simultaneously derived voltages of opposite polarity to differential amplifier 3 effectively doubles the output voltage that would be achieved if a single sided, ground referenced sensor were to be used.

Additionally, as the amplification arrangement 8 is typically located at a distance from the sensor, it is inevitable that any cable connecting the sensor 1 and the amplifier arrangement 8 will pick up noise, particularly if there is other electrical equipment operating nearby. The use of a differential amplifier 3 in the amplification arrangement 8 allows this problem to be minimised. In an ideal case the output of a differential amplifier is proportional to the voltage difference at its inputs and independent of the overall voltage level and thus there is no change in output if both inputs to the amplifier exhibit identical changes in voltage.

Hence, if the connecting cables from the sensor to the amplifier arrangement run in close proximity to each other or in a twisted pair arrangement, any noise picked up is induced in both cables and consequently is removed from the amplified signal.

The voltage gain of differential amplifier 3 is set by resistor 6, which may be a variable resistor if required.

Figure 1 shows the main component blocks used in the invention to calculate a breathing rate from the signal generated by sensor 1.

The power supply 16 is a 9 volt PP3 battery, but may be provided via an AC/DC mains adaptor. The voltage from power supply 16 is monitored using a potential divider resistor network 29 and a voltage comparator arrangement 30 the output of which is fed to a microcontroller 11 enabling the detection of a low battery voltage, or other power supply problems.

The power supply 16 voltage is also monitored by a voltage regulator 14. The power supply is regulated and inverted to provide a + 5 volts and a – 5 volts supply to the electronics of the device. The voltage regulator incorporates a power down pin controlled by a U-type flip-flop 15. The flip-flop 15 may be triggered to initiate a shut down either by microcontroller 11 or by a manual on/off switch 31.

The signals output by sensor 1 are boosted by the amplifier arrangement 8 of figure 2 and are then fed into a low pass filter 9. The effect of the low pass filter 9 is to exclude noise from the signal to be processed. In particular filter 9 excludes the noise generated by the piezoelectric qualities of sensor 1, whereby vibrations in the vicinity of sensor 1 such as those caused by the subject's breathing generate signals. These signals have a much higher frequency than the breath rate and hence may be excluded by a low pass filter.

The resulting boosted, low frequency signals output by the combination of amplifier arrangement 8 and filter 9 are input to signal conditioner 10. The signal conditioner 10 converts the bi-polar signal input to it into a unipolar square wave output. The output is high when a breathing event is detected and low otherwise and thus the frequency of the square wave signal is directly related to the breathing rate of the subject. The square wave signal output by signal conditioner 10 is input to microcontroller 11 for monitoring.

Microcontroller 11 is programmed to monitor the unipolar signal output by signal controller 10. In addition the microcontroller 11 performs a number of other tasks namely auto power on/off control keypad or other user interface control 13 liquid crystal display 12 control and alarm system 32 control.

Monitoring of the unipolar signal representing breathing is performed on an interrupt basis. The microcontroller 11 monitors the time between successive breaths (successive rising edges on the unipolar signal) and is able to calculate actual breaths per minute or an instantaneous breath rate.

The results of the calculations by microcontroller 11 may be displayed by liquid crystal display 12. The particular calculation result displayed may be chosen by an operator using the keypad 13.

The microcontroller 11 may also monitor the results of the calculations it performs such that if the breath rate drops below a first particular value, rises above a second particular value, or if no breath is detected for a particular time period then the microcontroller 11 may trigger an alarm 32. The alarm 32 is typically some form of buzzer, but may include a flashing light or other visual indication if preferred.

An alternative embodiment for calculating and outputting a breath rate is shown in Figure 3. Equivalent parts to parts of the first embodiment bear the same reference numerals.

Each breathing event results in the signal conditioning unit 10 generating a square wave output pulse V10 of fixed duration, dT shown in figure 4. The leading edges of successive pulses are separated by the inter-breath interval T. The duration of each square wave output pulse, dT, is chosen to be consistent with the required accuracy of the monitor.

During the time interval (T-dT) between successive outputs form the signal conditioner 10, the output of the clock 18, is passed by the gate 19, causing clock pulses to accumulate in the counter 20. The count accumulated during the time interval (T-dT) is transferred and latched into the D-to-A converter 21, where the count is converted to a directly proportional analogue voltage V21.

Consider an initial pulse generated by the signal conditioning unit 10. Throughout a timed fraction, dt of the next pulse duration dT, the analogue output voltage V21 of D-to-A 21 is held constant as a fixed reference voltage, at one input of the voltage comparator 25. The voltage V21 is directly proportional to the preceding inter-breath interval (T-dT).

The output of a free-running clock 22 is applied directly to a permanently enabled counter 23, thereby producing a continuous, step-wise incremented count. The output of counter 23 is applied to the input of a second D-to-A converter 24 which is permanently enabled. In response to the step-wise count, the D-to-A converter 24 outputs a linear ramp voltage of constant slope, determined by the frequency of clock 22. The output voltage of D-to-A converter 24 is applied to the second input of the voltage comparator 25, where it is continuously compared to the output voltage of D-to-A converter 21. When a step-wise increment from counter 23 causes the output of D-to-A converter 24 to exceed the output of D-to-A converter 21, the voltage comparator 25 outputs a suitably conditioned pulse, V25 as shown in Figure 5. The pulse is used to reset counter 23 and is simultaneously applied to gate 26.

The time taken for the output voltage of D-to-A converter 24 to equal the voltage output of D-to-A converter 21 is inversely proportional to (T-dT) and is therefore directly proportional to respiratory rate. The free-running counter 23 will resume counting immediately following the application of the reset pulse and continue in a loop; counting and resetting throughout the period dt.

During the period dt the voltage comparator 25 will output a number of reset pulses to the counter 23 the number being dependant on the frequency of the clock 22. The clock frequently is chosen to achieve the required respiratory-rate resolution, and simultaneously to be a simple multiple of the respiratory-rate.

The timer 27 generates a gating period pulse V27 which defines the timed fraction dt of dT as shown in figure 4. The output pulse from timer 27 is designed to have a leading edge that is synchronous with the leading edge of the pulse V10 is output from the signal conditioning unit 10. The timer 27 enables the gate 26 for a duration dt, thereby applying the pulses generated during dt, at the output of voltage comparator 25 to the input of counter 28. The trailing edge of the timer 27 is used to transfer and latch the count accumulated during dt by counter 28 into the display unit 12. During the time period (dT-dt) following the trailing edge, and prior to initiation of the next dt pulse, the output level of timer 27 is used to reset the counter 20 and the counter 28.

The provision of a means for displaying the subject's present breath rate allows an observer to very quickly tell if there is a problem with the regularity or frequency of the subject's breathing. The observer may then be in a position to take appropriate action to alleviate the problem earlier than might normally be the case. Additionally as an observer may assess the regularity and frequency of a subject's breathing more quickly an observer may safely watch over more subject's than would otherwise be the case.

It is however to be understood that the embodiments described above are by way of example only and that other embodiments are possible without departing from the scope of the invention.

Claims

1. A respiratory monitor comprising a sensor operative to produce a signal representative of a subjects breathing, a differential amplifier for amplifying the signal, the amplifier being operative to produce an output proportional to the difference between two inputs thereto, means for converting the output to indicate the subjects breathing rate and means for displaying the breathing rate, the inputs of the differential amplifier being connected to analogue circuit common ground by a filter comprising a capacitor and a resistor in parallel.

2. A respiratory monitor as claimed in claim 1, wherein the sensor is a pyroelectric sensor.

3. A respiratory monitor as claimed in claim 2, wherein the sensor comprises a lamina of poly vinylidene fluoride.

4. A respiratory monitor as claimed in claim 3, wherein the inputs of the differential amplifier are connected to opposite faces of the lamina via electrodes provided upon the faces of the lamina.

5. A respiratory monitor as claimed in any preceding claim, wherein means are provided for varying the gain of the differential amplifier.

6. A respiratory monitor as claimed in any preceding claim, wherein the differential amplifier is located at a distance from the sensor and the electrodes are connected to the sensor by electrically conductive cables.

7. A respiratory monitor as claimed in claim 6, wherein the cables run from the sensor to the differential amplifier in a twisted pair arrangement.

8. A respiratory monitor as claimed in any preceding claim, wherein a low pass filter is provided for filtering the signals output by the differential amplifier.

9. A respiratory monitor as claimed in any preceding claim, wherein the means for converting acts to convert a bipolar input signal to a unipolar square wave output.

10. A respiratory monitor as claimed in any preceding claim, wherein a microcontroller is provided, to calculate the subjects breathing rate.

11. A respiratory monitor as claimed in claim10, when appendant to claim 9, wherein the microcontroller is connected to the output of the means for converting and is operative to monitor the unipolar square wave signal output of the means for converting successive rising edges.

12. A respiratory monitor as claimed in any preceding claim, wherein a liquid crystal display is provided for displaying the calculated breathing rate.

13. A respiratory monitor as claimed in claim 12, wherein the liquid crystal display is operative to display the number of actual breaths per minute.

14. A respiratory monitor as claimed in claim 13, wherein a user interface is provided to control whether the calculated breathing rate or the number of actual breaths per minute is displayed on the LCD display.

15. A respiratory monitor as claimed in any preceding claim, wherein an alarm is provided.

16. A respiratory monitor as claimed in claim 15, wherein the alarm is operative when the breath rate drops below a first predetermined value.

17. A respiratory monitor as claimed in claim 15 or 16, wherein the alarm is operative when the breath rate rises above a second predetermined value.

18. A respiratory monitor as claimed in any preceding claim wherein the alarm is operative if no breath has been detected for a predetermined time period.

19. A respiratory monitor as claimed in any one of claims 1 to 9, wherein means are provided for generating a signal of successive pulses, the pulse rate being proportional to the subjects breathing rate.

20. A respiratory monitor as claimed in claim 19, wherein the means for generating comprises a voltage comparator provided for comparing a first analogue voltage with a second analogue voltage and outputting a pulse when the second analogue voltage is equal to the first analogue voltage.

21. A respiratory monitor as claimed in claims 19 or 20 wherein the means for generating comprises a first clock, a first counting unit and a first analogue to digital converter, the output of said first clock being input to said first counting unit, the output of said first counting unit being input to said first analogue to digital converter and the output of said first analogue to digital converter being said first analogue voltage.

22. A respiratory monitor as claimed in any of claims 19 to 21, wherein the means for generating comprises a second clock, a second counting unit and a second analogue to digital converter, the output of said second clock being input to said second counting unit, the output of said second counting unit being input to said second analogue to digital converter and the output of said second analogue to digital converter being said second analogue voltage.

23. A respiratory monitor as claimed in any of claims 19 to 22, wherein a third counting unit is provided, said third counting unit counting pulses generated by the voltage comparator.

24. A respiratory monitor as claimed in any of claims 19 to 23, wherein a timer is provided, said timer being operative to reset said third counting unit after a preset time interval.

25. A respiratory monitor as claimed in claim 24, wherein means are provided for adjusting the timer, thus varying the time interval.

26. A respiratory monitor as claimed in any of claims 21 to 25, wherein a liquid crystal display is provided for displaying the output of said third counting unit.