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Diab

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(54) **PLETHYSMOGRAPH PULSE RECOGNITION PROCESSOR**

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(58) **Field of Classification Search**

USPC 600/301, 500, 323

See application file for complete search history.

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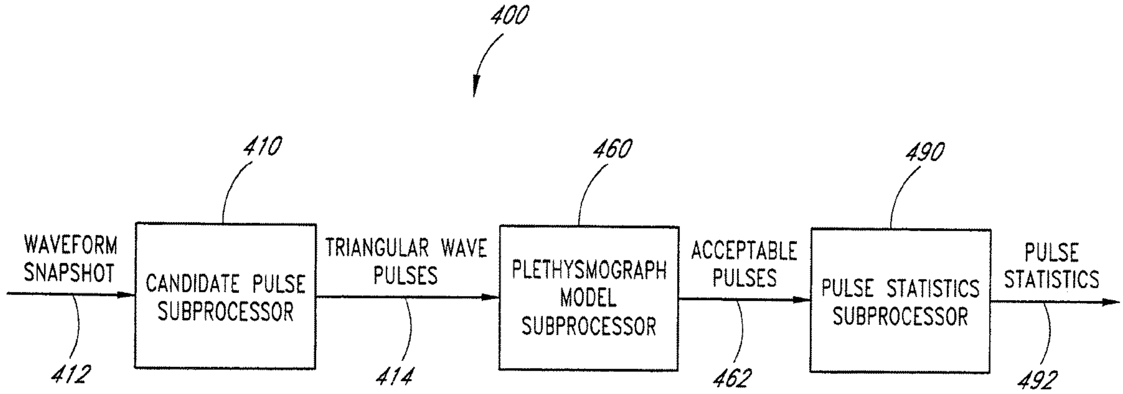
(57) **ABSTRACT**

A time domain rule-based processor provides recognition of pulses in a pulse oximeter-derived waveform.

4 Claims, 19 Drawing Sheets



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* cited by examiner

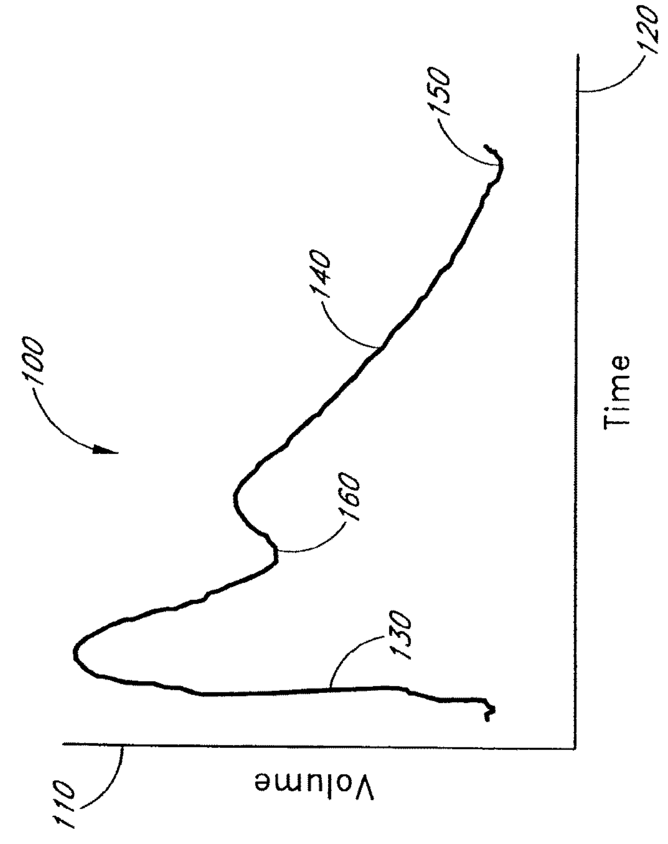

FIG._1

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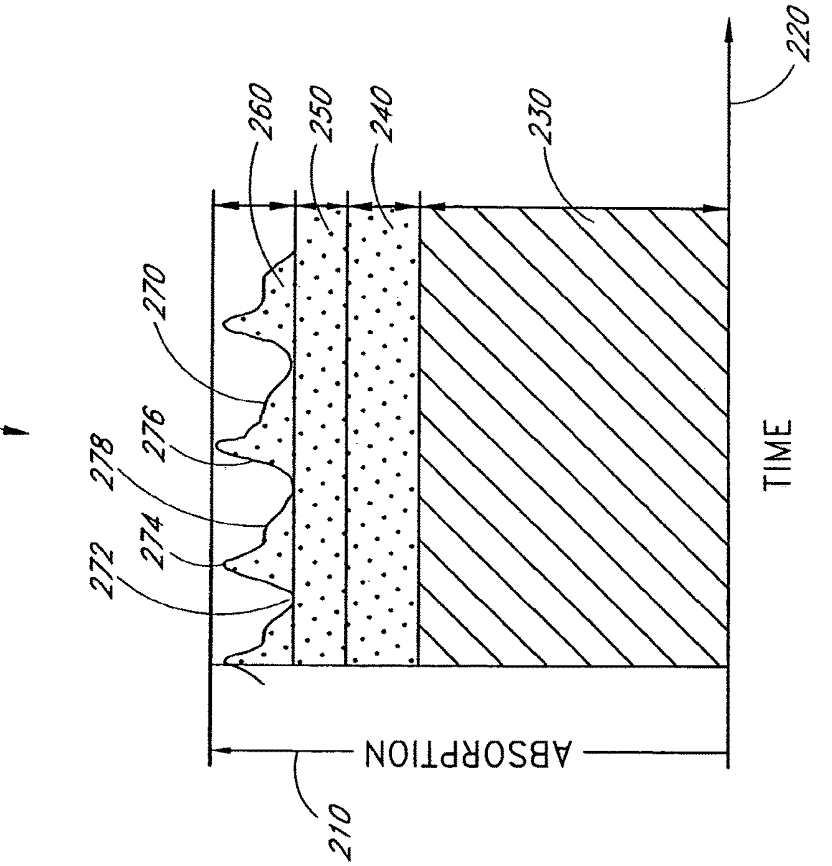


FIG. 2

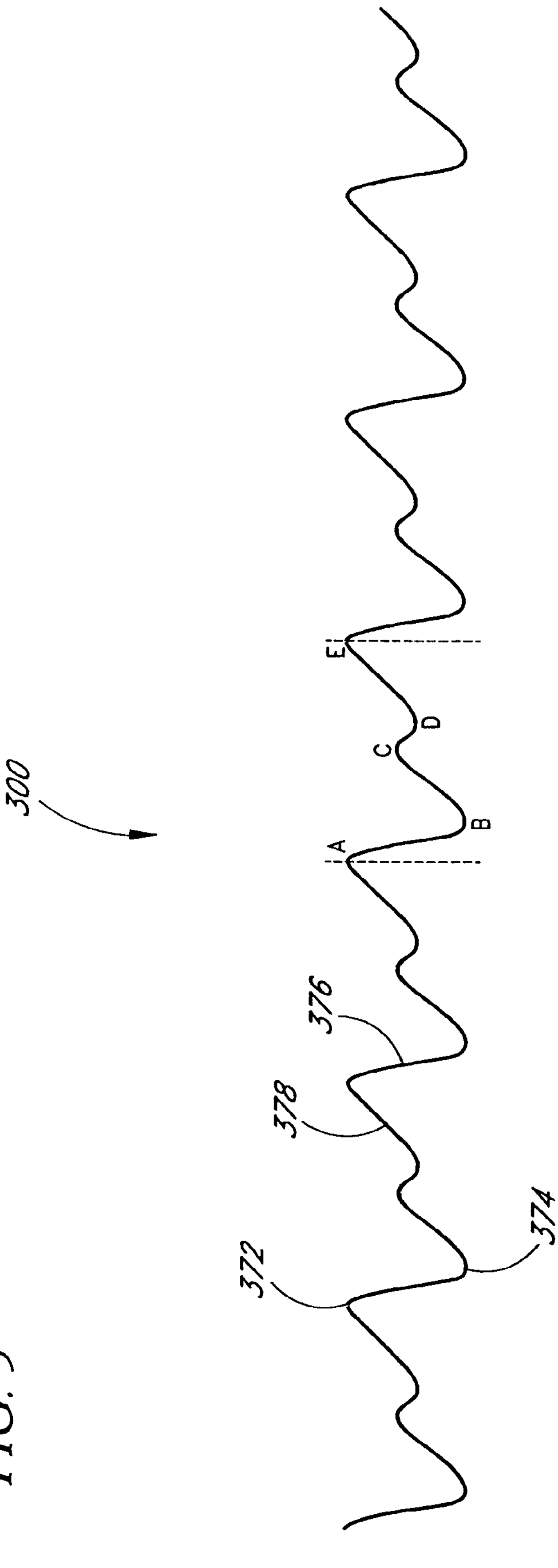
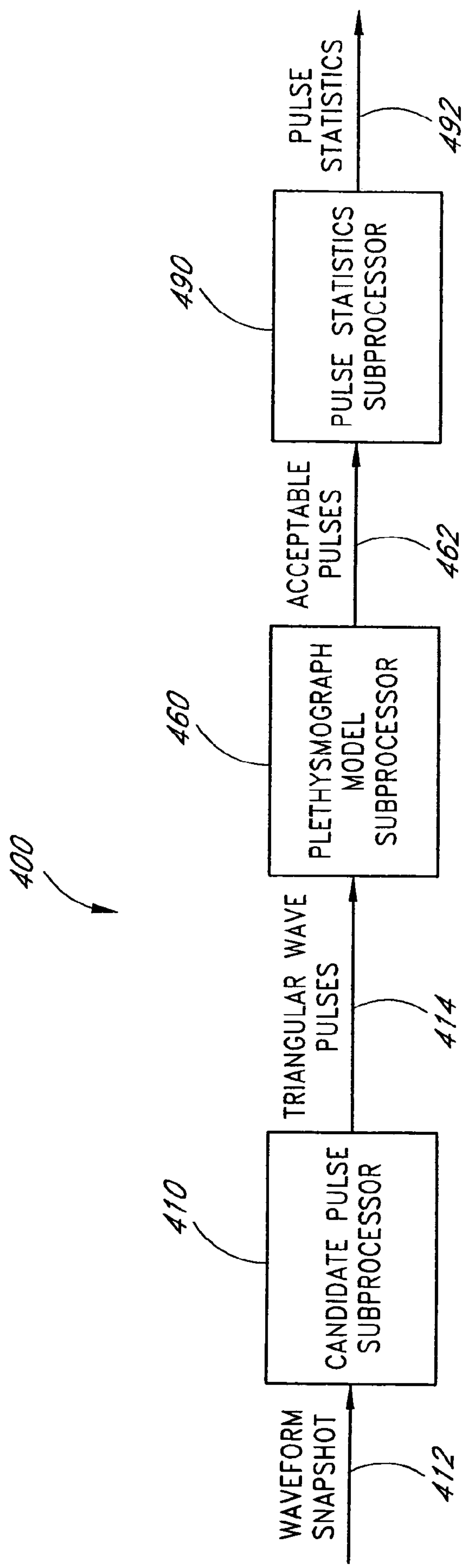


FIG. 3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412
CANDIDATE PULSE SUBPROCESSOR
410
TRIANGULAR WAVE PULSES
414
PLETHYSMOGRAPH MODEL SUBPROCESSOR
460
ACCEPTABLE PULSES
462
PULSE STATISTICS SUBPROCESSOR
490
PULSE STATISTICS
492
400

FIG. 4

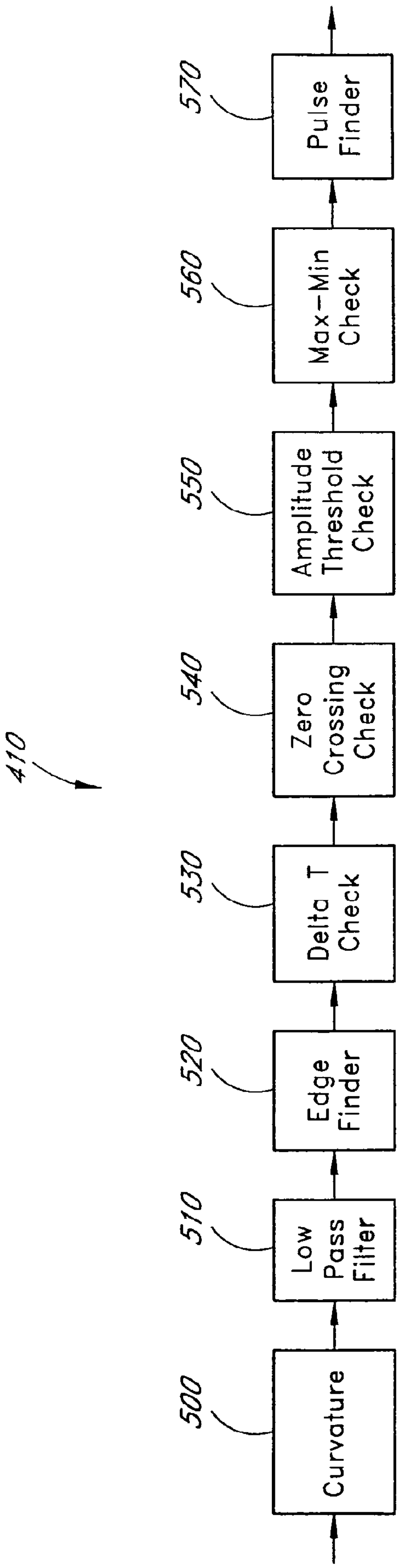
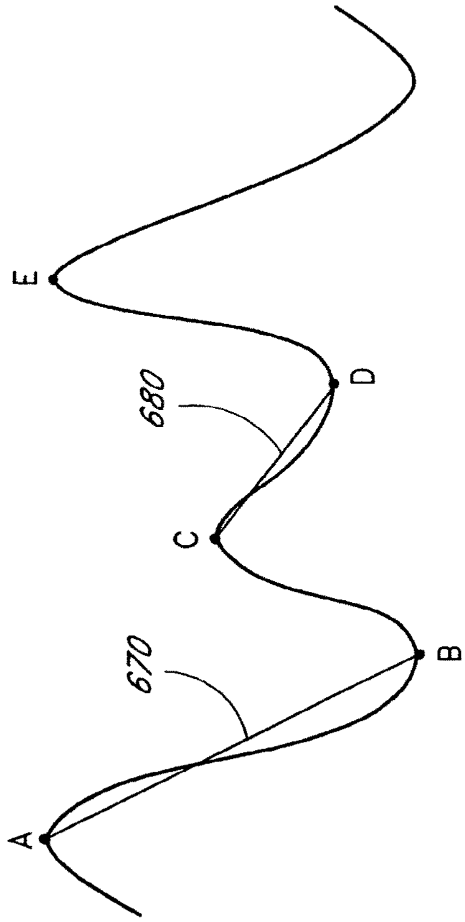

410
500
Curvature
510
Low Pass Filter
520
Edge Finder
530
Delta T Check
540
Zero Crossing Check
550
Amplitude Threshold Check
560
Max-Min Check
570
Pulse Finder

FIG. 5

FIG. 6



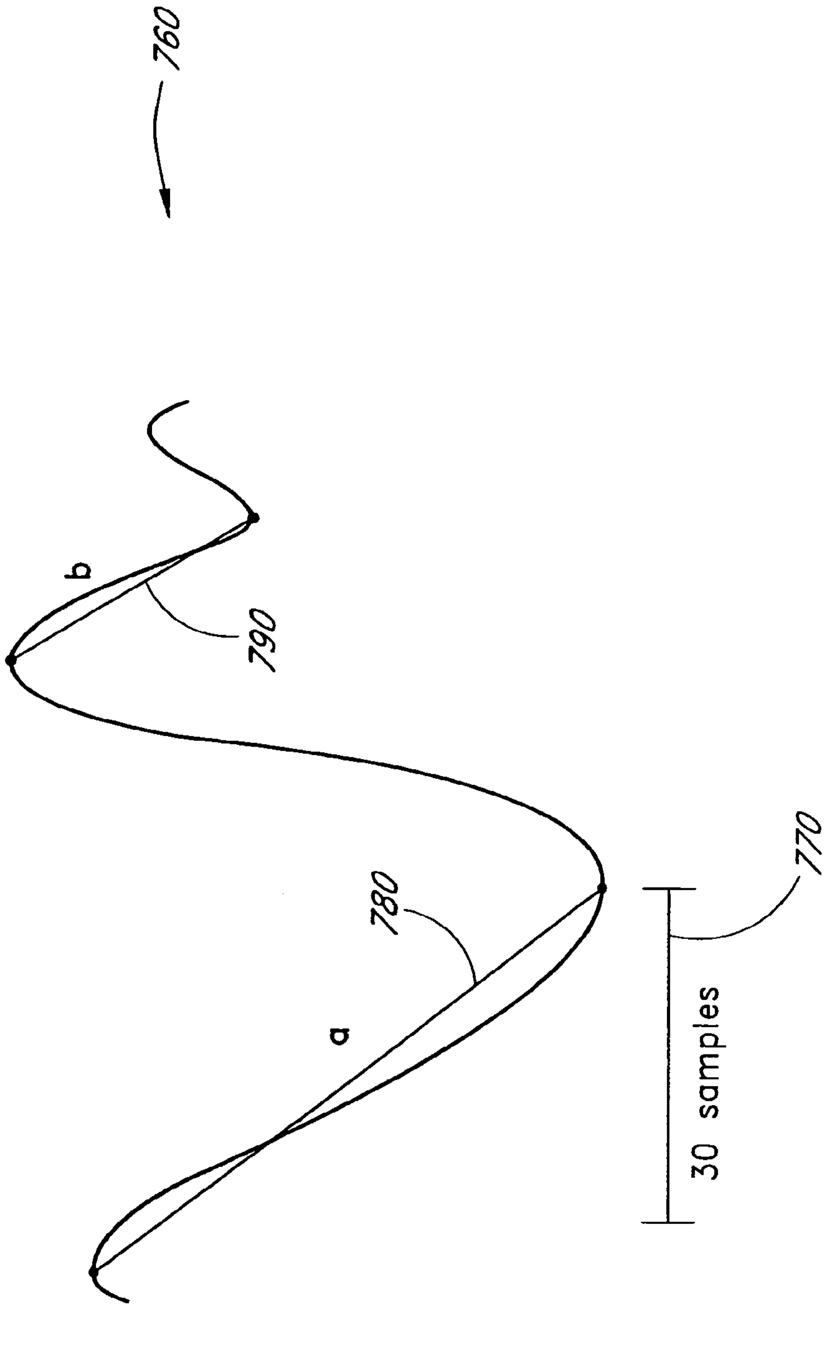


FIG. 7

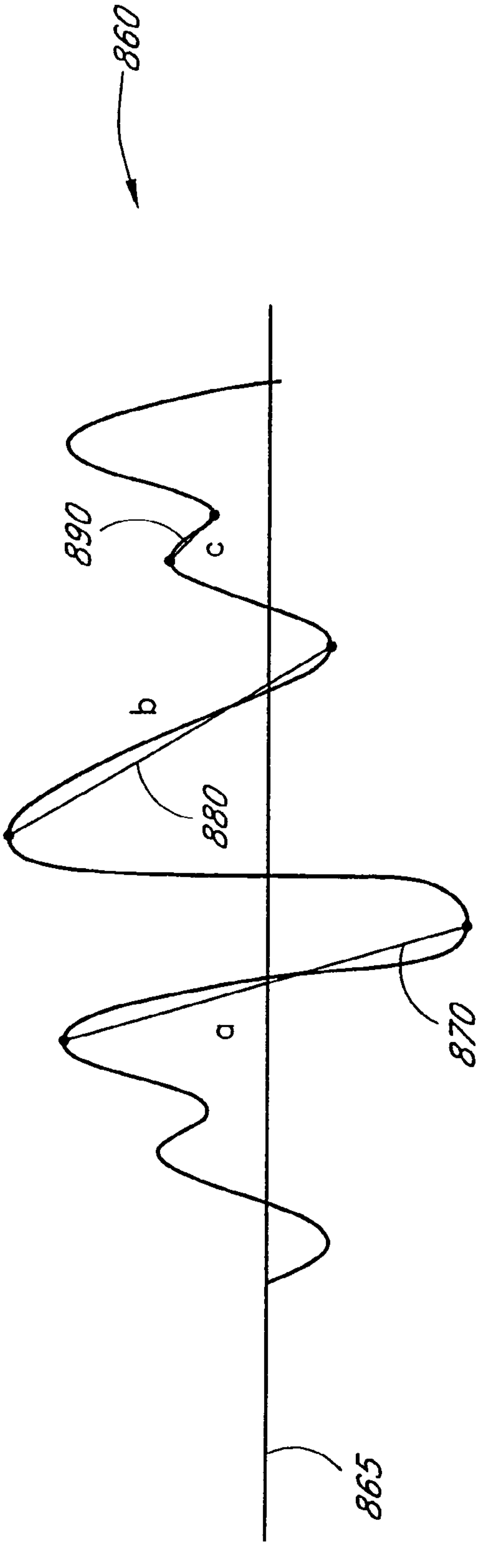


FIG. 8

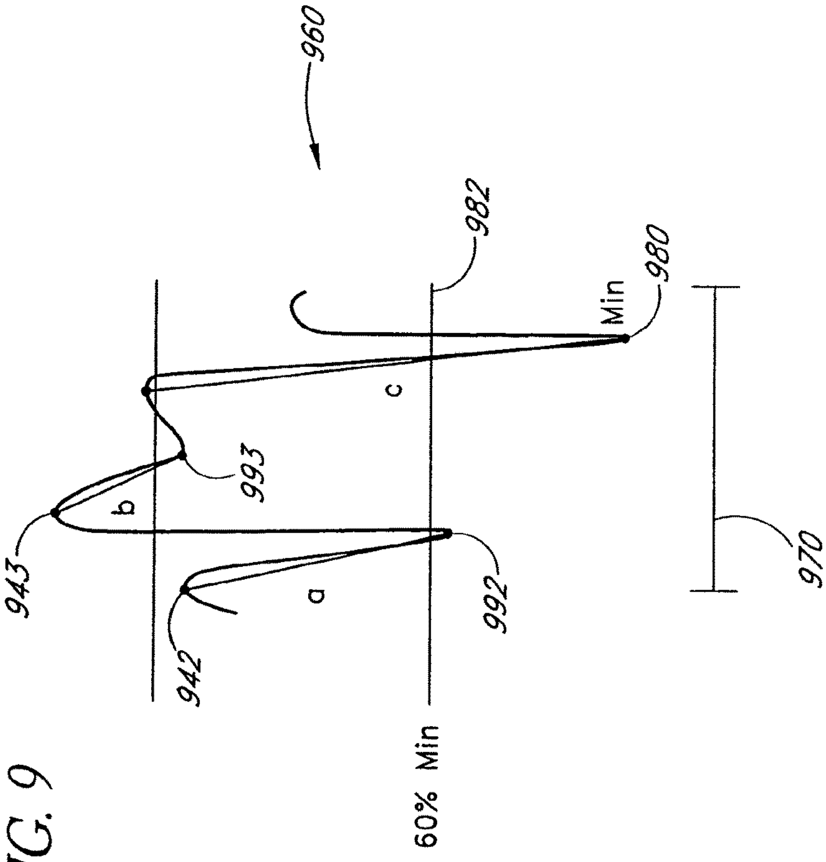
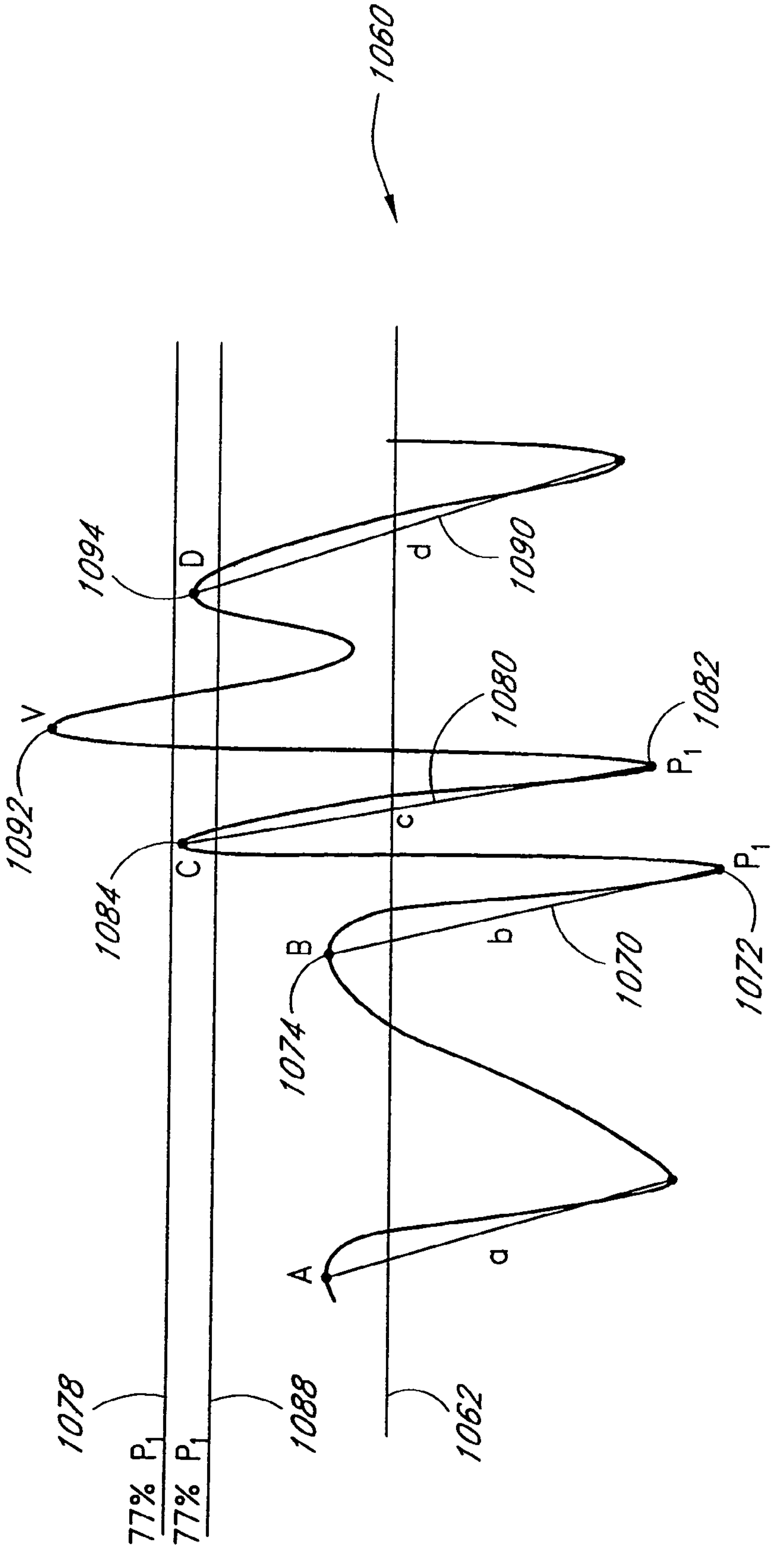


FIG. 9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1094
D
d
1090
1080
1082
V
P_1
1092
c
1084
C
P_1
B
b
1070
1072
1074
A
a
1078
1088
1062
77% P_1
77% P_1

FIG. 10

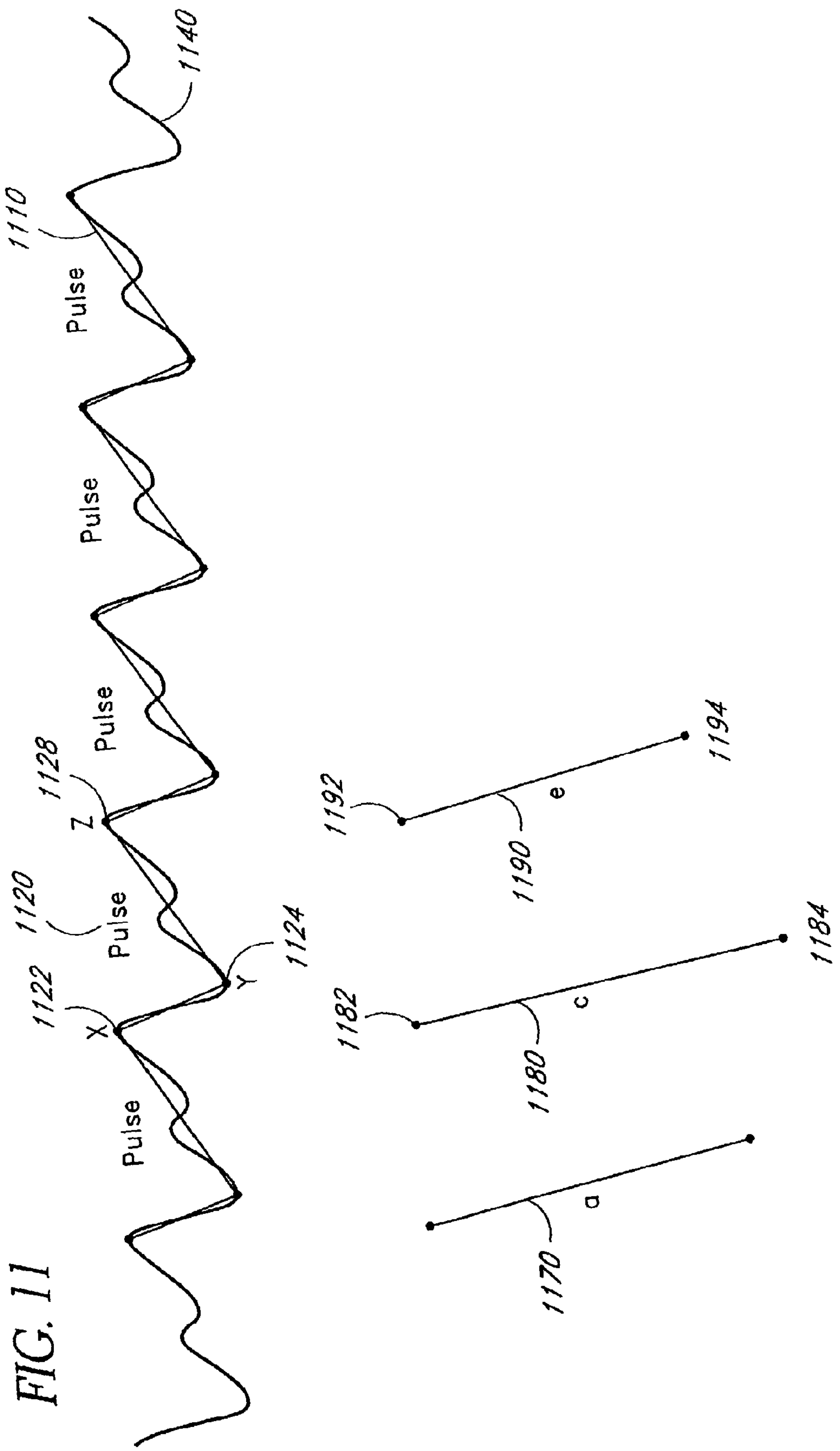

1140
1110
Pulse
Pulse
Pulse
1128
Z
1120
Pulse
1122
X
Y
1124
Pulse
1192
e
1194
1190
1182
c
1184
1180
a
1170

FIG. 11

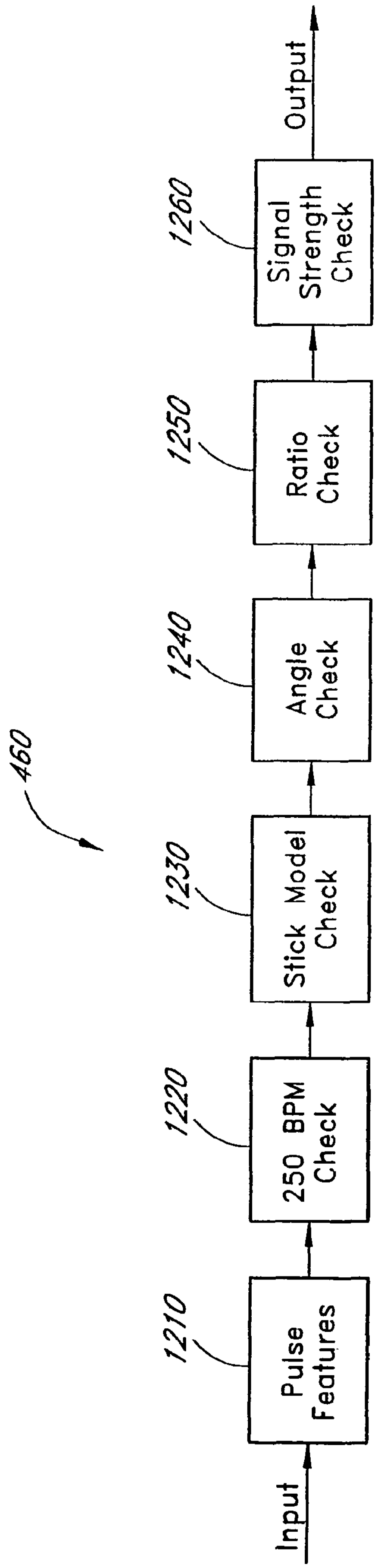

460
Input
1210
Pulse Features
1220
250 BPM Check
1230
Stick Model Check
1240
Angle Check
1250
Ratio Check
1260
Signal Strength Check
Output

FIG. 12

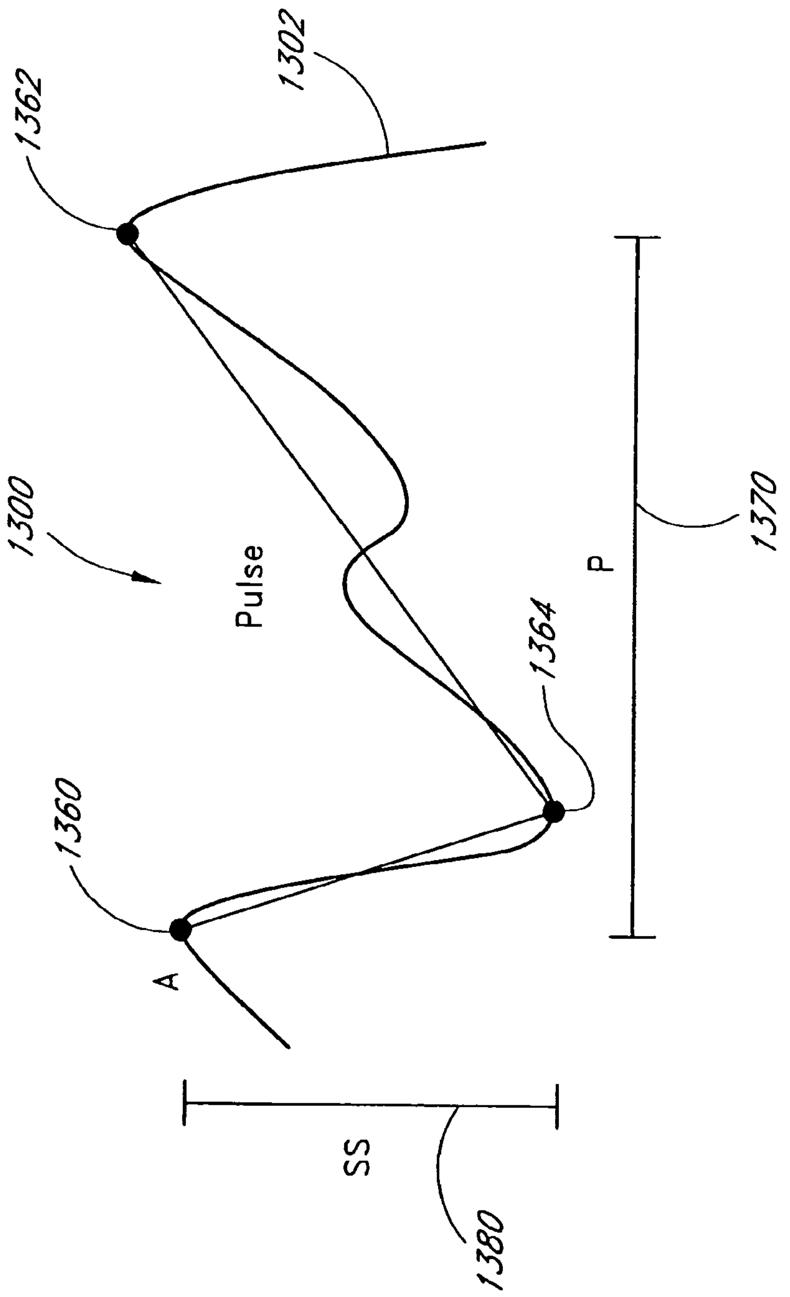


FIG. 13

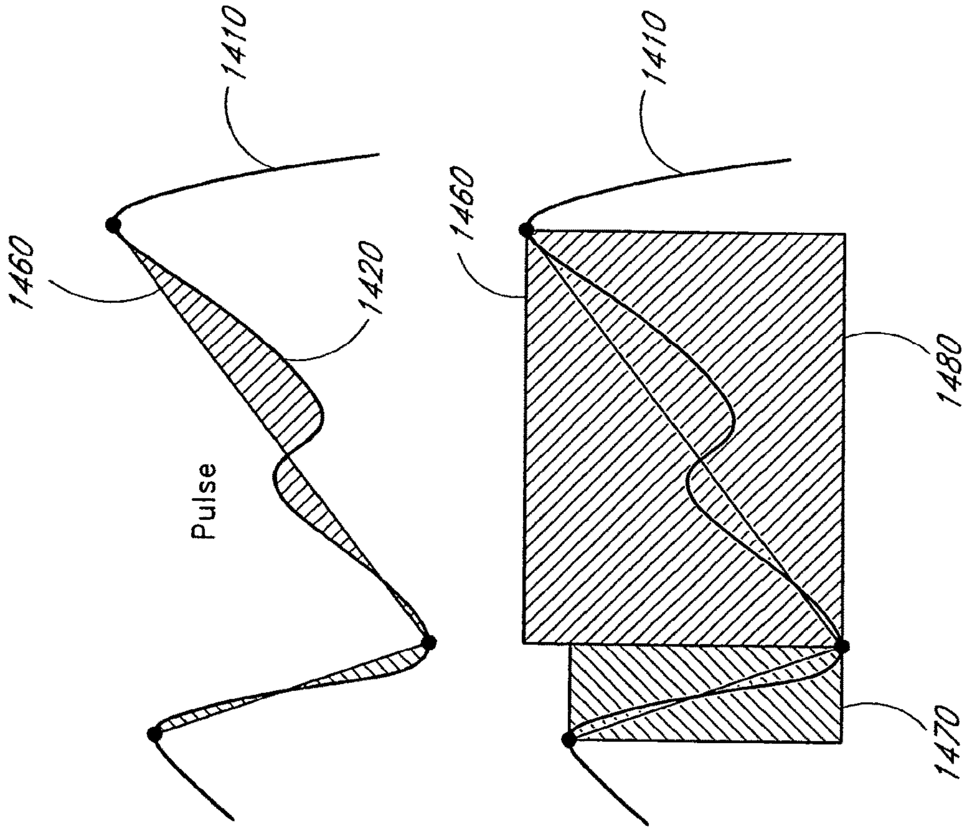


FIG. 14

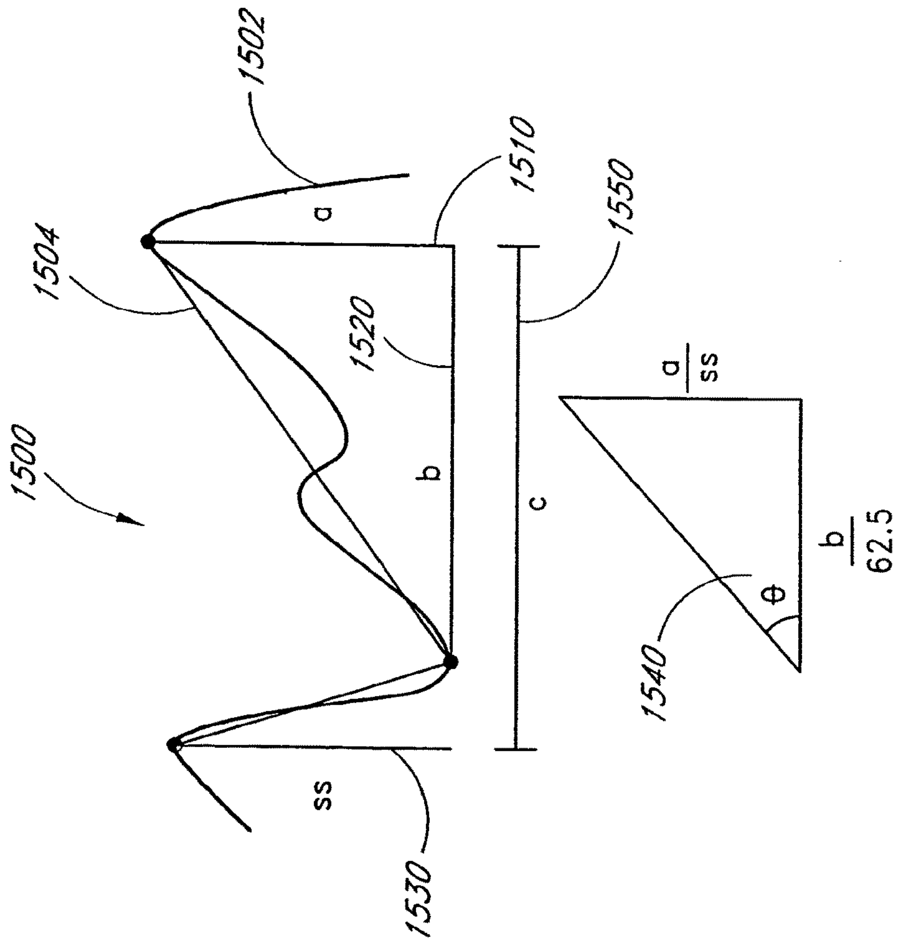


FIG. 15

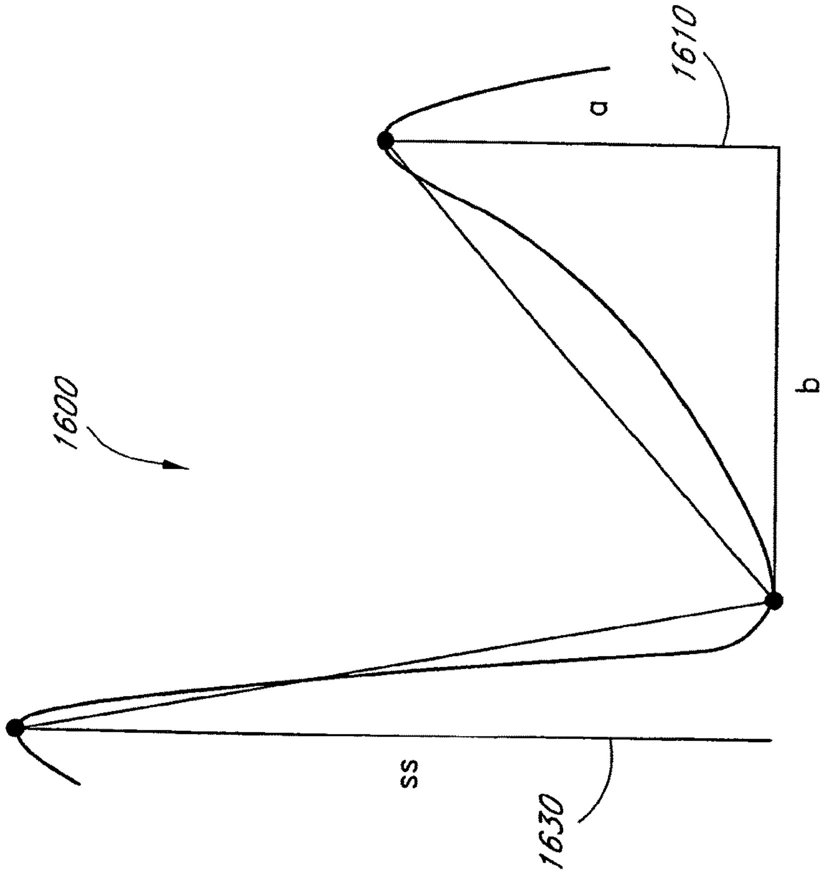


FIG. 16

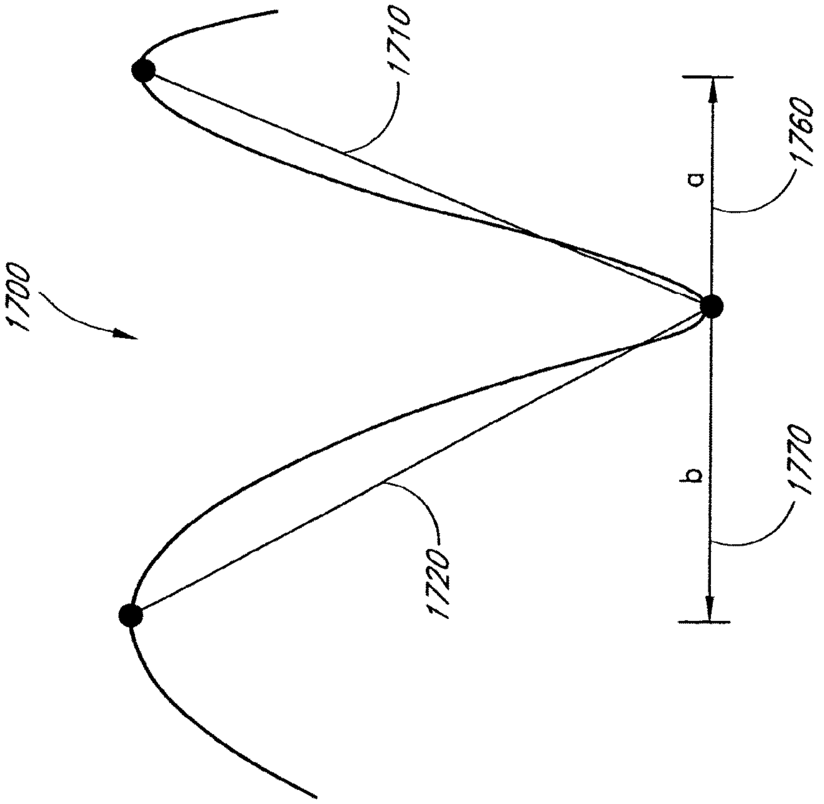
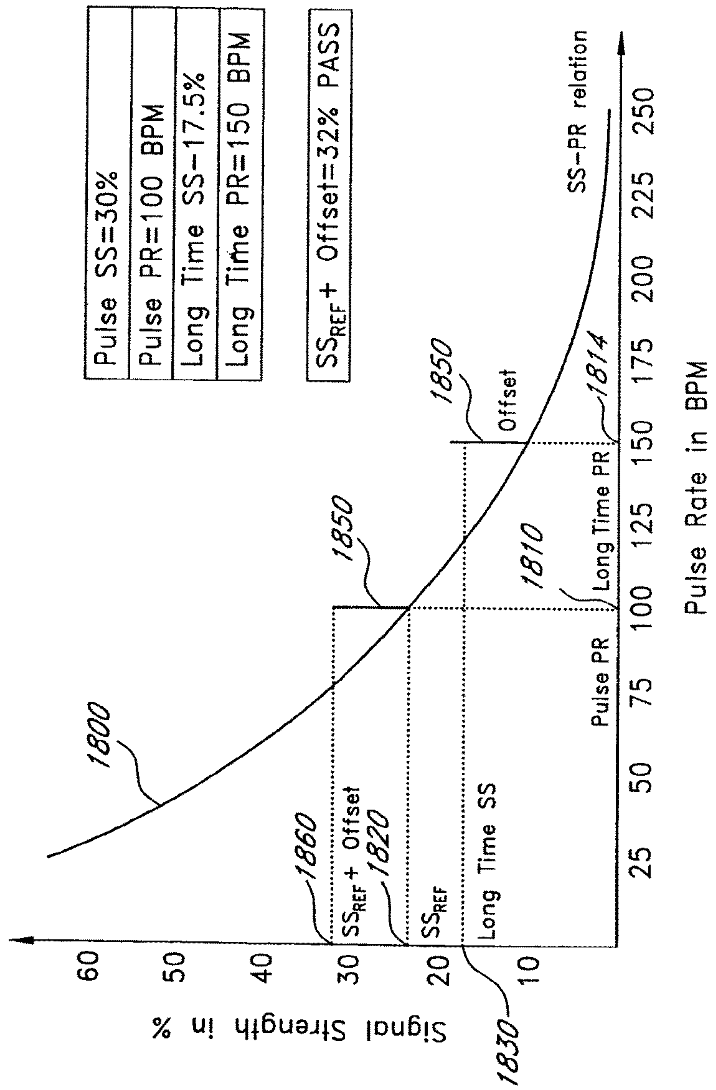


FIG. 17

FIG. 18



Pulse SS=30%
Pulse PR=100 BPM
Long Time SS-17.5%
Long Time PR=150 BPM
SS_REF + Offset=32% PASS
1800
1850
1850
Offset
SS-PR relation
1860
SS_REF + Offset
1820
SS_REF
Long Time SS
1830
1810
1814
Pulse PR
Long Time PR
Signal Strength in %
60
50
40
30
20
10
25
50
75
100
125
150
175
200
225
250
Pulse Rate in BPM

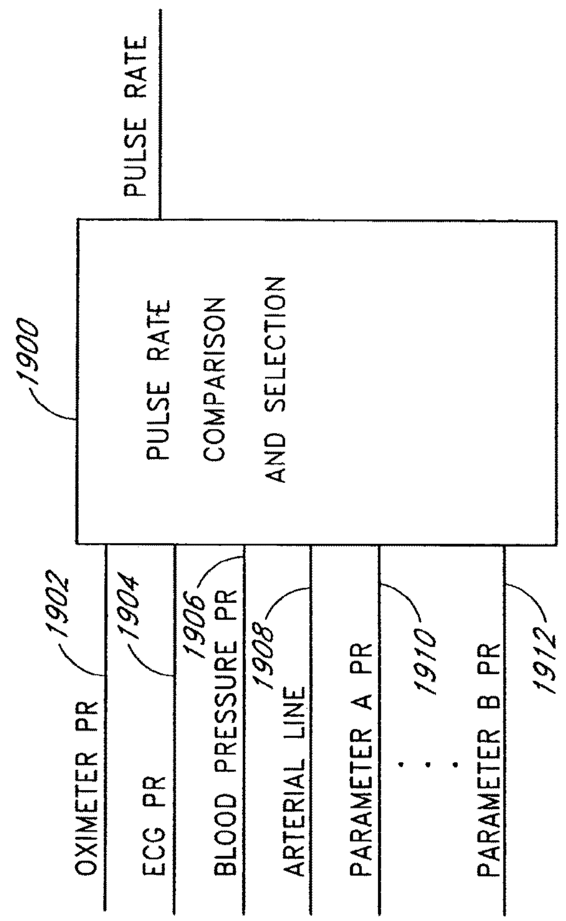

PULSE RATE
PULSE RATE COMPARISON AND SELECTION
1900
OXIMETER PR
1902
ECG PR
1904
BLOOD PRESSURE PR
1906
ARTERIAL LINE
1908
PARAMETER A PR
1910
PARAMETER B PR
1912

FIG. 19

PLETHYSMOGRAPH PULSE RECOGNITION PROCESSOR

CROSS-REFERENCE TO RELATED APPLICATION

The present application claims priority benefit under 35 U.S.C. §120 to, and is a continuation of U.S. patent application Ser. No. 11/418,328, filed May 3, 2006, entitled "Plethysmograph Pulse Recognition Processor," now U.S. Pat. No. 7,988,637, which is a continuation of U.S. patent application Ser. No. 10/974,095, filed Oct. 27, 2004, entitled "Plethysmograph Pulse Recognition Processor," now U.S. Pat. No. 7,044,918, which is a continuation of U.S. patent application Ser. No. 10/267,446, filed Oct. 8, 2002, entitled "Plethysmograph Pulse Recognition Processor," now U.S. Pat. No. 6,816,741, which is a continuation of U.S. patent application Ser. No. 09/471,510, filed Dec. 23, 1999, entitled "Plethysmograph Pulse Recognition Processor," now U.S. Pat. No. 6,463,311, which claims priority benefit under 35 U.S.C. §119(e) from U.S. Provisional Application No. 60/114,127, filed Dec. 30, 1998, entitled "Plethysmograph Pulse Recognition Processor." The present application also incorporates the foregoing utility disclosures herein by reference.

BACKGROUND OF THE INVENTION

Oximetry is the measurement of the oxygen status of blood. Early detection of low blood oxygen is critical in the medical field, for example in critical care and surgical applications, because an insufficient supply of oxygen can result in brain damage and death in a matter of minutes. Pulse oximetry is a widely accepted noninvasive procedure for measuring the oxygen saturation level of arterial blood, an indicator of oxygen supply. A pulse oximeter typically provides a numerical readout of the patient's oxygen saturation, a numerical readout of pulse rate, and an audible indicator or "beep" that occurs at each pulse.

A pulse oximetry system consists of a sensor attached to a patient, a monitor, and a cable connecting the sensor and monitor. Conventionally, a pulse oximetry sensor has both red and infrared (IR) light-emitting diode (LED) emitters and a photodiode detector. The sensor is typically attached to an adult patient's finger or an infant patient's foot. For a finger, the sensor is configured so that the emitters project light through the fingernail and into the blood vessels and capillaries underneath. The photodiode is positioned at the fingertip opposite the fingernail so as to detect the LED emitted light as it emerges from the finger tissues.

The pulse oximetry monitor (pulse oximeter) determines oxygen saturation by computing the differential absorption by arterial blood of the two wavelengths emitted by the sensor. The pulse oximeter alternately activates the sensor LED emitters and reads the resulting current generated by the photodiode detector. This current is proportional to the intensity of the detected light. The pulse oximeter calculates a ratio of detected red and infrared intensities, and an arterial oxygen saturation value is empirically determined based on the ratio obtained. The pulse oximeter contains circuitry for controlling the sensor, processing the sensor signals and displaying the patient's oxygen saturation and pulse rate. In addition, a pulse oximeter may display the patient's plethysmograph waveform, which is a visualization of blood volume change in the illuminated tissue caused by arterial blood flow over time. A pulse oximeter is described in U.S. Pat. No. 5,632,272 assigned to the assignee of the present invention.

SUMMARY OF THE INVENTION

FIG. **1** illustrates the standard plethysmograph waveform **100**, which can be derived from a pulse oximeter. The waveform **100** is a display of blood volume, shown along the y-axis **110**, over time, shown along the x-axis **120**. The shape of the plethysmograph waveform **100** is a function of heart stroke volume, pressure gradient, arterial elasticity and peripheral resistance. The ideal waveform **100** displays a broad peripheral flow curve, with a short, steep inflow phase **130** followed by a 3 to 4 times longer outflow phase **140**. The inflow phase **130** is the result of tissue distention by the rapid blood volume inflow during ventricular systole. During the outflow phase **140**, blood flow continues into the vascular bed during diastole. The end diastolic baseline **150** indicates the minimum basal tissue perfusion. During the outflow phase **140** is a dicrotic notch **160**, the nature of which is disputed. Classically, the dicrotic notch **160** is attributed to closure of the aortic valve at the end of ventricular systole. However, it may also be the result of reflection from the periphery of an initial, fast propagating, pressure pulse that occurs upon the opening of the aortic valve and that precedes the arterial flow wave. A double dicrotic notch can sometimes be observed, although its explanation is obscure, possibly the result of reflections reaching the sensor at different times.

FIG. **2** is a graph **200** illustrating a compartmental model of the absorption of light at a tissue site illuminated by a pulse oximetry sensor. The graph **200** has a y-axis **210** representing the total amount of light absorbed by the tissue site, with time shown along an x-axis **220**. The total absorption is represented by layers, including the static absorption layers due to tissue **230**, venous blood **240** and a baseline of arterial blood **250**. Also shown is a variable absorption layer due to the pulse-added volume of arterial blood **260**. The profile **270** of the pulse-added arterial blood **260** is seen as the plethysmograph waveform **100** depicted in FIG. **1**.

FIG. **3** illustrates the photo-plethysmograph intensity signal **300** detected by a pulse oximeter sensor. A pulse oximeter does not directly detect absorption and, hence, does not directly measure the standard plethysmograph waveform **100** (FIG. **1**). However, the standard plethysmograph can be derived by observing that the detected intensity signal **300** is merely an out of phase version of the absorption profile **270**. That is, the peak detected intensity **372** occurs at minimum absorption **272** (FIG. **2**), and the minimum detected intensity **374** occurs at maximum absorption **274** (FIG. **2**). Further, a rapid rise in absorption **276** (FIG. **2**) during the inflow phase of the plethysmograph is reflected in a rapid decline **376** in intensity, and the gradual decline **278** (FIG. **2**) in absorption during the outflow phase of the plethysmograph is reflected in a gradual increase **378** in detected intensity.

In addition to blood oxygen saturation, a desired pulse oximetry parameter is the rate at which the heart is beating, i.e. the pulse rate. At first glance, it seems that it is an easy task to determine pulse rate from the red and infrared plethysmograph waveforms described above. However, this task is complicated, even under ideal conditions, by the variety of physiological plethysmographic waveforms. Further, plethysmographic waveforms are often corrupted by noise, including motion artifact, as described in U.S. Pat. No. 2,632,272 cited above. Plethysmograph pulse recognition, especially in the presence of motion artifact and other noise sources, is a useful component for determining pulse rate and also for providing a visual or audible indication of pulse occurrence.

In one aspect of the pulse recognition processor according to the present invention, information regarding pulses within an input plethysmograph waveform is provided at a processor output. The processor has a candidate pulse portion that determines a plurality of potential pulses within the input waveform. A physiological model portion of the processor then determines the physiologically acceptable ones of these potential pulses. The processor may further provide statistics regarding the acceptable pulses. One statistic is pulse density, which is the ratio of the period of acceptable pulses to the duration of an input waveform segment.

The candidate pulse portion has a series of components that remove from consideration as potential pulses those waveform portions that do not correspond to an idealized triangular waveform. This processing removes irrelevant waveform features such as the characteristic dicrotic notch and those caused by noise or motion artifact. The candidate pulse portion provides an output having indices that identify potential pulses relative to the peaks and valleys of this triangular waveform.

The physiological model portion of the processor has a series of components that discard potential pulses that do not compare to a physiologically acceptable pulse. The first component of the model portion extracts features of the potential pulses, including pulse starting point, pulse period, and pulse signal strength. These features are compared against various checks, including checks for pulses that have a period below a predetermined threshold, that are asymmetric, that have a descending trend that is generally slower that a subsequent ascending trend, that do not sufficiently comply with an empirical relationship between pulse rate and pulse signal strength, and that have a signal strength that differs from a short-term average signal strength by greater than a predetermined amount.

In another aspect of the present invention, a pulse recognition method includes the steps of identifying a plurality of potential pulses in an input waveform and comparing the potential pulses to a physiological pulse model to derive at least one physiologically acceptable pulse. A further step of generating statistics for acceptable pulses may also be included. The generating step includes the steps of determining a total period of acceptable pulses and calculating a ratio of this total period to a duration of an input waveform segment to derive a pulse density value. The comparing step includes the steps of extracting pulse features from potential pulses and checking the extracted features against pulse criteria.

Yet another aspect of the current invention is a pulse recognition processor having a candidate pulse means for identifying potential pulses in an input waveform and providing a triangular waveform output. The processor also has a plethysmograph model means for determining physiologically acceptable pulses in the triangular waveform output and providing as a pulse output the indices of acceptable pulses. The pulse recognition processor may further have a pulse statistics means for determining cumulative pulse characteristics from said pulse output.

BRIEF DESCRIPTION OF THE DRAWINGS

The present invention will be described in detail below in connection with the following drawing figures in which:

FIG. **1** is a graph illustrating a single pulse of a plethysmograph waveform;

FIG. **2** is a graph illustrating the absorption contribution of various blood and tissue components;

FIG. **3** is a graph illustrating an intensity "plethysmograph" pulse oximetry waveform;

FIG. **4** is a block diagram of the plethysmograph pulse recognition processor according to the present invention;

FIG. **5** is a block diagram of the candidate pulse finding subprocessor portion of the present invention;

FIG. **6** is a graph illustrating the filtered, curvature of a plethysmograph pulse and the associated edges;

FIG. **7** is a graph illustrating the delta T check on the edges;

FIG. **8** is a graph illustrating the zero-crossing check on the edges;

FIG. **9** is a graph illustrating the amplitude threshold check on the edges;

FIG. **10** is a graph illustrating the max-min check on the edges;

FIG. **11** is a graph illustrating the output of the pulse finder;

FIG. **12** is a block diagram of the plethysmograph model subprocessor portion of the present invention;

FIG. **13** is a graph illustrating the parameters extracted by the pulse features component of the model subprocessor;

FIG. **14** is a graph illustrating the stick model check on the candidate pulses;

FIG. **15** is a graph illustrating an angle check on the candidate pulses;

FIG. **16** is a graph illustrating a pulse that would be discarded by the angle check;

FIG. **17** is a graph illustrating a pulse that would be discarded by the ratio check;

FIG. **18** is a graph illustrating one test of the signal strength check; and

FIG. **19** is a block diagram of a pulse rate selection and comparison module in accordance with a preferred embodiment of the present invention.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

FIG. **4** illustrates the plethysmograph pulse recognition processor **400** according to the present invention. The pulse processor **400** has three subprocessors, a candidate pulse subprocessor **410**, a plethysmograph model subprocessor **460**, and a pulse statistics subprocessor **490**. The candidate pulse subprocessor **410** applies various waveform criteria or "edge checks" to find candidate pulses in an input waveform "snapshot" **412**. In a particular embodiment, the snapshot is 400 samples of a detected intensity plethysmograph taken at a 62.5 Hz sampling rate. This snapshot represents a 6.4 second waveform segment. The output **414** of the candidate pulse subprocessor **410** is peaks and valleys of the input waveform segment representing a triangular wave model of identified candidate pulses. The candidate pulse output **414** is input to the plethysmograph model subprocessor **460**, which compares these candidate pulses to an internal model for physiological pulses. The output **462** of the plethysmograph model subprocessor **460** is physiologically acceptable pulses. The acceptable pulse output **462** is input to the pulse statistics subprocessor. The output **492** of the pulse statistics subprocessor is statistics regarding acceptable pulses, including mean pulse period and pulse density, as described below.

FIG. **5** illustrates the components of the candidate pulse subprocessor **410**. This subprocessor removes waveform features that do not correspond to an idealized triangular waveform, including the characteristic dicrotic notch. For example, as shown in FIG. **3**, the candidate pulse component must identify points ABE, discarding points CD. The candidate pulse subprocessor **410** first identifies "edges" within the input waveform segment. An edge is defined as a segment that connects a peak and subsequent valley of the filtered waveform signal. The candidate pulse processor **410** then discards edges that do not meet certain conditions.

As shown in FIG. **5**, the candidate pulse subprocessor has curvature **500**, low-pass filter **510** (in one embodiment) and edge finder **520** components that identify edges. In one embodiment, the curvature component **510** is implemented by convolving the waveform with the kernel [1,−2, 1]. In one embodiment, instead of a low-pass filter **510**, a band-pass filter can be used. For a kernel size of n, this can be represented as follows:

$$y_k = w y_{k-1} + u_k \quad (1)$$

where u_k is the kth input sample and y_k is the kth output sample and w is a fixed weight that determines the amount of filter feedback. The edge finder **520** identifies the peaks and subsequent valleys of the output of the filter **510**.

FIG. **6** illustrates the results of the curvature **500**, filter **510** and edge finder **520** components applied to a' couple waveform pulses **610**. The processed waveform **660** has peaks A and C and corresponding valleys B and D. There are two edges, a first edge is represented by a line segment **670** connecting A and B. A second edge is represented by a line segment **680** connecting C and D.

As shown in FIG. **5**, the candidate pulse portion also has delta T **530**, zero crossing **540**, amplitude threshold **550** and max-min **560** checks that eliminate certain of the identified edges. The delta T check **530** discards all the edges having a distance between end points that do not fall within a fixed interval. This is designed to eliminate pulse-like portions of the input waveform that are either too slow or too quick to be physiological pulses. In a particular embodiment, the interval is between 5 and 30 samples at the 62.5 Hz sampling rate, or 80-480 msec. That is, edges less than 80 msec. or greater than 480 msec. in length are eliminated.

FIG. **7** illustrates the delta T check **530** (FIG. **5**) described above. Shown is the processed waveform **760**, edge a **780** and edge b **790**, along with a maximum acceptable edge length interval **770** for comparison. In this example, edge a **780**, which is 35 samples in length, would be eliminated as exceeding in length the maximum acceptable interval **770** of 30 samples. By contrast, edge b **790**, which is 25 samples in length, would be accepted.

Also shown in FIG. **5**, the zero crossing check **540** eliminates all edges that do not cross zero. The zero crossing check eliminates small curvature changes in the input waveform segment, i.e. small bumps that are not peaks and valleys.

FIG. **8** illustrates the effect of the zero crossing check **540** (FIG. **5**). Shown is the processed waveform **860**. Edge a **870**, edge b **880** and edge c **890** are shown relative to the zero line **865** for the processed waveform **860**. In this example, edges a **870** and edge b **880** are accepted, but edge c **890** is eliminated because it does not cross the zero line **865**.

Shown in FIG. **5**, the amplitude threshold check **550** is designed to remove larger "bumps" than the zero crossing check **540**, such as dicrotic notches. This is done by comparing the right extreme (valley) of each edge within a fixed-length window to a threshold based on a fixed percentage of the minimum within that window. If the valley is not sufficiently deep, the edge is rejected. In a particular embodiment, the window size is set at 50 samples for neonates and 100 samples for adults in order accommodate the slower pulse rate of an adult. Also, a threshold of 60% of the minimum is used.

FIG. **9** illustrates an example of the amplitude threshold check **550** (FIG. **5**). Shown is a the processed waveform **960**. The starting point of the window **970** is set to the left extreme **942** (peak) of the first edge a. A minimum **980** within the window **970** is determined. A threshold **982** equal to 60% of the minimum **980** is determined. The right extreme **992** of edge a is compared with the threshold **982**. Edge a is kept because the right extreme **992** is smaller than (more negative) than the threshold **982**. The right extreme **993** of edge b is then compared with the threshold **982**. Edge b is removed because the right extreme **993** is greater than (less negative) than the threshold **982**. Similarly, edge c is kept. Next, the window **970** is moved to the left extreme **943** of edge b and the process repeated.

Also shown in FIG. **5**, the max-min check **560** applies another removal criteria to the edges. The max-min check **560** considers the interval of the processed waveform between the minimum of an edge being checked and the peak of the subsequent edge. The max-min check **560** finds the maximum of the processed waveform within this interval. The edge being checked is removed if the maximum is greater than a percentage of the absolute value of the right extreme (minimum) of that edge. In one embodiment requiring the most stringent algorithm performance, the threshold is set to 77% of the right extreme of the edge. In another embodiment with less stringent algorithm performance, the threshold is set to 200% of the right extreme of the edge. The max-min check **560** is effective in eliminating edges that are pulse-like but correspond to motion.

FIG. **10** illustrates an example of the max-min check **560** (FIG. **5**). Shown is the processed waveform **1060**. The max-min check **560** is applied to edge b **1070**. The interval B-C is considered, which is between point B **1074**, the peak of edge b **1070**, and point C **1084**, the peak of edge c **1080**. The maximum in the interval B-C is point C **1084**. Point C **1084** is compared to a first threshold **1078**, which in this example is 77% of the absolute value of point P**1** **1072**, the minimum of edge b **1070**. Edge b **1070** would not be discarded because point C **1084** is smaller than this first threshold **1078**. As another example, the max-min check **560** is applied to edge c **1080**. The interval C-D is considered, which is between point C **1084**, the peak of edge c **1080**, and point D **1094**, the peak of edge d **1090**. The maximum in the interval C-D is point V **1093**. Point V **1093** is compared to a second threshold **1088**, which is 77% of the absolute value of point P**2** **1082**, the valley of edge c **1080**. Edge c would be discarded because point V **1093** is greater than this second threshold **1088**.

As shown in FIG. **5**, the pulse finder **570** is the last component of the candidate pulse subprocessor **410**. The pulse finder **570** transforms the edges remaining after the various edge checks into candidate pulses in the form of an idealized triangular wave, which are fed into the plethysmograph model subprocessor **460** (FIG. **4**). From the information about the indices of the peaks of valleys of the remaining edges, it is simple to determine a pulse in the input waveform. The remaining edges are first divided into edge pairs, i.e. the first and second edges, the second and third edges, and so on. The first point of a pulse corresponds to the maximum of the waveform segment in the interval of indices determined by the peak and valley of the first edge of a pair. The second point is the minimum between the valley of the first edge and the peak of the second edge. The third and last point is the maximum between the peak and the valley of the second edge.

FIG. **11** illustrates the result of the pulse finder **570** (FIG. **5**) shown as a series of pulses **1110**, including a particular pulse XYZ **1120** appearing as a triangular wave superimposed on an input waveform segment **1140**. Also shown are the remaining edges a **1170**, b **1180** and c **1190**. In this example, pulse XYZ **1120** is formed from the pair of edges c **1180** and e **1190**. Point X **1122** is the maximum in the waveform segment **1140** in the time interval between the peak **1182** and valley **1184** of edge c **1180**. Point Y **1124** is the minimum in the waveform segment **1140** in the time interval between the valley **1184** of edge c **1180** and the peak **1192** of edge e **1190**. Point Z **1128** is the maximum in the waveform segment **1140** in the time interval between the peak **1192** and valley **1194** of edge e **1190**.

FIG. **12** illustrates the components of the plethysmograph model subprocessor **460**. This subprocessor takes as input the candidate pulses identified by the candidate pulse subprocessor **410** (FIG. **4**) and decides which of these satisfies an internal model for a physiological plethysmographic waveform. Although the candidate pulse subprocessor **410** (FIG. **4**) performs a series of checks on edges, the plethysmograph model subprocessor performs a series of checks on pulse features. The first component of the model subprocessor calculates relevant pulse features. The remainder of the model subprocessor checks these pulse features to identify physiologically acceptable features.

Shown in FIG. **12**, the pulse features component **1210** extracts three items of information about the input candidate pulses that are needed for downstream processing by the other components of the model subprocessor. The extracted features are the pulse starting point, period and signal strength.

FIG. **13** illustrates a candidate pulse **1300** and the three parameters extracted by the pulse features component **1210** (FIG. **12**). The pulse **1300** is shown overlaid on the input waveform **1302** for reference. The starting point A **1360** is the first peak of the pulse **1300**. The period P **1370** is the time difference between the time of occurrence of the first peak **1360** and the second peak **1362** of the pulse **1300**. The signal strength SS **1350** is the difference between the values of the first peak **1360** and the valley **1364** of the pulse **1300**. The signal strength SS **1350** is normalized by dividing this value by the value of the infrared raw signal data at the point corresponding to point A **1360**.

Also shown in FIG. **12** is the 250 BPM check **1220**. This component discards pulses having a period P **1370** (FIG. **13**) that is below 15 samples. This corresponds to an upper limit for the pulse rate set at 250 beats per minute. That is:

$$15\ \text{samples/beat}=(62.5\ \text{samples/sec.}\times 60\ \text{sec./min.})/250\ \text{beats per min} \qquad (2)$$

In addition, FIG. **12** shows the stick model check **1230**. This component discards pulses where the corresponding waveform does not closely fit a stick model, i.e. where a pulse cannot be represented by a triangular waveform. This component measures a normalized difference between the input waveform and the triangular wave representation of that waveform. The obtained value is compared to a threshold, and pulses are discarded where the normalized difference is greater than that threshold.

FIG. **14** illustrates the calculations performed by the stick model check **1230** (FIG. **12**). Shown is an input waveform pulse **1410** and the corresponding stick model pulse **1460**.

The stick model check **1230** (FIG. **12**) component computes a first value, which is a sum of the absolute differences, shown as the dark black areas **1420**, between the waveform pulse **1410** and the stick model pulse **1460**. This component also computes a second value, which is a sum of the first rectangular gray area **1470** enclosing the descending portion of the pulse **1410** and the second gray area **1480** enclosing the ascending portion of the pulse **1410**. The stick model check **1230** (FIG. **12**) then normalizes the first value by dividing it by the second value. This normalized value is compared with a threshold. A physiological pulse does not differ too much from the stick model at high pulse rates. This is not true at pulse rates much below 150 bpm because of the appearance of a dicrotic notch and other "bumps." Hence, the threshold is a function of pulse rate. In one embodiment, the threshold is:

$$0.15,\ \text{for pulse rate}<130 \qquad (3)$$

$$0.430455769e^{-0.008109302(\text{pulse rate})},\ \text{for } 130<\text{pulse rate}<160 \qquad (4)$$

$$0.1,\ \text{for pulse rate}>160 \qquad (5)$$

Shown in FIG. **12** is the angle check **1240**. The angle check **1240** is based on computing the angle of a normalized slope for the ascending portion of a pulse. This angle is compared with the same angle of an ideal pulse having the same period. This check is effective in discarding pulses that are extremely asymmetric.

FIG. **15** illustrates an example of the angle check **1240** (FIG. **12**). Shown is a single triangular pulse **1500** superimposed on the corresponding input waveform **1502**. The ascending pulse portion **1504** has a vertical rise a **1510** and a horizontal run b **1520**. The rise **1510** and run **1520** are normalized with respect to the pulse signal strength ss **1530** and the pulse frequency, which is 62.5 Hz. in this particular embodiment. An angle θ 1540 is computed as:

$$\theta=\arctan\left[(a/ss)/(b/62.5)\right]\times 180/\pi \qquad (6)$$

The angle θ is compared with the same angle of an ideal pulse having the same period, where a is equal to the signal strength and b is equal to the period c **1550** minus 6. Three degrees are added to this value as a threshold margin. Hence, θ is compared to θ_{ref} computed as follow:

$$\theta_{ref}=\arctan\left\{[a/ss]/[(c-6)/62.5]\right\}\times(180/\pi)+3 \qquad (7)$$

If $\theta<\theta_{ref}$, then the pulse is discarded. FIG. **16** illustrates an example pulse **1600** that would be discarded by the angle check, because the segment a **1610** is much smaller than the signal strength ss **1630**.

Also shown in FIG. **12** is the ratio check **1250**. The time ratio check component removes pulses in which the ratio between the duration of the ascending pulse portion and the duration of the descending pulse portion is less than a certain threshold. In a particular embodiment, the threshold is 1.1. The rationale for this check is that in every physiological pulse the ascending portion is shorter in time than the descending portion, which represents the ventricular contraction.

FIG. **17** illustrates an example pulse **1700** that would be discarded by the time ratio check **1250** (FIG. **12**). In this example, the duration a **1760** of the ascending portion **1710** is less than the duration b **1770** of the descending portion **1720**. Hence, the ratio of the ascending duration **1760** to the descending duration **1770**, a/b, is less than the threshold 1.1.

FIG. **12** further shows the signal strength check **1260**. The signal strength check **1260** assigns a confidence value to each pulse, based on its signal strength. There are two levels of confidence, high and low. The determination of confidence is based on two mechanisms. The first mechanism is founded on the observation that the higher the pulse rate, the lower the signal strength. This mechanism is implemented with an empirical relationship between pulse rate and signal strength. If the measured signal strength is greater than this empirical relationship by a fixed margin, the pulse confidence is low. The second mechanism incorporates the physiological limitation that signal strength cannot change too much over a short period of time. If the pulse signal strength is greater than a short-term average signal strength by a fixed margin, the pulse confidence is low. If the pulse meets both criteria, then the pulse has a high confidence. All pulses in a single waveform segment or snapshot have the same confidence value. Hence, if there is a least one pulse with a high confidence, then all pulses with a low confidence will be dropped.

FIG. **18** illustrates the first signal strength criteria described above. In one embodiment, the relationship between signal strength and pulse rate is given by curve **1800**, which is described by the following equation:

$$SS=110\cdot e^{-0.02131PR}+1 \quad (8)$$

First, the pulse rate, PR **1810**, is determined from the pulse period. Next, the corresponding signal strength, SS_{ref} **1820**, is determined from equation (8) and the pulse rate **1810**. Because equation (8) is empirically derived, it is shifted up and down to make it more applicable for individual patients. A long-term average signal strength, Long Time SS **1830**, and a long-term average pulse rate, Long Time PR **1840**, are derived. If Long Time SS **1830** is above the curve **1800** at the point corresponding to the Long Time PR **1840**, then the difference between the Long Time SS and the curve **1800** plus 2 becomes Offset **1850**. If the measured pulse signal strength, Pulse SS, is less than SS_{ref}+Offset **1860**, then this check is passed.

As shown in FIG. **4**, after the candidate pulse subprocessor **410** and the plethysmograph model subprocessor **460**, the pulse recognition processor **400** has identified inside the input waveform snapshot all of the pulses that meet a certain model for physiologically acceptable plethysmographs. From the information about these pulses, the pulse statistics subprocessor **490** can extract statistics regarding the snapshot itself. Two useful statistical parameters that are derived are the median value of the pulse periods and signal strengths. The median is used rather than the mean because inside a waveform snapshot of 400 points (almost 7 seconds) the period and signal strength associated with each pulse can vary widely. Another parameter is the signal strength confidence level, which in one embodiment is the same for all the recognized pulses of a snapshot. A fourth useful parameter is pulse density. Pulse density is the value obtained by dividing the sum of the periods of the acceptable pulses by the length of the snapshot. Pulse density represents that ratio of the snapshot that has been classified as physiologically acceptable. Pulse density is a value between 0 and 1, where 1 means that all of the snapshot is physiologically acceptable. In other words, pulse density is a measure of whether the data is clean or distorted, for example by motion artifact.

Finally, based on these described criteria, a pulse rate may be chosen. In a system with additional monitoring inputs, as depicted in FIG. **19**, a pulse rate selection and comparison module **1900** may be provided. For example, the oximeter pulse rate (and corresponding confidence information if desired) can be provided on a first input **1902**. In a multi-parameter patient monitor, there may also be pulse rate or pulse information (and possibly confidence information) from an ECG or EKG monitor on a second input **1904**, from a blood pressure monitor on a third input **1906**, from an arterial line on a fourth input **1908**, and other possible parameters **1910**, **1912**. The pulse rate module **1900** then compares the various inputs, and can determine which correlate or which correlate and have the highest confidence association. The selected pulse rate is then provided on an output **1914**. Alternatively, the pulse rate module **1900** may average each input, a selection of the inputs or provide a weighted average based on confidence information if available.

The plethysmograph pulse recognition processor has been disclosed in detail in connection with various embodiments of the present invention. These embodiments are disclosed by way of examples only and are not to limit the scope of the present invention, which is defined by the claims that follow. One of ordinary skill in the art will appreciate many variations and modifications within the scope of this invention.

What is claimed is:

1. A method of determining a pulse rate measurement of a monitored patient from a signal responsive to light absorption by tissue of the monitored patient, said method comprising:

receiving data from a plurality of sensors including a noninvasive optical sensor, wherein said data is responsive to light attenuated by tissue;

electronically processing said data using an electronic signal processor including:

identifying candidate pulses from the received data based on a triangular wave model;

extracting pulse features from the identified candidate pulses;

determining physiologically acceptable pulses from the identified candidate pulses based on the extracted pulse features;

extracting one or more pulse statistics from the determined physiologically acceptable pulses, wherein said one or more pulse statistics represent a confidence associated with the physiologically acceptable pulses;

selecting a pulse rate from pulse measurements derived from the plurality of sensors based on the extracted one or more pulse statistics representing the confidence associated with the physiologically acceptable pulses; and

displaying the selected pulse rate,

wherein said one or more pulse statistics comprise pulse density.

2. The method of claim **1**, wherein said pulse statistics comprise at least one of median value of pulse periods, signal strength of pulses, and signal strength confidence level.

3. A system for determining a rate measurement of a monitored patient from a signal responsive to light absorption by tissue of a monitored patient, said system comprising an electronic signal processor configured to:

receive data from a plurality of sensors including a noninvasive optical sensor, wherein said data is responsive to light attenuated by tissue, wherein said tissue may vary in optical density over time due to volumetric changes;

identify candidate pulses from the received data based on a triangular wave model;

extract pulse features from the identified candidate pulses;

determine physiologically acceptable pulses from the identified candidate pulses based on the extracted pulse features;

calculate one or more pulse statistics of the determined physiologically acceptable pulses;

select a pulse rate from pulse measurements derived from the plurality of sensors based on the calculated one or more pulse statistics of the determined physiologically acceptable pulses; and

display the selected pulse rate,

wherein the one or more pulse statistics comprise pulse density.

4. The system of claim **3**, wherein the pulse features comprise at least one of the following: pulse starting point, period, and signal strength.

* * * * *

专利名称(译)	体积描记器脉冲识别处理器		
公开(公告)号	US9675286	公开(公告)日	2017-06-13
申请号	US13/196220	申请日	2011-08-02
[标]申请(专利权)人(译)	戴铂MOHAMEDķ		
申请(专利权)人(译)	戴铂MOHAMED K.		
当前申请(专利权)人(译)	Masimo公司		
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发明人	DIAB, MOHAMED K.		
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其他公开文献	US20110288383A1		
外部链接	Espacenet USPTO		

摘要(译)

基于时域规则的处理器提供脉冲血氧计导出波形中的脉冲识别。