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**Brockway et al.**

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- (54) **METHOD AND APPARATUS FOR IDENTIFYING CARDIAC RISK**
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*A61B 5/0452* (2006.01)  
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*G06F 17/14* (2006.01)  
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- (52) **U.S. Cl.**  
CPC ..... *A61B 5/0452* (2013.01); *A61B 7/00* (2013.01); *A61B 5/0006* (2013.01); *A61B 5/0245* (2013.01); *A61B 5/04017* (2013.01); *A61B 5/04087* (2013.01); *G06F 17/14* (2013.01); *G06K 9/0051* (2013.01); *G06K 9/0053* (2013.01); *H03H 17/0248* (2013.01); *G06F 19/3431* (2013.01); *A61B 5/7253* (2013.01)

- (58) **Field of Classification Search**  
None  
See application file for complete search history.

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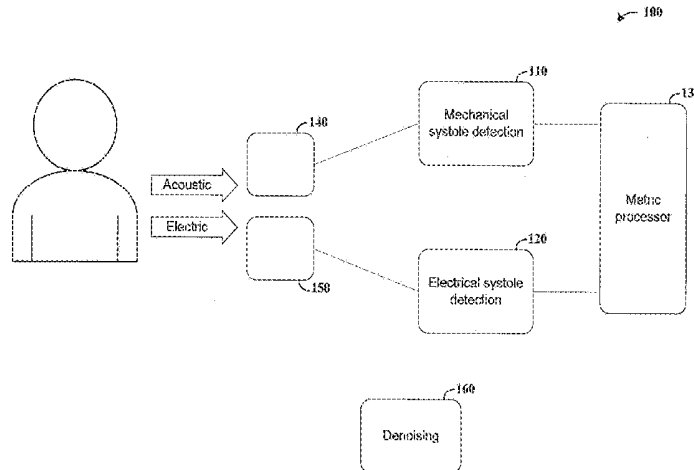
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(57) **ABSTRACT**

A cardiac-based metric is computed based upon characteristics of a subject's cardiac function. In accordance with one or more embodiments, the end of a mechanical systole is identified for each of a plurality of cardiac cycles of a subject, based upon an acoustical vibration associated with closure of an aortic valve during the cardiac cycle. The end of an electrical systole of an electrocardiogram (ECG) signal for each cardiac cycle is also identified. A cardiac-based metric is computed, based upon a time difference between the end of the electrical systole and the end of the mechanical systole, for the respective cardiac cycles.

**20 Claims, 19 Drawing Sheets**



**Related U.S. Application Data**

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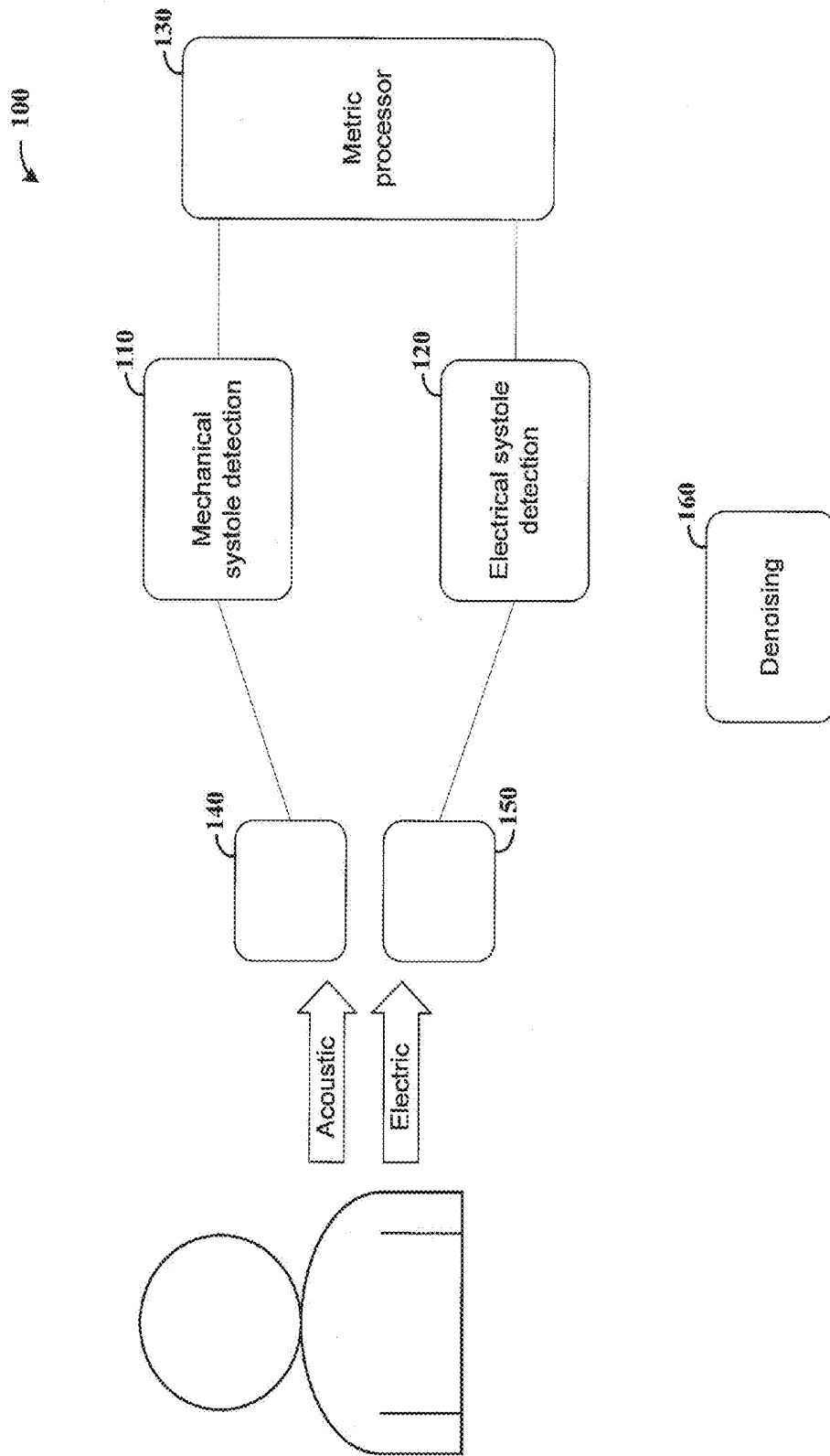


Figure 1A

Figure 1B

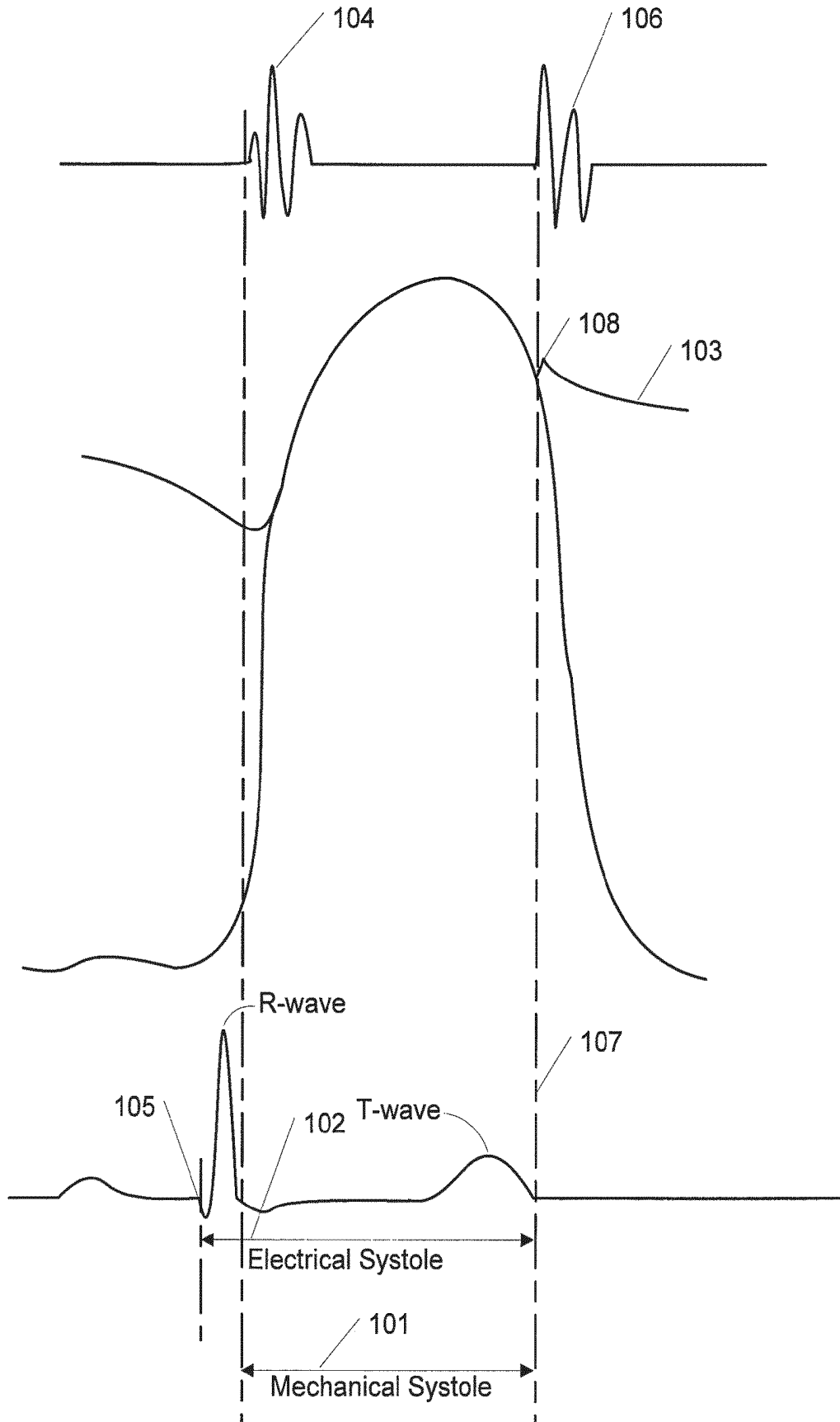


Figure 1C

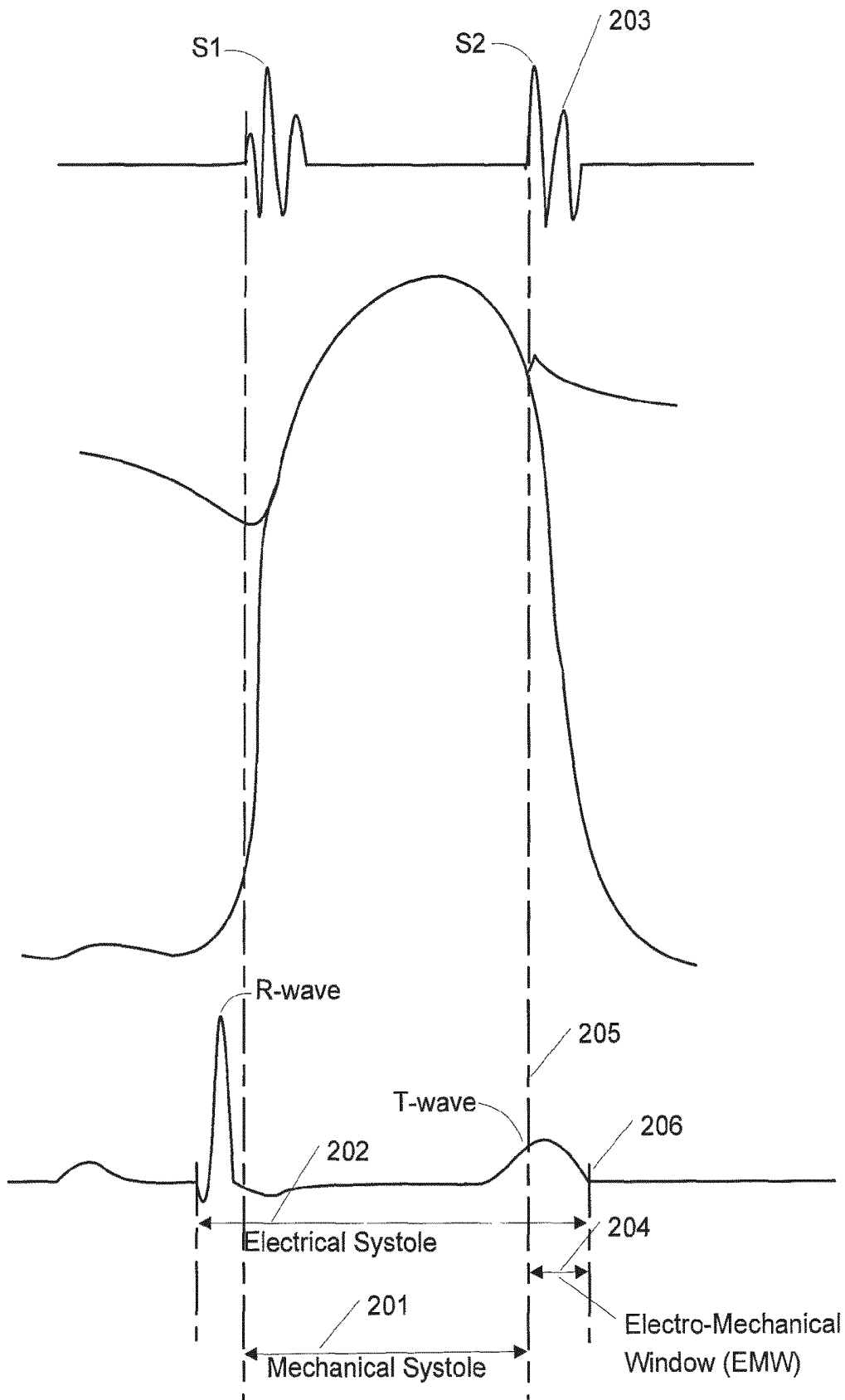


Figure 1D

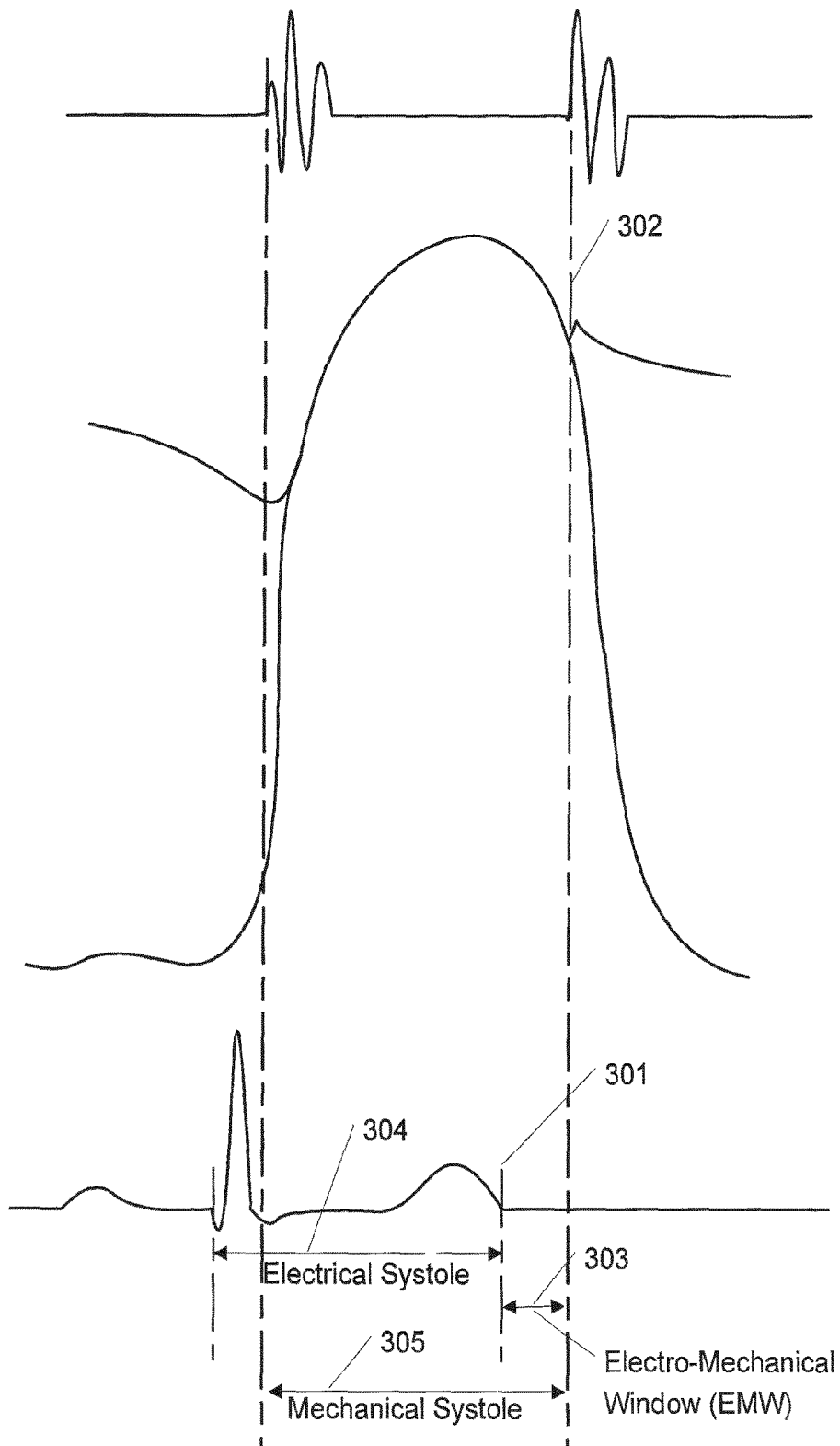
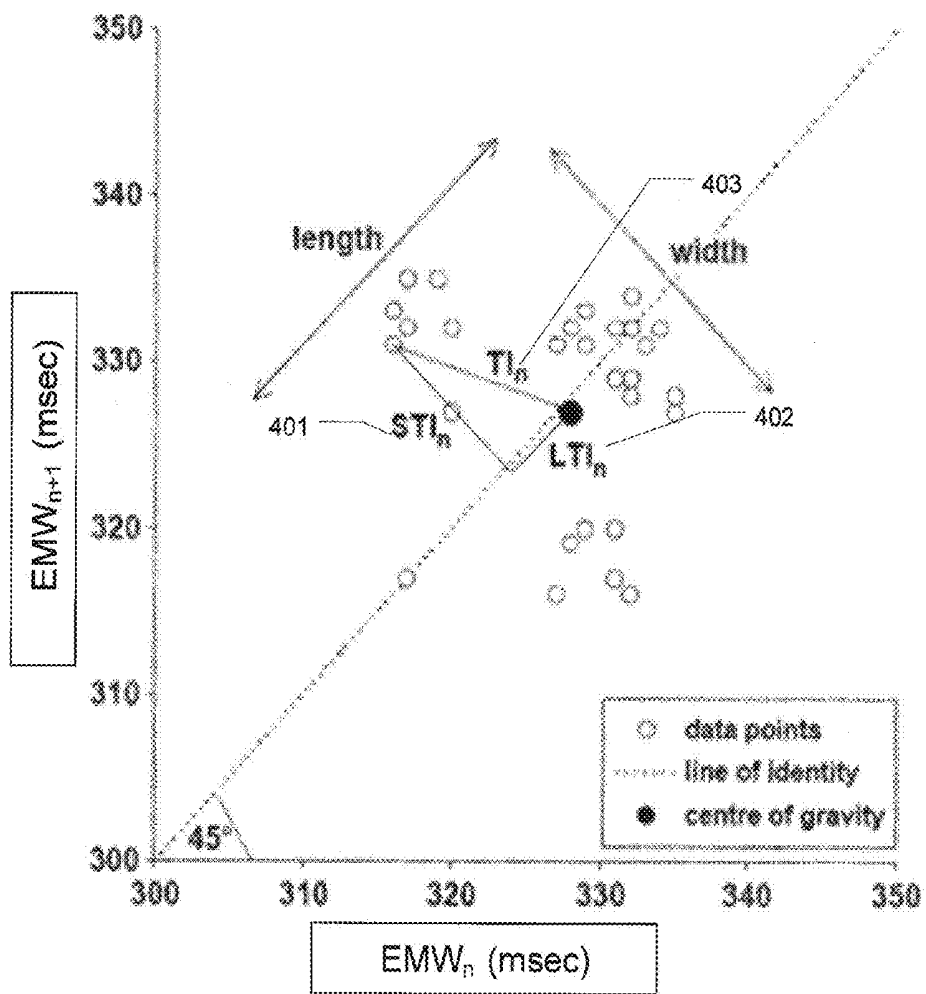


Figure 2



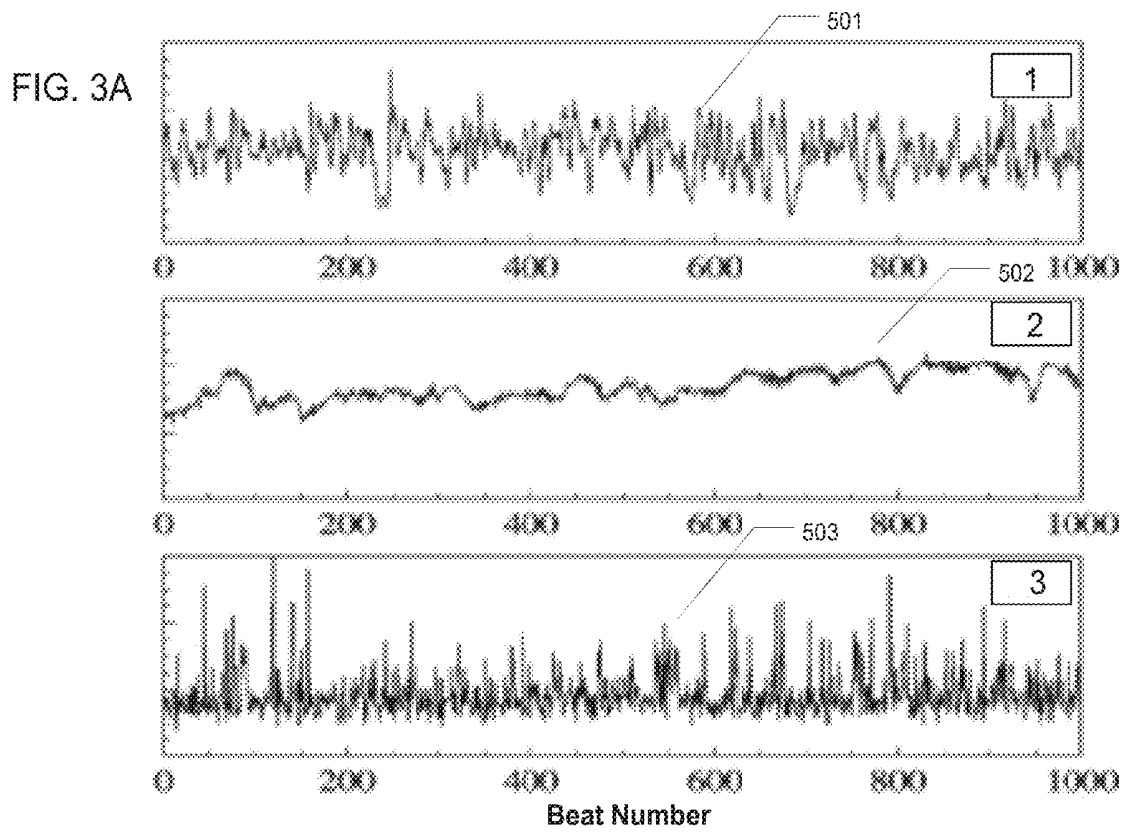
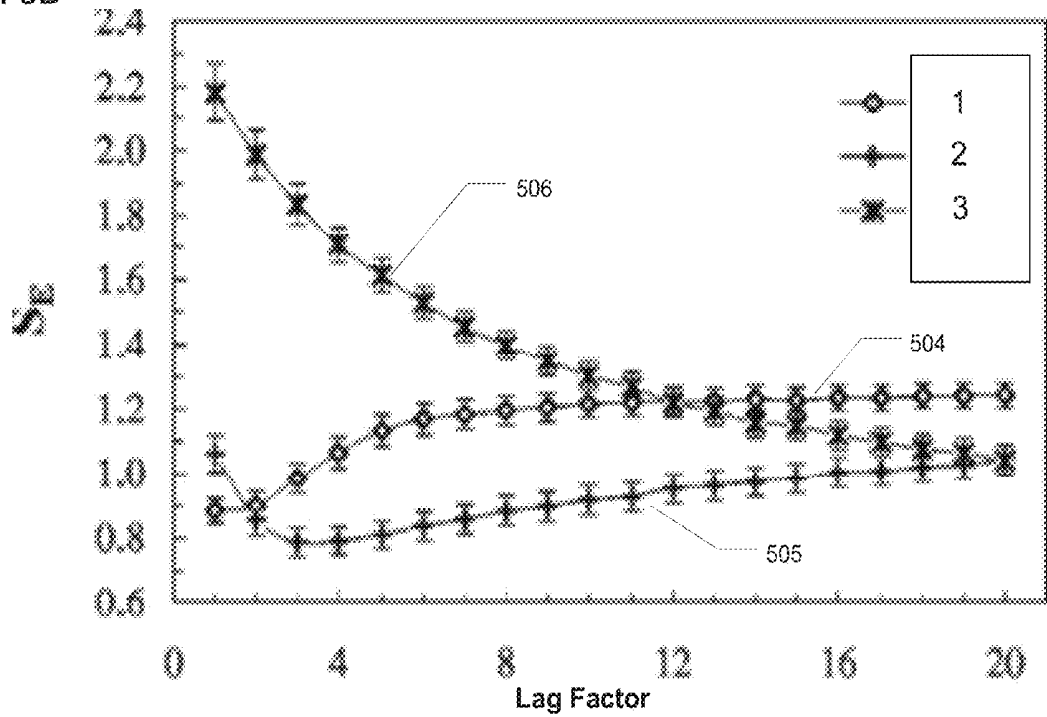


FIG. 3B



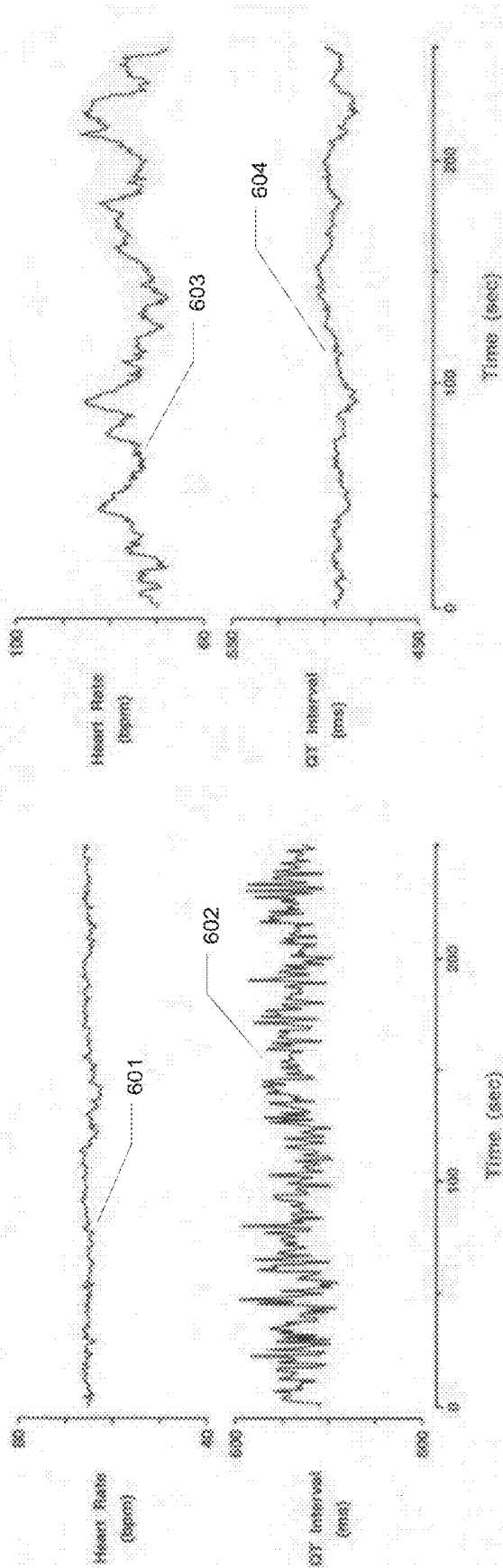


FIG. 4B

FIG. 4A

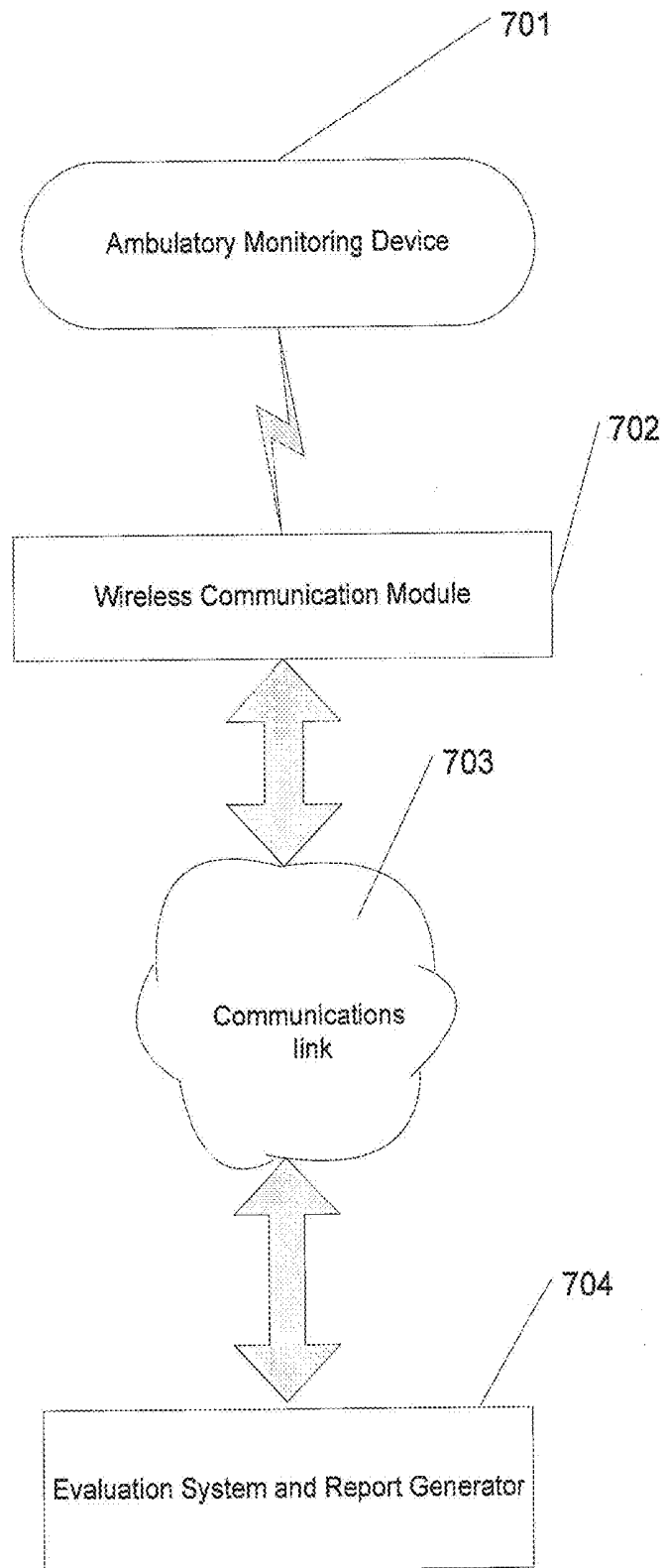


Figure 5

Figure 6

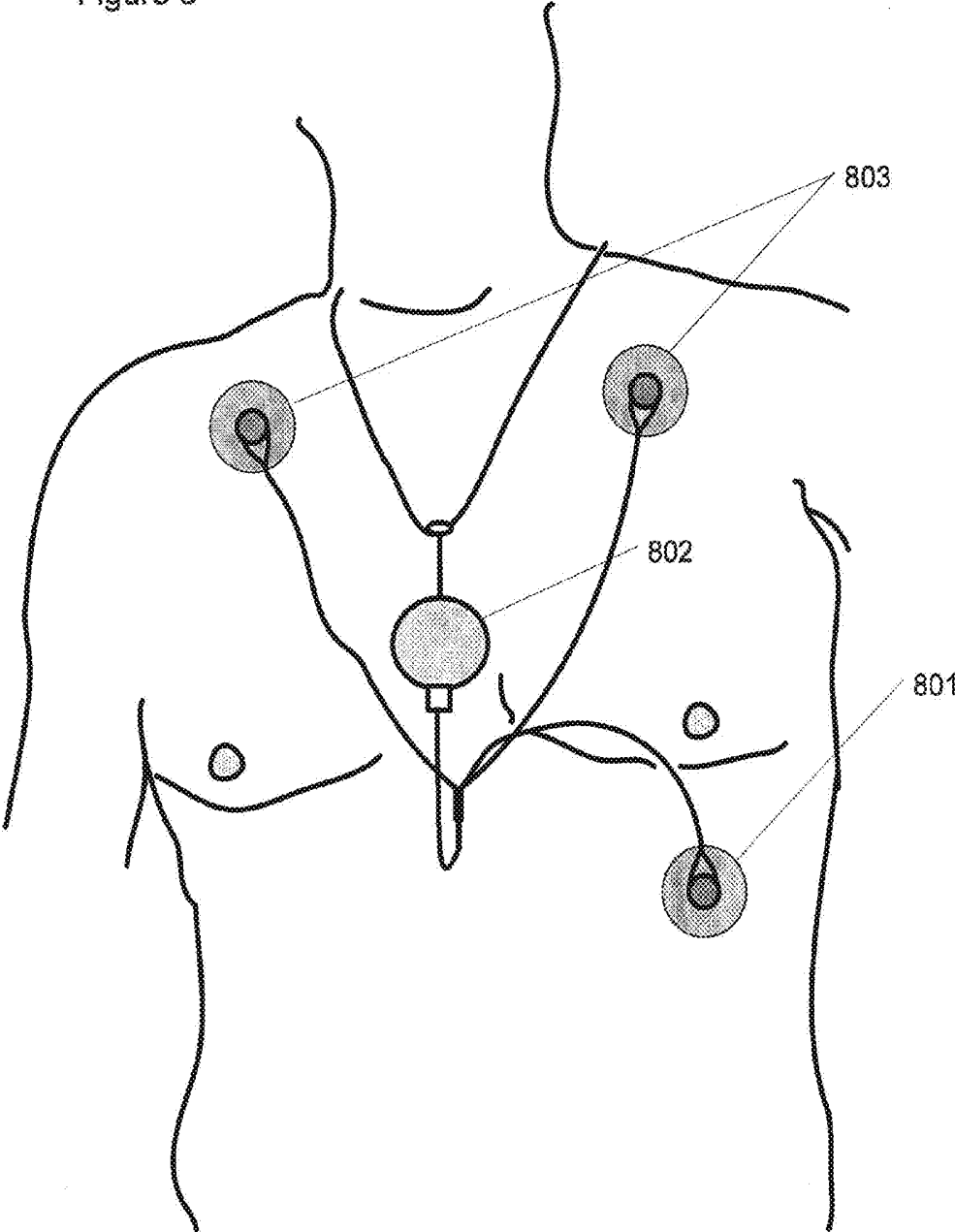


Figure 7

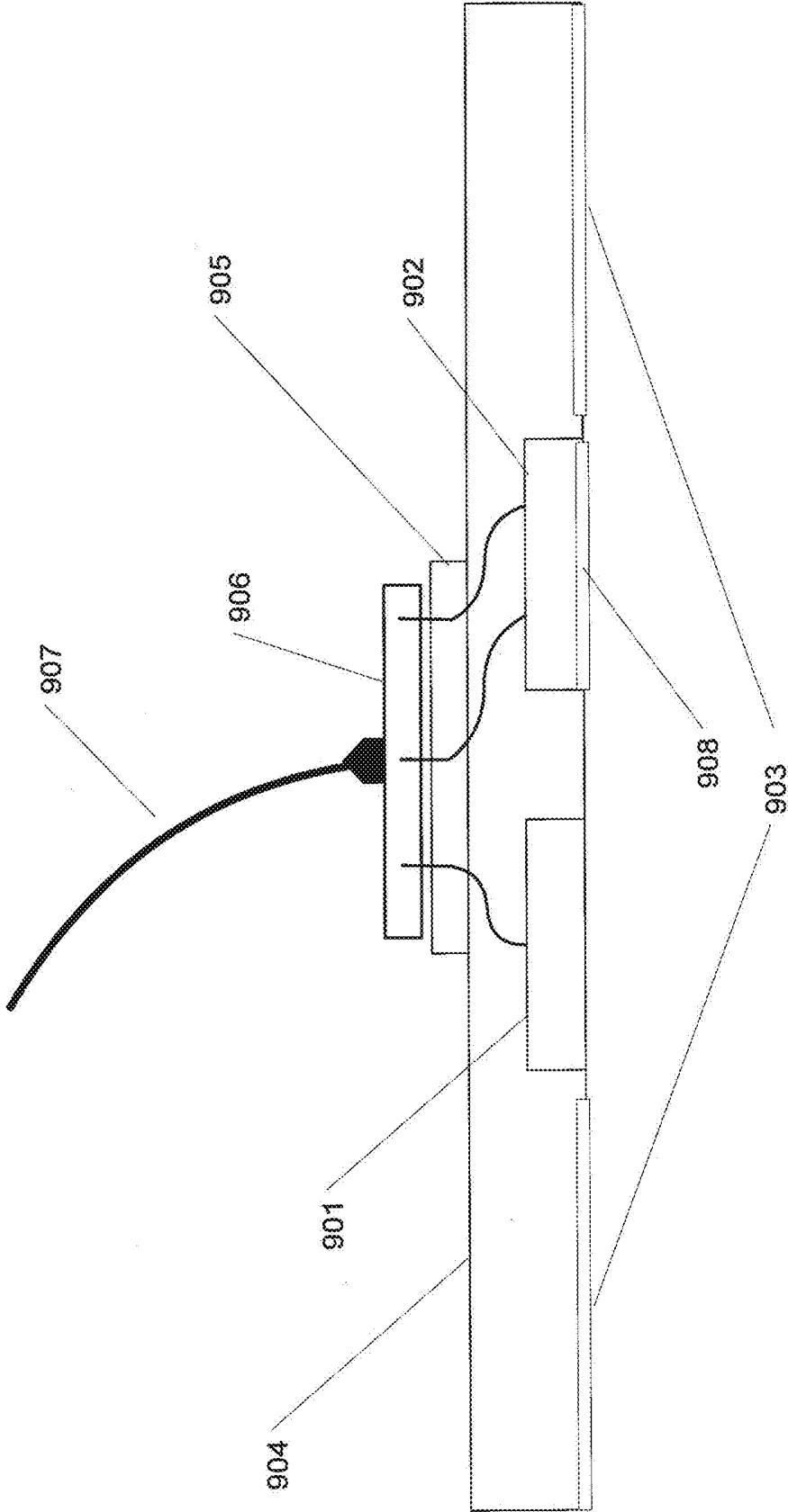


Figure 8

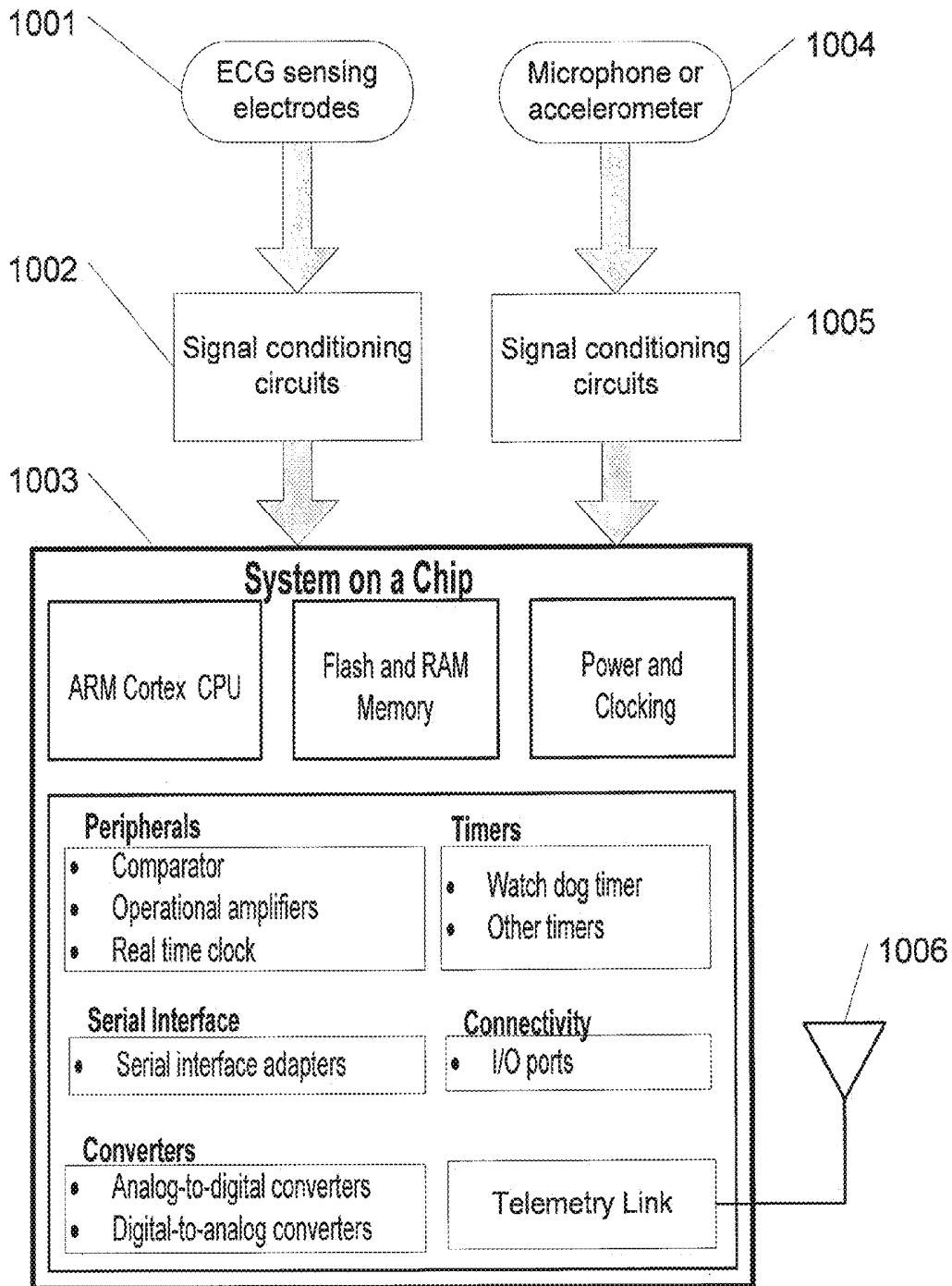


Figure 9

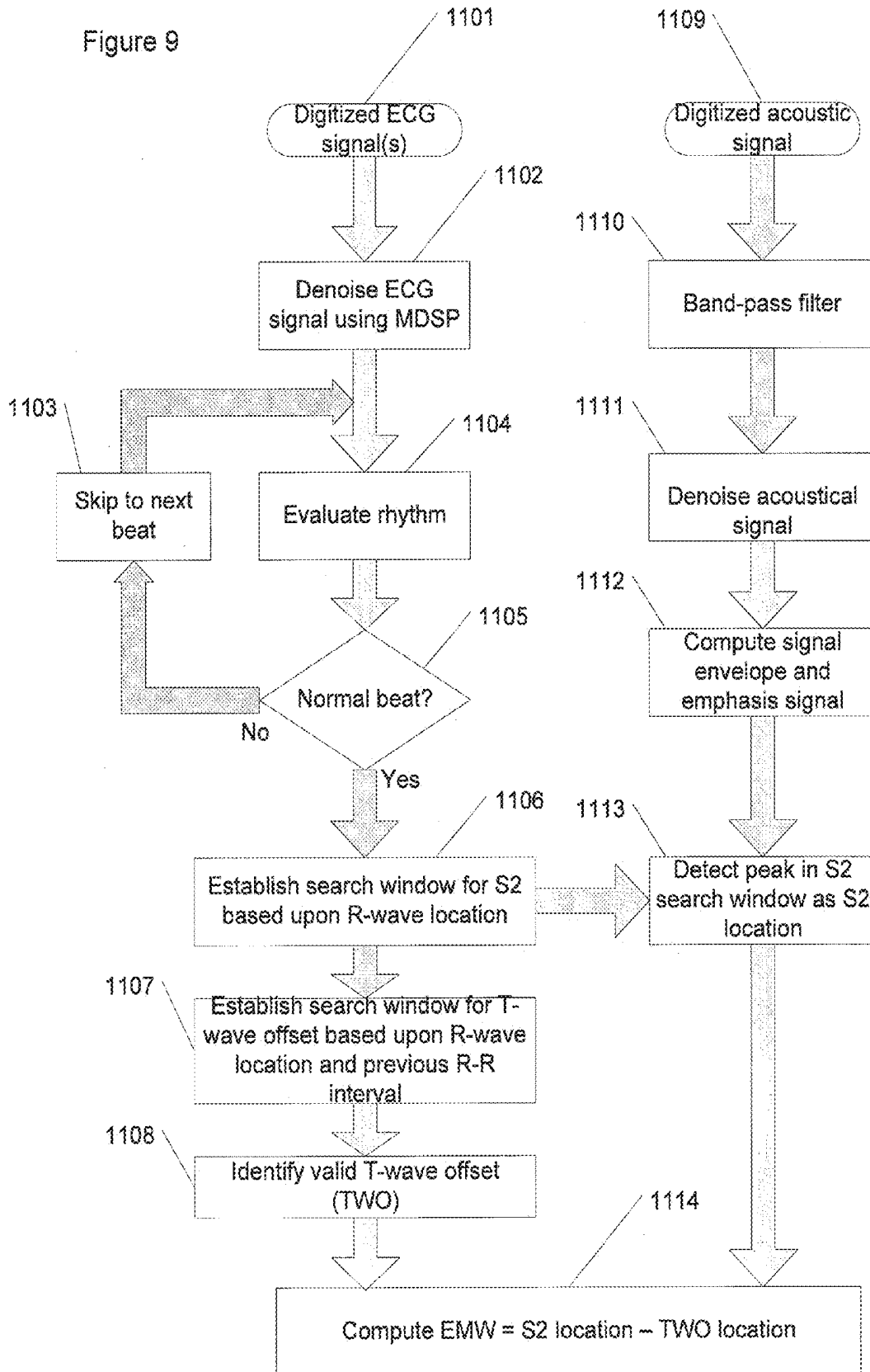


Figure 10

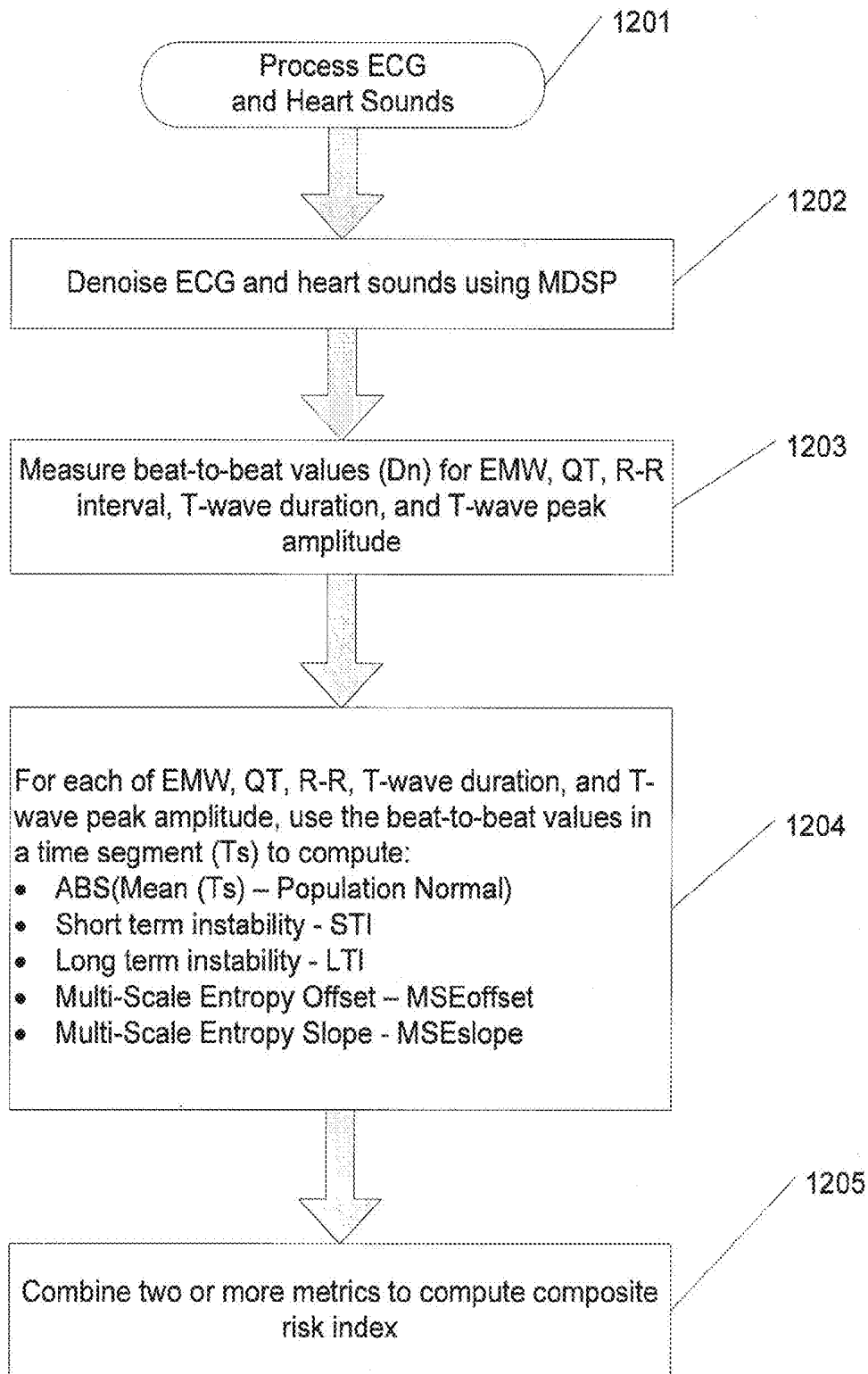


Figure 11

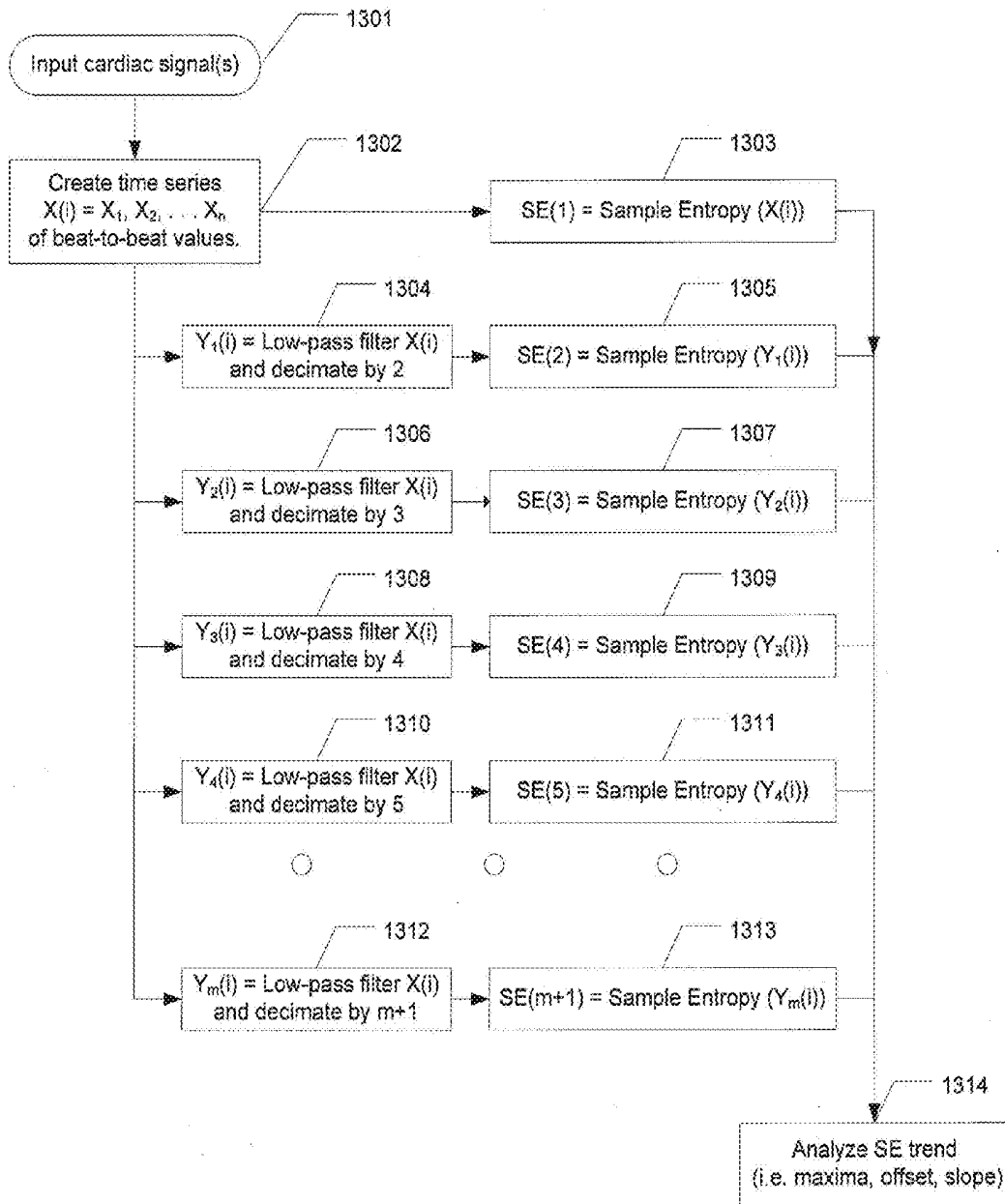
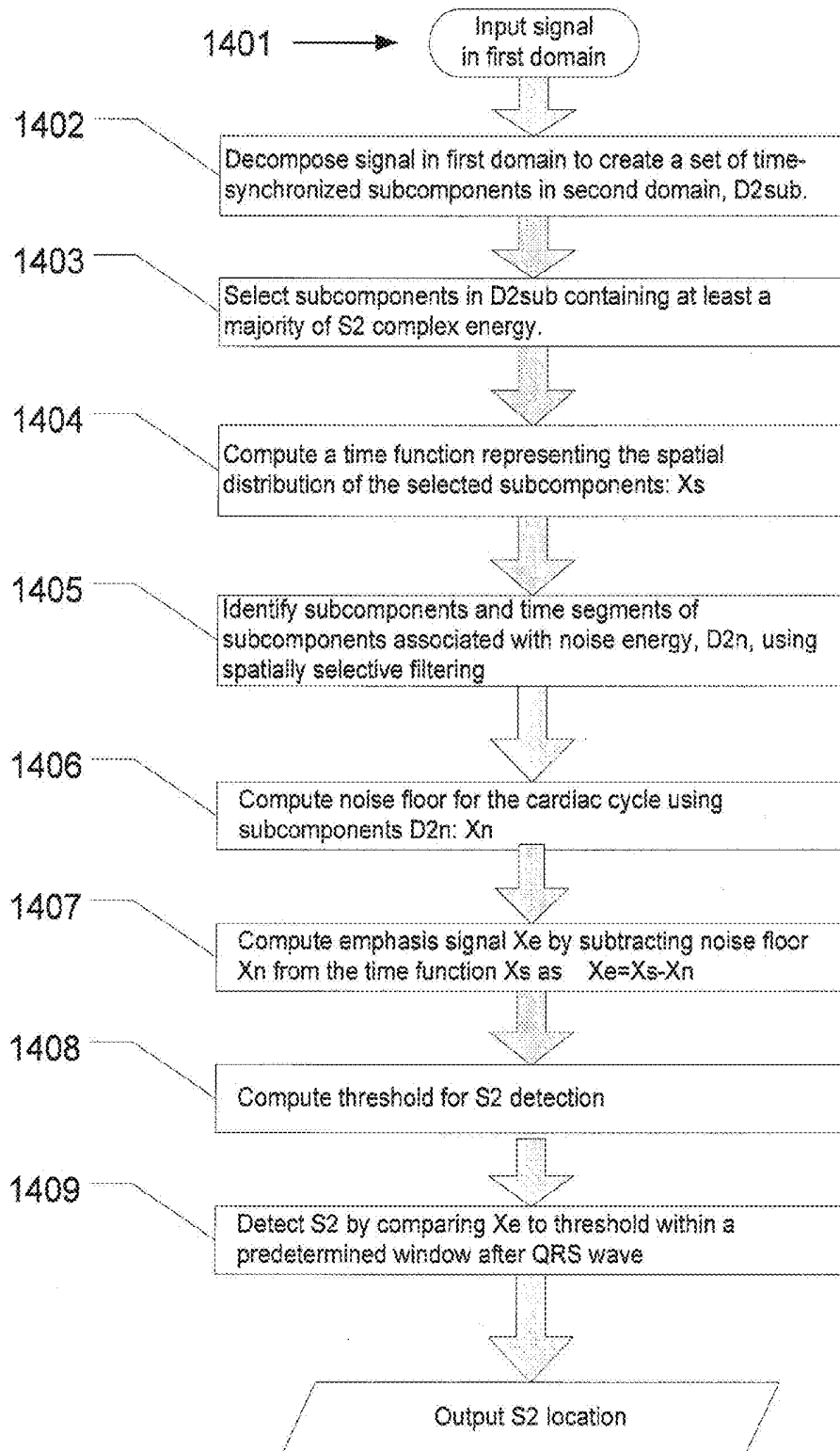


Figure 12



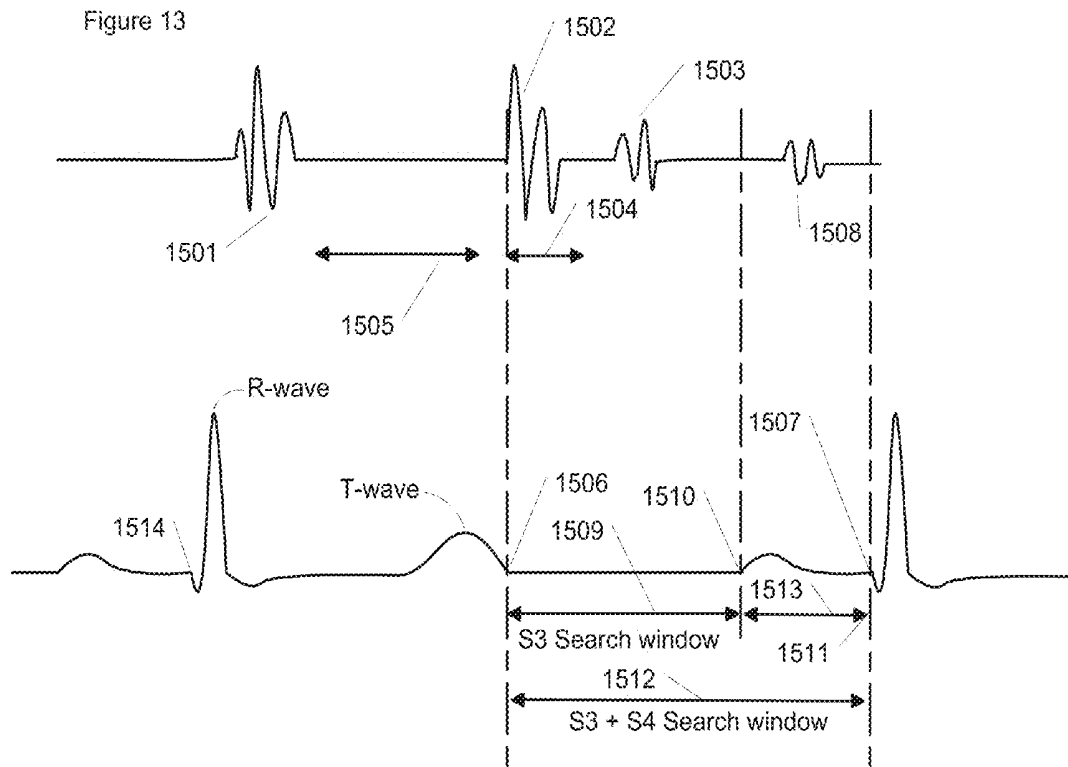


Figure 14

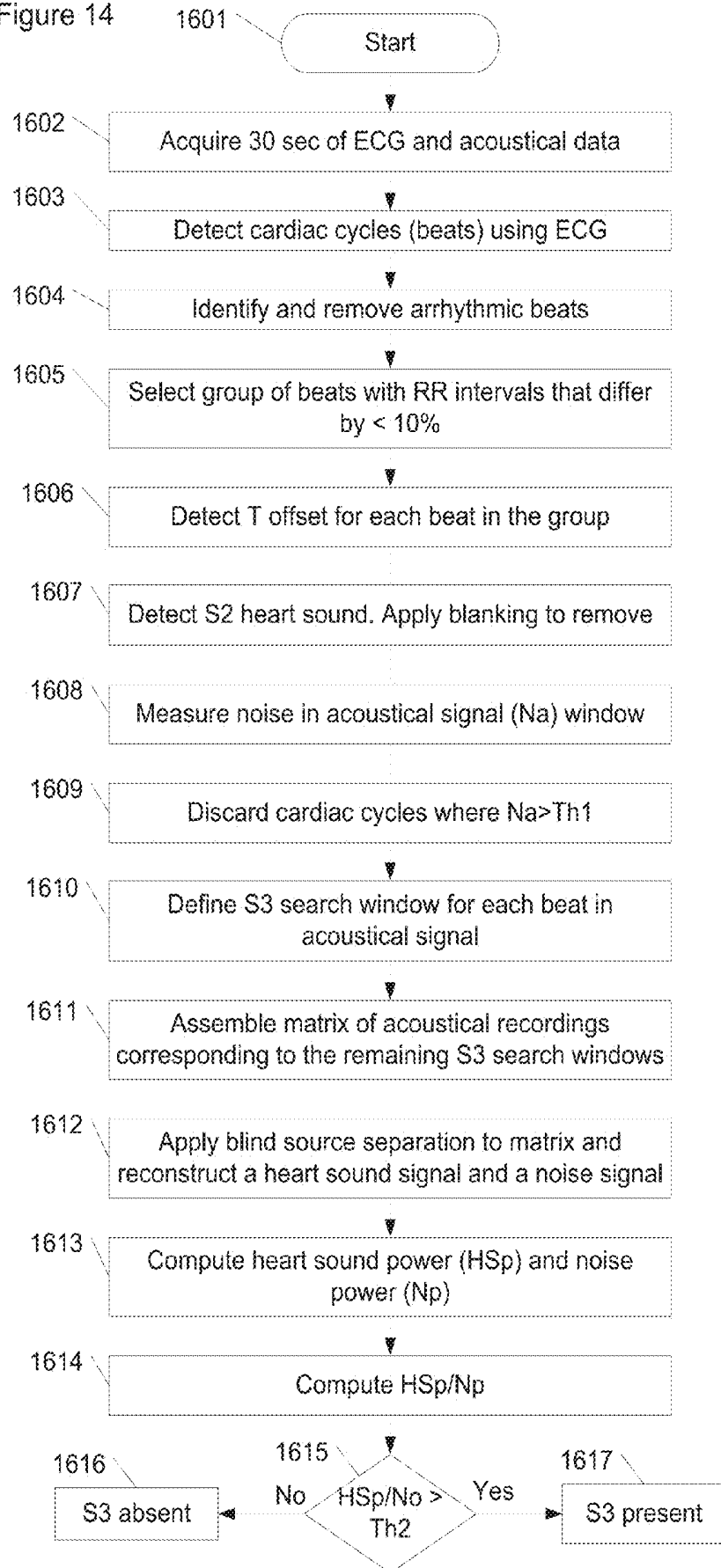
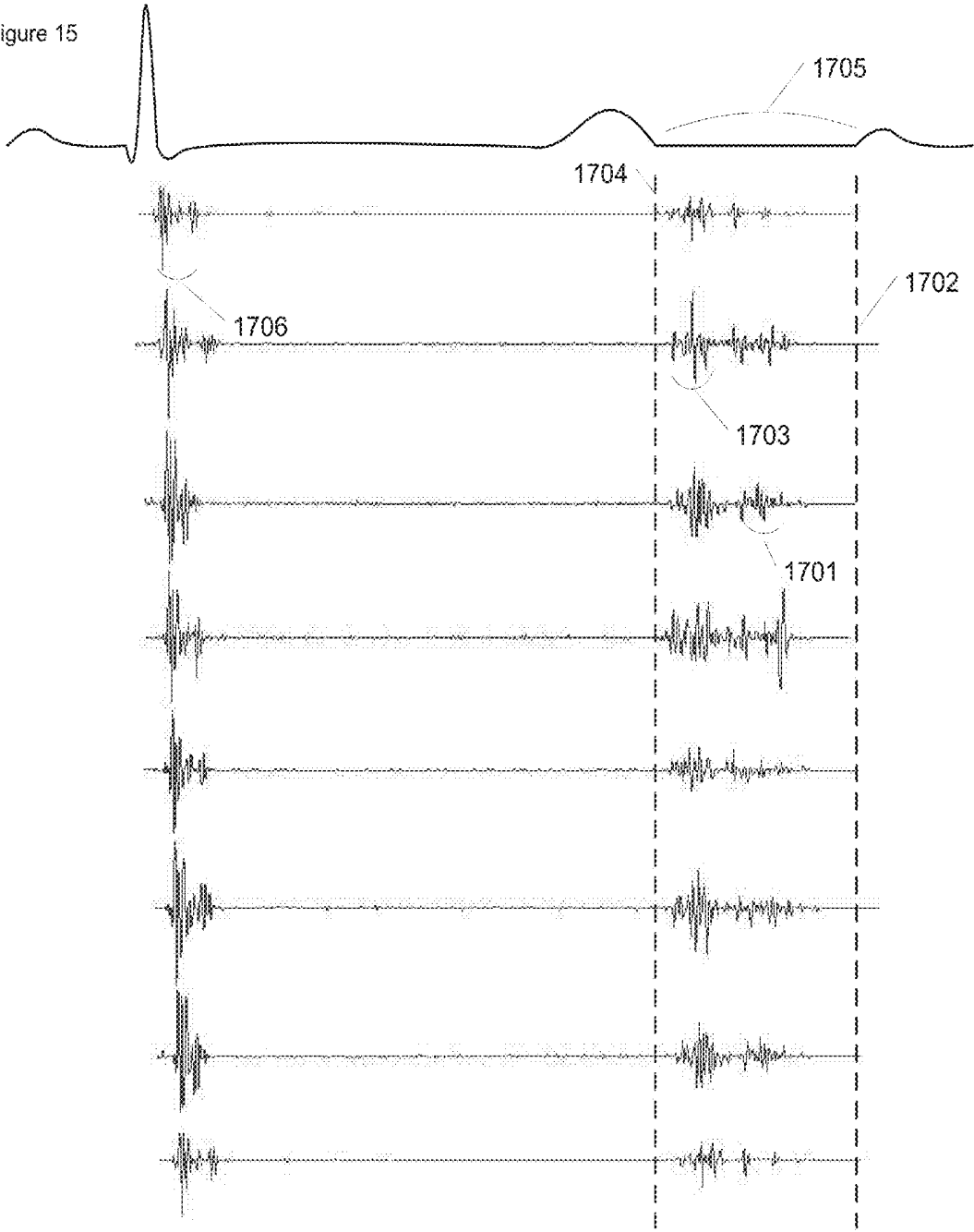
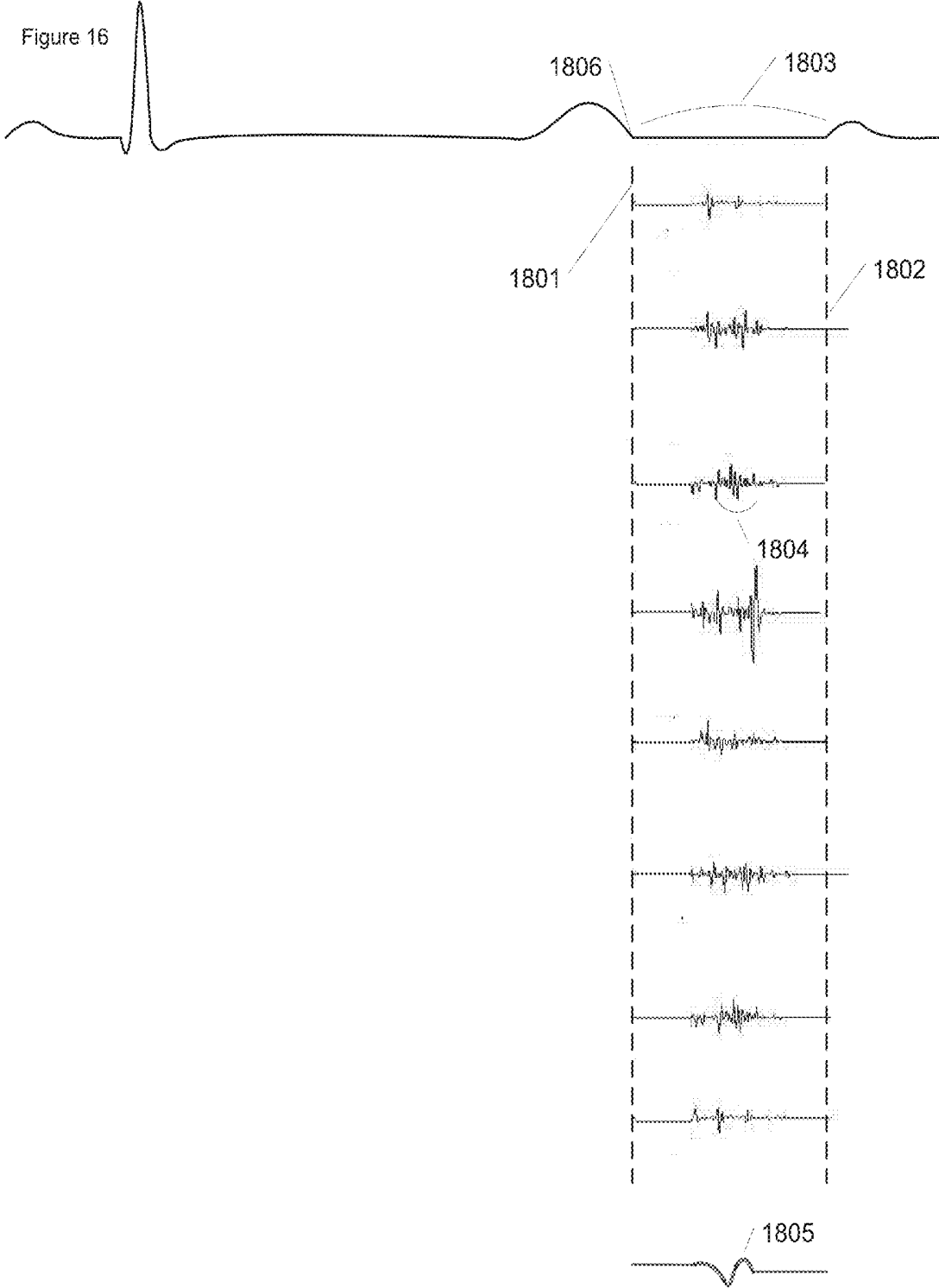


Figure 15





## METHOD AND APPARATUS FOR IDENTIFYING CARDIAC RISK

### FIELD

The present disclosure is related to characterizing cardiac-based function.

### BACKGROUND

Understanding the risk of arrhythmias, such as those that may stem from pharmaceuticals and cardiac pathologies, can be important in order to apply desirable and cost-effective therapeutic approaches and treat disease based upon patient-specific medical conditions and risks for developing a dangerous arrhythmia. For instance, understanding such risk can be helpful for patients diagnosed with cardiac diseases including heart failure and myocardial ischemia. The risk of arrhythmias is often assessed in both preclinical and clinical studies. For instance, the proarrhythmic risk of medications is often assessed in preclinical studies using several approaches. Clinical studies involving the QT interval of a cardiac cycle, such as those involving measurement of QT prolongation on healthy human subjects, can also be performed to assess the proarrhythmic risk of new medications.

However, such studies and assessments have been challenging to implement. It is often desirable to perform these assessments on ambulating human and animal subjects. However, performing these assessments on ambulatory subjects is difficult or impractical because either the required measurements are highly invasive or because the signals acquired using minimally invasive or non-invasive sensing techniques often result in signals that are sufficiently noisy that consistently accurate measurements are not possible. As evidence of these challenges, a significant percentage of pharmaceuticals that show no indication of proarrhythmic risk in preclinical studies eventually demonstrate evidence of proarrhythmic risk later in either development or post marketing. In addition, commonly used risk indicators are heart rate dependent and can hence be difficult to interpret. One of the unfortunate consequences of the lack of a reliable and sensitive cardiac risk metric is that preclinical studies sometimes falsely eliminate safe and effective drugs from the development pipeline based on metrics that have low predictive accuracy.

Techniques used to assess proarrhythmic risk in clinical care have also been challenging to implement in accurately assessing the risk of cardiac arrhythmias, such as for patients that have experienced myocardial infarction and those diagnosed with systolic heart failure and coronary artery disease. Unfortunately, the vast majority of deaths caused by dangerous arrhythmias occur in populations where existing techniques have proven ineffective and no practical and cost-effective options exist to accurately assess arrhythmic risk in these populations. Further, analyzing characteristics on ambulatory patients can be difficult. These and other characteristics have been challenging to the characterization of cardiac function, and risk associated therewith.

### SUMMARY

Various aspects of the present disclosure are directed to devices, methods and systems for assessing the risk of cardiac arrhythmias, in a manner that addresses challenges and limitations including those discussed above.

In accordance with one or more embodiments, a cardiac-based metric is computed for a subject as follows. The end of a mechanical systole is identified, for each of a plurality of

cardiac cycles of the subject, based upon an acoustical vibration associated with closure of an aortic valve during the cardiac cycle. The end of an electrical systole of an electrocardiogram (ECG) signal is also identified for the cardiac cycle. A time difference between the end of the electrical systole and the end of the mechanical systole for each of the plurality of cardiac cycles (e.g., collectively) is used to compute the cardiac-based metric. Using this approach, mechanical characteristics of the valve closing can be used together with electrical characteristics of the ECG, to provide an indication of cardiac function that measures electro-mechanical dysynchrony. In connection with this and other embodiments, it has been discovered that, by using this combined mechanical and electrical detection approach, challenges such as those above, as may be applied to measuring EMW, processing beat-to-beat information, and otherwise characterizing cardiac risk can be addressed. Further, the embodiments described here provide an approach that facilitates these cardiac electromechanical characteristics to be accurately measured on ambulating subjects.

In accordance with various example embodiments, mechanical and electrical dysynchrony is measured as the time difference (e.g., electro-mechanical window—EMW) between a point in a cardiac cycle that corresponds to the end of mechanical systole (MS) and a point that corresponds to the end of electrical systole (ES).  $EMW = \text{end of MS (MSend)} - \text{end of ES (ESend)}$ . In one embodiment, end of MS is identified by detecting the S2 heart sound. In one embodiment, end of ES is identified by detecting the end of the T-wave (i.e. T-wave offset). In one embodiment, short term and long term instability of the EMW or QT interval is computed to enhance the predictive value. In one embodiment, complexity of beat-to-beat dynamics of EMW or QT interval is quantified by computing multiscale entropy parameters and evaluating the trend of these parameters over multiple scales. In one embodiment EMW is combined with one or more of a) QRS duration, b) QT interval, c) short term QT variability and d) T-wave alternans to improve predictive value. In various embodiments, the S2 heart sound is sensed using a microphone or accelerometer and its occurrence is detected from the sensed signal using techniques such as those involving multi-domain signal processing (MDSP) techniques as discussed herein. For instance, the microphone or accelerometer can be integrated into an adhesive-backed ECG sensing electrode. In various embodiments T-wave offset is detected by denoising and processing an ECG, or by using MDSP techniques as discussed herein.

### BRIEF DESCRIPTION OF THE DRAWINGS

The disclosure may be more completely understood in consideration of the following detailed description of various embodiments in connection with the accompanying drawings, in which:

FIG. 1A shows an apparatus for characterizing a cardiac-based metric, in accordance with an example embodiment;

FIG. 1B shows a relationship between heart sounds, arterial blood pressure, left ventricular pressure, and ECG in a subject where EMW duration is <about 20 ms, as processed in connection with the apparatus **100** in FIG. 1A in connection with an example embodiment;

FIG. 1C shows a relationship between heart sounds, arterial blood pressure, left ventricular pressure, and ECG in a subject where EMW duration is longer than about 20 ms, as processed in connection with the apparatus **100** in FIG. 1A in connection with an example embodiment;

FIG. 1D shows a relationship between heart sounds, arterial blood pressure, left ventricular pressure, and ECG in a subject where EMW duration is negative, as processed in connection with the apparatus 100 in FIG. 1A in connection with an example embodiment;

FIG. 2 shows a relationship between EMW dynamics on successive cardiac cycles, consistent with an example embodiment;

FIG. 3A shows EMW dynamics in a normal heart, consistent with one or more example embodiments;

FIG. 3B shows EMW dynamics in a diseased heart, consistent with one or more example embodiments;

FIG. 4A shows an example of relationships between HR and QT dynamics in a diseased heart, consistent with one or more example embodiments;

FIG. 4B shows an example of relationships between HR and QT dynamics in a normal heart, consistent with one or more example embodiments;

FIG. 5 shows a block diagram of a system for evaluating arrhythmic risk, consistent with an example embodiment;

FIG. 6 shows a patient-worn component of a non-invasive system for evaluating arrhythmic risk, consistent with an example embodiment;

FIG. 7 shows a sensing element for attachment to the skin for sensing S2 heart sounds and ECG from a subject, consistent with an example embodiment;

FIG. 8 shows a block diagram of a patient-worn component of a system for evaluating arrhythmic risk, consistent with an example embodiment;

FIG. 9 shows a signal flow diagram for computing electro-mechanical window, consistent with an example embodiment;

FIG. 10 shows a block diagram for computing a composite metric of arrhythmic risk, consistent with an example embodiment;

FIG. 11 shows a signal flow diagram for computing and evaluating multiscale entropy of beat-to-beat values detected in a cardiac signal, consistent with another example embodiment;

FIG. 12 shows a flow diagram for detecting an S2 heart sound, in accordance with another example embodiment;

FIG. 13 provides an illustration of a timing relationship between heart sounds and ECG, as may be implemented in accordance with one or more embodiments;

FIG. 14 shows a signal flow diagram of a representative embodiment for processing an ECG and acoustical signal to detect S3 sounds, in accordance with another embodiment;

FIG. 15 shows an example of multiple segments of an acoustical recording containing S1, S2, and S3 heart sounds and their relationship in time with ECG, in accordance with another embodiment; and

FIG. 16 shows an illustration of multiple segments of an acoustical signal synchronized and assembled into an array for processing with S2 blanked, in accordance with another embodiment.

### DETAILED DESCRIPTION

Aspects of the present disclosure relate to methods and apparatuses involving measuring and detecting characteristics of cardiac function, such as alterations in cardiac function that precede the occurrence of arrhythmia or are indicative of an increased risk of arrhythmia. Certain aspects relate to methods and systems for measuring an electro-mechanical window (EMW) using heart sounds and evaluating beat-to-beat values of the EMW and other information such as QT interval to indicate arrhythmic risk. In some implementa-

tions, EMW is used to address a number challenges to implementing risk indicators as discussed above, by operating generally independent of heart rate.

In accordance with another example embodiment, a cardiac-based metric is computed using both mechanical and electrical systole for each of a plurality of cardiac cycles as follows. The end of a mechanical systole is detected using an acoustical vibration associated with closure of an aortic valve (e.g., the S2 heart sound) that occurs during the cardiac cycle.

The end of an electrical systole is also detected using electrocardiogram (ECG) signal for the cardiac cycle. The cardiac-based metric is computed using respective time differences between the end of the mechanical and electrical systoles for each of the plurality of cardiac cycles (e.g., by computing the time difference between the end of the electrical systole and the end of the mechanical systole for each cardiac cycle). These respective ends of the mechanical and electrical systole may, for example, be identified by processing signal data such as acoustical heart sound data and ECG data shown in and described in connection with figures below (e.g., identifying an S2 sound and a T-wave offset), within a computer-type circuit as described herein and using characteristics of the respective signals. Such a computer-type circuit can also be implemented to compute the cardiac-based metric using time differences collected from multiple cardiac cycles, and therein provide an indication of the collective differences as applicable to, for example, proarrhythmic risk.

The respective ends of the mechanical and electrical systoles are identified using one or more of a variety of approaches, in accordance with various example embodiments. For instance, the end of the mechanical systole can be identified using an acoustical signal containing energy associated with both the closure of the aortic valve and noise energy. In an embodiment, the acoustical signal is decomposed from a first domain into subcomponents of the acoustical signal in a second domain, and at least two of the subcomponents are identified as exhibiting an energy level of which at least half is associated with closure of the aortic valve. The identified subcomponents are mathematically combined to compute a time function that identifiably changes in value upon aortic valve closure.

In one embodiment, the end of the electrical systole can be identified using an ECG signal including a noise component and an ECG component originating from heart tissue of the subject. The ECG signal is decomposed from a first domain into subcomponents of the ECG signal in a second domain, and the location of a QRS complex of the cardiac cycle is identified using a spatial distribution of the subcomponents. A first time window in the cardiac cycle that includes the QRS complex is identified, as is at least one additional time window in the cardiac cycle that does not include the QRS complex. For each time window, subcomponents having more energy corresponding to the ECG component than noise energy are identified (e.g., those subcomponents within the respective window in which at least 50% of the energy thereof pertains to the actual ECG signal from the subject's heart, relative to noise). A denoised ECG is then constructed in the first domain by combining the identified subcomponents. The denoised ECG can then be analyzed using ECG analysis algorithms to identify the end of the electrical systole.

In another embodiment, the end of the electrical systole is identified by similarly decomposing an ECG signal from a first domain into subcomponents of the ECG signal in a second domain, and then identifying the location of the QRS complex of the cardiac cycle based upon a spatial distribution of the subcomponents. A T-wave offset search window is established, relative to the location of the QRS complex, and

at least two subcomponents are identified as having an energy value that is predominantly energy of a T-wave of the cardiac cycle. The at least two identified subcomponents are mathematically combined to compute an emphasis signal having an identifiable inflection corresponding to a location of the T-wave offset. The T-wave offset location is identified based upon a characteristic of the emphasis signal, and is used to identify the end of the electrical systole.

The cardiac-based metric is computed using one or more of a variety of approaches. In some embodiments, the metric is computed by computing one of a mean, a median, variance, standard deviation, and standard error of the time difference for each cardiac cycle. In certain embodiments, the cardiac-based metric is computed by computing a short-term instability metric based on one of the mean, standard deviation and root mean square of successive differences between beat-to-beat values in a window segment including the plurality of cardiac cycles, in which the beat-to-beat values correspond to heartbeats that define the successive start of the cardiac cycles. In another embodiment, a long-term instability metric is computed as a variance in the beat-to-beat values, multiplied by two and then subtracting the computed short term instability metric therefrom.

In connection with the above and other embodiments, it has been discovered that the measurement of synchrony between mechanical and electrical systole can be particularly useful in addressing issues as discussed above, and further that such approaches can be implemented together to obtain desirable characterizations of cardiac function under conditions in which noise has been challenging to address (e.g., with ambulatory subjects). It has further been discovered that these approaches can be achieved without necessarily involving invasive-type approaches, complex procedures such as the use of echocardiography to measure the velocity of heart tissue, and mitigates/avoids errors relating to changes in heart rate. Acoustical vibrations that occur as a result of aortic valve closure (e.g., S2 sounds) can be used in this context to measure EMW to achieve results similar to approaches involving left ventricular pressure (LVP) that are described in the literature. EMW, which is independent of heart rate, can be used to obtain a sensitive cardiac-based metric (e.g., by detecting changes in EMW that indicate an increased risk of arrhythmia on the order of 200%, relative to changes in QT and corrected QT(QT<sub>c</sub>) indicative of increased arrhythmic risk on the order of 30%).

Various embodiments are directed to non-invasive and minimally invasive measurements of EMW in ambulatory patients, such as for providing an assessment of arrhythmic risk. In some implementations, risk indicators are obtained on ambulatory subjects over a period of time (e.g., 24 hours). These approaches can be implemented, for example, to address alterations that can impact cardiac function and arrhythmic risk indicative of the changes that occur at the cellular level, which vary with time, stress, and other stimuli. Further, these risk indicators can be obtained from ambulatory patients while addressing noise from the patients' surroundings as well as noise that occurs due to respiration and patient movement (e.g., clothing rubbing). In some embodiments, such approaches are implemented using S2 heart sounds, an approach that allows for minimally invasive or non-invasive detection of MSend in ambulatory subjects. Such approaches are further facilitated by the use of MDSP signal processing techniques to accurately detect MSend and ESend when the signals are corrupted with noise.

Many embodiments described herein refer to signal processing approaches such as "multi-domain signal processing" (MDSP), which refers to one or more of various embodiments

described in U.S. patent application Ser. Nos. 12/938,995, 13/092,530, and 13/172,415, which may be implemented in accordance with one or more embodiments herein. These patent documents, as well as the patent documents therein to which benefit is claimed and the references cited therein, are fully incorporated herein by reference. In some embodiments, such an MDSP-based approach is used to process physiological information captured from ambulatory subjects in order to measure and detect alterations in cardiac function that are indicative of arrhythmic risk. In the following discussion, reference is made to cited references listed in a numbered order near the end of this document, which are fully incorporated herein by reference. These references may assist in providing general information regarding a variety of fields that may relate to one or more embodiments of the present disclosure, and further may provide specific information regarding the application of one or more such embodiments.

Various other embodiments are directed to methods involving the processing of physiological signals, as well as related apparatuses (e.g., with a computer type circuit and an input circuit that receives signals), which may be implemented as follows. Cardiac cycles are identified from an electrical signal representative of a subject's ECG. For each cardiac cycle, a T-wave offset is identified as is a segment of an acoustical vibration representative of heart sounds from the subject. The segment is identified based upon a T-wave offset time of a corresponding ECG synchronized with the heart sounds. An array of the identified segments from each of the plurality of cycles is constructed, and both heart sound and noise components of the acoustical vibration are computed from the segments using blind source separation. The presence of a heart sound is identified based upon energy in the heart sound components and the noise components (e.g., heart sound energy may be computed as the root mean square of computed heart sound components). In some implementations, such an approach involves detecting the presence of an S3 heart sound based upon characteristics of the energy in the heart sound and noise components being indicative of S3 heart sounds.

The heart sound and noise components are computed in a variety of manners, to suit particular applications. In some embodiments, heart sound and noise components of an acoustical vibration are computed using blind source separation with one or more of principal component analysis, eigenvalue decomposition, and independent component analysis. In certain embodiments, heart sound and noise components of the acoustical vibration are computed using blind source separation via principal component analysis followed by independent component analysis.

The segments of the acoustical vibration are determined or identified in a variety of manners. In some embodiments, segments are set to begin at a T-wave offset, and/or may terminate about 100 msec prior to the Q-wave onset of the subsequent cardiac cycle. In other embodiments, the segment is identified based on P-wave onset or P-wave peak of a subsequent cardiac cycle, and is terminated at the detected one of the P-wave onset or the P-wave peak.

In some embodiments, a time-frequency decomposition is applied to one of the above-discussed array of identified segments and the acoustical vibration, prior to computing the heart sound and noise components. This approach may, for example, involve using at least one of a wavelet related transform, a Gabor transform, a Fourier transform, a discrete cosine transform and a filter bank.

Various embodiments are directed to facilitating S3 heart sound detection. In some embodiments, the approximate location of an S2 heart sound is determined for at least one of

the identified segments of the acoustical vibration, and a portion of the S2 heart sound is blanked. S3 heart sounds are detected, based on the blanking of the at least a portion of the S2 heart sound.

Turning now to the figures, FIG. 1A shows an apparatus **100** for characterizing a cardiac-based metric of a subject (i.e., a patient), in accordance with another example embodiment. The apparatus **100** includes a mechanical systole module **110**, and electrical systole module **120**, and a processing module **130**. The mechanical systole module **110** operates to process acoustical data received from an acoustical signal acquisition module **140**, in which the acoustical data includes heart sound data of the subject, and to provide an indication of an end of the mechanical systole for the subject. In some implementations, the acoustical signal acquisition module includes a microphone or other audio detection-type device. The electrical systole module **120** operates to process electrical cardiac data, such as ECG data, for the subject and received via an electrical signal acquisition module **150**, and to provide an indication of an end of the electrical systole for the subject. For instance, the electrical signal acquisition module **150** may include ECG leads that are coupled to acquire an ECG signal from the subject. In some instances, the mechanical and electrical signal acquisition modules **140** and **150** are combined for application to the subject, such as via an adhesive component including both electrical and acoustical pickup circuits.

The processing module **130** is coupled to receive data from both the mechanical systole module **110** and the electrical systole module **120**, respectively indicative of the end of the mechanical systole and the electrical systole for the subject. The processing module **130** uses this data to compute a cardiac-based metric, such as may be related to proarrhythmic risk, based upon respective time differences between the end of the mechanical and electrical systoles for each of several cardiac cycles for which data is obtained from the subject. For instance, the time difference between the end of the mechanical and electrical systoles can be computed for each cardiac cycle, with the respective time differences used to compute the cardiac-based metric.

In some embodiments, the apparatus **100** includes a denoising module **160** that operates to denoise signals obtained from a subject as acquired by one or both of the acoustical signal acquisition module **140** and the electrical signal acquisition module **150**. This denoising module **160** may, for example, be implemented using a denoising approach as discussed herein and/or in the incorporated references/patent documents, such as those discussed in the context of MDSP. Further, this denoising module may be implemented in connection with other modules as shown (separately or together, or with two such denoising modules), such as within the mechanical systole module **110** and within the electrical systole module. In addition, one or more of the modules shown may be implemented in connection with processing circuits and/or a common processing circuit that executes programming to carry out respective functions, such as those involving the identification of systole characteristics, denoising, and the computation of a metric of cardiac risk.

Referring to FIG. 1B, input signals are shown as being processed in accordance with one or more example embodiments, for a normally functioning heart in which the electrical systole **102** begins with the onset of Q-wave **105**. The mechanical systole **101** begins immediately prior to an S1 heart sound **104**, which occurs concurrent with the opening of the subject's aortic valve. The end of the mechanical and electrical systole at **107** tightly coincide in time, each occurring within about 20 msec of the other, in such a normal heart.

An S2 heart sound **106** corresponds to approximately the end of the mechanical systole and the closing of the aortic and pulmonary valves. Inflection point **108** on arterial pressure signal **103** likewise corresponds to approximately the end of mechanical systole. Inflection point **108** is used to approximate the end of mechanical systole when arterial pressure is measured immediately adjacent to the heart. At points away from the heart, accommodation is made for delays in the transmission of the pressure wave through the arterial system that result in discordance of the inflection point with mechanical systole. In some embodiments, the diastolic pressure wave in a peripheral artery is monitored and used to approximate the inflection point of the mechanical systole.

FIGS. 1C and 1D respectively show signals as used in accordance with various embodiments, to characterize cardiac function in a heart predisposed to cardiac arrhythmias, respectively in which the end of the mechanical systole **201** (MSend) either precedes (as in FIG. 1C) or lags (as in FIG. 1D) the electrical systole **202** (ESend), which may depend upon the nature of the pathophysiology or mechanism of a drug effect. For instance, characteristics can be detected and used to identify a heart that has become predisposed to cardiac arrhythmias due to structural damage in myocardial tissue or by genetic or pro-arrhythmic drug alteration of myocyte ion channels. When this damage occurs, the complex electrophysiological and mechanical interactions that govern heart rhythm may become unbalanced, creating conditions that can lead to arrhythmias. The unbalanced electro-mechanical interactions may occur, at least in part, due to abnormal calcium handling. In the case where the MSend **205** (at the onset of heart sound **203**) precedes the end of electrical systole **206** (relative to the electro-mechanical window (EMW) **204**), as in FIG. 1C, the ventricular muscle is relaxing but is still electrically depolarized and is hence at risk of ventricular arrhythmia. Disassociation of the mechanical and electrical relaxation phases resulting in extension of repolarization into mechanical diastole can trigger Calcium (Ca) sparks in the myocardium, which can lead to Ca waves and Torsades de Pointe. Various embodiments are directed to detecting such characteristics and characterizing related cardiac risk.

Referring to FIG. 1D, the end of mechanical systole **302** lags the electrical systole **301**, which are detected via both mechanical and electrical detection and used as an indication that there is a time window in which heart tissue is electrically depolarized but has not yet started to relax (relative to the electrical systole **304**, EMW **303** and mechanical systole **305**). In some embodiments, disassociation of the mechanical and electrical relaxation phases as in this scenario are detected and used to identify conditions established in the myocardium in short QT syndrome (SQTS), such as described in Schimpf, et al., "Electromechanical coupling in patients with the short QT syndrome: Further insights into the mechano-electrical hypothesis of the U wave," Heart Rhythm, 2008 February; 5(2): 241-245, which is fully incorporated herein by reference. Changes in EMW on the order of tens of milliseconds can be highly significant. Accurate and consistent measurements of fiducial points are obtained in both electrical and hemodynamic/mechanical signals, using approaches as described herein.

In various embodiments, measurements of the QT interval (QTI) and the time from the Q onset to the S2 heart sound (QS2) are obtained and used to compute EMW as QS2-QTI. For instance, the Q onset can be detected in addition to the T-offset and S2 heart sound, to arrive at a similar result as in the above discussion.

Long-term (e.g., 24 hours or more) analysis is carried out in accordance with various example embodiments, to assess risk of arrhythmias from subjects ambulatory and/or going about normal activities. This approach involves characterizing the risk of arrhythmias as relative to a time-dependent component, including circadian variation, and is subject to environmental influences such as stress and dosing with a cardioactive drug. This approach can be used, for example, to detect arrhythmia risk markers such as QT prolongation and non-sustained ventricular tachycardia that may not be present during a spot check in the office or clinic. Often these risk markers are unmasked by other contributing factors present in everyday activity such as increased heart rate, stress, or medications. These approaches may be implemented, for example, to address challenges such as those discussed in J. Piccini, et al, "Predictors of sudden cardiac death change with time after myocardial infarction: results from the VALIANT trial," *European Heart Journal* (2009); and in R. Mayerburg, "Sudden cardiac death: exploring the limits of our knowledge," *Journal of Cardiovascular Electrophysiology*, Volume 12, No. 3, March (2001), which are fully incorporated herein by reference. Further, various embodiments are directed to accurately identifying T-wave offset in ambulatory subjects using an MDSP approach as described herein, in accordance with one or more embodiments as described in detail in the above-references, U.S. patent application Ser. No. 13/172,415, and as described in M. Brockway and R Hamlin, "Evaluation of an algorithm for highly automated measurements of QT interval," *Journal of Pharmacological and Toxicological Methods*, vol. 64, pp 16-24 (2011), which is fully incorporated herein by reference.

Various embodiments are directed to addressing challenges to accurately identifying the end of mechanical systole in ambulatory subjects. In one embodiment, EMW measurement is incorporated into an implantable device such as a pacemaker, implantable defibrillator (ICD), implantable cardiac monitor (ICM), or neurostimulation device to add diagnostic and monitoring capability to the device and/or to control therapy delivery based upon EMW measurements. In one embodiment, the device is a pacemaker or ICD with a lead extending into the heart, and the lead contains a pressure sensor to sense a right ventricular pressure. In another embodiment, a pressure sensor is placed in the left ventricle. In one embodiment, the end of mechanical systole is identified as minimum of right or left ventricular pressure (P). In another embodiment, the end of mechanical systole is estimated as a point of a maximum negative right or left ventricular pressure time derivative ( $\max -dP/dt$ ). In another embodiment, the endocardial lead or neurostimulation lead contains a microphone to sense heart sounds. In another embodiment, the lead contains an accelerometer to sense vibrations and movements that coincide with the end of mechanical systole.

In another embodiment the end of mechanical systole is identified by a sensor that is sensitive to mechanical vibrations that occur upon closure of the aortic valve (S2 heart sound). In some implementations, such a sensor includes an electronic microphone or an accelerometer placed either under the skin, or in contact with the outer surface of the skin. In some instances in which the sensor is placed under the skin, it is incorporated into an implantable therapeutic or monitoring device. In some implementations involving placement under or on the skin, the sensor includes one or more of a piezoelectric transducer, accelerometer, or microphone. In implementations in which the sensor is placed on the skin, an electronic microphone or accelerometer can be incorporated into an adhesive-backed patch as shown in FIG. 7. In another

embodiment, an acoustic sensor is placed in an elastic strap or tight fitting garment that holds the microphone in close contact with the skin.

In one embodiment an accelerometer, microphone, or other sensor capable of converting vibration into an electrical signal is placed subcutaneously or on the skin surface. The signal from the sensor is amplified and filtered to remove noise. In some embodiments an envelope of heart sound signal is computed using low pass-filtering or Hilbert transform. In some embodiments, derivative-based methods are applied to the envelope to generate an emphasis signal. In some embodiments peaks, valleys, and zero crossings of the emphasis signal are evaluated to detect the location of the S2 heart sound.

In some embodiments, a confidence signal is computed and used to assess the validity of a detected S2 location in a manner similar to that described in U.S. patent application Ser. No. 12/938,995, referenced above. If the confidence signal indicates that the detection is invalid or potentially invalid, the system discards the measurement of EMW for that cardiac cycle. In certain embodiments, a confidence signal is computed for a detected T-wave offset in a manner described in U.S. patent application Ser. No. 13/172,415, referenced above. If the confidence signal indicates that the detection is invalid or potentially invalid, the system can discard the EMW measurement for the corresponding cardiac cycle.

In some embodiments, EMW is measured in an implantable cardioverter defibrillator (ICD) and used to predict the onset of ventricular tachycardia (VT) or ventricular fibrillation (VF), and for initiating antitachycardic pacing (antitachy pacing) or overdrive pacing. Predicting the onset of VT or VF seconds or minutes in advance and initiating antitachy or overdrive pacing can be implemented to arrest an arrhythmia without the need for a painful shock delivered by the ICD. In one embodiment, end of electrical systole is detected in the ICD from the endocardial ECG (electrogram) or subcutaneous ECG with electrodes located on or near the ICD. The end of the mechanical systole is detected using one of a number of approaches including an acoustical sensor or accelerometer incorporated into an ICD lead wire, an acoustical sensor incorporated into the ICD can, or a hemodynamic sensor capable of measuring endocardial pressure in communication with the ICD. When using an acoustical sensor or accelerometer, circuitry within the ICD detects the S2 heart sound to indicate MSend. In one embodiment, a hemodynamic sensor measures right ventricular pressure (RVP) and end of mechanical systole is detected as the minimum RVP immediately following the downslope of the RVP waveform in a cardiac cycle or the maximum negative derivative of RVP in a cardiac cycle. In one embodiment, EMW is measured in the ICD or a pacemaker. When the absolute value of EMW consistently exceeds a predetermined threshold, the risk of a life threatening arrhythmia is increased and the ICD delivers antitachy pacing or overdrive pacing is effected for a predetermined period of time to avert the occurrence of a dangerous arrhythmia and the need to deliver a painful defibrillation shock.

In some embodiments, EMW measurements from multiple cardiac cycles are combined to obtain a measure of arrhythmic risk. In other embodiments, measurements of EMW are combined with measurements obtained from other predictors of arrhythmic risk such as QRS duration, QT interval, interval from T-peak to T-offset (TpTo), and T wave alternans (TWA). In one embodiment, a time series of beat-to-beat EMW values is averaged over a predetermined time period (e.g., 30 seconds) to compute a mean EMW value for comparison to a reference normal value as an indicator of risk.

The relationships between successive EMW dynamics and the measures on instability based on lag 1 dynamics are implemented to characterize cardiac function as graphically illustrated in FIG. 2 (e.g., as may be implemented with the apparatus 100 as shown in FIG. 1A). In some embodiments, the short term instability (STI) of EMW is computed as an indicator of arrhythmic risk. In one embodiment, STI 401 is computed as the mean of successive differences between beat-to-beat EMW values in a window segment including several (e.g., 10 to 500) cardiac cycles as follows:

$$STI_D = \sum_{i=1}^N |D_{n+1} - D_n| / [N\sqrt{2}],$$

where  $D_n$  is an EMW measured in the n-th beat and N is the number of beats in the segment.

In another embodiment, short term instability is computed as the standard deviation of successive differences between beat-to-beat EMW values in a segment of N beats. In another embodiment, short term instability is measured as the root mean square of successive differences between beat-to-beat EMW values in a segment of N beats.

In some embodiments, long-term instability (LTI) 402 is computed for a segment of N cardiac cycles in duration as an indicator of arrhythmic risk. LTI is measured as

$$LTI_D = \sum_{i=1}^N |D_{n+1} + D_n - 2D_{mean}| / [N\sqrt{2}],$$

where  $D_n$  is the value of EMW measured for the n-th beat of a segment having a duration of N beats.

In some embodiments a total instability (TI) 403 value is computed by combining STI and LTI, in which TI is computed as the square root of sum of squares of STI and LTI such as

$$TI_D = \sqrt{STI_D^2 + LTI_D^2} / \sqrt{2}$$

In some embodiments, the complexity of beat-to-beat interval dynamics is quantified for various risk metrics including QT interval, RR interval, and EMW. In an example embodiment, entropy-based analysis is used to quantify complexity of interval dynamics. For general information regarding entropy-based analysis, and for specific information regarding entropy-based analyses that may be implemented in accordance with one or more example embodiments, reference may be made to the Multiscale Entropy (MSE) approaches as described in M. Costa, A. L. Goldberger, and C.-K. Peng, "Multiscale Entropy Analysis of Complex Physiologic Time Series," Phys. Rev. Lett. 89, 6, (2002), which is fully incorporated herein by reference.

Referring to FIGS. 3A and 3B, various example embodiments involving beat-to-beat intervals are shown with respect to plots that are implemented, for example, in computing a cardiac metric (e.g., as may be implemented with the apparatus 100 shown in FIG. 1A). In FIG. 3A, plots 501, 502, and 503 illustrate beat-to-beat R-R interval dynamics with different characteristics: plot 501 shows an example of high variability and high complexity (normal healthy heart); plot 502 shows an example of low variability and low complexity (diseased heart with heart failure); and plot 503 shows an example of high variability and low complexity (diseased heart with atrial fibrillation). Plots 501 and 503 in FIG. 3A are both characterized by high variability. However, plot 501

corresponds to normal dynamics of a healthy heart, and plot 503 illustrates R-R interval dynamics of a diseased heart. These two plots demonstrate that both the healthy and diseased heart can both be characterized by high R-R interval variability, but their physiologic status can be discriminated by assessing complexity of R-R interval dynamics.

Complexity can be quantified using tools, such as multiscale entropy (MSE), that measure system entropy at various lags. In FIG. 3B, MSE is shown for the corresponding plots 504, 505, and 506, with lags ranging from 1 to 20. In FIG. 3B, plot 504 shows MSE for plot 501 of FIG. 3A, plot 505 shows MSE for plot 502, and plot 506 shows MSE for plot 503. In some embodiments involving these examples, the MSE trend is approximated by a linear equation for the first few lags, and the offset and slope of the linear equation can be used to assess the complexity of interval dynamics. In an example illustrated in plot 506, the low complexity of interval dynamics is characterized by high offset and large negative slope of the linear equation approximating multiscale entropy. High complexity dynamics illustrated in plot 504 are characterized by a relatively lower offset and a positive slope. Referring to FIG. 4A, a diseased heart is identified as exhibiting low complexity interval dynamics as illustrated in plots 601 and 602 in comparison to plots 603 and 604 of FIG. 4B representing normal heart dynamics, such as may be implemented in connection with the apparatus 100 shown in FIG. 1A.

In an example embodiment, and referring to FIG. 11, multiscale entropy (MSE) is computed and analyzed for assessing risk of arrhythmias or another characteristic of the heart such as atrial fibrillation, coronary ischemia, and an autonomic disturbance due to heart failure. One or more cardiac signals, such as ECG, endocardial pressure, and heart sounds, are input at 1301. A time series X(i) including beat-to-beat values is derived from the cardiac signals in 1302. A time series may include one of QT interval, QS2 interval, R-R interval, T-wave peak amplitude, T-wave area, and EMW=QS2-QT. Time series X(i) may include between 40 and 500 beats, although fewer or more beats may be used.

Time series X(i) is processed to compute sample entropy SE(1) in step 1303. Time series X(i) is further processed to compute sample entropy at multiple lags. For example, in 1304 X(i) is low pass filtered (LPF), decimated to remove every other point, and sample entropy SE(2) is computed for the resulting time series Y1(i) in step 1305. In one embodiment, the frequency cutoff of the LPF is 0.5/(level of decimation). Steps 1306 through 1313 mirror steps 1304 and 1305, and in which the low-pass filter cutoff and the level of decimation are set, for example, with the LPF cutoff in 1304 being ¼ and the level of decimation being 2. The level of decimation corresponds to a lag at which dynamics are evaluated and is also referred to as the scale of the entropy estimate. In 1306, the LPF cutoff is ⅓ and the level of decimation is 3 (2 of every 3 points is removed). In 1308, the LPF cutoff is ¼ and the level of decimation is 4 (3 of every 4 points is removed). In one embodiment the LPF is an IIR filter such as Butterworth filter. In another embodiment the LPF is an FIR filter such as moving average filter. The number of scales (m+1) as discussed above may be implemented to suit various applications. In one embodiment, the number of scales (m+1) is 10. The resulting trend of sample entropy values, SE(1), SE(2), SE(3), . . . SE(m+1) is analyzed at 1314 to assess maxima, slope, and offset. In one embodiment, a method used to compute sample entropy is the same for all scales and may be implemented in accordance with the method described in J. S. Richman and J. R. Moorman, "Physiological time-series

analysis using approximate entropy and sample entropy,” Am. J. Physiol. 278, H2039 (2000), which is fully incorporated by reference.

Referring to FIG. 10, an example embodiment is shown whereby multiple risk metrics are combined to form a single arrhythmic risk index. In one embodiment, EMW is combined with one or more of a) QRS duration, b) QT interval, c) short term QT instability, and d) TWA. ECG and heart sound signals input at 1201 are processed in 1202 using MDSP and beat-to-beat values of one or more time series Ts of EMW, QT interval, R-R interval, T-wave duration, T-wave area, and T-wave peak amplitude, which are identified in 1203 for a time segment (e.g., 60 seconds). In step 1204, the beat-to-beat values in the time segment are processed to compute a number of risk metrics, such as one or more of: ABS (Mean(Ts)-Normal), Short term instability (STI), Long-term instability (LTI), multi-scale entropy (MSE) offset, and multi-scale entropy (MSE) slope.

In one embodiment, the mean of valid beat-to-beat values (e.g., those obtained from normal cardiac cycles and not corrupted by noise) is computed for the time segment. The population normal value for the metric is subtracted from the mean computed for the segment. The deviation from the normal value is compared to one or more thresholds to assess the level of arrhythmic risk. In another embodiment, STI and LTI are computed and combined to form a total instability metric as described earlier.

In another embodiment, multiscale entropy parameters are used to assess cardiac risk, slope and offset of the linear equation approximating the MSE trend are computed for a QT interval and compared to multiscale entropy parameters of RR interval. Referring to FIG. 4A, an embodiment is shown involving RR interval and QT interval dynamics during heart failure (plots 601 and 602) and normal condition (plots 603 and 604 of FIG. 4B). Referring to 601 and 602 of FIG. 4A, compromised cardiac function is characterized by low variability RR interval dynamics and high variability QT with low complexity. In this case, MSE of QT exhibits high offset and high negative slope while MSE of RR interval exhibits low offset. In contrast, normal cardiovascular function is characterized by high variability RR interval dynamics and low variability QT dynamics as illustrated in plots 603 and 604. In this case, MSE of QT exhibits low offset and normal slope while MSE of RR interval exhibit normal offset and positive slope. An example relationship between the proposed parameters for compromised and normal cardiac function is illustrated in Table 1.

Note that the statistical measures of cardiac intervals shown in Table 1 demonstrate that EMW provides a consistent indicator of cardiac status, whereas QT and RR are often inconsistent.

TABLE 1

Statistical Measures of Cardiac Intervals						
	Mean	STI	LTI	TI	MSE Offset	MSE Slope
<b>Abnormal Cardiac Function</b>						
ABS (EMW)	High	High	High	High	High	High, negative
QT	High or low	High	High	High	High	High, negative
RR	High	Low	Low	Low	Low	Negative
<b>Normal Cardiac Function</b>						
EMW	Low	Low	Low	Low	Normal	Normal
QT	Normal	Normal	Normal	Normal	Low	Normal
RR	Normal	Normal	Normal	Normal	Normal	Positive

FIG. 5 shows an embodiment of a system for assessing arrhythmic risk. Ambulatory monitoring device (AMD) 701 receives sensed ECG and heart sounds. A computing device within AMD 701 processes these signals to compute beat-to-beat values for one or more of EMW, QT, R-R, and QRS duration. In one embodiment, AMD 701 also evaluates the validity of these computed values using MDSP techniques described in one or more references above (characterized with MDSP). These values are telemetered from AMD 701 to a wireless communication module 702 where they are forwarded via communication link 703 to an evaluation system and report generator (ESRG) 704. The ESRG may, for example, be located in a clinic, laboratory, or service bureau where the received values are reviewed by trained personnel and may be further processed by a computer. In one embodiment, two or more of the received beat-to-beat values are combined in 704 to compute a composite risk index. In some embodiments, AMD 701 detects events of interest, such as arrhythmias, and transmits ECG strips containing those events to wireless communication module 702 and on to ESRG 704. In other embodiments, AMD 701 transmits either segments of the ECG signal (e.g., a 60 second strip) at regular intervals, or a full disclosure ECG and the beat-to-beat values are extracted in ESRG 704.

FIG. 6 shows an embodiment involving the use of AMD 701 in an implementation in which it is worn by a subject, as implemented with device 802 which includes AMD 701 as well as signal conditioning electronics to process sensed ECG and signals received from acoustical sensors or accelerometers, a computing circuit to denoise the sensed signals and extract beat-to-beat values, a communication circuit to wirelessly telemeter information to wireless communication module 702, and a battery to power the circuits. AMD 701 receives sensed ECG from electrodes 803 and sensed ECG and acoustical or accelerometer signals from combined ECG electrode and acoustical sensor 801. In some embodiments, a combined ECG and acoustical sensor 801 is implemented at each electrode location. This can provide redundancy when sensing the acoustical signal and improve noise immunity by processing signals received from multiple acoustical sensors concurrently to improve the accuracy of S2 heart sound detection. In some embodiments, AMD 701 employs more or less (but not less than two) ECG sensing electrodes.

In one embodiment, referring to FIG. 7, 801 includes adhesive backed form 904 at 2 to 3 mm thick with an ECG sensing electrode 901 and an acoustical sensor 902 mounted in a cavity within the foam. Combined ECG and acoustical sensor 801 provides a single patch that can be adhered to the skin surface for measuring ECG and heart sounds, and used to assess the risk of arrhythmias. Sensor 801 includes foam pad 904 with adhesive applied selectively in region 903. ECG sensing area 901 includes a gel material that couples charge from the skin to a silver-silver chloride material that is electrically connected to device 802 via cable 907. Acoustical sensor 902 includes a microphone, accelerometer, or other sensor capable of converting the mechanical vibrations produced by the heart during closing of the aortic valve into an electrical signal. In some embodiments, sensor 902 is coated with an impedance matching material 908 that matches the impedance of the sensing element in sensor 902 with the acoustical impedance of the skin in order to improve the signal-to-noise ratio of the output of sensor 902. Connecting wires extend from sensing elements 901 and 902 to connector 905. Mating connector 906 connectively couples the signals into cable 907 where they are forwarded to the electronics contained within device 802.

FIG. 8 shows an embodiment involving electronic circuitry that may be contained within AMD 701. Input signals 1001 and 1004 are amplified and filtered to remove noise outside the bandwidth of the signals in signal conditioning circuits 1002 and 1005. In one embodiment, system-on-a-chip 1003 contains analog-to-digital converters to digitize the ECG and acoustical or accelerometer signals, an ARM Cortex microcontroller and associated memory to denoise (e.g. remove in-band noise) and extract beat-to-beat values from the combined ECG and acoustical or accelerometer signals, and a telemetry link and antenna 1006 to wirelessly communicate information to wireless communication module 702.

In one embodiment, referring to FIG. 9, EMW is computed using an ECG signal and a digitized acoustical or accelerometer signal. Digitized ECG input signal(s) 1101 are denoised in process 1102 using MDSP to remove in-band noise. The rhythm is evaluated in process 1104 to determine if the current beat is normal or arrhythmic (e.g., premature ventricular contraction, ventricular tachycardia, etc.). If in decision process 1105 it is determined that the beat is arrhythmic, then EMW is not measured and signal flow is directed to the next heart beat via process 1103. If the beat is normal, an S2 search window is established relative to the location of the R-wave in process 1106. For a human being, in some embodiments, the S2 search window begins about 250 msec after the R-wave

and extend for about 250 msec. The location of the search window relative to the R-wave is varied based upon species. In some embodiments, the search window location is adjusted on an ongoing basis depending upon the prior RR interval length. In addition, a search window for T-wave offset is also established relative to the R-wave location and the previous R-R interval in process 1107. T-wave offset is identified within the search window and tested for validity using MDSP techniques in process 1108 as described in U.S. patent application Ser. No. 13/172,415, as discussed above.

In one embodiment, digitized input acoustic or accelerometer signal(s) 1109 is (are) band-pass filtered in process 1110 to remove noise outside the pass-band of the S2 heart sounds. In one embodiment band-pass filtering is accomplished using an infinite impulse response filter. In one embodiment, the resulting signal is denoised using signal averaging techniques in process 1111 to remove noise. An envelope of the denoised signal is computed in 1112 using, for example, a Hilbert transform. In some embodiments, derivative-based methods are applied to the envelope to generate an emphasis signal. The resulting emphasis signal is subjected to a peak or threshold detector to identify the S2 heart sound within the search window established in process 1106. EMW is computed in process 1114 as the time difference between the S2 location identified in 1113 and the T-wave offset location identified in 1108. In an alternate embodiment, an emphasis signal is computed based upon process 1407 in FIG. 12 and the peaks, valleys, and zero crossings of the resulting emphasis signal are evaluated to determine the location of the S2 sound.

In another embodiment, referring to FIG. 12, input signal at 1401 is decomposed at 1402 to create time synchronized subcomponents in a second domain using one of: a) a discrete cosine transform, b) a wavelet related transform, c) a Karhunen-Loeve transform, d) a short-time Fourier transform, and e) a filter bank. In some embodiments the decomposition is performed in a way that preserves time synchronization of the subcomponents used to compute the time function 1404 in order to achieve an accurate representation of the spatial distribution of the selected subcomponents. To this end, the decomposition uses techniques such as non-orthogonal waveform, undecimated wavelet transform, or stationary wavelet transform.

Subcomponents containing at least a majority of S2 complex energy are selected in 1403. In one embodiment, these subcomponents are selected based upon a priori knowledge of the frequency content of the S2 complexes. Selected subcomponents are combined in 1404 to form a time function  $X_s$  that emphasizes the S2 complex. In some embodiments, the time function is formed by computing a point-wise product of the selected subcomponents. In another embodiment, the time function is formed by computing a sum of the selected subcomponents. In yet another embodiment, the time function is formed by computing a cross-correlation function between the selected subcomponents.

Subcomponents and time segments of subcomponents associated with noise energy are identified in step 1405 using spatially selective filtering, such as one or more of the above-referenced patent documents. In step 1406, the noise floor  $X_n$  is computed by combining the subcomponents  $D_{2n}$  identified in step 1405. In one embodiment, the noise floor is computed on a sample-by-sample basis. In another embodiment, the noise floor is determined by computing the sum of the squares of the subcomponents  $D_{2n}$ . In another embodiment, an emphasis signal is computed in step 1407 as the difference  $X_e = X_s - X_n$ . The threshold of S2 detection is computed in step 1408. In some embodiments the threshold for S2 detection in 1408 is adaptive and changes based on distance from

the prior QRS wave. The occurrence of S2 is detected in step 1409 when the time function  $X_e$  exceeds the threshold computed in step 1408 within a predetermined window starting from the prior QRS wave.

Various embodiments are directed to using S3 heart sounds for prediction and detection of heart failure decompensation. Patients at risk of decompensation are examined using a mechanical or electronic stethoscope for determining whether a heart sound is present. In some embodiments, S3 heart sounds are automatically detected from an acoustical signal acquired using a microphone, accelerometer, or electronic stethoscope placed on the chest. In an example such embodiment, and referring to FIG. 10, ECG sensing electrodes and an acoustic sensor are placed on the chest of a patient. The ECG sensing electrodes can be standard adhesive backed electrodes, or they can be dry electrodes. In some embodiments, the dry electrodes are attached to the electronic stethoscope or microphone/accelerometer to provide an apparatus that can be held in one hand and applied to a patient for simultaneous recording an acoustical signal and ECG. In some embodiments, the dry electrodes can be similar to those available from Orbital Research (Cleveland, Ohio) and can be mounted in a structure that is integral to the acoustical sensor or stethoscope.

In some embodiments the ability to automatically detect S3 sounds is incorporated with a home monitoring device or telemedicine work station whereby the patient is directed to place an electronic stethoscope on the thorax at the approximate position of the apex of the heart (e.g., in the vicinity of V4) while leaning forward sitting in a chair or lying on their left side. In some embodiments, ECG electrodes are positioned cranial and caudal to the location of the microphone along a line approximately parallel to a Lead II ECG vector. ECG and acoustical signals can be processed by a hand-held battery powered device and the results of the assessment communicated to a remotely located care provider. Alternatively, the digitized ECG and acoustical signals are communicated for processing to a computing device located remote from the patient. In some embodiments, the device provides an indication that S3 sounds are present and alert a skilled medical professional to listen to the received heart sound recording or a live transmission of heart sounds via a telemonitoring workstation or cloud-enabled electronic stethoscope (e.g. such as 3M TeleSteth or American Telecare Caretone) to confirm the presence of S3 sounds. If S3 sounds are detected, a medical professional can then take appropriate action to medically manage the patient to avoid hospitalization. In other embodiments, an S3 heart sound detector is incorporated into an electronic stethoscope to aid less skilled medical professionals in recognizing the presence of S3 sounds.

In another embodiment a home monitoring system or telemonitoring workstation is configured to provide a more complete cardiovascular assessment, using heart sounds. For example, various embodiments described herein may be implemented in connection with aspects disclosed in U.S. Pat. Nos. 8,632,465; 8,688,202; 8,433,395; and U.S. provisional patent application Ser. No. 61/944,253 to which benefit is claimed (and which are fully incorporated herein), such as to provide a home-based system for assessment of patients suffering from heart failure and commonly associated comorbidities. These co-morbidities include atrial fibrillation (AF) and risk of sudden cardiac death (SCD), in addition to the risk of heart failure decompensation (HFD). AF is a common trigger of HFD, increases risk of stroke, and is often asymptomatic. Various embodiments are directed to automatically detecting AF in the home monitoring system, which

can improve patient medical management. In one embodiment, AF is detected from the ECG signal as described in U.S. provisional patent application Ser. No. 61/944,253 as referenced above. The risk of SCD is assessed in the home monitoring system. In one embodiment, risk of SCD is assessed as described in U.S. Pat. No. 8,688,202 by analysis of the ECG and acoustical signals. These aspects can facilitate patient medical management. In one embodiment, respiratory rate is derived from the ECG using MDSP techniques as described in U.S. Pat. No. 8,632,465, with increased respiratory rate and labored breathing being used as an indicator of heart failure decompensation and other respiratory issues.

FIG. 13 illustrates a scenario for the relative timing of S1 (1501), S2 (1502), S3 (1503), and S4 (1508) heart sounds and the corresponding ECG waveform, in accordance with one or more embodiments. Referring to FIG. 14, a recording of ECG and hearts sounds is obtained using an apparatus similar to that shown in FIG. 8. In some embodiments, a 30 second recording is obtained in step 1602, but the recording could be as short as 15 seconds or as long as 2 minutes. In some embodiments, the recording is sufficiently long so that at least 8 relatively noise-free cardiac cycles, free of arrhythmias and having similar RR intervals, are obtained for processing. In this example embodiment, the location and length of cardiac cycles are determined by detection of the QRS complex in the ECG. However, in other embodiments these parameters (e.g., cardiac cycle length and absence of arrhythmias) are determined from a non-invasive blood pressure measurement device or oxygen saturation measurement device. The recording is analyzed in 1604 to identify if any arrhythmic beats are present. Any beat that is arrhythmic (e.g., contains a wide complex QRS or is premature) is discarded and not used for further analysis. In some embodiments, if the RR interval is shorter than a predetermined value (e.g., 500 msec), that beat is excluded from analysis. In yet other embodiments, noise level is evaluated in both ECG and the acoustical signal and a beat is excluded from analysis if the noise level exceeds a threshold. In such embodiments it may be useful to employ MDSP to measure noise as described in U.S. Pat. No. 8,632,465. If fewer than 8 beats remain following exclusion of certain beats, a new recording is obtained. To improve consistency of the information in each cardiac cycle, a candidate group of beats is selected in 1605 with RR intervals that differ by <10%. T-wave offset is identified in 1606 for each beat in the group in a manner similar to that described elsewhere in this disclosure and in U.S. Pat. No. 8,433,395.

The acoustical recording is processed in 1607 to identify the onset of the S2 heart sound for each beat in the group. In certain implementations, at least a portion of the S2 sound is blanked from the recording during time window 1504 to avoid or reduce leakage of power from the S2 sound to the S3 sound during processing, addressing issues relating to the energy in the S2 sound being greater than the energy in the S3 sound. In one embodiment, blanking window 1504 is 75 msec long. In other embodiments the blanking window ranges from 50 msec to 150 msec. In another embodiment, blanking window 1504 is timed based upon the T-wave offset rather than S2.

A noise measurement window 1505 precedes the onset of the S2 heart sound. Window 1505 is positioned to provide for measurement of energy present in acoustical signal between the S1 and S2 heart sounds. In some embodiments, window 1505 terminates about 50 to 100 msec prior to detection of S2 onset to avoid including any energy of the S2 heart sound in window 1505. Likewise, initiation of window 1505 should follow cessation of acoustical energy from the S1 sound in the cardiac cycle. Acoustical signal energy in window 1505 is

measured in step **1608**. All energy measured in this window is assumed to be noise. In one embodiment, energy in window **1505** is measured by computing the root-mean-square of the acoustical signal. If noise energy  $N_a$  is greater than a predetermined threshold  $Th_1$ , the cardiac cycle is considered to be too noisy to obtain reliable S3 detection and is removed from the candidate group of beats in **1609**. In one embodiment, window **1505** is 250 msec in duration. In other embodiments, window **1505** can range from 200 msec to 300 msec. It may be useful in some embodiments to determine the threshold  $Th_1$  empirically. In other embodiments  $Th_1$  can be determined by establishing a baseline noise reading under circumstances when the sensed acoustical energy is known to be relatively noise free or when a nominal noise level is present.

In step **1610**, S3 search window **1509** is established for each cardiac cycle of the acoustical recording. If the RR interval for a cardiac cycle is too short, that cardiac cycle is excluded from evaluation of S3 sounds. In one embodiment, window **1509** spans the acoustical signal for a cardiac cycle from T-wave offset **1506** to about atrial depolarization **1510** (i.e. P wave onset or P-wave peak). In one embodiment, the termination of window **1509** is established as about 100 msec prior to the Q-wave onset time **1507** of the subsequent cardiac cycle. In other embodiments, the termination of window **1509** can be established as about 250 msec prior to the onset of the S1 heart sound of the subsequent cardiac cycle. In some embodiments, the S1 heart sound is detected as described in U.S. Pat. No. 8,688,202. In alternate embodiments, the S3 search window is initiated upon onset of the S2 heart sound. In an alternate embodiment, the time that the S2 heart sound extends beyond T-wave offset can be measured or estimated as  $TS_2$ . The start time of the S3 search window can hence be established as T-wave offset **1506**+ $TS_2$ . In some embodiments,  $TS_2$  corresponds to the duration of blanking interval **1504** and hence the S3 search window starts following termination of blanking interval **1504**.

In yet another embodiment, T-wave offset **1506** and Q-wave onset **1514** are identified for each cardiac cycle in the ECG recording. A group of cardiac cycles is identified with nearly identical RR intervals (e.g. RR intervals differ by <10%) and the mean QT interval for this group of cardiac cycles is computed. The S3 search window for a cardiac cycle is subsequently based upon the location of Q-onset for that cardiac cycle and the mean QT interval for the group.

FIG. **15** shows an example acoustical recording with S1 (**1706**), S2 (**1703**), and S3 (**1701**) heart sounds present aligned in time with a representative ECG recording. S3 search windows **1705** spans from mark **1704** to mark **1702**. Note that for illustrative purposes, S2 heart sounds **1703** have not been blanked in FIG. **15**. Referring to FIG. **16**, 8 acoustical recording segments spanning S3 search window **1803** from time **1801** to **1802** have been synchronized based upon T-wave offset **1806** of the corresponding ECG signal and blanking window **1504** has been applied. The matrix of these acoustical recording segments is formed in step **1611**. Creating a matrix results in creation of a multidimensional space where signal and noise can be orthogonalized via blind source separation techniques, which are (in one or more embodiments) carried out as described in U.S. Pat. No. 8,632,465. For instance, noise may be removed from a pseudoperiodic signal by segmenting the signal and combining the segments into a matrix, as characterized therein. In some embodiments, the matrix formed in **1611** includes between 8 and 64 search windows, with more or fewer cardiac cycles used for other embodiments.

In one embodiment, the matrix is processed using one or more blind source separation (BSS) methods such as, princi-

pal component analysis (PCA), independent component analysis (ICA), or eigenvalue decomposition is applied in step **1612**. The process of constructing a matrix of segmented quasiperiodic signals followed by application of blind source separation (BSS) techniques to extract signal from noise is described in U.S. Pat. No. 8,632,465 included herein by reference. The output of the blind source separation provides two signals: a denoised heart sound signal and a noise signal. Signal **1805** is representative of the denoised heart sound signal produced by BSS.

In an alternate embodiment the matrix is decomposed using a time-frequency transform, such as a wavelet transform, followed by application of BSS on the resulting sub-components. In yet another embodiment, a time-frequency transform is applied to the acoustical signal prior to segmentation and formation of the matrix. The use of a time-frequency transform allows for signal components to be separated in time and frequency and creates additional dimensions in the space to facilitate better signal and noise separation. The BSS results in orthogonalization of independent components and separation of signal from noise based upon relative energy. The orthogonalized sub-components with low signal power are often associated with noise and are removed to achieve denoising. The orthogonalized sub-components with high signal power are used to reconstruct the denoised heart sound signal using an inverse time-frequency transform.

Once the heart sound and noise signals are separated, the amplitude or energy of the heart sounds signal corresponding to S3 location is measured and compared to the amplitude or power of ambient noise from outside of the S3 location. In one embodiment, heart sound power ( $H_{Sp}$ ) and noise power ( $N_p$ ) are computed at **1613** as the root mean square of each signal. A ratio  $H_{Sp}/N_p$  is computed in **1614** and compared to threshold  $Th_2$  in **1615** to determine if S3 sounds are present. In one embodiment S3 power for a patient is used to measure a relative severity score. In some embodiments, the history of this severity score is tracked to assess the trend of the patient's condition.

In one embodiment, an S3 search window is established in the acoustical signal for each cardiac cycle spanning from T-wave offset to QRS onset of the following cycle. In some embodiments the S3 search window begins after the S2 heart sound, starting at the T-wave offset plus a time slightly longer than the duration of a typical S2 sound in order to eliminate at least a portion of the S2 heart sound from the acoustical recording.

In some embodiments, the presence of either S3 (reference **1503** in FIG. **13**) or S4 (reference **1508** in FIG. **13**) heart sounds is detected. If either S3 or S4 heart sounds were detected, a message could be sent to a care provider that the patient may warrant closer examination to assess their medical condition. In one embodiment for detecting the presence of either one or a combination of S3 and S4 heart sounds, search window **1512** is employed in order to include the portion of the cardiac cycle where S3 heart sound **1503** and S4 heart sound **1508**, if present, would occur. S3+S4 search window **1512** would hence replace "S3 search window" in steps **1610** and **1611**, "S3 and S4 absent" would replace step **1616**, and "S3 or S4 present" would replace step **1617**. Other steps described in FIG. **14** are implemented similarly to those used for detection of S3 in isolation.

In some embodiments, S4 is detected in isolation. This can be accomplished by defining a search window **1513** that comprises the time segment of the cardiac cycle containing the S4 sound. In one embodiment, S4 search window **1513** is

defined as the time from about P-wave onset **1510** to Q-wave onset **1507**. In another embodiment, an S4 search window is defined as starting at T-wave offset time **1506** and blanking the acoustical signal for the duration of S3 search window **1509**. Extraction of the S4 signal and detection of its presence in either of these embodiments may proceed as described previously for detection of S3.

The various embodiments as discussed herein may be implemented using a variety of structures and related operations/functions. For instance, one or more embodiments as described herein may be computer-implemented or computer-assisted, as by being coded as software within a coding system as memory-based codes or instructions executed by a computer processor, microprocessor, PC or mainframe computer. Such computer-based implementations are implemented using one or more programmable circuits that include at least one computer-processor and internal/external memory and/or registers for data retention and access. One or more embodiments may also be implemented in various other forms of hardware such as a state machine, programmed into a circuit such as a field-programmable gate array, and/or implemented using electronic circuits such as digital or analog circuits. In addition, various embodiments may be implemented using a tangible storage medium that stores instructions that, when executed by a processor, performs one or more of the steps, methods or processes described herein. These applications and embodiments may also be used in combination; for instance certain functions can be implemented using discrete logic (e.g., a digital circuit) that generates an output that is provided as an input to a processor.

Various modules may be implemented to carry out one or more of the operations and activities described herein and/or shown in the figures. In these contexts, a "module" is a circuit that carries out one or more of these or related operations/activities (e.g., ascertaining a signal characteristic, or computing a value based upon such ascertained characteristics). For example, in certain of the above-discussed embodiments, one or more modules are discrete logic circuits or programmable logic circuits configured and arranged for implementing these operations/activities, as in the circuit modules shown in the Figures. In certain embodiments, such a programmable circuit is one or more computer circuits programmed to execute a set (or sets) of instructions (and/or configuration data). The instructions (and/or configuration data) can be in the form of firmware or software stored in and accessible from a memory (circuit). As an example, first and second modules include a combination of a CPU hardware-based circuit and a set of instructions in the form of firmware, where the first module includes a first CPU hardware circuit with one set of instructions and the second module includes a second CPU hardware circuit with another set of instructions.

Certain embodiments are directed to a computer program product (e.g., nonvolatile memory device), which includes a machine or computer-readable medium having stored thereon instructions which may be executed by a computer (or other electronic device) to perform these operations/activities.

Based upon the above discussion and illustrations, those skilled in the art will readily recognize that various modifications and changes may be made to the present invention without strictly following the exemplary embodiments and applications illustrated and described herein. For example, different types of signal collecting devices may be used. Such modifications do not depart from the true spirit and scope of the present invention, including that set forth in the following claims.

What is claimed is:

1. A method comprising:

identifying a plurality of cardiac cycles in an electrical signal representative of an electrocardiogram (ECG) from a subject;

for each of the plurality of cardiac cycles,

identifying T-wave offset of the ECG, and

identifying a segment of an acoustical vibration representative of heart sounds from the subject, based upon a T-wave offset time of a corresponding ECG synchronized with the heart sounds;

constructing an array of the identified segments from each of the plurality of cycles;

computing heart sound and noise components of the acoustical vibration using blind source separation; and

detecting the presence of a heart sound based upon energy in the heart sound components and the noise components.

2. The method of claim 1, wherein detecting the presence of a heart sound includes detecting the presence of a S3 heart sound based upon characteristics of the energy in the heart sound and noise components being indicative of S3 heart sounds.

3. The method of claim 1, wherein computing heart sound and noise components of the acoustical vibration using blind source separation includes using at least one of:

principal component analysis;

eigenvalue decomposition; and

independent component analysis.

4. The method of claim 1, further including applying a time-frequency decomposition to one of the array of identified segments and the acoustical vibration, prior to computing the heart sound and noise components.

5. The method of claim 4, wherein applying a time-frequency decomposition includes using at least one of:

a wavelet related transform;

a Gabor transform;

a Fourier transform;

a discrete cosine transform; and

a filter bank.

6. The method of claim 1 wherein computing heart sound and noise components of the acoustical vibration using blind source separation includes performing principal component analysis followed by independent component analysis.

7. The method of claim 1 wherein identifying the segment of the acoustical vibration based upon the T-wave offset time includes beginning each identified segment at T-wave offset.

8. The method of claim 1, further including identifying the approximate location of an S2 heart sound for at least one of the identified segments of the acoustical vibration and blanking at least a portion of said S2 heart sound, wherein detecting the presence of a heart sound includes detecting the presence of an S3 heart sound, based on the blanking of the at least a portion of the S2 heart sound.

9. The method of claim 1, wherein identifying the segment includes identifying a segment that terminates about 100 msec prior to a Q-wave onset of a subsequent cardiac cycle.

10. The method of claim 1, wherein identifying the segment includes detecting one of P-wave onset or P-wave peak of a subsequent cardiac cycle, and terminating the segment at the detected one of the P-wave onset or the P-wave peak.

11. The method of claim 1, further including computing the energy of the heart sounds as the root mean square of the computed heart sound components.

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12. An apparatus comprising:  
 an input circuit;  
 a computer circuit configured and arranged with the input circuit to:  
 receive an electrical signal representative of an electrocardiogram (ECG) from a subject;  
 identify a plurality of cardiac cycles in the electrical signal;  
 for each of the plurality of cardiac cycles,  
 identify T-wave offset of the ECG, and  
 identify a segment of an acoustical vibration representative of heart sounds from the subject, based upon a T-wave offset time of a corresponding ECG synchronized with the heart sounds;  
 construct an array of the identified segments from each of the plurality of cycles;  
 compute heart sound and noise components of the acoustical vibration using blind source separation; and  
 detect the presence of a heart sound based upon energy in the heart sound components and the noise components.

13. The apparatus of claim 12, wherein the computer circuit is configured and arranged to detect the presence of a heart sound by detecting the presence of an S3 heart sound based upon characteristics of the energy in the heart sound and noise components being indicative of S3 heart sounds.

14. The apparatus of claim 12, wherein the computer circuit is configured and arranged to apply a time-frequency decomposition to one of the array of identified segments and the acoustical vibration, prior to computing the heart sound and noise components.

15. The apparatus of claim 12 wherein the computer circuit is configured and arranged to compute heart sound and noise components of the acoustical vibration using blind source separation by performing principal component analysis followed by independent component analysis.

16. The apparatus of claim 12 wherein the computer circuit is configured and arranged to identify the segment of the

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acoustical vibration based upon the T-wave offset time by beginning each identified segment at T-wave offset.

17. The apparatus of claim 12, wherein the computer circuit is configured and arranged to:

5 identify the approximate location of an S2 heart sound for at least one of the identified segments of the acoustical vibration, and blank at least a portion of said S2 heart sound, and

10 detect the presence of a heart sound by detecting the presence of an S3 heart sound, based on the blanking of the at least a portion of the S2 heart sound.

18. The apparatus of claim 12, wherein the computer circuit is configured and arranged to identify the segment by identifying a segment that terminates about 100 msec prior to the Q-wave onset of the subsequent cardiac cycle.

19. The apparatus of claim 12, wherein the computer circuit is configured and arranged to identify the segment by detecting one of P-wave onset or P-wave peak of a subsequent cardiac cycle, and terminating the segment at the detected one of the P-wave onset or the P-wave peak.

20. An apparatus comprising:  
 means for identifying a plurality of cardiac cycles in an electrical signal representative of an electrocardiogram (ECG) from a subject;

25 means for, for each of the plurality of cardiac cycles, identifying T-wave offset of the ECG, and identifying a segment of an acoustical vibration representative of heart sounds from the subject, based upon a T-wave offset time of a corresponding ECG synchronized with the heart sounds;

30 means for constructing an array of the identified segments from each of the plurality of cycles;

means for computing heart sound and noise components of the acoustical vibration using blind source separation; and

35 means for detecting the presence of a heart sound based upon energy in the heart sound components and the noise components.

\* \* \* \* \*

专利名称(译)	用于识别心脏风险的方法和设备		
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[标]申请(专利权)人(译)	VIVAQUANT		
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当前申请(专利权)人(译)	VIVAQUANT LLC		
[标]发明人	BROCKWAY MARINA BROCKWAY BRIAN		
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IPC分类号	A61B5/04 A61B5/00 A61B5/0245 A61B5/0408 A61B5/0452 A61B7/00 G06F17/14 G06F19/00 G06K9/00 H03H17/02		
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优先权	61/555165 2011-11-03 US 61/944253 2014-02-25 US 61/257718 2009-11-03 US 61/359462 2010-06-29 US 61/370026 2010-08-02 US 61/366052 2010-07-20 US		
其他公开文献	US20140296726A1		
外部链接	<a href="#">Espacenet</a> <a href="#">USPTO</a>		

摘要(译)

基于心脏的度量基于受试者的心脏功能的特征来计算。根据一个或多个实施例，基于与心动周期期间主动脉瓣闭合相关联的声学振动，针对受试者的多个心动周期中的每一个识别机械收缩的末端。还识别每个心动周期的心电图 ( ECG ) 信号的电收缩的结束。基于基于心脏的度量，基于电心收缩的结束与机械收缩的结束之间的时间差，针对相应的心动周期计算。

