US 20120071767A1

(19) United States

(12) Patent Application Publication (10) Pub. No.: US 2012/0071767 A1

Popov et al. (43) Pub. Date: Mar. 22, 2012

(54) PULMONARY ARTERY PRESSURE ESTIMATOR

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(21) Appl. No.: **13/053,202**

(22) Filed: **Mar. 21, 2011**

Related U.S. Application Data

(63) Continuation of application No. 11/578,462, filed on Aug. 13, 2007, now Pat. No. 7,909,772, filed as application No. PCT/CA2005/000568 on Apr. 15, 2005.

(60) Provisional application No. 60/562,538, filed on Apr. 16, 2004.

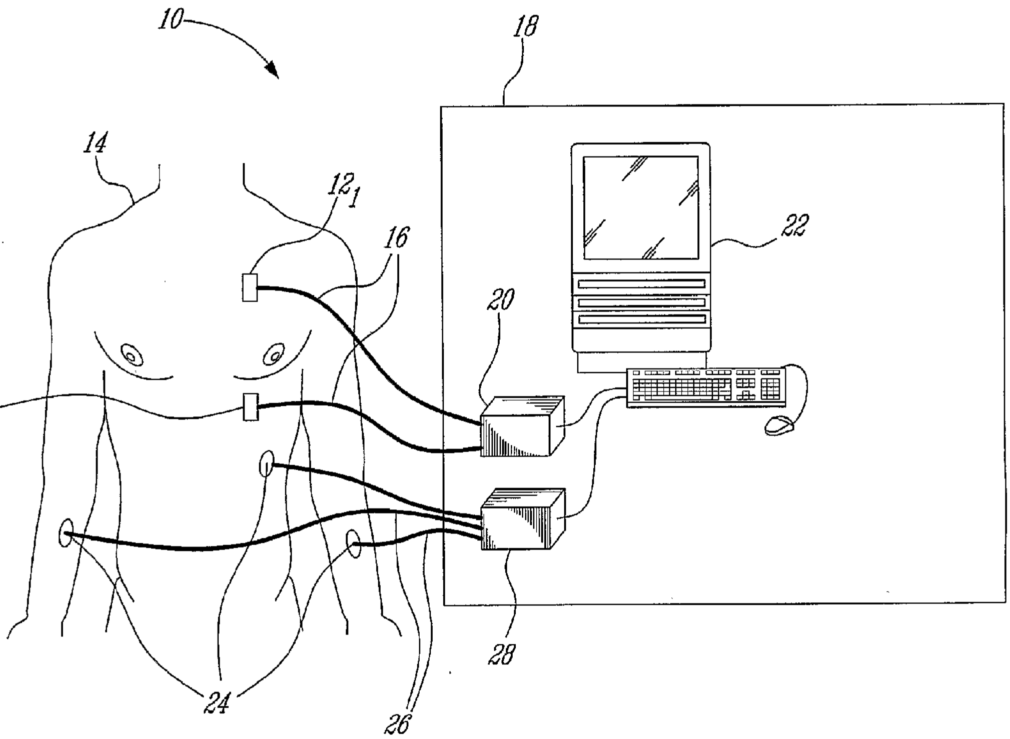
Publication Classification

(51) **Int. Cl.**
A61B 5/021 (2006.01)

(52) **U.S. Cl.** **600/485**

(57) **ABSTRACT**

A pulmonary artery pressure estimator attaches a plurality of acoustic sensors to a patient so as to measure a second heart sound. The sensors are arranged so that an A2 component of the second heart sound is maximized from at least one of the sensors and a P2 component is maximized from at least another one of the sensors. Electrodes are also attached to the patient so as to measure a cardiac interval. A splitting interval is derived from the A2 and P2 components, which is normalized by the cardiac interval. The normalized splitting interval provides an estimation of the pulmonary artery pressure (PAP).



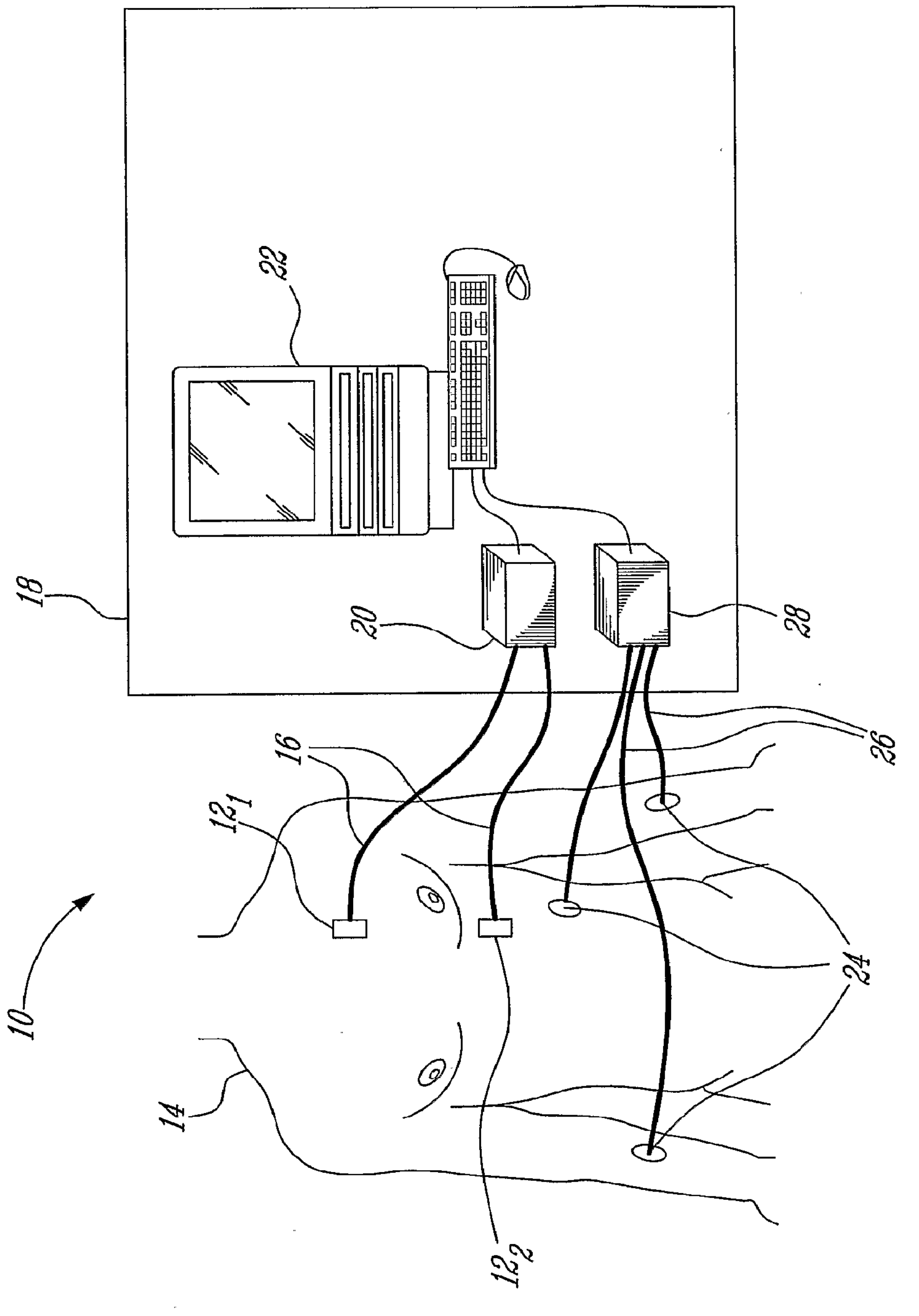


FIG. 1

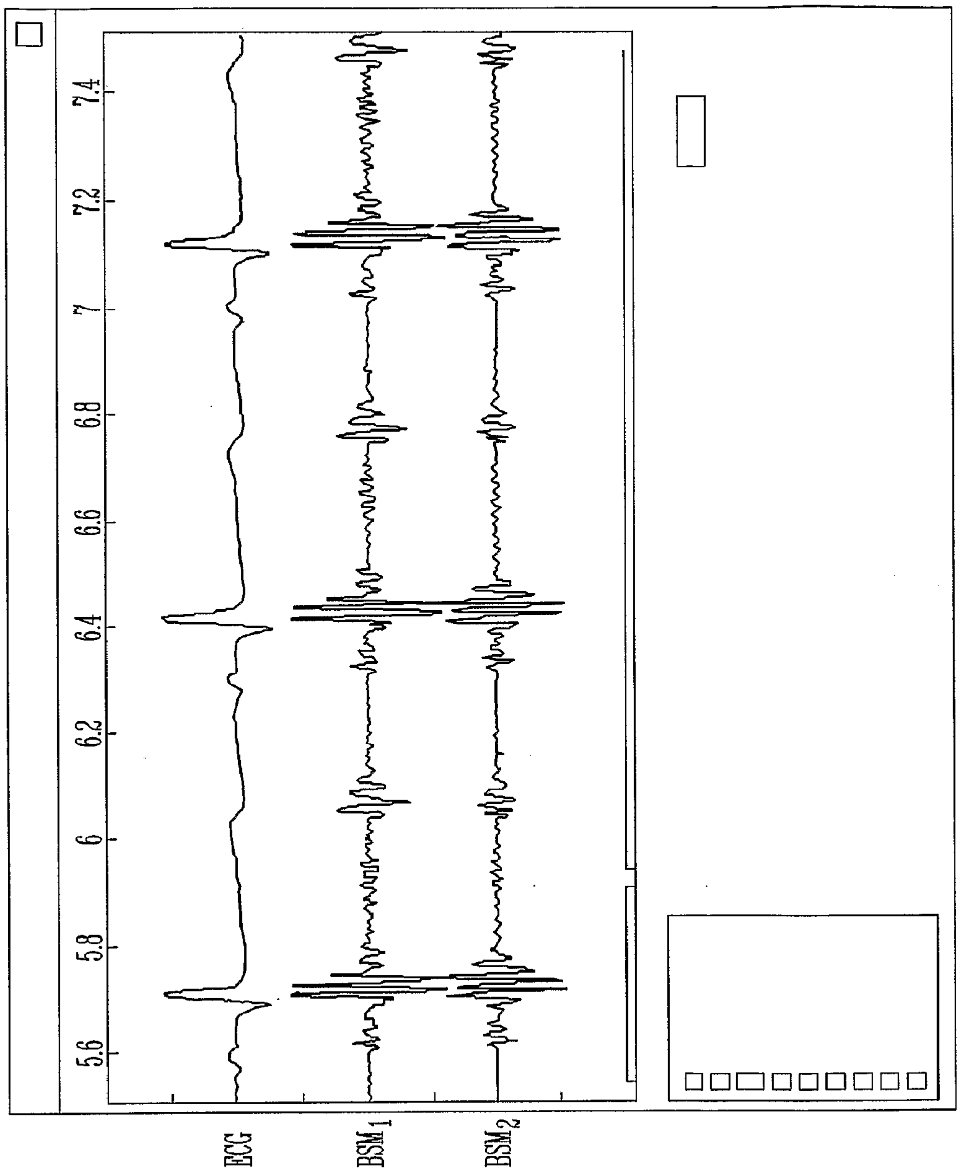


FIG. 2

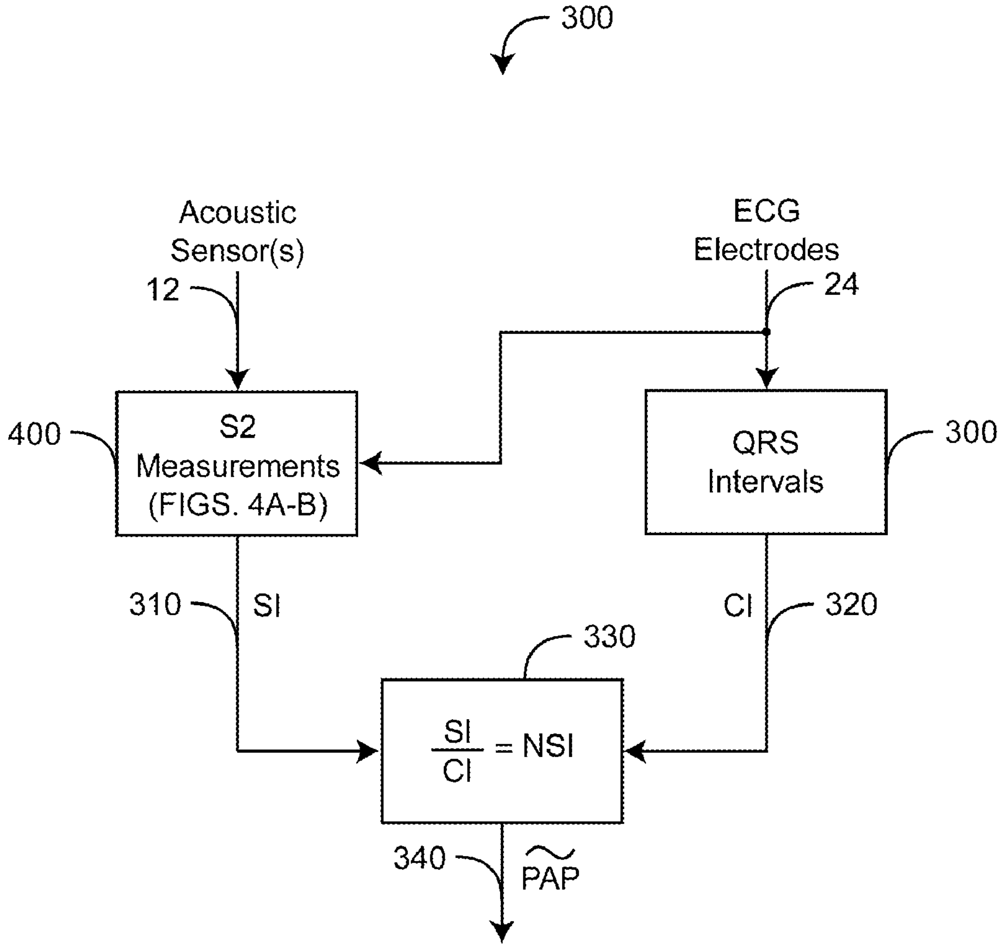

300
Acoustic Sensor(s)
12
ECG Electrodes
24
400
S2 Measurements (FIGS. 4A-B)
QRS Intervals
300
310
SI
CI
320
330
$\frac{SI}{CI}$ = NSI
340
$\widetilde{PAP}$

FIG. 3

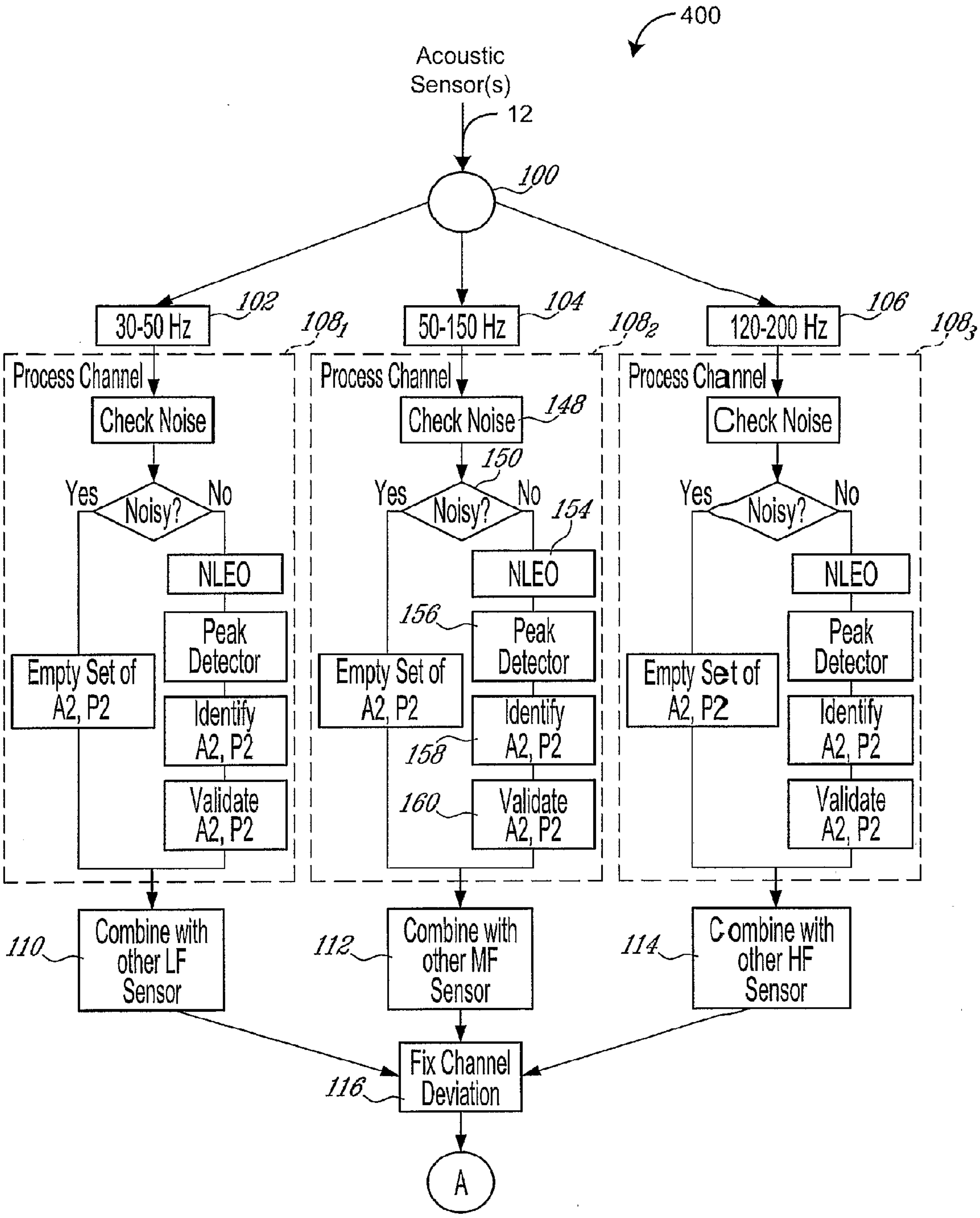

400
Acoustic
Sensor(s)
12
100
30-50 Hz
102
50-150 Hz
104
120-200 Hz
106
108_1
108_2
108_3
Process Channel
Check Noise
148
150
Yes
Noisy?
No
154
NLEO
156
Peak
Detector
Empty Set of
A2, P2
Identify
A2, P2
158
160
Validate
A2, P2
110
Combine with
other LF
Sensor
112
Combine with
other MF
Sensor
114
Combine with
other HF
Sensor
116
Fix Channel
Deviation
A

FIG. 4A

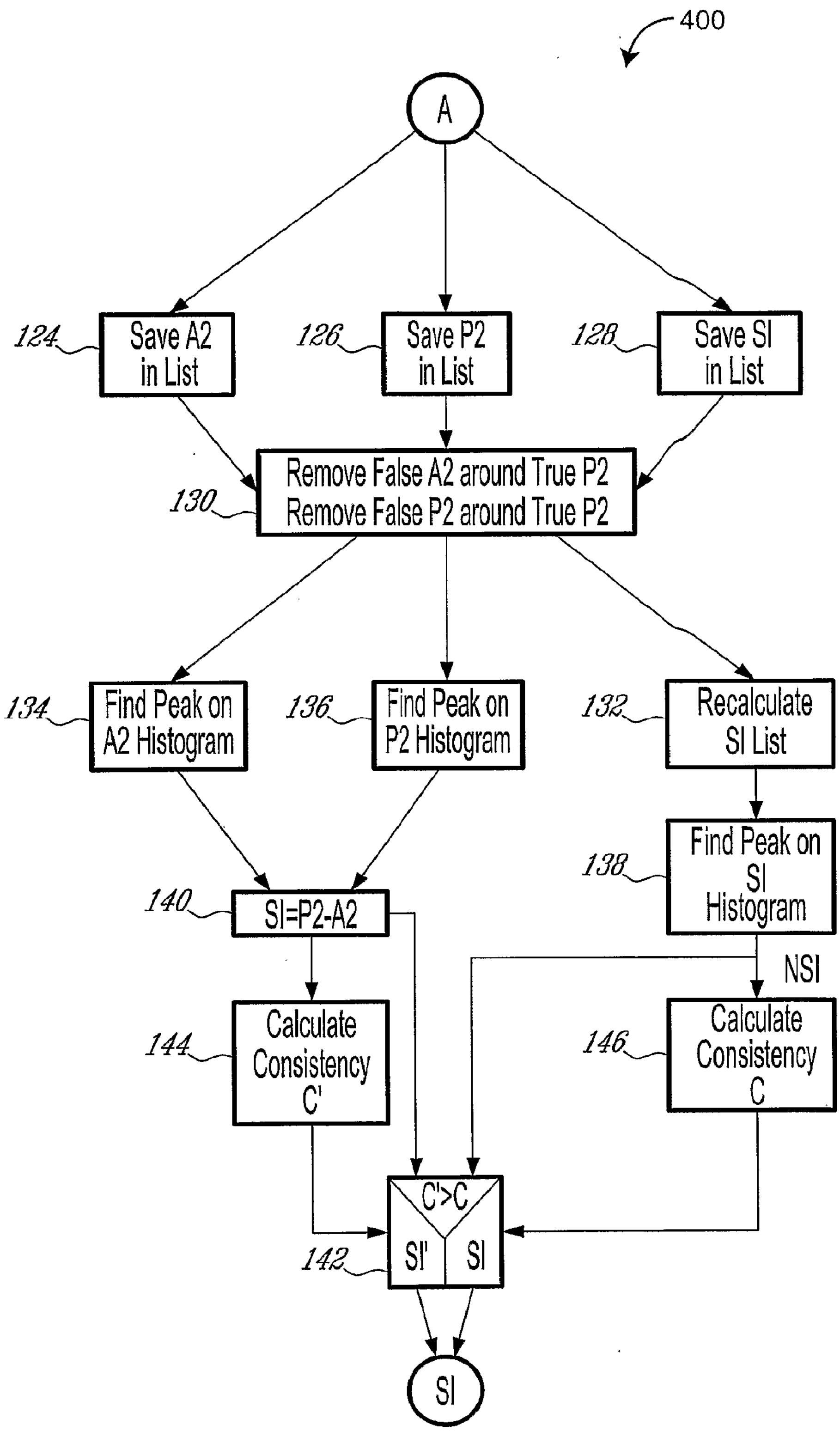

400
A
124
Save A2 in List
126
Save P2 in List
128
Save SI in List
130
Remove False A2 around True P2
Remove False P2 around True P2
134
Find Peak on A2 Histogram
136
Find Peak on P2 Histogram
132
Recalculate SI List
138
Find Peak on SI Histogram
140
SI=P2-A2
NSI
144
Calculate Consistency C'
146
Calculate Consistency C
C'>C
142
SI'
SI
SI

FIG. 4B

PULMONARY ARTERY PRESSURE ESTIMATOR

PRIORITY CLAIM TO RELATED APPLICATIONS

[0001] The present application is a continuation of U.S. patent application Ser. No. 11/578,462 titled *Non-Invasive Measurement of Second Heart Sound Components*, issuing Mar. 22, 2011 as U.S. Pat. No. 7,909,772, which is the national stage of International Application No. PCT/CA2005/000568 filed Apr. 15, 2005; which claims priority benefit of U.S. Provisional Patent Application No. 60/562,538 titled PAP Estimator, filed Apr. 16, 2004; all the above-referenced provisional, non-provisional and international applications are hereby incorporated by reference herein.

BACKGROUND OF THE INVENTION

[0002] The highly publicized problem of cardio-vascular diseases, an increased population living excess of eighty and the predominance of the heart disease as a leading cause of death have increased the importance of the clinical practitioner's ability to recognize abnormal heart conditions. Traditionally, non-invasive heart diagnostics is based upon auscultation and a physician's ability to use a stethoscope to recognize specific patterns and phenomena. Through advances in technology, many of these abilities have been automated. For some of these auscultation methods, however, a stable automated procedure has yet to be found.

SUMMARY OF THE INVENTION

[0003] Pulmonary artery hypertension is caused by several heart or pulmonary diseases including dysfunction of prosthetic or native heart valves, left ventricular dysfunction, congenital abnormalities of the heart and great vessels, chronic obstructive pulmonary disease, and adult respiratory distress syndrome. Pulmonary artery hypertension is a serious cardiovascular dysfunction that is difficult to assess non-invasively. In patients requiring continuous monitoring or with those suspected of having pulmonary artery hypertension, the pulmonary artery pressure (PAP) is usually measured using a pulmonary arterial catheter. This is an invasive surgical procedure that is associated with significant morbidity and modality. A pulmonary arterial catheter can be left in place for a few days to allow continuous monitoring of PAP of patients in the critical care unit. However, due to the potential risk to patients, catheterization is not recommended for repeated measurements. Nonetheless, regular evaluation of the PAP is very important to identify pulmonary hypertension and subsequently to follow the evolution of the disease and to assess the efficacy of treatment. Consequently, the development of non-invasive methods that allow frequent and accurate measurement of PAP are particularly advantageous.

[0004] Doppler echocardiography is one non-invasive method that can be used to estimate the systolic PAP. If tricuspid regurgitation is detected, it is possible to estimate the systolic pressure gradient across the tricuspid valve using continuous-wave Doppler. The right ventricular systolic pressure can be calculated by sliding the systolic tricuspid valve gradient to the estimated right atrial pressure. The right ventricular systolic pressure can be considered equivalent to the systolic PAP when the systolic pressure gradient across the pulmonary valve is negligible. This noninvasive method can provide a high degree of correlation ($0.89<r<0.97$) and a standard error estimate (SEE) varying from 7 to 12 mmHg with pulmonary artery catheterization (for systolic PAP range: 20-180 mmHg). However, the estimation of PAP by Doppler echocardiography has several important limitations. First, PAP cannot be estimated by Doppler in approximately 50% of patients with normal PAP, 10 to 20% of patients with elevated PAP and 34 to 76% of patients with chronic obstructive pulmonary disease. This is due to the absence of tricuspid regurgitation, a weak Doppler signal or a poor signal-to-noise ratio. To improve the feasibility of the Doppler echocardiography in some of these patients, it is necessary to use contrast agent enhancement. Second, Doppler echocardiography tends to overestimate PAP in patients with normal PAP and significantly underestimate the PAP in patients with severe pulmonary arterial hypertension.

[0005] A PAP estimation method is based on the spectral properties of the P2 component of the second heart sound. The basic principle supporting this approach is based on Laplace's Law and assumes that the tension of the pulmonary wall is proportional to the corresponding intra-arterial blood pressure. Similar to a stretched drumhead, it is expected that the resonant frequency of the blood column in the pulmonary artery is proportional to the tension in the arterial wall and thus to the arterial pressure. Spectral methods have shown a high correlation between spectral features of P2 and the systolic PAP measured by PA catheterization. The two features were the dominant frequency and the quality factor of the spectrum of P2. In patients with prosthetic heart valves it has been demonstrated that systolic PAP estimated by Doppler echocardiography can be predicted from the spectral features extracted from P2.

[0006] For diagnostic of cardiac events, one of the most interesting sounds is the second heart sound. This sound comprises two components which are generally of interest: the aortic component and the pulmonary component. Detection and recognition of those components provides the possibility of measuring the systole and diastole duration for both the left- and right heart. These values are very important for many applications such as detection of pulmonary artery hypertension, dysfunction of heart valves, left and right ventricular dysfunction, etc.

[0007] As described hereinabove, the second heart sound and the components A2 and P2 thereof have significant clinical value. However, these components are very often masked by noises and other acoustic components of both the heart sounds and other parts of human body. As result, typically only specially trained and experienced clinicians can distinguish the A2 and P2 components. As a result, an automated computer-based procedure for A2 and P2 components would be desirable in clinical practice.

[0008] Cardiac catheterization and echocardiography, which have provided an accurate diagnosis of both right- and left heart abnormalities, have added a new dimension to usefulness of the phonocardiogram in assessing the presence and severity of cardiovascular abnormalities. Although cardiac catheterization generally provides the decisive evidence of the presence and severity of cardiac abnormalities, the external sound recordings correlate sufficiently well with the internal findings for them to serve, in many instances, as diagnostic tool per se. In this regard, phonocardiography often provides information complementary to that obtained by echocardiography. With this enhanced diagnostic accuracy, simpler and less painful external techniques can be used to determine when a patient needs more extensive cardiac treatment. Even in those cases where cardiac catheterization is deemed necessary, the knowledge gained beforehand through phonocardiography and other non-invasive studies can lead to much more efficient and fruitful invasive study.

[0009] An advantageous approach for estimating PAP, described herein, uses advanced signal processing techniques to determine a normalized splitting interval (NSI), which is normalized time interval between the two components of the second heart sound (S2), i.e. the aortic (A2) and pulmonary (P2) components. The splitting interval is normalized by a cardiac interval (CI) derived from an ECG. In an embodiment, the cardiac interval is derived from an average of intervals between a predetermined number of consecutive QRS waves of the ECG.

[0010] A PAP estimator determines a location of pulmonary and aortic components of second heart sounds of a patient over an interval. The method comprises the steps of producing an electronic representation of heart sounds of the patient over the interval, identifying at least one second heart sound in the interval using the electronic representation, for each identified second heart sound calculating a frequency weighted energy (FWE), normalizing the FWE, identifying peaks in the FWE, determining a maximum peak from the identified peaks and retaining the maximum peak and peaks having an amplitude within a predetermined amount of an amplitude of the maximum peak, wherein if two or more peaks are retained, two largest peaks are selected, a first peak as a candidate value for the aortic component and a second peak as a candidate value for the pulmonary component, wherein the first peak is prior to the second peak and wherein if only a single peak is retained, the single peak is selected as a candidate value for the aortic component, and generating an estimated value for a location of the aortic component and the pulmonary component from the candidate values.

[0011] In an embodiment of this method, peaks having an amplitude of a predetermined value between 90% and 100% of the amplitude of the maximum peak are retained. Peaks having an amplitude less than a predetermined value between 90% and 100% of the amplitude of the maximum peak are discarded. Similarly, in an embodiment of this method, peaks are retained which have an amplitude lower than the amplitude of the maximum peak by a value between 0 and 10% of the amplitude of the maximum peak.

[0012] There is also provided a method for estimating a location of pulmonary and aortic components of second heart sounds of a patient over an interval. The method comprises the steps of producing an electronic representation of heart sounds of the patient over the interval, dividing the electronic representation into a plurality of sub-channels, for each of the sub-channel representations, identifying at least one second heart sound in the interval using the electronic representation and extracting an estimated location of a sub-channel aortic component and a sub-channel pulmonary component from the at least one second heart sound, combining the estimated sub-channel aortic component locations to form the estimated aortic component location and the estimated sub-channel pulmonary component locations to form the estimated pulmonary component location.

[0013] Additionally, there is provided a method for estimating a location of pulmonary and aortic components of second heart sounds a patient over an interval. The method comprises the steps of positioning a first transducer at a first position on the patient, the first transducer producing a first electronic representation of heart sounds of the patient over the interval, positioning a second transducer at a second position on the patient, the second transducer producing a second electronic representation of heart sounds of the patient over the interval, for the first electronic representation identifying at least one second heart sound in the interval, for each identified second heart sound calculating a FWE, normalizing the FWE, identifying peaks in the FWE, determining a maximum peak from the identified peaks and retaining the maximum peak and peaks having an amplitude within a predetermined amount of an amplitude of the maximum peak, wherein if two or more peaks are retained, two largest peaks are selected, a first peak as a candidate value for the aortic component and a second peak as a candidate value for the pulmonary component, wherein the first peak is prior to the second peak and wherein if only a single peak is retained, the single peak is selected as a candidate value for the aortic component, and generating a first estimated value for a location of an aortic component and a pulmonary component from the candidate values and for the second electronic representation identifying at least one second heart sound in the interval, for each identified second heart sound calculating a FWE, normalizing the FWE, identifying peaks in the FWE, determining a maximum peak from the identified peaks and retaining the maximum peak and peaks having an amplitude within a predetermined amount of an amplitude of the maximum peak, wherein if two or more peaks are retained, two largest peaks are selected, a first peak as a candidate value for the aortic component and a second peak as a candidate value for the pulmonary component, wherein the first peak is prior to the second peak and wherein if only a single peak is retained, the single peak is selected as a candidate value for the aortic component and generating second estimated values for a location of the aortic component and the pulmonary component from the candidate values and combining the first and second estimated aortic location values and the first and second estimated pulmonary location values wherein the estimated location of the aortic components is the combined first and second estimated aortic location values and the estimated location of the pulmonary components is the combined first and second estimated pulmonary location values.

[0014] Furthermore, there is provided a method for estimating pulmonary artery pressure of a patient over an interval. The method comprises the steps of producing an electronic representation of heart sounds of the patient over the interval, identifying at least one second heart sound in the interval using the electronic representation, for each identified second heart sound calculating a FWE, normalizing the FWE, identifying peaks in the FWE, determining a maximum peak from the identified peaks and retaining the maximum peak and peaks having an amplitude within a predetermined amount of an amplitude of the maximum peak, wherein if two or more peaks are retained, two largest peaks are selected, a first peak as a candidate value for the aortic component and a second peak as a candidate value for the pulmonary component, wherein the first peak is prior to the second peak and wherein if only a single peak is retained, the single peak is selected as a candidate value for the aortic component and generating an estimated value for a location of an aortic component and a location of pulmonary component from the candidate values, determining a splitting interval as a time between the aortic component location and the pulmonary component location, normalizing the splitting interval, and estimating the systolic pulmonary artery pressure using a predetermined function which describes a relationship between the normalized splitting interval and the systolic and diastolic pulmonary artery pressures. Also, there is provided an apparatus implementing any of the above methods.

BRIEF DESCRIPTION OF THE DRAWINGS

[0015] FIG. **1** is a general illustration of a pulmonary artery pressure (PAP) estimator embodiment;

[0016] FIG. **2** are graphs of typical signals detected by a PAP estimator embodiment;

[0017] FIG. **3** is a block diagram of a PAP estimator based upon a normalized splitting interval (SI); and

[0018] FIGS. **4**A-B are flow charts of the A2, P2 and SI detection embodiments.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

[0019] FIG. **1** generally illustrates a pulmonary artery pressure (PAP) estimator embodiment **10** having two identical biological sound (acoustic) sensors **12**, for example those described in U.S. Pat. No. 6,661,161, incorporated by reference herein, are provided for, although in a given application a single sensor or more than two sensors may be preferable. In the case of a multiple sensor schema, sensors are placed at different locations on the patient **14**, where there is expected to be a maximal intensity of the aortic component of the second heart sound A2 or the pulmonary component of the second heart sound P2 or both A2 and P2 signals. In the illustrated example one sensor **121** is positioned at the apex of heart, where the A2 component of the S2 sound is likely at its maximal in intensity and P2 component is minimal. A second sensor **122** is placed to maximize the P2 component intensity (between the 3rd and 4th left intercostal space). The best sensor locations are obtained by experimenting with different positions while observing S2 sound signals, so as to achieve the maximal signal intensity.

[0020] The sensors **12** are attached via appropriate leads as in **16** to a data acquisition system **18** comprised of an analog to digital converter **20** and personal computer **22**. Data collected by the sensors **12** is digitized by the analog to digital converter **20**, illustratively using a sampling rate of 2 kHz with 12 bits of resolution. Additionally, electrocardiogram (ECG) signals are also collected via a series of electrodes **24**, leads **26** and a second analog to digital converter **28**. Similar to the acoustic data collected by the biological sound sensors **12**, data collected by the ECG electrodes **24** is digitized by the analog to digital converter **28**, illustratively using a sampling rate of 2 kHz with 12 bits of resolution. As will be seen below, the electrocardiogram is used as the reference signal to frame the second heart sound (S2).

[0021] FIG. **2** illustrates typical signals detected by a PAP estimator embodiment, including an ECG reading displayed along side readings from first and second biological sound (acoustic) sensors. The ECG is used to provide the reference signal to frame the second heart sound. The ECG is also to determine a normalizing cardiac interval for the calculated splitting interval (SI). The beat signal in the description below means the part of acoustic signal between two consecutive QRS complexes on the ECG. Depending on the selected approach, the 'beat signal" can be defined as the Q-Q' (distance between two Q markers) or as the R-R' (distance between two R markers). In the following description Q-Q' provides the beat signal. For each beat signal the first heart signal (S1) is detected and removed. The remaining sounds, including the second heart sounds and possibly murmurs and the like, are then used as input.

[0022] FIG. **3** illustrates a pulmonary artery pressure (PAP) estimator **300** having acoustic sensor(s) **12** and ECG sensors **24** inputs. The acoustic sensor(s) **12** are used to measure second heart sounds **400** (FIGS. **4**A-B) so that the data acquisition system **18** (FIG. **1**) derives a splitting interval (SI) **310** output. The ECG sensors **24** are used by the data acquisition system **18** (FIG. **1**) to frame the second heart sounds so as to derive S2 measurements **400** and to derive a cardiac interval (CI) **320**. In an embodiment, CI **320** is based upon successive QRS intervals **300** of the ECG sensors **24** input. The data acquisition system **18** (FIG. **1**) utilizes CI to normalize SI **330** (NSI) so as to generate a PAP estimate **340** output.

[0023] FIGS. **4**A-B illustrate aortic component (A2), pulmonary component (P2) and splitting interval (SI) detection embodiments. The illustrative method supports input signals from the single or multiple sensor(s) **12**, each of them comprised of signals of heart sounds in the frequency range 30-200 Hz, although this range could be wider without any changes in the approach. If that range is narrower, however, the method should be adapted to those limitations.

[0024] As shown in FIG. **4**A, sounds related to heart beats are collected at **100** via a sensor(s) **12** and illustratively divided into three sub channels **102**, **104** and **106** (or frequency bands). These bands are: Low Frequency (LF, 30-50 Hz), Medium Frequency (MF, 50-150 Hz), and High Frequency (HF, 120-200 Hz).

[0025] Each sub-channel is relayed to a "Process Channel" block as in **1081**, **1082**, and **1083**, (these will be described separately below). The process channel block can be based on a variety of methods including a Chirplet method, Nonlinear Energy Operator (NLEO) method, or any other suitable method capable of extracting and discriminating A2 and P2 components from second heart sound S2. The illustrated embodiment applies the NLEO method.

[0026] The output values of A2 and P2 from the process channel blocks as in **1081**, **1082**, and **1083** are analyzed. If both components A2, P2 are clearly detectable in at least one of the sub channels, these are the values for A2, P2. If both components are not clearly detectable then the outputs of the process channel blocks as in **1081**, **1082**, and **1083** are compared sub-channel by sub-channel with the output of the process channel blocks for other sensors (not shown) of the same sub channels at blocks **110**, **112**, and **114**. In the case at hand, there are illustratively two sensors (the second sensor not shown) the outputs of the process blocks of which are thus compared pair wise.

[0027] Illustratively, the comparison is carried out on each frequency band according to the following set of rules, although it should be understood that this is an example and not intended to be limiting. If the output of **108** for both sensors reveals A2 and P2 components and the positions of A2 and P2 in each sensor output are the same, then these positions provide the values of A2 and P2. If one of the outputs of **108** for both sensors reveals A2 and P2 components, but the other does not, then the positions of these A2 and P2 provide the values of A2 and P2. If the output of **108** for both sensors reveals only one A2 or one P2 component then, as it is unknown whether the component is A2 or P2, then the value of A2 is the position of the first component and the value of P2 the position of the second component. If the output of **108** for one of the sensors reveals both A2 and P2 components while the output of **108** for the other sensor reveals only one (A2 or P2) component, then the readings for both sensors are combined (superimposed). If the result reveals only two components (A2 and P2) then the positions of these A2 and P2 provide the values of A2 and P2. If the result still reveals three components (where one or two of the results are A2 and/or P2 and the remainder the result of biological noise), then the readings are combined (superimposed) and the two components with the greatest FWE are selected as A2 and P2, the positions of these A2 and P2 provide the values of A2 and P2.

[0028] If the output of **108** for both sensors reveals A2 and P2 components but the positions of A2 and P2 are different, then, if the Splitting Interval (SI) of both sensors is less than 10 ms then the value of A2 is the position of A2 and the value of P2 is the position of P2 as determined via one of the sensors. If at least one of the SI from first or second sensor is greater than 10 ms, all components (A2 and P2) within 10 ms are merged. If only one component results, then the value of both A2 and P2 is the position of this one component and resulting SI is equal to zero. If two components result, then the value of A2 is the position of the first component and the value of P2 the position of the second component. If three components result, then the values of A2 and P2 are the positions of the two components with the greatest FWE. If four components result, then the values of A2 and P2 are the positions of A2 and P2 from the sensor where the amplitude of components FWE is greater than that of the other sensor. A similar approach is used in the case of multiple sensors.

[0029] Also shown in FIG. **4**A, the SI for each sub-channel, including combined channels, is also calculated. The A2 and P2 components in the LF, MF, and HF sub-channels have small variations in positioning because of different frequency content. As a result, at block **116**, heuristic rules are used to correct those deviations and produce A2 and P2 single values from the combination of A2 and P2 from all sub-channels (LF, MF, HF) as well as any combined values which may have been generated. An illustrative example of the heuristic rules applied at block **116** is as follows. If no values for both A2 and P2 are available in the MF and HF sub-channels and the SI of the LF channel>120 msec, then discard the SI of the LF channel. If values for both A2 and P2 are available in the LF and HF sub-channels and the SI of the LF channel>1.4*SI of the HF channel, then discard the SI of the LF channel. If values for both A2 and P2 are available in the LF and MF sub-channels, and the SI of the LF channel>1.4*the SI of the MF channel, then the SI of the LF channel=1.4*the SI of the MF channel. If values for both A2 and P2 are available in the MF and HF sub-channels, and the SI of the MF channel<1.4*the SI of the HF channel, then the SI of the HF channel=(1/1.4)*the SI of the MF channel.

[0030] As shown in FIG. **4**B, the values of A2, P2 and SI for the current beat are stored at blocks **124**, **126** and **128**. Illustratively, values of A2, P2 and SI calculated for beats during the previous minute are retained. At the same time consistency of solution and signal-to-noise ratio (SNR) for each sub-channel is estimated and stored in separate lists. In this regard, for each sub-range the SNR is estimated. Consistency indicates the percentage of beats not rejected due to high noise. Illustratively, in order to determine the SNR, the S2 sound is first detected as well as the precise position of the start and end of S2. The signal component (S) is calculated as the energy between the start and end of S2, divided by the duration of S2 (in msec). The noise component (N) is calculated as the energy within 50 msec segment before the start of S2 added to the energy within 50 msec segment after the end of S2 divided by 100 msec. The resulting signal-to-noise ratio is calculated as SNR=S/N.

[0031] After all beats within the time averaging interval (in the case at hand illustratively 1 minute) have been processed in the above manner, a series of values of A2, P2 and SI are ready for statistical validation. At a first step of the validation process the distributions of A2 and P2 are estimated and a threshold location in time from the start of S2 value T calculated using a Bayesian criterion. Typically between 50-200 beats are present during a one minute sampling interval. Histograms are used in order to provide an estimation of the distributions. The distribution law of SI is used for additional control of the T value in the case of multi-peak distribution of A2 or P2.

[0032] At block **130**, any values of A2 which are located at a time greater than T from the start of S2 and values of P2 located at a time of less than time T from the Start of S2 are discarded from the stored values. The SI values are then recalculated at block **132** using only those A2 and P2 values which still have pairs.

[0033] At blocks **134**, **136** and **138** the central peaks on the A2, P2 and SI histograms are estimated using a two-iteration method. During a first iteration the central peak of each histogram is identified. During a second iteration, 20% of the input values, those which are the most distant from each central peak are removed. The histogram is rebuilt using only the remaining input values. Then at block **140** the value SI'=P2−A2 is calculated.

[0034] At block **142**, SI' is compared with the peak value of SI calculated at block **138**. If the difference between SI and SI' is less than 1% of the average beat duration, the mean value of SI and SI' is produced as the final output value for SI. If the difference between SI and SI' is greater than 1% of the average beat duration, the values of SI, SI' having a higher consistency value, as previously calculated at blocks **144**, **146** provides the final output value.

[0035] As shown in FIG. **4**A, as stated above, the process channel block **108** can be based on a variety of methods including a Chirplet method, NLEO method, or any other suitable method capable of extracting and discriminating A2 and P2 components from second heart sound S2. Illustratively, the NLEO method is described and comprises the following processing steps. Referring to block $\mathbf{108_2}$, the Signal to Noise Ratio (SNR) is determined at block **148**. The NLEO method is described in "Adaptive Segmentation of Electroencephalographic Data Using a Nonlinear Energy Operator" by Agarwal, et al., Proceedings IEEE ISCAS '99, Orlando, Fla., 1999, incorporated by reference herein.

[0036] At decision block **150**, if the SNR is below a predetermined value (illustratively 1.5), the current beat in the channel being processed is discarded and no further processing steps carried out. Alternatively, if the SNR is above a predetermined value the NLEO function is calculated at block **154** using the current beat's signal. In this regard, the NLEO or any other individual implementation of FWE or any other individual implementation of the general family of Autocorrelators may be used. NLEO is a manipulation of digital signal described in the general case by:

$$\psi[n]=x(n-l)\cdot x(n-m)-x(n-p)\cdot x(n-q) \text{ for } l+m=p+q \tag{1}$$

[0037] One of NLEO's special properties is the ability to compactly describe the notion of a Frequency Weighted Energy (FWE), which is different from the mean-square energy as it reflects both the amplitude as well as the frequency content of a signal. For the special case where $l \neq m$ and $p \neq q$, given an input of additive white Gaussian noise (AWGN) the expected value of NLEO output is zero. Thus it has the ability to suppress noise. If we consider the case of an amplitude modulated short duration sinusoidal burst in the presence of random noise and structured sinusoidal interference (as in the case of the aortic and the pulmonary components of the S2 sound in the midst of noise), it is anticipated that the NLEO output will enhance FWE of each of these components while suppressing AWGN interference and provide a constant baseline for sinusoidal interference. The time-varying nature of amplitude (Gaussian) and chirping of the dominant rhythm will modulate the NLEO output and produce a detectable burst corresponding to each component in contrast to background clutter. It will then be possible to apply detection strategies on the NLEO output with S2 sound input. Illustratively, NLEO with parameters $l=2$, $m=3$, $p=1$, $q=4$ was applied.

[0038] Once the NLEO function is calculated, at block **156** the highest peak (maximum of NLEO output for given beat signal) is determined and those peaks having values within 0.05 of highest peak value are retained. In this regard, 0.05 provides good results, although other values may also provide adequate results. If more than two peaks remain, the A2 and P2 candidates are identified at block **158**. If only one peak is detected, then this is passed to the output and determined as A2 or P2 according to the procedure described hereinabove at paragraph **18**.

[0039] Finally, at block **160** the values of A2 and P2 are validated using list of heuristic rules. An illustrative example of such rules are if the time interval between A2 and P2 on NLEO is greater than 100 msec the component with lower FWE is considered invalid and if the time interval between A2 and P2 on NLEO is less than 10 msec, the component having a lower FWE is considered invalid.

[0040] A non-invasive PAP estimator has been disclosed in detail in connection with various embodiments. These embodiments are disclosed by way of examples only and are not to limit the scope of the claims that follow. One of ordinary skill in art will appreciate many variations and modifications.

What is claimed is:

1. A pulmonary artery pressure estimator comprising:

a first biological sound sensor located on a person proximate the apex of the heart so that an A2 component of a second heart sound is maximal in intensity and a P2 component of the second heart sound is minimal in intensity;

a second biological sound sensor located on the person proximately between the third and fourth left intercostal space so that the P2 component is maximal in intensity;

a plurality of electrodes located on the person so as to generate ECG signals;

a data acquisition system in communications with the sound sensors and electrodes so as to derive the second heart sounds components from the sound sensors and an ECG from the electrode ECG signals;

the data acquisition system derives a splitting interval (SI=P2−A2) from the sound sensors;

the data acquisition system derives a cardiac interval (CI) from the electrodes; and

the data acquisition system estimates a pulmonary artery pressure from the splitting interval normalized by the cardiac interval.

2. A pulmonary artery pressure estimating method comprising:

attaching a first acoustic sensor and a second acoustic sensor to a person so as to measure a second heart sound;

locating the first acoustic sensor on the person so as to maximize an A2 component of the second heart sound and minimize a P2 component of the second heart sound;

locating the second acoustic sensor on the person so as to maximize the P2 component;

attaching electrodes to the person so as to measure an ECG;

deriving a splitting interval from the first and second acoustic sensors;

deriving a cardiac interval from the electrodes; and

estimating a pulmonary artery pressure from the splitting interval normalized by the cardiac interval.

* * * * *

patsnap

专利名称(译)	肺动脉压力估算器
公开(公告)号	US20120071767A1
公开(公告)日	2012-03-22
申请号	US13/053202
申请日	2011-03-21
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IPC分类号	A61B5/021 A61B5/00 A61B5/0456 A61B5/0476 A61B7/04
CPC分类号	A61B5/021 A61B7/04 A61B5/0476 A61B5/0456
优先权	PCT/CA2005/000568 2005-04-15 WO 60/562538 2004-04-16 US 2464634 2004-04-16 CA
外部链接	Espacenet USPTO

摘要(译)

肺动脉压力估计器将多个声学传感器附接到患者以便测量第二心音。传感器布置成使得第二心音的A2分量从至少一个传感器最大化，并且P2分量从至少另一个传感器最大化。电极也附着在患者身上，以便测量心脏间隔。从A2和P2分量导出分裂间隔，其由心脏间隔归一化。归一化的分裂间隔提供肺动脉压（PAP）的估计。