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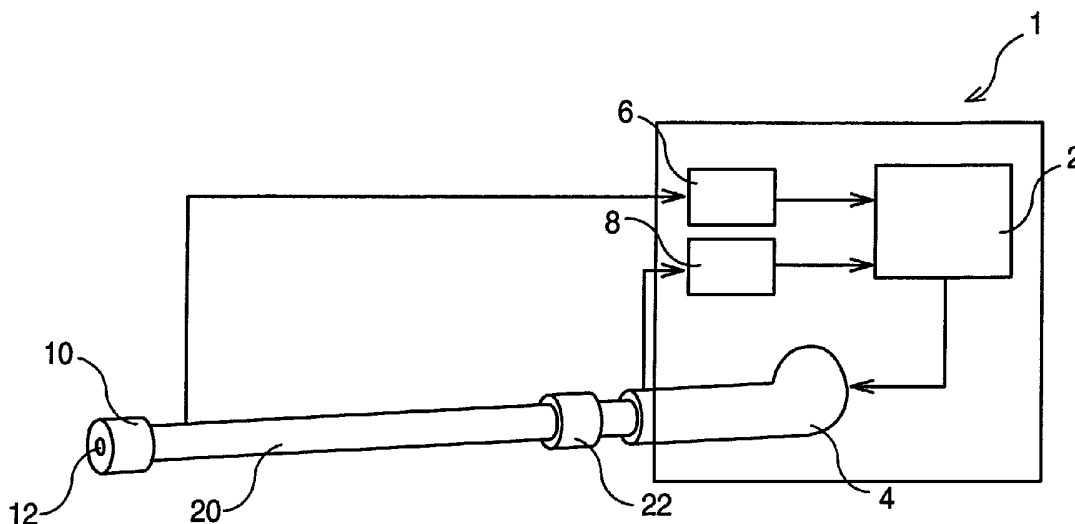
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(54) Title: AN AIR ASSISTANCE APPARATUS FOR COMPUTING THE AIRFLOW PROVIDED BY ONLY MEANS OF PRESSURE SENSORS



(57) Abstract: The invention concerns an apparatus (1) to assist a patient respiration by delivering air to a patient through a mask, said mask being designed to be connected on one first extremity of a tube, said apparatus comprising: - a control unit (2) to adjust the pressure delivered by the blower (4) of said apparatus, - a first pressure sensor (6) for sensing the pressure PM at said first tube extremity and being connected to said control unit, and - a second pressure sensor (8) for sensing the pressure PB at the air output of said blower and being connected to said control unit; in order that, when a tube is connected to said mask and connected to said apparatus on its said second extremity, the air flowing from the apparatus to the mask, said control unit is able to calculate the airflow at said second extremity of the tube from said pressures PM and PB and from the airflow resistance coefficient K<sub>t</sub> of said tube.



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*For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.*

**An air assistance apparatus for computing the airflow  
provided by only means of pressure sensors**

5 Technical field

This invention concerns the field of apparatus to assist a patient respiration and more specifically an apparatus able to calibrate the tubes connected to the patient's mask and to the blower and to determine the airflow of the patient's from  
10 pressure measurement and from the tube calibration.

Background art

In many treatments apparatus are used to provide patients with air. More frequently it is used for patients  
15 with a breathing deficiency caused for example by the weakness of the breathing system or by obstructive apneas during the sleep. In those cases it is important to control the pressure of the air delivered to the patient. With respiratory insufficient patients, apparatus providing air  
20 at a higher pressure help to compensate the weakness of the patients lungs. In the case of patients suffering of sleep apneas, providing the air at a higher pressure removes the obstruction of the upper airways.

In order to provide a correct treatment, it is  
25 required to accurately know the value of airflow the patient is provided with. Usually the apparatus determine the airflow by measuring airflow in the patient circuit (between the blower and the mask) using an airflow sensor. Airflow sensors can be based on high sensitivity  
30 differential pressure sensors measuring the pressure drop across a low resistance or using pitot tubes, or hot wire airflow sensors. Another commonly used mean to evaluate the airflow is by measuring the blower parameters such as speed, current consumption and power.

Summary of the invention

The first object of the invention is to determine the airflow at the patient's mask, without using sensors like airflow sensors that are expensive or motor consumption or rotation sensors that are not accurate especially at low flows.

The invention thus concerns to assist a patient respiration by delivering air to a patient through a mask, said mask being designed to be connected on one first extremity of a tube, said apparatus comprising :

- a control unit to adjust the pressure delivered by the blower of said apparatus,
- a first pressure sensor for sensing the pressure  $P_M$  at the first tube extremity and being connected to the control unit,
- and
- a second pressure sensor for sensing the pressure  $P_B$  at the air output of said blower and being connected to said control unit.

These elements are comprised in the apparatus in order that, when a tube is connected to the mask and connected to said apparatus on its second extremity, the air flowing from the apparatus to the mask, said control unit is able to calculate the airflow at said second extremity of the tube from the pressures  $P_M$  and  $P_B$  and from the airflow resistance coefficient  $K_T$  of the tube

An other implementation of the apparatus according to the invention is to provide an apparatus being able to calculate the coefficient  $K_T$  by using a shell with a traversing hole having a known airflow resistance coefficient  $K_s$ . This enables to use tubes of different sizes, and even tubes with different standards of airflow resistance coefficients.

A further implementation of the apparatus according to the invention is that the pressure control unit comprises an estimation module connected to the means for detecting the patient's breathing parameters, in order that the estimation module is able to determine when the patient is inspiring or

expiring and in response the pressure to apply to the patient's mask, so that the control unit adjust the pressure delivered by the blower.

Further implementations enable to modulate the pressure of the provided air in response to the patient's breathing parameters and events which occur in the patient's breathing.

#### Brief description of figures

The purposes, objects and characteristics of the invention will become more apparent from the following description when taken in conjunction with the accompanying drawings in which:

Figure 1 represents the apparatus according to the present invention when used according to a preferential implementation,

Figures 2 represents the flowchart of the calibration of a tube by the apparatus,

Figure 3 represents a device of the apparatus according to present invention,

Figure 4 represents the electric schema of the device represented in figure 3,

Figure 5 represents an implementation of the apparatus,

Figure 6 represents the way the apparatus reacts to events occurring in patient's breathing, and

Figure 7 represents how the apparatus operates to detect the presence of a patient at the mask.

#### Detailed description of the invention

Ordinary tubes used in air assisting apparatus usually comprise a pressure sensing tube to measure the pressure at the end of the tube. The apparatus according to the present invention is based on this characteristic. As represented on figure 1, the apparatus 1 is connected to the tube 20 by the air inlet and connected to the pressure sensing tube of the tube. The pressure sensing tube is connected to a first pressure sensor 6 comprised in the apparatus. If no sensing

tube is comprised, a manometer will be connected at a first pressure sensor of the apparatus. Because the tube diameter value is much smaller than the tube length value, the pressure drop in the tube is defined by the following equation:

$$\Delta p = K_T \cdot \text{airflow}^2$$

wherein :

$K_T$  represents a constant coefficient characteristic of the tube,

$\Delta p$  represents the difference of pressure between the two tube extremities, and

airflow is the volume of air per time crossing the tube.

The apparatus has sensors that measure the pressure PB at the apparatus air outlet. As the apparatus is sensing pressure on both sides of the tube, knowing the tube coefficient  $k$  is allowing the system to compute the airflow by measuring the pressure drop. The present implementation consists in operating in a mode where precise airflow is required. A shell 10 which is a cap which comprises a small hole 12 at its top is placed to close one extremity of the tube, the other extremity of the tube being connected to the air outlet of the blower 4 of the apparatus 1. This calibration shell 10 has an airflow resistance coefficient  $K_s$  which is a characteristic of it.

Considering that PB being the pressure sensed at the output of the apparatus and PM being the pressure sensed at the calibrated termination,  $S_0$  being known, we have :

$$\text{Airflow} = K_s \cdot \sqrt{PM}$$

$$\text{as } \Delta p = K_T \cdot \text{airflow}^2 \text{ then } K_T = \frac{(PB - PM)}{(K_s^2 \cdot PM)}$$

The apparatus is thus able to determinate the tube coefficient  $K_T$ .

The calibration process takes advantages of having a number of measurements made at different pressures levels and by averaging them can get a more accurate  $K_T$  value.

Example of a 1.8m tubing Ø15mm:

The calibration termination has a *SO* coefficient of 18  
(airflow units are in LPM)

*PB* is measured at 9.90 hPa

5      *PM* is measured at 7.72 hPa

This is giving  $k=0.000871$

Meaning then that the airflow is : 50.02 LPM

When using a new tube, the calibration process is  
10 entered, notably at the request of a clinician or a qualified  
user. The apparatus is expecting the calibration shell to be  
hooked as described in figure 1. The flowchart represented in  
figure 2 is showing an example of a series of measurements  
resulting on the  $K_T$  coefficient calculation, then the tube  
15 coefficient  $K_T$  is recorded and all the upcoming sessions will  
be based on this new tube coefficient. When the tube 20 and  
the shell 10 are installed, the value *J* corresponding to the  
number of one measure is set equal to 1 and associated to a  
number *I* corresponding to the value of the pressure provided  
20 by the blower 4, this *I* number being for example equal to 4  
hPa when the calibration starts. Then the Pulse Width  
Modulation voltage is applied in order to deliver the pressure  
*PM* sensed at the calibrated termination which is the shell 10.  
The pressure *PM* and the pressure *PB* are measured and  
25 associated to the *J* number corresponding to the measure. If *J*  
does not equals the number of measures *N* required to calculate  
the  $K_T$  average, the next measure is taken which means that *J*  
is incremented of 1. The next measure is preferentially made  
for a pressure incremented of 2 hPa, which means that *I* is  
30 incremented of two. When the number of measures required *N* is  
reached, the data associated to each measure are computed and  
enable to associate a tube coefficient  $K_T$  value to each  
corresponding measure *J*. This enables to calculate an average  
of the  $K_T$  value. For example if eight measures are required, *J*  
35 and *I* will be incremented 7 times before calculating the tube  
coefficient *k* value. This also enables to reject the tube if

its standard does not correspond to the kind of tube which can be used with the apparatus 1. The  $K_T$  value will then be used for the airflow computation.

5       The apparatus according to any of the previous claims, wherein the control unit comprises offset compensation means for compensating the possible difference of gauging between the two pressure

Another embodiment of the present invention is  
10   improving the sensitivity of the airflow measurement at low values, by gauging the pressure sensors.

Low flows accuracy is important in air delivery apparatus specially when triggering between inspiration and expiration where sensitivities as low as 5 l/mn are required.

15       A classical 22mm diameter breathing tube will present a pressure drop (PB-PM) of 0.01 hPa with an airflow of 5 l/mn, so it is mandatory to consider a signal amplification after the subtraction.

But this amplification cannot be done without a prior  
20   offset compensation. The apparatus according to the preferential implementation enables to correct this difference of gauging.

As previously described, using two pressure sensors at both ends of a breathing circuit tube, can give an accurate  
25   value of airflow, but due to physical constraints, for low flows accuracy, there is a need for amplification.

Due to the manufacturing process of airflow sensors, most of them are presenting a Voltage vs. Pressure relationship like:

30       
$$V_{out} = V_{offset} + K_{ps} \cdot Pr$$

With

$V_{out}$  being the output voltage

$V_{offset}$  being the constant that can change drastically from one sensor to another within the same lot and that drifts  
35   slowly due to aging

**Kps** : the gain in volts/hPa of the pressure sensor that is usually stable.

**Pr** : the difference between pressure sensors 6 and 8

5        Given the previous equation, subtracting the two voltages of the two pressure sensors will then give:

$$VPbPm = VoffPm - VoffPb + Kps * (Pb - Pm)$$

With

**VPbPm** is the Voltage result of the subtraction

10        **VoffPm** is the offset of the Pm pressure sensor

**VoffPb** is the offset of the PB pressure sensor

**Kps** is the gain in volts/hPa of the pressure sensors

By offset it is mean the constant which corresponds to  
15        the difference between the pressure measured by one sensor and the absolute value of pressure.

The invention takes advantages of a well know state of the apparatus when no patient is connected to the apparatus, and no pressure is generated by the blower. During this state  
20        The PM and PB values have the same value than the ambient pressure. The control unit 2 comprises offset compensation means for compensating the possible difference of gauging between the two pressure sensors 6 and 8.

Preferentially the offset compensation means comprise :

25        - a digital to analog converter 32 connected to a microprocessor 30 in order to convert microprocessor's digital data in analog data,

- an analog subtractor 34 having inputs connected to the second pressure sensor, to the first pressure sensor, and to  
30        said digital to analog converter;

said microprocessor calculating, when the blower is not functioning, the difference between the two pressures measured by said first and second pressure sensors and then sending the value C of this difference to said digital to analog  
35        converter, which converts said value C in analog data and drive it to said analog subtractor, which subtract the

pressure  $P_B$  measured by said second pressure sensor and said value  $C$  to the pressure  $P_M$  measured by said second pressure sensor and send the corresponding result  $D$  to the microprocessor, which will modify the  $C$  value until said  $D$  result equals zero, said microprocessor capturing the  $C$  value when said  $D$  result equals zero, enabling the microprocessor which uses the pressure sensors to control the apparatus to correct the difference of offsets between the pressure sensors.

10 Preferentially the apparatus comprises an analog amplifier 36 connected said analog subtractor 34 in order to amplify the signal corresponding to said  $D$  result and to send it to the microprocessor 30, thus enabling the microprocessor to have an accurate adjustment of said value  $C$  until the  
15 result  $D$  reaches the value zero.

The apparatus can also comprise analog to digital converters 42, 44 and 40 connected between the microprocessor 3) and the said first pressure sensor, between the microprocessor and the said second pressure sensor, and  
20 between the microprocessor and the said analog amplifier, so that the microprocessor is provided with only digital data.

With a non standard calibration tube, the process for calibrating a tube used in apparatus to assist patient's respiration, comprising :

- 25 - connecting a first tube's 20 extremity to the blower of an apparatus according to any of claim 1 to 5,
- if the tube is not provided with a pressure sensor at its second extremity, placing said second pressure sensor at said second extremity,
- 30 - connecting said second extremity to a shell 10 with a traversing hole 12 having a known airflow resistance coefficient  $K_s$ ,
- switching the blower on, and instructing said control unit 2 to measured the pressures on both said first and second  
35 pressure sensors,

- calculating the value of the tube airflow resistance coefficient  $K_t$  from these measured pressures and from the said coefficient  $K_s$ .

5           The apparatus can also comprise a Frequency Shift Keying (FSK) modulator which transforms the binary data send by the apparatus sensors or elements in a modulation of the frequency of the tension applied on a voltage controlled current source, connected to the external power supply, so that the voltage  
10 controlled current source transmit the modulation corresponding to the data, a FSK demodulator converting the voltage frequency modulation into binary data and transmit to the elements, so that each sensor or module connected to the power source is able to receive or transmit information.

15           The apparatus can also be used in a set for calibrating a tube used in apparatus to assist patient's respiration comprising the apparatus according to present invention and a shell (10) with a traversing hole (12) having a known airflow resistance coefficient  $K_s$ .

20

          The apparatus enables to modulate the pressure to the patient in respect to the illness to treat. Due to the airflow computation, the apparatus has the capacity to differentiate the two basic states of the respiration:  
25 inspiration and expiration. The sensors provided in the apparatus enables the pressure control unit to control the pressure of the air delivered. The outputs of the E Estimator are the value of the inspiration pressure  $P_I$  which is the pressure maintained at the patient's mask  
30 during the inspiration, and the value of the expiration pressure  $P_E$  which is the pressure maintained at the patient's mask during the expiration. The data of the pressures  $P_M$  and  $P_B$  which are sensed at the extremities of the tube and the data of the tube coefficient  $K_T$  enable the  
35 airflow computation. This computation enables the

computation of the inspiration and expiration, this latest computation enables the estimation module 100 to determinate, which step of the patient's breathing is occurring. A breath estimation step is qualifying a breath in shape, energy (volume) and frequency. The clinician or a qualified user enters parameters of the delivered pressures for the expiration phase and the inspiration phase. The clinician enters also parameters defining how the estimation module 100 is going to react following events detected in the breath estimator. It is well known that a feedback of the patient with his treatment is helping compliance, thus the patient can have an access to a parameter ranging from min to max that is qualified to be "comfort vs. efficiency". This patient setting is having the weight that the clinician is giving to it, from pure placebo effect to some level of effects. Basically the patient settings are applied in the normal breath situation or/and have a limited action on the pressure regulation. It is also possible that the airflow is an input to the estimation module 100. Thus, with the data inputs concerning the breath estimation (and clinical symptoms or event associated with), the inspiration/expiration computation and the clinical settings, and possibly the airflow computation and patient settings, the control unit 2 by the estimation module 100 is able to determinate the pressures required PI and PE. Those two values can be addressed to two different outputs where a switch is able, relative to the inspiration/expiration computation, to connect to the required output regarding if the patient is breathing in or out. The control unit 2 comprises a pressure control loop which, by comparing the pressure measured in the mask and the value of pressure required PI or PE, is able to adjust the tension PWM in order to obtain the correct pressure in the mask. The figure 6 represents one pattern of the pressure of treatment provided according to the airflow due to the patient breathing. In this

example, the clinical technician has set a special modulation of pressures  $P_i$  and  $P_e$  during respectively the inspiration and the expiration; after a while as no special event occurs the values of the two pressures are changed.

5       The apparatus has a two steps strong recognition process in order to prevent false start of the apparatus when the mask is not on the patient's face and to prevent starting a new treatment session. When the apparatus is started by the patient by using the keyboard, for example and as represented  
10 in figure 7 by using the start key, the blower 33 is kept turning at a very low speed, waiting for some activity on the mask pressure sensors 6. When an activity is detected, the apparatus is instantaneously trying to bring the pressure at the apparatus outlet at a minimum starting pressure SP of 4  
15 hPa. When this pressure is reached the apparatus tries to identify at least one breath to start the process according to the settings. When the mask is not applied against something, like a hand or the patient's face, no activity is detected. Then if a maximum time since the apparatus start has been  
20 spend (the timeout is reached) then the blower is stopped. On the contrary, the apparatus keeps on waiting for an activity on the pressure sensors. When the mask is not applied correctly the pressure can not reach 4 hPa. Then if the timeout is reached the blower is stopped, on the contrary the  
25 apparatus waits to detect some activity on the pressure sensors. When the mask is not applied on the patient's face no breath pattern is recognized. Then the blower is stopped if the timeout is reached. On the contrary the apparatus waits to detect some activity on the pressure sensors. The timeout  
30 checking prevents the blower keeping turning on if the patients does not start the treatment and forget to start the blower. A further advantage of this implementation is that if a patient is connected, bringing the pressure instantaneously to 4 hPa will prevent CO<sub>2</sub> rebreathing.

The following examples demonstrates the way the estimation module 100 modulates the pressure value  $P_M$  to apply to the patient's mask.

5 Example 1 : Variations of the pressure of the average pressure of treatment as a function time according to detected events

In a preferential embodiment the average pressure of treatment on one breathing step is not constant in time and  
10 will be modulate by the estimation module 100 according to the events occurring, such as snoring or apneas.

The apparatus will try to reduce the average pressure of treatment value, thus enhancing the patient comfort while breathing against the apparatus. The  
15 clinician set a minimum average pressure of treatment  $P_{tmin}$  and a coefficient  $NOEK$  expressed in hPa/s.

As represented on figure 6 , when no events are detected the average pressure of treatment value will follow the equation:  
20

$$PT(t) = MAX(PT(t-\varepsilon) - (NOEK \times \varepsilon), PT \min)$$

$\varepsilon$  being the sampling time, the MAX function is returning the greatest value of its two members.

25 The average pressure value  $P_M$ , corresponding to the pressure of treatment  $PT$  on one breath thus decrease linearly until it reaches the minimum set by the clinician and stays constant until an event occurs. The average pressure is change each sampling time which correspond to  
30 one single breath (one inspiration and one consecutive expiration).

If an event is detected, a 3 steps process is initiated:

- Step 1: the estimation module, looking in the clinical settings is defining if the event has to affect the  
35 average value of pressure  $PT$ .

- Step 2 : If so the estimation module looking in clinical settings, will define a persistence delay  $D_p$ .

In the example of figure 6 a snore is detected, the persistence delay could be set to 2 minutes.

- 5 • Step 3 : A  $E_k$  parameter in hPa/s is extracted from clinical settings and balanced with an eventual ongoing event. The estimation module will determine the  $E_k$  corresponding to the event which occurs or has occurred and linearly increase the average pressure of treatment with  $E_k$  as slope coefficient. AVP will then follow the equation:

$$PT(t) = \text{MIN}(PT(t - \varepsilon) + (E_k \times \varepsilon), AVP \text{ max})$$

$\varepsilon$  being the sampling time, the MIN function is returning the smallest value of its two members.

- 15 During the persistence delay even if no event occurs the estimation module will keep on increasing the average pressure of treatment PT.

#### Example 2 : d1, d2 auto adjustment

- 20 On the figures 6 d1 and d2 have two different behaviors, according to t2 and t4 definitions. t2 is defined when the absolute value of airflow starts to decrease within the inspiration phase or shows a fixed delay after t1. t4 is defined when the absolute value of airflow starts to decrease within the expiration phase or shows a fixed delay after t3.

- 25 If t2 and t4 are defined according to the airflow waveform then no auto-adjustment is occurring. If not, following the same rules than AVP, the d1 and d2 can also be affected by the events. In the case when d1 & d2 are not locked by the breath waveform following the same process, d1 & d2 can also be adjusted according to events in the case.

## CLAIMS

1. An apparatus (1) to assist a patient respiration by  
5 delivering air to a patient through a mask, said mask being  
designed to be connected on one first extremity of a tube,  
said apparatus comprising :
- a control unit (2) to adjust the pressure delivered by the  
blower (4) of said apparatus,
  - 10 - a first pressure sensor (6) for sensing the pressure  $P_M$  at  
said first tube extremity and being connected to said control  
unit, and
  - a second pressure sensor (8) for sensing the pressure  $P_B$  at  
the air output of said blower and being connected to said  
15 control unit;
- in order that, when a tube is connected to said mask and  
connected to said apparatus on its said second extremity, the  
air flowing from the apparatus to the mask, said control unit  
is able to calculate the airflow at said second extremity of  
20 the tube from said pressures  $P_M$  and  $P_B$  and from the airflow  
resistance coefficient  $K_T$  of said tube.
2. The apparatus according to claim 1, wherein when at  
least one filter (22) is placed at one tube's extremity, said  
control unit (2) is able to calculate the airflow at said  
25 second extremity of the tube (20) from these measured  
pressures  $P_M$  and  $P_B$  and from the airflow resistance  
coefficient  $K_T$  of said tube and from the airflow resistance  
coefficient  $K_F$  of said filter.
3. The apparatus according to claim 1 or 2, wherein when  
30 a tube is connected between said apparatus (1) and a shell  
(10) with a traversing hole (12) having a known airflow  
resistance coefficient  $K_S$ , the air flowing from the apparatus  
to said shell, the measured pressures  $P_M$  and  $P_B$  are sent to  
said control unit (2) which calculates the tube airflow  
35 resistance coefficient  $K_T$  from these measured pressures and  
from the said coefficient  $K_S$ .

4. An apparatus according to any one of claim 1 to 3, wherein said control unit (2) comprises a non volatile memory in which the control unit stores, as a couple of values, the two pressures  $PM(J)$  and  $PB(J)$ , measured at each said pressure sensors (6 and 8) when said control unit forces the blower to deliver a determined constant pressure  $I$  at one of the two sensors (6 or 8), so that when at least two couples of pressures corresponding to two different said determined constant pressure  $I$  are stored, the control unit is able to calculate an average of said coefficient  $K_T$ .

5. The apparatus according to any of the previous claims, wherein the control unit (2) comprises offset compensation means for compensating the possible difference of gauging between the two pressure sensors (6 and 8).

6. An apparatus (1) according to claim 5, wherein said offset compensation means comprise :

- a microprocessor (30)
- a digital to analog converter (32) connected to said microprocessor (30) in order to convert microprocessor's digital data in analog data,
- an analog subtractor (34) having inputs connected to the second pressure sensor (8), to the first pressure sensor (6), and to said digital to analog converter, said microprocessor calculating, when the blower is not functioning, the difference between the two pressures measured by said first and second pressure sensors and then sending the value  $C$  of this difference to said digital to analog converter, which converts said value  $C$  in analog data and drive it to said analog subtractor, which subtract the pressure  $PB$  measured by said second pressure sensor and said value  $C$  to the pressure  $PM$  measured by said second pressure sensor and send the corresponding result  $D$  to the microprocessor, which will modify the  $C$  value until said  $D$  result equals zero, said microprocessor capturing the  $C$  value when said  $D$  result equals zero, enabling the control unit to correct the difference of offsets between the pressure sensors.

7. An apparatus according to claim 6, further comprising an analog amplifier (36) connected to said analog subtractor (34) in order to amplify the signal corresponding to said D result and to send it to said microprocessor (30), thus  
5 enabling said microprocessor to have an accurate adjustment of said value C until said result D reaches the value zero.

8. An apparatus according to claim 7, further comprising analog to digital converters (42, 44 and 40) connected between the microprocessor (30) and the said first pressure sensor  
10 (6), between the microprocessor and the said second pressure sensor (8), and between the microprocessor and the said analog amplifier (36), so that the microprocessor is provided with only digital data.

9. Process for calibrating a tube used in apparatus to  
15 assist patient's respiration by using the apparatus (1) according to any of claim 1 to 5, said process comprising :  
- connecting a first tube's (20) extremity to the blower (4) of said apparatus,  
- connecting said first pressure sensor (6) to measure the  
20 pressure PM at a second tube's extremity,  
- connecting said second extremity to a shell (10) with a traversing hole (12) having a known airflow resistance coefficient  $K_s$ ,  
- switching the blower on,  
25 - instructing said control unit (2) to measured the pressures on said first pressure sensor and on the second pressure sensor (8), which is measuring the pressure PB at the blower's apparatus outlet, and  
- calculating the value of the tube airflow resistance  
30 coefficient  $K_t$  from these measured pressures PM and PB and from the said coefficient  $K_s$ .

10. Process for calibrating the tube used in apparatus to assist patient's respiration, and for calibrating the tube by using the apparatus (1) according to any of claim 1 to 5,  
35 said process comprising :  
- connecting a first tube's (20) extremity to the blower (4) of said apparatus,

- connecting said first pressure sensor (6) to measure the pressure PM at a second tube's extremity,
- connecting said second extremity to a shell (10) with a traversing hole (12) having a known airflow resistance coefficient  $K_s$ ,
- switching the blower on,
- fixing at a value I the pressure provided and measured on one pressure sensor,
- instructing said control unit (2) to measured the pressures on said first pressure sensor and on the second pressure sensor (8), which is measuring the pressure PB at the blower's apparatus outlet,
- storing these measures PM(J) and PB(J) as a couple of measures associated to said value I,
- repeating a number of time N the steps 5 to 6 of said process, said value I being different for each time, so that each couples of pressures is associated with one value I,
- calculating (20) on average of the airflow resistance coefficient  $K_T$  from these measured pressures PM and PB and from the said coefficient  $K_s$ .

11. The apparatus according to claim 1 to 8, wherein said pressure control unit (2) comprises an estimation module (100) connected to the means for detecting the patient's breathing parameters (110), in order that the estimation module is able to determine when the patient is inspiring or expiring and in response the pressure to apply to the patient's mask, so that the control unit adjust the pressure delivered by the blower.

12. The apparatus according to claim 11 or 12, wherein the control unit comprises a non volatile memory (120) in which the clinician can enter clinical settings (120) comprising at least the treatment pressure and possibly the pressure to apply according to the patient's breathing parameters, said estimator applying the pressure according to these clinical settings and to the patient's breathing parameters.

13. The apparatus according to claim 13, wherein the patient can enter patient settings (122) in said non volatile memory, said estimator applying the pressure according to these patient settings and to the patient's breathing parameters within bounds given by the clinician settings.

14. The apparatus according to any one of claim 11 to 14, in which the estimation module 100 is able to determine that an event ( $E_1$ ,  $E_2$  or  $E_3$ ) occurs in patient's breathing thus enabling said control unit to adjust the tension to apply to the blower to adjust the pressure at patient's mask.

15. The apparatus according to any one of claim 11 to 15, wherein said means (6) for detecting the patient's breathing parameters enable the control unit (2) to compute the airflow at patient's mask (20), said comparator determining that an event ( $E_1$ ,  $E_2$  or  $E_3$ ) is occurring with the airflow parameters or shape.

16. The apparatus according to claim 11 to 17, wherein said estimation module has an inspiration out put (102) where said estimation module set the mask pressure PM value during inspiration and wherein said estimation module has an expiration out put (102) where said estimation module set the mask pressure PM value during expiration, said control unit comprising a switch which is connected alternatively to the inspiration out put (102) or expiration out put (102) according to patient's breathing.

17. The apparatus according to claim 11 to 18, wherein the apparatus further comprise a starting mean which when actuated enables the estimation module (100) to determine if a breathing activity is detected, the estimator module sending the instruction to stop the blower if no activity is sensed after a given delay.

18. The apparatus according to any one of the previous claim, further comprising a Frequency Shift Keying (FSK) modulator 50 which transforms the binary data send by the apparatus sensors or elements in a modulation of the frequency of the tension applied on a voltage controlled current source 52, connected to the external power supply, so that the

voltage controlled current source 52 transmit the modulation corresponding to the data, a FSK demodulator converting the voltage frequency modulation into binary data 61 and transmit to the elements, so that each sensor or module connected to  
5 the power source is able to receive or transmit information.

19. Set for calibrating a tube used in apparatus to assist patient's respiration comprising :

- an apparatus according to claim 1 to 6
- a shell (10) with a traversing hole (12) having a known  
10 airflow resistance coefficient  $K_s$ .

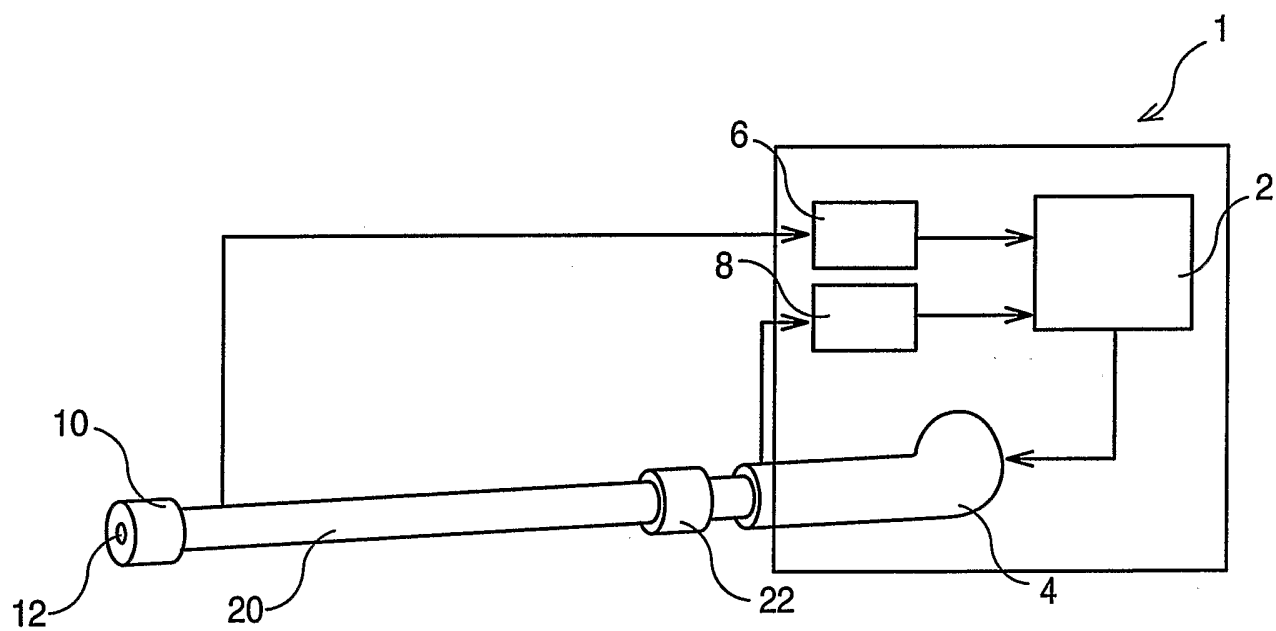


FIG. 1

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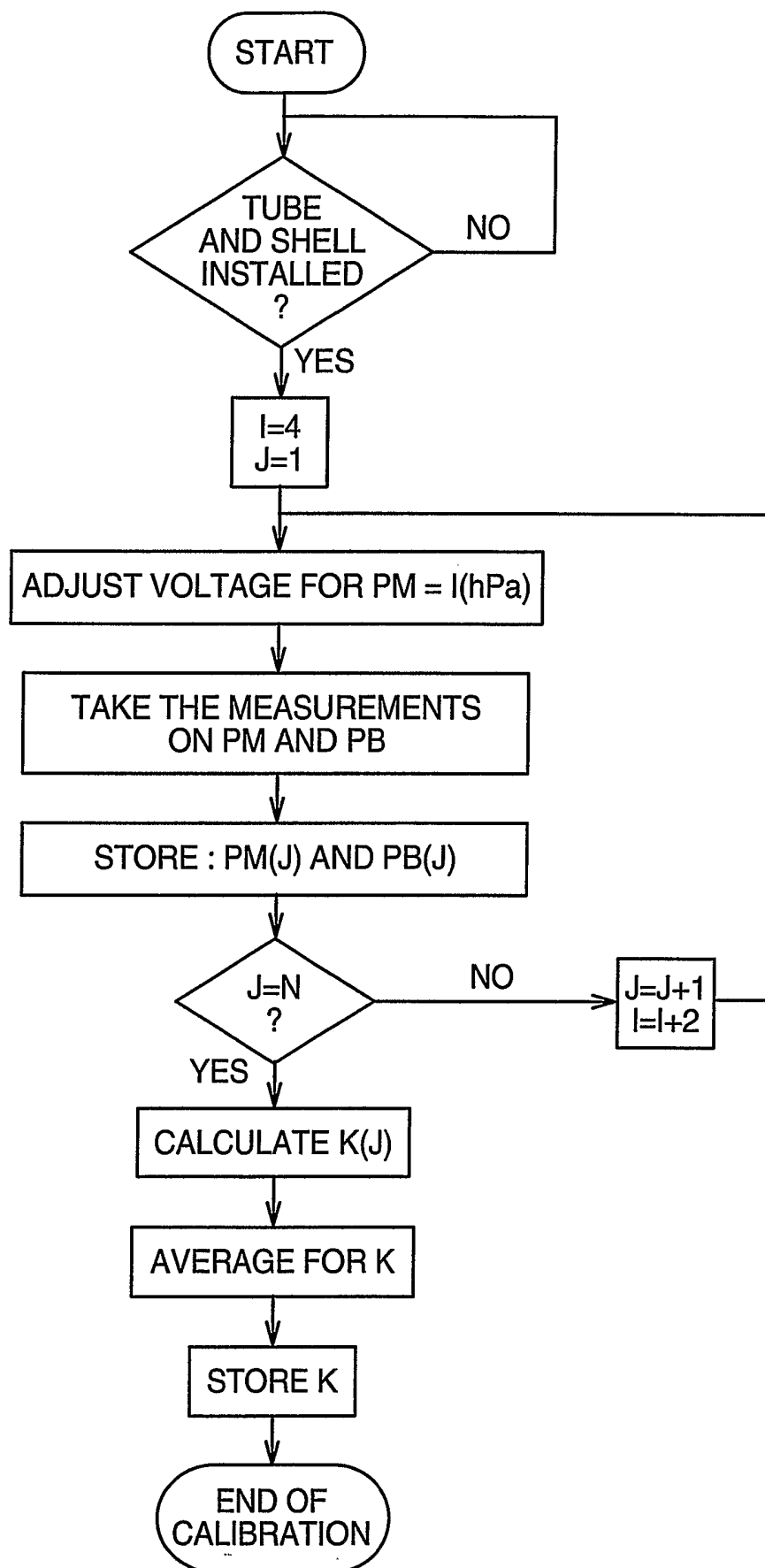


FIG. 2

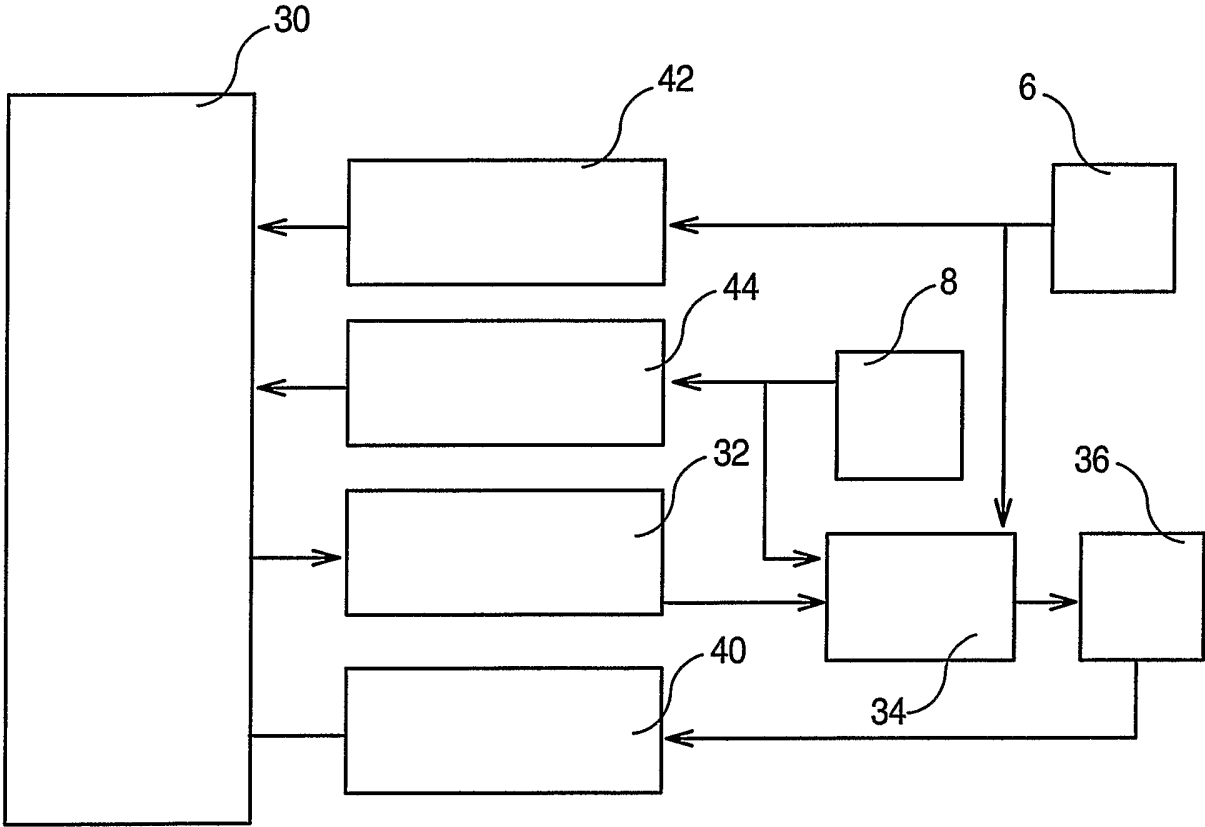


FIG. 3

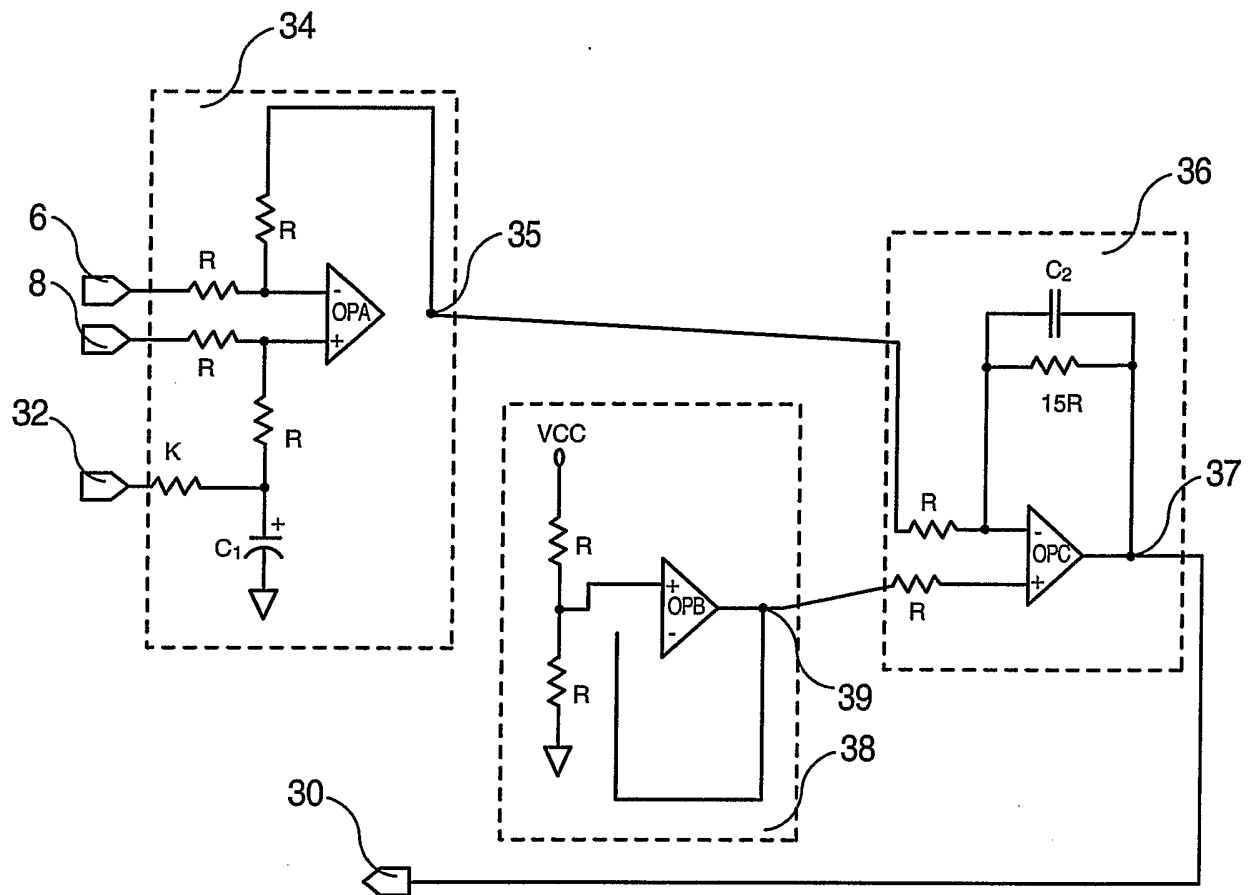


FIG. 4

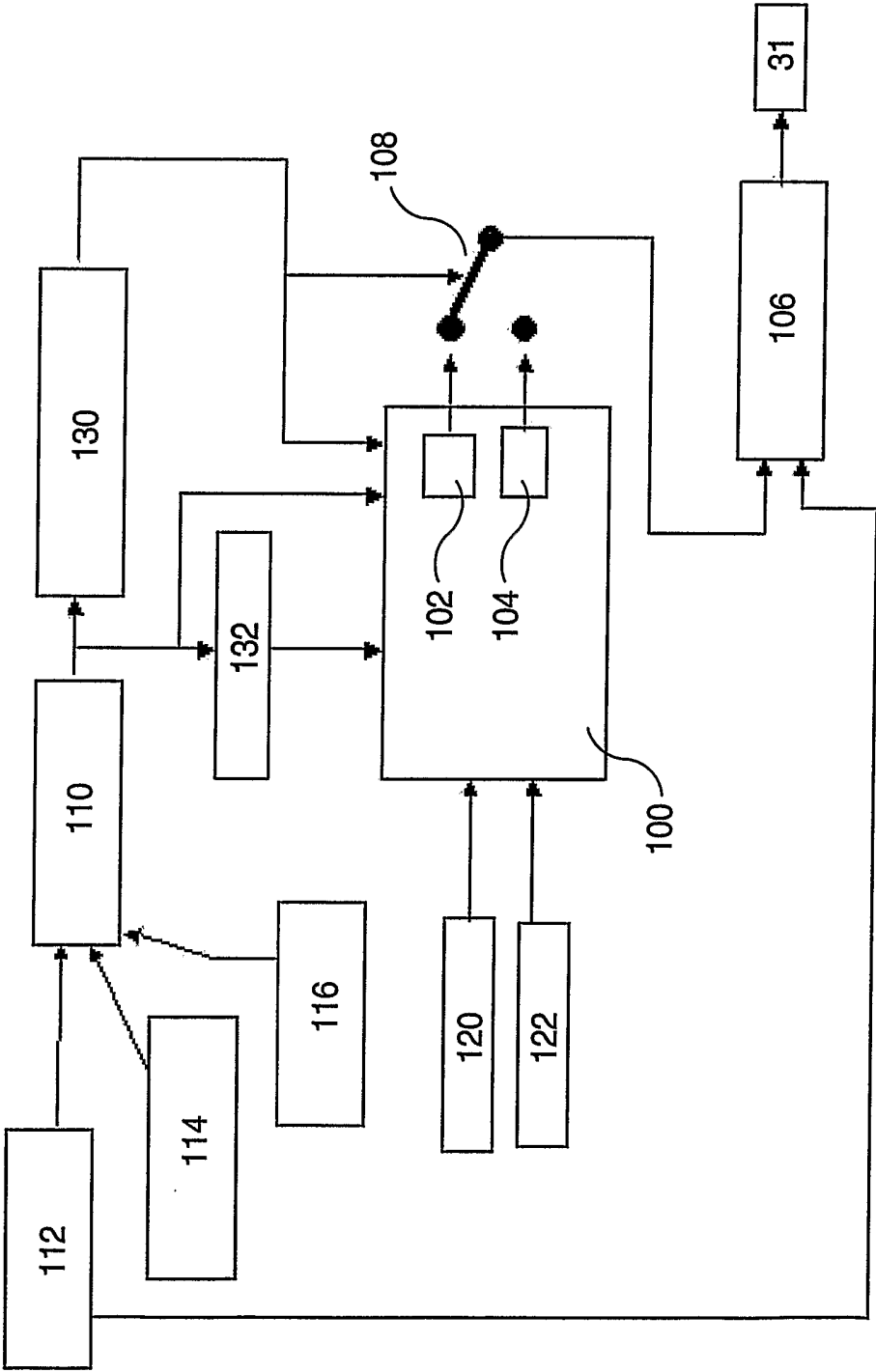


FIG. 5

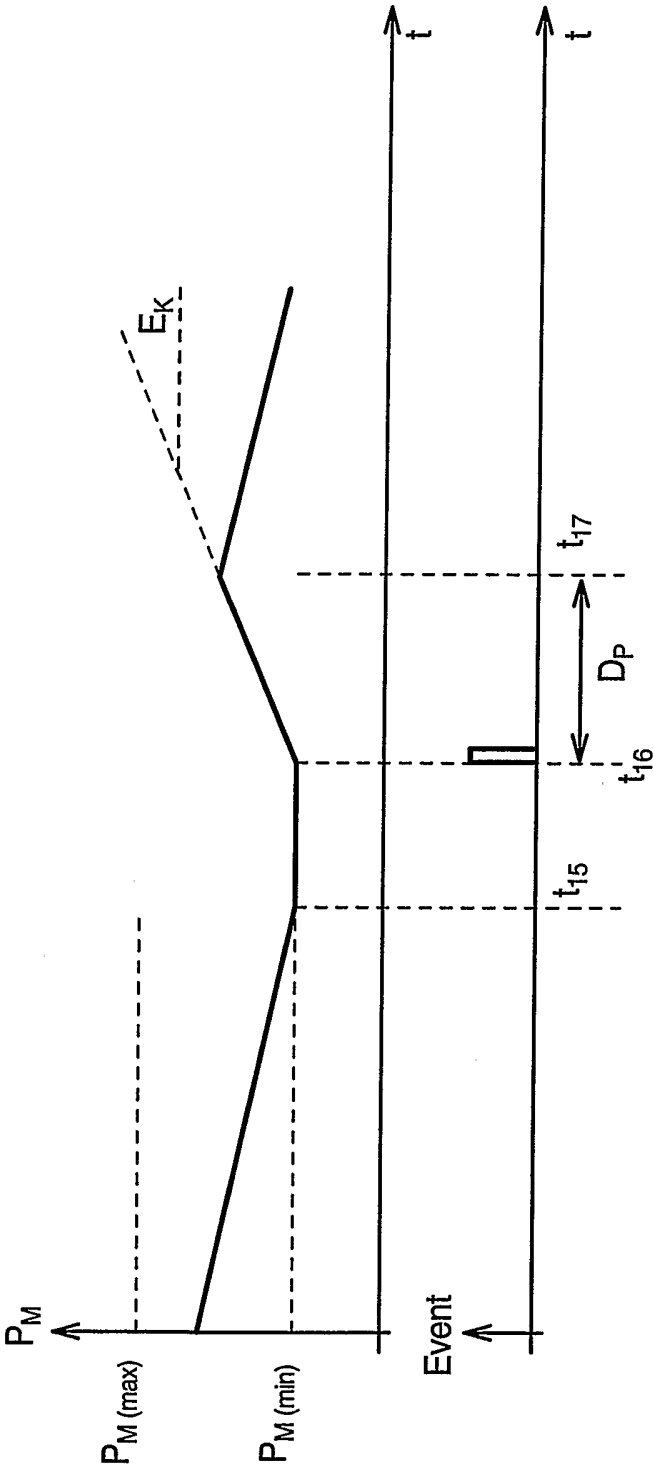


FIG. 6

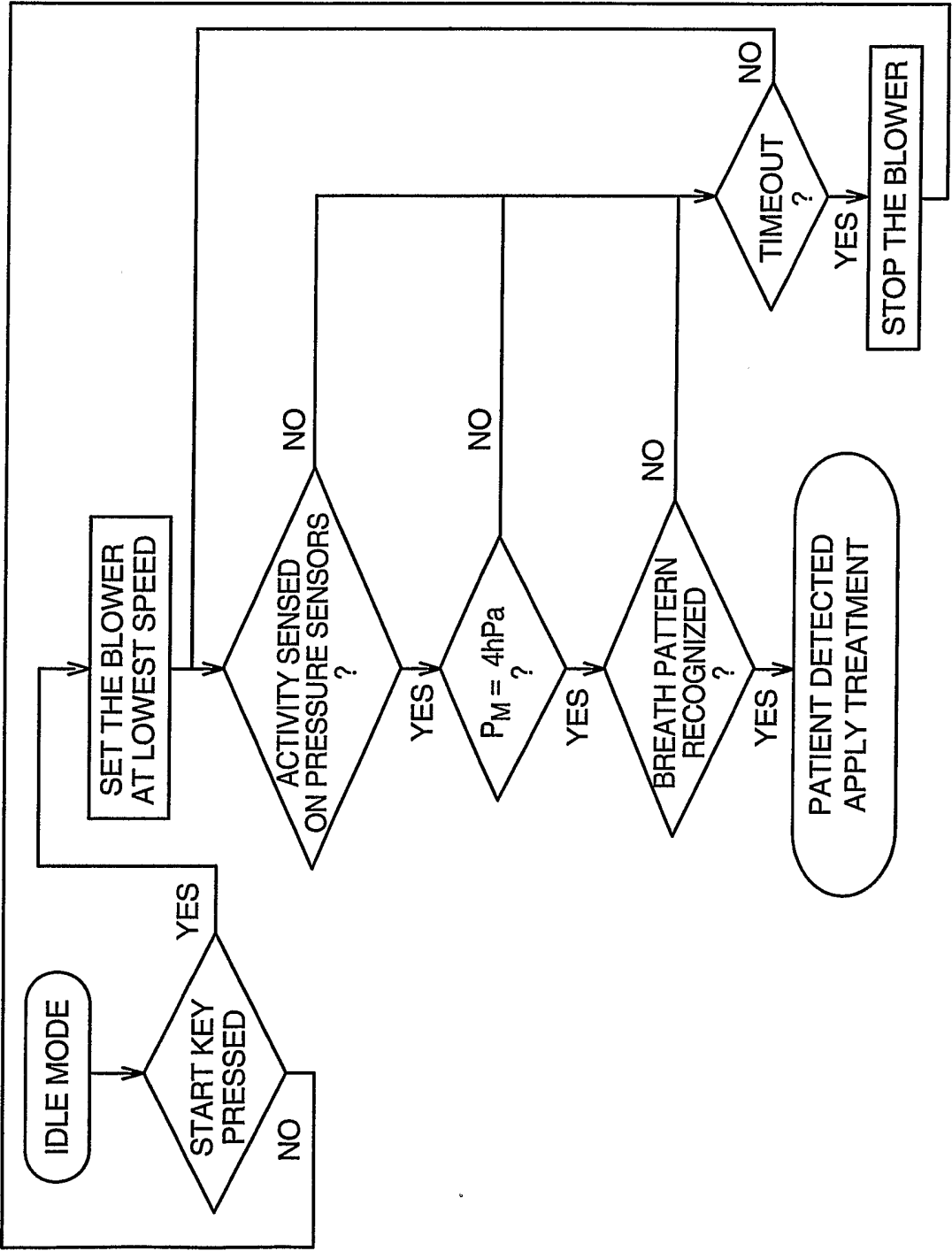


FIG. 7