(12) STANDARD PATENT (11) Application No. **AU 2004222800 B2**
(19) AUSTRALIAN PATENT OFFICE

(54) Title
Method and apparatus for determining conditions of biological tissues

(51) International Patent Classification(s)
A61B 8/00 (2006.01) ***A61B 5/08*** (2006.01)

(21) Application No: **2004222800** (22) Date of Filing: **2004.10.21**

(43) Publication Date: **2004.11.11**
(43) Publication Journal Date: **2004.11.11**
(44) Accepted Journal Date: **2008.05.29**

(62) Divisional of:
2001252025

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(56) Related Art
US 5620004

ABSTRACT

Method and Apparatus for Determining Conditions of Biological Tissue

The present invention relates to methods and apparatus of determining characteristics of biological tissues, including especially the respiratory system, in humans and animals. In particular, it relates to determining the characteristics of the lungs and airways by introducing sound into the body and by recording the velocity and/or attenuation of the sound.

[Fig. 1]

AUSTRALIA

Patents Act

COMPLETE SPECIFICATION
(ORIGINAL)

Class Int. Class

Application Number:
Lodged:

Complete Specification Lodged:
Accepted:
Published:

Priority

Related Art:

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Invention Title:

METHOD AND APPARATUS FOR DETERMINING CONDITIONS OF BIOLOGICAL TISSUES

Our Ref : 731067
POF Code: 459639/459639

The following statement is a full description of this invention, including the best method of performing it known to applicant(s):

METHOD AND APPARATUS FOR DETERMINING CONDITIONS OF BIOLOGICAL TISSUES

This application is a divisional application of Australian Patent Application 2001252025 the entire contents of which are herein incorporated by reference.

The present invention relates to a method of determining characteristics of biological tissues in humans and animals. In particular, it relates to determining the characteristics of tissues such as the lungs and airways by introducing a sound to the tissue, and recording the sound. The invention further includes apparatus capable of such measurement. In a particularly preferred aspect, the present invention relates to the detection of sleep apnea.

BACKGROUND OF THE INVENTION

Non-invasive determination of the condition of biological tissues is useful in particular where the patient is unable to co-operate or the tissue is inaccessible for easy monitoring.

Techniques presently used in determining the characteristics of biological tissues include x-rays, magnetic resonance imaging (MRI) and radio-isotopic imaging. These are generally expensive and involve some degree of risk which is usually associated with the use of X-rays, radioactive materials or gamma-ray emission. Furthermore, these techniques are generally complicated and require equipment which is bulky and expensive to install and, in most cases, cannot be taken to the bedside to assess biological tissues in patients whose illness prevents them being moved.

Sound waves, particularly in the ultra-sound range have been used to monitor and observe the condition of patients or of selected tissues, such as the placenta or fetus. However, the process requires sophisticated and sometimes expensive technology and cannot be used in tissues in which there is a substantial quantity of gas, such as the lung.

Every year in Australia about 5000 newborn infants require a period of intensive care (ANZNN Annual Report, 1996-1997). Respiratory failure is the most common problem requiring support and is usually treated with a period of mechanical ventilation. Over the last decade the mortality of infants suffering respiratory failure has shown an impressive decline, attributable at least in part to improved techniques used in mechanical ventilation, and the introduction of surfactant replacement therapy (Jobe, 1993). The vast majority of infants now survive initial acute respiratory illness, but lung injury associated with mechanical ventilation causes many infants to develop 'chronic lung disease'. Chronic lung disease is characterised by persisting inflammatory and fibrotic changes, and causes over 90% of surviving infants born at less than 28 weeks gestation, and 30% of those of 28-31 weeks gestation, to be dependent on supplementary oxygen at 28 days of age. Of these, over half still require supplementary oxygen when they have reached a post-menstrual age of 36 weeks gestation (ANZNN Annual report, 1996-1997). Assistance with continuous positive airway pressure (CPAP) or artificial ventilation is also commonly required.

Historically, barotrauma and oxygen toxicity have been considered to be the primary culprits in the aetiology of chronic lung disease (Northway et al, 1967; Taghizadeh & Reynolds, 1976). However, trials of new strategies in mechanical ventilation which were expected to reduce barotrauma and/or exposure to oxygen have often had disappointingly little impact on the incidence of chronic lung disease (HIFI Study Group,1989; Bernstein et al, 1996; Baumer, 2000). Comparison of strategies of conventional mechanical ventilation in animals (Dreyfuss et al, 1985) have indicated that high lung volumes may be more damaging than high intrapulmonary pressures, and has led to the concept of 'volutrauma' due to over-inflation of the lung. At the same time, experience with high frequency oscillatory ventilation (HFOV) has indicated that avoidance of under-inflation may be equally important. HFOV offers the potential to reduce lung injury by employing exceptionally small tidal volumes which are delivered at a very high frequency. However, this technique fails to confer benefit, if the average lung volume is low (HIFI Study Group, 1989), yet it appears to be successful if a normal volume is maintained (McCulloch et al, 1988; Gerstmann

et al, 1996). This highlights the importance of keeping the atelectasis-prone lung 'open' (Froese, 1989). Evidence of this kind has led to the concept that a 'safe window' of lung volume exists within which the likelihood of lung injury can be minimised. The key to preventing lung injury may lie in maintaining lung volume within that safe window thereby avoiding either repetitive over-inflation or sustained atelectasis. (See Figure 1).

Attempts to maintain an optimal lung volume in the clinical setting are frustrated by a lack of suitable methods by which the degree of lung inflation can be monitored. In current practice, evaluation of oxygen requirements and radiological examination of the lungs are the principal techniques employed. However, oxygen requirements may be influenced by factors other than lung volume (for example intra- or extracardiac right to left shunting), and the hazards of radiation exposure prevent radiological examination being performed with the frequency required.

Monitoring of infants during mechanical ventilation has been significantly improved over the last decade by the incorporation of a pneumotachograph or hot-wire anemometer into the design of many neonatal ventilators. Although this provides a valuable tool for monitoring tidal volume and compliance, it gives only the most indirect indication (from the shape of the pressure-volume curve) of whether that tidal volume is being delivered in a setting of under-inflation, optimal inflation, or over-inflation. Furthermore, while absolute lung gas volume can be measured using 'gold-standard' techniques of Nitrogen (N_2) washout or Helium (He) dilution, these are impractical for routine clinical use.

Even when lung volume is maintained in the "safe window", changes in the lung condition may manifest due to the general damaged or underdeveloped condition of the lung. Fluid and blood may accumulate in the lung, posing additional threats to the patient. Evaluation with a stethoscope of audible sounds which originate from within the lung (breath sounds) or are introduced into the lung (by percussion, or as vocal sounds) forms an essential part of any routine medical examination. However, in the sick newborn, the infant's small

size, inability to co-operate and the presence of background noise greatly limits the value of such techniques.

Aside from the difficulty of determining and monitoring lung condition in newborn babies and adults, determining airway condition in human subjects is equally challenging, particularly when the patient is unconscious or unable to cooperate. The need to monitor airway patency is particularly important in patients with the disorder of obstructive sleep apnea (OSA), in which the upper airway obstructs periodically during sleep, causing disruption of breathing and an associated fall in blood oxygen stores. In extreme cases the patient may be repetitively aroused from sleep and blood oxygen may fall to dangerously low levels. As a result, patients may suffer from adverse effects including excessive daytime sleepiness, and an increased incidence of cardiovascular disease and stroke.

The current method for diagnosing OSA involves an overnight polysomnographic recording conducted in a sleep laboratory. However, identification of obstructions requires a rigorous comparison of multiple traces by an experienced polysomnographer. As a result, this method is expensive and time-consuming and may lead to inaccurate results. Other known approaches (such as the oesophageal balloon method) are too invasive for routine clinical use.

A need exists for simple, non-invasive and convenient methods by which the condition of the respiratory system, including the degree of inflation of the lungs and the patency of the upper airway, can be closely monitored in the clinical setting.

According to an aspect of the present invention, there is provided a method for monitoring airway patency including the steps of:

introducing an audible frequency sound signal into the airway at a first location;

detecting the sound signal on the chest at a second location after the sound has travelled through at least a portion of the airway; and

determining airway patency based on the attenuation of said sound signal between the first and second locations.

According to another aspect of the present invention there is provided apparatus for monitoring airway patency, the apparatus including:

sound generating means for applying an audible frequency sound signal in the airway at a first location;

sound detection means for detecting the sound signal on the chest at a second location after the sound has travelled through at least a portion of the airway; and

analysis means for determining airway patency based on the attenuation of said sound signal between said first and second locations.

According to a further aspect of the present invention, there is provided CPAP delivery apparatus for delivering continuous positive airway pressure to a subject, the CPAP apparatus including airway patency monitoring apparatus that includes means for applying sound to the airway of the subject and means for detecting, on the chest of the subject, the attenuation of the applied sound after transmission through at least a portion of the airway.

BRIEF DESCRIPTION OF THE FIGURES

Figure 1 shows a pressure-volume curve of a moderately diseased lung illustrating two hazardous regions of lung volume, and indicating an optimal "safe" window there between (from Froese, 1997).

Figure 2 shows (A) Sound pressure level (dB) and sound velocity (m/s) versus frequency (Hz) for pooled results taken from 5 adult subjects during breath-holds at residual volume (RV), functional residual capacity (FRC) and total lung capacity (TLC). (B) shows results from an infant of 26 weeks gestation with healthy lungs, each data point representing the pooled mean ± S.E. of 5 measurements. The results were obtained from a reference position in the adult with the transducer at the 2nd right intercostal space on the anterior chest wall and in the newborn over the right upper chest. In both adult and infant, the microphone was placed on the opposite wall of the chest directly in line with the transducer.

Figure 3 illustrates the relationship between sound velocity and the volumetric fraction of tissue hand the average lung density.

Figure 4 (a) illustrates an electric circuit which models the acoustic characteristics of the thorax. Figure 4 (b) illustrates (1) large, (2) moderate and (3) small acoustic losses as measured using the electric circuit model and which represents the output SPL as would be measured at a chest microphone when the input SPL is 105dB.

Figure 5 (a) shows the SPL measured at a chest microphone, recorded before (pre) and after (post) administration of surfactant in 3 preterm infants, wherein the sound level produced by the transducer was 105dB (Sheridan 2000). Figure 5 (b) shows the electric model simulation of Figure 5 (a), demonstrating the change in the SPL measured at the chest wall following a 3-fold increase in lung gas compliance, wherein the sound level produced by the transducer was, again, dB.

Figure 6 shows the relationship between frequency and the attenuation coefficient α plotted with tissue fraction h as a parameter.

DETAILED DESCRIPTION

In an aspect of the invention there is provided apparatus for monitoring airway patency, the apparatus including:

(a) sound generating means for applying an audible frequency sound signal in the airway at a first location;

(b) sound detection means for detecting the sound signal at a second location on the chest, after the sound has travelled through at least a portion of the airway; and

(c) analysis means for determining airway patency based on the attenuation of said sound signal between said first and second locations.

Characteristics of biological tissues can be determined by measuring the velocity and attenuation of a sound as it propagates through the tissue. This can be achieved by introducing a sound to a particular location or position on the tissue, allowing the sound to propagate through the tissue and measuring the velocity and attenuation with which the sound travels from its source to its destination, wherein the destination includes a receiver which is spatially separated from the sound's source.

In an embodiment, airway patency may be monitored by measuring attenuation of the applied sound signal.

Characteristics of the biological tissue may include a feature of the tissue including but limited to its make-up, volume, condition or position in the body.

Biological tissues may include any single tissue or a group of tissues making up an organ or part or region of the body. The tissue may comprise a homogeneous cellular material or it may be a composite structure such as that found in regions of the body including the thorax which for instance can include lung tissue, gas, skeletal tissue and muscle tissue. However, it is particularly preferred that the tissue is porous which comprises a composite structure made up of tissue and gas or has regions of high and low density such as that found in bone tissue.

Preferably the tissue is of the respiratory system. More preferably the tissue is lung tissue or from the upper airway of the respiratory system. Preferably the upper airway includes the buccal region extending to the trachea before entering the lungs.

Throughout the description and claims of this specification, the word "comprise" and variations of the word, such as "comprising" and "comprises", is not intended to exclude other additives, components, integers or steps.

An understanding of the theoretical aspects of sound transmission in tissue is essential for the best use of bio-acoustic data which is obtained using the present invention.

A unique feature of sound propagation through the lung parenchyma is that the sound velocity is less than that expected for either tissue (1500 ms^{-1}) or air (343 ms^{-1}). This can be explained, in part, by examining the basic relationship between sound velocity v and the physical properties of the lung tissue through which the sound is propagating. This relationship is:-

$$v = \frac{1}{\sqrt{\rho C}} \qquad (1)$$

where ρ is the density and C is the volumetric compliance or inverse volumetric stiffness per unit volume. In determining the velocity of sound in air, substituting an air density of 1.2 kgm^{-3} and an air compliance of 7.14×10^{-6} Pa^{-1} yields a sound velocity in air of 342 ms^{-1}.

Rice(1983) has shown that this relationship also holds for composite porous materials with a closed cell structure which is similar to that of the lung, but where ρ and C are replaced by the tissue's average or composite values. Expressing these values in terms of the volumetric fraction of tissue h and of gas $(1-h)$ and the constituent densities and compliances gives tissue density:

$$\rho = (1-h)\rho_g + h\rho_t \qquad (2)$$

and volumetric compliance:

$$C = (1-h)C_g + hC_t \qquad (3)$$

where ρ, ρ_g, ρ_t are the composite, gas and tissue densities respectively and C, C_g, C_t are the composite, gas and tissue volumetric compliances respectively.

Substituting equations (2) and (3) into equation (1) yields an expression which relates sound velocity through a composite structure to the volumetric fraction

and the physical properties of both the tissue and gas which compose the material:

$$v = \frac{1}{\sqrt{((1-h)\rho_g + h\rho_t)((1-h)C_g + hC_t)}} \qquad (4)$$

It must also be noted that the density of air is approximately 3 orders of magnitude less than that of most tissues and the volumetric compliance of air is some 4 orders of magnitude larger than that of most tissues. This can be used to determine the velocity of sound propagation through the lung for a range of volumetric fractions which are likely to be seen in the lung, (0.05 at TLC to 0.5 to 0.9 for a fully atelectatic/collapsed lung). These velocities can be determined by simplifying equation 4 as follows:

$$v = \frac{1}{\sqrt{h(1-h)}} \frac{1}{\sqrt{\rho_t C_g}} \qquad (5)$$

Equation 5, in combination with Figure 3 illustrates the dependence that sound velocity has on the volumetric fraction of tissue, the volumetric fraction of air, the tissue density and the gas compliance. The tissue compliance and the gas density play essentially no role in the determination of velocity.

Sound velocity in composite materials is determined in part by the product of the tissue density and the gas compliance. The result of this is that the lung parenchyma appears to act *like homogeneous mass-loaded air* as far as sound propagation is concerned, such that the velocity of sound propagation through the tissue is markedly slower than through air. Substitution of known values for tissue density, ρ_t and gas compliance, C_g in equation 5 gives:

$$v = \frac{11.82}{\sqrt{h(1-h)}} \qquad (6)$$

Differentiation of v in equation 6 with respect to h determines a minimum value for velocity at $h = 0.5$ where $v = 23.6\ \mathrm{ms}^{-1}$. For values of $h < 0.5$ the velocity increases with decreasing lung density and conversely for $h > 0.5$ the velocity decreases with decreasing lung density. This is clarified by way of illustration in Figure 3.

The quadratic properties of equation 6 result in the presence of two values for h for any particular value of measured velocity. These values are:

$$h = 0.5 \pm \sqrt{0.25 - 139.56/v^2} \qquad (7)$$

Therefore, the determination as to whether h is above or below 0.5 must be made on physical grounds or by making paired velocity measurements where h is changed between measurements. The direction of the associated change in velocity (increasing or decreasing) can then be used to indicate whether h is above or below 0.5. Therefore, the volumetric fraction of tissue and gas in the lung and hence lung density can be determined directly from measuring the velocity of sound as it propagates through the tissue.

The sound may be introduced in any non-invasive manner, such as by percussion, or using any mechanical, electrical or other transducer which is capable of generating acoustic sounds. It is preferable that the sound which is introduced to the tissue possesses properties which allow it to easily be distinguished from environmental noise which may be present. Examples may include a single tone or a sinusoidal wave. In a preferred embodiment of the invention, a pseudo-random noise is produced by an electro-acoustic transducer and introduced into the tissue. The transducer is preferably attached to the surface of the biological tissue through which sound velocities are being measured. It is preferred that the pseudo-random noise signal which is used has characteristics which are similar to a white noise signal, but with mathematical properties which allow its amplitude to be defined at any moment in time. Furthermore, it is preferred that introduction of the pseudo-random noise signal to the tissue occurs in bursts, preferably of 0.1 to 20 seconds duration, and the sounds are produced preferably with frequencies which range from 20 Hz to 25kHz and at a sound pressure level of between 1 and 100 Pascal.

The sound can then be recorded at a location spaced from the position at which the sound is introduced, preferably on the surface of the biological tissue which is spatially distinct from the location of the transducer, using a sound detection means such as a microphone or a vibration detector, such as an accelerometer, which has a frequency response that is flat in the acoustic region, preferably between 20 Hz and 25 kHz. It is preferred that there are at least two of these detectors used to measure the sound, wherein one detector is positioned near a sound-generating acoustic transducer, and another is located at a position spaced from the first position of the tissue being assessed. This enables the sound pressure level, phase, and frequency content of the signal which is produced by the acoustic transducer (the input signal) to be accurately defined before it is detected by the spatially separated second detector. Placement of the second detector is preferably substantially in line with the acoustic transducer and the first detector.

The detector or preferably a microphone output may be amplified using low noise isolation amplifiers and band-pass filtered with cut-off frequencies and roll-off characteristics which depend on the acoustic properties of the tissue which is being assessed. For example, for measurements made on the neonatal lung, the pass band is preferably between 50Hz and 5KHz with a roll-off which corresponds to that of a 4^{th} order linear phase filter. These filters remove any very low frequency environmental noise (e.g. below 10Hz) that can adversely affect the performance of auto-scaling amplifiers into which the filtered signal may be fed.

The amplified output signal from the detector or microphone can then be processed by any means necessary, and a cross-correlation analysis of the input and output signals performed.

The cross-correlation function can be calculated using the output of the microphone which is located in close proximity to the acoustic transducer as the input signal, $x(t)$ and the output of the second microphone located on the other side of the tissue as the output signal, $y(t)$ wherein the cross-correlation function can be calculated as

$$R_{xy}(\tau) = \lim_{T\to\infty} \frac{1}{T} \int_0^T x(t)y(t+\tau)dt$$

where T is the observation time, and τ is the delay time between $x(t)$ and $y(t)$ at which $R_{xy}(\tau)$ is calculated.

The impulse response of the system in the time domain can also be determined. It is preferable that the impulse response then undergoes Fast Fourier Transformation so that the signal is transformed into the frequency domain and the transfer function of the tissue can be determined. This transfer function provides a quantitative indication of the characteristics of the tissue, wherein:

(a) the magnitude of the transform provides data relating to the transmission of the sound as it propagates through the tissue as a function of frequency (Rife and Vanderkooy, 1989); and

(b) the phase of the transform (after "unwrapping") can be used to calculate the phase difference, time delay and velocity of the sound for each frequency that is present in the psuedo-random noise signal which is introduced to the tissue by the acoustic transducer.

Commercially available acoustic hardware and software packages may be used to generate the psuedo-random noise signal, and to perform initial data processing. External noise which is not introduced to the tissue as part of the psuedo-random noise signal is strongly suppressed by the cross-correlation process thereby improving the quality of the measurements made.

A separate analysis of the relative transmission of the sound through the tissue can be used to identify resonant and anti-resonant frequencies of the tissue which is being assessed. Changes in these frequencies can then be used to assess regional differences in tissue topology which may be related to pathology.

Despite numerous experimental investigations (Kraman 1983, Goncharoff et al. 1989, Wodicka and Shannon 1990) of trans-pulmonary sound transmission where the source of sound is placed at the mouth, there has been no theoretical model which described sound transmission through the thorax. The present invention uses a simple model, based on the double wall transmission model that is used in architectural acoustics (Fahy 1985) to describe the sound attenuating effect of double walls separated by a compliant air layer, as is present in the lung.

The essential features of this model as it relates to the thorax can be represented by an electrical equivalent circuit that can be used to describe the pertinent features of sound transmission through the thorax. This model is illustrated in Figure 4(a). This approach to the analysis of acoustic transmission across the thorax facilitates analysis using sophisticated circuit emulation software such as SPICE to explore the effect of changing model parameters. In the equivalent electric circuit model where:

R_{cw} is the loss component associated with the chest wall and parenchyma;

M_{cw}, M_p is the surface mass of the chest wall and parenchyma respectively;

C_{gl} is the lung gas compliance;

P_{in}, P_o are the acoustic input and output sound pressure levels respectively; and

R_0 is the acoustic impedance of free space (414 MKS Rayls).

As illustrated in Figure 4(b), the model can be used to simulate the effect that changing R_{cw} has on the transfer function of the equivalent circuit which represents the chest. This transfer function can be described mathematically as $P_o(f)/P_{in}(f)$ where f is the frequency and $P_{in}(f)$ and $P_o(f)$ are the input (transducer) and output (chest microphone) sound pressure levels (SPL) respectively. As R_{cw} is decreased, the transfer function becomes progressively more peaked or resonant as illustrated by curves 1 to 3 in figure 4(b).

At sufficiently high frequencies, the output sound pressure level for all three curves falls asymptotically at a rate of 60dB per decade. As the frequency is increased above the resonant frequency, the response is dominated by the inertial mass of the proximal and distal chest walls, and the shunt gas compliance of the lung. These act together to produce the 60 dB per decade fall-off, such that the thorax is, in effect, acting like a third order low-pass electrical filter. Analysis of the equivalent circuit, neglecting losses, shows that the resonant frequency of the thorax, f_0, can be determined using:

$$f_0 = \frac{1}{2\pi}\sqrt{\frac{2}{C_{gl}(M_{cw}+M_p)}} \qquad (8)$$

Furthermore, if the transfer function is measured at f_0 and at another frequency well above f_0, say, $3f_0$ then using an analysis of the equivalent circuit, an explicit expression for lung gas compliance, C_{gl} ,can be deduced in the form

$$C_{gl} = \frac{4.18\times10^{-2}G}{f_0} \qquad (9)$$

where $G = |P_o(f)/P_{in}(f)|$ and is the magnitude of the transfer function of the thorax measured at $3f_0$. This equation has been verified using SPICE simulation.

It follows that gas volume V_{gl} can be computed using equation 9:

$$V_{gl} = \gamma P_0 C_{gl} \qquad (10)$$

where γ is the adiabatic gas constant and P_0 is the atmospheric pressure

A further important application of this model is illustrated in Figures 5(a) and 5(b). Figure 5(a) shows the experimentally measured thorax transfer function in a preterm infant soon after delivery but before surfactant administration (pre) and after the administration of surfactant (post) (Sheridan 2000). There is a steep fall-off in sound transmission for frequencies above 1000 Hz pre-surfactant and the leftward shift of this fall-off accompanied by an increase in attenuation of 10 dB following surfactant administration. A similar 10dB change

can be simulated in the model by increasing C_{gl} by about a factor of three while maintaining other parameters constant as illustrated in Figure 5(b). Although a measurement of lung gas compliance was not made during these experiments, and is not feasible using currently available technology, it would be expected that such an increase in compliance (associated with an increase in gas volume) would occur after surfactant administration.

An important component of acoustic transmission which can be modelled using the equivalent electric circuit is the loss component Rcw illustrated in Figure 4(a) which includes acoustic loss in the chest wall and parenchyma. Because the chest wall is acoustically thin, the dissipative loss in the wall is negligible but the loss in the parenchyma, which includes a large number of serial mass-compliance interfaces formed from the tissue and gas comprising the parenchymal structure, may be considerable. One model that has been proposed to account for acoustic loss in the parenchyma comprises air bubbles in water, for which an analysis already exists. In this model, absorption occurs because acoustic work is required to alternately compress and expand these bubbles.

It has been shown (Wodicka 1989) that the plane wave attenuation produced by N bubbles over distance x is given by:

$$P(x) = P_0 e^{-(\frac{N\sigma}{2})x} \qquad (11)$$

where $\sigma = 16\pi^2 r_o^4 \rho_t c_t R / \{R^2 + (\omega M - 1/\omega C)^2\}$

$P(x)$ is the SPL at x

P_o is the SPL at $x = 0$

r_0 bubble radius

c_t sound speed in tissue

R, M, C are the effective mechanical resistance, mass and compliance of the bubbles respectively

Attenuation, $\alpha = \frac{P(x)}{P_0}$ in dB/cm can then be written as:

$$\alpha = 4.35N\sigma \qquad (12)$$

This is a complex function of R,M,C but a simplified expression for the attenuation can be deduced by recognising that the acoustic vibration of the bubbles (alveoli) is dominated by bubble compliance C at frequencies which are much lower than resonance (ie. < ≈ 10kHz for realistic alveoli sizes). Therefore, attenuation can be reduced to:

$$\alpha = 2.36\times10^{-2} r_0^{\,6} f^3 N \qquad (13)$$

The number of bubbles per unit volume N is approximately related to the gas fraction $(1-h)$ by:

$$N = \frac{3(1-h)}{4\pi r_0^{\,3}} \qquad (14)$$

hence equation 13 can be written as

$$\alpha = 1.35\times10^{-3} \frac{f^3(1-h)^2}{N} \qquad (15)$$

From these equations, it can be seen that:

(a) absorption is related to the square of the gas fraction $(1-h)$; a small increase in the tissue fraction h is associated with a marked decrease in high frequency attenuation (Figure 6). This may explain the increased transmission of sound across the chest wall which can be observed clinically at high frequencies, following pneumonic consolidation of the lung; and

(b) attenuation is a strong function of both the frequency f and the alveolar radius r_0. This may explain, in part, the rapid fall-off in transmitted sound at high frequencies seen in both adult and neonatal subjects. The dependence on

bubble radius may explain the reduced transmission through the thorax during emphysema.

Furthermore, these equations indicate that:

(a) absorption is related to the square of the gas fraction (1-h); and

(b) sound transmission attenuation is s strong function of both the frequency and the alveolar radius.

Using these relationships between sound transmission velocity in tissues and the tissue characteristics themselves, it is possible to obtain a workable relationship between acoustic measurements and lung pathology or the pathology or condition of other biological tissues.

This method provides a virtually continuous real-time measurement of tissue characteristics by analysing the velocity and attenuation of a defined sound as it propagates through the tissue. The method is applicable in both adults and infants, and for humans and animals. In particular, the present invention can be used in the determination of respiratory conditions in infants who cannot cooperate with presently available conventional stethoscopic methods of respiratory condition analysis which requires vocal co-operation. It is also useful where the patient is critically ill, is unconscious, or is unable to respond or generate a sound which can be used to determine lung condition.

An embodiment of the present invention, provides a method of determining a state of the upper airways in a respiratory tract in a patient in situ, said method including:

introducing a sound at first position in the upper airways;

detecting the sound after it has travelled through the upper airways at another position spaced from the first position;

calculating the velocity and attenuation of the sound travelled through the upper airways from the first position to another position; and

correlating the velocity and attenuation of the sound to the state of the upper airways.

In a preferred embodiment, there is provided a method for monitoring airway patency including the steps of:

(a) introducing an audible sound signal into the airway at a first location;

(b) detecting the sound signal on the chest at a second location after the sound has travelled through at least a portion of the airway; and

(c) determining airway patency based on the attenuation of said sound signal between the first and second locations.

The state of the upper airways may include any condition of the upper airways such as obstructed or open airways. Measurement of the closure or collapse of the upper airway is particularly useful for conditions such as in obstructive sleep apnoea or OSA.

Apnoea, and particularly Obstructive Sleep Apnoea (OSA) is associated with closure of the upper airway and lapses in respiration during sleep. Using the present invention, a pseudo-random noise may be introduced into the airway using an acoustic transducer which conducts the sound from a location in the upper airway preferably via a Silastic nosepiece adapter. During normal respiration, the airway is open and the sound is transmitted via the airway to the lung via the trachea, where it subsequently propagates through the lung parenchyma and thorax to the surface of the chest. A sound-detection device such as a microphone may be attached in the chest region. Variations in the sound level which is measured at the chest region can then be used to model the degree of upper airway patency. The chest region may include the region extending from below the buccal cavity to below the lung.

Preferably, the microphone is placed on the upper chest region generally below the neck and just above the lung.

When the airway is closed, the transmission of sound through the tissue decreases so that it may be undetectable by a microphone located on the chest. Therefore, when the sound falls below a certain value, it is likely to indicate the closure of the airway. When the signal which is detected by a microphone detector or located on the chest region falls below a certain preset limit, an alarm is activated indicating obstruction of the airway. This alarm may

wake up the subject which will most often result in the subsequent reopening of the airway, or it may alert attending staff to a patient who is being monitored for OSA or any other airway dysfunction. There are several benefits associated with this method for detecting airway obstruction or closure which include:

(a) the technique is non-invasive;

(b) the technique can be used in new-borns and adults alike, and in humans or in animals;

(c) the technique monitors patency of the airway, not depletion of oxygen or lack of movement as is the case in other apnoea detection devices. As a result of this, the susceptibility of the subject to oxygen depletion is detected before depletion itself occurs, reducing the likelihood of discomfort and tissue damage which can be caused by extended lapses or pauses in regular respiration and oxygen deprivation. This method can also be used to set the optimal level of CPAP to apply to a patient in order to maintain airway patency.

Previous work shows that measurement of sound velocity alone may provide a technique for assessing lung density and gives an insight into the degree of lung inflation. However, no attempt has been made to evaluate the potential utility of sound velocity and attenuation as a clinical tool.

Lung condition may be selected from the group including but is not limited to:

(a) lung tissue density;

(b) lung gas volume;

(c) regional collapse (atelectasis);

(d) regional blood volume, interstitial oedema ; and

(e) focal lung pathology such as tumour and global lung disease such as emphysema.

These lung conditions may then be compared with the condition of a normal, healthy lung.

To measure lung condition, the method of the present invention is preferably applied by introducing a sound to the thorax and hence to the lung preferably by applying an acoustic transducer to the thorax on one side of the chest and calculating the sound velocity and attenuation using a detector or microphone which is attached to the other side of the chest and which detects the transmitted sound. Previous measurements of lung condition or volume have been made by introducing sound to the lung tissue via the trachea. However, there are problems associated with this method for the lung which result from the unknown distance between the trachea and chest wall, and an inability to selectively distinguish the effects of the airway from the effects of the lung parenchyma on the velocity of the introduced sound. In other measurement techniques, the sound is generated by the subject by respiration, coughing or speech, or is introduced through percussion. However, this presents a key limitation because the acoustic properties of these sounds are subject-dependent and beyond control, particularly in the newborn infant, who is unable to reliably produce the desired sound on command.

The present invention exhibits a novel approach to examining the acoustic properties of the biological tissues, including the upper airways and of the thorax, by introducing sounds with a known and precisely defined spectral content as the investigative tool. For the lung, by utilising this sound which is introduced directly to the wall of the thorax, and by recording the sound after it is transmitted across the thorax, uncertainties associated with noise introduced via the trachea are eliminated. Without being restricted by the theory, research suggests that the lung tissue type which is primarily responsible for changes in sound velocity as it propagates through the thorax is the lung parenchyma; the contribution to changes in sound wave velocity and attenuation which is made by the airways is insignificant.

Many lung diseases are associated with characteristic features that can be detected using auscultation of the chest (Lowe and Robinson, 1970). In the normal lung, frequencies above 300-400Hz are heavily attenuated by thoracic tissue, and on auscultation, respiration sounds are soft, conversational sounds are muffled, and whispered sounds are inaudible. By contrast, pneumonic

consolidation greatly reduces the attenuation of high frequency sounds, resulting in characteristic respiration sounds known as 'bronchial breathing' and strong transmission of whispered (high-frequency) sounds known as 'whispering pectriloquy'. A pleural effusion on the other hand, classically gives rise to increased attenuation of low frequency sound, causing vocal sounds to have a high pitched nasal quality known as 'aegophony'.

Studies have been published which examine the effect of lung condition on sound attenuation in the healthy human lung. However, these studies have failed to measure the effect of lung inflation on sound attenuation. The present invention utilises transthoracically introduced sound and preferably measures the sound velocity and sound attenuation to determine lung condition. Lung conditions assessed using the present invention may include lung density and lung volume. However, other lung conditions may be determined by correlating changes in sound velocity and sound attenuation which are associated with known lung conditions with sound velocities and attenuation which are measured using a normal, healthy lung.

Tissue density may be measured using sound velocity alone. However, sound attenuation may also be introduced as a parameter for the determination of tissue density. Tissue density may be a measure of the amount of fluid or blood in the tissue. In the lung, it may also indicate gas volume, regional collapse (atelectasis), regional blood volume, interstitial oedema and both focal

lung disease (eg tumour) and global lung disease (eg emphysema) which may be compared with a normal, healthy lung.

Lung gas volume is inversely proportional to lung density and may be measured using sound velocity and preferably sound attenuation. Furthermore, measurement of the velocity of a sound as it propagates from one side of the thorax through the lung tissue to the other side of the thorax can be correlated with a change in lung volume (inflation). This may be done in isolation, or during or after clinical interventions which alter the degree of lung inflation.

Measurements taken may include:

(a) before and at intervals after treatment with surfactant;

(b) before and at intervals after commencement of Continuous Positive Airway Pressure (CPAP) to recruit lung volume in the presence of hyaline membrane disease and/or atelectasis;

(c) before and at intervals after the commencement of mechanical ventilation; and

(d) before and immediately after endotracheal tube suctioning.

The degree of change in the sound velocity and preferably also of sound attenuation may be used together to provide a more conclusive indication of the degree to which the lung is inflated. Lung inflation may be determined using a single measurement, or it may be determined continuously, thereby

enabling the monitoring of progress of lung disease and its treatment. This has particular value in the treatment and monitoring of lung disorders in premature babies over a period of time.

Abnormal lung density due to over-or under-inflation of the lung may be associated with increased lung injury and the propensity for development of chronic lung disease in infants. Therefore, measurements of sound velocity and attenuation (which relate to lung density) in a premature infant may allow inflation to be optimised and risk of chronic lung disease to be reduced.

Measurements of the sound velocity and sound attenuation may be made on days 1, 2, 3, 5, 7, 10 and 14 or any interval thereof and then at weekly intervals until about 36 weeks. As a comparison, and to complement measurements made using the present invention, absolute lung volume may be measured using the gold-standard and long-established helium dilution technique at the time of the acoustic measurements. Results taken from infants who subsequently develop chronic lung disease (defined either as oxygen dependency at 28 days or at a postmenstrual age of 36 weeks) may be compared with results from those who do not.

A similar technique can be used to assist in diagnosing lung disease wherein again, a sound is introduced to the thorax such that it travels from one side of the thorax, through the lung, to another side of the thorax. The sound velocity

and preferably attenuation which is measured is then compared with that of a normal, healthy lung. Since lung disease often manifests in reduced lung volume, a comparison can be used, again, to provide an indication as to whether a subject's lung exhibits a propensity for lung disease. Common lung diseases may include emphysema, asthma, regional collapse (atelectasis), interstitial oedema and both focal lung disease (e. g. tumour) and global lung disease (e.g. emphysema). Each of these may be detectable when measurements of the velocity and attenuation of a sound which is transmitted through a diseased lung is compared with that of a lung in normal condition.

The present invention provides a reliable method for monitoring lung density and volume *in situ*. However, it can also be used to provide a method of preventing lung injury by again, introducing a sound transthoracically so that the sound travels from one side of the thorax through the lung to another side of the thorax. The velocity of the sound can be measured as it travels from one side of the thorax through the lung to the other side of the thorax, and the measurement can be used to indicate the volume of the lung which can then be used in the maintenance of an optimal lung volume which is substantially free of atelectasis or over-inflation (volutrauma). These optimal lung volumes are illustrated graphically in Figure 1, wherein there exists a window inside which the possibility of causing lung injury can be minimised. This window is framed by under-inflation and over-inflation lung volumes. If lung volume is maintained inside this window, the likelihood of lung injury will be reduced.

However, to ensure the volume does not rise excessively and does not drop to the level of atelectasis, it is necessary to constantly monitor the lung's volume.

The present invention can be used to provide a monitoring system which measures sound velocity and preferably combines sound velocity data with measurements of sound attenuation in order to determine the level of lung inflation in a subject. Spectral analysis of the impulse response can indicate frequency components in the sound signal which are more prominent than others and which may be an indicator of pathological or abnormal tissue.

Preferably the lung condition is monitored by an independent measure of lung density or lung volume.

The benefits associated with the application and detection of acoustic signals to biological tissues is not limited to the lungs, airways and other tissues associated with respiration. The present invention can be used to detect densities of other porous structures and composite biological tissues which have high or low densities, wherein the ratio of solid to porous tissue gives rise to the change in velocity and sound attenuation which is measured.

The present invention will be more fully described with reference to the accompanying examples and figures. It is to be understood that the description following is illustrative only and should not be taken to be limiting in any way, or as a restriction on the generality of applications for the invention previously described.

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EXAMPLES

Example 1 – Measurement of Lung Volume in Adults

In 5 healthy adult subjects, the velocity and attenuation of sound which was transmitted from one side of the chest to another, in a range of frequencies from 50–1000Hz was measured at a number of defined positions on the chest. These measurements were taken while the lung volume was varied between Residual Volume (RV) and Total Lung Capacity (TLC). A reference position was established over the right upper zone of the chest. Using this position, a region in the frequency spectrum (around 100-125Hz) where sound attenuation was much reduced and where the degree of attenuation was directly related to lung inflation (see Fig 2A, upper panel) was found. The difference in attenuation between RV and TLC was approximately 7.5dB and statistically significant

(P=0.028). Further, it was found that sound velocity was low, averaging around 30 m/sec, and it showed a clear and strong sensitivity to the degree of lung inflation, being appreciably faster at TLC than at RV (Fig 2A, lower panel). In this study evidence was found which indicated that the effect of inflation on velocity and attenuation varies at different locations in the thorax, particularly in the lower zones. It is likely that this is, in part, attributable to the location of the heart and liver (at RV) in the sound path.

The method of analysis permits determination of phase shift, and therefore velocity as a function of frequency. This work has shown that the speed of sound in the lung parenchyma is dispersive, or frequency dependent, over the range of frequencies studied. This is of considerable importance, since it is theorised that the relationship between velocity and frequency is dependent on regional compliance and inertial (ie mass dependent) properties of the lung. These properties may provide valuable information about the lung since they are partly determined by the condition of the alveolar septum, the degree of fluid infiltration of the lung parenchyma, and the extent of atelectasis.

Preliminary pilot data were collected from newborn infants in the neonatal intensive care unit. Figure 2B represents a sample result from an infant of 26 wks gestation with healthy lungs, illustrating that measurements can be made using the present invention with a subject who cannot co-operate and who must be studied in the noisy intensive care setting. Interestingly, the frequency region over which sound attenuation is least in the newborn is higher (approximately 300Hz) than in the adult. In addition, although the relationship between velocity and frequency has a nadir at about 300 Hz compared with 125 Hz in the adult, the dispersive nature of sound velocity which is evident in the adult is also present in the infant.

Example 2 – Measurement of Lung Density in Rabbits

Experiments were conducted in 1-2 kg New Zealand white rabbits. These animals were chosen for their similarity in size to the human newborn and their widespread use as a model of neonatal surfactant deficiency. Animals were anaesthetised with intravenous thiopentone, before performing a tracheostomy during which a 3 mm endotracheal tube was inserted into the airway to allow ventilation using a conventional neonatal ventilator (Bournes BP200). Maintenance anaesthesia was be achieved with intravenous fentanyl. The chest was shaved and a microphone and transducer secured in various pre-defined positions, including a reference position over the right upper chest. The animal was then placed in a whole body plethysmograph to monitor absolute lung gas volume at intervals throughout the experiment. Tidal volume was monitored continuously with a pneumotachograph attached to the tracheostomy tube. The sound velocity and attenuation was determined at each location of the where a microphone was situated, and each observation was the average of 10 repeated measures.

The effect of changes in lung density as a result of lung disease on sound velocity and attenuation was examined by comparing results from 3 groups of rabbits with differing lung conditions:

Group 1 – Normal lungs (n=10)

Group 2 – Lungs rendered surfactant deficient by saline lavage (n=10)

Group 3 – Lungs rendered oedematous by inflation of a left atrial balloon catheter (n=10).

Within each group of animals the effect of changes in lung density, resulting from changes in degree of lung inflation, was examined by making measurements under dynamic and static conditions.

(1) Dynamic measurements during mechanical ventilation. Sound velocity and attenuation may be measured during mechanical ventilation at various levels of positive end-expiratory pressure (PEEP) including 0, 5, 10, 15 and 20 cmH_2O. Absolute lung volume at end expiration, and tidal volume may be

determined for each level of PEEP. A wide range of PEEP can be employed to ensure that observations are made over a wide range oflung volumes, from under-inflation to over-inflation and including optimal inflation.

(2) *Static measurements during apnoea.* Sound velocity and attenuation was measured while the lung was transiently held at constant volume after spontaneous respiratory effort had been suppressed by a brief period of hyperventilation. Various lung volumes from below functional residual capacity (FRC) to TLC were achieved by varying airway pressure between −10 and +30cmH_2O. Studying the lung under static conditions allows observations to be made at the extremes of lung volume. These results were directly comparable to observations during breath-hold in adult subjects and enables verification of the cross-correlation technique used in the present invention which increases the system's robusticity against interference from breath sounds.

(3) *Static measurements post-mortem.* At the completion of (2) above, a lethal dose of anaesthetic was administered and observations of sound velocity and attenuation were repeated across the same range of lung volumes as in (2). The trachea was then clamped at an inflation pressure of 10 cmH_2O before dissecting the lungs so that they were free from the chest and so that their weight and density could be determined. In order to address the question of the regional differences in sound velocity and attenuation observed in the adult human study, final measurements were made of the acoustic properties of the excised lung at the same levels as those studied in the intact thorax. An important aspect of this analysis is that it allowed comparison of results obtained before and after death to establish whether the cross-correlation technique used is resistant to interference from cardiac sounds.

Example 3 – Measurement of Lung Inflation in Infants

To be a valuable clinical tool, measurements of sound velocity and attenuation must be sensitive to changes in lung inflation that are of a clinically relevant magnitude. A test of whether measurable changes in sound velocity and attenuation which occurred after clinical interventions which were confidently

predicted alter the degree of lung inflation was conducted. It was found that clinical interventions which cause a significant change in lung inflation are associated with changes in sound transmission and velocity which are measurable using the present invention.

Example 4 – Prediction of Chronic Lung Disease

It is also necessary to determine whether evidence from acoustic measurements of abnormal lung density are indicative of either under-inflation or over-inflation, and associated with development of chronic lung disease as a result. It was found that abnormal lung density in the first few days of life was more common in infants who subsequently developed chronic lung disease than in those who did not. Serial measurements of sound velocity and attenuation in a population of pre-term infants (n=30) who, by virtue of their gestation (<30 weeks), are at high risk of developing chronic lung disease were made. In this population and using the present invention, it was estimated that about 65% of the population will still be oxygen dependent at 28 days of age, and about 30% will still be oxygen dependent at a postmenstrual age of 36 weeks.

Finally it is to be understood that various other modifications and/or alterations may be made without departing from the spirit of the present invention as outlined herein.

The claims defining the invention are as follows:

1. A method for monitoring airway patency including the steps of:

(a) introducing an audible frequency sound signal into the airway at a first location;

(b) detecting the sound signal on the chest at a second location after the sound signal has travelled through at least a portion of the airway; and

(c) determining airway patency based on the attenuation of said sound signal between the first and second locations.

2. A method according to claim 1, wherein the airway is the upper airway which includes the buccal region extending to the trachea before entering the lungs.

3. A method according to claim 1 or claim 2, wherein the first airway location is in a nasal region of a subject.

4. A method according to any one of claims 1 to 3, wherein the second location is in the chest region of a subject, which region includes a region extending from below the buccal cavity to below the lung.

5. A method according to claim 4, wherein the second location is in the upper chest region, generally below the neck and just above the lung.

6. A method according to any one of claims 1 to 5, wherein the introduced

audible signal is pseudo-random noise.

7. A method according to any one of claims 1 to 6, wherein determining airway patency involves a cross-correlation analysis of the detected sound signal with the introduced sound signal.

8. A method according to any one of claims 1 to 7, wherein the audible frequency sound signal has frequencies in the range of 20 Hz to 25kHz.

9. A method according to any one of claims 1 to 8, wherein the introduced audible frequency sound signal has a sound pressure level between 1 and 100 Pascal.

10. A method according to any one of claims 1 to 9, further including the step of detecting when the sound is attenuated below a pre-defined threshold level, said attenuation being indicative of obstruction of the airway.

11. Apparatus for monitoring airway patency, the apparatus including:

(a) sound generating means for applying an audible frequency sound signal in the airway at a first location;

(b) sound detection means for detecting the sound signal on the chest at a second location after the sound signal has travelled through at least a portion of the airway; and

(c) analysis means for determining airway patency based on the attenuation of said sound signal between said first and second locations.

12. Apparatus according to claim 11, wherein the airway is the upper airway

which includes the buccal region extending to the trachea before entering the lungs.

13. Apparatus according to claim 11 or claim 12, wherein the sound generating means is adapted to apply said sound signal at a first location in the nasal region of a subject.

14. Apparatus according to any one of claims 11 to 13, wherein the sound detection means is adapted to detect the sound signal at a relevant location that is in the chest region, which region includes a region extending from below the buccal cavity to below the lung.

15. Apparatus according to claim 14, wherein the sound detection means detects the sound signal in the upper chest region, generally below the neck and just above the lung.

16. Apparatus according to any one of claims 11 to 15, wherein the sound generating means generates an audible frequency sound signal which is pseudo-random noise.

17. Apparatus according to any one of claims 11 to 16, wherein the analysis means includes correlation means for performing cross-correlation of detected sound signals with sound signals of the generating means to determine an amount of signal attenuation during travel from the first airway location to the second location.

18. Apparatus according to any one of claims 11 to 17, wherein the sound generating means generates a sound signal at a sound pressure level between 1 and 100 Pascal.

19. Apparatus according to any one of claims 11 to 18, further including a threshold detector to determine when the sound has been attenuated to below a threshold level which is indicative of obstruction of the airway.

20. Apparatus according to claim 19, further including an alarm configured to sound when the sound attenuation falls below the threshold level.

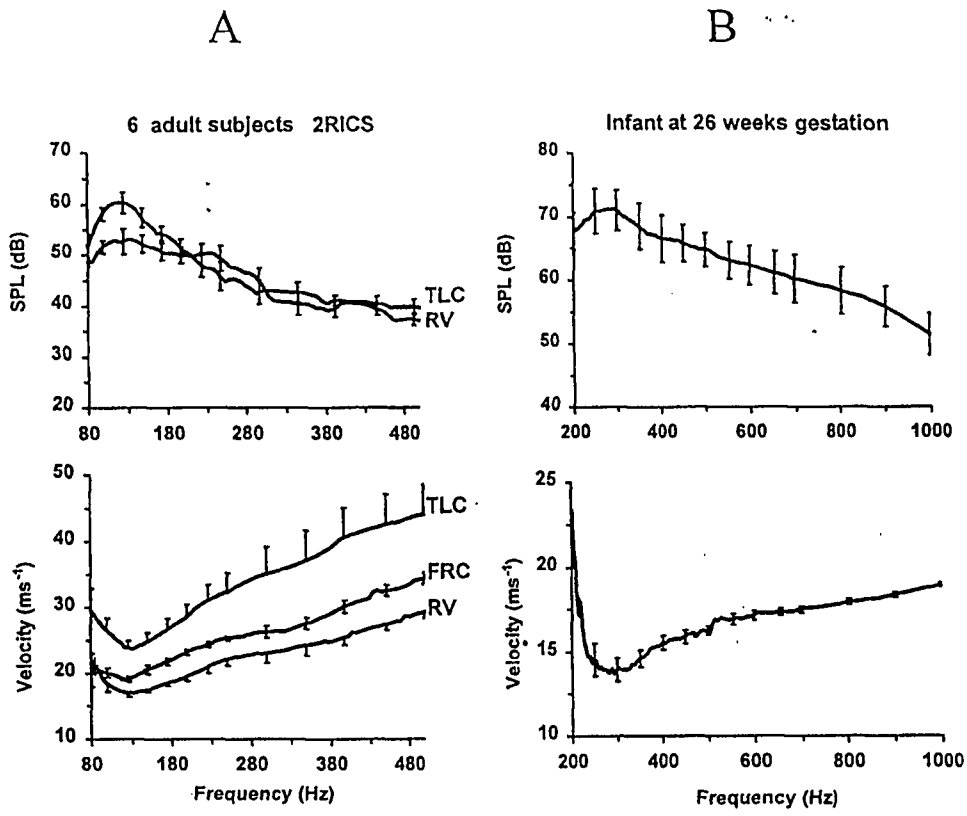
21. CPAP delivery apparatus for delivering continuous positive airway pressure to a subject, the CPAP apparatus including airway patency monitoring apparatus that includes means for applying sound to the airway of the subject and means for detecting on the chest of the subject the attenuation of the applied sound after transmission through at least a portion of the airway.

Figure 1

Fig 1. Pressure-volume curve of a moderately diseased lung illustrating two hazards of lung volume

with an optimal "safe" window between (from Froese, 1997).

FIGURE 2



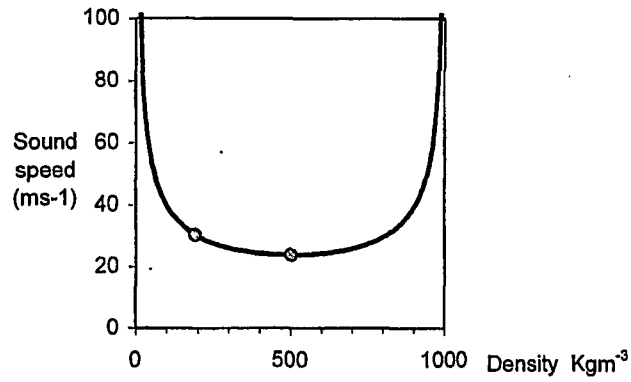


Figure 3

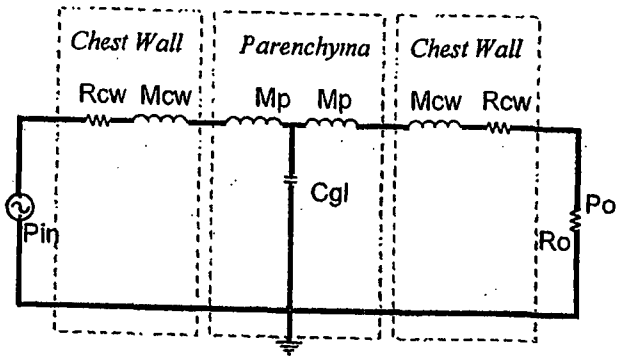


Figure 4(a)

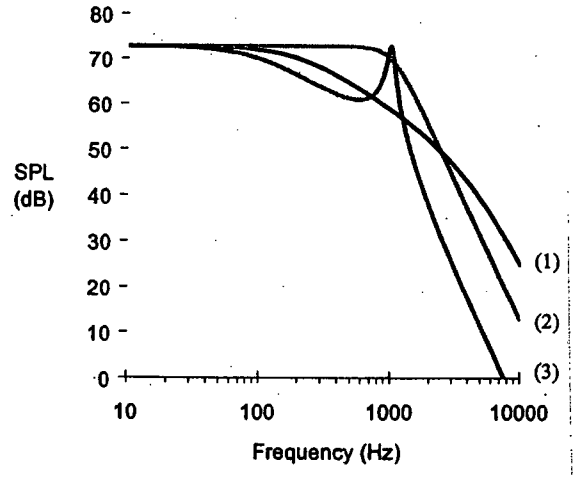


Figure 4(b)

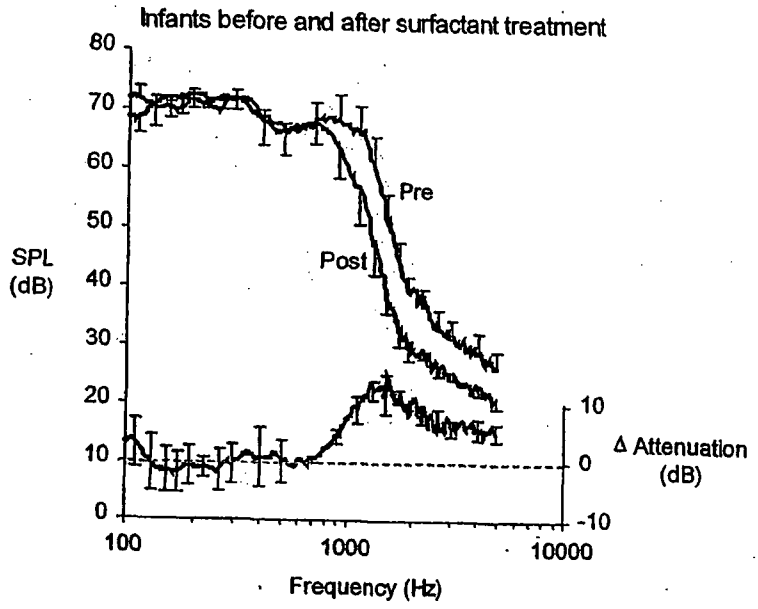
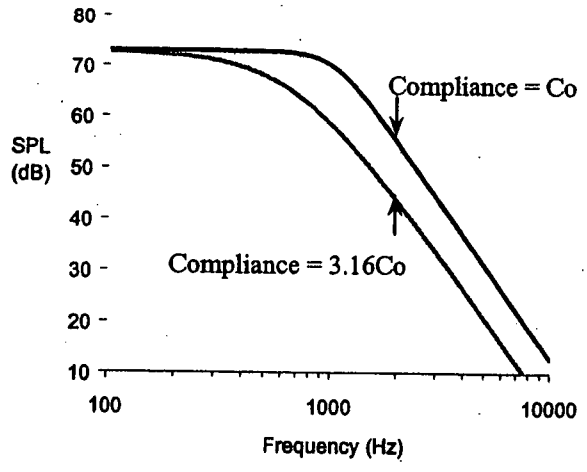


Figure 5(a)

Figure 5(b)



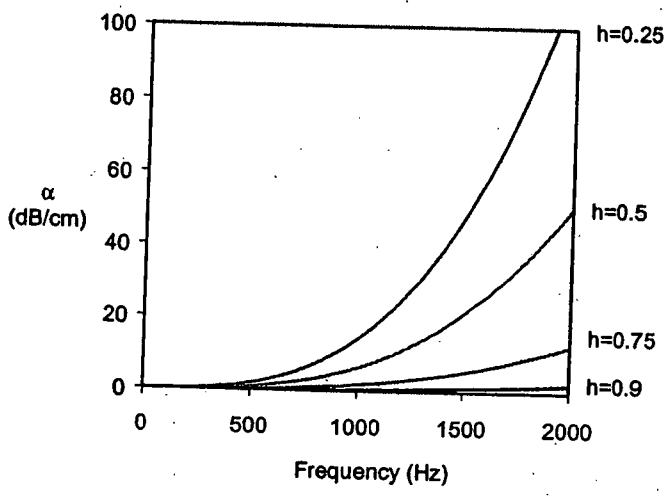


Figure 6