US006261238B1

(12) **United States Patent** (10) **Patent No.:** **US 6,261,238 B1**
Gavriely (45) **Date of Patent:** **Jul. 17, 2001**

(54) **PHONOPNEUMOGRAPH SYSTEM**

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(*) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 0 days.

(21) Appl. No.: **09/269,513**

(22) PCT Filed: **Sep. 30, 1997**

(86) PCT No.: **PCT/IL97/00318**

§ 371 Date: **Jun. 1, 1999**

§ 102(e) Date: **Sep. 30, 1997**

(87) PCT Pub. No.: **WO98/14116**

PCT Pub. Date: **Apr. 9, 1998**

Related U.S. Application Data

(63) Continuation-in-part of application No. 08/729,651, filed on Oct. 4, 1996, now Pat. No. 6,168,568.

(51) **Int. Cl.⁷** **A61B 5/08**
(52) **U.S. Cl.** **600/532**; 600/538; 600/586
(58) **Field of Search** 600/525, 532–4, 600/538, 586

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Primary Examiner—Robert L. Nasser
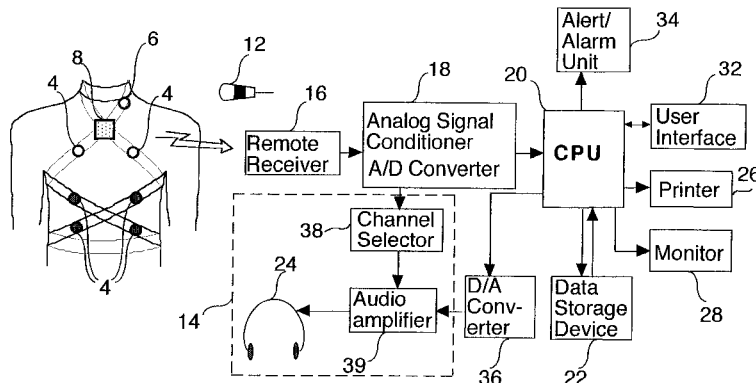(74) *Attorney, Agent, or Firm*—Fenster & Company Patent Attorneys, Ltd.

(57) **ABSTRACT**

A method of analyzing breath sounds produced by a respiratory system, the method comprising: measuring breath sounds produced by the respiratory system; tentatively identifying a signal as being caused by a breath sound of a given type if it meets a first criteria characteristic of the breath sound of the given type; and confirming said identification if a tentatively identified signal meets a second criteria characteristic of the breath sound of the given type.
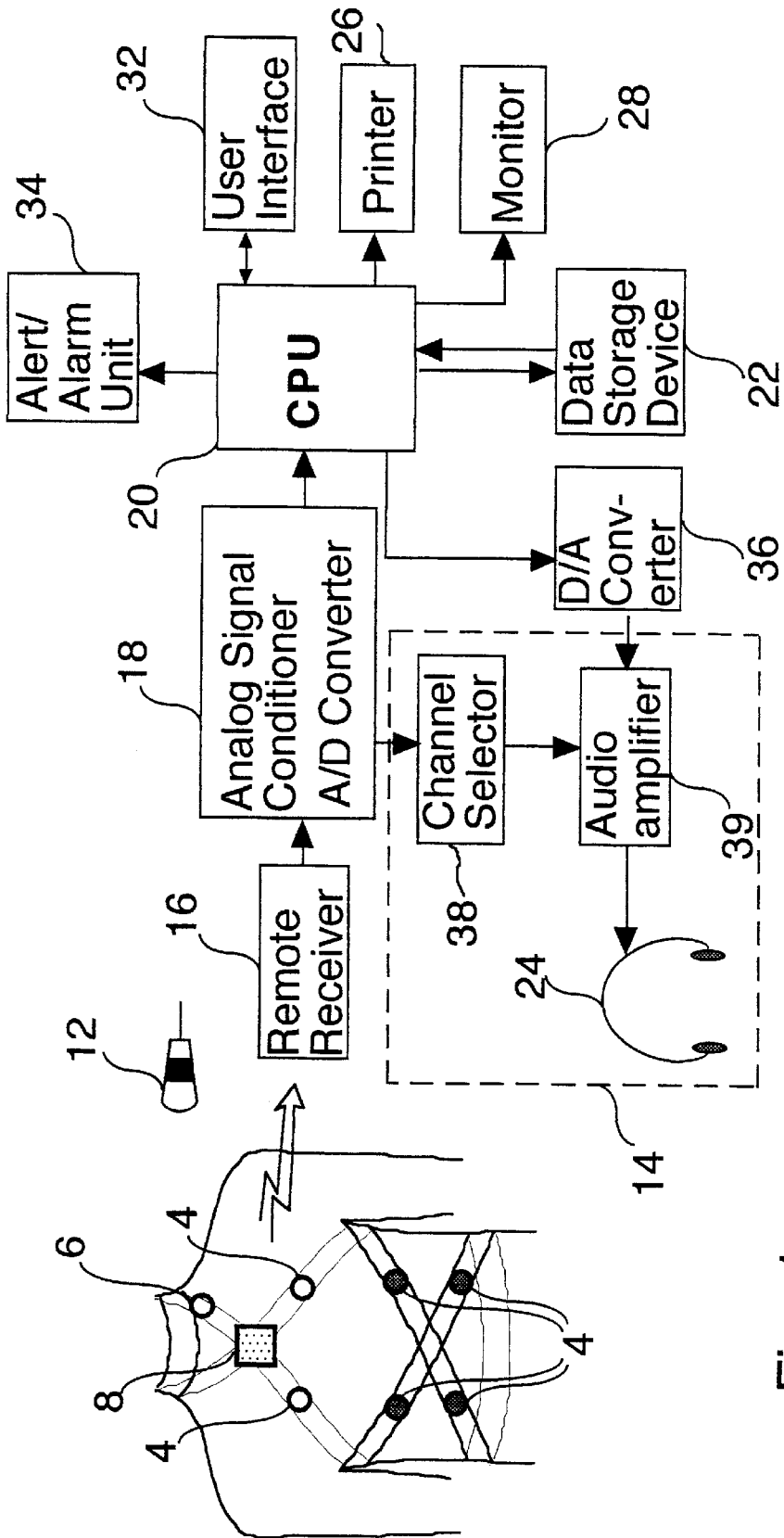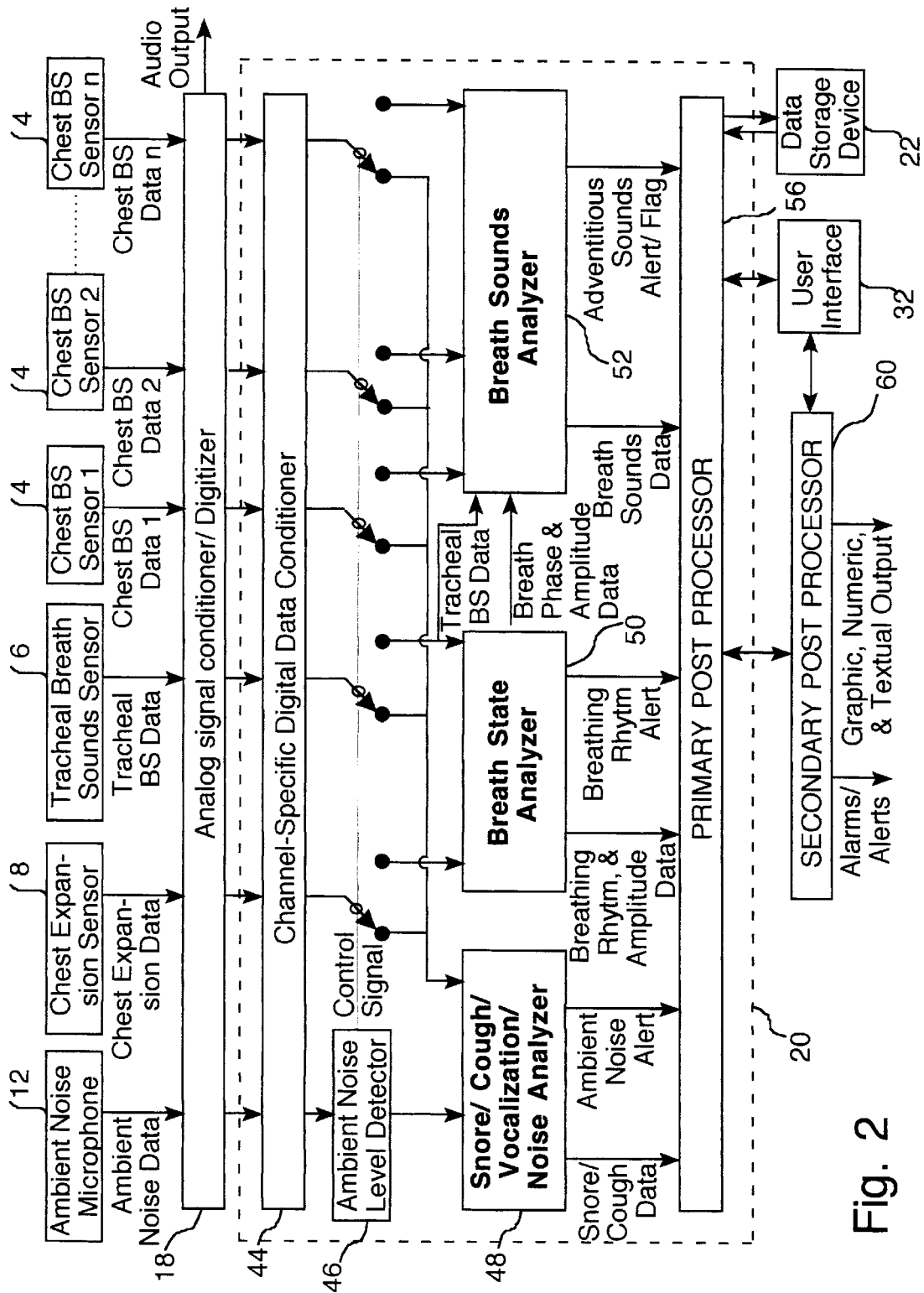
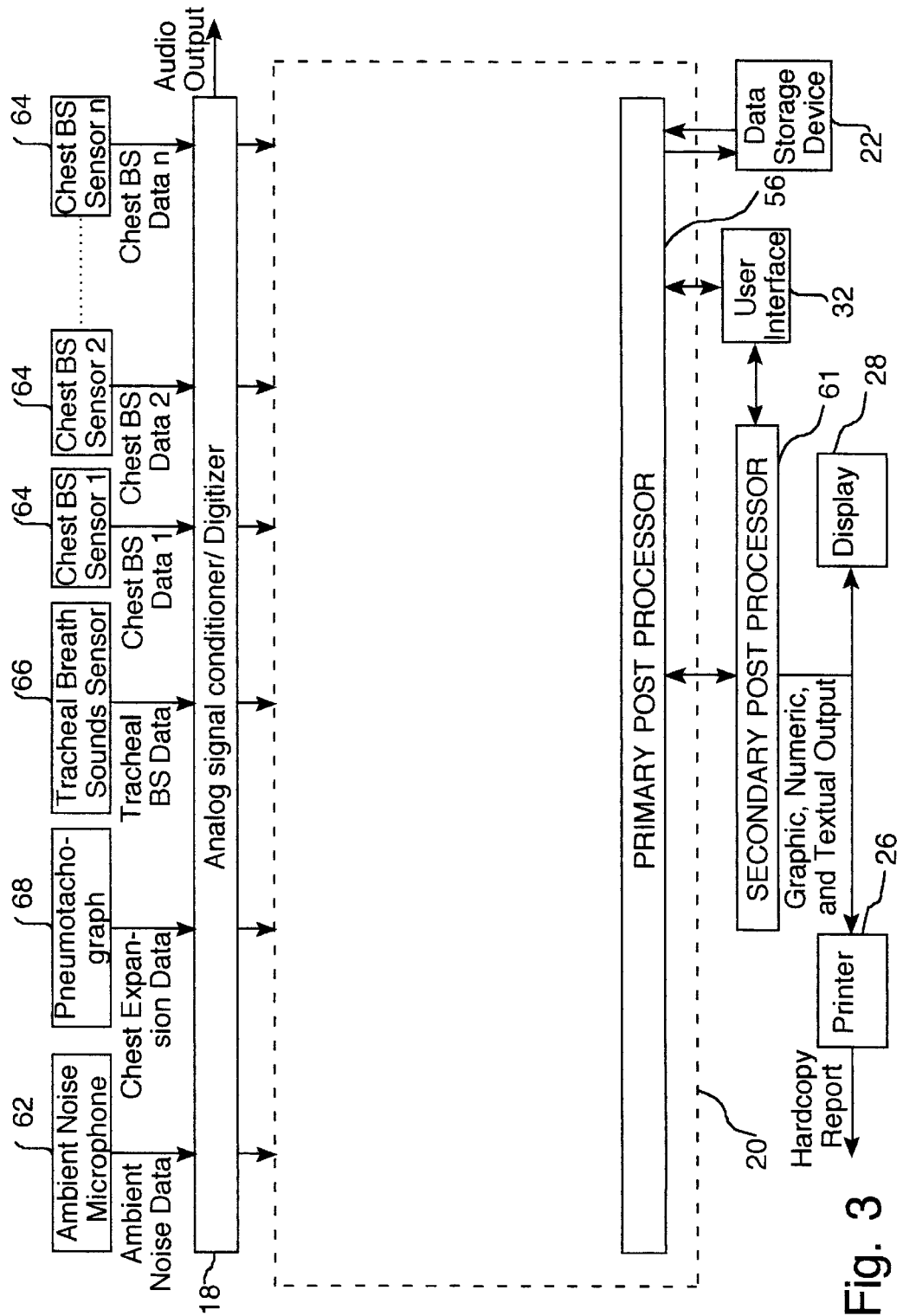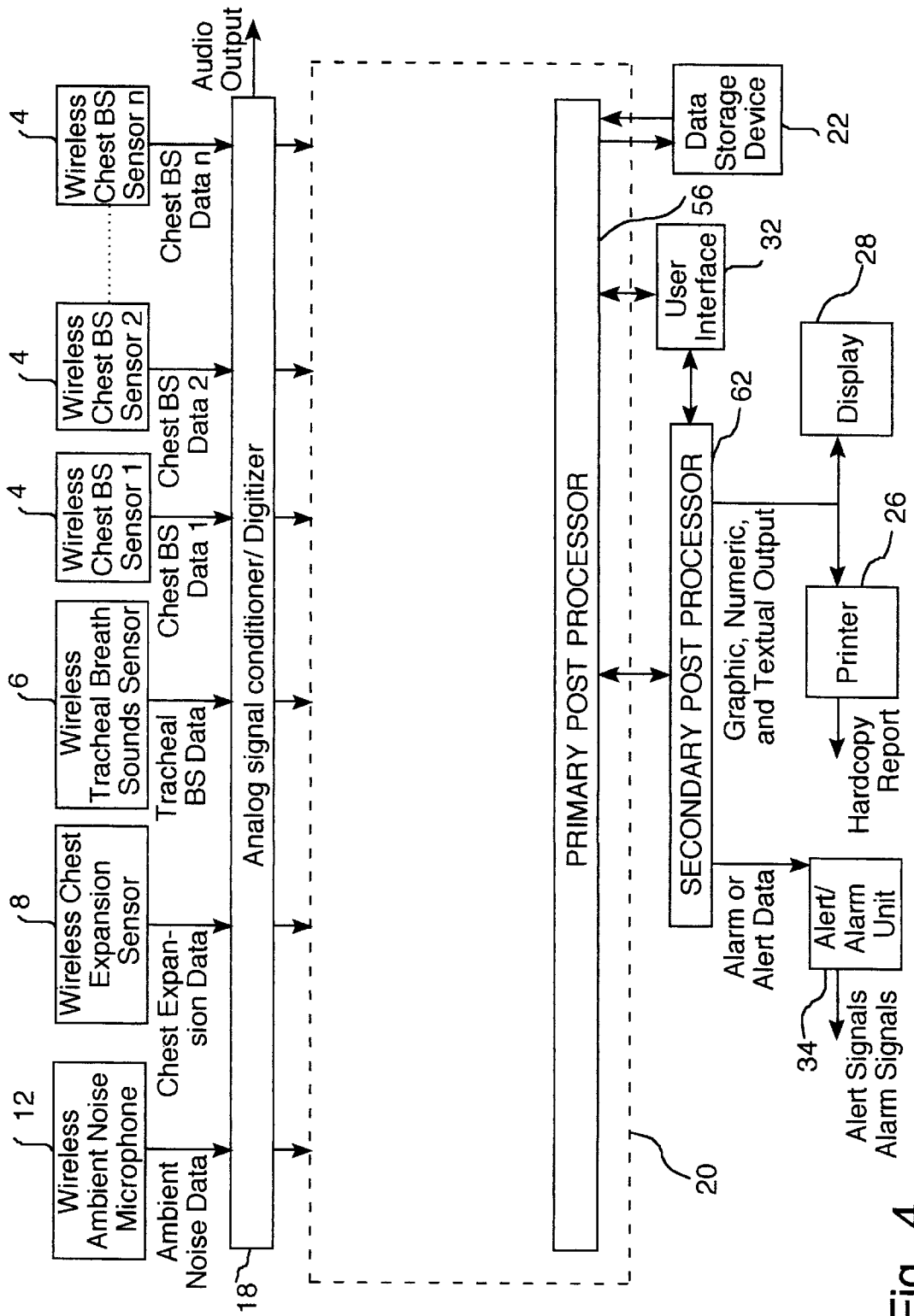**65 Claims, 30 Drawing Sheets**

Fig. 1

Fig. 2

Fig. 3

Fig. 4

Fig. 5

Fig. 6

Fig. 7A

Fig. 7B

Fig. 7C

Fig. 8A

Fig. 8B

Fig. 9A

From D
and F

99
Next

A

Pick up chest expansion	Pick up ambient noise	Pick up Tracheal BS
110	112	114
Ampify and filter	Amplify and filter	Ampify and filter
116	118	120
Digitize	Digitize	Digitize
122	124	126
Calculate smoothed Impedance=Ii	Rectify & integrate = Ai	Rectify & integrate = Ti
128	130	132
To B.134	To C.136	To D.158

From C.136

Fig. 9B

From A.128

From A.130

134 — del(I)= Ii-Ii-1

136

Ai > Threshold ?

No → **To A.132**

Yes ↓

| del(I) | > Threshold ?

138

No

Yes

Set Pi=0

140

del(I) > zero ?

142

No

Set Pi=-1

146

Yes

Set Pi=1

144

B

Activate ambient noise alert

C   152

G

Activate snore/ cough/ stridor Method

From E.160

From E.168

To F.148

Fig. 9C

Fig. 9D

From
D.158

```
              Ti
               >
           High volume ────No──→  Calculate ──── 162
           Threshold            spectrum
               ?
         160                         │
              Yes                    ▼
                               Curve fit ──── 164
                               to model
                                     │
                                     ▼
To G                           Calculate ──── 166    ──── E
                               Chi and R
To G                                 │
                                     ▼
                    Yes           Chi/R  ── 168
←─────────────────────────          >
                                  Alpha
                                    ?
                                    │
         172                        No
  Reset apnea clock.          │
  Activate breath    No   Si-1 =      Set  ──── 170
  length clock     ←──     +          Si=+
  Register prev.           ?
  apnea length. ── 174
To                              Yes
F.178    │
         ▼
     Advance         ←─────────────
     breath clock ── 176
```

Fig. 9E

To A.99

F

Activate breath length alert

186

Activate hardware alarm

184

No

Hardware OK ?

182

Yes

Yes

Breath duration Tbr > t ?

180

No

From E.176

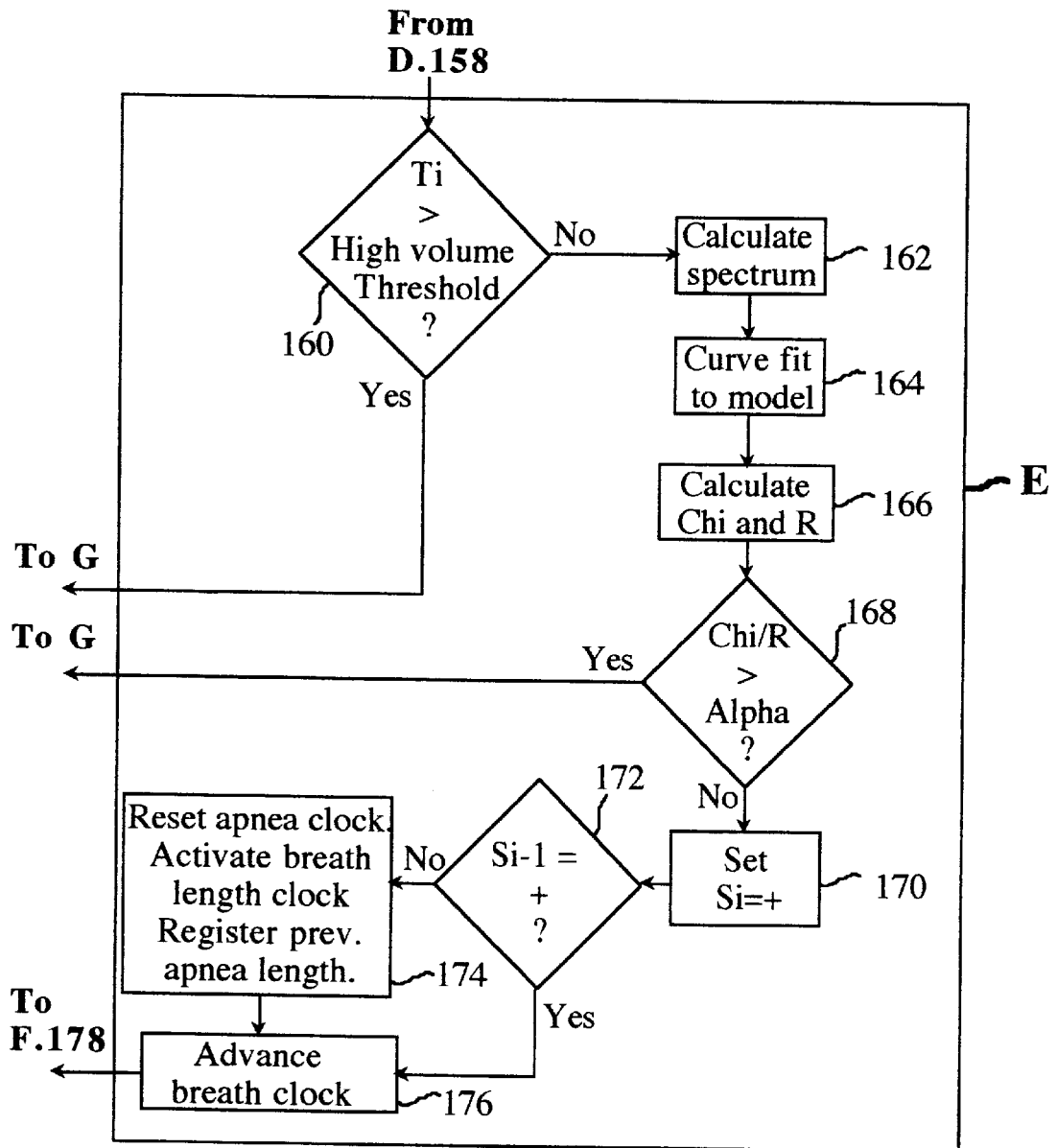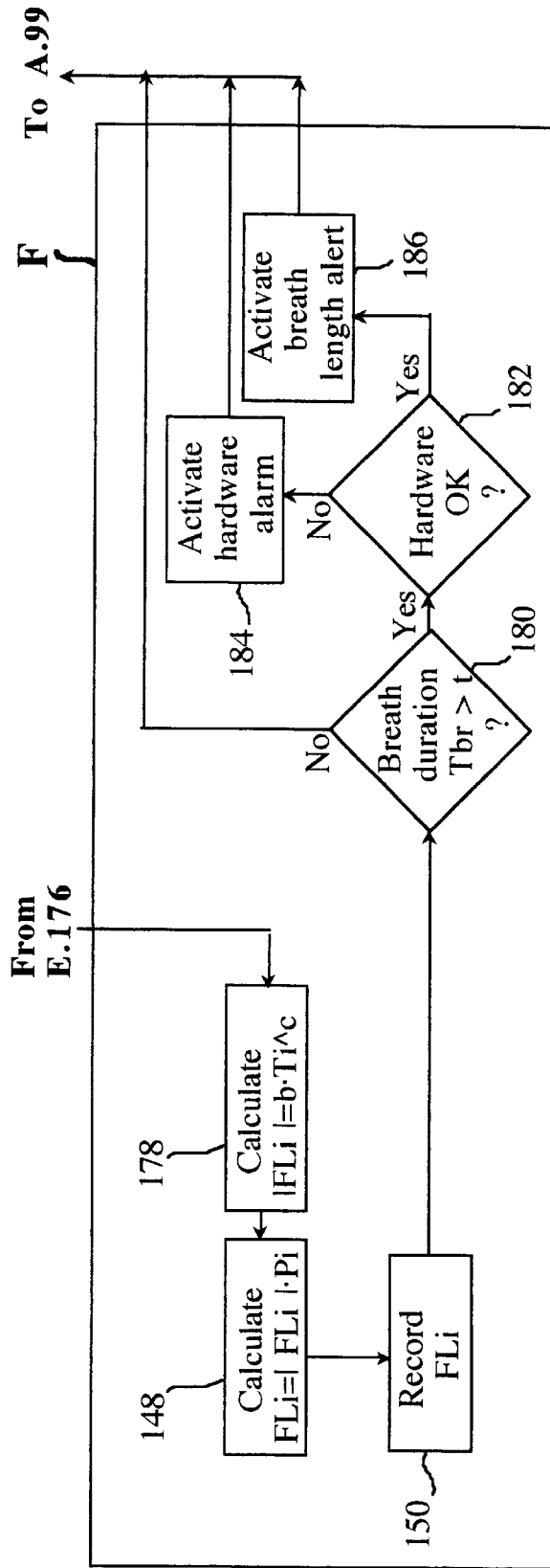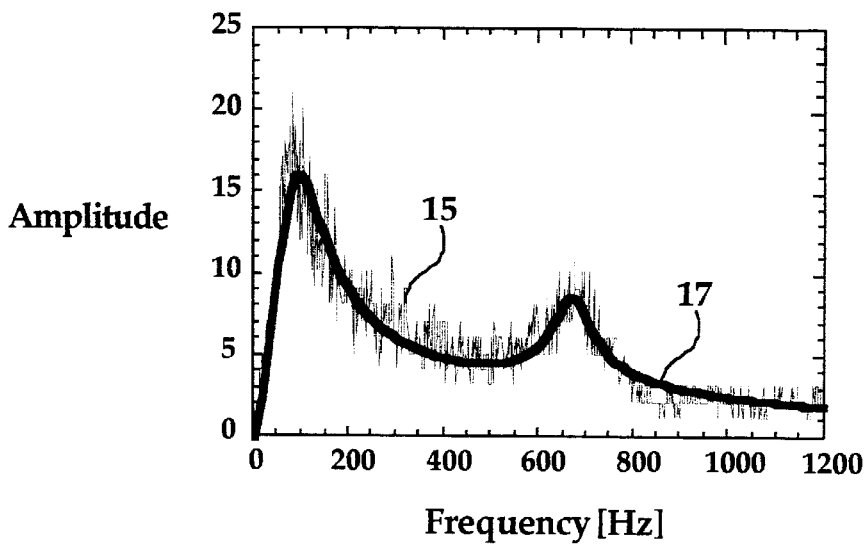Calculate IFLi I=b·Ti^c

178

Calculate FLi=I FLi I·Pi

148

Record FLi

150

Fig. 9F

Fig. 10

Fig. 11D

Fig. 11A

Fig. 11B

L

From J.264 →

Is previous segment a breath ? — **266**

Yes → Subtract average BG spectum from average BS spectrum — **270**

No ↓

Average BG spectra — **268**

From K.298 →

↓ Defilter BS — **274**

276 — Calculate BS curve fit

Is goodness of fit acceptable ? — **278**

280 — Are parameters within normal bounds ?

Yes → Record parameters — **282**

294 ↓ Reset BS aver. spectra

Record parameters — **286**

Yes → is hardware OK ? — **284**

292 Activate normal BS alert ← Activate wheeze/rhonchi detector — **290**

No ↓

Activate hardware alert — **288**

To H.250 ↓

Fig. 11C
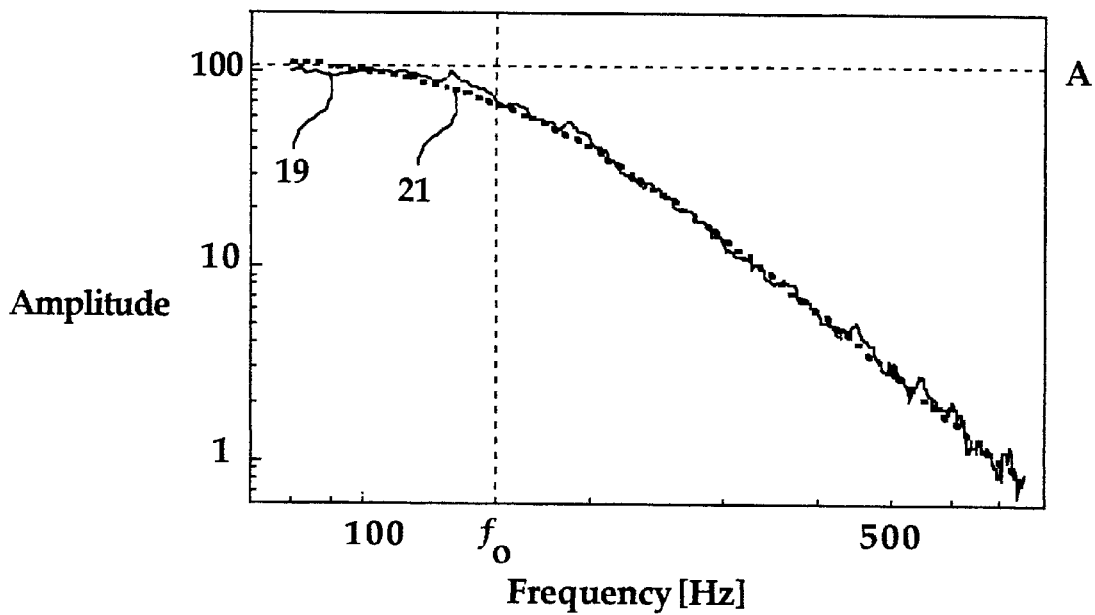
Fig. 12

Fig. 13A

Fig. 13B

Fig. 14

Obtain Ai data segment and Si data segment — 400

Calculate transfer function between Si and Ai data — 404

Calculate coherence of transfer function — 406

Do high coherence frequency ranges correspond to the peaks in the amplitude spectrum of Si ? — 408

Yes

Go to Snore/ vocalization Methods — 410

No

Record rhonchi parameters — 412
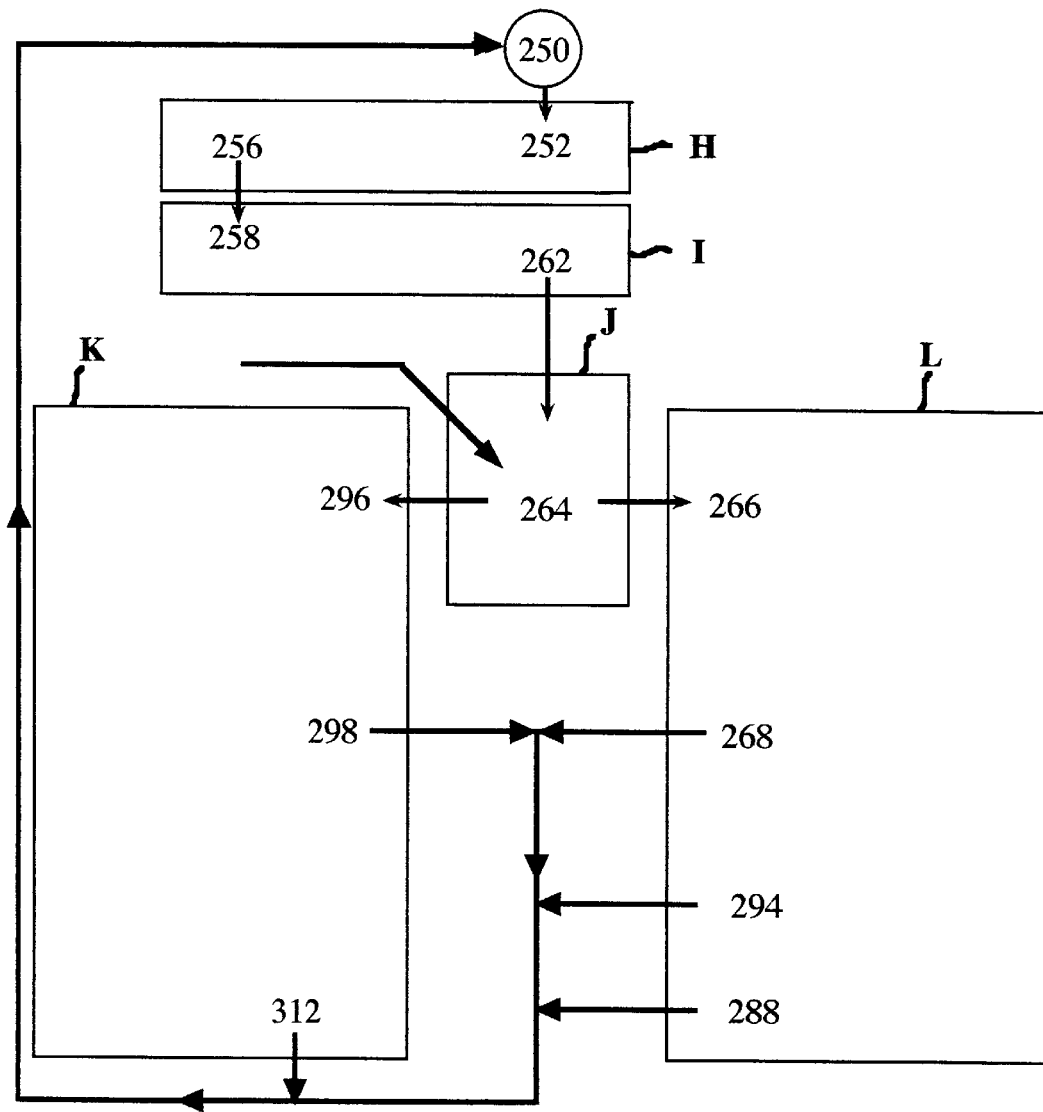
Breath phase and flow data
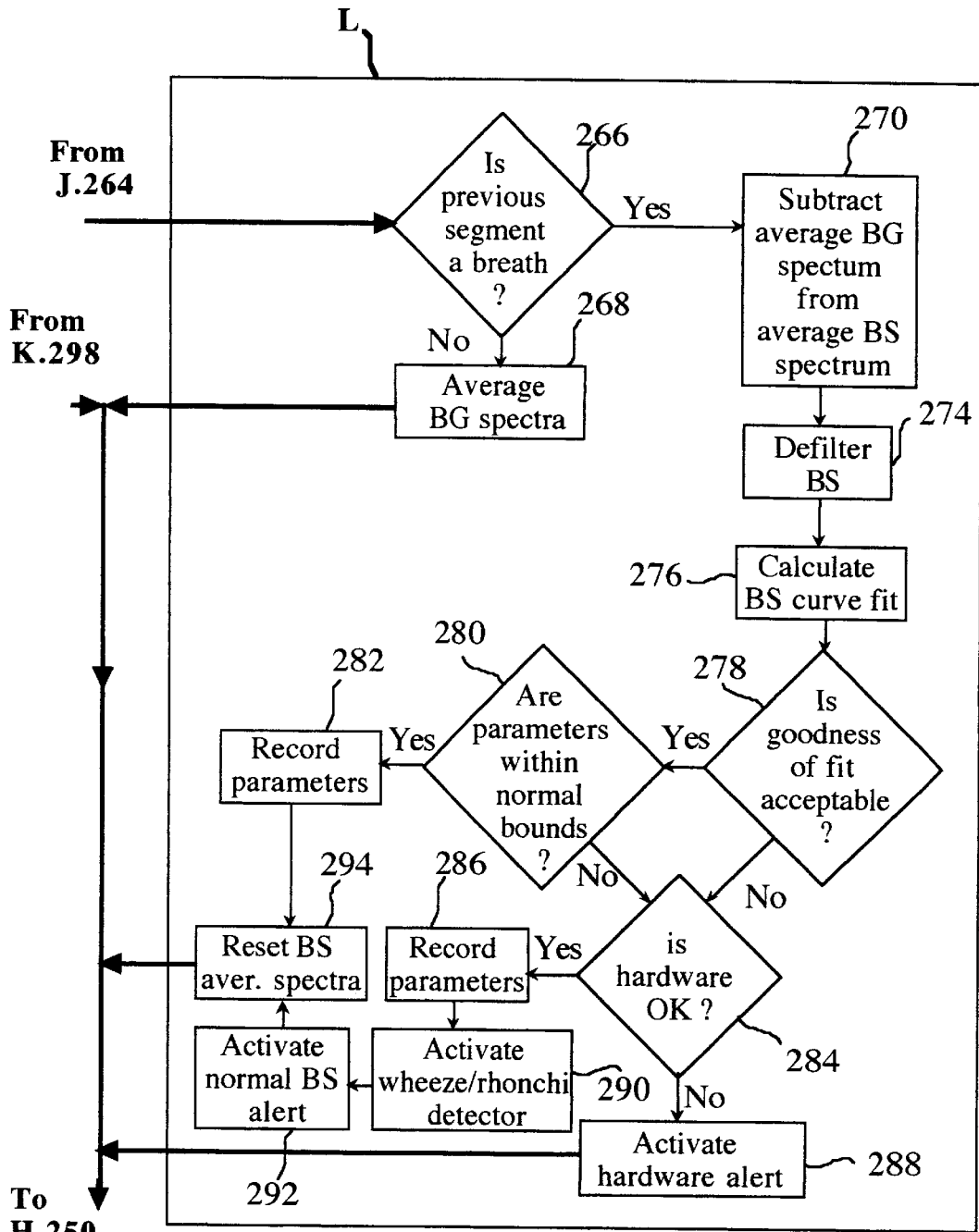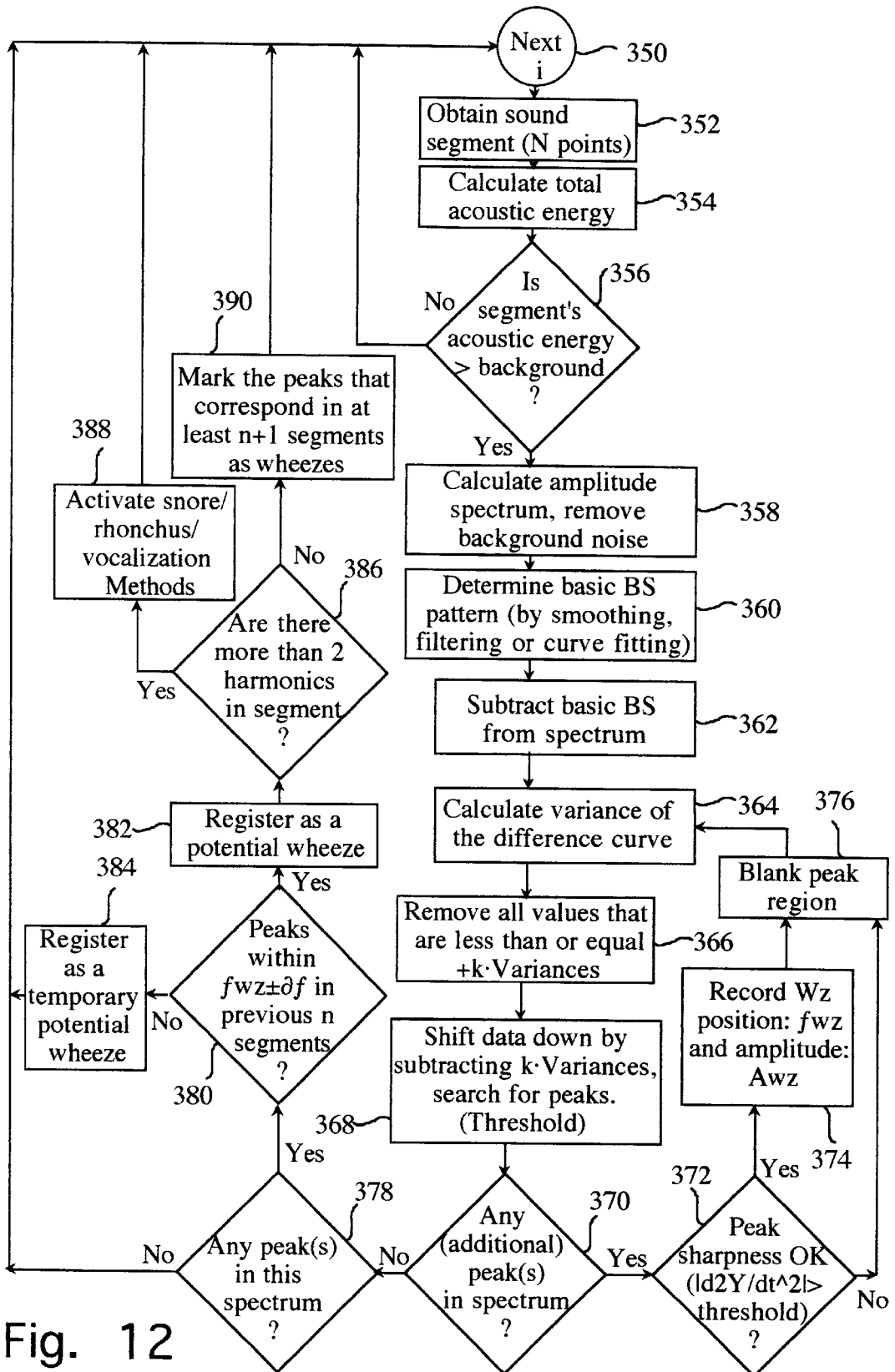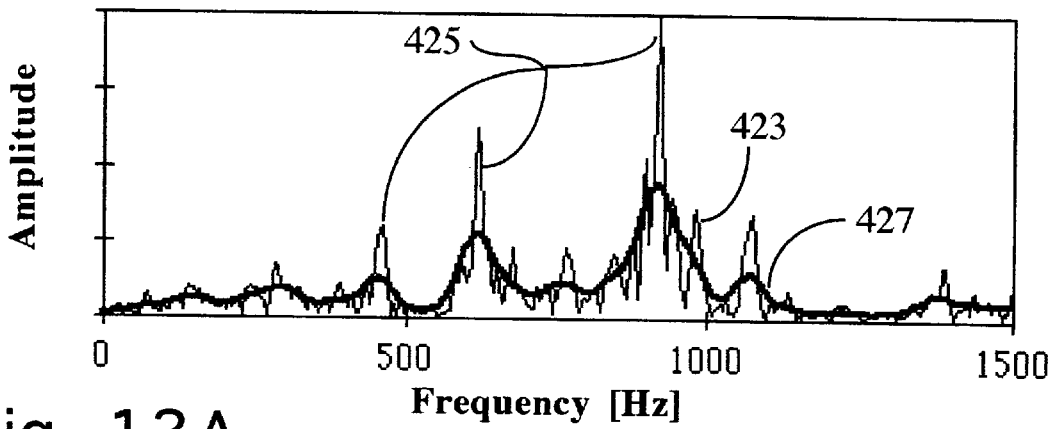
Activate rhonchi alarm — 414
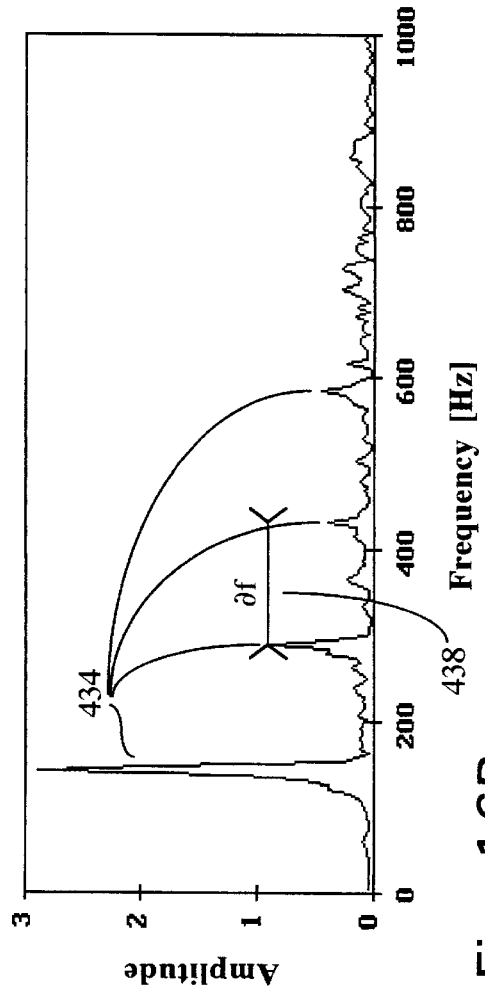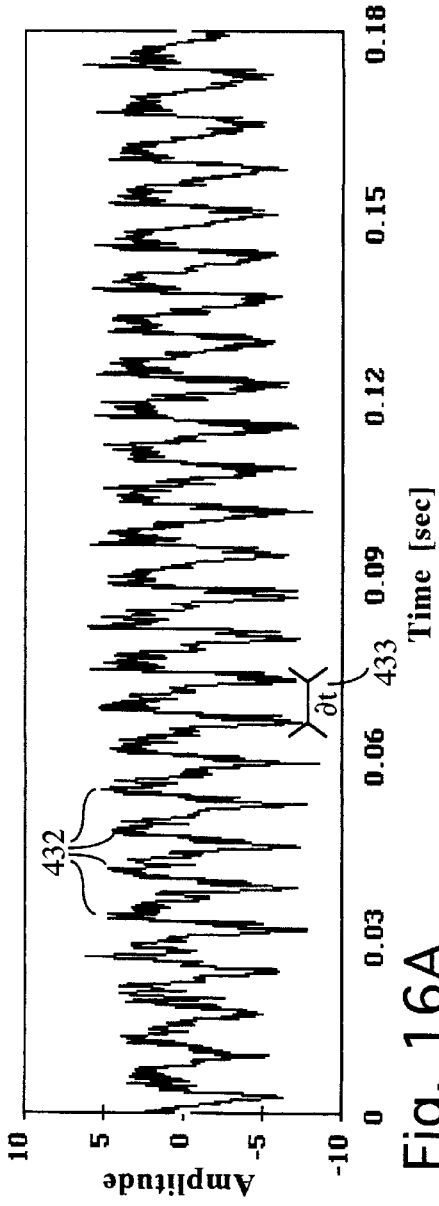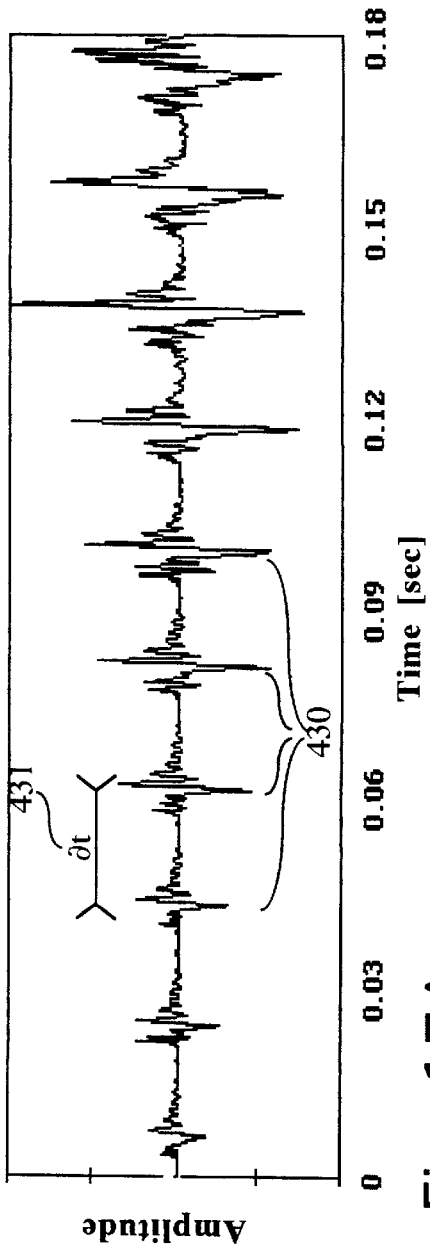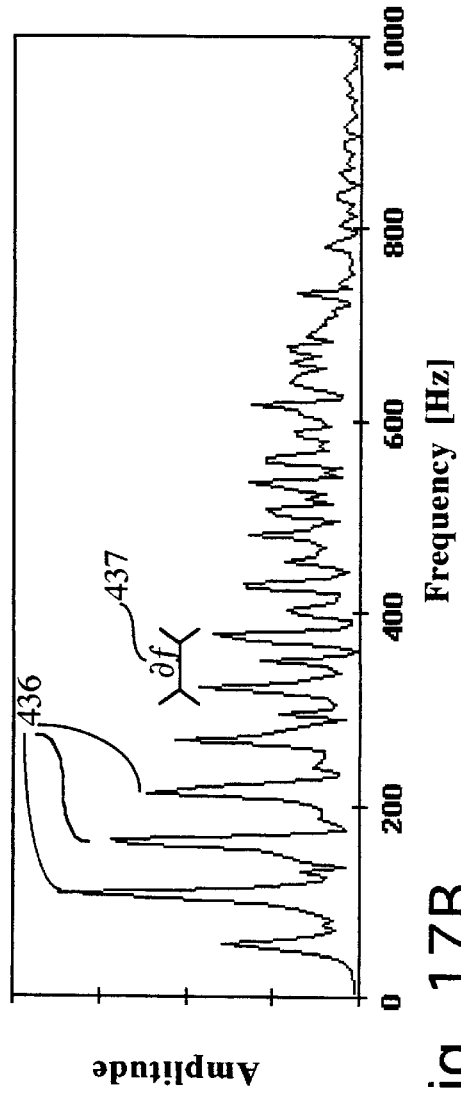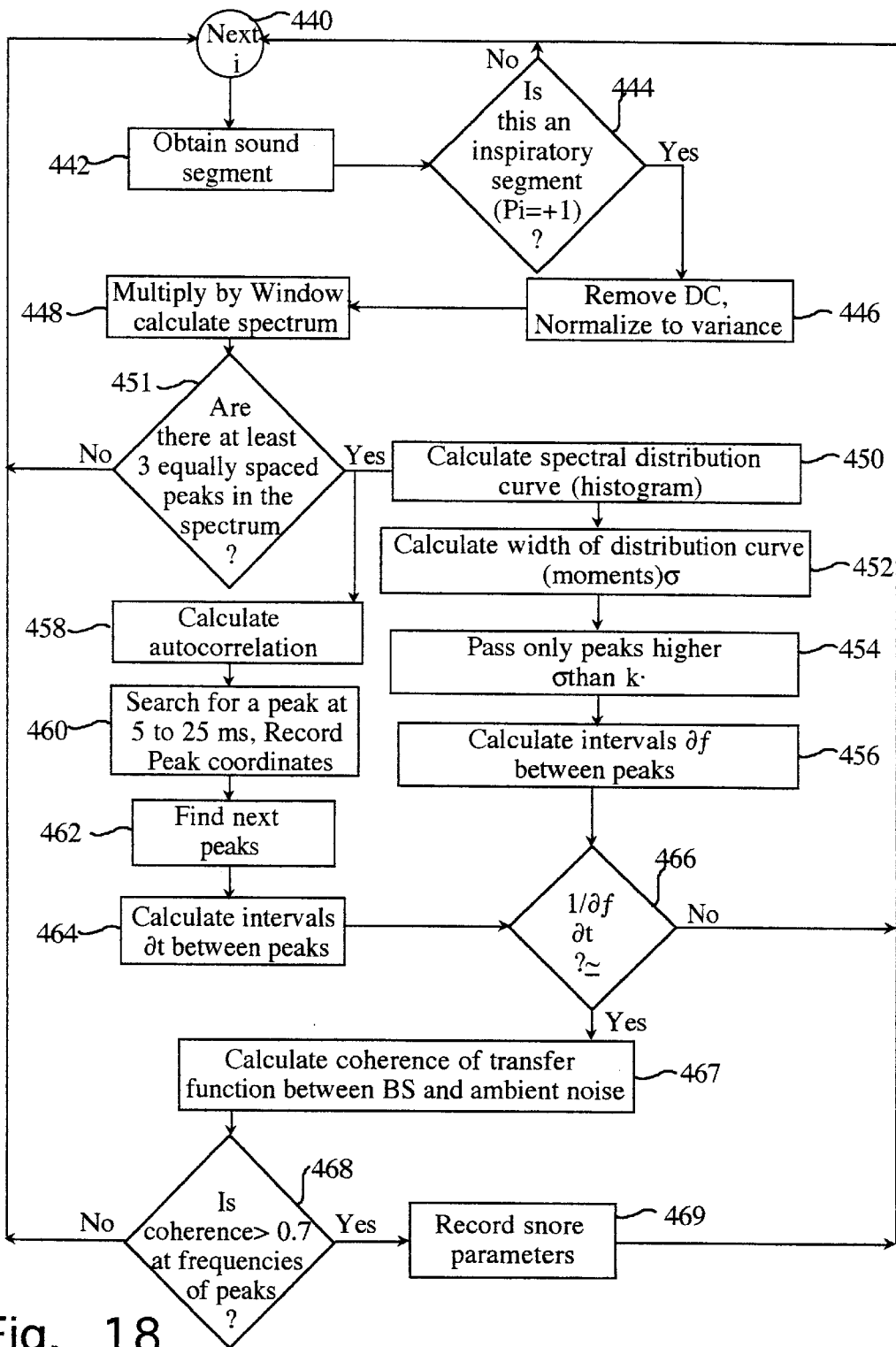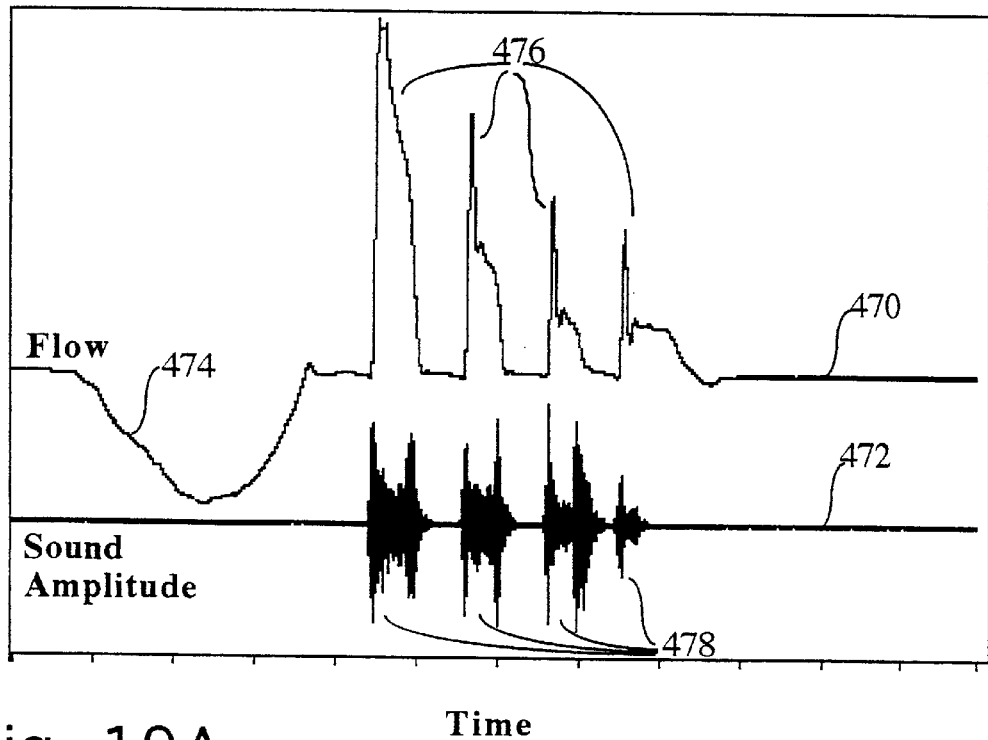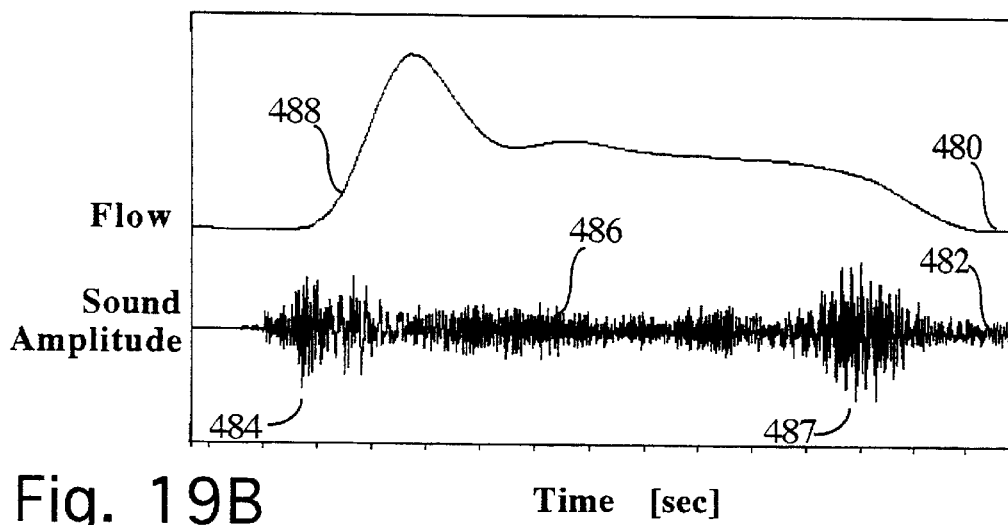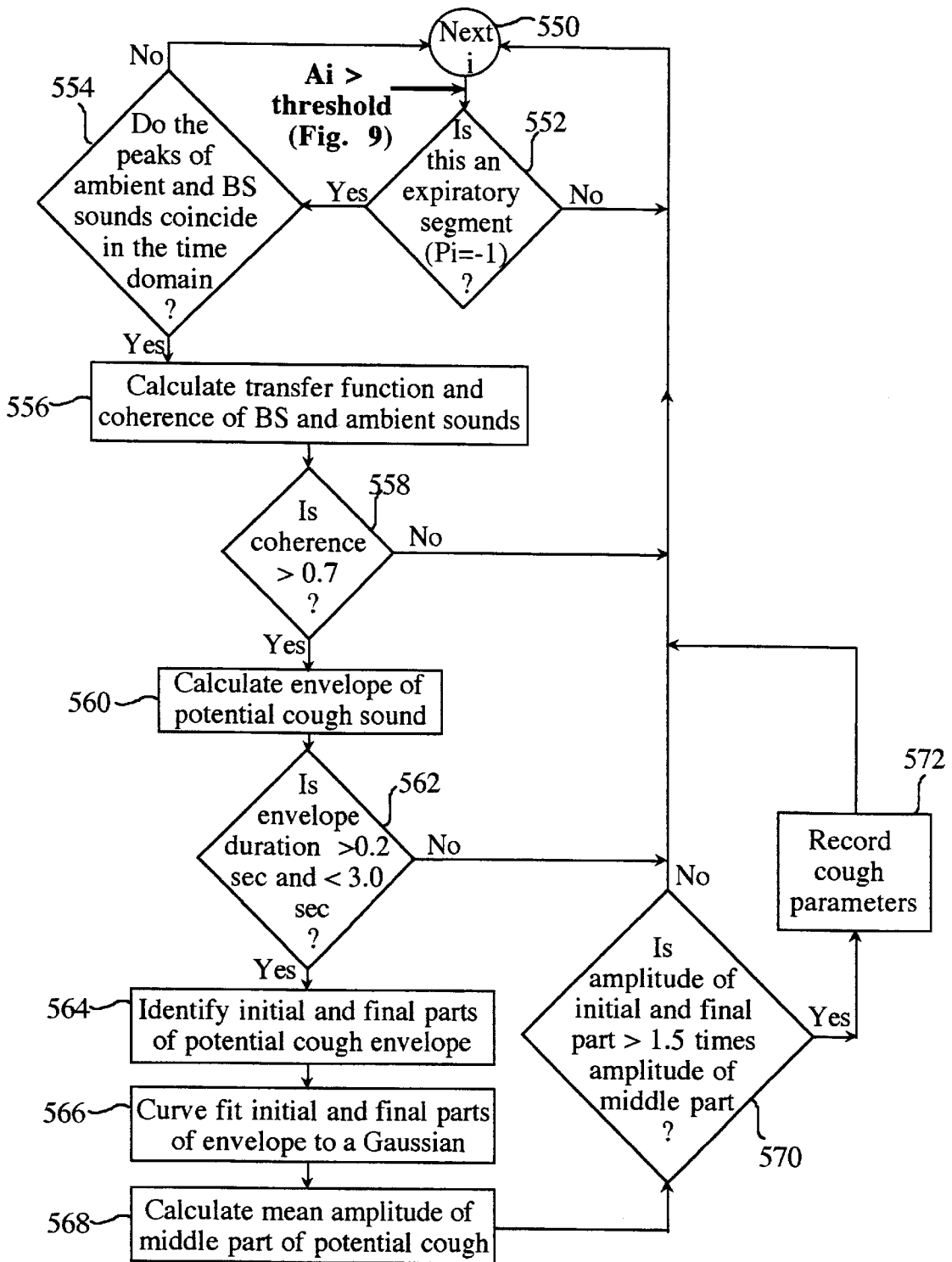
Fig. 15

Fig. 16A

Fig. 16B

Fig. 17A

Fig. 17B
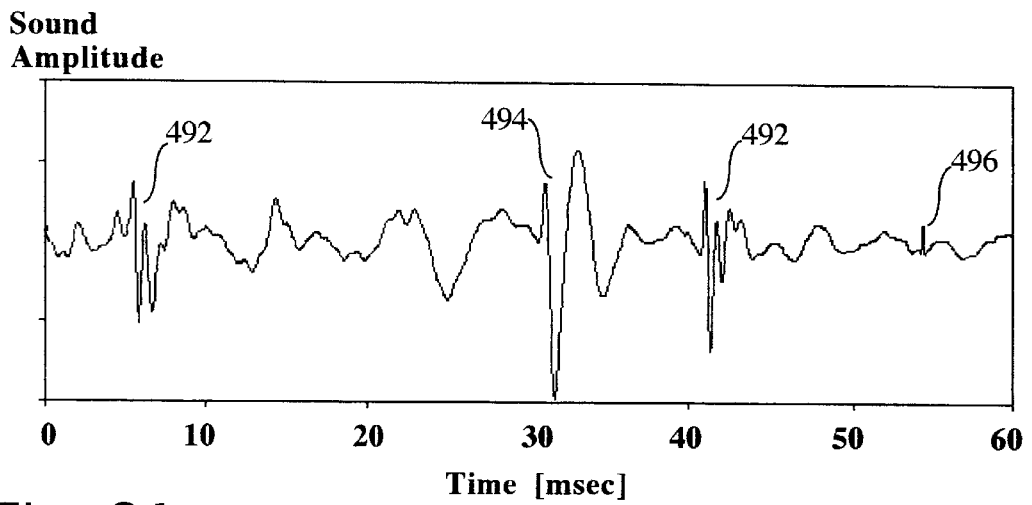
Fig. 18

Fig. 19A

Fig. 19B
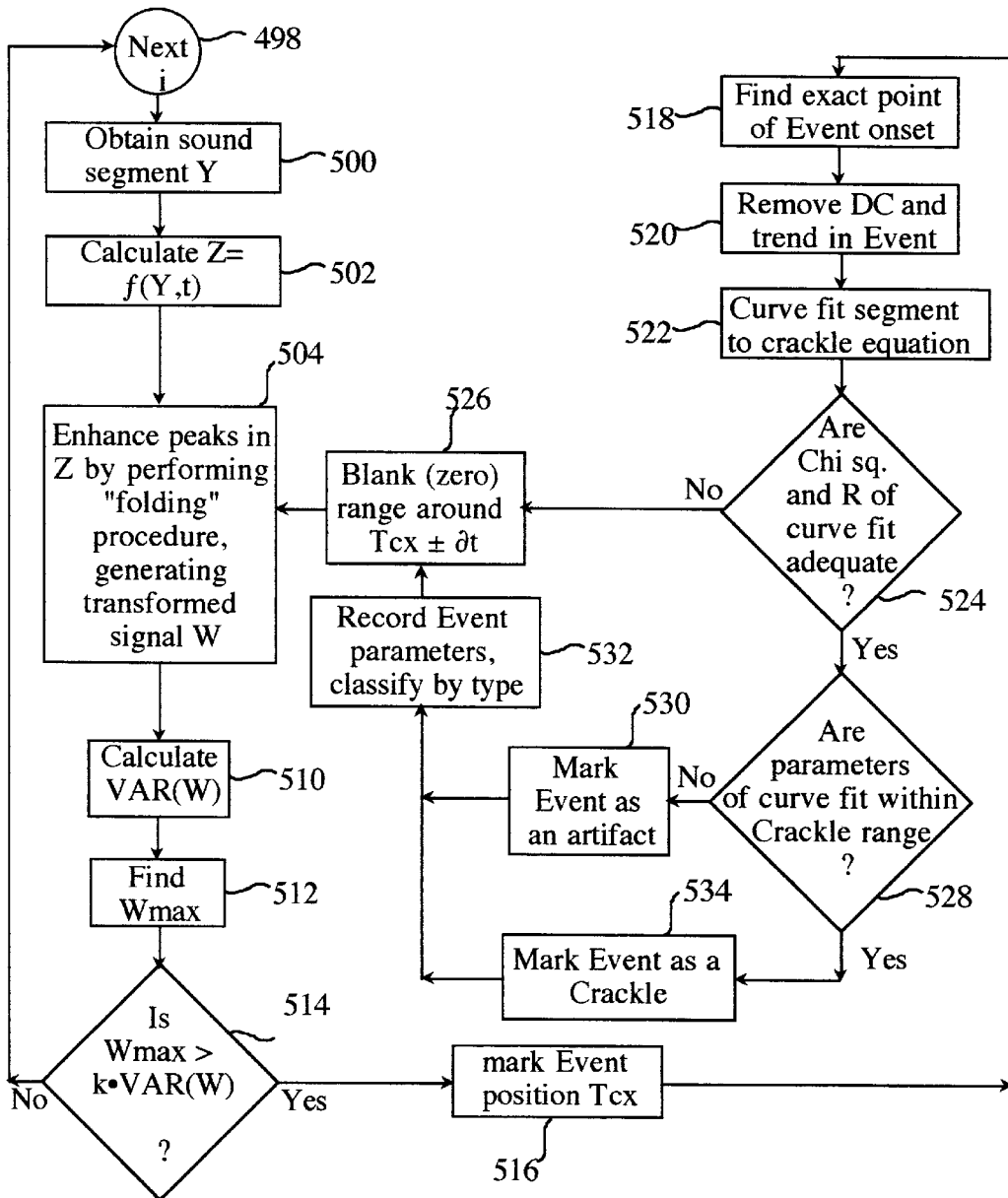
Fig. 20

**Sound
Amplitude**
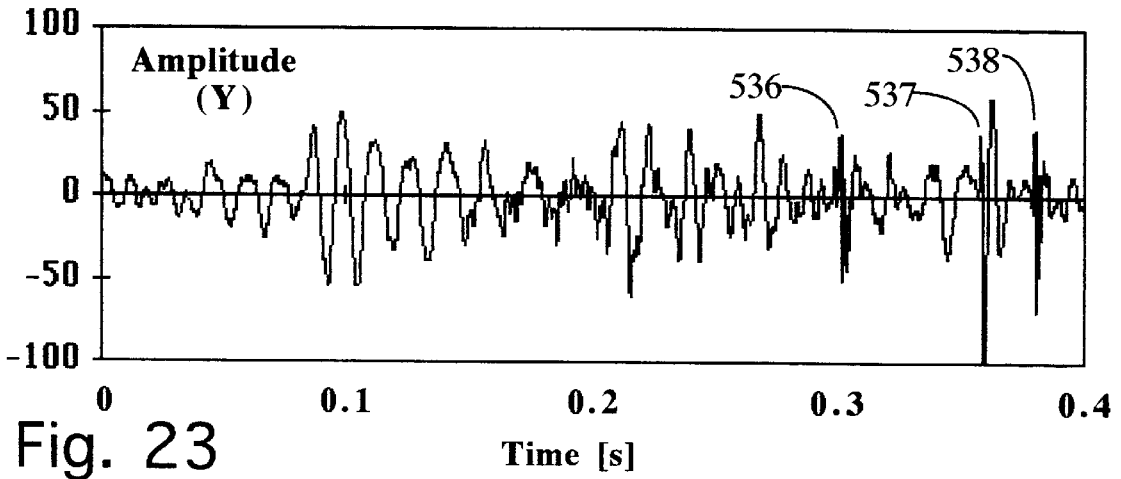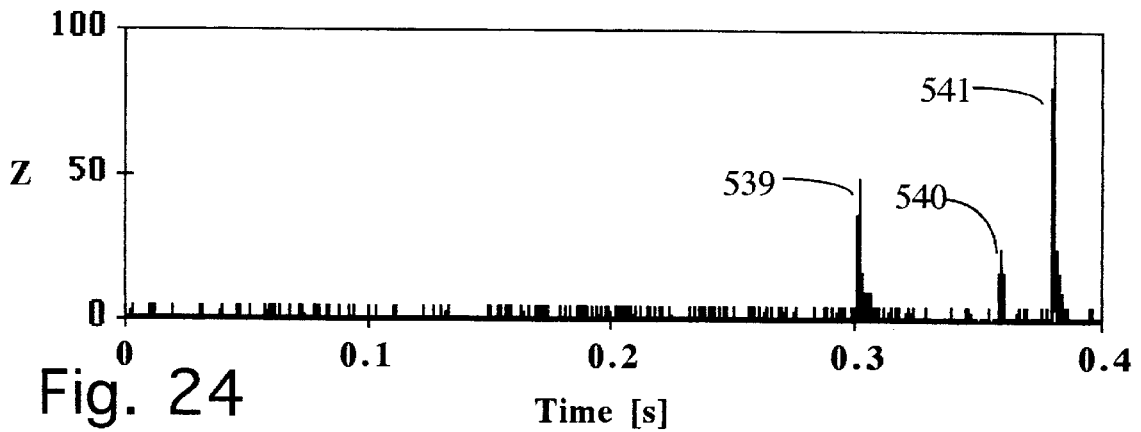

Time [msec]

Fig. 21

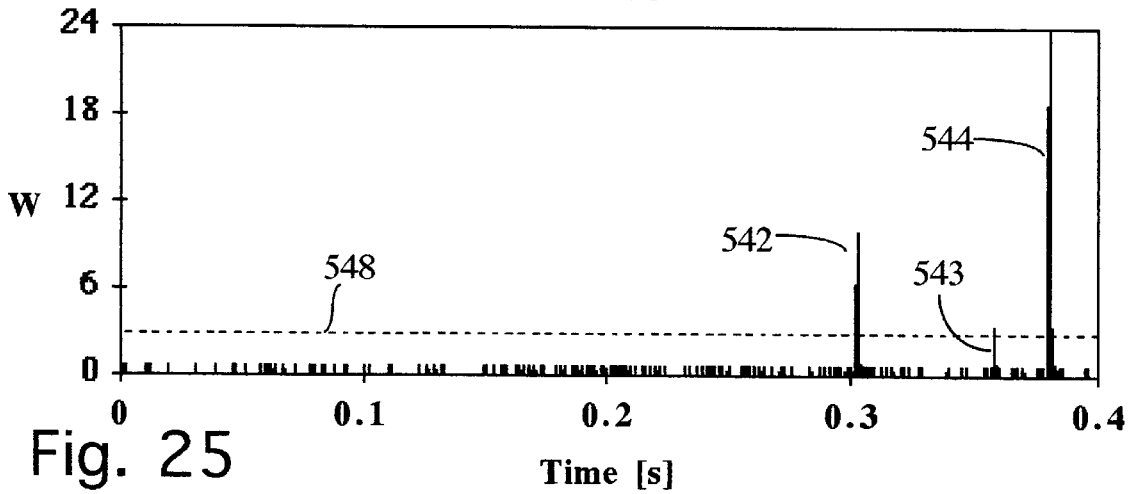Fig. 22

Fig. 23

Fig. 24

Fig. 25

PHONOPNEUMOGRAPH SYSTEM

This application is a national stage application of PCT application PCT/IL97/00318, filed Sep. 30, 1997 which is a continuation-in-part of U.S. Ser. No. 08/729,651, filed Oct. 4, 1996, now U.S. Pat. No. 6,168,568.

FIELD OF THE INVENTION

The present invention relates to a method, system and apparatus for the detection and analysis of body sounds in general and to a method, system and apparatus for the automatic detection and analysis of breath sounds in particular.

BACKGROUND OF THE INVENTION

The art of listening to body sounds, or auscultation, has been used by physicians for thousands of years, for diagnosing various diseases.

Auscultation was initially performed by placing the physician's ear directly on the skin of the patient. At the beginning of the 19th century, R. T. Laennec introduced a tool, the stethoscope, for transmitting of body sounds to the ear.

Currently used stethoscopes include a "chest piece" brought into contact with the patient's skin, and two flexible tubes, terminating in the physicians ears. Pulmonary sounds are typically classified into normal breath sounds and adventitious (abnormal) breath sounds. The type of adventitious breath sounds, their temporal location relative to the inspiration and expiration phases of the respiratory cycle and their rate of occurrence are used to diagnose the nature and severity of pulmonary diseases.

Adventitious breath sounds are usually divided into continuous and discontinuous sounds depending on their duration Continuous sounds are further subdivided into wheezes, which are higher pitched musical sounds indicating the presence of airway narrowing and rhonchi, which are low-pitched, grinding sounds. Discontinuous adventitious breath sounds are similarly divided into coarse crackles, which are short intermittent explosive sounds having a lower pitch, and fine crackles, which are less loud, shorter in duration, and higher in pitch. Crackles are usually indicative of obstructive airway diseases or restrictive lung diseases (depending on their timing in the respiratory cycle). In addition, cough and snores are respiration—related acoustic signals whose presence reflect on the status of well-being of the pulmonary system.

The use of various sensors which transform the acoustic signals of the body into electrical voltages is well known in the art Various types of transducers have been used in implementing body sound sensors, including both air coupled and contact microphones or accelerometers. An improved contact sensor for body sounds has been disclosed by the present inventor in U.S. patent application 08/654, 643 filed on May 29, 1996 and entitled "A Contact Sensor for Body Sounds". The disclosure of this application is incorporated herein by reference.

The introduction of computerized signal processing methods has facilitated quantitative and objective analysis of breath sounds. Methods have been developed which transform the acoustic signals to the frequency domain, characterize the signals' temporal and spectral patterns, and extract features that distinguish the various classes of normal and abnormal breath sounds, as described in the book, *Breath Sounds Methodology* by Noam Gavriely, CRC Press Inc,

1995. Pages 1–196, Chapters 1–13, which is incorporated herein by reference.

U.S. Pat. No. 5,213,108 to Bredesen and Schmerler discloses a visual display stethoscope which enables the visual display of acoustic data including breath sounds and manual measurement of certain parameters of the acoustic waveform. Methods involving the manual analysis of visually displayed digitized waveforms are time consuming, require the analysis to be performed by an expert and lack objective and uniform criteria for quantitative identification of adventitious sounds.

Published PCT Application WO 91/03981 to Murphy discloses a system and method for automatically detecting and identifying certain adventitious sounds. The method of crackle identification is based on two primary parameters, the amplitude and duration of the crackle wave (half cycle), and on one secondary parameter, the slope of the crackle waves.

Crackle detection methods based on wave amplitude analysis as a primary detection criterion have an inherent disadvantage due to the fact that the amplitude of crackles is often similar to or even smaller than the amplitude of the underlying breath sounds. A simple threshold crossing criterion will miss the majority of crackles, thus rendering the rest of the analysis useless.

Nocturnal respiratory symptoms are common, yet difficult to assess. They include nighttime dyspnoea caused by cardiac, gastrointestinal and pulmonary disease processes. Paroxysmal nocturnal dyspnoea (PND) is a symptom of congestive heart failure (CHF) where water accumulates in the lung of the supine patient and interferes with alveolar gas exchange. This process leads to a sudden onset of breathlessness, usually during the third part of the night, which is relieved by shifting to an upright posture. Gastro-esophageal reflux of acidic content from the stomach in supine patients is often associated with aspiration of the acid into the airways that causes an acute onset of cough, wheezing and dyspnoea. A nighttime onset of asthma attacks is common in children. It is often suspected by the physician after hearing a description of the course of events by a parent, but an objective diagnosis is difficult. Especially since at least some of these children have normal physical examination and spirometry during the day. Another class of nocturnal breathing disorders is the obstructive sleep—apnea (OSA) syndrome and its related conditions hypopnea, snoring). In these conditions, the patient's upper airway collapses or becomes narrowed or flutter develops, leading to a complete or partial interruption of the flow. OSA causes interference with the normal pattern of sleep, multiple (some time as many as hundreds) arousals during the night and reduced oxygenation.

Each of these clinical conditions have distinct breath sound features. Progressive pulmonary water overload is initially associated with the generation of inspiratory crackles, with an increasing range of chest wall distribution, and eventually, the emerging of expiratory crackles and wheezes. In addition, there is a gradual increase in the respiratory rate (tachypnea). An acute onset of cough, wheezing and secretion sounds (rhonchi and expiratory crackles) is a marker of aspiration. Finally, a gradual onset of wheezing with or without coughing is associated with bronchial asthma.

A disadvantage of the prior art of manual auscultation is that it can not provide a reliable objective and continuous detection and documentation of the occurrence of nocturnal adventitious breath sounds with respect to time, and while

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the patient is sleeping. Thus, prior art auscultation methods do not provide a reliable objective method for accurately diagnosing and documenting respiratory symptoms.

SUMMARY OF THE PRESENT INVENTION

One aspect of the present invention involves the determination of the types of breath sounds, i.e., wheezes, crackles, snores produced by a patient based on their generation and transmission mechanisms.

This aspect of the invention is based on an understanding and analysis of the underlying physics, aerodynamics, acoustics and physiology of breath sounds. Since basic breath sounds are generated by acoustic emission from turbulent flow they generally have a broad band spectrum at the site of their generation These sounds often reverberate in the airways which results in broad resonant frequencies. Furthermore, sounds transmitted to the chest wall are attenuated by intervening tissue, air and bone which acts as a low pass acoustic filter. When the breath sounds are fitted to characteristic equations of breath sounds the change in attenuation properties (for example by the presence of pneumonia) are reflected in modification of coefficients of the equations.

In another aspect of the invention, breath sounds are detected and characterized using a two step process. Preferably these two steps include an initial screening based on the detection of one or more specific properties of the sought signals. This screening can be both positive (i.e., the presence of a sound pattern which is characteristic of the particular breath sound) or negative (i.e., the presence of a sound pattern which is never present as part of the particular breath sound).

The second step is a verification process in which the match between the detected event (breath sound) and the features and parameter values of the particular breath sound are determined. This may include the characteristics of peaks, their number, placement, spectral properties of the sounds, etc.

In some preferred embodiments of the invention, the fitting of the breath sound with a mathematical function is used as the basis for the second step. The parameters of the match can serve as a way of grouping the sounds or further characterizing them so that further diagnostic information as to their causes and origins may be determined.

In preferred embodiments of the invention, the two step process serves as a starting point for the search for adventitious sounds.

One preferred embodiment of the invention includes the search for crackle sounds. A crackle is a non-stationary event with a specific waveform. The characteristics of the waveform reflect the mechanism of crackle generation as an abrupt opening of an airway that was previously closed due to a collapse or a liquid barrier. The waveform also reflects the attenuation and refraction of the original sound as it is transmitted through the lung and chest wall. With this in mind, a preferred embodiment of a method of detecting a crackle is based on a rapid search for non-stationary events, followed by a verification process that is designed to eliminate artifacts i.e., other non-stationary events that do not fit a mathematical description of a true crackle and to obtain the values of parameters of the mathematical description to sort the crackles into subgroups.

A second preferred embodiment of the invention includes a search for coughs. Cough detection is initially based on the simultaneous detection of a sudden onset of loud noise in a tracheal sound sensor and ambient microphone and the

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detection of a rapid motion of the patient's chest. Following this rapid screening, the algorithm verifies the detection by curve fitting the sound signal envelope to a specific mathematical expression and by verifying that the spectral content of the sound is that of a cough wad band noise) rather than that of a snore and vocalization.

A third preferred embodiment of the invention is concerned with the detection of wheezes. The first step in the detection of wheezes is the is a screening in the frequency domain in which peaks of power above an underlying spectrum of basic sounds are sought If any are detected, they are evaluated to determine whether these peaks correspond to true wheezes, for example as to their sharpness and prominence over the underlying spectrum. A sub-set of wheezes is called "squeaks." Squeaks and wheezes are similar except that squeaks last between about 80 and 150 milliseconds and wheezes last longer than 150 milliseconds and they have other distinctions as well.

A fourth preferred embodiment of the invention is concerned with the detection of breath activity. The breath detector combines data from the tracheal breath sounds sensor and from the chest expansion sensor to obtain a tentative detection of breathing activity (as oppose to apnea) and the respiratory phase. This detection is then validated by verifying that the tracheal breath sounds spectral characteristics match the mathematical template of authentic tracheal breath sounds. The template of tracheal breath sounds reflects an understanding of tracheal sound generation.

Chest wall breath sound detection is secondary to the breath detection process. If breathing activity (inspiratory or expiratory) is detected (see above), the spectra of the chest wall breath sounds are calculated. The verification is done by matching the spectra to a low pass filter template. The quality of the match and the values of the filter parameters determine whether the sounds can actually be classified as authentic chest wall breath sounds.

A fifth preferred embodiment of the invention is concerned with snore detection. Snores are tentatively identified when loud tracheal and ambient sounds are detected at the same time, but without a coinciding sudden chest motion. However, this detection has to be validated by positively identifying certain features and excluding others. In particular, the snore detector searches for equally spaced sound structures in the time domain, multiple resonance peaks in the frequency domain, high coherence between the ambient and tracheal sounds and inspiratory chest motion. It should be noted that while snores may also be expiratory, the latter are indistinguishable from certain types of vocalizations. Therefore for absolute positive validation of a sound as a snore it should coincide with inspiratory chest motion.

A sixth preferred embodiment of the invention is concerned with rhonchi detection. Tentative detection of rhonchi is a secondary process, that is they are suspected based on a search for another sound. They are suspected if continuous adventitious breath sounds with multiple resonance peaks (>3) are detected by the wheeze detection method. To be confirmed the absence of coinciding loud sounds at the trachea and/or the ambient sounds sensor must be verified to exclude snores and vocalization.

Other preferred embodiments of the invention use the verification process to separate between different, generally related, breath sounds. Thus, for example, basic chest wall breath sounds can be classified as normal and bronchial. The determination is based on the characteristics of the low pass filter which is the best fit for the detected sounds. Fine cackles can be distinguished from coarse cackles by the

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value of an initial cackle frequency, the rate of increase of the cackle frequency and other factors. Similarly, wheezes can be separated into "low", "high" and "ultra high" frequency sub-groups coughs can be separated into "dry", "productive" and "barking types" and snores can be separated into simple and complex types.

The initial features are preferably sought using relatively rapid computational methods that do not require much CPU time. Once a potential event is detected it is evaluated, using algorithms which generate a high degree of confidence in the determination.

The new approach to Phonopneumography disclosed herein is substantially different from other methods described in existing patents and previously published manuscripts by its focus on the narrow repertoire of basic and adventitious breath sounds, using specific tools and mathematical descriptors of the sought sounds, while essentially eliminating the many other sounds that exist in the environment that do not fit the exact definitions of breath sounds. In addition, while the present approach is comprehensive in the sense that it deals with all the known breath sounds, including cough and snore (i.e. almost all the sounds that are associated with gas motion in the airways), it uses tailor-made signal processing method for each sound. This is different from other methods whereby a single signal processing method is used to classify the sounds into clusters or groups, based on features that are extracted from the signal (e.g. autoregressive (AR or ARMA) parameters).

In a further aspect of the invention, breath sounds are analyzed using acoustic signals and information regarding chest expansion. This combination results in a greatly enhanced ability to characterize sounds since a knowledge of the phase of breathing and the timing of events is very useful in the characterization of basic and adventitious breath sounds. For example, the clinical significance of early inspiratory crackles is completely different from late inspiratory or expiratory ones.

There is thus provided, in accordance with a preferred embodiment of the invention, a method of analyzing breath sounds produced by a respiratory system, the method comprising:

measuring breath sounds produced by the respiratory system;

tentatively identifying a signal as being caused by a breath sound of a given type if it meets a first criteria characteristic of the breath sound of the given type; and

confirming said identification if a tentatively identified signal meets a second criteria characteristic of the breath sound of the given type.

Preferably tentatively identifying comprises comparing the breath sound to a plurality of first criteria, each said criteria being characteristic of a breath sound of a given type, wherein said breath sound is tentatively identified as being of the type for which it meets the first criteria.

In a preferred embodiment of the invention, confirming said identification comprises comparing breath sounds tentatively identified as being of a given type to one of a plurality of second criteria each of which is characteristic of the given type and confirming that the breath sound is of the second type if it meets the second criteria characteristic of the given type.

Preferably the method comprises segmenting said breath sound data into segments and wherein tentatively identifying and confirming are based on time segments of breath sound data.

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In a preferred embodiment of the invention the given breath type comprises a wheeze. Preferably, tentatively identifying the breath sound as a wheeze comprises detecting narrow peaks within the spectrum of the breath sound and determining if the narrow peaks are located within a narrow frequency range over a number of consecutive time periods. Preferably confining comprises determining if the narrow peaks of said tentatively identified wheeze have less than three harmonics each.

Preferably said consecutive time periods span at least 150 ms. Preferably, said narrow frequency range is not greater then 64 Hz among any two consecutive time periods.

In accordance with a preferred embodiment of the invention a breath sound is confirmed as a squeak type wheeze when said consecutive time periods span between 80 and 150 ms.

Preferably, a breath sound is classified as low frequency wheeze when the frequency of the narrow peak is less than 400 Hz. Preferably a breath sound is classified as a high frequency wheeze when the frequency of the narrow peak is between 400 Hz and 1600 Hz. Preferably, a breath sound is classified as an ultra-high frequency wheeze if the frequency of the narrow beak is above 1600 Hz.

In accordance with a preferred embodiment of the invention, the given breath type is a rhonchus. Preferably, tentatively identifying comprises detecting narrow peaks within a spectrum of said segment and determining if said narrow peaks are located within a predetermined small frequency range across during consecutive time periods. Preferably confirming the identification of a breath sound as a rhonchus comprises, if there are more than two harmonics in each of the consecutive time periods:

generating a transfer function in the frequency domain between said breath sound data and measured ambient sound in the space surrounding the patient;

determining a coherence graph of said transfer function;

confirming each narrow peak as a rhonchus if the frequency range of high coherence of said coherence graph does not correspond to the frequency range of said narrow peaks.

In a preferred embodiment of the invention, the breath sound is a cough. Preferably tentatively identifying comprises:

coincidentally detecting sudden loud ambient noise, a sudden loud breath sound, and a sudden chest motion.

Preferably, confirming that a breath sound is a cough comprises:

generating an envelope of said breath sound data and determining the duration of said envelope;

determining that the sound is a cough if it fulfills all of the following conditions:

a) the breath sound takes place during expiration;

b) the breath sound peaks generally coincide with the ambient noise peaks;

c) the envelope of the breath sound data has a double hump shape;

d) the duration of the envelope is within a predetermined time range; and

e) determining that the frequency spectra of the sound are broad band with at least a predetermined high coherence level between the ambient and the breath sounds.

Preferably, said predetermined time range is 0.2–3.0 seconds.

Preferably, said predetermined coherence level is 0.7 or greater.

In a preferred embodiment of the invention the cough is classified as a productive cough if the second hump has a variance greater than 40 msec and is skewed to later times. Alternatively, the cough is classified as a dry cough if it has a variance of less than 40 msec and is not substantially skewed. Alternatively, the cough is classified as a barking cough if it is has a duration of between 200 and 350 msec and has a second hump of 10%–25% above the value between the humps.

In a preferred embodiment of the invention, the breath sound is a crackle. Preferably, tentatively identifying a breath sound as a crackle comprises:

finding the locations of abrupt changes in said breath sound data;

Preferably, confirming that a breath sound is a crackle comprises:

matching breath sound data following the said abrupt changes to the following curve:

$$y=A*B(t)*C(t)$$

where y is said breath sound data starting at the abrupt change at which t begins, A is an amplitude parameter, B(t) is an envelope function and C(t) is an oscillatory function.

In a preferred embodiment of the invention,

$$B(t) = \frac{t^n}{1 + (mt)^k}$$

and

$$C(t) = \sin\left(\frac{2\pi f_0 t}{1 + Ct}\right),$$

where f_0, C, n, m and k are parameters.

A preferred embodiment of the invention includes identifying the type of crackle from the values of said parameters. Preferably, the cackle is identified as a fine cackle if f_0 is above 600 Hz. Preferably, the cackle is defined as a coarse cackle if the if C, the rate of change of the cackles internal frequency is greater than 100.

In accordance with a preferred embodiment of the invention the method includes identifying the portion of the breathing cycle during which the crackle occurs.

In a preferred embodiment of the invention, the given breath sound is a snore. Preferably tentatively identifying a breath sound as a snore comprises:

determining that the breath sound occurs during an inspiratory period; and

determining that the breath sound is highly correlated with ambient noise in the region surrounding the patient

Preferably, that the tentatively identified breath sound is a snore comprising:

identifying peaks in the inspiratory period and determining the average peak-to-peak time delta_t;

identifying at least three peaks in the power spectrum of the inspiratory period which are significantly large and determining the average peak-to-peak frequency delta_f;

generating a coherence graph for a transfer function in the frequency domain between said breath sound data and measured ambient noise of the space where said breath sound data was gathered;

identifying a snore if delta_t is close to the inverse of delta_f and if the frequency range of high coherence of said coherence graph corresponds to the frequency range of said narrow peaks.

In a preferred embodiment of the invention, identifying peaks in the inspiratory period comprises:

calculating the inverse Fourier transform of said power spectrum of the inspiratory period and generating thereby cleaned data;

searching for a first peak in said cleaned data beginning between 5 to 25 ms from the start of said segment; and

searching for peaks following said first peak.

Preferably, identifying three peaks in the power spectrum comprises:

calculating the histogram of the power spectrum;

determining the variance of the histogram; and

defining as peaks those points having values which are at k variance or higher, where k is at least tree.

In a preferred embodiment of the invention the snore is classified as a simple snore if the spectrum contains 3 or 4 resonance peaks. Preferably, the snore is classified as a complex snore if the spectrum contains 5 or more resonance peaks.

In a preferred embodiment of the invention, the breath sound is a chest wall breath sound. Preferably, the breath sound is identified as a chest wall breath sound if it fits a low pass filter shaped curve having the parameters of amplitude, corner frequency and slope.

Preferably, the breath sound is confirmed as a bronchial chest wall breath sound if the corner frequency is higher than a normal frequency for the age of the patient.

There is further provided, in accordance with a preferred embodiment of the invention, a phonopneumograph for analyzing breath sounds produced by a respiratory system, the phonopneumograph comprising:

at least one auditory breath sensor which receives sound related to breath sounds and produces signals in response to the received sounds;

at least one first sound signal analyzer which analyzes said signals to initially screen said breath sounds said initial screening being operative to produce a tentative identification of the signals as being caused by a breath sound of a given type if it meets a first criteria; and

at least one second sound signal analyzer which further analyzes those of said signals which have been tentatively identified to confirm if the signals are caused by a breath sound of the given type.

Preferably, the at least one breath sensor comprises a plurality of breath sensors placed at different positions with respect to the respiratory system.

Preferably, the at least one first sound analyzer comprises a plurality of first sound analyzers, each of which is operative to initially screen said breath sounds, each of said first sound analyzers producing a tentative identification of a breath sound of a different type.

Preferably, the at least one second sound analyzer comprises a plurality of second sound analyzers, each of which is operative to initially screen said breath sounds, each of said second sound analyzers producing a confirmed identification of a breath sound of a different type.

In one preferred embodiment of the invention the breath sound of a given type comprises a crackle. Alternatively or additionally the breath sound of a given type comprises a cough. Alternatively or additionally, the breath sound of a given type comprises a snore. Alternatively or additionally, the breath sound of a given type comprises a rhonchus. Alternatively or additionally, the breath sound of a given type is a wheeze. Alternatively or additionally, the breath sound of a given type is a breath.

In accordance with a preferred embodiment of the invention, the first sound analyzer divides the sound into time segments and analyzes the sound on a segment by segment basis.

There is further provided, in accordance with a preferred embodiment of the invention, a phonopneumograph for analyzing breath sounds, produced by a respiratory system the phonopneumograph comprising:

at least one breath related sensor placed adjacent the respiratory system of a patient the sensor measuring breath related activity, said at least one sensor producing breath sound data;

a breath analyzer which matches said breath sound data to a plurality of breath sound templates each of which is characteristic of and paremetrizes one type of breath sound and which produces a signal indicative of the presence of a particular regular or adventitious breath sounds only when said breath sound data matches, within predetermined goodness of fit criteria, one or more of said breath sound templates.

Preferably, the at least one breath related sensor comprises a plurality of sensors placed at different positions with respect to the respiratory system.

Preferably, said at least one sensor includes at least one sensor of at least one of the following types of sensors: chest expansion sensors, breath sounds sensors, tracheal sound sensors, flow meters and spirometers.

In a preferred embodiment of the invention, said plurality of breath sound templates are individually characteristic of a breath sound chosen from the following breath sounds: regular chest wall breath sound, regular tracheal breath sound, a wheeze, a cough, a rhonchus, a snore and a crackle.

Preferably, said system additionally includes an ambient noise microphone which measures ambient noise data in a space surrounding the patient.

In a preferred embodiment of the invention, the at least one template comprises a template for regular chest wall breath sound, said template for regular chest wall breath sound comprising a curve in the frequency domain of said breath sound data having the shape of a low pass filter, wherein the parameters are the amplitude, cutoff frequency and slope of the low pass filter.

In a preferred embodiment of the invention, the at least one template comprises a template for regular tracheal breath sound, said template for regular tracheal breath sound comprising a curve in the frequency domain of said breath sound data of an ensemble of second order resonant systems wherein the parameters stored are a set of amplitude coefficients, a set of damping coefficients and a set of resonance frequencies. In a preferred embodiment of the invention, the at least one template comprises a template for a wheeze, said template for a wheeze comprising a narrow peak in the frequency domain of said breath sound data whose width at half height spreads less than a predetermined frequency width, which has fewer than three harmonics, said narrow peak occurring within at least three time segments comprising a predetermined time period, the frequency of said narrow peaks varying less than 1.5 Hz per msec within said at least three said occurrences.

In a preferred embodiment of the invention, the at least one template comprises a template for a rhonchus, said template for a rhonchus comprising a repetitive sound in said breath sound data having generally evenly spaced peaks in the time and frequency domains that has a significant lack of correlation in the time and frequency domains with ambient noise.

In a preferred embodiment of the invention, the at least one template comprises a template for a snore, said template for a snore comprising a repetitive sound in said breath sound data having generally evenly spaced sound structures in the time domain and evenly spaced peaks in the frequency

domain, whose average spacing between sound structures in the time domain is generally equivalent to the inverse of the average spacing between peaks in the frequency domain and wherein said breath sound data is significantly correlated with ambient noise.

In a preferred embodiment of the invention, the at least one template comprises a template for a cough, said template for a cough comprising a sound occurring during a sudden expiratory chest motion which endures 0.2–3 seconds, has a double hump envelope in the time domain, has a relatively flat spectrum and which has a significant correlation with ambient noise.

In a preferred embodiment of the invention, the at least one template comprises a template for a crackle, said template for a crackle comprising a curve, whose onset point begins as an abrupt change in said breath sound data, and which generally matches the function:

$$y = A * B(t) * C(t)$$

where y is said breath sound data beginning at said onset point, t is the time measured from said onset point, A is an amplitude parameter, B(t) is an envelope function and C(t) is an oscillatory function. Preferably,

$$B(t) = \frac{t^n}{1 + (mt)^k}$$

and

$$C(t) = \sin\left(\frac{2\pi f_0 t}{1 + Ct}\right),$$

where f_0, C, n, m and k are parameters.

Preferably, the phonopneumograph according to a preferred embodiment of the invention also comprises a timing analyzer for determining the timing of breathing activity and the relative timing and duration of at least one of said regular and adventitious breath sounds.

Preferably, the phonopneumograph according to a preferred embodiment of the invention comprises:

a storing unit for storing at least the parameters of detected adventitious breath sounds and the times at which they occurred.

Preferably, said times are related to a the breathing cycle.

Preferably, said breath analyzer comprises a trends analyzer for analyzing trends in said breath sound data signals over said relatively long time period.

There is further provided, in accordance with a preferred embodiment of the invention, a phonopneumograph for analyzing breath sounds, produced by a respiratory system, the phonopneumograph comprising:

at least one breath related sensor placed adjacent the respiratory system of a patient which at least one sensor produces breath sound data signals in response to breath sounds;

an ambient noise microphone placed near said patient which microphone produces an ambient noise signal in response to ambient noise;

an ambient noise level detector which quantifies the level of the ambient noise signal;

a loud noise analyzer which indicates the presence of snores or coughs in said breath sound data signal when said ambient noise level detector simultaneously detects loud ambient noise based on the ambient noise signal; and

a breath analyzer which is operative to determine regular breathing activity and/or adventitious breath sounds in said

breath sound data signals when the ambient noise level detector detects a low level of noise.

In a preferred embodiment of the invention, said adventitious breath sounds include at least one of the following sounds: a wheeze, a cough, a rhonchus, a snore and a crackle.

Preferably, the phonopneumograph according to a preferred embodiment of the invention comprises a timing analyzer for determining the timing of breathing activity and the relative timing and duration of at least one of said regular and adventitious breath sounds.

Preferably, the phonopneumograph according to a preferred embodiment of the invention comprises a display which indicates regular breathing and which indicates when in the respiratory cycle an adventitious breath sound has occurred.

There is flier provided, in accordance with a preferred embodiment of the invention, a breath sounds monitor comprising:

at least one breath related sensor, placed proximate the respiratory system of a patient, the at least one sensor producing breath sound data signals in response to breath sounds; and

a breath analyzer which continuously analyzing said breath sound data signals, said breath sound analyzer being operative to determine the presence of breathing, and the presence of adventitious breath sounds in said breath sound data and

an alarm which provides alert indications responsive to a determination of the presence of abnormal breath sounds or adventitious breath sounds by said breath sound analyzer.

Preferably, the at least one breath related sensor comprises at a pair of sensors placed on opposite lateral sides of the chest and said breath analyzer detects asymmetry and imbalance in the breath sound data output from the pair of sensors and wherein the alarm provides an alert responsive to a determination of asymmetry by the analyzer.

Preferably the monitor flyer comprises a display which shows an indication of breathing activity and which indicates the occurrence, identity and extent of at least one said adventitious breath sound.

In a preferred embodiment of the invention the breath analyzer comprises a template matcher which matches the breath sound data produced by said breath related sensors to a plurality of breath sound templates each of which is characteristic of and parametrizes one type of breath sound and which produces a signal indicative of the presence of a particular regular or adventitious breath sounds only when said breath sound data matches, within predetermined goodness of fit criteria, one or more of said breath sound templates. Preferably, the breath analyzer comprises a memory which stores template parameters of at least said detected adventitious sounds. Preferably, the monitor comprises a trend analyzer which analyzers a plurality of said stored template parameters and determines trends of said detected adventitious sounds.

In a preferred embodiment of the invention, the breath analyzer comprises an apnea monitor which determines absence of breathing for a time longer than a given time and wherein and wherein the alarm is activated whenever such absence occurs. Preferably, the breath analyzer comprises a training unit which defines an initial state of breathing of a patient and a change unit which determines that the current state of breathing is significantly different from said initial state of breathing.

Preferably the monitor comprises a stethoscope converter, comprising at least a channel selector and at least one

speaker, which receives said breath sound data signals and provides said signals in audio form to the ears of an operator. Preferably, the monitor comprises a raw data recorder unit which records said breath sound data signals.

There is further provided, in accordance with a preferred embodiment of the invention, an apnea monitor comprising:

at least one breath related sensor placed around the respiratory system of a patient for measuring breath sound data signals; and

a breath analyzer which continuously matches the breath sound data signals produced by said at least breath related sensor to at least a regular breath sound template to determine the presence of breathing and which provides an alert indication when no breathing is present for a time period longer than a given period.

There is further provided, in accordance with a preferred embodiment of the invention, a method of determining the state of breathing of a patient, the method comprising:

determining the inspiration/expiration phase of a breath from chest movement data and defining a breath phase variable therefrom;

if the tracheal breath sound data are significant and if the external noise is low:

determining if the tracheal breath sound data has a generally normal shape; and

if so, generating breath flow data from said tracheal breath sound data and from said breath phase variable;

otherwise:

determining if the lack of flow indicates the presence of apnea and, if so, setting an apnea alarm.

Preferably, the method includes generating a loud noise indication if a) the ambient noise is high, b) the breath shape is not normal or c) the tracheal sound is too high.

Preferably, the breath flow data is defined as the tracheal sound data to a power in the range of 0.45–0.67 and the direction of the flow is defined by said breath phase variable.

There is further provided, in accordance with a preferred embodiment of the invention, a phonopneumograph system comprising:

a plurality of piezoelectric sensors placed around the respiratory system of a patient which measures breath related activity; and

a verification unit connected to said sensors operative which individually activates the sensors to produce a sound wherein with the non-activated ones of said sensors detect said sounds, wherein in a verification mode, the unit compares the measured sounds with those received during a training mode.

There is further provided, in accordance with a preferred embodiment of the invention, a hardware verification method for a system having a plurality of sound transducers placed around an object, the method comprising:

in a training mode:

individually activating at least some of said transducers to produce a sound;

for, each sound produced, measuring said sound with non-activated ones of said transducers; and

storing said measured sounds;

in a verification mode:

individually activating at least some of said transducers to produce a sound;

for, each sound produced, measuring said sound with non-activated ones of said transducers; and

identifying malfunctioning transducers by comparing the measured sounds with those received during said training mode.

There is further provided, in accordance with a preferred embodiment of the invention, a method for analyzing breath data, the method comprising:

generating the spectrum of the current segment;

determining if the current segment of data is a background segment, representing background noise, or a breath segment representing breath sounds;

averaging the spectra of the segments of each type to produce an average background spectrum and an average breath sound specimen;

for said average breath sound spectrum, subtracting said average background spectrum therefrom to produce a relatively noiseless breath spectrum;

fitting said relatively noiseless breath spectrum to a predetermined normal curve and determining the quality of the fit;

activating an irregular breath sounds detector if said quality of fit is poor.

Preferably, said predetermined normal spectrum is selected from a plurality of predetermined normal spectra defined by the group to which the patient belongs, wherein said group is one of the following: male, female and child

There is further provided, in accordance with a preferred embodiment of the invention, a method of analysis of breath sound data comprising

determining the presence of a first breath sound in the breath sound data;

adjusting the breath sound data by reducing the effect of the first breath sound in the breath sound data; and

determining the presence of a second breath sound in the adjusted breath sound data

In one preferred embodiment of the invention the first breath sound is a wheeze. Preferably, the second breath sound is a wheeze.

In a preferred embodiment of the invention the first breath sound is a crackle. Preferably, the second breath sound is a crackle.

In a preferred embodiment of the invention, reducing the effect of the first breath sound comprises substantially removing the effect of the first breath sound.

BRIEF DESCRIPTION OF THE DRAWINGS

The present invention will be understood and appreciated more fully from the following detailed description taken in conjunction with the drawings in which:

FIG. 1 is a schematic diagram illustrating a Phonopneumograph (PPG) system in accordance with a preferred embodiment of the present invention;

FIG. 2 is a schematic block diagram illustrating the PPG system of FIG. 1 in greater detail;

FIG. 3 is a schematic block diagram illustrating a PPG system configured as a PPG Meter in accordance with a preferred embodiment of the present invention;

FIG. 4 is a schematic block diagram illustrating a PPG system configured as a PPG Monitor in accordance with a preferred embodiment of the present invention;

FIG. 5 is a schematic block diagram illustrating a PPG system configured as a PPG Recorder in accordance with a preferred embodiment of the present invention;

FIG. 6 illustrates an exemplary report generated by the PPG Meter of FIG. 3 in accordance with an exemplary embodiment of the present invention;

FIGS. 7A, 7B and 7C are graphs illustrating an exemplary screen configuration of the PPG Monitor of FIG. 4;

FIGS. 8A and 8B are schematic block diagrams illustrating two additional configurations of the PPG System of FIG.

2 in accordance with two additional preferred embodiments of the present invention;

FIG. 9A is a block flow chart which illustrates the steps of a Breath Detection method in accordance with a preferred embodiment of the present invention;

FIGS. 9B–9F illustrate portions of the flow chart of FIG. 9A in greater detail;

FIG. 10 which is an exemplary graph illustrating curve fitting of the tracheal breath sound in accordance with a preferred embodiment of the present invention;

FIG. 11A is a block flow chart illustrating the steps of an exemplary breath sound analyzing method of the PPG system of FIGS. 1 and 2 in accordance with a preferred embodiment of the present invention;

FIGS. 11B and 11C illustrate portions of the flow chart of FIG. 11A in greater detail.

FIG. 11D is an exemplary graph illustrating curve fitting of an exemplary chest breath sound power spectrum in accordance with a preferred embodiment of the breath sounds analyzing method of the present invention;

FIG. 12 is a flow chart illustrating the steps of an exemplary wheeze detection method in accordance with a preferred embodiment of the present invention;

FIGS. 13A and 13B are graphs illustrating an exemplary amplitude spectra calculated by different steps of the wheeze detection method of FIG. 12 in accordance with a preferred embodiment of the present invention;

FIG. 14 is a graph illustrating part of an exemplary rhonchus sampled in a patient with pneumonia and lung cancer,

FIG. 15 is a flow chart illustrating the steps of an exemplary rhonchus detection method in accordance with a preferred embodiment of the present invention;

FIGS. 16A and 16B are graphs illustrating exemplary curves of a "simple" snore in the time and frequency domains, respectively;

FIGS. 17A and 17B are graphs illustrating exemplary curves of a "complex" snore in the time and frequency domains, respectively;

FIG. 18 is a flow chart illustrating the steps of an exemplary snore detection method in accordance with a preferred embodiment of the present invention;

FIGS. 19A and 19B are graphs illustrating an exemplary temporal structure of the sound and air flow data of a recorded cough;

FIG. 20 is a flow chart illustrating the steps of an exemplary cough detection method in accordance with a preferred embodiment of the present invention;

FIG. 21 is a graph illustrating the sounds of exemplary fine and coarse crackles recorded from a patient with pulmonary fibrosis;

FIG. 22 is a flow chart illustrating the steps of an exemplary crackle detection method in accordance with a preferred embodiment of the present invention;

FIG. 23 is a graph of a breath sounds segment prior to being processed by the crackle detection method of FIG. 22;

FIG. 24 is a graph illustrating the second power of the second derivative (Z) calculated from the raw data values of FIG. 23 by the crackle detection method of FIG. 22; and

FIG. 25 is a graph illustrating the values W calculated by the crackle detection method of FIG. 22, by "folding" the values Z of FIG. 24.

DETAILED DESCRIPTION OF PREFERRED EMBODIMENTS OF THE INVENTION

The Phonopneumograph System

Reference is now made to FIG. 1 which is a schematic illustration of a Phonopneumograph (PPG) system constructed and operative for detecting, analyzing, and monitoring normal and adventitious breath sounds, documenting the results of the breath sounds analysis and recording selected portions of breath sounds in accordance with a preferred embodiment of the present invention.

The PPG system preferably includes a plurality of preferably wireless, chest, breath sound (BS) sensors 4, placed in contact with the patients skin at predetermined back and chest positions, for picking up breath sounds. The PPG system preferably also includes a preferably wireless tracheal BS sensor 6, placed at a predetermined tracheal position, for picking up tracheal breath sounds. The PPG system preferably further includes a preferably wireless chest expansion (CE) sensor 8, such as a chest impedance plethysmograph sensor, suitably attached to the patient's chest for picking up chest expansion.

The PPG system preferably also includes a wireless ambient noise microphone 12, such as a suitable air coupled microphone, placed near the patient for picking up ambient acoustic noise.

The sounds picked up by the BS sensors 4 and 6, the CE sensor 8 and the ambient noise microphone 12 are transmitted to a remote receiver 16 by any suitable means for wireless transmission, such as radio, ultrasonic or infrared transmission, or b wire. The PPG system preferably also includes an analog signal conditioner 18 connected to receiver 16 for amplifying, filtering and digitizing the sensors' analog signal. Preferably receiver 16 is a remote receiver and signals are transmitted to it by wireless transmission.

The PPG system preferably further includes a central processing unit (CPU) 20 suitably connected to analog signal conditioner 18 and to a data storage device 22 for processing the digitized sensors' signals, preferably using a method for analysis of breath sounds as described in detail hereinbelow. It is noted that the CPU 20 can be a CPU of a commercially available personal computer or any other suitable, commercially available CPU.

The PPG system preferably-further includes a display 28 connected to the CPU 20 for on-line or off-line displaying of the results of the breath sound analysis, an alert/alarm unit 34 suitably connected to CPU 20 for generating of audible or visible alert and alarm signals and a user interface 32.

The user interface 32 preferably receives input from an operator and can be any suitable type of interface such as a keyboard, a mouse, a touch sensitive screen a control panel or any combination thereof.

The PPG system preferably additionally includes an electronic stethoscope 14 suitably connected, preferably, to analog signal conditioner 18 and to a D/A converter 36 for enabling the PPG operator to listen to the breath sounds picked up by any BS sensor selectable from the plurality of BS sensors. The electronic stethoscope 14 preferably includes a channel selector 38 connected between analog signal conditioner 18 and an audio amplifier 39 which, together, select the desired sensor out of the plurality of BS sensors. The audio amplifier 39 is preferably connected to earphones 24 or a loudspeaker (not shown) for listening to the breath sounds picked up by the selected BS sensor. Preferably the operator can use the user interface 32 to select

between a first mode, in which he can listen in real time to the sounds picked up by a selected BS sensor, and a second mode, in which he can listen to selected, previously digitized and recorded sounds from any selected BS sensor which are stored in the digital data storage device 22.

Listening to selected, previously digitized and recorded sounds is useful for simultaneous auscultation and visual monitoring of the breath sounds of a selected sensor. Additionally, this feature is useful for teaching and training purposes where it might be desirable for a student or trainee to simultaneously follow both the audible sound from a sensor and the analyzed features of breath sounds on the monitor.

It is noted that, in situations where the patient is acoustically isolated from the PPG system, for example when they are positioned in separate rooms or when playing back stored data, it is also possible to connect the audio output through a suitable amplifier to a speaker system, thus enabling a group of people to listen to the amplified sound while simultaneously following the monitor data display.

The data storage device 22 can be used for storing certain selected segments of digitally recorded sensor data as well as the results of the breath sound analysis. It is noted that the digital storage device 22 can be any type of suitable digital data storage device such as a magnetic, optical, or magneto-optical digital data storage device.

The PPG system preferably also includes a printer 26 for generating a hard-copy printout of the data output of the breath sound analysis such as graphs, tables and statistics of the results of the breath sound analysis. The hard-copy printout can be used as a condensed or detailed report for the physician, for comparison with previously generated reports and for archival purposes. Additionally, the storage device may be of the removable medium type providing a further type of permanent archival of data.

It is noted that BS sensors 4 and 6, CE sensor 8 and ambient noise microphone 12 can be any suitable type of breath sound sensor, chest expansion sensor or microphone, respectively, having a suitable frequency response. In an exemplary embodiment of the present invention, the BS sensors are contact sensors having a generally flat frequency response in the range 75–2500 Hz (±3 dB), the CE sensor 8 is an impedance chest expansion sensor having a generally flat frequency response of 0–3 Hz (±3 dB), and the ambient noise microphone 12 is an air coupled microphone having a generally flat frequency response of 50–5000 Hz (±3 dB).

Preferably, the BS sensors are of the wireless contact sensor type as shown in FIG. 1 and disclosed by the present inventor in U.S. patent application 08/654,643 filed on May 29, 1996 and entitled "A Contact Sensor for Body Sounds". The CE sensor 8 and ambient noise microphone 12 are also preferably wireless. However, any other type of wireless or wired BS sensor, CE sensor or ambient noise microphone, having a suitable frequency response characteristics and a suitable signal to noise ratio (SIN), can be used.

Preferably, the output of the BS sensors is independent of the magnitude of the contact force exerted by the patient's skin on the sensor.

It is noted that, in accordance with another preferred embodiment of the present invention, the BS sensors 4, the CE sensor 8 and the ambient noise microphone 12 can be electrically connected directly to the analog signal conditioner 18 by suitable wires. In this preferred embodiment, the remote receiver 16 is not included. It is noted that, if wired sensors are used, caution should be exercised to avoid introduction of artifactual noise due to friction and motion of the wires.

Reference is now made to FIG. 2 which illustrates a general schematic functional block diagram of the PPG system of FIG. 1. It is noted that the following description is a general functional description of the mode of operation of the PPG system, based on FIGS. 1 and 2, while the methods of operation of the preferred embodiments of the PPG system are separately described and illustrated in the figures hereinafter.

The signals of the BS sensors 4 and 6, CE sensor 8 and the ambient noise microphone 12 are transmitted to the receiver 16 (not shown for the sake of clarity of illustration) and sent to the analog signal conditioner/digitizer 18, where they are filtered, amplified and digitized. The sensors' signals are amplified, preferably at a fixed gain of X1,000–X1,000 or with an automatic gain control, band-pass filtered, preferably with a bandwidth of approximately 75–2500 Hz, 18 dB/octave and preferably digitized by an A/D converter for example a 12–16 bit ADC at 5,500–11,000 samples per channel per second.

Consecutive digitized data segments from all the sensors are sent to the CPU 20 for processing. The duration of the data segment can vary depending on the specific detection method used for detection of specific types of breath sounds. An exemplary data segment is 50 ms but it can vary in the range of 25–500 ms or more. For example the snore detection method uses a segment duration of 250±50 ms. Thus, the segment duration is preferably optimized to suit each specific adventitious sound detection method depending on the type of mathematical analysis which is used by the method for detecting specific breath sounds and on the duration and frequency content of the analyzed breath sound.

The data segments are sent to a channel specific digital data conditioner 44 where the data segments from the different sensors and the ambient noise microphone are individually processed as described in detail hereinafter. For the sake of clarity, the differently conditioned data segments of all the sensors and the ambient noise microphone are hereinafter collectively referred to as the conditioned data segments. The data segments are transferred to a memory (not shown) linked to the CPU 20 and can be also stored in the data storage device 22.

A segment of the digitized data of the ambient noise microphone 12 is rectified and integrated by the channel specific digital data conditioner 44 and sent to an ambient noise level detector 46 for evaluation. The amplitude of the conditioned ambient noise data is compared to a reset or adaptive threshold. If the amplitude of the conditioned ambient noise data is above the threshold, the system generates a control signal which diverts the conditioned data segment to a snore/cough/vocalization/noise analyzer 48 for evaluation of the acoustic characteristics of the noise and detection of snoring, cough and vocalization sounds that may be generated by the patient or by external non-patient generated noise.

The snore/cough/vocalization noise analyzer 48 receives the conditioned data segments from the tracheal BS sensor 6 and the ambient noise microphone 12. If a snore or cough are detected, the parameters of the snore or cough are calculated and sent to a primary post-processor 56 where the snore or cough count, timing and time distribution are updated and logged (by storing in data storage device 22). If no snore or cough is positively identified, the system suspends the analysis of all the conditioned data segments for as long as the noise level exceeds the threshold value. The system also records the noise duration. If the duration of the

noise is longer than a user determined preset value, an "ambient noise alert" or an "ambient noise alarm" signal can be sent to the primary post-processor 56 and issued by the system (depending upon the specific system configuration).

If the ambient noise level is below the threshold, the system proceeds with the analysis of the sensors' data segments. The conditioned data segments of the CE sensor 8 and the tracheal BS sensor 6 are sent to a breath state analyzer 50. The breath state analyzer 50 receives the conditioned data segments from tracheal BS sensor 6 and the CE sensor 8 and determines the presence or absence of breathing for detection of periods of apnea (cessation of breathing). The breath state analyzer 50 also calculates the breathing rhythm (breathing rate), the breath duration and amplitude and the breathing regularity, and outputs the data to the primary post-processor 56 for further processing.

In the case of detected apnea, a breathing rhythm alert is also sent to the primary post-processor which may initiate the generation of an "apnea alarm". Additionally the breath stage analyzer determines the breath phase (inspiration or expiration), its duration and its timing which are subsequently used as references for determining the temporal relationship of detected adventitious sounds to the breathing cycle.

The breath phase and amplitude data of the breath stage analyzer are sent as input to a breath sounds analyzer 52. The breath sounds analyzer 52 also receives the conditioned data segments from the tracheal BS sensor 6 and the plurality of chest BS sensors 4. The breath sounds analyzer characterizes the normal and abnormal breath sounds, detects adventitious sounds (wheezes, crackles and rhonchi) in the data segments, calculates the parameters and timing of the various detected adventitious sounds and outputs these data and an adventitious sound alert or flag to the primary post-processor 56 for further processing and logging. The breath sounds analyzer 52 also analyzes the normal breath sounds as disclosed in detail hereinafter and outputs the analyzed breath sound data to the primary post-processor 56 for further processing and logging.

It is noted that the data segments of the various BS sensors 4 and 6 are separately conditioned by the channel specific digital data conditioner 44 and are also separately analyzed by the breath sounds analyzer 52.

The primary post-processor 56 receives the analyzed breath sound data of each of the BS sensors 4 and 6 and performs an analysis of channel inter-relationships, calculates the statistics of events and the adventitious sound distribution curves (histograms), performs a trend analysis of the various breath sound and adventitious sound data, and calculates three quality control parameters. The first quality control parameter defines the degree of matching of the normal breath sound power spectrum curve to a template. The second quality control parameter defines the degree of matching of a detected crackle sound to a crackle template. The third quality control parameter defines the degree of matching a detected wheezing sound to a wheeze template.

The primary post-processor 56 also further processes the various alarm, alert and flag conditions and sends instructions to a secondary post processor 60 for issuing appropriate alarm or alert signals.

The secondary post-processor 60 receives data from the primary post-processor 56 and operator instructions from the user interface 32 and processes the data and the instructions to generate graphic, numerical and textual data.

The graphic, numerical and textual data generated by the secondary post-processor 60 can be sent as output to the

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display 28 for display, to the printer 26 for generating a hard copy report or as data files to be transferred electronically to another remote computer or telemedicine center, by modem or any other suitable means for electronic data transfer as described in detail hereinafter. Additionally, the data files created by the secondary post-processor 60 can be stored on any suitable means for data file storage, such as removable magnetic media, opto-magnetic media or other types for media, for later printing or archival purposes.

Reference is now made to FIGS. 3, 4 and 5 illustrating three different configurations of the PPG system of FIGS. 1 and 2. It is noted that, for a better understanding, like components are designated by like reference numerals throughout the various figures.

It is generally noted that, in the various configurations of the PPG system shown in FIGS. 3, 4 and 5 the data processing is similarly performed. The differences between the configurations are in the preferred type of sensors, the wired or wireless transmission of the sensors' signals, the secondary post-processor and the type of available output devices, such as the printer, display monitor and alarm/alert unit.

In accordance with a preferred embodiment of the present invention, selected breath sound data and parameters may be continuously displayed on display 28 as graphs or numerically, or printed out as part of a report, depending on the specific configuration of the PPG system.

The PPG Meter

FIG. 3 illustrates a PPG system configured as a PPG Meter in accordance with a preferred embodiment of the present invention. The PPG Meter preferably includes a plurality of, preferably, wired chest BS sensors 64, a preferably wired tracheal BS sensor 66, a preferably wired pneumotachograph 68 and a preferably wired ambient noise microphone 62.

It is noted that, although the preferred embodiment of the PPG Meter uses the pneumotachograph 68 for flow measurement, any other sensor suitable for flow detection can be used, such as an impedance spirometry sensor or any other suitable chest expansion detecting sensor.

It is also noted that, although in the preferred embodiment of the PPG Meter of FIG. 3 the BS sensors 64 and 66, the ambient noise microphone 62 and the pneumotachograph 68 are all wired, other preferred embodiments of the PPG Meter may use wireless sensors, a wireless microphone or a wireless pneumotachograph, together with a suitable wireless receiver (FIG. 1) for receiving the respective signals thereof and sending them to the analog signal conditioner/digitizer 18.

The PPG Meter also preferably includes the analog signal conditioner 18 which is connected to the CPU 20 and preferably to the electronic stethoscope 14 (not shown for clarity of illustration). The CPU 20 operates as described hereinabove and is preferably connected to a secondary post-processor 61 which is preferably connected to the printer 26 and the display 28.

The secondary post-processor 61 further processes the data from the primary post-processor 56 and the operator input to generate graphic, textual and numeric output which is displayed on the screen of display 28 or can be printed as a hard copy report by the printer 26.

The PPG3 Meter is used to obtain a spot measure of the patient[]s breath sounds. A plurality of sensors are attached to the patients chest and anterior neck and the pneumotachograph 68 (a flow detecting device) is activated.

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While the patient breathes, the system obtains and analyzes the sounds to provide an optional immediate output in the form of a hard copy report showing the features of the sounds in graphic and parametric notation.

The breath sounds are analyzed by the CPU 20 preferable to determine the spectral pattern of the basic chest wall and tracheal sounds and to detect and characterize adventitious sounds. The parameters of the basic patterns are preferably calculated and recorded by breath sounds analyzer 52 (FIG. 2). Any adventitious sounds detected are classified, the relevant quantitative parameters (e.g., wheeze duration, timing and frequency range, crackle timing, count, and parameters, etc.) are determined, presented and recorded using standard notations.

Reference is now made to FIG. 6 which illustrates an exemplary report generated by the PPG Meter of FIG. 3. The exemplary report includes eight graphs arranged in two rows. The top row of four graphs, labeled TR-I, CR-I, BR-I and BL-I, represents the averaged inspiratory breath sound (BS) power spectra of the operator selected tracheal, right anterior chest, right posterior lung base and left posterior lung base sensors, respectively. The lower row of four graphs, labeled TR-E, CR-B, BR-E and BL-E, represents the averaged expiratory breath sound (BS) power spectra of the same sensors as the top row, respectively. In all the -eight graphs the vertical axes represent the log amplitude. The horizontal axes represent either the frequency (TR-I and TR-E), or the log frequency (CR-I&E, BR-I&-E and BL-I&E). Each of the graphs includes a curve labeled 73 representing the calculated time average power spectrum of the respiratory breath sound picked up by a selected sensor. If the calculated curve fit was significant, the graph also includes an additional curve labeled 75, representing the calculated fitted curve. The exemplary report of FIG. 6 also includes four numerical parameter reports, labeled 77, which include the numerical parameters fit to the spectra shown in the graphs. The parameters reported are A, $\int_0$, S and Q as hereinafter described It is noted that the report can also include the number of detected crackles and their relative timing within the respiratory cycle, the parameters of detected wheezes and a textual description of the sounds.

The PPG Meter can be used to evaluate the pulmonary health status of patients of all age groups from neonates to the very elderly.

The PPG Meter can be used in primary care clinics to evaluate any patient who presents with respiratory symptoms. (An analogy may be the use of the electrocardiogram to evaluate any patient who presents cardiac-related symptoms).

The PPG Meter may also be used for follow-up of the conditions of patients with established pulmonary ailments, using their previous quantitative breath sound records as a baseline.

The PPG Meter can further be used for evaluating the effectiveness of a drug treatment on a specific individual or in a group of patients. The quantitative records of the PPG Meter allows a quantitative comparison of pretreatment with post-treatment results. This is a considerable improvement over the prior art use of auscultation results for drug treatment evaluation since it allows the comparison of quantitatively determined breath sound parameters, thus eliminating possible bias of the results due to personal differences in the experience, training and hearing acuity of the evaluating physicians.

An additional advantage of the PPG Meter over the prior art is the fact that the records of the PPG Meter can be used as documentation and evidence when legal issues arise.

The PPG Monitor

FIG. 4 illustrates a PPG system configured as a PPG Monitor in accordance with another preferred embodiment of the present invention. The PPG Monitor includes the same components as described for the PPG system hereinabove and illustrated in FIG. 2 except that a secondary post-processor 62 replaces the secondary post-processor 60 of FIG. 2 and that the PPG Monitor preferably additionally includes the display 28, the printer 26 and the alarm/alert unit 34 of FIG. 1.

It is noted that, in accordance with a preferred embodiment of the present invention, the plurality of sensors of the PPG Monitor may also include an esophageal sensor (not shown). For example, in cases of anaesthetized and artificially ventilated patients with endotracheal intubation, an esophageal sensor may be used.

It is also noted that the CE sensor 8 of the PPG Monitor of FIG. 4 may be any suitable chest expansion or air flow measuring sensor, such as a piezoelectric chest expansion sensor, a strain gauge chest expansion sensor, an electric impedance plethysmograph sensor, a bellows with a pressure transducer, a magnetometer or an inductive plethysmograph.

The PPG Monitor is a respiratory acoustic monitor. It allows on-line, breath-by-breath monitoring and determination of the health and integrity of the respiratory system. It is intended for use in situations where continuous monitoring of respiration is important. Examples include general anesthesia with endotracheal intubation; adult, pediatric, and neonatal critical care medicine; emergency medicine; pulmonary medicine where bronchial provocation tests are used, and in individuals (infants and adults) who are at increased risk for sudden death due to breathing irregularity. The PPG Monitor performs the analysis of breath sounds as generally described hereinabove for the PPG system and illustrated in FIG. 2.

An additional feature of the PPG Monitor is that it initially "leans" the features of the sounds of the specific patient. A catalog is generated which is subsequently used as a reference for evaluation of changes and trends. If abnormal sounds are detected during the initial period, generally referred to as the "training" period hereinafter, the PPG Monitor provides an alert signal during this stage by activating the alert/alarm unit 34. Following the initial "training" period, the PPG Monitor carries out a breath-by-breath analysis of the breath sounds.

The CPU 20 evaluates the intensity, timing, spectral content, and specific properties of the sounds from each sensor, with respect to the ambient noise and in comparison with the reference catalog. The PPG Monitor searches for specific events such as wheezes or crackles, breath sound intensity shifts (throughout all sensors or in one compared with others), breaching irregularities such as apnoca or periodic breathing, and the presence of rhonchi, cough, stridor, secretion sounds or snores. Once the onset of such an event is detected, the system generates an alarm signal and may also record information regarding the timing and frequency content of the event in a data log. The primary post-processor 56 calculates and presents the trends and statistical information of the respiratory rate, apnea duration distribution, relative duration of wheezes and the counts of crackles, cough, and snores.

The PPG Monitor screens for and parametrizes the following features of the breath sounds:
1. The presence, timing, intensity, and spectral content of ambient noises.

2. The presence of breath sounds.
3. The correlation between the sounds picked up by the chest, tracheal (and possibly the esophageal) sensors.
4. The matching of the spectral patterns of the breath sounds to those of the corresponding normal templates.
5. The matching of the spectral patterns to those of the corresponding templates identified during the most recent "training" period.
6. The presence of wheezes and their parameters.
7. The presence of crackles and their parameters.
8. The presence of secretion sounds (expiratory crackles, rhonchi).
9. An imbalance between the signals reaching the lateral chest wall sensors.
10. Temporal irregularity of the breathing sounds.
11. The presence of coughs.
12. The presence of snores.

The outcome of the screening procedures is a decision tree that determines whether the breath sounds have been altered in a clinically significant fashion that evokes the triggering of an alarm, as is described in detail hereinafter.

The PPG Monitor preferably acquires data continuously from the BS sensors 4 and 6, the ambient noise microphone 12, and the CE sensor 8. The data are sampled in contiguous or partially overlapping segments. The signal processing of each segment is performed while the next segment is being acquired, so that there is a short delay, of a duration of less than one segment (approximately 50–100 ms), between the data acquisition and their presentation. Exceptions are the wheeze detection method and the snore detection method that compile data from more than one segment (approximately 3–5 segments) to verify the presence of the corresponding sounds.

Reference is now made to FIGS. 7A, 7B and 7C which together illustrate an exemplary screen configuration of the PPG Monitor's display. Three graphs are shown simultaneously on the monitor's display in this exemplary screen configuration, the wheeze monogram graph (FIG. 7A), the BS-based flow indicator graph (FIG. 7B) and the crackles per segment graph (FIG. 7C).

FIG. 7A illustrates the wheeze sonogram graph. The vertical axis represents the wheeze frequency and the horizontal axis represents time. The wheezes trace 113 indicates the presence of wheezes and shows the wheeze frequency as a function of time. The duration of wheezes compared with the total duration of active breathing (Twh/Ttot), labeled %Wz, is shown next to the wheeze sonogram traces 113. A color coding may be used to indicate the confidence level of the wheeze identification (which represents the degree of matching of the detected sound to the wheeze template). For example, a red color may represent absolute confidence, an orange color may represent "probable" wheeze detection, and a yellow color may represent "possible" wheeze detection.

FIG. 7B illustrates the flow indicator graph. The vertical axis represents the flow rate and the horizontal axis represents time. The flow indicator trace 111 shows the timing and regularity of breathing. When the breath state analyzer 50 is calibrated against a conventional flow-measuring device, such as a pneumotachograph, spirometer, or a mechanical ventilation apparatus, the PPG Monitor can also generate a quantitative measure of the flow amplitude. The respiratory rate, labeled RR, is given next to the flow indicator trace 111.

FIG. 7C illustrates the crackles per segment graph. The vertical axis represents the number of crackles per segment and the horizontal axis represents time. The graph displays the presence of crackles as vertically stacked marks 115,

each mark representing the number of crackles detected in the segment The number of crackles per minute, labeled Cx/Min, is shown next to the crackles per segment marks **115**.

It is noted that the time axes of the tree graphs of FIGS. 7A, 7B and 7C are synchronized on the display of the PPG Monitor so that the user can discern the timing of the crackles or wheezes that are detected within the respiratory cycle.

When excessive ambient noise is detected, the PPG Monitor indicates this condition by changing the background color of all the panels on the screen of display **28** and by not displaying the traces representing the detected breath sounds. The presence of a cough, rhonchi, or snores is indicated using specific icon notation on a separate strip displayed on the bottom of the breath detection (flow) panel (not shown).

Two types of alarm are used by the PPG Monitor, an absolute alarm signal and a relative alert signal.

An absolute alarm signal is generated when an apnea that is longer tan a predetermined duration (approximately 6–10 seconds) is detected. A relative alert signal, which indicates a change in the sounds relative to the conditions during the most recent "training" period, is generated under the following circumstances:

a) Detection of wheezes that were not previously present.

b) A significant change (for example a 20% increase) in the duration of wheezes compared with their duration in the most recent "training" period.

c) Detection of crackles that were not previously present.

d) A significant change (for example a 20% increase) in the number of crackles per breath or per time unit compared with the most recent "training" period.

e) A significant change of the respiratory rate, for example a 20% increase or decrease.

f) The presence of secretion sounds (rhonchi or expiratory crackles).

g) Detection of an imbalance between the outputs of a plurality of sensors placed on the two opposite sides of the thorax. For example, an imbalance between the output of the left and right anterior chest sensors might be sensed.

Whenever an alarm or alert signal is issued, a specific explanatory text message is simultaneously displayed on the display **28**.

According to a preferred embodiment of the present invention, the PPG Monitor, in addition to the evaluation of the data from each individual sensor as described hereinabove, detects imbalance between sounds from different sites over the chest. In particular, detection of changes in the sounds detected on the left and right sides of the thorax in a patient who has been intubated by an endotracheal tube is important. Such changes may indicate a malposition of the tip of the tube to preferentially ventilate one of the lungs compared to the other. To detect these changes, the PPG Monitor keeps a log of the parameters of the sounds from the two sides of the chest. Whenever an imbalance is suspected, the primary post-processor **56** (FIG. **2**) calculates the transfer function of the sounds (magnitude, phase and coherence) between the two sites over the thorax and between the chest sites and the tracheal sensor.

Any significant change (increase or decrease) in the magnitude of the transfer function, phase or coherence relative to the "training" period is identified by the primary post-processor **56** and induces an alarm.

The spectral content of the basic breath sounds is continuously evaluated by the breath sounds analyzer **52** which performs a curve fit to a template equation and calculates a

set of parameters from the best-fit equation disclosed in detail hereinafter. Any significant changes in the values of the parameters of the best-fit equation compared with their values obtained during the most recent "training" period are identified.

The PPG Recorder

We now return to FIG. **5** illustrating the PPG system of FIG. **2**, configured as a PPG Recorder in accordance with a preferred embodiment of the present invention. The PPG Recorder preferably includes the same BS sensors **4** and **6**, the ambient noise microphone **12** and the CE sensor **8**. Preferably, the BS sensors **4** and **6** are wireless contact sensors, the ambient noise microphone **12** is preferably a wireless microphone and the CE sensor **8** is preferably a wireless chest expansion sensor (FIG. **1**) for reducing artifactual wire function noises caused by patient movements during sleep.

The PPG Recorder preferably also includes the remote receiver **16** of FIG. **1** (not shown in FIG. **5** for the sake of clarity of illustration) for receiving the signals from the sensors **4, 6** and **8** and the ambient noise microphone **12** and transmitting them to the analog signal conditioner/digitizer **18**, where the signals are amplified, conditioned and digitized as described hereinabove. The data is then sent to the CPU **20** for on-line analysis as described hereinabove for the PPG System of FIG. **2**.

The PPG Recorder can include the electronic stethoscope **14** of FIG. **1** for remote verification of sensor placement and adequate skin contact thereof.

The primary post-processor **56** receives the results of the data analysis for firer processing and preferably provides graphic presentations, parametric summaries, analyses of temporal trends, and histograms. The input parameters include information on each sound segment that include the following parameters, described in detail hereinafter, from each sound pick up location:

The time of day;

The environmental noise amplitude;

The amplitude of the tracheal sound signal;

Inspiratory, Expiratory, or Apnea segment designation [−/+/ 0];

Amplitude of the sound signal in this channel (RMS amplitude);

Parameters of the basic breath sound spectrum (Amplitude, $\int_0$, slope);

Wheeze presence or absence in the segment;

If wheezes are present, Wheeze frequency (frequencies): $\int_1$, $\int_2$, $\int_3$, . . in segment;

Crackles presence or absence in the segment;

If crackles are present, number of crackles in segment;

If crackles are present, the parameters of best fit equation of each crackle;

Rhonchi presence or absence in segment;

Snore presence or absence in segment;

If snore is present in segment, the snore classification (simple/complex);

Cough presence in segment;

Transfer functions (if two or more sensors are used);

Magnitude, Coherence and phase of the transfer function.

The primary post processor **56** performs a short range, for example a minute-by-minute analysis, and a longer range analysis, for a selectable duration ranging from approximately 15 minutes to 24 hours. The short range analysis generates only numerical data on the last minute, for example, the last running minute, updated every 10 seconds. These analyses include the mean respiratory rate; the mean ratio of inspiratory and expiratory duration (T_{insp}; T_{exp},

respectively) to the total duration of active breathing (T_{total}) (T_{isnp}/T_{total} and T_{exp}/T_{total}, respectively); mean inspiratory and expiratory amplitudes (A_{insp}, A_{exp}, respectively); the total duration of wheezes (T_{Wh}) relative to the duration of active breathing (T_{Wh}/T_{total}); the duration of inspiratory wheezes to the active inspiratory duration ($T_{msp\text{-}Wh}/T_{insp}$); the duration of expiratory wheezes to the active expiratory duration ($T_{exp\text{-}Wh}/T_{exp}$); the total number of crackles per minute (N_{Cx}); the number of inspiratory crackles per minute ($N_{insp\text{-}Cx}$); the number of expiratory crackles ($N_{exp\text{-}Cx}$); the duration of snores relative to the total duration (T_{Snore}/T_{total}); the number of coughs per minute (N_{Cough}); and an indication of ambient noise level. The primary post processor 56 also calculates inter-channel parameter relationships, frequency distribution (histograms) of the respiratory rate RR, the breathing amplitude V_T and minute ventilation V_E and statistics of all the parameters, and sends the data to a data log stored in storage device 22. The primary post-processor 56 is also preferably connected to a secondary post-processor 63 for further processing of the data to the appropriate form for data logging and outputting in a suitable file format for creating a permanent record of the logged data on a removable data storage medium, such as a floppy diskette, a removable hard disk cartridge, a removable opto-magnetic disk or any other suitable removable data storage medium.

Additionally, the secondary post-processor controls the electronic transfer of data log files by receiving appropriate operator instructions through the user interface 32. The secondary post-processor also controls the printing of a PPG recorder report by a suitable printer.

The parameters of the analyzed breath sounds are stored as a data log in the data storage device 22.

The PPG Recorder can thus perform a continuous, quantitative evaluation of the breathing parameters and parameters of the adventitious sounds of sleeping patients with suspected nocturnal respiratory symptoms.

Reference is now made to FIGS. 8A and 8B illustrating two additional configurations of the PPG System of FIG. 2.

FIG. 8A illustrates a PPG System 70 constructed and operative in accordance with a preferred embodiment of the present invention. PPG System 70 includes a plurality of sensors 72 which are suitably attached to a patients body as described hereinabove and illustrated in FIG. 1. The sensors 72 are suitably connected to an analog signal conditioner/ digitizer 74 which amplifies, conditions and digitizes the sensors' signals as described hereinabove and sends the amplified and conditioned and digitized signals through a modem 76 to another remote modem 82. The modems 76 and 82 communicate either directly through the Public Switched Telephone Network (PSTN) or by using digital data packet switching protocols of a Wide Area Network (WAN).

The modem 82 is a part of a Telemedicine Center 80 and is connected to a CPU 84 which is a part of the Telemedicine Center's PPG System (the parts thereof are not shown for the sake of clarity of illustration). The CPU 84 performs the analysis of breath sounds as described hereinabove and illustrated in FIG. 2. Thus, the PPG system of the Telemedicine Center 80 can generate all the reports, alarms, alerts, and data-logging functions of any of the configurations of the PPG System of the present invention described hereinabove and illustrated in FIGS. 2–5.

FIG. 8B illustrates a PPG System 90 constructed and operative in accordance with an additional preferred embodiment of the present invention. The PPG System 90 includes a plurality of sensors 92 suitably connected to an

analog signal conditioner/digitizer 94 for amplifying filtering and digitizing the signals of sensors 92. The PPG System 90 further includes a secondary CPU 96 suitably connected between the analog signal conditioner/digitizer 94 and a modem 98. The modem 98 communicates with a modem 102 through a suitable communication line 100. The modem 102 is also suitably connected to a primary CPU 104 which is a part of a Telemedicine Center 106. The secondary CPU 96 performs a part of or all of the breath sound analysis as described hereinabove and illustrated in FIG. 2 and communicates the partially processed data or the completely processed data, respectively, to the primary CPU 104 of the Telemedicine center 106 where the data processing is completed and the data is logged or the completely processed data is logged, respectively.

It is noted that, in accordance with a preferred embodiment of the present invention, the analog breath sounds which are picked up by the sensors can be communicated using a telephone line to a telemedicine center where the received analog breath sounds are fed as input to a PPG system for further analysis.

It is flirter noted that either the primary CPU 104 or the secondary CPU 96 may be equipped with any of the combinations of the input or output devices described hereinabove and illustrated in FIGS. 1–5.

It is also noted that the sensors 72 and 92 of the PPG Systems 70 and 90, respectively, can be implemented as wired or wireless sensors in accordance with different preferred embodiments of the present invention.

It is still further noted that, in accordance with a preferred embodiment of the present invention, the data log recorded by the PPG recorder can be electronically transferred to a telemedicine center (FIG. 8B) by a modem using a direct modem-to-modem link or the digital data packet protocols of a wide area network (including the Internet). This feature has the advantage that the stored data log does not need to be physically transported on a removable storage medium and can be quickly available to a remotely located physician for immediate interpretation and use.

Breath Analysis Methods

Reference is now additionally made to FIGS. 9, 11, 12, 15, 18 and 22, illustrating in detail the flow control and data processing steps of the breath analysis methods and the adventitious sound detection methods used by the PPG System in accordance with a preferred embodiment of the present invention.

The Breath Detection Method

Breath detection, in accordance with a preferred embodiment of the invention, combines data from the tracheal breath sounds sensor and from the chest expansion sensor to obtain a tentative detection of breathing activity (as oppose to apnea) and the respiratory phase. This detection is then validated by verifying that the tracheal breath sounds spectral characteristics match the mathematical template of authentic tracheal breath sounds. The template of tracheal breath sounds reflects an understanding of the aerodynamics of tracheal sound generation.

The breath detection method accepts as input the signals picked up by the tracheal BS sensor 6 or 66, the ambient noise microphone 12 or 62 and the CE sensors 4 or 64 of the PPG Systems of FIGS. 1–5 and processes the signals to yield as output the following parameters: the breath flow rate, the times of onset of the inspiration and expiration phases of breathing, the breathing rate (breaths/time unit), the breath amplitude, the breathing regularity and the timing and duration of apnoea.

The breathing regularity is defined as: $(\sigma_{RR}/RR)\text{-}100$, where σ_{RR} is the variance of the breathing rate and RR is the breathing rate.

The breath detection method also outputs control signals sent to the alert/alarm unit **34** of FIG. **1** (if appropriate to the specific configuration) for initiating a variety of alarms or alerts. The breath detection method also determines whether adventitious sounds will be searched for.

FIG. **9A** is a block schematic flow chart which illustrates the details of the Breath Detector method. FIGS. **9b–9F** illustrate portions of the flow chat of FIG. **9A** in greater detail. Hereinafter, the term FIG. **9** is used to refer to one or more of FIGS. **9A–9F**.

The Breath Detector method is divided into seven major blocks A, B, C, D, E, F and G. Block A picks up the analog signals of the BS and CE sensors and the ambient noise microphone, conditions and digitizes the signals and then performs channel specific conditioning operations on each of the individual digitized data segments.

The system in block A preferably picks up the chest expansion signal from the CE sensor **8** or **68** (step **110**), and amplifies and band-pass filters the signal (step **116**). The system then digitizes the signal (step **122**) and digitally filters the signal for obtaining the smoothed impedance parameter labeled I_i (step **128**) which is output to block B.

It is noted that the chest expansion sensor can be any suitable CE sensor such as an electrical impedance sensor, an inductive plethysmograph, a magnetometer sensor, a flow sensor or a bellows sensor or any other sensor that can sense breathing movement.

The system in block A preferably also picks up the analog signal from the ambient noise microphone (step **112**), amplifies the signal and band-pass filters it (step **118**). The system then digitizes the signal (step **124**) and rectifies and then integrates the signal for obtaining the rectified and integrated ambient noise parameter labeled A_i (step **130**) which is output to block C.

The system in block A also preferably picks up the analog signal from the tracheal BS sensor (step **114**), amplifies and band-pass filters the signal (step **120**). The system then digitizes the signal (step **126**) and rectifies and then integrates the signal for obtaining the rectified and integrated tracheal breath sound amplitude parameter labeled T_i (step **132**) which is output to block D.

Note that while in block A, the signals are first digitized and then rectified, smoothed and/or integrated, an alternative preferred system consists of performing the digitizing into the computer after one or more of the other steps.

Block C analyzes the noise level and controls the further processing of the data segments by the system.

Block C preferably receives the parameter A_i from step **130** and compares it to a preset or adaptive noise threshold value (step **136**). If A_i is larger than the noise threshold value, the system generates an ambient noise alert (step **152**) and diverts control to block G which analyzes the ambient noise for acoustic characteristics of a snore, cough or stridor sound that may be generated by the patient. If A_i is smaller or equal to the noise threshold value, the system proceeds with the analysis of the tracheal breath sounds by transferring control to step **132** of block A.

In block D, the system preferably first compares the T_i parameter to a preset or adaptive breath amplitude threshold (step **158**). If T_i is larger than the breath sound amplitude threshold, the segment is identified as a potential breath and control is transferred to block E. If T_i is not larger than the breath sound amplitude threshold, the segment is identified as an apnea segment. The system then checks whether this segment is the first segment in a sequence of apnea segments by comparing the value of the amplitude parameter T_{i-1} of the previous segment to the breath amplitude threshold (step **188**).

If T_{i-1} is larger than the breath sound amplitude threshold, this means that the current segment (segment i) is the first in a new sequence of apnea segments and the system activates the apnea clock (step **200**), registers the previous breath length (step **202**), resets the breath length clock (step **204**) and returns to step **99** for sampling the next segment. If T_{i-1} is not larger than the breath amplitude threshold, his means that the previous segment was also an apnea segment and the system advances the apnea clock (step **190**). The system then checks the value of the apnea duration T_{ap} (step **192**). If T_{ap} is not longer than a preset time threshold (operator determined with a preset default value), the system returns to step **99**. If T_{ap} is longer than the preset time threshold, the system runs a brief hardware verification (step **194**) as described in detail hereinafter. If no hardware malfunction is detected, the system initiates an "apnea alarm" and returns to step **99**. If the system detects a hardware failure, it initiates a "hardware alarm" (step **198**) and returns to step **99**.

Block E preferably receives the T_i value as input and checks its amplitude. In case of excessive amplitude it transfers control to block G for detection of possible snores, cough, or stridor. Otherwise, the block analyzes the segment's spectrum and determines whether the segment is to be identified as a valid tracheal breath sound segment.

Block E preferably receives T_i from step **158** and compares it to a preset or adaptive high-volume threshold (step **160**). If T_i is larger than the high-volume threshold, control is transferred to block **13** for further analyzing of a potential snore, cough, stridor or vocalization.

If T_i is below the high-volume threshold, the system continues to evaluate the segment by calculating the segment's spectral pattern (step **162**).

Step **162** includes a determination of the spectral pattern calculated by fast Fourier transform (FFT) or from the autoregressive (AR) or autoregressive, moving average (ARMA) coefficients of the segment. The system then verifies that the spectral pattern of the segment corresponds to the known pattern of tracheal BS. This is done preferably by fitting the data to equation 1 (step **164**):

$$y = \sum_{i=1}^{n} \frac{m_{1i} \cdot f}{\sqrt{m_{2i} \cdot f^2 + (m_{3i}^2 - f^2)^2}}, n = 3 \pm 1 \tag{1}$$

wherein m_{1i} is a set of amplitude coefficients, f is the frequency, m_{2i} is a set of damping coefficients, m_{3i} is the set of resonance frequencies, i is the serial number of the resonance frequencies and n is the total number of resonance frequencies (n=3±1).

Alternatively, the system can verify that the spectral pattern of the segment corresponds to the known pattern of tracheal BS, by mapping the zeroes and poles of the actual segment relative to the known distribution for tracheal BS.

Reference is now made to FIG. **10** which is an exemplary graph illustrating the results of curve fitting of the tracheal breath sound. Curve **15** represents the spectrum of tracheal BS and curve **17**, superimposed on curve **15**, represents the best fit line **17**, calculated from equation 1 for n=2. The horizontal axis represents the frequency and the vertical axis represents the amplitude. The system calculates goodness-of-fit coefficients—Chi^2 and regression coefficient R (step **166**). The system evaluates the match of the signal to the model of equation 1 (step **168**). If the match is below predetermined acceptability thresholds, for example if Chi/R>Alpha, where Alpha is a preset or adaptive goodness-of-fit threshold, for example Alpha=10, the system determines

that this segment is not a tracheal breath sound segment and diverts control to block G for evaluating the acoustical characteristics of a possible snore, cough, stridor or vocalization. If the match is good, for example when Chi/R is not larger than Alpha, the system identifies the segment as representing tracheal breath sounds and assigns a logical (+) value to the segment parameter S_i (step **170**). The system checks if the current segment is the first in a sequence of breath-representing segments by checking the value of the previous segment parameter S_{i-1}. If S_{i-1} is not a logical (+), the system records the duration of the apnea that just ended, resets the apnea duration clock and activates the breath length counter (step **174**). The system then advances the breath duration clock (step **176**) and transfers control to block F. If the current segment is not the first in a sequence that represents breathing (e.g. the parameter S_{i-1} is not a logical (+)), the system advances the breath duration clock (step **176**) and transfers control to block F.

Blocks B and F preferably determine the respiratory phase and the flow amplitude, respectively. Block B preferably receives the smoothed impedance parameter I_i of the CE sensor (or any qualitative chest expansion or flow detector) from step **128**. The system preferably calculates the first derivative of the smoothed CE signal, for example by calculating the difference between the parameter I_i of the current segment and of the parameter I_{i-1} of the previous segment (step **134**). The difference is calculated as del(I)= $I_i - I_{i-1}$. The system evaluates del(I) for its absolute magnitude $0.5 \, del(I)^{1/2}$ and for its sign (step **138**). If $0.5 del(I)^{1/2}$ is smaller than a threshold value, which was determined during the "training" period, the chest is not moving and the segment is identified as a breath-hold and a breath phase parameter P_i is set to zero (step **140**). If $0.5 del(I)^{1/2}$ is larger than the threshold, the segment represents breathing. If del(I) is positive the system designates the segment as expiratory and sets the value of P_i to +1 (step **144**). If del(I) is negative, the system designates the segment as inspiratory and sets P_i to −1 (step **146**).

Block F receives the values of P_i and T_i as input from blocks B and E, respectively, and calculates the segment flow amplitude FL_i (step **178**) by using the equation $|FL_i|= b \times T_i^c$, where $|FL_i|$ is the absolute value of the flow, b is a calibration factor and c is a calibration exponent determined during the "training" period. Exemplary values of the calibration factor and the calibration exponent are $b=10^{0.5}$ and $c=0.57$. The system then preferably calculates the segment flow amplitude by using the equation $FL_i=|FL_i| \times P_i$ (step **148**). The value of FL_i is recorded for further use (step **150**).

It is noted that the value of the calibration exponent c can be determined by a calibration procedure in which the patient breathes into a suitable quantitative flow measuring device, such as a pneumotachograph or a respirometer.

The system then performs an additional evaluation of the breath length parameter T_{br} (step **180**). If T_{br} is longer than a pre-determined duration of t seconds, for example t equals three times the value of T_{br} determined in the "training" period, the system performs a hardware check (step **182**). If no hardware malfunction is detected, the system generates a breath duration alert (step **184**) and transfers control to step **99**. If a hardware malfunction is detected the system generates a hardware failure alarm (step **186**) and the system transfers control to step **99** for the pick up and evaluation of a new segment.

It is noted that, in accordance with a preferred embodiment of the present invention, the system will use a "training period" to determine values for some of the parameters and indices stated above. In particular, the training period will be

used to run the hardware testing procedure, to determine threshold values for T_i and A_i, and to obtain reference values of breath and apnea durations. In addition, the system also evaluates the values of the tracheal sound resonance frequencies. The tracheal sound resonance frequencies are indicative of the respiratory system's integrity. For example, these frequencies might show a significant change in cases like pneumothorax or blockage of a major airway. The use of the "training" period has the advantage of improving the speed of calculation and the accuracy of detection of the system.

It is noted that the breath detector method is used by the PPG system of the present invention, including the PPG Monitor, Meter and Recorder, but can also be used as a "stand-alone" apnea monitor in clinically relevant situations (sleep apnea diagnosis and follow-up, servo information for nasal CPAP treatment or upper airway electrical stimulation, apnea monitoring of infants and senior citizens, and more). The breath detection method can also be used to detect breathing abnormalities in individuals who are involved in high risk occupations or environments, for example, undersea divers, astronauts, and miners.

The Breath Sound Analyzer Method

Reference is now made to FIG. **11A** which illustrates in block form a breath sound analyzer according to a preferred embodiment of the invention. FIGS. **11B** and **11C** illustrate the blocks of FIG. **11A** in greater detail. Hereinafter, the term FIG. **11A** is used to refer to one or more of FIGS. **11A–11C**. The breath sound analyzer preferably includes blocks H, I J, K and L. Block H preferably receives the signals picked up by chest wall BS sensors (step **252**), amplifies and filters the analog signal (step **254**) and digitizes it, as described hereinabove, to yield a data segment (step **256**).

Block I preferably receives the digitized segment and preferably calculates the magnitude of the spectrum by first calculating the envelope of the time domain signal (step **258**). The system then performs a stationarization of the segment data by normalizing the signal to its envelope (step **260**). The system normalizes the segment data by multiplying it by a window function, for example, a Hanning or Blackmann window function and calculates the amplitude spectrum of the segment (step **262**).

Block J preferably checks whether the data segment is a breath or a background (breath hold) segment by checking the sign of the parameter S_i which it receives from step **170** of block E of the breath detector method of FIG. **9** (step **264**). If S_i is negative, block J diverts control to block L for further processing. If S_i is positive block J diverts control to block K for further processing. If the current segment is identified as a breath segment block K checks the previous segment (step **296**). If the previous segment is not a background segment, this indicates that the current segment is not the beginning of a new breathing phase and the system averages the current BS segment's amplitude spectrum with the previous ones after denormalizing the data for the stationarization factor (step **298**), and transfers control to step **250** for sampling the next segment. If the previous segment is a background segment, this indicates that the current segment is the beginning of a new breathing phase. The system stores the current BS spectrum for averaging (step **300**) and proceeds to evaluate the preceding background (breath hold) phase by calculating the curve fit parameters of the averaged background amplitude spectrum of the preceding background phase segments (step **302**) and checking the goodness of fit to a model equation which is empirically found to be a good representation of the amplitude spectrum of the background sound (step **306**).

A preferred model equation used to calculate the curve fit of the background data is equation 2:

$$y = \frac{A}{1 + (f/f_0)^S} \qquad (2)$$

wherein:

A=Spectrum's amplitude at the plateau

$\int_0$=Spectrum's curve deflection point

S=The spectrum's power. (S should be multiplied by 6.02 to get the slope in dB/oct, for example, slope =3.6×6.02= 21.6 dB/oct).

The system preferably evaluates the fit parameters Chi^2, R and their ratio relative to a threshold. The threshold can be a preset threshold, for example the threshold value can be 10, or can be determined by the system during the "training" period If the goodness of fit is acceptable, the system checks whether the values of A, $\int_0$, and S are within a certain range of values which is empirically determined from average fit parameters of normal background data of patients (step **308**), If the fit parameters are within the normal range, the system records the parameters A, $\int_o$, S and Chi^2/R (step **312**) and transfers control to step **250** for sampling the next segment. Typical values of the parameters for background sounds as obtained with a PPG system using a contact sensor as disclosed by the present inventor in U.S. patent application 08/654,643 filed on May 29, 1996 and entitled "A Contact Sensor for Body Sounds", are: $f_o \approx 120$ Hz and $S \approx 25$ dB/octave. The value of A depends on the gain of the PPG system.

If the goodness of fit is unacceptable or the parameters A, $\int_o$, and S are not within the normal range, the system preferably generates a background problem alert (step **310**) and then records the parameters A, $\int_0$, S and Chi^2/R (step **312**) and transfers control to step **250** for sampling the next segment.

If block J identified the current segment as a background segment, block L checks if the previous segment was a bread or a background segment (step **266**). If the previous segment was a background segment, this indicates that the breath hold phase is not yet ended and the system averages the current background segment's amplitude space with the previous ones after denormalizing the data for the stationarization factor (step **268**). The system transfers control to step **250** for sampling the next segment If the previous segment was a breath segment, this indicates that a new breath hold phase has begun and the system proceeds to analyze the accumulated averaged BS spectra The system initially subtracts the average background spectrum of the n previous segments from the BS spectrum (step **270**). This step is done in the linear frequency domain by arithmetic subtraction. If the result of the subtraction is less than a specific value determined by the specifications of the A/D converter, for example 0.5×(amplitude resolution), the system eliminates the point. If more than a specific number of successive points, for example 3 points, are less than the threshold, the frequency of the first such point is designated $\int_{max}$ and recorded.

The system then preferably performs a defiltration operation to compensate for the effect of the high pass filter on the sound (step **274**). The defiltration is done by dividing each data point by the frequency response of the filter or by the reconstructed response from the best-fit parameters of the filter. The best fit regression equation for this step is equation 3:

$$y = \frac{\text{Gain}}{1 + (\text{"3dBpoint"}/f)^{\text{Rolloff}/6.02}} \qquad (3)$$

where Gain is the amplification factor of the filter, 3 dB point is the deflection point of the filter and Rolloff is the slope of the filter in dB/octave.

The system then preferably analyzes the spectral pattern of the averaged BS spectrum (step **276**). The analysis is done by calculating the best fit of the BS signal to the equation 4:

$$y = \frac{A}{1 + (f/f_0)^S} \qquad (4)$$

wherein:

A=amplitude at the plateau

$\int_0$=deflection point

S=power (S is multiplied by 6.02 to get the slope in dB/oct, for example 3.6×6.02=21.6 dB/oct).

Reference is now made to FIG. **11D** which is a graph demonstrating the results of fitting equation 4 to an exemplary chest breath sound power spectrum. The vertical axis of the graph represents amplitude and the horizontal axis represents frequency. The curve **19** represents the power spectrum of the defiltered data picked up by a chest contact sensor of the PPG system of FIG. **1**. The value of the spectrum's amplitude at the plateau is indicated by the horizontal dashed line, labeled A, and the value of the spectrum's curve deflection point (3 dB point) is indicated by the vertical dashed line, labeled $\int_0$. The curve marked **21** represents the curve fitted to the data using the model equation 4.

The system preferably evaluates the fit parameters Chi^2, R and their ratio Chi^2/R, relative to preset or adaptive thresholds (step **278**). If the goodness of fit is acceptable, for example if (Chi^2/R)<10, the system checks whether the values of A, $\int_0$, and S are within a certain range of values which is empirically determined from average fit parameters of normal breath sound data of patients (step **280**), If the fit parameters are within the normal range, the system preferably records the parameters A, $\int_0$, S and Chi^2/R (step **282**), resets the BS averaged spectrum (step **294**) and transfers control to step **250** for sampling the next segment If the goodness of fit is unacceptable or the parameters A, $\int_0$, or S are not within the normal range, the system preferably performs a hardware check (step **284**). If the hardware is found to malfunction, the system generates a hardware alert (step **288**) and transfers control to step **250**. If no hardware malfunction found, the system preferably records the parameters A, $\int_0$, S and Chi^2/R (step **286**), transfers control to the wheeze/rhonchi detection method (step **290**), preferably generates an abnormal BS alert (step **292**), resets the BS average spectrum (step **294**) and transfers control to step **250** for sampling the next segment.

It is noted that the typical values of the parameters $\int_0$ and S vary for men, women and children and can also vary for different sensor locations. Additionally, the typical values of the parameters $\int_0$ and S can be different for inspiratory and expiratory breath sounds. Exemplary ranges are $\int_0$=175 Hz±25 Hz, with an average value of approximately 175 Hz for adult males, approximately 182 Hz for adult females and 235±35 Hz for children. An exemplary range for the slope is S=18±4.5 dB/Octave. A is a system specific parameter which depends on the gain of the system and on the loudness of the breath sounds.

It is noted that the method of fitting the BS power spectrum data to equation 4 yields better results (better fit)

than the prior art method of fitting the BS power spectrum data to two straight lines as disclosed by the present inventor in the book *Breath Sounds Methodology* by Noam Gavriely, CRC Press Inc, 1995.

It is also noted that, from equation 4 and the values of A, $\int_0$ and S, it is possible to derive other descriptive parameters of BS such as median frequency, quartile frequencies, for example frequencies of 25%, 50% and 75% of power and percentile frequencies, for example 95% of power or 99% of power frequencies, by analytic calculation This is a new method of calculation that is more accurate and specific for basic breath sounds evaluation than the digital integration method currently in use, since it is less sensitive to spurious peaks of power in the spectrum such as may be caused by the presence of wheezes or artifacts. The calculation of the quartile and median frequencies from the best fit equation is also faster if the curve fitting is calculated anyway for other purposes. The Quartile and Median frequencies are an alternative parametric representation of the spectral content of basic breath sounds that is currently in use.

The Wheeze (and squeak) detection method

The first step in the detection of wheezes is the is a screening in the frequency domain in which peaks of power above an underlying spectrum of basic sounds are sought. If any are detected, they are evaluated to determine whether these peaks correspond to true wheezes, for example as to their sharpness and prominence over the underlying spectrum.

Reference is now made to FIG. **12** which illustrates a wheeze detection method in accordance with a preferred embodiment of the present invention.

Wheezes may be multiple ("polyphonic") or single ("mono-wheeze"), have a constant frequency or a varying frequency and may be localized or widely distributed over the chest. The time-domain characteristics of a single wheeze are usually similar to a pure sinusoidal wave. It may, however, contain a small number of harmonics. The frequency domain (Amplitude spectrum) pattern of wheezes is therefore characterized by a single or a few sharp and narrow peaks.

The wheeze detection method preferably receives as input a sound segment (N data points) digitized from the amplified and preferably conditioned signal of a tracheal or a chest BS sensor as described hereinabove (step **352**). The system preferably calculates the total acoustic energy within the segment (step **354**) and preferably evaluates the segment's acoustic energy (step **356**). If the segment's total acoustic energy is less than or equal to the most recent value of the background's acoustic energy, the segment is identified as 'below level' and the system returns control to step **350** for sampling the next segment. This "low-level" designation may be due to breath hold, "silent lung" (a condition with extreme bronchoconstriction where no breath sounds or even wheezes are heard), or due to equipment failure. The system uses the breath detection method of FIG. **9** to deal with these situations as described and illustrated hereinabove. If the segment's total acoustic energy is larger than the background, the system calculates the amplitude spectrum of the segment and preferably subtracts the background noise amplitude spectrum from it (step **358**).

The system then preferably searches for peaks of power in the spectrum. It does so by preferably calculating the pattern of the amplitude spectrum of the underlying basic breath sound by preferably curve fitting, as described hereinabove, or by low pass filtration-f the spectrum, for example by a Hamming or a 10% FIR filter, or by smoothing (step **360**), or by using the average spectrum described hereinabove, and

subtracts it from the segment's amplitude spectrum (step **362**), generating an array of difference values between the spectrum and the underlying basic breath sounds pattern. The system preferably proceeds to search for narrow spectal peaks by preferably calculating the variance of the array of differences (the residuals) (step **364**). Significant spectral peaks are always positive, so that, based on statistical considerations, when no peaks are present, the residuals are randomly distributed around zero and there will generally be no values that are greater than 4 variances of the mean value of the residuals within a segment or greater than 5 variances within an ensemble of segments.

The system then preferably equates all the amplitudes within the segment whose values are less than k•(variance) to k•(variance) (step **366**) and subtracts k•(variance) from all the amplitude values within the segment (step **368**). An exemplary value that can be used for this procedure is k=5. Thus, the only peaks remaining in the segment have positive values whereas the values in between them become zero. The system then searches the spectrum for peaks by comparing the spectrum data to a threshold which has the value of m.(variance), for example, m=0.5. Other threshold values including zero or near zero values can also be used under certain circumstances. The system identifies a spectrum amplitude value in excess of m.(variance) as a peak (step **370**).

Note that other methods of peak detection may be applied to detect spectral peaks characteristic of wheezes, in accordance with other embodiments of the invention.

It is further noted that, as an alternative for comparing the spectrum's data to a threshold in step **370**, the system can calculate the 4th moment of the difference (residual) array which is the outcome of step **362**. If the 4th moment of the difference (residual) array is significantly greater than 3, for example if the 4th moment is 6, the system searches for peaks.

If a peak is detected by any method, its sharpness is evaluated using the absolute value of the first or second derivative of the difference (residual) array as an evaluation criterion (step **372**). If the absolute value of the first or second derivative of the difference (residual) array at the frequency of the peak is larger than a predetermined threshold value, the system registers the peak's frequency $\int_{wz}$ and its spectral amplitude A_{wz} (step **374**) and blanks the peak region by equating to zero a range within the residual array that is centered at $\int_{wz}$ and spans $\pm\Delta\int$ around $\int_{wz}$ (step **376**), for example, $\Delta\int=32$ Hz.

$\Delta\int$ is determined by N which is the number of data points in the segment and by the sampling rate of the analog to digital converter. If the absolute value of the first or second derivative of the residuals is not larger than a predetermined threshold value, the system blanks the peak region (step **376**). The system then continues to search for additional (secondary) peaks within the spectrum by repeating the peak searching steps **364–376** until no more significant peaks are found.

Reference is now made to FIGS. **13A** and **13B** which illustrate a graphic representation of an amplitude spectra calculated by different steps of the wheeze detection method. In FIGS. **13A** and **13B** the vertical axis represents the amplitude and the horizontal axis represents the frequency. FIG. **13A** illustrates two superimposed curves. The first curve, labeled **423**, represents the amplitude spectrum curve of the breath sound segment calculated by step **358** of FIG. **12**. The second curve, labeled **423**, represents the amplitude spectrum of the basic BS pattern calculated by step **360** of FIG. **12**. The three largest peaks of curve **423** are labeled

425. FIG. 13B illustrates the output of step 368 of the wheeze detection method of FIG. 12 after the first pass through steps 364–376. The largest spectral peak which was detected is labeled 429. It is noted that the range around and including the first detected peak 429 is blanked on subsequent passes so that the secondary peaks may be detected (not shown).

Returning to FIG. 12, if no more peaks are found in the segment, the system transfers control to step 378. Step 378 is based upon experimental observations of the duration of actual wheezes and of their trends within a breath. The method evaluates each peak to assure that it is part of a group of peaks that appear at approximately the same frequency in preceding or subsequent segment's amplitude spectra. A peak is accepted as a wheeze only if it appears consecutively in preferably three or more segments that span at least 150 ms but no more than 2500 ms (2.5 seconds). Thus, the system checks if there was any peak in the current segment's spectrum (step 378) using the information recorded by step 374. If there is no peak in the current segment's spectrum, the system transfers control to step 350 for obtaining the next segment. If any peaks were detected in the current segment's spectrum, the method checks the record of the previous n segments for the presence of significant peaks at $\int_{wz} \pm \delta f$ (step 380), where δf may be set as a fraction of the Nyquist frequency (half the sampling rate) or as a constant, for example, $\delta f = 64$ Hz which give 64 Hz per 50 ms or 1.28 Hz per ms. Values of 0.5 to 2 Hz per ms are preferably used for the wheeze frequency rate of change in time. An exemplary value of n is n=2 where the total number of segments checked including the current segment is 3. Thus, for an -exemplary segment duration of 50 ms, the total duration checked by the system for the presence of significant peaks is preferably 150 ms.

Note that shorter wheezes, often called squeaks, may be detected by this method, but since they are shorter than wheezes (they have a length of between 80 and 150 msec), n is preferably equal to one. Squeaks also differ from wheezes in the peaks associated with squeaks may appear in a time period comprising only 2–3 segments of 50 msec. Squeaks are clinically relevant in artificially ventilated patients and represent air "sneaking" out (or in) in an almost collapsed airway or around the cuff of the endotracheal tube at the end of inspiration if the cuff is not sufficiently inflated.

If corresponding peaks appear in previous segments, the peak is registered as potential wheeze (step 382). If no matching peaks are found in the previous segments the registration is temporary (step 384), pending the findings in the next segments.

The final part of the method's preferred wheeze verification procedure is the separation of true wheezes from other signals that generate sharp prominent peaks in the amplitude spectrum. The latter include speech (vocalization), snoring, and rhonchi. All three signals are periodic but non-sinusoidal in the time domain, displaying repetitive, relatively complex, sound structures. These signals are represented in the frequency domain by a series of sharp peaks that are uniformly spaced in the amplitude spectrum. These peaks are distinguished from the peaks of true wheezes by the number of harmonics which is the number of peaks that are spaced exactly Δf Hz apart. While speech, snores and rhonchi always have more than three such peaks, wheezes usually have one peak with an occasional additional harmonic that is substantially attenuated. The separation of non-wheeze "peaked" signals into the subgroups of rhonchi, snores, and speech is disclosed in detail hereinafter. Thus, after the system registered a potential wheeze (step 382), the

system checks whether there are more than two harmonics in the segment (step 386). If there are more than two harmonics in the segment, the segment is not designated as belonging to a potential wheeze, the system activates the rhonchi/snore/vocalization detecting methods (step 388) and returns control to step 350 for obtaining the next segment. If there are no more than two harmonics in the segment, the segment is designated as belonging to a wheeze and the system registers the peaks that correspond in at least n+1 segments as wheezes (step 390), where (n+1) (segment duration)$\geq$150 ms, and transfers control to step 350 for obtaining the next segment.

It is noted that since the rate of sound structures in rhonchi and snores is low, being in the approximate range of 40–120 sound structures per second, it is important to evaluate segments of larger duration than those used for detection of wheezes and other higher frequency sounds. An exemplary suitable segment duration is equal to or larger than 200 ms.

Once the evaluation and verification of a segment is completed, the system moves on to analyze the next sound segment. It is noted tat, in accordance with a preferred embodiment of the present invention, the analysis of a segment is completed before the end of the acquisition of the data of the next consecutive segment, so that quasi real-time monitoring is accomplished. The feature, in particular, enables the continuous on-line monitoring of the results of the breath analysis performed by the preferred embodiments of the present invention,

The Rhonchi Detection Method

Rhonchi are continuous adventitious breath sounds that are heard over the chest and the trachea of patients with a variety of lung diseases. The sounds are different from wheezes in their lack of smooth musical tone which is replaced by a rough, grinding-like quality. Rhonchi may be inspiratory or expiratory. In the time domain, rhonchi consist of non-sinusoidal, periodic waves (repetitive sound structures) and in the frequency domain by multiple peaks of power.

Tentative detection of rhonchi is a secondary process, that is they are suspected based on a search for another sound. They are suspected if continuous adventitious breath sounds with multiple resonance peaks (>3) are detected by the wheeze detection method. To be confirmed the absence of coinciding loud sounds at the trachea and/or the ambient sounds sensor must be verified to exclude snores and vocalization.

Reference is now made to FIG. 14 illustrating a graph of a section from a rhonchus (sampled in a patient with pneumonia and lung cancer). The graph's vertical axis represents amplitude and the horizontal axis represents time. It is noted that the amplitude trace contains a train of multiple repetitive sound structures that are similar in shape and are generally equally spaced as indicated by the horizontal bars 401.

Repetitive sound structures are also characteristic of snores and vocalization (speech), so that rhonchi cannot be distinguished and identified solely based on these features. However, unlike both snores and vocalization, rhonchi are only detectable by contact sensors placed over the chest wall and cannot be picked up by an ambient microphone. In addition, rhonchi are often localized sounds picked up over a certain chest region, but not over other locations. Thus, a particular feature of some aspects of the method of detection of rhonchi of the present invention, is that the positive identification of rhonchi utilizes the signals picked up by the ambient noise microphone, in addition to the signals picked up by the chest-wall BS sensor.

Reference is now made to FIG. **15** which is a flow chart illustrating the steps of a rhonchi detection method in accordance with a preferred embodiment of the present invention.

The rhonchi detection method is designed to identify the presence of rhonchi in breath sounds. The method uses specific characteristics of the sounds to distinguish them from wheezes on the one hand and from snores and vocalization sounds on the other. When rhonchi are detected, the system performs specific actions which may include the activation of an alarm (in the PPG monitor embodiment), the recording of the rhonchi parameters (in the PPG recorder embodiment) or the generation of graphic and parametric information to be printed or displayed (in the PPG meter or monitor embodiments).

The rhonchi detection method is a second-line method. It is invoked only if called by the primary method—the wheeze detection method. The rhonchi detection method is activated by the wheeze detection method if three or more equally-spaced peaks are detected in the amplitude spectrum or by the crackle detection method if closely and equally spaced multiple crackles are detected. The system verifies that the presence of a train of sound structures actually represents a rhonchus by receiving the ambient noise data segment A_i and the breath sound data segment S_i of the tracheal BS sensor or a chest BS sensor (step **400**) and calculating the transfer function between the breath sound data segment B and the sounds data segment A_i picked up simultaneously by the ambient noise microphone (step **404**). The system then calculates the coherence of the transfer function (step **406**). The system compares the frequency ranges of peaks in the coherence data with the frequency ranges of peaks in the amplitude spectrum of S_i (step **408**). If the frequency ranges of high coherence, for example the frequency ranges for which the coherence>0.7, match, within the frequency resolution of the system, for example 32Hz, the frequency ranges of the peaks in the amplitude spectrum of the chest wall sounds, the system rejects the hypothesis that these peaks represent rhonchi and transfers control to the snore and vocalization methods (step **410**). The peak correlation verification procedure becomes optional if the ambient sound level is low and less than a threshold value as shown and described for step **136** of FIG. **9**, since snores and vocalization are usually loud enough to cause violation of the ambient sound threshold

Alternatively, the system performs a cross correlation between the chest wall sound data and the ambient noise data and searches for peaks at intervals that correspond to the reciprocal of the frequency interval between the peaks in both spectra. When such correlation is found, the system rejects the hypothesis that the sounds represent rhonchi and transfers control to the snore/vocalization detection methods (step **410**). If there is no correspondence between the frequency ranges of the coherence peaks and the amplitude spectrum peaks, the system identifies the segment as containing rhonchi and records the parameters of the rhonchus (step **412**). The parameters recorded are the rhonchus duration, estimated from the number of contiguous segments that contain rhonchi and the relative positioning of the rhonchus within the respiratory cycle (expiratory or inspiratory rhonchus). It is noted that, at step **412**, the system uses the breath phase and flow data calculated by the breath detection method of FIG. **9**.

It is noted that if rhonchi are detected and verified, the system activates a "rhonchi alarm" in the Monitor configuration (step **414**), otherwise, the system records or presents the rhonchi parameters (step **412**).

The Snore Detection Method

Snores are continuous adventitious breath sounds that are generated during sleep due to oscillations of certain upper airway structures. These are loud sounds that are detectable by an ambient noise microphone as well as over the chest by a contact sensor.

Snores are tentatively identified when loud tracheal and ambient sounds are detected at the same time, but without a coinciding sudden chest motion. However, this detection has to be validated by positively identifying certain features and excluding others. In particular, the snore detector searches for equally spaced sound structures in the time domain, multiple resonance peaks in the frequency domain, high coherence between the ambient and tracheal sounds and inspiratory chest motion. It should be noted that while snores may also be expiratory, the latter are indistinguishable from certain types of vocalizations. Therefore for absolute positive validation of a sound as a snore it should coincide with inspiratory chest motion.

Reference is now made to FIGS. **16A** and **16B**, illustrating the curves of a "simple" snore in the time and frequency domains, respectively, and to FIGS. **17A** and **17B**, illustrating the curves of a "complex" snore in the time and frequency domains, respectively.

In the time domain, snores are characterized as a periodic wave consisting of simple or complex sound structures. FIG. **16A** and **17A** are graphs illustrating the time domain curves of a simple snore and a complex snore, respectively, where the vertical axis represents the snore amplitude and the horizontal axis represents time. Some of the simple sound structures occurring repetitively in FIG. **16A** are labeled **432** while some of the repetitively occurring complex sound structures of FIG. **17A** are labeled **430**. The periodic sound structures of simple and complex snores have a period A, labeled **433**, in FIG. **16A**, and **431**, in FIG. **17A**.

The snores may be superimposed on random noise, such as that picked up by the tracheal BS sensor, with varying relationships between the snore sound structures and the noise.

In the frequency domain, snores are represented by a series of equally spaced sharp and narrow peaks. The frequency interval between the peaks is equal to the reciprocal of the time interval between the sound structures in the time domain.

FIG. **16B** and **17B** are graphs illustrating the frequency domain (power spectrum) curves of the simple snore of FIG. **16A** and the complex snore of FIG. **17A**, respectively, where the vertical axis represents the amplitude and the horizontal axis represents the frequency. FIG. **16B** illustrates the four largest peaks, labeled **434**, of the power spectrum of the simple snore. FIG. **17B** illustrates the three largest of the multiple peaks, labeled **436**, of the power spectrum of the complex snore. The highest peak is not always the first in the series as seen in FIG. **17B**. The number of the peaks and their distribution is determined by the frequency content of each individual sound structure. The frequency interval of between adjacent peaks in the power spectrum is labeled **438** in FIG. **16B** and **437** in FIG. **17B**.

A snore detecting method in accordance with a preferred embodiment of the invention is designed to identify the presence of snores during monitoring of breath sounds and to distinguish snores from other adventitious or environmental sounds. The method is used by the PPG Monitor and the PPG Recorder, but may also be used as a stand-alone snore detection device, or as part of a sleep disorders evaluation system. The output of the method includes the timing, duration, rate of sound structures per unit time, and the character of the snore (for example simple or complex).

The snore detection method is a second-line method. It is invoked only if called by a primary method such as the wheeze detection method, the crackle detection method, the rhonchus detection method or the breath detector method.

Reference is now made to FIG. **18** which is a flow chart illustrating the steps of a snore detection method in accordance with a preferred embodiment of the invention. Preferably, the snore detection method is based on detection of segments in which uniformly spaced and similar sound structures are present.

It is noted that the method is limited to detection of inspiratory snores since expiratory snores cannot be effectively distinguished from vocalization.

The system obtains a sound segment of duration T (step **442**). It is noted that, in order to function properly, the sound segment duration T should preferably be approximately 200±50 ms. These limits are set in order to facilitate capture of a sufficient number of sound structures even in complex snores which usually have a relatively low rate of sound structure generation and, on the other hand, in order to limit the duration of the segment so that a train of equally spaced sound structures may be isolated. If the duration is too long the rate of sound structures may change and the detection will become less accurate. The duration T may consist of three to five 50 ms segments and the calculation is done in a running fashion with T-50 ms overlap between successive snore detection segments. The system determines whether the segment is an inspiratory segment (step **444**) by checking the parameter P_i which was calculated by block B of FIG. **9**. If the segment is not an inspiratory segment, the system returns control to step **440** for obtaining the next segment.

If the segment is an inspiratory segment, the system preferably removes the DC component of the segment, calculated as an average of the sound signal over several segments, and normalizes the segment's data by its variance (step **446**), multiplies the normalized segment by a window, for example a Blackman window, and calculates the power spectrum of the normalized window multiplied segment (step **448**). The system then checks the calculated power spectrum for the presence of equally spaced peaks (step **451**). If the power spectrum does not contain at least three approximately equally spaced peaks, the system rejects the possibility that the current segment is a snore and returns control to step **440** to obtain the next segment. If the power spectrum contains at least three approximately equally spaced peaks, the system calculates the autocorrelation of the normalized, window multiplied signal (step **458**).

The system can calculate the autocorrelation using values of time delay of 5 to 25 ms, or by calculating the inverse transform of the non-truncated power spectrum. The system checks for peaks in the calculated autocorrelation (step **460**). If a significant peak is found within 5 to 25 ms from the beginning of the autocorrelation, the system searches for one or more subsequent equally spaced peaks (step **462**) and calculates the time interval between them as at (step **464**). In parallel to the checking of the autocorrelation for peaks (steps **458–464**), the system evaluates the power spectrum calculated in step **448** in search of a train of sharp peaks. The system calculates the distribution curve (histogram) of the spectrum (step **450**). The system calculates the width a of the distribution curve (step **452**) from the zeroth, first and second moments μ_0, μ_1, & $\mu2$, respectively, of the distribution curve, using equations 5 and 6.

$$\sigma = \sqrt{\frac{\mu_2}{\mu_0} - \left(\frac{\mu_1}{\mu_0}\right)^2} \tag{5}$$

$$\mu_0 = \frac{1}{n}\sum_{i=1}^{n} x_i; \; \mu_1 = \frac{1}{n}\sum_{i=1}^{n} i \cdot x_i; \; \mu_2 = \frac{1}{n}\sum_{i=1}^{n} i^2 \cdot x_i \tag{6}$$

where n is the number of elements in the distribution and xi are the values of the elements. The system accepts as peaks only peaks that are greater than k-c where k is a constant, for example k-4 (step **454**). The system then calculates the frequency intervals between the accepted peaks δʃ (step **456**). The system verifies the calculated intervals between the peaks by comparing between δt and $(\delta\!\int)^{-1}$ (step **466**). If the values of δt and $(\delta\!\int)^{-1}$ are not within close proximity, the system does not identify the segment as a snore and returns control to step **440** to obtain the next segment. This may happen in a noisy environment with knocking or grinding ambient sounds, as may happen when the bed fellow of the index subject also snores.

If the two values δt and $(\delta\!\int)^{-1}$ are within close proximity, for example when δt-$(\delta\!\int)^{-1}$<1 ms, the system calculates the coherence of the transfer function between the current BS segment and the current ambient noise segment (step **467**). The system then checks the coherence value at the frequencies of the peaks (step **468**). If the coherence at the frequencies of the peaks is not greater than 0.7, the system rejects the possibility that the current segment is a snore and returns control to step **440** to obtain the next segment. If the coherence at the frequencies of the peaks is greater than 0.7, the system accepts the segment as containing a snore, logs the snore parameters (step **469**) and returns control to step **440** for obtaining the next segment. The logged snore parameters are the amplitude and duration of the snore, the snore type (complex or simple) and the respiratory phase. The system distinguishes between a complex and simple snore by using a peak count criterion. If the system detects 3–4 peaks in the power spectrum, the system registers the snore as a simple snore. If the number of peaks detected by the system is equal to or greater than 5, the system registers the snore as a complex snore.

The Cough Detection Method

Cough detection is initially based on the simultaneous detection of a sudden onset of loud noise in a tracheal sound sensor and ambient microphone and the detection of a rapid motion of the patient's chest Following this rapid screening, the algorithm verifies the detection by curve fitting the sound signal envelope to a specific mathematical expression and by verifying that the spectral content of the sound is that of a cough (broad band noise) rather than that of a snore and vocalization.

The cough detection method disclosed hereinbelow as a preferred embodiment of part of the PPG system of FIG. **2** is designed to detect cough sounds in patients and to determine their timing, number and parameters. The cough detection method can be used as part of the PPG Monitor of FIG. **4**, the PPG Recorder of FIG. **5** and the PPG Meter of FIG. **3**. The cough detection method can also be performed by a stand-alone device. The detection and count of coughs is important in the follow-up of asthma and other lung diseases and their treatment.

Reference is now made to FIGS. **19A** and **19B** illustrating the temporal structure of the sound and air flow data of a cough. A cough is an explosive exhalation associated with noise generated from the thorax and upper airways. It is usually preceded by an inspiration and has 1–4 sound

components. Each component is accompanied by a rapid motion and change in chest volume. FIG. 19A is a graph illustrating the flow curve 470 recorded by a pneumotachograph and the sound curve 472 recorded by an ambient noise microphone. The horizontal axis represents time, the left vertical axis represents the sound amplitude of curve 472 and the right vertical axis represents the flow of curve 470. In the flow curve 470, an inspiration 474 precedes the cough which is composed of four components, labeled 476, each of which has an abrupt onset. In the sound curve 472, the cough includes four sound components 478, each having an initial and final louder parts and a less loud middle part FIG. 19B illustrates a graph of a single component of a cough. The axes are similar to the axes of FIG. 19A except that the time axis is expanded relative to the time axis of FIG. 19A. The flow curve 480 illustrates the abrupt onset, labeled 488, of the cough component while the sound curve 482 illustrates the initial louder part, labeled 484, the final louder part, labeled 487, and the less loud middle part, labeled 486. Thus, the envelope of a cough sound can be said to have a characteristic "double hump" shape.

A cough detection method in accordance with a preferred embodiment of he invention uses information from an ambient noise microphone and from one or more contact sensors placed on the subject's body. The system also uses data from a chest expansion sensor or flow detecting sensor, for example an impedance measurement, to verify that the signal actually represent a cough. The ambient noise microphone is placed near the subject. If the noise detection method detects a loud environmental sound, it activates the cough detection method. The cough detection method positively identifies coughs and records their timing, duration, and number of components.

Reference is now made to FIG. 20 which is a flow chart illustrating the methodology and the various steps of a cough detection method in accordance with a preferred embodiment of the invention.

When the cough detection method is activated by one of the primary methods, for example when the A_i parameter of the step 136 of FIG. 9 is larger than threshold as described hereinabove, the system preferably checks whether the current segment is an expiratory segment by checking the value of the parameter P_i (step 552). If P_i is not equal to −1, the system returns control to step 550 for obtaining the next segment If P_i=−1, the system preferably evaluates the rate of change of chest dimensions, or flow rate in comparison with a predetermined value or to a value previously calculated during quiet breathing. If the rate of chest dimension change, or the flow rate is not sufficiently high, the system return control to step 550 for obtaining the next segment. The System then preferably further checks whether the peaks of the ambient sound and the breath sound coincide in the time domain (step 554). If the peaks of the ambient sounds and the breath sounds do not coincide within 50 ms in the time domain, the system transfers control to step 550 for obtaining the next segment. If the peaks of the ambient sounds and the breath sounds coincide within 50 ms in the time domain, the segment is potentially a cough. To verify that, the system preferably calculates the transfer function and coherence of the breath sounds and the ambient sounds (step 556). The system then checks the coherence value of the transfer function (step 558). If the coherence is not larger than 0.7, the system transfers control to step 550 for obtaining the next segment. If the coherence is greater than 0.7, the system calculates the envelope of the potential cough sound (step 560). The system then checks the duration of the potential cough sound envelope (step 562). If the duration of the

potential cough sound envelope is not greater than 0.2 seconds and not smaller Man 3.0 seconds, the system transfers control to step 550 for obtaining the next segment. If the duration of the potential cough sound envelope is greater than 0.2 seconds and smaller than 3.0 seconds, the system identifies the initial and final parts of the potential cough envelope (step 564), curve fits each of the initial and final parts of the potential cough envelope to a Gaussian (step 566) and calculates the mean amplitude of the middle part of the potential cough envelope (step 568). It is noted that the initial, middle and final parts of the calculated envelope of a cough (not shown) approximately coincide in the time domain with the initial, middle and final parts, respectively, of the cough sounds from which the envelope was calculated. The initial, middle and final parts are best seen in the exemplary cough sound of FIG. 19B and are labeled 484, 486 and 487, respectively. The envelope calculation of step 560 can be performed, for example, by rectifying and suitably filtering the breath sound data.

The system then checks the amplitudes of the initial middle and final parts of the calculated envelope of the potential cough (step 570). If the amplitude of the initial and the final parts of the envelope are not greater than k times the middle part of the envelope of the potential cough sound, where k is a parameter with a value of, for example, 1.5, the system transfers control to step 550 to obtain the next segment. If the amplitude of the initial and the final parts of the envelope are greater than k times the middle part the of the envelope of the potential cough sound, the system registers the segment as a cough segment, records the cough parameters (step 572) and transfers control to step 550 to obtain the next segment It is noted that the recorded cough parameters are the cough duration and amplitude.

Note that weak coughs (grunts) may be detected where the cough has only the initial component. Such coughs are detected by the combination of ambient noise level violation, tracheal noise level violation, expiratory activity, and rate of change of chest dimension, but are not verified as having the double hump appearance. Such weak coughs are registered but not verified.

The Crackle Detection Method

A crackle is a non-stationary event with a specific waveform. The characteristics of the waveform reflect the mechanism of crackle generation as an abrupt opening of an airway that was previously closed due to a collapse or a liquid barrier. The waveform also reflects the attenuation and refraction of the original sound as it is transmitted though the lung and chest wall. With this in mind, a preferred embodiment of a method of detecting a crackle is based on a rapid search for non-stationary events, followed by a verification process that is designed to eliminate artifacts i.e., other non-stationary events that do not fit a mathematical description of a true crackle and to obtain the values of parameters of the mathematical description to sort the crackles into subgroups.

Crackles are discontinuous adventitious breath sounds with an abrupt onset and a short duration. They are heard over the chest wall of patients with cardiopulmonary diseases. The presence of crackles usually represents an abnormality. Reference is now made to FIG. 21 illustrating examples of typical crackles recorded from a patient with pulmonary fibrosis. In FIG. 21 the vertical axis represents the amplitude of the sound and the horizontal axis represents time. The sound curve contains two "fine" crackles labeled 492, one "coarse" crackle labeled 494 and an artifactual sound labeled 496 having an abrupt onset and a very short duration.

The crackle detection method described herein as part of a preferred embodiment of the PPG system is designed to identify the presence of crackles in breath sounds, determine their timing within the respiratory cycle, count their number, and characterize their waveform. The crackle detection method is used by the PPG monitor, the PPG meter and the PPG recorder, and can also be used as a "stand-alone" module or in a "stand-alone" device. The crackle detection method can be used during examination of the chest at shifting body postures to detect gravity-dependent migration of crackles. It can be used in the monitoring of patients during general anesthesia to detect fluid overload from IV fluids or from irrigation fluid during trans-urethral resection of prostate (TURP) procedure. The crackle detection method can also be used to monitor patients after myocardial infarction in the coronary care unit. Such patients often develop congestive heart failure.

The crackle detection method can also be used to monitor patients at home who complain of orthopnea or paroxysmal nocturnal dyspnoea (PND). The crackle detection method can further be used to monitor accumulation of secretions in the airways of intubated patients in the intensive care unit (ICU).

It is noted that the crackle detection method can be applied for monitoring and analyzing the breath sounds of multiple sensors positioned at different sites of the patients body simultaneously, or of a single sensor positioned at a single site.

Reference is now made to FIG. 22 which is a schematic block diagram illustrating in detail the steps of a crackle detection method in accordance with a preferred embodiment of the invention. The input signals for the method are digitized breath sounds from one or more sensors placed over the chest wall. Amplification, filtration and digitization of the analog signals are preferably performed as described in detail hereinabove. The data is processed on-line or off-line to detect crackles. Selected segments may be archived for documentation or post-processing.

The system obtains a segment Y of N data points, for example N=1000 (step **500**). The system first calculates Z which is a measure of an abrupt change in the signal. Z is defined by one of a family of equations 7:

$$Z = \left(\frac{d^m Y}{dt^m}\right)^n \left(\frac{d^k Y}{dt^k}\right)^l \tag{7}$$

where m, n, l, and k are parameters. Each of the parameters m, n, l, and k can have the values 0, 1 and 2.

The discrete form of the family of equations 7 is represented by the family of equations 8:

$$Z_{i+1} = \frac{1}{\Delta t^{(m \cdot n + k \cdot l)}} \cdot (Y_{i+m} - Y_i)^n \cdot (Y_{i+k} - Y_i)^l \tag{8}$$

where Y_i are the data points, Δt is the sampling interval, Z_{i+1} are the derived values for the i+1 points in the segment and m, n, l, and k are the parameters of the family of equations 7 described hereinabove.

The decision as to which values of the parameters m, n, l, and k are to be used for calculating Z_i of equation 8 depends on the balance between the desired accuracy, which represents the sensitivity and specificity of detection, and the desired speed of calculation. Increasing m, n, l and k increases the sensitivity of detection but may overload the verification step with many events that are not necessarily

crackles. In a preferred embodiment of the invention, the system calculates Z_i which is the second power of the second derivative of the signal for each point i in the segment (step **502**), by using equation 9:

$$Z_i = \frac{1}{\Delta t^4}(Y_{i+1} - Y_{i-1})^2 \quad i = 2 \to N - 1 \tag{9}$$

where m=2, n=2, k=0 and l=0.

In order to detect the timing of crackle events, the system then preferably calculates ($\sigma(Z)$, which is the variance of Z, using equation 10:

$$\sigma(Z) = \frac{1}{N-2} \cdot \left[\sum_{i=2}^{N-1} (Z_i - \bar{Z})^2\right]^{1/2} \tag{10}$$

where $\bar{Z}$ is the mean value of Z within the segment (step **504**).

The system applies a "folding" procedure to Z (step **504**) as follows: $\sigma(Z)$ which is the variance of Z is subtracted from each value of Z. The system calculates a new series of values Z using equation 11:

$$Z^* = |Z - \sigma(Z)| \tag{11}$$

This "folding" procedure is subsequently repeated m times on the resulting series of the previous step, where m varies between 2 to 4, for gradually diminishing the smaller fluctuations near the time axis while enhancing the peaks (if peaks are present in the segment). The final results of the repeated "folding" procedure is referred to hereinafter as W.

Reference is now made to FIGS. 23 and 24 illustrating the processed data resulting from some of the different steps of data processing performed by the crackle detection method of FIG. 22. FIG. 23 illustrates the "raw data" of a segment of breath sounds that is obtained by the system in step **500**. The vertical axis represents the amplitude of the digitized sound data and the horizontal axis represents time. The sound data of FIG. 23 contains three crackles labeled **536**, **537** and **538**. FIG. 24 illustrates a graph of the values of the second power of the second derivative (Z) calculated by step **502** of the crackle detection method. The vertical axis represents Z and the horizontal axis represents time. The data of FIG. 24 includes three peaks, labeled **539**, **540** and **541** which correspond to the crackles **536**, **537** and **538** of FIG. 23, respectively. It is noted that, although the amplitude of the second crackle **537** in the raw data of FIG. 23 has the highest amplitude, the second, corresponding, peak **540** of FIG. 24, does not have the highest amplitude because the second crackle **537** is a coarse crackle and has a lower frequency content.

Going back to FIG. 22, the system calculates $\sigma(W)$ representing the variance of the folded data (step **510**) which is used as a basis for finding peaks. The system then finds the maximal value W_{max} of the segment (step **512**). The system checks whether the value of W_{max} is greater than $k \cdot \sigma(W)$ where k is a constant, for example k3.8 (step **514**).

Reference is now made to FIG. 25 illustrating a graph representing the values W of the folded second power of the second derivative which were calculated by step **508** of FIG. 22 from the Z values of FIG. 24. The vertical axis represents the calculated values of the parameter W and the horizontal axis represents time. The graph includes three peaks, labeled **542**, **543** and **544**, corresponding to the three crackles **536**, **537** and **538**, respectively, of FIG. 23. The dashed line **548**

represents a threshold of 4 times the variance used for screening the peaks in step **514** of FIG. **22**.

Going back to FIG. **22**, if W_{max} exceeds $k \cdot \sigma(W)$, the peak is registered as potentially representing a crackle and the position of the peak T_{cx} is recorded (step **516**). The system proceeds through steps **518–528** for evaluating the shape of the "potential" crackle by template matching.

The aim of this part of the method is to distinguish crackles from other signals that have abrupt onset, for example squawks, wheezes, snores and artifacts, but do not conform to the waveform of crackles as hereinafter defined. In addition, the parameters of each potential crackle are determined, compared to a range of acceptable values obtained from an empirically determined data base of crackling breath sounds, and recorded for diagnostic or other uses.

The system first finds the exact point of onset of the crackle (step **518**). This is done by searching for the first value of W that is greater than $\mu \cdot k \cdot \sigma(W)$, within the t milliseconds preceding the point of maximal value W_{max} used as the index point for this crackle. μ is a predetermined onset factor, for example μ=0.7, and t is a predetermined time interval for example, in accordance with a preferred embodiment of the present invention, t=5 ms.

The system removes the DC level from the sound segment and trends that may exist in the crackle containing portion of the signal segment (step **520**). This is preferably done by subtracting the mean value of amplitudes within the segment from all its elements, by determining the linear regression of the segment and subtracting a value that is equal to the regression line amplitude from all the elements of the segment. The system subjects the "cleaned up" signal segment to a curve fitting routine (step **522**) which fits a curve, generally defined by equation 12 hereinbelow, to the crackle in the segment

$$Y = A \cdot B(t) \cdot C(t)^* \tag{12}$$

wherein the first element A of equation 12, defines the crackle amplitude, the second element B(t) of the equation defines its envelope characteristics, and the third element C(t) of the equation defines the oscillatory component of the crackle.

Preferably, the system performs the curve fit by using a form of the general equation 12 which is defined by equation 13:

$$Y = A \cdot \frac{t^n}{1 + (m \cdot t)^k} \cdot \sin\left(\frac{2\pi f_0 t}{1 + C \cdot t}\right) \tag{13}$$

where A is an amplitude parameter, t is the time in seconds within the segment (starting with the crackle onset), f_0 is the initial crackle frequency in Hz, C determines the rate of increase or decrease of the crackle's internal frequency. Larger values of C cause faster diminution of frequency if C is positive and faster increase of frequency if C is negative. When C=0, the frequency remains constant at f_0 and n determines the acuteness at which the crackle amplitude rises (small values of n are associated with faster rise, since t<<1). k determines the rate of crackle decay where increasing k speeds up the rate of crackle envelope amplitude reduction beyond its peak. m influences the position of the peak of the crackle envelope (smaller values of m move the peak to later tines). It is noted that k and m interact and have mutual effects on the rate of crackle decay.

The system determines the goodness of fit of the potential crackle by checking the calculated values of Chi^2 and R of the fit (step **524**). If Chi^2 and R are not adequate, for example

if R<0.85, Chi^2>20,000, the system blanks the Z values within the range of values $T_{cx} \pm \Delta t$ of the currently recorded maximal peak by equating them to zero (step **526**) and transfers control to step **504**. The system then repeats the peak searching steps for locating the subsequent maximal peak of Z thereby searching for additional crackles within the segment.

This process is repeated until there are no more maxima that are greater than $k \cdot \sigma(W)$, or until more than a certain predetermined portion of the segment, for example half of the segment, has been blanked.

If the goodness of fit is adequate, for example if R>0.85, Chi^2<20,000, the system checks whether the calculated parameters k, A, m, n, C and f_0 are within the empirically determined range for crackles individually, and performs a scoring procedure (step **528**).

The scoring procedure calculates a score for example by using equation 14:

$$SCORE = \frac{1}{n} \sum_{i=1}^{n} \left| \frac{P_i - \overline{P_i}}{\overline{P_i}} \right| \cdot D_i \tag{14}$$

where P_i designates the following parameters P_1=k, P_2=s, P_3=n, P_4=c and P_5=f_0, D_i designates the corresponding weighing factors representing the relative importance of each of the parameters P_i as determined empirically and $\overline{P_i}$ are the corresponding empirically determined values of the parameters of respiratory crackles.

If the values of the parameters P_i or of the score are not within the empirically determined crackle range, the system marks the current potential crackle as an artifact (step **530**), records the artifact's parameters (step **532**) and transfers control to step **526**.

If the values of the parameters P_i or of the score are within the empirically determined crackle range, tlhe system marks the potential crackle as a verified crackle, records the crackle's parameters (step **532**), blanks the range of values around the current peak as described hereinabove (step **526**) and transfers control to step **504** for detecting additional potential crackles. The values of the parameters are then used to distinguish between fine and coarse crackles. In particular, the value of f_0 is used, where crackles having values of f_0 that are greater than a specific threshold, for example crackles for which f_0>500 Hz, are designated as fine crackles, while crackles having values of f_0 that are smaller than the specific threshold, for example: crackles for which f_0<500 Hz, are designated as coarse crackles.

The use of the breath sounds template parameters to sub-classify the sound types

General

While the goodness of fit to the breath sounds to their templates are used in this various aspects of the present invention to validate the detection of the various sound types, the parameter values of the matched templates are used in order to subdivide the sound types into subclasses. These subclasses have diagnostic and clinical significance as described below.

Separation of 'normal' and 'bronchial' chest wall breath sounds

Basic chest wall breath sounds fit a low pass filter that has three parameters, the amplitude, the corner ('3 dB') frequency, and the slope. The normal values of these parameters are specific for inspiration and expiration. The amplitude parameter is hardware system specific, but within one system the inspiratory value is 2.4 fold larger than the expiratory one. The normal corner frequency is the same for

inspiration and expiration, but is higher in, children than in adults. The slope parameter is also generally the same for inspiration and expiration. Each of these parameters have independent or combined clinical significance. For example, high corner frequency is characteristic of what is called in stethoscopic auscultation 'bronchial' breath sounds frequently seen in consolidation of the lungs due to lobar pneumonia. On the other hand, low corner frequency values are found in patients with severe pulmonary emphysema with or without reduction of the values of the amplitude parameter. Smokers who do not have any other respiratory symptoms have changes in their basic breath sounds that are seen as higher value of the slope parameter.

Separation of 'fine', and 'coarse' crackles

Crackle timing within the respiratory cycle has previously been identified as important clinical diagnostic information It is also well known that crackle acoustic character is different in different disease states. At the least, crackles are clinically divided into 'fine' and 'coarse. Crackles have been classified using waveform analysis (See for example U.S. Pat. Nos. 3,990,435 and 5,165,417 to Murphy). The template used for crackle detection validation is also useful in separating them into their subclasses. The parameters that are important in this classification are $\int_0$—the initial crackle frequency in Hz (higher values (e.g. above 600 Hz) are associated with fine crackles), C—the rate of increase or decrease of the crackle's internal frequency (higher positive values (e.g. above 100) are associated with coarse crackles), n—the acuteness of crackle amplitude rise towards its peak (faster rise (e.g. n above 1) is associated with fine crackles), k—the rate of crackle decay and m—influences the position of the peak of the crackle envelope (slower decay (k smaller than 4) and delayed envelope peak (e.g. m less than 400) are associated with coarse crackles). The range between fine and coarse is a continuum, not a distinct division.

The Wheeze Rate parameter and separating wheezes into 'low', 'high' and 'ultra high' frequency subgroups

Wheeze rate, defined as the duration of wheezing as percent of breathing time and its subdivisions into inspiratory and expiratory wheeze rates defined as the duration of inspiratory and expiratory wheezes as percent of total inspiratory and expiratory breathing times, respectively has been previously shown to correlate well with the severity of airway obstruction in asthmatic patients. The detection and validation of wheeze-containing segment is used to determine these parameters in the present invention. Another subdivision that is based on duration is the separation between squeaks (80–150 msec) and wheezes (above 150 msec). Furthermore, the frequencies of the wheezes determined by the wheeze method and are used to classify the wheezes into three frequency ranges: low frequency (less than 400 Hz), high frequency (400–1600 Hz) and ultra high frequency (above 1600 Hz). The ultra high wheezes are specific to the tracheal site.

Separation of 'dry', 'productive' and 'barking (singe hump)' coughs

Coughs are a clinically important sign that frequently bring patients to doctors but is rarely evaluated in quantified way. The new method of cough detection not only permit cough counting and distribution analysis, but separation into three subgroups based on the parameters of the template matching. In particular, the distinction between dry and productive coughs is based no the shape of the second hump of the template. Using the double Gaussian template, if the second (later) Gaussian is broad (variance greater than 40 msec) and skewed to the right (i.e. contains a long (>150 msec) tail), the cough is generally more 'productive' in

character while coughs that have a narrow second hump (variance lesser than 40 msec) and have no tail are 'dry' in character. A third category is the 'barking' type cough. The template matching parameters reveal this type of cough by showing a short signal (200–350 msec) with small second hump that is often only 10–25% above the inter-hump plateau.

Separation of snores into 'simple' and 'complex' types

Snores are due to oscillations of the soft components of the upper airways. The two types of oscillations—those that are centered around a certain airway diameter without actual closure of the channel and are called 'flutter', and those that are actually associated with repeated closure (collapse) and reopening of the upper airways and are called 'flapping flutter'. The acoustic properties of each type are different and their medical significance is thought to be different too. The snore acoustic template as disclosed in the present invention is used to differentiate the two types. The flutter type is associated with 'simple' snores that have 3–4 resonance peaks in the spectrum and the 'complex' snores that have 5 or more resonance peaks in their spectrum and are caused by the flapping flutter type mechanism and complete momentary airway closure. The use of spectral resonance peak count to subdivide snores is another novel feature of the present invention.

The Hardware Verification Method

It will be appreciated that the contact sensors **4, 6, 64** and **66** of the present invention can move during the course of use, particularly when the use is extended, such as in the case of the monitor or recorder. Shifts of position or contact with the body may cause a substantial change in the amplitude or frequency response of the breath sounds. The present invention includes a hardware verification system and method which determines that all of the sensors are operating properly. The hardware verification can be performed at any time and is useful for verifying that a possible alarm is justified and was not generated by a problem with the equipment.

In accordance with a preferred embodiment of the present invention, the piezoelectric contact sensors **4** and **6** are individually utilized, during hardware verification, as sound emitters through switching of the input and output of the contact sensors and providing thereto a voltage waveform. In response, the sound emitting "sensor" produces a sound which is transmitted through the body tissues and is detected by the remaining sensors. Features of the emitted sound are relatively constant as long as the system is stable. However, if the position of either the emitting sensor or one or more of the receiving sensors has changed, the received waveforms will be altered from baseline waveforms produced during a setup period. It will be appreciated that the emitting sensor can produce multiple sounds to increase the signal-to-noise ratio of the received waveforms and that all of the sensors can sequentially act as an emitting sensor for the other sensors.

The received waveforms are compared to the baseline waveforms using any suitable comparison method. For example, a cross-correlation can be performed between each received waveform and its corresponding baseline waveform, using the timing of the emitted sound as a trigger. Other methods can be utilized to determine the degree of similarity between the old and new signals. For example, each new waveform can be subtracted from its corresponding baseline waveform, followed by integration of the squares of the residuals, etc. If the difference between the old and new waveforms is greater than a predetermined threshold, there is a malfunction and the system generates an alert.

It will be appreciated by persons skilled in the art that the present invention is not limited to what has been particularly shown and described hereinabove. Furthermore it should be understood that the various methods of determining various breathing characteristics may be utilized separately or together and that the various features of the preferred embodiments of the invention may be utilized separately or combined in other ways than those disclosed in the above description of the preferred embodiment of the invention. Rather the scope of the present invention is defined only by the claims which follow:

What is claimed is:

1. A method of analyzing breath sounds produced by a respiratory system, the method comprising:

measuring breath sounds produced by the respiratory system;

tentatively identifying a signal as being caused by a breath sound of a given type if it meets a first criteria that tentatively identifies only the breath sound of the given type; and

confirming said identification if a tentatively identified signal meets a second criteria characteristic of the breath sound of the given type.

2. A method according to claim 1 comprising segmenting said breath sound data into segments and wherein tentatively identifying and confirming are based on time segments of breath sound data.

3. A method according to claim 1 wherein confirming said identification comprises comparing breath sounds tentatively identified as being of a given type to at least one of a plurality of second criteria each of which is characteristic of the tentatively identified given type and confirming that the breath sound is of the tentatively identified given type if it meets the second criteria characteristic of the given type.

4. A method according to claim 1 wherein tentatively identifying comprises comparing the breath sound to a plurality of first criteria, each said criteria being tentatively identify a breath sound as being of a given type, wherein said breath sound is tentatively identified as being only of the type for which it meets the first criteria.

5. A method according to claim 1 or claim 4 wherein the breath sound comprises a wheeze.

6. A method according to claim 5 wherein tentatively identifying the breath sound as a wheeze comprises detecting narrow peaks within the spectrum of the breath sound and determining if the narrow peaks are located within a narrow frequency range over a number of consecutive time periods.

7. A method according to claim 6 wherein confirming comprises determining if the narrow peaks of said tentatively identified wheeze have less than three harmonics each.

8. A method according to claim 6 wherein said consecutive time periods span at least 150 ms.

9. A method according to claim 6 wherein a breath sound is confirmed as a squeak type wheeze when said consecutive time periods span between 80 and 150 ms.

10. A method according to claim 6 wherein a breath sound is classified as low frequency wheeze when the frequency of the narrow peak is less than 400 Hz.

11. A method according to claim 6 wherein a breath sound is classified as a high frequency wheeze when the frequency of the narrow peak is between 400 Hz and 1600 Hz.

12. A method according to claim 6 wherein a breath sound is classified as an ultra-high frequency wheeze if the frequency of the narrow beak is above 1600 Hz.

13. A method according to claim 5 wherein said narrow frequency range is not greater then 64 Hz among any two consecutive time periods.

14. A method of according to claim 1 or claim 4 wherein the given breath type comprises a rhonchus.

15. A method according to claim 14 wherein tentatively identifying comprises detecting narrow peaks within a spectrum of said segment and determining if said narrow peaks are located within a predetermined small frequency range across during consecutive time periods.

16. A method according to claim 15 wherein confirming the identification of a breath sound as a rhonchus comprises, if there are more than two harmonics in each of the consecutive time periods:

generating a transfer function in the frequency domain between said breath sound data and measured ambient sound in the space surrounding the patient;

determining a coherence graph of said transfer function;

confirming each narrow peak as a rhonchus if the frequency range of high coherence of said coherence graph does not correspond to the frequency range of said narrow peaks.

17. A method according to claim 1 or claim 4 wherein the breath sound comprises a cough.

18. A method according to claim 17 wherein tentatively identifying comprises:

coincidentally detecting sudden loud ambient noise, a sudden loud breath sound, and a sudden chest motion.

19. A method according to claim 18 wherein confirming that a breath sound is a cough comprises:

generating an envelope of said breath sound data and determining the duration of said envelope;

determining that the sound is a cough if it fulfills all of the following conditions:

a) the breath sound takes place during expiration;

b) the breath sound peaks generally coincide with the ambient noise peaks;

c) the envelope of the breath sound data has a double hump shape;

d) the duration of the envelope is within a predetermined time range; and

e) determining that the frequency spectra of the sound are broad band with at least a predetermined high coherence level between the ambient and the breath sounds.

20. A method according to claim 19 and wherein said predetermined time range is 0.2–3.0 seconds.

21. A method according to claim 19 or claim 20 wherein said predetermined coherence level is 0.7 or greater.

22. A method according to claim 19 wherein the cough is classified as a productive cough if the second hump has a variance greater than 40 msec and is skewed to later times.

23. A method according to claim 19 wherein the cough is classified as a dry cough if it has a variance of less than 40 msec and is not substantially skewed.

24. A method according to claim 19 wherein the cough is classified as a barking cough if it is has a duration of between 200 and 350 msec and has a second hump of 10%–25% above the value between the humps.

25. A method according to claim 1 or claim 4 wherein the breath sound comprises a crackle.

26. A method according to claim 25 wherein tentatively identifying a breath sound as a crackle comprises:

finding the locations of abrupt changes in said breath sound data.

27. A method according to claim 26 wherein confirming that a breath sound is a crackle comprises:

matching breath sound data following the said abrupt changes to the following curve:

$$y = A * B(t) * C(t)$$

where y is said breath sound data starting at the abrupt change at which t begins, A is an amplitude parameter, B(t) is an envelope function and C(t) is an oscillatory function.

28. A method according to claim **27** wherein:

$$B(t) = \frac{t^n}{1 + (mt)^k}$$

and

$$C(t) = \sin\left(\frac{2\pi f_0 t}{1 + Ct}\right)$$

where $\int_0$, C, n, m and k are parameters.

29. A method according to claim **28** including identifying the type of crackle from the values of said parameters.

30. A method according to claim **29** wherein the cackle is identified as a fine cackle if $\int_0$ is above 600 Hz.

31. A method according to claim **29** wherein the cackle is defined as a coarse cackle if the if C, the rate of change of the cackles internal frequency is greater than 100.

32. A method according to claim **26** and including identifying the portion of the breathing cycle during which the crackle occurs.

33. A method according to claim **1** or claim **4** wherein the breath sound comprises a snore.

34. A method according to claim **33** wherein tentatively identifying a breath sound as a snore comprises:

determining that the breath sound occurs during an inspiratory period; and

determining that the breath sound is highly correlated with ambient noise in the region surrounding the patient.

35. A method according to claim **34** wherein confirming that the tentatively identified breath sound is a snore comprising:

identifying peaks in the inspiratory period and determining the average peak-to-peak time delta_t;

identifying at least three peaks in the power spectrum of the inspiratory period which are significantly large and determining the average peak-to-peak frequency delta_f;

generating a coherence graph for a transfer function in the frequency domain between said breath sound data and measured ambient noise of the space where said breath sound data was gathered;

identifying a snore if delta_t is close to the inverse of delta_f and if the frequency range of high coherence of said coherence graph corresponds to the frequency range of said narrow peaks.

36. A method according to claim **35** ad wherein identifying peaks in the inspiratory period comprises:

calculating the inverse Fourier transform of said power spectrum of the inspiratory period said generating thereby cleaned data;

searching for a first peak in said cleaned data beginning between 5 to 25 ms from the start of said segment; and

searching for peaks following said first peak.

37. A method according to claim **35** wherein identifying three peaks in the power spectrum comprises:

calculating the histogram of the power spectrum, determining the variance of the histogram; and

defining as peaks those points having values which are at k variance or higher, where k is at least three.

38. A method according to claim **35** wherein the snore is classified as a simple snore if the spectrum contains 3 or 4 resonance peaks.

39. A method according to claim **35** wherein the snore is classified as a complex snore if the spectrum contains 5 or more resonance peaks.

40. A method according to claim **1** or claim **4** wherein the breath sound comprises a chest wall breath sound.

41. A method according to claim **40** wherein the breath sound is identified as a chest wall breath sound if it fits a low pass filter shaped curve having the parameters of amplitude, corner frequency and slope.

42. A method according to claim **41** wherein the breath sound is confirmed as a bronchial chest wall breath sound if the corner frequency is higher than a normal frequency for the age of the patient.

43. A Phonopneumograph for analyzing breath sounds produced by a respiratory system, the phonopneumograph comprising:

at least one auditory breath sensor which receives sound related to breath sounds and produces signals in response to the received sounds;

at least one first sound signal analyzer which analyzes said signals to initially screen said breath sounds said initial screening being operative to produce a tentative identification of the signals as being caused by a breath sound of a given type if it meets a first criteria; and

at least one second sound signal analyzer which further analyzes those of said signals which have been tentatively identified to confirm if the signals are caused by a breath sound of the given type.

44. A Phonopneumograph according to claim **43** wherein the at least one first sound analyzer comprises a plurality of first sound analyzers, each of which is operative to initially screen said breath sounds, each of said first sound analyzers producing a tentative identification of a breath sound of a different type.

45. A phonopneumograph according to claim **43** wherein the at least one breath sensor comprises a plurality of breath sensors placed at different positions with respect to the respiratory system.

46. A phonopneumograph according to claim **43** or claim **44** wherein the at least one second sound analyzer comprises a plurality of second sound analyzers, each of which is operative to further screen breath sounds of the tentatively identified given type and to produce a confirmation of the tentative identification if the breath sound additional criteria characteristic of the second type.

47. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a crackle.

48. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a cough.

49. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a snore.

50. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a rhonchus.

51. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a wheeze.

52. A phonopneumograph according to claim **43** or claim **44** wherein the breath sound of a given type comprises a breath.

53. A phonopneumograph according to claim **43** or claim **44** wherein the first sound analyzer divides the sound into time segments and analyzes the sound on a segment by segment basis.

54. A method of analysis of breath sound data comprising:

determining the presence of a first breath sound in the breath sound data;

adjusting the breath sound data by reducing the effect of the first breath sound in the breath sound data; and

determining the presence of a second breath sound in the adjusted breath sound data.

55. A method according to claim **54** wherein the first breath sound is a wheeze.

56. A method according to claim **54** or claim **55** wherein the second breath sound is a wheeze.

57. A method according to claim **54** wherein the first breath sound is a crackle.

58. A method according to claim **54** or **57** wherein the second breath sound is a crackle.

59. A method according to claim **54** wherein the first breath sound and the second breath sound are of the same type.

60. A method according to claim **54** wherein the first breath sound and the second breath sound are adventitious breath sounds.

61. A method according to claim **54** wherein reducing the effect of the first breath sound comprises substantially removing the effect of the first breath sound.

62. A method according to claim **54** wherein determining the presence of a breath sound comprises segmenting the breath sound data segments and making the determination based on time segments of breath sound data.

63. A method according to any of claim **54, 55, 61, 59** or **60** wherein determining comprises:

tentatively identifying a signal as being caused by a breath sound of a given type if it meets a first criteria that tentatively identifies only the breath sound of a given type; and

confirming said identification if a tentatively identified signal meets a second criteria characteristic of the breath sounds of the given type.

64. A method according to claim **63** wherein confirming said identification comprises comparing breath sounds tentatively identified as being of a given type to at least one of a plurality of second criteria each of which is characteristic of the tentatively identified given type and confirming that the breath sound is of the tentatively identified given type if it meets the second criteria characteristic of the given type.

65. A method according to claim **63** wherein tentatively identifying comprises comparing the breath sound to a plurality of fir criteria, each said criteria being used to tentatively identity a breath sound as being of a given type, wherein said breath sound is tentatively identified as being only of the type for which it meets the first criteria.

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