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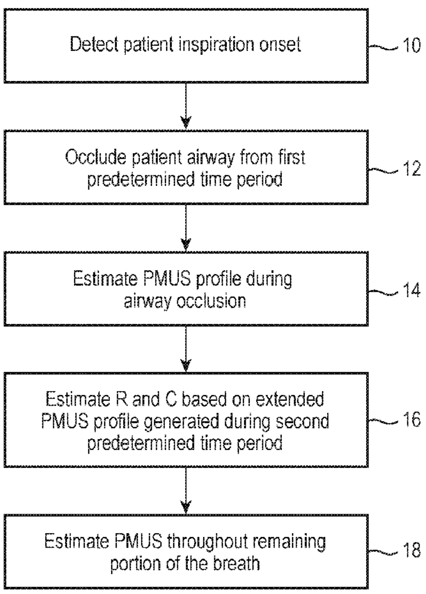

FIG. 1

(57) Abstract: When estimating respiratory muscle pressure and respiratory mechanics using a $P0_{.1}$ maneuver patient inspiration onset for a patient connected to a ventilator (64) is detected, and the airway of the patient is occluded for a first predetermined time period. A first respiratory muscle pressure (P_{mus}) profile is estimated during the airway occlusion. Resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period are then estimated. A third P_{mus} profile is estimated during a third predetermined time period that extends from the end of the second predetermined time period until the end of inspiration. $P_{mus}(t)$ over an entire breath is estimated by concatenating the first, second and third P_{mus} profiles, and the estimated R and C values and the estimated P_{mus} profiles are output on a display.

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SYSTEM AND METHODS FOR ESTIMATION OF RESPIRATORY MUSCLE PRESSURE AND RESPIRATORY MECHANICS USING $P_{0.1}$ MANEUVER

FIELD

[0001] The present invention finds application in patient ventilation systems and methods. However, it will be appreciated that the described techniques may also find application in other patient care systems, other patient parameter estimation techniques, and the like.

BACKGROUND

[0002] Estimating respiratory muscle pressure ($P_{mus}(t)$) is of paramount importance in support modalities of mechanical ventilation, such as Pressure Support Ventilation (PSV), where patient and ventilator share the mechanical work performed on the respiratory system. Quantitative assessment of $P_{mus}(t)$ can be used to select the appropriate level of ventilation support in order to prevent both atrophy and fatigue of the respiratory muscles. One clinical parameter commonly used to assess the effort made by the patient per breath is known as Work of Breathing (WOB) and can be computed once the estimate of $P_{mus}(t)$ is available for the breath (e.g., WOB can be obtained from $P_{mus}(t)$ by integration of the latter over the inhaled volume). One conventional approach for $P_{mus}(t)$ and WOB estimation relates to measuring the esophageal pressure (P_{es}) via insertion of a balloon-tipped catheter in the patient's esophagus. The measured $P_{es}(t)$ is assumed to be a good proxy for the pleural pressure (P_{pl}) and can be used, in conjunction with an estimate of chest wall compliance, to compute the WoB via the so-called Campbell diagram or, equivalently, via explicit computation of $P_{mus}(t)$ and then of WOB.

[0003] Estimates of R and C are important per se, as they provide quantitative information to the physician about the mechanical properties of the patient's respiratory system and they can be used to diagnose respiratory diseases and better select the appropriate ventilation modalities and therapeutic paths. Moreover, R and C can also be used to estimate $P_{mus}(t)$ as a non-invasive alternative to the use of the esophageal catheter. Assuming R and C are known, it is indeed possible to estimate $P_{mus}(t)$ via the following equation (known as the Equation of Motion of the lungs):

$$P_{aw}(t) = R \cdot \dot{V}(t) + E \cdot V(t) + P_{mus}(t) + P_0 \qquad (1)$$

where $P_{aw}(t)$ is the pressure measured at the airway opening, $\dot{V}(t)$ is the flow of air into and out of the patient's respiratory system (measured again at the airway opening), $V(t)$ is the net volume of air delivered to the patient (measured by integrating the flow signal over time), E is the elastance (inverse of the compliance C) and P_0 is a constant term to account for the pressure at the end of expiration (needed to balance the equation but not interesting per se).

[0004] Previous attempts to use equation (1) for a non-invasive estimation of $P_{mus}(t)$ relied on a two-step approach, where R and C are estimated first and then equation (1) is applied to compute $P_{mus}(t)$ using the estimated values of R and C. Estimation of R and C was performed either by applying the End-Inspiratory Occlusion (EIP) maneuver or via Least-Squares (LS) fitting of equation (1) to flow and pressure measurements under specific conditions, where the term $P_{mus}(t)$ was assumed to be zero. These conditions included:

1. periods of patient paralysis and Continuous Mandatory Ventilation (CMV);
2. periods of high Pressure Support Ventilation (PSV) level;
3. specific portions of every pressure-supported breath that extend both during the inhalation and the exhalation phases;
4. exhalation portions of pressure-supported breaths, where the flow signal satisfies specific conditions that are indicative of the absence of patient inspiratory efforts.

[0005] The quantitative assessment of the mechanical properties of the respiratory system and of the inspiratory efforts in patients under mechanical ventilation offers invaluable information for the clinician to tailor ventilation strategy and settings. The current state of the art for the assessment of respiratory mechanics consists of computing two parameters, namely resistance (R) and compliance (C), via the EIP technique. This technique, however, not only interferes with the normal operation of the ventilator, but it requires the respiratory muscles to be fully relaxed in order to provide accurate R and C estimates. Hence, because of the presence of respiratory activity from the patient, the EIP often leads to biased results. Assessment of inspiratory patient efforts, on the other hand, is traditionally obtained by inferring the pressure generated by the respiratory muscles (P_{mus}) from the pressure measured in the esophagus (P_{es}). Quantitative assessment of patient effort is then obtained on a breath-by-breath basis by computing Work of Breathing (WOB) from the P_{mus} waveform. The main limitation of such approach is that measurement of P_{es} requires the insertion of an esophageal catheter, with consequent discomfort for the patient in addition to the need for special instrumentation and skilled personnel.

[0006] Other methods have been developed to allow simultaneous estimation of R, C and $P_{mus}(t)$ from measured airway pressure and flow waveforms during regular ventilation without requiring esophageal pressure measurements. These methods are based on the use of the traditional first-order single-compartment model of respiratory mechanics and its associated Equation of Motion in (1). They all face the fundamental difficulty of the simultaneous estimation approach related to the underdetermined nature of the mathematical problem (more unknowns than available equations). In these methods, the use of constrains based on physiological assumptions have been advocated to render the mathematical problem solvable. However, these methods have been shown to work only under specific conditions. Particularly, when the ventilator cycles off before the patient has completely released his respiratory muscles (i.e., P_{mus} has returned to zero baseline value), these conventional methods are not reliable. This may limit their applicability to all clinical scenarios.

[0007] The disadvantage of the conventional invasive procedure of esophageal pressure measurement is apparent, since the insertion of an esophageal balloon requires experienced personnel and implies discomfort and risk for the patient.

[0008] The two-step estimation technique, where the EIP maneuver is first performed to get R and C and equation (1) is then used to compute $P_{mus}(t)$, has the following main drawbacks:

1) The patient's respiratory muscles ought to be fully relaxed during the EIP maneuver in order for the R and C computation to be valid.

2) The EIP maneuver is performed in a specific ventilation mode (Volume Assisted Control, VAC) and the resulting R and C values might not be representative of the corresponding values that determine the dynamics of the lung mechanics under other ventilation modes, such as PSV. Therefore, the accuracy of $P_{mus}(t)$ computed via equation (1) during PSV operation can get compromised.

3) The EIP maneuver interrupts the regular ventilation pattern needed by the patient.

[0009] Finally, the above mentioned two-step techniques that apply LS fitting under specific conditions or to portions of the breath, where $P_{mus}(t)$ is theoretically negligible, present limitations. In particular:

1) Repeated periods of paralysis, after the patient has recovered, plus CMV are not clinically feasible.

2) Repeated periods of high PSV interfere with the normal operation of the ventilator and may not be beneficial to the patient.

3) The assumption of negligible $P_{mus}(t)$ during pressure-supported breaths is debatable, especially during the inhalation phase.

[0010] The present application provides new and improved systems and methods that facilitate noninvasively estimating of R, C, and P_{mus} using an airway occlusion pressure maneuver ($P_{0.1}$) having a predetermined duration, thereby overcoming the above-referenced problems and others.

BRIEF SUMMARY

[0011] Still further advantages of the subject innovation will be appreciated by those of ordinary skill in the art upon reading and understand the following detailed description.

[0012] According to one embodiment, a method for estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver comprises detecting patient inspiration onset for a patient connected to a ventilator, occluding the airway of the patient for a first predetermined time period, and estimating a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion. The method further comprises estimating resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period, estimating a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of the breath, and estimating $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles. The estimated R and C values and the estimated P_{mus} profiles are output on a display.

[0013] According to another embodiment, a system that facilitates estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver comprises a ventilator to which a patient is connected, and one or more processors configured to detect patient inspiration onset for a patient connected to a ventilator and occlude the airway of the patient for a first predetermined time period. The one or more processors are further configured to estimate a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion, estimate resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period, and estimate a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of the breath. Additionally, the one or more processors are configured to estimate $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles, and output the estimated R and C values and the estimated P_{mus} profiles on a display.

[0014] According to another embodiment, a processor is configured to execute computer-executable instructions for estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver. The instructions comprise detecting processor patient inspiration onset for a patient connected to a ventilator, occluding the airway of the patient for a first predetermined time period, and estimating a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion. The instructions further comprise estimating resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period, and estimating a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of the breath. Additionally, the instructions comprise estimating $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles, and outputting the estimated R and C values and the estimated P_{mus} profiles on a display.

BRIEF DESCRIPTION OF THE DRAWINGS

[0015] The drawings are only for purposes of illustrating various aspects and are not to be construed as limiting.

[0016] FIGURE 1 is a flowchart illustrating a method for estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver in accordance with one or more aspects described herein.

[0017] FIGURE 2 illustrates a graph summarizing steps of the method of Figure 1.

[0018] FIGURE 3 illustrates exemplary results from method of Figure 1 on one illustrative breath, where the estimated P_{mus} profile is compared against the gold-standard P_{mus} waveform measured inside the vessel.

[0019] FIGURE 4 is a graph showing error that may be introduced during an occlusion period when a polynomial model of P_{mus} is fit to the airway pressure measurements during the occlusion period.

[0020] FIGURE 5 illustrates a system that facilitates estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver in accordance with one or more aspects described herein.

[0021] FIGURE 6 shows a system that facilitates estimating work of breathing (WOB) in a patient hooked up to a ventilator with automated software for the $P_{0.1}$ maneuver.

[0022] FIGURE 7 illustrates a system that facilitates estimating work of breathing (WOB) and power of breathing (POB) in a patient hooked up to a ventilator with automated software for the $P_{0.1}$ maneuver wherein the ventilator is operating in Proportional Assist Ventilation (PAV) mode.

DETAILED DESCRIPTION

[0023] The need for estimation of the respiratory system parameters (resistance R and compliance C) and patient inspiratory efforts (respiratory muscle pressure $P_{mus}(t)$) is well-known in the medical community. In order to overcome the above-described problems in the art, the herein-described systems and methods relate to an alternative approach for the noninvasive estimation of R, C, and P_{mus} that makes use of an airway occlusion pressure maneuver ($P_{0.1}$) having a predetermined duration (e.g., less than 150ms or the like) to circumvent the inherent difficulty of the simultaneous estimation approach. The described method involves, inter alia, the following steps: 1) In the first step, the patient's airway is occluded at end of exhalation as soon as zero flow condition is detected; occlusion is maintained for a first predetermined time period, (e.g., 100ms) and the airway pressure waveform during these 100ms is used to estimate the coefficients of a polynomial model of $P_{mus}(t)$; 2) once the occlusion has been released, the estimated $P_{mus}(t)$ profile is extended (in time) for a second predetermined time period (e.g., for an additional 100ms) and airway pressure and flow waveforms are used together with the extended P_{mus} profile to estimate R and C using the equation of motion via a standard Least-

Square method; 3) the estimated R and C are used in conjunction with airway pressure and flow waveforms to reconstruct a P_{mus} profile for a third predetermined time period (e.g., throughout the remaining portion of the breath) based on the standard equation of motion. The $P_{0.1}$ maneuver can be intermittently repeated at a variable or fixed rate (e.g., every X number of breaths) while the values of R and C estimated during the previous maneuver can still be used to compute an estimate of P_{mus} between each consecutive $P_{0.1}$ maneuver. This also allows computation of WOB (or power of breathing (POB)) from the estimated P_{mus} profile on a breath-by-breath basis. In one embodiment, the claimed systems and methods are employed in hospital and home ventilators for real-time patient monitoring, ventilation optimization and closed-loop control.

[0024] The herein-described systems and methods overcome the aforementioned limitations of conventional approaches by; not requiring an esophageal balloon; explicitly accounting for the presence of P_{mus}; and not requiring a change ventilation mode during the maneuver so that the resulting R and C estimates are still related to the current ventilation operating conditions. Moreover, unlike the EIP, the $P_{0.1}$ maneuver does not modify the patient's natural breathing pattern. Unlike other conventional approaches, the $P_{0.1}$ is still reliable even when the ventilator cycles off before P_{mus} has returned to zero baseline value.

[0025] The described systems and methods facilitate performing noninvasive estimation of R, C and P_{mus} in patients receiving mechanical ventilation and able to breathe spontaneously. The R, C and P_{mus} estimates can be used for real-time patient monitoring, ventilation optimization and closed-loop control. The described systems and methods can be implemented as part of software or firmware running on a ventilator, anesthesia machines, or patient monitoring products (including remote patient monitors, e.g. eICU). The described systems and methods improve ventilator function by improving the accuracy of estimated R, C and P_{mus} values.

[0026] FIGURE 1 is a flowchart illustrating a method for estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver in accordance with one or more aspects described herein. The method facilitates performing noninvasive estimation of R, C and $P_{mus}(t)$ from airway pressure and flow measurements on a breath-by-breath basis. At **10**, patient inspiration onset is detected, such as by sensing a characteristic pressure profile and flow profile from ventilator circuit pressure and flow sensors disposed in the patient circuit. At, **12**, the patient's airway is occluded for a first predetermined time period by use of an occluder device such as a valve or flap disposed in the ventilator airflow path to the patient and under control of software for automatic operation. The first predetermined time period may be any suitable time period (e.g., less than approximately 150ms, etc.). In the remainder of the document, a predetermined time period of 100ms will be discussed but is not to be construed in a limiting sense. At **14**, the initial inspiratory P_{mus} profile during airway occlusion is estimated. At **16,** after the occlusion period, R and C are estimated based on an extended P_{mus} profile generated during a second predetermined time period. The second predetermined time period may be any suitable time period (e.g., less than approximately 150ms, etc.) and need not be equal to the first predetermined time period in duration. At **18**, P_{mus} is estimated using data collected during a third predetermined time period (e.g., throughout the remaining portion of the breath) following the second predetermined time period.

[0027] When performing estimation of the initial inspiratory P_{mus} profile during airway occlusion at **14**, the patient's airway is occluded (at **12**) at end of exhalation, as soon as the patient's inspiratory effort is detected (at **10**). The occlusion is then maintained for, e.g., 100ms, during which the patient is essentially trying to inhale against a closed airway. In one embodiment, the $P_{0.1}$ maneuver is software-automated. During the occlusion, since there is no airflow between the point at which the airway pressure is measured and the patient's lungs, the negative deflection in pressure is measured at the airways (P_{aw}) and is essentially a reflection of the P_{mus} developed by the patient's respiratory muscles (gas decompression can be neglected) such that:

$$P_{aw}(t) = P_{mus}(t) \qquad \text{for } 0 \leq t \leq 100\ ms$$

[0028] The small duration of the occlusion (e.g., less than 150ms) ensures that the patient's natural respiratory P_{mus} output is not affected by the application of the occlusion. Hence, it is possible to fit a polynomial model of P_{mus} to the airway pressure measurements during the 100ms occlusion and estimate the initial inspiratory P_{mus} profile via standard Least-Square (LS) technique. For instance, a 2^{nd} order polynomial P_{mus} model could be assumed and its unknown coefficients could then be estimated as shown below:

$$P_{mus}(t) = a_1 + a_2 \cdot t + a_3 \cdot t^2 \quad \text{for } 0 \leq t \leq 100\ ms$$

$$Y = \begin{bmatrix} P_{aw}(t_1) \\ P_{aw}(t_2) \\ . \\ . \\ . \\ . \\ P_{aw}(t_k) \end{bmatrix} = \begin{bmatrix} 1 & t_1 & {t_1}^2 \\ 1 & t_2 & {t_2}^2 \\ & . & \\ & . & \\ & . & \\ & . & \\ 1 & t_k & {t_k}^2 \end{bmatrix} \begin{bmatrix} a_1 \\ a_2 \\ a_3 \end{bmatrix} = X \cdot \theta$$

$$\hat{\theta} = [X^T X]^{-1} X^T Y$$

where θ is the vector of unknown parameters $[a_1\ a_2\ \ a_3]$ (i.e., the coefficients of the polynomial P_{mus} model), Y is the vector containing the airway pressure measurements, k is the total number of samples collected during the 100 ms occlusion and t_1, t_2 .....t_k are the time at which the airway pressure signal is sampled (i.e., t_1=0, t_2=T, t_3=2T, ...t_k=(k-1)T with T being the sample period).

[0029] When estimating R and C after the occlusion based on a 100ms extended P_{mus} profile at **16**, after the 100ms occlusion period, the airways are released and air flows to the lungs under the pressure gradient established by both the patient's own P_{mus} drive and the contribution of the ventilator. In such conditions, it may be difficult to estimate R, C and P_{mus} simultaneously from flow and pressure measurements based on the simple equation of motion because the underlying LS problem is underdetermined. However, it is reasonable to assume that for a very short period of time (e.g., 100ms), the profile of P_{mus} remains unchanged compared to the profile estimated during the previous 100ms occlusion period. Hence, it is possible to extend the P_{mus} profile based on the previously estimated polynomial coefficients and attain an estimate of P_{mus} during this additional 100ms post-occlusion period such that:

$$\hat{P}_{mus}(t) = a_1 + a_2 \cdot t + a_3 \cdot t^2 \quad \text{for } 100\ ms \leq t \leq 200\ ms$$

[0030] The extended $P_{mus}(t)$ profile can be used together with the airway pressure and flow waveforms to estimate R and C using the equation of motion via a LS method such that:

$$P_{aw}(t) - \hat{P}_{mus}(t) = \hat{R} \cdot \dot{V}(t) + \hat{E} \cdot V(t) + P_0 \quad \text{for } 100\,ms \le t \le 200\,ms$$

$$\bar{Y} = \begin{bmatrix} P_{aw}(t_{k+1}) - \hat{P}_{mus}(t_{k+1}) \\ P_{aw}(t_{k+2}) - \hat{P}_{mus}(t_{k+2}) \\ \vdots \\ P_{aw}(t_{k+K}) - \hat{P}_{mus}(t_{k+K}) \end{bmatrix} = \begin{bmatrix} \dot{V}(t_{k+1}) & V(t_{k+1}) & 1 \\ \dot{V}(t_{k+2}) & V(t_{k+2}) & 1 \\ & \vdots & \\ \dot{V}(t_{k+K}) & V(t_{k+K}) & 1 \end{bmatrix} \begin{bmatrix} R \\ E \\ P_0 \end{bmatrix} = \bar{X} \cdot \bar{\theta}$$

$$\hat{\bar{\theta}} = [\bar{X}^T \bar{X}]^{-1} \bar{X}^T \bar{Y}$$

where $P_{aw}(t)$ is the pressure measured at the airway opening, $\dot{V}(t)$ is the flow of air into and out of the patient's respiratory system (measured again at the airway opening), $V(t)$ is the net volume of air delivered to the patient (measured by integrating the flow signal over time), E is the elastance (inverse of the compliance C) , P_0 is a constant term to account for the pressure at the end of expiration (needed to balance the equation but not interesting per se), $\bar{\theta}$ is the vector of unknown parameters $[R \ \ E \ \ P_0]$, K is the number of samples collected during the 100ms post-occlusion period, and t_{k+1}, t_{k+2} .....t_{k+K} are the times (within the 100ms post-occlusion period) at which airway pressure and flow signals are sampled.

[0031] When estimating P_{mus} throughout the remaining portion of the breath at **18**, the values of R and C computed during the previous step are used in conjunction with airway pressure and flow waveforms to compute an estimate of P_{mus} throughout the remaining portion of the breath based on the standard equation of motion such that:

$$\hat{P}_{mus}(t) = P_{aw}(t) - \hat{R} \cdot \dot{V}(t) - \hat{E} \cdot V(t) - P_0 \quad \text{for } 200\,ms \le t \le t_{end} \qquad (2)$$

where t_{end} is the last available time sample (time at the end of the breath).

[0032] While the systems and methods discussed herein have been described with respect to certain embodiments, it is to be understood that said systems and methods are not limited to the disclosed embodiments and examples. To the contrary, described systems and methods are intended to cover various modifications and equivalent arrangements included within the spirit and scope of the described algorithm. For instance: The degree of the polynomial P_{mus} model used in step **14** of Figure 1 (occlusion period) can be different than 2. For instance, a first order polynomial model (i.e., a line) could also be used.

[0033] In another embodiment, the duration of step **16** (post-occlusion period) is not limited to being 100ms. A short duration is useful in order for the assumption of unaltered P_{mus} profile from step **14** to step **16** to be as valid as possible. In fact, the initiation of pressurization, provided by the ventilator after release of the airways occlusion, can induce changes in the patient's own P_{mus} drive via mechanical reflexes (e.g. Hering-Breuer reflex). However, the activation of such reflexes and the manifestation of their effects on P_{mus} may occur on a time scale that is larger than 100ms. On the other hand, a too short duration of step 16 may induce noise in the measurements that could compromise the LS procedure and lead to biased R and C estimates.

[0034] The final estimated P_{mus} profile does not necessarily need to be constructed by concatenating the three P_{mus} profiles obtained during steps **14**, **16**, and **18**, respectively. According to one embodiment, the values of R and E, which is essentially the inverse of C, from step **16** are used to compute the estimated P_{mus} profile over the entire breath according to:

$$\hat{P}_{mus}(t) = P_{aw}(t) - \hat{R} \cdot \dot{V}(t) - \hat{E} \cdot V(t) - P_0 \quad \text{for } 0 \leq t \leq t_{end} \qquad (3)$$

[0035] FIGURE 2 illustrates a graph **30** summarizing steps **14**, **16**, and **18** of the method of Figure 1. Exemplary results of the herein-described estimation algorithm have been generated using an in-vitro hydraulic model of the lungs. The in-vitro model was comprised of a rigid vessel within which an elastic balloon was placed. The balloon was characterized by a certain elastance value and its behavior was approximated as linear within a certain range of pressure values. The system was connected to a mechanical ventilator (e.g., Esprit, Philips-Respironics) via a linear resistor. The pressure within the vessel and external to the balloon was artificially controlled via an automated vacuum and compressed air system. Hence, a specific nominal P_{mus} profile can be generated externally to the balloon. The ventilator was then operated in Pressure Control Mode (note that any other suitable modes can be selected) and a $P_{0.1}$ maneuver was performed via the automatic software embedded in the ventilator. Pressure and flow measurements were collected via dedicated sensors placed at the Y connection between the ventilator and the in-vitro lung model.

[0036] FIGURE 3 illustrates exemplary results **40** from method of Figure 1 on one illustrative breath, where the estimated P_{mus} profile is compared against the gold-standard P_{mus} waveform measured inside the vessel. The level of agreement between the two waveforms is acceptable in this example (RMSE=0.7297) even though a certain degree of error can be observed. Furthermore, the R and E estimates provided by the algorithm ($\hat{R} = 24.85$, $\hat{E} = 56.84$) are very close to the corresponding gold standard values ($R_{gs} = 22.35$, $E_{gs} = 54.11$) computed via LS using the gold standard P_{mus} waveform.

[0037] The small error between the P_{mus}, R and E estimates and the corresponding gold standard values can be in part ascribed to the existence of a non-zero flow during the occlusion (see dashed arrow in Figure 3). This non-zero flow is essentially due to decompression of the air contained within the system when both the inhalation and the exhalation valves are closed. In fact, due to non-zero flow during the occlusion, Eq. (1) is no longer valid:

$$P_{aw}(t) \neq P_{mus}(t) \quad \text{for } 0 \leq t \leq 100\ ms$$

[0038] Hence, when a polynomial model of P_{mus} is fit to the airway pressure measurements during the occlusion period, error may be introduced as shown in the graph **50** of FIGURE 4. This error can be mitigated by reducing the length of the tubing circuit thus reducing the magnitude of the non-zero flow during the occlusion.

[0039] FIGURE 5 illustrates a system **60** that facilitates estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver in accordance with one or more aspects described herein. According to the embodiment of Figure 5, a patient **62** is connected to a ventilator **64** having one or more pressure sensors **63** and one or more flow sensors **65** in the patient circuit that respectively sense a characteristic pressure profile and a flow profile in the patient circuit. The ventilator is equipped with software and/or hardware configured to perform automatically a $P_{0.1}$ maneuver. The airway pressure and flow signals are measured in real-time; for example, volume can be computed by numerical integration of the flow signal. The system further comprises an estimation module **66** that includes a breath segmentation algorithm or module **68** that is used to isolate the current breath, starting from the moment at which the airways are occluded (e.g., at the beginning of patient's inspiratory activity) to the moment at which the exhalation is completed. To do so, specific flags from the ventilator (e.g. start of inspiration (SOI), start of expiration (SOE), etc.) are used. These flags contain the timestamps of the opening and closing of the inhalation and exhalation valves. The breath segmentation algorithm also divides the airflow and pressure data from the current breath into 3 different subsets related to the 3 regions identified in Figure 2, e.g.: 1) 100ms occlusion region; 2) 100ms post-occlusion region; 3) remaining portion of the breath. Flow and pressure data from the 3 segmented breath regions are provided as input to 3 estimation routines or modules, including a P_{mus} profile estimation routine or module **70**, an R and C estimation routine or module **72**, and a P_{mus} remainder-of-breath (ROB) routine or module **74** for estimating P_{mus} throughout the remaining portion of the breath. Each routine executes one of the 3 aforementioned estimation steps **14**, **16**, **18** of Figure 1 sequentially. Once the three steps have been executed, an estimate of $P_{mus}(t)$ over the entire breath can be computed by concatenating the 3 $P_{mus}(t)$ profiles computed during each step 14, 16, and 18. Finally, the R, C and $P_{mus}(t)$ estimated by the algorithm are provided as output. These can be displayed directly on the ventilator screen, or on a separate patient monitor.

[0040] The estimation algorithm **66** illustrated in the embodiment of Figure 5 can run on a ventilator processor or on a separate patient monitor. Additionally, the estimation algorithm **66** can run continually on a breath-by-breath basis, and a $P_{0.1}$ maneuver can be performed at every breath. This allows breath-by-breath updating of the estimated R and C and tracking of potential changes in patients' respiratory mechanics from one breath to the next one. Alternatively, the $P_{0.1}$ maneuver can be performed intermittently (e.g., at every X number of breaths, where X is an integer), while the values from the latest R and C estimation procedure are assumed valid over the next subsequent breaths and used to compute an estimate of $P_{mus}(t)$ based on the equation of motion (as in Eq. 2) until a new $P_{0.1}$ maneuver is performed.

[0041] The system further comprises a processor **76** that executes, and a memory **78** that stores, computer executable instructions for carrying out the various functions and/or methods described herein. The memory **78** may be a computer-readable medium on which a control program is stored, such as a disk, hard drive, or the like. Common forms of computer-readable media include, for example, floppy disks, flexible disks, hard disks, magnetic tape, or any other magnetic storage medium, CD-ROM, DVD, or any other optical medium, RAM, ROM, PROM, EPROM, FLASH-EPROM, variants thereof, other memory chip or cartridge, or any other tangible medium from which the processor **76** can read and execute. In this context, the described systems may be implemented on or as one or more general purpose computers, special purpose computer(s), a programmed microprocessor or microcontroller and peripheral integrated circuit elements, an ASIC or other integrated circuit, a digital signal processor, a hardwired electronic or logic circuit such as a discrete element circuit, a programmable logic device such as a PLD, PLA, FPGA, Graphics processing unit (GPU), or PAL, or the like.

[0042] FIGURE 6 shows a system **90** that facilitates estimating work of breathing (WOB) in a patient **62** hooked up to a ventilator **64** having one or more pressure sensors **63** and one or more flow sensor **65** and equipped with automated software for the $P_{0.1}$ maneuver. The P_{mus} output from the estimation algorithm **66** is used at WOB estimating step **92** to compute an estimate the Work of Breathing (WOB) by integrating the product between $P_{mus}(t)$ and $\dot{V}(t)$ over the inhalation phase of the current breath. From the computed WOB, Power of Breathing (POB) can also be attained by summing WOB over one minute. The estimated WOB/POB can ultimately be displayed on the ventilator screen or used internally by the ventilator as input to a closed-loop controller. The system further comprises a processor **76** that executes, and a memory **78** that stores, computer executable instructions for executing the various modules, algorithms, routines, etc. of Figure 6.

[0043] FIGURE 7 illustrates a system **100** that facilitates estimating work of breathing (WOB) and power of breathing (POB) in a patient **62** hooked up to a ventilator **64** with automated software for the $P_{0.1}$ maneuver wherein the ventilator is operating in Proportional Assist Ventilation (PAV) mode. The ventilator further comprises one or more pressure sensors **63** and one or more flow sensors **65** in the patient circuit that respectively sense a characteristic pressure profile and a flow profile in the patient circuit. The R and C values estimated by the algorithm **66** can be used to compute a desired airway pressure signal proportional to P_{mus} and drive the mechanical ventilator in PAV mode. The system further comprises a processor **76** that executes, and a memory **78** that stores, computer executable instructions for executing the various modules, algorithms, routines, etc. of Figure 6.

[0044] The innovation has been described with reference to several embodiments. Modifications and alterations may occur to others upon reading and understanding the preceding detailed description. It is intended that the innovation be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

CLAIMS:

1. A method for estimating respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver for a patient connected to a ventilator, comprising:

detecting patient inspiration onset;

automatically occluding the airway between the ventilator and the patient for a first predetermined time period responsive to the detecting step,

estimating a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion;

estimating resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period;

estimating a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of inspiration;

estimating $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles; and

outputting the estimated R and C values and the estimated P_{mus} profiles on a display.

2. The method according to claim 1, wherein estimating the first P_{mus} profile during the airway occlusion comprises fitting a first polynomial model of P_{mus} to the airway pressure measurements during the airway occlusion and estimating the first P_{mus} profile via a Least-Square (LS) technique.

3. The method according to claim 2, wherein estimating the second P_{mus} profile during the airway occlusion comprises extending the first polynomial model of P_{mus} in time.

4. The method according to any one of the preceding claims, wherein the first and second time periods are less than approximately 150ms in duration.

5. The method according any one of the preceding claims, wherein the first and second time periods are approximately 100ms in duration.

6. The method according to any one of the preceding claims, further comprising estimating a work of breathing (WOB) by integrating a product between $P_{mus}(t)$ and $\dot{V}(t)$ over an inhalation phase of a current breath.

7. The method according to any one of the preceding claims, wherein the ventilator (64) is operating in proportional assist ventilation (PAV) mode, and further comprising computing a desired airway pressure signal proportional to P_{mus} for driving the ventilator in PAV mode.

8. A processor (76) or computer-readable medium (78) having stored thereon computer-readable instructions for performing the method of any one of the preceding claims.

9. A system that facilitates estimating a patient's respiratory muscle pressure and respiratory mechanics using a $P_{0.1}$ maneuver, comprising:

a ventilator (64) having a pressure sensor and a flow sensor; and

one or more processors (76) in communication with the ventilator and configured to:

detect patient inspiration onset for a patient connected to the ventilator;

automatically occlude the airway of the patient for a first predetermined time period responsive to the detect,

estimate a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion;

estimate resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period;

estimate a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of inspiration;

estimate $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles; and

output the estimated R and C values and the estimated P_{mus} profiles on a display.

10. The system according to claim 9, wherein the one or more processors are further configured to estimate the first P_{mus} profile during the airway occlusion by fitting a polynomial

model of P_{mus} to the airway pressure measurements during the airway occlusion and estimate the initial inspiratory P_{mus} profile via a Least-Square (LS) technique.

11. The system according to either of claims 9 or 10, wherein the first and second time periods are less than approximately 150ms in duration.

12. The system according to any one of claims 9-11, wherein the first and second time periods are approximately 100ms in duration.

13. The system according to any one of claims 9-12,wherein the one or more processors are further configured to estimate a work of breathing (WOB) by integrating a product between $P_{mus}(t)$ and $\dot{V}(t)$ over an inhalation phase of a current breath.

14. The system according to any one of claims 9-13, wherein the ventilator is operating in proportional assist ventilation (PAV) mode.

15. The system according to claim 14, he one or more processors are further configured to compute a desired airway pressure signal proportional to P_{mus} for driving the ventilator (64) in PAV mode.

16. A processor (76) configured to execute computer-executable instructions for estimating respiratory muscle pressure and respiratory mechanics of a patient connected to a ventilator (64) using a $P_{0.1}$ maneuver, the instructions comprising:

detecting processor patient inspiration onset for;

automatically occluding the airway of the patient for a first predetermined time period responsive to the detecting step,

estimating a first respiratory muscle pressure (P_{mus}) profile during the airway occlusion;

estimating resistance (R) and compliance (C) values and a second P_{mus} profile generated during a second predetermined time period;

estimating a third P_{mus} profile during a third predetermined time period that extends from the end of the second predetermined time period until the end of inspiration;

estimating $P_{mus}(t)$ over an entire breath by concatenating the first, second and third P_{mus} profiles; and

outputting the estimated R and C values and the estimated P_{mus} profiles on a display.

17. The processor (76) according to claim 16, wherein estimating the first P_{mus} profile during the airway occlusion comprises fitting a polynomial model of P_{mus} to the airway pressure measurements during the airway occlusion and estimating the initial inspiratory P_{mus} profile via a Least-Square (LS) technique.

18. The processor (76) according to either one of claims 16 or 17, wherein the first and second time periods are greater than approximately 50ms and less than approximately 150ms in duration.

19. The processor (76) according to any one of claims 16-18, the instructions further comprising estimating a work of breathing (WOB) by integrating a product between $P_{mus}(t)$ and $\dot{V}(t)$ over an inhalation phase of a current breath.

20. The processor (76) according to any one of claims 16-19, the instructions further comprising computing a desired airway pressure signal proportional to P_{mus} for driving the mechanical ventilator in proportional assist ventilation (PAV) mode when the ventilator is operating in PAV mode.

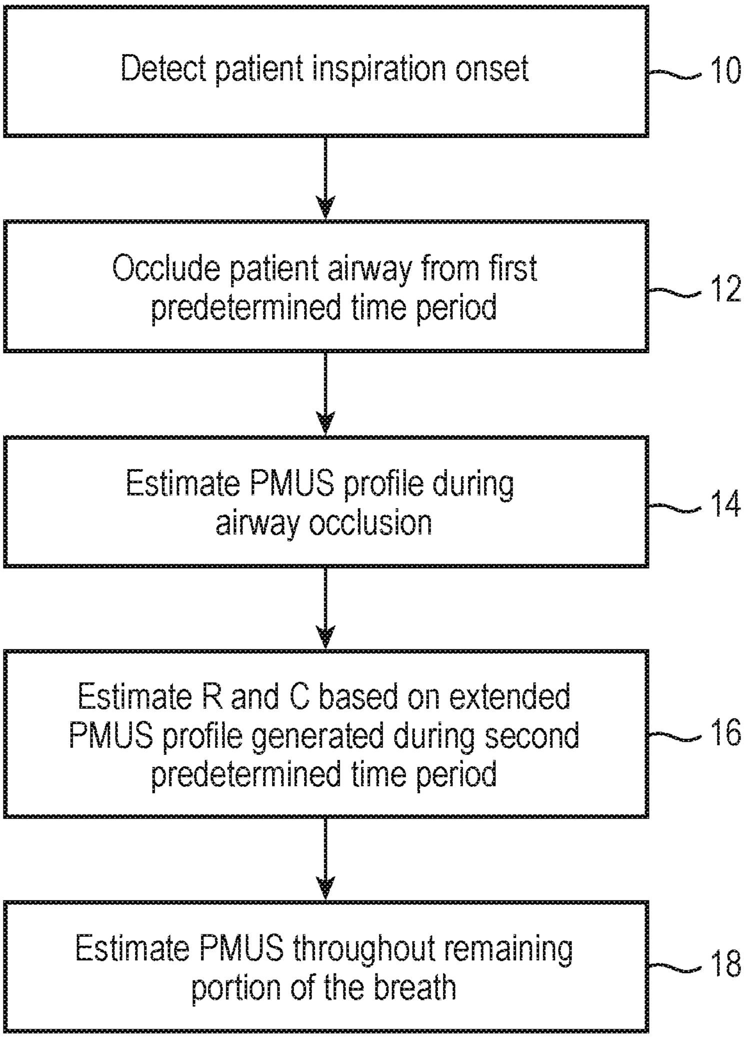
Detect patient inspiration onset
10
Occlude patient airway from first predetermined time period
12
Estimate PMUS profile during airway occlusion
14
Estimate R and C based on extended PMUS profile generated during second predetermined time period
16
Estimate PMUS throughout remaining portion of the breath
18

FIG. 1

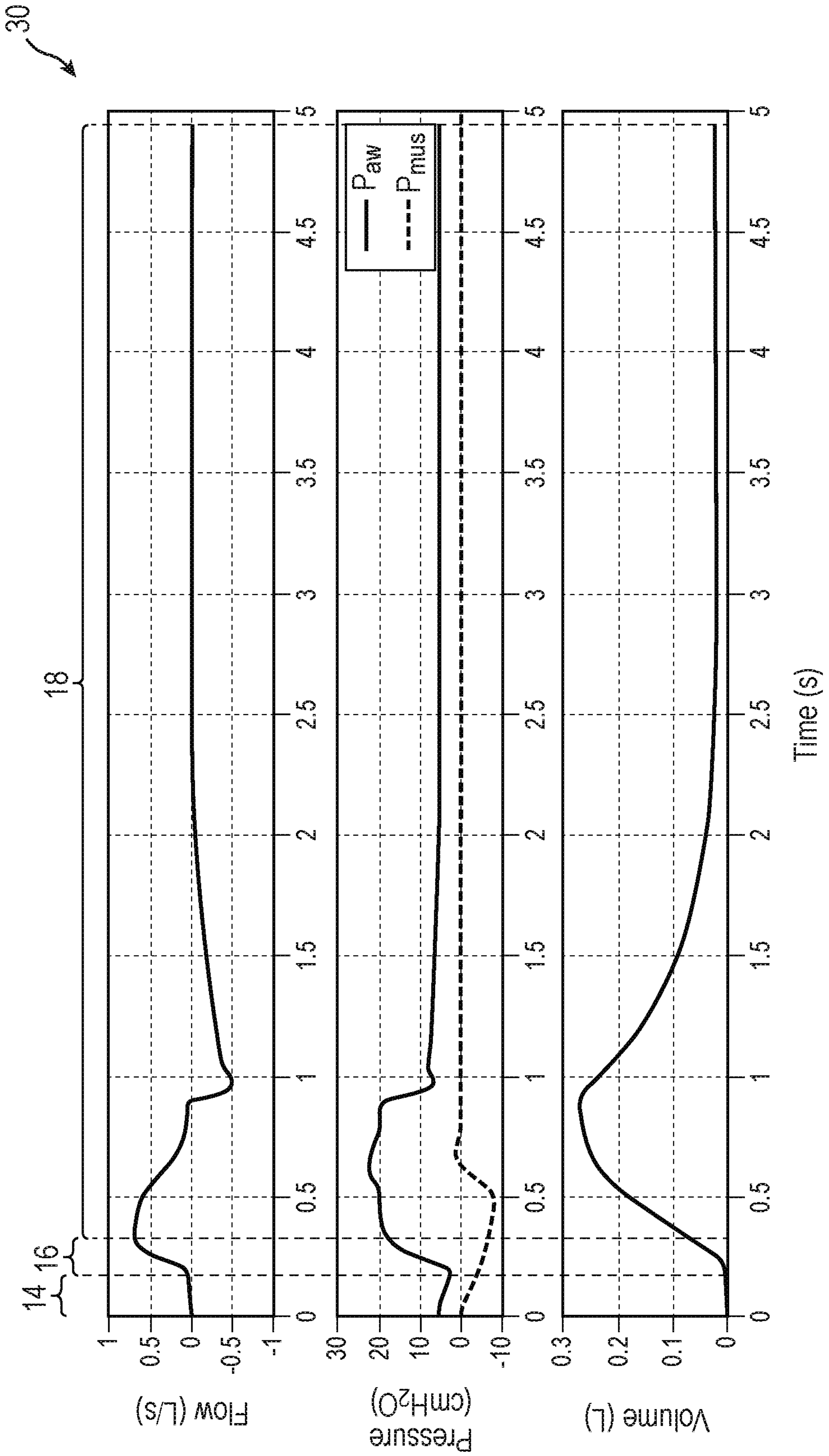
30
14
16
18
Flow (L/s)
1
0.5
0
-0.5
-1
Pressure (cmH$_2$O)
30
20
10
0
-10
P_{aw}
P_{mus}
Volume (L)
0.3
0.2
0.1
0
0
0.5
1
1.5
2
2.5
3
3.5
4
4.5
5
Time (s)

FIG. 2

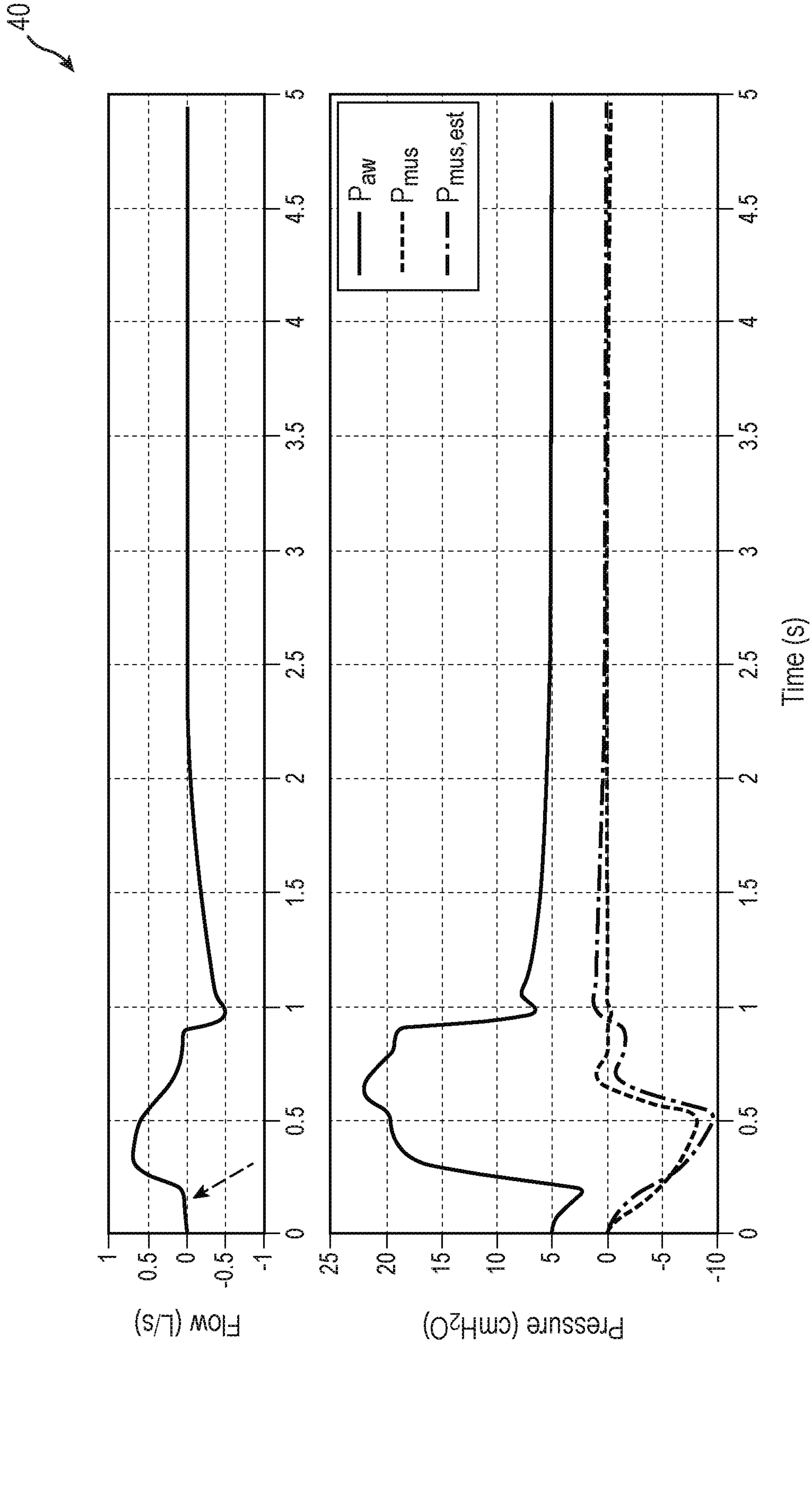
40
Flow (L/s)
Pressure (cmH2O)
Time (s)
Paw
Pmus
Pmus,est
1
0.5
0
-0.5
-1
25
20
15
10
5
0
-5
-10
0
0.5
1
1.5
2
2.5
3
3.5
4
4.5
5

FIG. 3

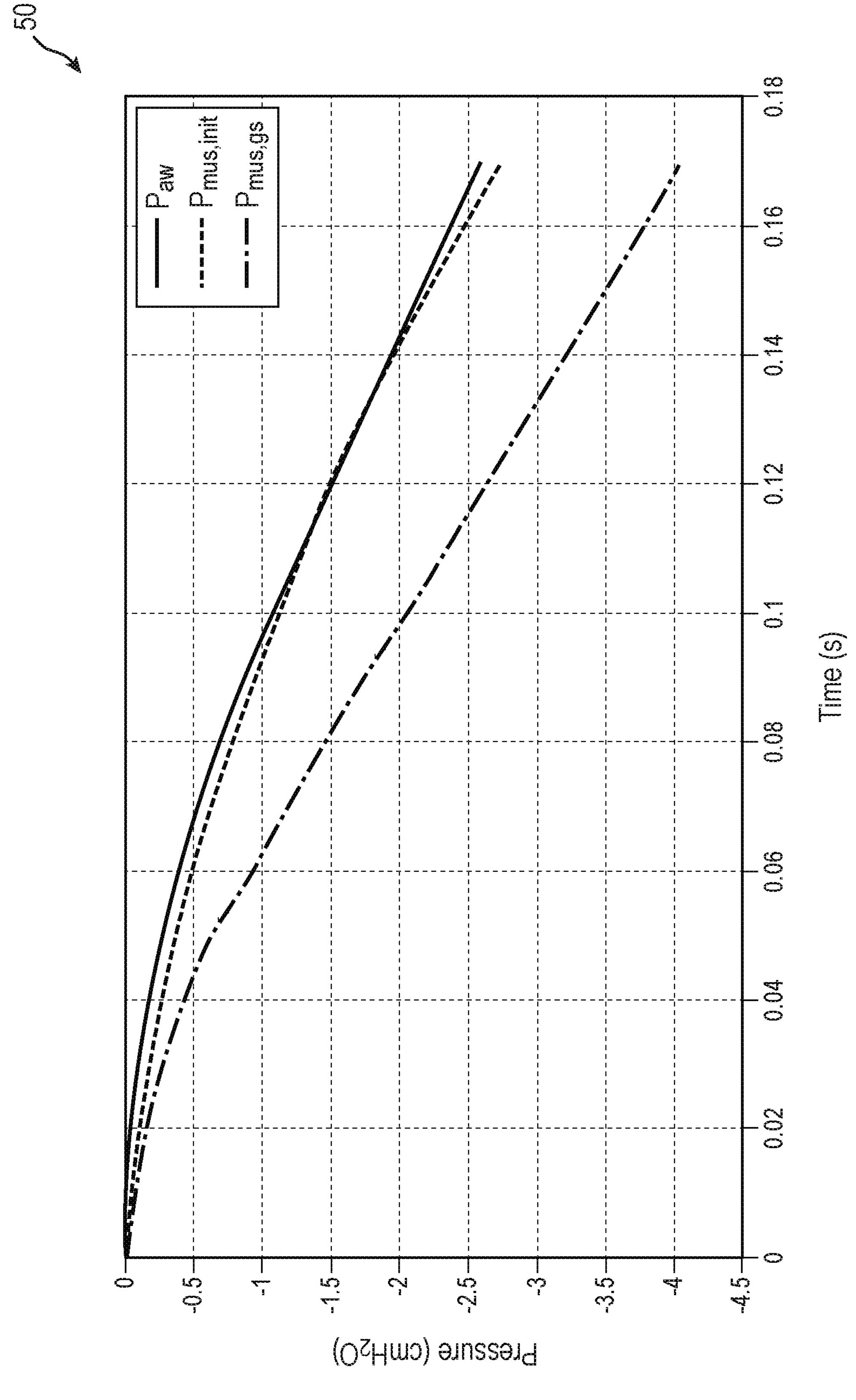
50
P_{aw}
$P_{mus,init}$
$P_{mus,gs}$
Pressure (cmH_2O)
0
-0.5
-1
-1.5
-2
-2.5
-3
-3.5
-4
-4.5
Time (s)
0
0.02
0.04
0.06
0.08
0.1
0.12
0.14
0.16
0.18

FIG. 4

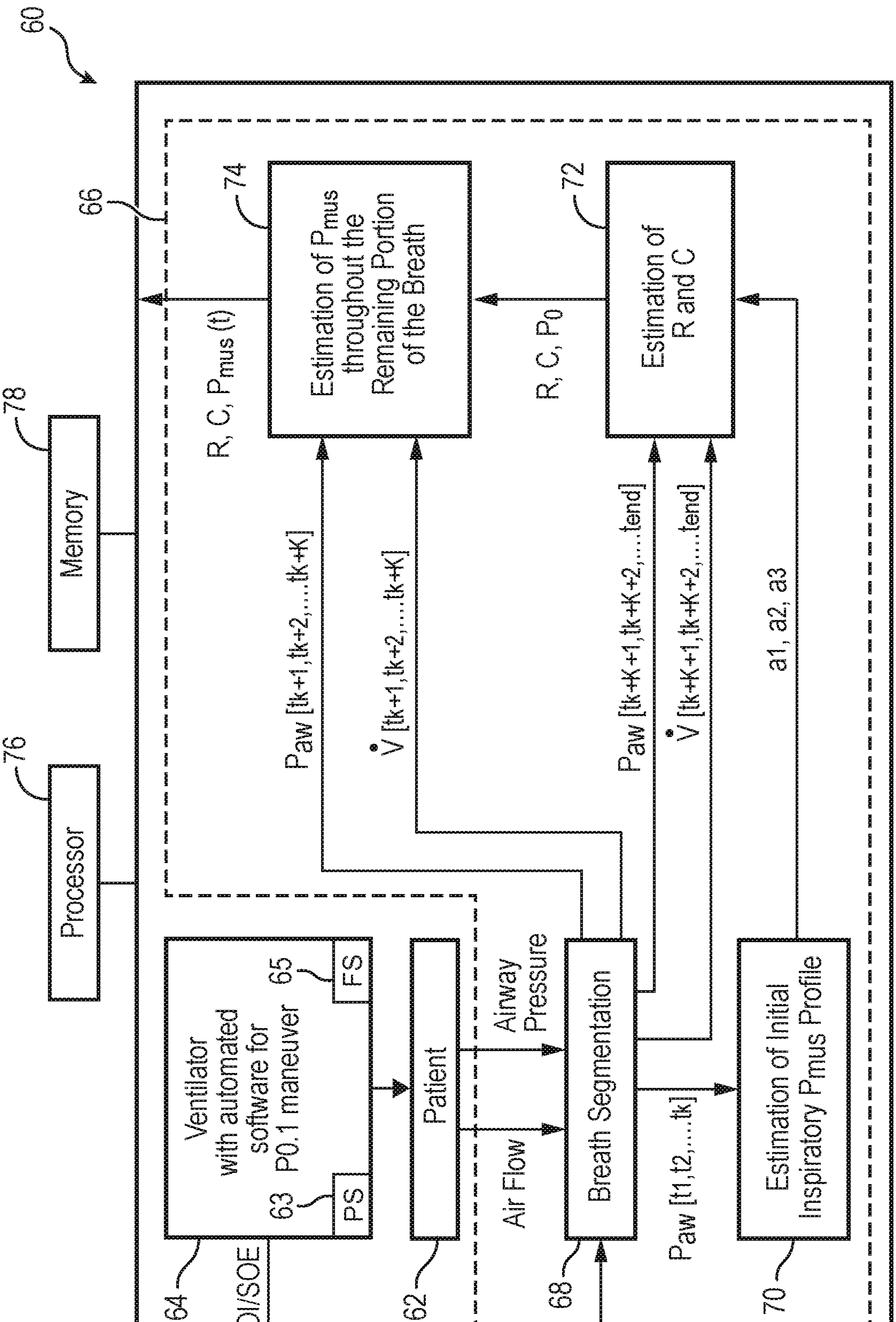
60
76
Processor
78
Memory
66
74
Estimation of P_mus throughout the Remaining Portion of the Breath
R, C, P_mus (t)
72
Estimation of R and C
R, C, P_0
Paw [tk+1, tk+2,....tk+K]
V̇ [tk+1, tk+2,....tk+K]
Paw [tk+K+1, tk+K+2,....tend]
V̇ [tk+K+1, tk+K+2,....tend]
a1, a2, a3
Ventilator with automated software for P0.1 maneuver
63
PS
65
FS
64
SOI/SOE
Patient
62
Airway Pressure
Air Flow
Breath Segmentation
68
Paw [t1, t2,....tk]
Estimation of Initial Inspiratory Pmus Profile
70

FIG. 5

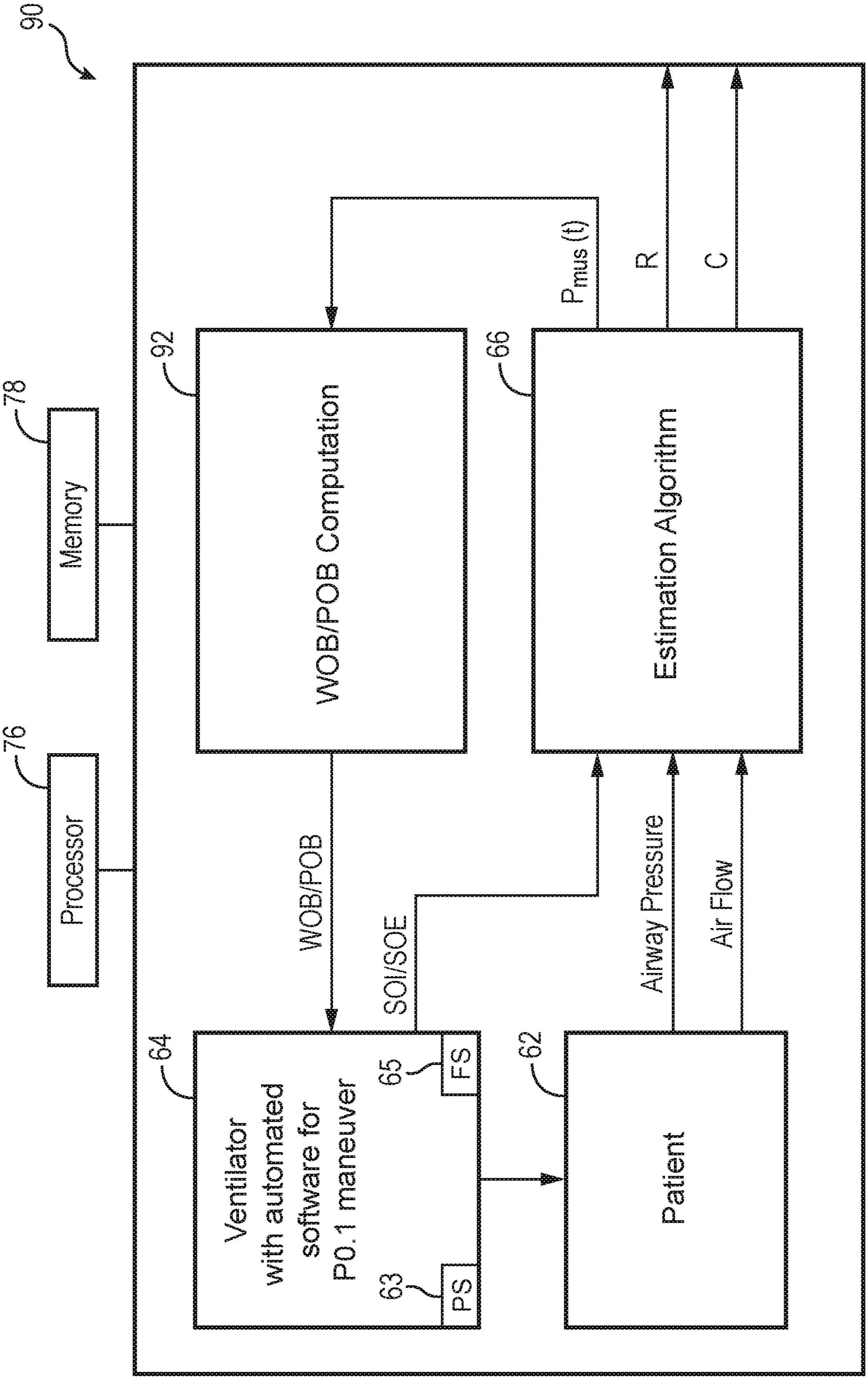
90
78
Memory
76
Processor
92
WOB/POB Computation
WOB/POB
64
Ventilator
with automated
software for
P0.1 maneuver
65
FS
63
PS
SOI/SOE
P_{mus} (t)
R
C
66
Estimation Algorithm
62
Patient
Airway Pressure
Air Flow

FIG. 6

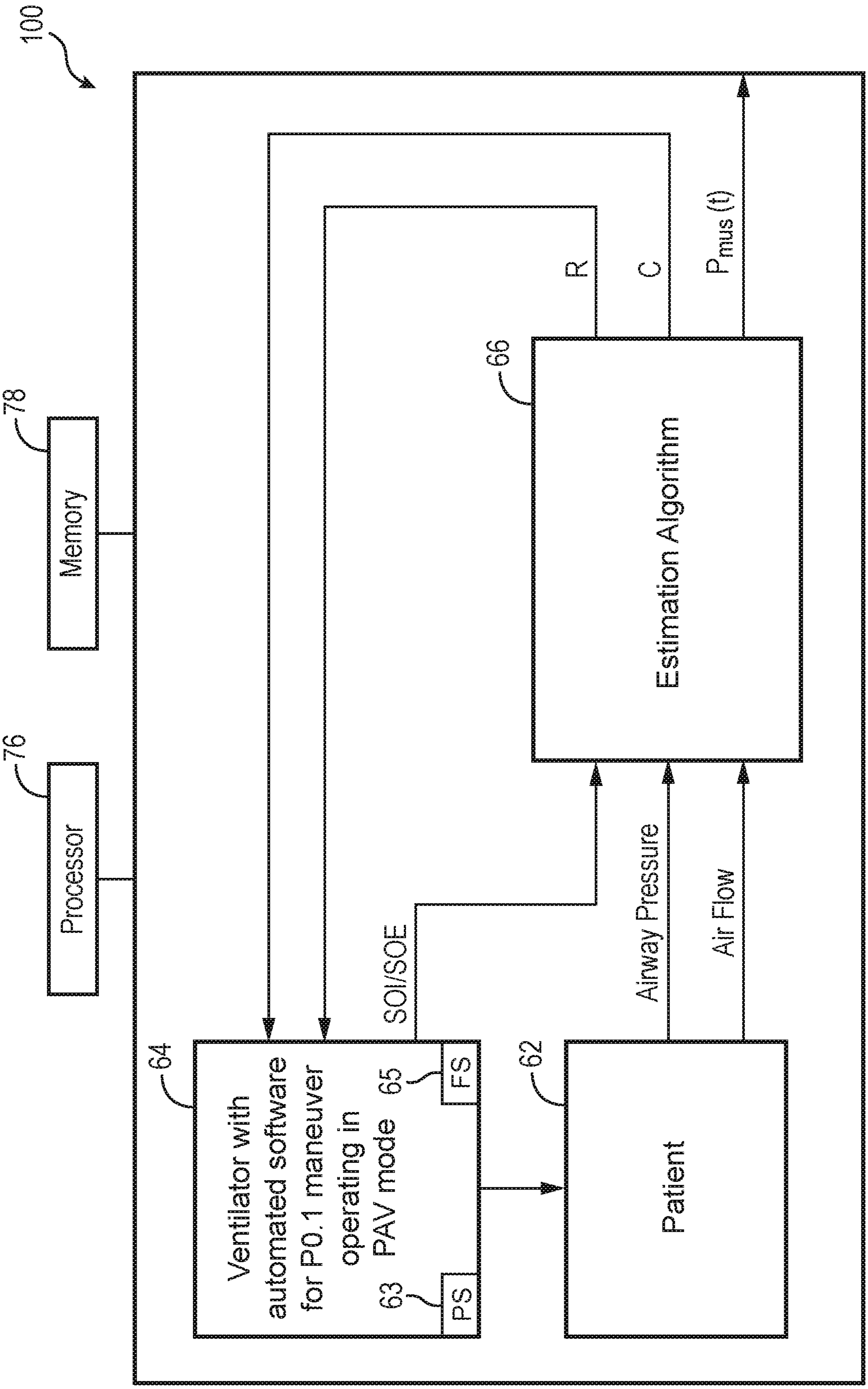
100
Processor
76
Memory
78
Ventilator with automated software for P0.1 maneuver operating in PAV mode
64
PS
63
FS
65
Patient
62
Estimation Algorithm
66
SOI/SOE
Airway Pressure
Air Flow
R
C
P_{mus} (t)

FIG. 7

INTERNATIONAL SEARCH REPORT

International application No
PCT/IB2017/056562

A. CLASSIFICATION OF SUBJECT MATTER
INV. A61B5/085 A61B5/03 A61M16/00
ADD. A61B5/087 A61B5/00

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
A61B A61M

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

EPO-Internal, WPI Data, BIOSIS, COMPENDEX, EMBASE, INSPEC

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	EP 1 972 274 A1 (DRAEGER MEDICAL AG [DE]) 24 September 2008 (2008-09-24)	1-5, 7-12, 14-18,20
Y	paragraph [0006] - paragraph [0008] paragraph [0012] - paragraph [0015] paragraph [0017] - paragraph [0022] figures -----	6,13,19
X	US 5 876 352 A (WEISMANN DIETER [DE]) 2 March 1999 (1999-03-02)	1
A	column 3, line 15 - column 4, line 11 column 5, line 11 - column 7, line 14 figures -----	2-20

-/--

[X] Further documents are listed in the continuation of Box C. [X] See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier application or patent but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
12 January 2018	19/01/2018

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Görlach, Tobias

1

INTERNATIONAL SEARCH REPORT

International application No
PCT/IB2017/056562

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indioation, where appropriate, of the relevant passages	Relevant to olaim No.
Y	Antonio Albanese: "Physiology-based Mathematical Models for the Intensive Care Unit: Application to Mechanical Ventilation", , 1 January 2014 (2014-01-01), XP055332137, ISBN: 978-1-303-94555-7 Retrieved from the Internet: URL:https://academiccommons.columbia.edu/download/fedora_content/download/ac:199653/CONTENT/Albanese_columbia_0054D_12039.pdf [retrieved on 2017-01-04]	6,13,19
A	page 101 - page 155	1-5, 7-12, 14-18,20

1

INTERNATIONAL SEARCH REPORT

Information on patent family members

International application No

PCT/IB2017/056562

Patent document cited in search report		Publication date	Patent family member(s)		Publication date
EP 1972274	A1	24-09-2008	EP 1972274	A1	24-09-2008
			US 2008234595	A1	25-09-2008
US 5876352	A	02-03-1999	DE 19808543	A1	19-11-1998
			US 5876352	A	02-03-1999

patsnap

专利名称(译)	使用p0.1演算估算呼吸肌压力和呼吸力学的系统和方法		
公开(公告)号	EP3544502A1	公开(公告)日	2019-10-02
申请号	EP2017794086	申请日	2017-10-23
[标]申请(专利权)人(译)	皇家飞利浦电子股份有限公司		
申请(专利权)人(译)	皇家飞利浦N.V.		
当前申请(专利权)人(译)	皇家飞利浦N.V.		
[标]发明人	ALBANESE ANTONIO BUIZZA ROBERTO VICARIO FRANCESCO		
发明人	ALBANESE, ANTONIO BUIZZA, ROBERTO VICARIO, FRANCESCO		
IPC分类号	A61B5/085 A61B5/03 A61M16/00 A61B5/087 A61B5/00		
CPC分类号	A61B5/03 A61B5/085 A61B5/087 A61B5/4836 A61M16/00 A61M16/0866 A61M2230/46 A61M16/022 A61M2016/0015 A61M2016/0039		
代理机构(译)	德哈恩波尔ERIK		
优先权	62/412927 2016-10-26 US		
外部链接	Espacenet		

摘要(译)

当使用P0.1操纵估计呼吸肌压力和呼吸力学时，检测连接到呼吸机（64）的患者的患者吸气开始，并且患者的气道被阻塞第一预定时间段。在气道阻塞期间估计第一呼吸肌压力（Pmus）曲线。然后估计在第二预定时间段期间产生的电阻（R）和顺应性（C）值和第二Pmus曲线。在第三预定时间段期间估计第三Pmus轮廓，该第三预定时间段从第二预定时间段的结束延伸到吸气结束。通过连接第一，第二和第三Pmus轮廓来估计整个呼吸的Pmus（t），并且在显示器上输出估计的R和C值以及估计的Pmus轮廓。