SYSTEM AND METHOD FOR DETECTING OF PULMONARY DISEASES

Field of the Invention

[0001] The present invention generally relates to systems and method for detecting of pulmonary diseases, and more particularly to non-invasive systems and methods for detecting of pulmonary diseases by analyzing inhalation and exhalation.

Background of the Invention

[0002] Lung is a vital organ allowing exchanges of O_2 and CO_2 between the alveolar sockets and bloods. When the lung is impaired, e.g., viral or bacterial infections, its physical appearance and/or functions will be altered, resulting in pulmonary diseases. The current detection of pulmonary diseases is mainly done by X-ray. However, the X-ray is not only invasive, but also only able to detect the physical alterations of lungs. Thus, the X-ray is not applicable for detecting pulmonary diseases at early stages. In addition, there are a broad range of so-called pulmonary function tests. For example, spirometry measures how well the lungs exhale. Lung volume measurement permits detection of restrictive lung diseases. Testing the diffusion capacity permits as estimate of how efficiently the lungs are able to transfer oxygen from the air into the bloodstream. However, all the available function tests are mere measures of systemic metabolism rather than of pulmonary functions.

[0003] It is well known that the components of exhalation could be utilized for diagnosis of certain diseases. For example, U.S. Pat. 4,823,803 discloses a device for testing human exhalation for halitosis by using sensors that are sensitive to malodorant gases of predetermined chemical compositions for producing signals variable with the detected concentrations of the malodorant gases. However, this patent is directed to the detection of malodorant gases rather than the lung functions.

[0004] There are situations in which an early detection of viral infection is critical for treatment of the victim and more importantly for the control of the epidemic. For example, the recent epidemic of SARS (Severe Acute Respiratory Syndrome) has raised concerns about effective detection of such relevant diseases, especially at a much earlier

stage of infection in human bodies. Upon early detection, scarce and precious medical resources can be focused on the infected persons. In the early stages of the infections, the X-ray screening may not be useful because the pathogenic damages on the lungs may not be evident enough to be detected by the machine. The current available function tests may not be able to detect the early signs of infections because the functions tests measure the systemic metabolism only. While PCR is powerful in detecting the early infection, it is prone to mutations. Massive mutations will handicap this technique severely.

[0005] Therefore, there is an imperative need to develop non-invasive systems and methods for detecting pulmonary diseases, especially ones inflicted by viral or bacterial infections. Furthermore, the detection is independent of the mutations of infectious agents. This invention satisfies this need by disclosing systems and methods of detecting pulmonary diseases by analyzing of the inhalation and exhalation of a test person. Other advantages of this invention will be apparent with reference to the detailed description.

Summary of the Invention

[0006] The present invention provides non-invasive systems and methods for detecting of pulmonary diseases by analyzing inhalation and exhalation. In one aspect, the present invention provides a pulmonary disease detection system for detecting breathing deficiencies of a test person. The pulmonary disease detection system comprises a breath analyser for analysing the inhale and exhale airs of the test person, a computer processor for receiving from the breath analyser the information and processing the received analysis data to give values to different parameters of the inhale and exhale airs of the test person, and a medical database for storing different parameters of breaths of the public and normal ranges for healthy persons; thereby the computer processor compares the values of different parameters of the inhale and exhale airs of the test person with the ones stored in the medical database so as to yield a test result of whether the test person is suffering breath deficiencies. In one embodiment, the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the breathing rate, oxygen consumption rate and carbon dioxide generation rate from the composition information of the inhale and exhale airs of the test person; thereby comparing the rates of the test person with the ones stored in the medical database so as to

conclude whether the test person is suffering any breath deficiencies. In another embodiment, the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the composition values of the exhale air from the composition values of the inhale air and the assumed normal rates including breathing rate, oxygen consumption rate and carbon dioxide generation rate; thereby comparing the composition values of the exhale air with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies. In yet another embodiment, the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the volume compliance and air-flow resistance of the test person from the composition values and pressure values derived from the inhale air rate and exhale air rate; thereby comparing the volume compliance and air-flow resistance values with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

[0007] In another aspect, the present invention provides a method of detecting a pulmonary disease of a test person by investigating the breath deficiencies of the test person. The method comprises acquiring the information of inhale and exhale airs of the test person, processing the acquired information to give values of designated aspects of the inhale and exhale airs of the test person, and comparing the calculated values with the ones stored in a medical database so as to conclude whether the test person is suffering any pulmonary diseases. In one embodiment, the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the breathing rate, oxygen consumption rate and carbon dioxide generation rate from the composition information of the inhale and exhale airs of the test person; thereby comparing the rates of the test person with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies. In another embodiment, the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the composition values of the exhale air from the composition values of the inhale air and the assumed normal rates including breathing rate, oxygen consumption rate and carbon dioxide generation rate; thereby comparing the composition values of the exhale air with the ones stored in the medical

database so as to conclude whether the test person is suffering any breath deficiencies. In yet another embodiment, the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the volume compliance and air-flow resistance of the test person from the composition values and pressure values derived from the inhale air rate and exhale air rate; thereby comparing the volume compliance and air-flow resistance values with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

[0008] The objectives and advantages of the invention will become apparent from the following detailed description of preferred embodiments thereof in connection with the accompanying drawings.

Brief Description of the Drawings

[0009] Preferred embodiments according to the present invention will now be described with reference to the Figures, in which like reference numerals denote like elements.

[0010] FIG 1 is a block diagram of the pulmonary disease detection system (PDDS) in accordance with one embodiment of the present invention.

[0011] FIG 2 shows a breath analyzer configured in accordance with one embodiment of the present invention.

[0012] FIG 3 shows a flowchart of detecting pulmonary diseases on the basis of altered oxygen consumption and carbon dioxide generation in accordance with one embodiment of the present invention.

[0013] FIG 4 shows another functional flowchart of detecting pulmonary diseases on the basis of lung-volume compliance and airflow resistance in accordance with another embodiment of the present invention.

[0014] FIG 5a shows a diagram illustrating the directional relationships among the different pressures that may be detected or deduced from the air flow rates detected by the breath analyzer as shown in FIG 2.

[0015] FIG 5b shows that different pressures are functions of time.

[0016] FIG 6 shows one functional configuration of the disease identification software system and databases in accordance with one embodiment of the present invention.

[0017] FIG 7 shows a flowchart illustrating the performance of the disease identification software system as shown in FIG 6.

Detailed Description of the Invention

[0018] The present invention may be understood more readily by reference to the following detailed description of certain embodiments of the invention.

[0019] Throughout this application, where publications are referenced, the disclosures of these publications are hereby incorporated by reference, in their entireties, into this application in order to more fully describe the state of art to which this invention pertains.

[0020] The present invention provides systems and methods for detecting pulmonary diseases. While there are provided more details about the systems and methods hereinafter, it is to be appreciated that the present systems and methods are based on the understanding that anyone developing any pulmonary diseases would demonstrate certain detectable breathing deficiencies. The deficiencies may be manifested by the changes of the oxygen consumption and carbon dioxide generation, or the differences of lung compliance and airflow-resistance.

[0021] There is provided a block diagram of the pulmonary disease detection system (PDDS) as shown in FIG 1 in accordance with one embodiment of the present invention. The PDDS **100** comprises a breath analyzer **102**, a computer processor **103**, and a medical database **104**. The breath analyzer **102** will take in the breath from a test person **101** and output the information of components of the breath from the person to the computer processor **103**. The computer processor **103** contains algorithms for manipulating the information of the breath and comparing the manipulated results with the medical database **104**, so that the computer processor provides the results of diagnosis **105**.

[0022] The breath analyzer **102** may be any apparatus that can obtain breathing information from a test person that is sufficient for the application of the algorithms embedded in the computer processor **103**. For example, for the application of an algorithm

based on oxygen consumption and carbon dioxide generation, the breath analyzer **102** basically comprises of gas sensors such as sensors for oxygen, carbon dioxide and water vapor. The compositions of both inhale and exhale gases are analyzed. Then the respective compositions are further processed by the computer processor **103**.

[0023] FIG 2 shows a breath analyzer **102** configured in accordance with one embodiment of the present invention. The breath analyzer **102** comprises a mask **1**, a data acquisition unit **7**, and an air tank **11**. The mask **1** is configured to cover the nose and mouth of a test person so that maximum fresh air is delivered to the test person and minimum exhaled air is lost before proper measurement is completed. As shown in FIG 2, the mask **1** includes an air outlet membrane **2** as a seal for preventing air within the mask from leaking; an exhaust flip valve **3** that will open to allow all the expired air to flow out **4** of the mask when the test person breaths out; an air inlet membrane **21** as a seal for preventing air within the mask from leaking; an inhale flip valve **22** that will open to allow the fresh air from the air tank **11** to flow in **23** when the test person breaths in; an oxygen electrode **25** for detecting the oxygen in the air composition; a carbon dioxide gas electrode **24** for detecting the carbon dioxide in the air composition; a nitrogen gas electrode **17** for detecting the nitrogen in the air composition; a water vapor electrode **18** for detecting the water vapor in the air composition; an inspired air flow rate electrode **19** for determining the flow rate of exhale air; an expired air flow rate electrode **20** for determining the flow rate of exhale air; and signal conductors **5** that transmit the information from the electrodes to the data acquisition unit **7**. The signal conductors may be ultra-low impedance conductor or fiber optic.

[0024] The air tank **11** contains pressurized air so as to ensure a measurable and controllable air supply to the test person. An air delivery pipe **16** connects the air tank with the mask so as to deliver a stream of controlled air from the air tank to the mask. The air tank **11** also includes four electrodes **12, 13, 14, 15** for detecting the oxygen, carbon dioxide, nitrogen, and water vapor respectively. The signal conductors **10** transmit the data from the four electrodes to the data acquisition unit **7**. The signal conductors may be ultra-low impedance conductor or fiber optic.

[0025] The data acquisition unit **7** includes connectors **6, 9** for connecting to the signal conductors **5, 10** so that it will receive all the information from the air tank and the mask. Then the data acquisition unit **7** transmits the received signals to the computer

processor **103** which acts as the central processing unit (CPU). The transmitted information may be digitalized packets.

[0026] After the CPU receives the data of the air compositions from the data acquisition unit **7**, it will process the air composition data. In one aspect of the present invention, the detection of pulmonary diseases is based on the understanding that anyone developing any pulmonary diseases would demonstrate certain detectable breathing deficiencies. The inventors of the present invention further discovered that the breathing deficiencies are manifested by altered oxygen consumption and carbon dioxide generation. FIG 3 shows a flowchart of detecting pulmonary diseases on the basis of altered oxygen consumption and carbon dioxide generation.

[0027] Referring now to FIG 3, when the PDDS **100** starts **301**, it obtains through the data acquisition unit **7** the information including oxygen composition, carbon dioxide composition, nitrogen composition, water vapor composition, and air flow rate **302**. It is to be noted that the air flow rate data will be discussed hereinafter when the air flow rate will be used to calculate the volume compliance and air-flow resistance in another algorithm of the present invention. Then the CPU will calculate the overall relatives of all gases compositions **303**. Then the processed data is searched against the stored database to determine whether the test person has pulmonary diseases **304**. Then diagnostic results will be outputted **305** and the operation comes to an end **306**. It is noted from FIG 3 that the stored database is continuously updated so that the database will become more useful when more data is collected.

[0028] Now there is provided a more detailed description of detecting pulmonary diseases on the basis of altered oxygen consumption and carbon dioxide generation. The basic assumption is that the composition of expired air from a patient such as a person with SARS infection is different from that of a normal person. It is further assumed that with an expired air, (a) its O_2 content (or % vol.) will be greater (because of less O_2 consumed from alveoli) and closer to that of inspired air; (b) its CO_2 content will be lesser, and more akin to that of inspired air; and (c) the transfer coefficients for O_2 & CO_2 will be lesser as compared to a medically normal person. Therefore, the mass balance analysis involves (i) compositions of air breathed in and out; and (ii) consumption or generation of O_2, CO_2 and H_2O.

[0029] For calculation of inhale and exhale compositions, there are a few general assumptions: (1) Breathing Rate (BR) = 12 breaths/min; (2) P_{H2O} at 37°C = 47mmHg; (3) O_2 metabolic consumption rate at (at BTP) = 284 m*l*/min; and (4) CO_2 production rate (at BTP) = 227 m*l*/min. Thus, the expected compositions of the expired air can be calculated from the atmospheric air or vice versa. For example, as shown in Table 1, the expected expired air compositions can be calculated from the numbers of the atmospheric air column:

[0030] N_2 = 393.1 m*l*

[0031] O_2 = 104.2-(284/12) = 80.53 m*l*

[0032] CO_2 = 0.2+(227/12) = 19.12 m*l*

[0033] Total = 492.75 m*l* (1)

[0034] Ratio of water vapor/dry gas in the expired air = 49.5mmHg/(760-47) mmHg.= 49.5/713 = 6.94% (2)

[0035] Volume of water vapor in the Expired air = (1)x(2)=492.75 0.0694=34.21*ml* (3)

[0036] Total Expired air = [1]+[3]=492.75+34.21=526.96 *ml* (4)

[0037] Thus the percentage of the gases components in the expired air can be calculated as follows:

[0038] N_2 = 393.1/526.96 = 74.6%;

[0039] O_2 = 80.52/526.96 = 15.28%;

[0040] CO_2 = 19.12/526.96 = 3.63%; and

[0041] H_2O = 34.21/526.96 = 6.49%.

[0042] All the numbers of the atmospheric air and expired air are presented in Table 1.

[0043] Table 1. An exemplary air compositions

[0044]

Respiratory gases	Atmospheric Air		Humidified Air		Alveolar Air		Expired Air	
	mmHg	m*l*/%	mmHg	m*l*/ %	mmHg	m*l*/%	mmHg	m*l*/%
N_2	597	393.1 78.62%	563.3	117.8 78.5%	569	275.3 74.9%	597.5	393.1 78.62
O_2	159	104.2 20.84%	149.3	31.3 20.9%	104	49.3 13.6%	84	55.3 11.06
CO_2	0.3	0.2 0.04%	0.3	0.09 0.06%	40	19 5.3%	29	19.1 3.82
H_2O	3.7	2.5	47.0	0.81	47	22.7	49.5	32.5

		0.5%		0.54%		6.2%		6.5
Total	760	500 100%	760	150 100%	760	366.3 100%	760	500 100%

[0045] So far it has been shown that if the rates of oxygen consumption and carbon dioxide generation are known, the ideal composition of an expired air can be calculated from the original composition of the atmospheric air. This is important for initializing the stored databases because in the early stages, the PDDS may have to generate part of the database by calculating the components of the expired airs from the atmospheric airs based on certain assumptions. With the gradual accumulation of the database, more and more actual data will supplement or substitute the calculated ones. As discussed early, the breath analyzer **102** of the present invention can acquire the data of individual gas components of the expired air from the test person and the CPU can manipulate the data to give the percentages of each gas component as to the expired air. The processed data of each gas components will be compared with the stored medical database. If the test person suffers from any illness that affects pulmonary functions, it is expected that the processed data of the expired air from a test person will be deviated from that of the stored database for a normal person. If this is so, the CPU will output the diagnostic results showing that the test person is probably having pulmonary diseases. Then the test person can seek further examinations to determine which kind of pulmonary diseases he/she is developing.

[0046] As shown in Table 1, it is apparent that the rates of oxygen consumption and carbon dioxide generation may be derived from the actual measurements of individual gas components of the atmospheric air and expired air. It is noted that all of the data of individual gas components can be obtained by the breath analyzer as shown in FIG 2. The calculated rates can be compared with the stored database to determine whether the test person is suffering any pulmonary diseases.

[0047] As mentioned earlier, the choice of a breath analyzer **102** for the present PDDS is limited by its function only. For example, when the lung compliance and airflow-resistance are used for detecting pulmonary diseases, the breath analyzer **102** may be a commercially available spirometer. Now referring to FIG 4, there is provided another functional flowchart of detecting pulmonary diseases in accordance with another embodiment of the present invention. When the PDDS starts **400**, it obtains data from the breath analyzer **401** (the same as the one shown in FIG 2) to determine the volume breath

characteristics of the test person **402**, and further extract the parameters of P0, P1, P2, Ra, Ca and W **403**. If the extracted parameters are not the best filtered results **404**, the program will go back to step **403** to try to get the best parameters. If the extracted parameters are the best filtered results **404**, then the VTI1, VTI2 and VTI3 will be determined **405**. Then the program will conduct diagnosis **407** by utilizing the medical databases **406, 408**, and output the results **409**, resulting the end of the program **410**.

[0048] FIG 5a shows a diagram illustrating the directional relationships among the different pressures that may be detected or induced from the air flow rates detected by the breath analyzer as shown in FIG 2. FIG 5b shows that different pressures are functions of time. From there are derived a few fundamental equations that are the foundation for the algorithm shown in FIG 4:

[0049]

$$(i)\ (P_a - P_p) - P_{el} = 0$$
$$(ii)\ P_{el} = (2\ a\ h) / R\ r = 2\ T / r = V/C + P_{el,\ 0}$$
$$(iii)\ (P_o - P_a) = R(dV/dt)$$
$$(iv)\ P_L = Po - Pp$$
$$(v)\ R(dV/dt) + VIC = P_L - P_{el,\ 0}$$

[0050] Now there is provided a more detailed description of determination of the volume breath characteristics of the test person **402**. In one embodiment of the present invention, the lung ventilation function is analyzed by means of a very simple model represented by a first-order differential equation (*Deq*) in lung-volume (*V*) dynamics in response to the driving pressure (P_L = atmospheric pressure - pleural pressure), as shown in FIG 5

[0051] First, the model governing equation derived from the basic equations is as follows:

$$R\dot{V} + V/C = P_L(t) - P_{el,\ 0} = P_o + P_1 \cos\omega t + P_2 \sin\omega t - P_{el\ 0} \qquad (1)$$

[0052] wherein (i) P_o, P_1 and P_2 are obtained from the given P_L ($= P_o - P_p$) data; (ii) the parameters of this Governing Deq are lung compliance (C) and airflow-resistance (R), wherein in the equation both R and C are instantaneous valves; (iii) $V = V(t) - Vo$ (the lung volume at the end-expiration); and (iv) $P_{el,o}$ is the lung elastic recoil pressure at the end of expiration and $P_{el,o} = P_{el} - V/C$.

[0053] At end-expiration when $\omega t = \omega T$, $P_L = P_{el,o}$. Hence,

[0054] $P_{el,o} = P_o + P_1$, and the governing equation (1) becomes:

$$R\dot{V} + V/C = -P_1 + P_1 \cos\omega t + P_2 \sin\omega t = P_n \qquad (2\text{-a})$$

[0055] where the right-hand side represents the net pleural pressure. $(P_n = P_{atm} - P_p - P_{el,o})$ curve. This P_n is in fact the driving pressure $(P_o - P_p)$ normalized with respect to its value at end-expiration. Equation(2-a) can be rewritten as follows:

$$\dot{V} + V/RC = -P_1/R + (P_1/R)\cos\omega t + (P_2/R)\sin\omega t \; ; \quad RC=\tau \qquad (2\text{-b})$$

[0056] wherein the P(t) clinical data displayed in FIG 5b is assumed to be represented by:

[0057] $Po = 9.84$ cm H2 O, $P1 = -1.84$ cm H2 O, and $P2 = 3.16$ cm H2 O (3)

[0058] If, in equation (1), R_a and C_a are designated as the average values (R and C) for the ventilatory cycle, then the solution of equation (1) is given by:

$$V(t) = -P_1 C_a + P_1 C_a [(\cos\omega t + \omega R_a C_a \sin\omega t)/(1+\omega^2 R_a^2 C_a^2)] + P_2 C_a$$

$$[(\sin\omega t - \omega R_a C_a \cos\omega t)/(1+\omega^2 R_a^2 C_a^2)] + H e^{-t/R_a} \qquad (4)$$

[0059] wherein the term (R_aC_a) is denoted by τ_a , and $\omega = 1.55$ *rad/s* (based on the data in FIG 5b. If $V = 0$ at $t = 0$, then, putting V (at $t = 0$) $= 0$ gives us:

$$H = \{ C_a \omega \tau_a / (1+\omega^2 \tau_a^2) \} (P_2 + P_1 \omega \tau_a) \qquad (5)$$

[0060] Then from equations (4) and (5), the overall expression for $V(t)$ becomes:

$$\boldsymbol{V(t)} = -P_1 C_a + \{ P_1 C_a (\cos\omega t + \omega \tau_a \sin\omega t)/(1+\omega^2\tau_a^2) \}$$

$$+\{ P_2 C_a (\sin\omega t - \omega \tau_a \cos\omega t)/(1+\omega^2\tau_a^2) \}$$

$$+\{ e^{-t/\tau a} \omega C_a \tau_a (P_2 + P_1 \omega \tau_a) / (1+\omega^2\tau_a^2) \} \qquad (6)$$

[0061] If $dV/dt = 0$ at $t = 0$, implying no air-flow at the start of inspiration, then equation (6) can be differentiated into:

$$\dot{V} = \{P_1 C_a/(1+\omega^2\tau_a^2)\}(-\omega \sin\omega t + \omega^2 \tau_a \cos\omega t)$$

$$+\{P_2 C_a/(1+\omega^2\tau_a^2)\}(\omega\cos\omega t + \omega^2 \tau_a \sin\omega t)$$

$$+\{- C_a\, e^{-t/\tau a} / (1+ \omega^2 \tau_a^2)\} (\omega P_2 + P_1 \omega^2 \tau_a) \qquad (7)$$

[0062] From equation (7), we get: $\dot{V} = 0$ at $t = 0$, thereby also satisfying this initial condition. By matching the above $V(t)$ expression (6) with the given $V(t)$ data (in FIG 5b), and carrying out parameter-identification, the in vivo values of R_a and C_a can be determined. As a check, it can be verified that the substitution of (6) and (7) satisfies equation (2).

[0063] However, we can also analytically evaluate R_a and C_a by satisfying some conditions. For this purpose, we first note that V is maximum (=Tidal Volume, TV) at about $t\ (= t_v) = 1.6s$, i.e. at $\omega t_v = 2.48\ rad$. Now, for $\omega t_v = 2.48\ rad$, we get : $\sin(\omega t_v)=0.62$, $\cos(\omega t_v) = -0.79$ and $\tan(\omega t_v) = -0.78$. Also, for $\omega t_v = 2.48\ rad$ (and based on the knowledge of the range of τ_a), the exponential term $e^{-t/\tau a}$ (in equation 6) becomes of the order of e^{-3} and less; hence, we decide to neglect it. So then, by and putting $\dot{V} = 0$ in equation. (7), we obtain:

$$\tan(\omega t_v) = (P_2 + \omega \tau_a P_1) / (P_1 - P_2 \omega \tau_a) = -0.78 \qquad (8)$$

[0064] Upon substituting the values of P_1 and P_2 from equation (3), and putting ω = 1.55 rad s^{-1} , we obtain the value of $\boldsymbol{\tau_a = 0.26s}$. We can also put $\ddot{V}=0$ at $t \cong 0.58$ or $\omega t=93$ and obtain a similar value for τ. Then, we also note that at $t_v = 1.6s$ (for which $dv/dt = 0$), $V = 0.6l$. Hence upon substituting for $\cos(\omega t_v) = -0.79$ and $\sin(\omega t_v) = 0.62$ in equation(7), and again neglecting the exponential term we get the following algebraic equation:

$$- P_1 C_a - (0.54 P_1 C_a / D) + (0.94 P_2 C_a / D) = 0.6; \qquad (9)$$

[0065] wherein $D = 1 + \omega^2 \tau_a^2$, $\omega = 1.55\ rad/s$, and $\tau_a = 0.26s$; this equation can hence be rewritten as:

$$C_a (-1.54 P_1 + 0.94 P_2) = 0.7 \qquad (10)$$

[0066] We can substitute, therein, the values of P_1 & P_2 from equation (3), and obtain the value of $\boldsymbol{C_a = 0.12\ L\ (cm\ H_2O)^{-1}}$. Since we have computed $\tau_a = 0.26\ s$, therefore $\boldsymbol{R_a = 2.20\ (cm\ H_2O)\ s\ L^{-1}}$. These are the average values of resistance- to – airflow and lung compliance during the ventilatory cycle shown in FIG 5b.

[0067] Since Lung disease will influence the values of *R* and *C*, these parameters can be employed to diagnose lung diseases. For instance in the case of emphysema, the destruction of lung tissue between the alveoli produces a more compliant lung, and hence results in a larger value of *C*. In asthma, there is increased airway resistance *(R)* due to contraction of the smooth muscle around the airways. In fibrosis of the lung, the membranes between the alveoli thicken and hence lung compliance *(C)* decreases. Thus by determining the normal and diseased ranges of the parameters R and C, we can employ this simple Lung-ventilation model for differential diagnosis. Let us, however formulate just one non-dimensional number to serve as a ventilatory performance index VTI_1 (to characterize ventilatory function), as:

$$VTI_1 = [\,(R_a C_a)(\text{Ventilatory rate in } s^{-1})\ 60\,]^2 = \tau^2 (BR)^2\, 60^2 \quad (11)$$

where *BR* is the breathing rate. Now, let us obtain its order-of-magnitude by adopting representative values of R_a *and* C_a in normal and disease states. Let us take the above computed values of $R_a = 2.2(cm\ H_2O)\ s\ L^{-1}$ *and* $C_a = 0.12\ L(cm\ H_2O)^{-1}$ *and* $BR = 12m^{-1}$ *or* $0.2s^{-1}$, computed for the data of Fig(1) and equation(3). Then, in a supposed normal situation, the value of VTI_1 is of the order of 9.75. In the case of obstructive lung disease, (with increased R_a), let us take $R_a = 3cm\ H_2O\ s\ L^{-1}$, $C_a = 0.12\ L\ (cm\ H_2O)^{-1}$ and $BR = 0.3s^{-1}$; then we get $VTI_1 = 42$. For the case of emphysema (with enhanced C_a), let us take $R_a = 2.0cm\ H_2O\ s\ L^{-1}$, $C_a = 0.2\ L(\ cm\ H_2O)^{-1}$ *and* $BR = 0.2s^{-1}$; then we obtain $VTI_1 = 23.04$. In the case of lung fibrosis (with decreased C_a), we take $R_a = 2.0cm\ H_2O\ s\ L^{-1}$, $C_a = 0.08\ L(\ cm\ H_2O)^{-1}$ and $BR = 0.2s^{-1}$; then we obtain $VTI_1 = 3.7$. We can, hence summarize that VTI_1 would be in the range of 2-5 in the case of fibrotic lung disease, 5-15 in normal persons, 15-25 for the case of emphysema, 25-50 in the case of obstructive lung disease. This would of course be needed to be verified by analyzing a big patient population.

[0068] Now, all of this analysis requires pleural pressure data, for which the patient has to be intubated. If now we evaluate the patient in an outpatient clinic, in which we can only monitor lung volume and not the pleural pressure, then we have to develop a non-invasively obtainable Ventilatory index.

[0069] In order to formulate a non-invasively determinable Ventilatory index from equation (1), we need to redesignate the model parameters, and indicate their identification procedure. So we make use of the following features from the volume-time data to facilitate evaluation of the following three parameters: : $(P_1 C)$, $(P_2 C)$ and τ:

[0070] At $t = t_v = 1.6s$ & $\omega t_v = 2.48$, V is *max* & $dV/dt = 0$; hence we rewrite equation (9) as:

[0071]

$$\tan(\omega t_v) = -0.78 = (P_2 + \omega t P_1) / (P_1 - P_2 \omega t) \qquad (12)$$

[0072] At $t = t_m$, $\ddot{V} = 0$; hence by differentiating equation. (7), without the exponential term ,we obtain:

$$\ddot{V} = 0 = \frac{P_1 C(-\omega^2 \cos \omega t_m - \omega^3 \tau \sin \omega t_m) + P_2 C(-\omega^2 \sin \omega t_m - \omega^3 \tau \sin \omega t_m)(1+\omega^2\tau^2)}{}$$

i.e. $$\tan \omega t_m = (-P_1 + \omega\tau P_2) / (P_1 \omega\tau + P_2) \qquad (13)$$

[0073] At $t = 1s$ & $\omega t = \pi/2$, $V = V_1$ (whose value is obtainable from FIG 5b); this condition yields (without the exponential term):

$$V_1 = -(P_1 C) - \{\omega \tau (P_1 C)/(1+\omega^2\tau^2)\} + \{(P_2 C)/(1+\omega^2\tau^2)\} \qquad (14)$$

[0074] At $t = 2s$ & $\omega t = \pi/2$, $V = V_2$ (whose value is obtainable from FIG 5b); this condition yields (without the exponential term):

$$V_2 = -(P_1 C) - \{(P_1 C)/(1+\omega^2\tau^2)\} + \{\omega\tau (P_2 C)/(1+\omega^2\tau^2)\} \qquad (15)$$

[0075] At $t = 0.3s$ & $\omega t = 270$, $V = V_3$ gives:

$$V_3 = -(P_1 C) - \{\omega\tau (P_1 C)/(1+\omega^2\tau^2)\} - \{(P_2 C)/(1+\omega^2\tau^2)\} \qquad (16)$$

[0076] From equations (12) & (13) and any one of the equation(s) (14-16), we can only obtain the values of $\omega\tau$ (or of τ, since $\omega = 1.55$) and of $(P_1 C)$ & $(P_2 C)$ but not of P_1 & P_2 by themselves. On the other hand, by also fitting equation (6), (without the exponential term) to the V(t) data, we obtain:

$$\tau = rad\ s^{-1};\quad P_1 C = L,\quad P_2 C = L \qquad (17)$$

[0077] We can nevertheless formulate another non-invasively-determinable non-dimensional ventilatory index ($V\,T\,I_2$) in terms of these parameters. ($\omega\tau$, $P_1\,C$ and $P_2\,C$) as follows:

[0078]

$$V\,T\,I_2 = 60\,(\omega\tau)\,(T\,V)^2 / 2\pi\,(P_1\,C)(P_2\,C) \qquad (18)$$

$$= 30\,\omega\,(R/C)\,(T\,V)^2 / \pi\,P_1\,P_2$$

[0079] It is seen that VTI_2 can in fact be expressed in terms of P_1, P_2 and R, C. This VTI_2 index can be evaluated by computing the values of τ, alongwith ($P_1\,C$) & ($P_2\,C$) as given by equation (17). Then, after evaluating $V\,T\,I_2$ for a number of persons, and patients its distribution can enable us to categorize and differentially diagnose patients with various lung disorders and diseases. Between the two indices VTI_1 and VTI_2, we can employ the one that enables more distinct separation of subjects with different ventilatory disorders.

[0080] Thus far, we have adopted the average cyclic values C_a and R_a for our DE_q model parameters. However, we expect that C will vary with lung volume (V), and that R will perhaps vary with the airflow-rate or ($\dot{V}$) or even ω. Hence, for a true representation of the lung properties C & R, let us determine their values for different times during the ventilatory cycle, and compare them with their average values C_a & R_a, so as to make a case for a non-linear ventilatory-function model.

[0081] Let us hence compute the instantaneous value of compliance (***C***) at mid-inspiration at (***t = tm***), and compare it with that of its average cyclic value of C_a. For this purpose let us differentiate equation (2-a), giving :

[0082]

$$R\,\ddot{V} + \dot{V}/C = -\,P_1\,\omega Sin\omega t + P_2\,\omega Cos\omega t \qquad (19)$$

[0083] Now at about mid-inspiration, when $t \cong 0.87\ s$ and $\omega\,t = \omega\,t_m \cong 1.32\ rad$ or 78°, $\ddot{V} = 0$ and $\dot{V} = 0.5\ L/s$ (based on fig1). By substituting for $\ddot{V}$ and $\dot{V}$ in equation (19), we obtain, $\boldsymbol{C \cong 0.14 L/cm\ H_2O}$ (compared to its C_a value of 0.118). Now, in order to also compute R at $\omega\,t_m = 1.32$ we substitute $\dot{V} = 0.5\ L/s$ *and* $V = 0.3L$ (from the fig1 data) into equation (2-a), to obtain:

[0084]

$$0.5\,R \;+\; 0.3/C \;= -P_1 + P_1\,Cos\,78^o \;+ P_2\,Sin\,78^o \;=\; 3.8 \qquad (20)$$

[0085] Now, since $C(at\ \omega\, t_m = 1.32) \cong 0.14 L/cm$ H_2O, we obtain $\boldsymbol{R=4.6(cmH_2O)\ s\ L^{-1}}$, compared to $R_a = 2.20$. This gives us some idea of the order of magnitude of R & C, in comparison to their average values C_a & R_a. We could also expect C at mid-inspiration to be higher than its value at end-inspiration, when the lung is fully inflated. Also, we could expect the flow-resistance to be maximum at mid-inspiration, when $\dot{V}$ is maximal.

[0086] We can hence represent lung compliance *(C)* and resistance *(R)* as follows:

[0087]

$$C = n_1\,(V)^{n_2} - C_o \quad \text{or} \quad C = C_o\, e^{n(V)} \quad , \qquad (21\text{-a})$$

$$R = s_1 V_2^{s_2} \quad \text{or} \quad R = R_o e^{s\dot{V}} \qquad (21\text{-b})$$

[0088] wherein $\dot{V}$ can also be varied by having the subjects breathe at different tidal volumes (TV_s) and ventilation frequency (ω)

[0089] We note as per the conventional formulation of compliance, given by equation(ii) in FIG 5b as:

$$P_{el} \;= V/C \;+\; P_{el,\,0}\ ; \qquad (22)$$

[0090] In the above formulation, we assume that C and $E(= 1/C)$ remains constant throughout the ventilation cycle. However at the start of inspiration, $C = C_o$ at $t = 0$, and it decreases as the lung volume increases, based on the lung (static) volume vs pressure curve. So let us improve upon this equation(22) model, by making P_{el} to be a non-linear function of volume, as follows:

[0091]

$$P_{el} = P_{el,o} + E_0 e^{kv} \qquad (23)$$

[0092] Employing the above format of compliance, the governing DE_q (1) becomes

[0093] $$R\dot{V} + E_0\, e^{kV} = P_L(t) - P_{el,0} = P_o + P_1 \cos\omega t + P_2 \sin\omega t - P_{el,0} \qquad (24)$$

[0094] Again at end-expiration, $P_{el,\,0}$ = intra-pulmonary pressure $= (P_0 + P_1)$. Hence equation.(24) becomes:

[0095]

$$R\dot{V} + E_0 e^{kV} = -P_1 + P_1 \cos \omega t + P_2 \sin \omega t \qquad (25)$$

[0096] whose RHS is similar to that of equation(2-a), and the values of P_1 & P_2 are given by equation(3) for the FIG 5b data.

[0097] In order to evaluate these parameters k & E_0 ,we again bring to bear the situation that at end-inspiration, for $t = t_v = 1.6\ s$ (for which $\omega t = \omega t_v = 2.48\ rad$, $\sin \omega t_v = 0.62$ & $\cos \omega t_v = -0.79$), we have

$\dot{V} = 0$ and $V = V_{max} = TV = 0.6\ L$. Hence, from fig (1) data, and equations (3 & 25), we obtain:

[0098]

$$E_0 e^{0.6k} = 8.75 \qquad (26)$$

[0099] Let us now employ the volume data point at which $\ddot{V} = 0$. For this purpose, we differentiate equation (25) , to obtain:

[00100] $$R\ddot{V} + E_0 k e^{kV} = -P_1 \omega \sin(\omega t) + P_2 \omega \cos(\omega t) \qquad (27)$$

[00101] From the Fig (1) data at about mid-inspiration, for which $t = t_m = 0.87s$ & $\omega t_m = 1.32 = 78^o$ with $\cos(\omega t_m) = 0.2$ & $\sin(\omega t_m) = 0.98$), we have $\ddot{V} \cong 0$, $\dot{V} = 0.5\ Ls^{-1}$, $V = 0.3\ L$, from fig(1) data. Substituting these values into equation.(27), we get :

[00102]

$$E_0 e^{0.3k} = 3.8 \qquad (28)$$

[00103] Now from equation(s) (27) & (28), we get:

$$e^{-0.3k} = 1.38 \qquad (29)$$

[00104] *for which,* $k = 1.07$ *and (from equation 26 or 28)* $E_0 = 2.75$ (30)

[00105] Hence, by employing the non-linear formulation,

$$P_{el} = P_{el,o} + E_o e^{kv}$$

[00106] we obtain the following expression for lung compliance (or elastance):

[00107]

$$\frac{dP_{el}}{dV}=E=\frac{1}{C}=E_o k e^{kV}=2.94e^{1.07V} \quad (31)$$

[00108] Based on this expression, we obtain ,for $t = t_m$ & $V = 0.3L$:

$E = 1/C = 4.06cm\ H_2O / L$; and $\boldsymbol{C = 0.25\ L/cm\ H_2\ O.}$ (32)

[00109] Equation(32) can now provide us a more realistic characterization of lung compliance as follows:

At $t = 0$ and $V = 0$, we compute $E = 1/C = 2.94$ and $C = 0.34\ cm\ H_2\ O / L$.

At $t = t_m = 0.87s$ & $V = 0.3L$, $E = 1/C = 4.06$, and $C = 0.25$. (33)

At $t = t_v = 1.6s$ & $V = 0.6L$, and $E = 1/C = 5.6$ and $C = 0.18$,

which corresponds to the value of C_a.

[00110] Our non-linear formulation of lung compliance, as depicted by equation. (31 & 33) , indicates that compliance decreases from $0.34\ cm\ H_2\ O/L$ at start-inspiration to $0.25\ cm\ H_2\ O/L$ at about mid-inspiration, and then to $0.18\ cm\ H_2\ O/L$ at the end of inspiration. What this also tells us is that the ventilatory model equation.(1) gives the correct reading of the compliance at V_{max} , i.e. at end-inspiration. At other times of inspiration and expiration, the C_a parameter underestimates the instantaneous value of lung compliance. Now how about obtaining an analytical solution of equation(25) for V(t), and fitting the expression for V(t) to the lung volume data, to evaluate the parameters (i) R, E_o & k for an intubated patient and (ii) R, E_o k & P_1 & P_2 for a non-intubated patient in the out patient clinic.

[00111] Finally, while it is important to determine the normal and pathological diagnostic ranges of C_a & R_a , or better still of the parameters (E_0 & k) of the C vs V and R vs $\dot{V}$ relationships, it would be more useful to construct and employ a non-dimensional ventilatory index We have already formulated $VT\ I_1$ & $VT\ I_2$ in equations (11) & (18), respectively. We will now formulate yet another index :

[00112]

$$VT\ I_3 = [60\ (R_a / C_a)\ (BR)\ (TV)^2] / P_1\ P_2 \quad (34)$$

[00113] wherein (i) BR (the breathing rate in $\#/sec) = 0.5\ \omega/\pi$ (ii) and $|P_1\ P_2|$ is the absolute value of the product $P_1\ P_2$ (because of P_1 being negative). For the fig (1)

clinical data, of $BR= 0.25$, with $TV=0.6L$ & $/P_1 P_2/^{(3)} = 18.1$, and for the computed value of $R_a/C_a=18.33cm\ H_2O\ s\ L^{-1}$, we obtain $VTI_3 = 5.47$. Between VTI_1 and VTI_3 , we can decide which index enables us to better differentially diagnose subjects with ventilatory disorders.

[00114] Now, let us go one step further and recognize that, for non-intubated patients, we cannot monitor P_1 and P_2 , and hence cannot evaluate R_a & C_a as demonstrated in [0051] to [0068]. However for evaluating ventilatory index in out-patient clinics, we can infact adopt $(P_1\ C)$ and $(P_2\ C)$ to be the model-system parameters, and evaluate them as delineated in [0069] to [0083]. We can hence adopt the non-invasively-obtainable ventilatory-performance index $VT\ I_2$ (given by equation 18):

[00115]

$$\boldsymbol{VTI_2} = 30\ \omega\tau\ (TV)^2 / \pi\ (P_1\ C)\ (P_2 C);\ \ \omega = 2\pi(BR)(in\ \#s^{-1})$$

$$= 60\ (BR)(R / C)\ (TV)^2 / (P_1\ P_2) = \quad VT\ I_3 \qquad (35)$$

[00116] which is noted to be the same expression as for index VTI_3 , except that it can be evaluated without intubating the patient. Hence, it would be even more useful to determine the distribution of VTI_2 for patients with a wide range of lung pathologies and ventilatory disorders. Then, we can delineate the normal and pathological ranges of this index, and employ this information to diagnose patients into different disease categories.

[00117] Now referring to FIG 6, there is provided a more detailed description of the functional configuration **600** of the disease identification software system and databases in accordance with one embodiment of the present invention. When the user **610** inputs all the parameters through the GUI **620** into the computer processor, the inputs will be queued in the query module **640**. Then, when the parameters are transformed into indexes as discussed above, the indexes will go to the disease identification software system **650** through the intelligent DBMS to obtain the matched diseases; the indexes at the same time are compared with the expert DB **630** to verify the diagnosis. The final diagnosis will be output **105** as shown in FIG 1.

[00118] Now referring to FIG 7, there is provided a more detailed description of the performance **700** of the disease identification software system as shown in FIG 6. The index **710** first fires up numerous search engines in finding the best match diseases in the intelligent DBMS **720**; the search engines will search the disease identification software

system (DISS) **730**. The DISS contains many types of diseases **750**, **760**, **770** where the search engines will search all the types of diseases and output the search results to the Query Module **780**. There is a possibility that there is more than one type of diseases being suspected. Statistical means are used to determine which is the best match diseases and thereby putting all these best fit diseases to the expert database system to further refine the possibilities **740**. Then the verified diagnostic results will be output to the user **800** through the GUI **790**.

[00119] While the present invention has been described with reference to particular embodiments, it will be understood that the embodiments are illustrative and that the invention scope is not so limited. Alternative embodiments of the present invention will become apparent to those having ordinary skill in the art to which the present invention pertains. Such alternate embodiments are considered to be encompassed within the spirit and scope of the present invention. Accordingly, the scope of the present invention is described by the appended claims and is supported by the foregoing description.

CLAIMS

What is claimed is:

1. A pulmonary disease detection system for detecting breathing deficiencies of a test person, comprising:

a breath analyser for analysing the inhale and exhale airs of the test person;

a computer processor for receiving from the breath analyser the information and processing the received analysis data to give values to different parameters of the inhale and exhale airs of the test person; and

a medical database for storing different parameters of breaths of the public and normal ranges for healthy persons;

thereby the computer processor compares the values of different parameters of the inhale and exhale airs of the test person with the ones stored in the medical database so as to yield a test result of whether the test person is suffering breath deficiencies.

2. The pulmonary disease detection system of claim 1, wherein the breathing deficiencies may be caused by bacterial infection, viral infection, physical injuries and cancer.

3. The pulmonary disease detection system of claim 1, wherein the breath analyser comprises:

a mask for covering the nose and mouth of the test person so as to maximizing the delivery of fresh air and minimizing the loss of the exhaled air;

an air tank for supplying measurable and controllable air to the mask; and

an acquisition unit electrically connected with the mask and the air tank so as to receive all the information from the air tank and the mask.

4. The pulmonary disease detection system of claim 3, wherein the mask comprises a group of electrodes including an oxygen electrode, a carbon dioxide gas electrode, a nitrogen electrode, a water vapor electrode, an inspired air flow rate electrode and an

expired air flow rate electrode; wherein each electrode detects each designated component of the inhale and exhale airs.

5. The pulmonary disease detection system of claim 4, wherein the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the breathing rate, oxygen consumption rate and carbon dioxide generation rate from the composition information of the inhale and exhale airs of the test person; thereby comparing the rates of the test person with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

6. The pulmonary disease detection system of claim 4, wherein the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the composition values of the exhale air from the composition values of the inhale air and the assumed normal rates including breathing rate, oxygen consumption rate and carbon dioxide generation rate; thereby comparing the composition values of the exhale air with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

7. The pulmonary disease detection system of claim 4, wherein the computer processor is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the volume compliance and air-flow resistance of the test person from the composition values and pressure values derived from the inhale air rate and exhale air rate; thereby comparing the volume compliance and air-flow resistance values with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

8. The pulmonary disease detection system of claim 7, wherein the derived pressure values include pleural pressure and alveolar pressure.

9. The pulmonary disease detection system of claim 8, wherein the volume compliance and air-flow resistance are calculated by the equations (1) – (10) so that VTI_1 is computed as follows:

$$VTI_1 = [\ (R_a\ C_a\ )(\text{Ventilatory rate in } s^{-1})\ 60\ ]^2 = \tau^2\ (BR)^2\ 60^2$$

wherein ***VTI1*** is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein R_a and C_a are designated as the average values (R and C) for the ventilatory cycle, and BR is the breathing rate.

10. The pulmonary disease detection system of claim 8, wherein ($\omega\tau$), (P_1 C) and (P_2 C) are calculated by the equations (12) – (17) so that V T I_2 is computed as follows:

$$V\ T\ I_2 = 60\ (\omega\tau)\ (T\ V)^2\ /2\pi\ (P_1\ C\ )(P_2\ C) = 30\ \omega\ (R/C)\ (T\ V)^2\ /\pi\ P_1\ P_2$$

wherein V T I_2 is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein TV is tidal volume; P_1& P_2 as pleural pressures and $\omega\tau$.

11. The pulmonary disease detection system of claim 8, wherein all the required parameters involved are calculated by the equations (1) to (10) and (12)-(17) so that V T I_3 is computed as follows:

$$VT\ I_3 = [60\ (\ R_a\ /\ C_a\ )\ (BR)\ (TV)^2]/\ P_1\ P_2$$

wherein VTI3 is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein TV is tidal volume; and P_1& P_2 as pleural pressures.

12. A method of detecting a pulmonary disease of a test person by investigating the breath deficiencies of the test person, said method comprising the steps of:

acquiring the information of inhale and exhale airs of the test person;

processing the acquired information to give values of designated aspects of the inhale and exhale airs of the test person; and

comparing the calculated values with the ones stored in a medical database so as to conclude whether the test person is suffering any pulmonary diseases.

13. The method of claim 12, wherein the breathing deficiencies may be caused by bacterial infection, viral infection, physical injuries and cancer.

14. The method of claim 12, wherein the information of inhale and exhale airs of the test person is acquired by a breath analyser; and wherein the breath analyser comprises:

a mask for covering the nose and mouth of the test person so as to maximizing the delivery of fresh air and minimizing the loss of the exhaled air;

an air tank for supplying measurable and controllable air to the mask; and

an acquisition unit electrically connected with the mask and the air tank so as to receive all the information from the air tank and the mask.

15. The method of claim 14, wherein the mask comprises a group of electrodes including an oxygen electrode, a carbon dioxide gas electrode, a nitrogen electrode, a water vapor electrode, an inspired air flow rate electrode and an expired air flow rate electrode; wherein each electrode detects each designated component of the inhale and exhale airs.

16. The method of claim 15, wherein the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the breathing rate, oxygen consumption rate and carbon dioxide generation rate from the composition information of the inhale and exhale airs of the test person; thereby comparing the rates of the test person with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

17. The method of claim 15, wherein the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the composition values of the exhale air from the composition values of the inhale air and the assumed normal rates including breathing rate, oxygen consumption rate and carbon dioxide generation rate; thereby comparing the composition values of the exhale air with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

18. The method of claim 15, wherein the processing is executed within a computer processor that is embedded with an algorithm for processing the information from the breath analyser; wherein the algorithm calculates the volume compliance and air-flow resistance of the test person from the composition values and pressure values derived from the inhale air rate and exhale air rate; thereby comparing the volume compliance and air-flow resistance values with the ones stored in the medical database so as to conclude whether the test person is suffering any breath deficiencies.

19. The method of claim 18, wherein the derived pressure values include pleural pressure and alveolar pressure.

20. The method of claim 19, wherein the volume compliance and air-flow resistance are calculated by the equations (1) – (10) so that V T I_1 is computed as follows:

$$VTI_1 = [\,(R_a\, C_a)(\text{Ventilatory rate in } s^{-1})\; 60\,]^2 = \tau^2\,(BR)^2\,60^2$$

wherein $\boldsymbol{VTI_1}$ is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein R_a and C_a are designated as the average values (R and C) for the ventilatory cycle, and BR is the breathing rate.

21. The method of claim 19, wherein ($\omega\tau$), (P_1 C) and (P_2 C) are calculated by the equations (12) – (17) so that V T I_2 is computed as follows:

$$V\,T\,I_2 = 60\,(\omega\tau)\,(T\,V)^2 / 2\pi\,(P_1\,C)(P_2\,C) = 30\,\omega\,(R/C)\,(T\,V)^2 / \pi\,P_1\,P_2$$

wherein V T I_2 is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein TV is tidal volume; P_1& P_2 as pleural pressures and $\omega\tau$.

22. The method of claim 19, wherein all the required parameters involved are calculated by the equations (1) to (10) and (12)-(17) so that V T I_3 is computed as follows:

$$VT\,I_3 = [60\,(R_a / C_a)\,(BR)\,(TV)^2] / P_1\,P_2$$

wherein VTI3 is denoted as ventilatory performance index; C as lung-volume compliance; and R as air-flow resistance; and wherein TV is tidal volume and P_1& P_2 as pleural pressures.