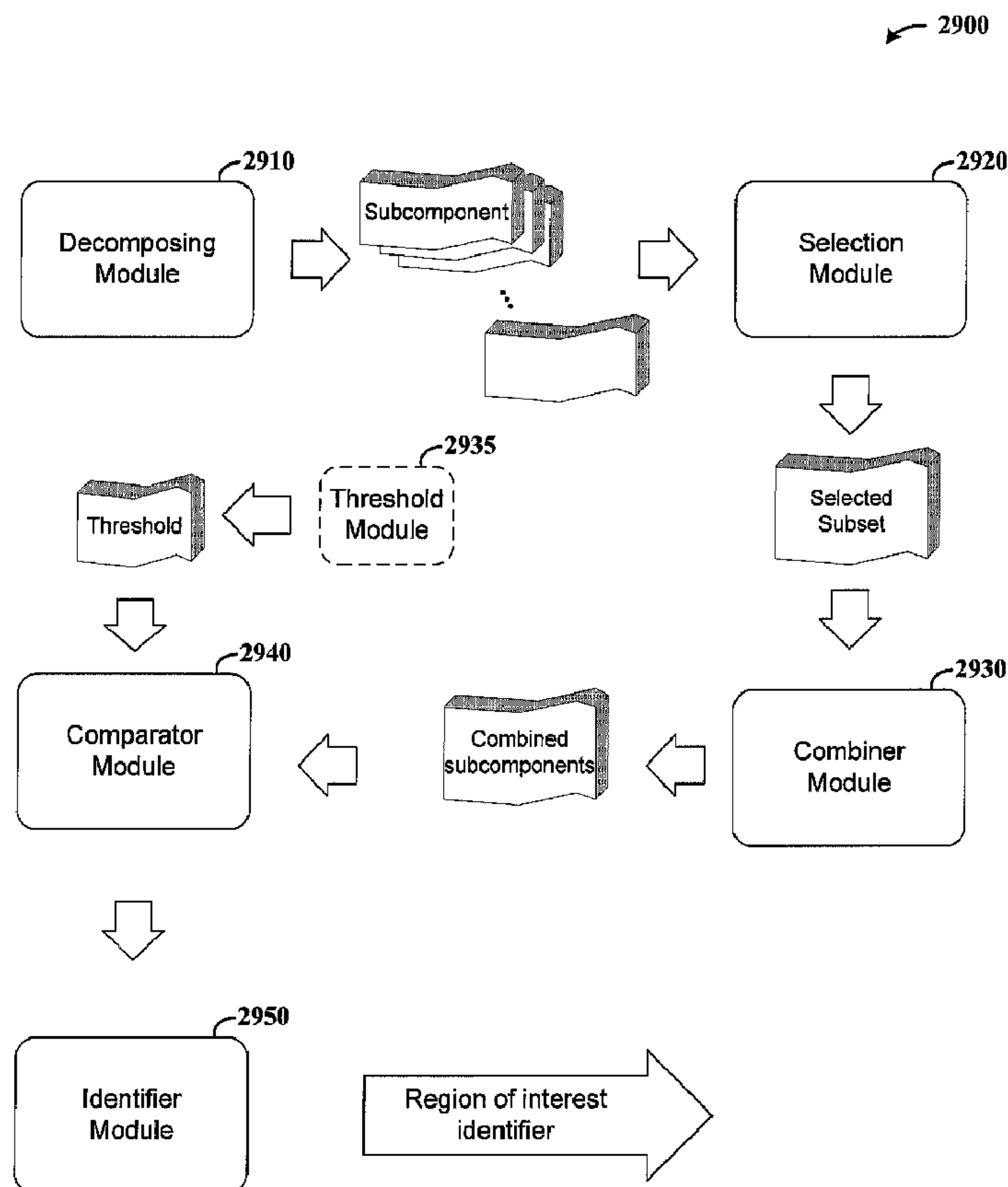




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(54) **Titre : DETECTION DE POINTS DE REPERE DANS DES SIGNAUX PHYSIOLOGIQUES**
(54) **Title: DETECTING FIDUCIAL POINTS IN PHYSIOLOGICAL SIGNALS**



(57) **Abrégé/Abstract:**

Various aspects are directed to identifying a region of interest in a physiological signal. As may be consistent with one or more embodiments, the physiological signal is decomposed into subcomponents, and a subset of the subcomponents is selected based

(57) **Abrégé(suite)/Abstract(continued):**

upon overlap of spectral energy with expected spectral energy of the region of interest, in at least one of the subcomponents. At least two of the subcomponents in the subset are combined and compared to a threshold, with the comparison being used to identify the location of the region of interest.

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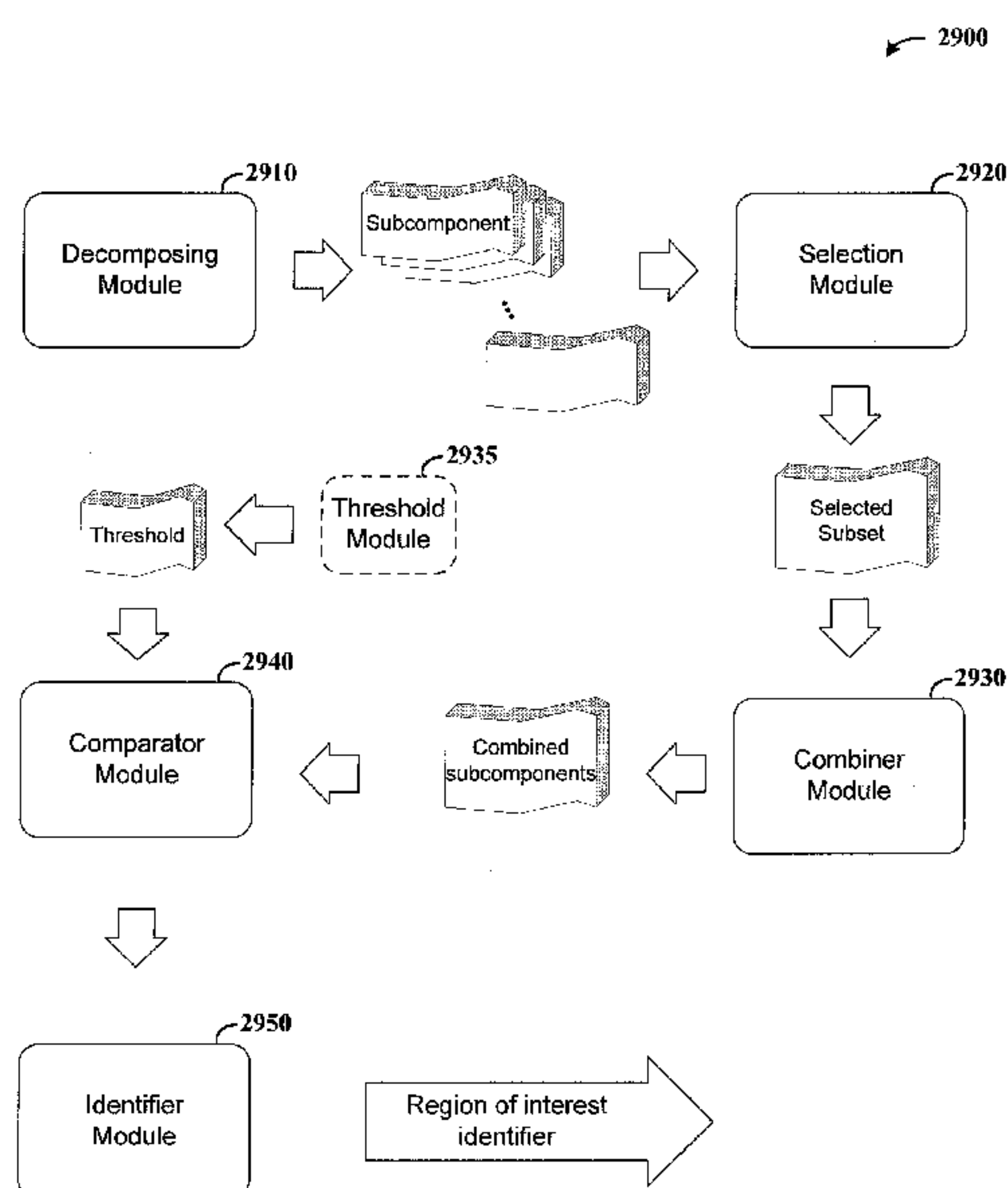


FIG. 29

(57) Abstract: Various aspects are directed to identifying a region of interest in a physiological signal. As may be consistent with one or more embodiments, the physiological signal is decomposed into subcomponents, and a subset of the subcomponents is selected based upon overlap of spectral energy with expected spectral energy of the region of interest, in at least one of the subcomponents. At least two of the subcomponents in the subset are combined and compared to a threshold, with the comparison being used to identify the location of the region of interest.

DETECTING FIDUCIAL POINTS IN PHYSIOLOGICAL SIGNALS

FIELD

5 Various aspects of the present disclosure relate to the processing of physiological signals, and more particular aspects relate to detection of fiducial points in quasi-periodic signals, such as identifying QRS complexes in an ECG.

BACKGROUND

10 Implantable and external devices are used to monitor physiologic signals of human and animal subjects. These devices may incorporate various types of sensors and can measure and record signals from those sensors for processing by a system or monitoring center located remote from the subject, or in other cases, the device may perform some or all of the desired signal processing to extract information from the measured signals and
15 forward the resulting information to a remote system for display, recording, or further processing.

The signal processing is performed to extract information from the signal in order to assess the physiological condition of the monitored subject and often to evaluate the response of the subject to a therapy or experimental protocol. Examples of quasi-periodic
20 physiological signals include respiration, ECG, blood pressure, blood flow, and photoplethysmography measurements of blood gases. Clinicians can use this information to make therapy decisions and researchers can use this information to assess the safety and utility of experimental therapies. This information may also be used for closed loop control of therapy delivery. In other examples, measurements of peripheral nerve activity (PNA),
25 respiration, blood oxygen, blood glucose, EEG, EMG, heart sounds and blood flow signals are processed to extract information for clinical or research purposes.

There is an increasing reliance on automatic processing to extract information in order to reduce labor and costs and to more consistently and accurately evaluate the physiologic condition of the subject. In therapeutic devices automatic detection of fiducial
30 points, such as the QRS complex of an ECG, is often useful for feedback control. Accurate QRS detection in ECGs is also the foundation for identifying arrhythmias and measuring intervals in ECGs recorded for research or diagnostic purposes. However, highly accurate QRS detection is often compromised when the ECG contains significant levels of in-band noise (*e.g.*, ambulatory ECG recordings). This often results in high rates of false positive and

false negative detection of arrhythmias mandating manual (*e.g.*, human operator) over-read of the signals following processing by an algorithm.

In some physiologic signal processing applications, automated analysis is complicated by the fact that measured signals are the result of activity of multiple sources, referred to as multi-source signals. An example of a multi-source signal is ECG measured on the surface of the body where electrical activity is sensed from the atria and ventricles as well as skeletal muscles. It is useful, for example, to observe atrial activity independent of ventricular activity in order to improve the detection of atrial arrhythmias. Current techniques for providing signal source extraction of multi-source signals, such as independent component analysis (ICA), assume independence of sources and performance is compromised when this assumption is invalid, such as is the case when separating atrial and ventricular activity in an ECG. In addition, ECGs are often recorded from ambulatory subjects using a small number of sensing leads, further complicating signal source extraction and fiducial point detection (*e.g.*, atrial activity detection) due to the mixing of sources inherent in a small number of leads.

Other signals, such as peripheral nerve activity (PNA) and brainstem auditory response, have proven difficult to analyze because of very low signal-to-noise ratio (SNR). Visual analysis of these signals is often inadequate to detect important features and obtain a quantitative evaluation.

Many physiological signal processing techniques have been difficult to successfully implement under certain conditions, particularly when processing signals from ambulatory subjects where the signals are often quite noisy. For example, measurements of ECG parameters such as heart rate, QT interval, and PR interval may contain errors due to inaccurate QRS detection. When noise is present due to patient movement or when there is background acoustical noise (*e.g.*, when the patient is being transported in a helicopter, ambulance, or automobile), measurements of blood pressure obtained from an arm cuff may contain errors as a result of inaccurate detection of the changes in acoustical sounds or other measurements that correlate with systole or diastole. Detection of ventricular and atrial arrhythmias in ECG may have excessive incidence of false positives due to the inability of a signal processing algorithm to provide accurate detection, particularly in the presence of noise. Because of lack of confidence in the accuracy of results, human review has often been used to confirm results or correct errors made by automated analysis algorithms.

Inaccuracies in performance can also result in excess telecom costs when monitoring ambulatory subjects. For example, some types of ambulatory ECG monitoring devices

employ on-board signal processing to detect arrhythmias and forward the detected arrhythmias to a monitoring center where they are further processed and reviewed by a human being using a data review system. Because of limitations in existing algorithms, there is a high rate of false positive arrhythmia detections in the ambulatory device that results in a high volume of data transmitted from the patient to a monitoring center. These aspects can result in excessive telecommunications expense, the need for additional memory in the ambulatory monitoring device, and additional expense to manually review the data received at the monitoring center.

Data compression is often used to reduce the volume of data needed for storage or transmission. Various methods of ECG data compression are limited in their ability to provide high levels of compression with minimal signal distortion in part due to the presence of noise that hinders accurate QRS detection. Accurate QRS detection can facilitate the use of more efficient data compression techniques that can reduce the volume of data that must be stored in memory on an ambulatory monitoring device as well as reduce the volume of data transmitted from the monitored subject. In certain applications, this can result in a reduction in telecom expense and a reduction in power consumption in the ambulatory monitoring device, leading to a reduction in the device size and extension of battery life.

The presence of noise in physiological signals and its negative impact on accurate fiducial point detection can be a limiting factor in providing accurate and consistent computerized evaluations and extraction of information. The noise is especially problematic when its frequency content falls within the bandwidth of the signals of interest (referred to as in-band noise). For example atrial signals can be contaminated by electrical activity of the ventricle, and ECG signals can be contaminated by EMG from the skeletal muscles. The plethora of signal sources contained within a limited number of channels measured in a surface ECG, with each channel containing mixed interdependent signals, renders accurate detection of fiducial points in these multisource signals a very difficult problem. This problem is further complicated when signals are acquired from closely spaced electrodes and are contaminated by noise, as is often the case when monitoring patients outside a clinic or hospital. This characterization is not only common to electrocardiogram (ECG) signals acquired with surface or subcutaneous leads but is also common to electrograms (EGM) measured with intracardiac leads, non-invasive blood pressure signals, pulse oximetry signals, peripheral nerve activity (PNA) recordings, signals representing non-invasive measurements of intracranial pressure, and other physiologic signals collected from ambulatory subjects. Current filtering techniques such as bandpass filtering are effective in

removing noise without distorting the signal when the spectral content of the noise and signal are separated in the frequency domain. Many filtering techniques capable of removing in-band noise such as independent component analysis require that noise and signal content are uncorrelated and independent, an inaccurate assumption for most physiological signals.

Certain efforts have been made toward improving detection of fiducial points in noisy quasi-periodic physiological signals, but the success of these efforts has been limited. These and other matters have presented challenges to the design and implementation of devices, systems and methods for processing physiological signals.

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SUMMARY

Various aspects of the present disclosure are directed to devices, methods and systems involving physiological signal processing, in a manner that addresses challenges and limitations including those discussed above.

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Various aspects of the present invention are directed to identifying a region of interest in a physiological signal, such as identifying a QRS complex in an electrocardiogram (ECG). In an embodiment, the physiological signal is decomposed into subcomponents, and a subset of the subcomponents is selected based upon overlap of spectral energy in one or more of the subcomponents with expected spectral energy of the region of interest. Two or more of the subcomponents are combined, and the combination is compared to a threshold. The location of the region of interest within the signal is identified based on the comparing.

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Another embodiment is directed to an apparatus including a computer-based circuit and a plurality of modules as follows, for identifying such a region of interest in a physiological signal. A decomposing decomposes the physiological signal into subcomponents, and a selection module selects a subset of the subcomponents based upon a degree of overlap of spectral energy (in one or more of the subcomponents) with expected spectral energy of the region of interest. A combiner module combines at least two of the subcomponents in the subset, and a comparator module compares the combined subcomponents to a threshold. An identifier module identifies the location of the region of interest in the physiological signal, based on the comparing.

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The above summary is not intended to describe each embodiment or every implementation of the present disclosure. The figures and detailed description that follow more particularly exemplify various embodiments.

BRIEF DESCRIPTION OF THE DRAWINGS

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

5 FIG. 1 shows a data flow diagram of Multi-Domain Signal Processing applied to denoising a captured signal, consistent with an example embodiment of the present disclosure;

 FIG. 2 shows a data flow diagram of Multi-Domain Signal Processing applied to extraction of a signal from a captured physiological signal, consistent with an example
10 embodiment of the present disclosure;

 FIG. 3 shows scatter plots, in which FIG. 3a shows a scatter plot of subcomponents of a noisy signal before application of PCA and ICA, and FIG. 3b shows a scatter plot of subcomponents of the same noisy signal of FIG. 3a after application of PCA and ICA, in accordance with one or more example embodiments of the present disclosure;

15 FIG. 4 illustrates a data flow diagram for computing a dynamic signal-to-noise ratio, according to an example embodiment of the present disclosure;

 FIG. 5 shows an example of denoising and QRS detection in a rabbit ECG with non-sustained ventricular tachycardia, and illustrates a confidence signal computed using dynamic signal-to-noise ratio updated for every cardiac cycle, according to an example
20 embodiment of the present disclosure;

 FIG. 6 shows an example of denoising on a 3-lead ECG signals using PCA, ICA, and MDSP, according to another example embodiment of the present disclosure;

 FIG. 7 shows results representing denoising of MDSP, PCA, and bandpass filtering, in connection with another example embodiment of the present disclosure;

25 FIG. 8 shows denoising for a single channel noisy ECG signal, in connection with an example embodiment of the present disclosure;

 FIG. 9 shows an example of atrial activity extraction from a surface ECG signal containing atrial flutter, in connection with another example embodiment of the present disclosure;

30 FIG. 10 shows an example of atrial activity extraction from a surface ECG signal containing atrial fibrillation, in connection with another example embodiment of the present disclosure;

FIG. 11 shows an example of denoising on single channel peripheral nerve activity (PNA) recording corrupted with noise, in connection with another example embodiment of the present disclosure;

FIG. 12 shows a data flow diagram for computing an emphasis signal and finding a feature point in a signal waveform, and example signal waveforms and corresponding emphasis signals and feature point markers, according to another example embodiment of the present disclosure;

FIG. 13 shows an example of QRS detection for a challenging ECG recording, according to another example embodiment of the present disclosure;

FIG. 14 shows a data flow diagram for computing an emphasis signal corresponding to a respiration pattern, according to an example embodiment of the present disclosure;

FIG. 15 shows an example of respiration signal extraction in an ECG recording, according to another example embodiment of the present disclosure;

FIG. 16 illustrates an apparatus for efficient wireless communication of an ECG along with data flow diagrams for ECG signal compression, according to an example embodiment of the present disclosure;

FIG. 17 shows an example of 3D ECG cycle plot as an intermediate step in ECG compression, in connection with another example embodiment of the present disclosure;

FIG. 18 shows a system for evaluating ECG strips captured by an ambulatory monitoring device along with data flow diagrams for determining if a strip contains an arrhythmia and assessing the validity of the result, according to an example embodiment of the present disclosure;

FIG. 19 illustrates an apparatus for improving the signal-to-noise ratio of an ECG signal and related flow chart for processing an ECG signal, in accordance with one or more example embodiments of the present disclosure;

FIG. 20 illustrates an approach for spatially selective filtering, and partitioning of a cardiac cycle of an ECG signal, according to another example embodiment of the present disclosure;

FIG. 21 illustrates a signal flow diagram for an embodiment of denoising applied to an ECG signal, according to another example embodiment of the present disclosure;

FIG. 22 illustrates a system for denoising an ECG signal, in accordance with one or more example embodiments of the present disclosure;

FIG. 23a shows plots representing SNR_A improvement and QSR for MDSP denoising of ECG recordings with controlled levels of imputed band-limited white noise, in accordance with a particular embodiment;

FIG. 23b shows example ECG recording according to a controlled application with 0 dB SNR_A imputed band-limited white noise (top) and post-MDSP denoising (bottom), in accordance with another example embodiment;

FIG. 24 shows modules/flow functions for denoising a signal, in accordance with another example embodiment of the present invention;

FIG. 25 illustrates the processing steps for an embodiment of QRS detection in an ECG signal;

FIG. 26 illustrates an example of an input ECG signal and intermediate processing results;

FIG. 27 illustrates an example of an input ECG signal, computed emphasis function, and adaptive threshold function for QRS detection;

FIG. 28 shows expected spectral energy and noise spectral energy, in accordance with one or more embodiments; and

FIG. 29 shows an apparatus having several modules, in accordance with another example embodiment.

While the disclosure is amenable to various modifications and alternative forms, specifics thereof have been shown by way of example in the drawings and will be described in detail. It should be understood, however, that the intention is not to limit the disclosure to the particular embodiments described. On the contrary, the intention is to cover all modifications, equivalents, and alternatives falling within the scope of the disclosure including aspects defined in the claims.

DETAILED DESCRIPTION

Various example embodiments of the present disclosure relate to circuits, devices and systems that process physiological signals, and in many implementations, that process such signals to reduce noise, extract information, characterize signals, and/or compress the volume of data. While the present disclosure is not necessarily limited to such applications, various aspects of the disclosure may be appreciated through a discussion of examples using this context.

As discussed hereinabove, various circuits, devices and systems that process noisy physiological signals (*e.g.*, ambulatory ECG recordings) are incapable of accurately

identifying signal characteristics, such as those relating to an event represented by signals (e.g., a cardiac event) or a feature point of the signal (e.g., T-wave offset), in a manner that is sufficiently reliable to permit accurate computer-based identification of the signal characteristics. This presents a significant challenge to the implementation of such circuits, devices and systems for carrying out signal evaluation. Accordingly, various aspects of the present invention are directed to addressing these challenges, with various implementations directed to denoising signals and providing the denoised signals for evaluation in identifying signal characteristics such as events. Other aspects are further directed to using the denoised signals to automatically identify such characteristics or events with a degree of reliability facilitated via the denoising.

Many embodiments described here are referred to as including an approach involving “multi-domain signal processing” (MDSP), for which many different embodiments are described by way of example (e.g., as involving signal processing in two or more domains). In connection with various embodiments, the term multi-domain filtering (MDF) is used herein to refer to embodiments that use one or more MDSP-based embodiments to denoise physiologic signals. Various embodiments are also directed to circuits, devices and systems for processing a broad range of physiological signals including but not limited to signals corresponding to ECG, blood pressure, respiratory, heart sounds, EEG, peripheral nerve activity, activity, temperature, photoplethysmography, tissue impedance, blood glucose, and EMG. Certain embodiments are further directed to one or more of: improving the accuracy and consistency of information provided under a broad range of use scenarios, extending battery life of monitoring devices, providing a desirably-sized monitoring device (e.g., a reduced, or small size), and improving and/or reducing the need for human review of analysis results and the associated expense of doing so.

In the following discussion, reference is made to cited references listed in a numbered order near the end of this document. 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.

According to an example embodiment, physiological signals of a subject are captured by an implantable or external device. This device can be implemented as a part of a system that measures, processes, and evaluates physiological data from animal or human subjects for research, therapy titration, diagnosis, or delivery of medical care. The device may

temporarily store the processed signals for later transmission to a system remote from the subject for further processing, display, and reporting to a medical care provider or researcher. The processed signals may be transmitted in real time or they may be used within a therapy delivery device as part of a system to control or advise administration of a therapy. In yet another embodiment, unprocessed or partially processed signals may be transmitted to a device or system located outside the body of the subject for processing, display, review, reporting, or retransmission to another system.

In many applications, physiological signals processed in accordance with the embodiments discussed herein are multisource signals, meaning that the signal observed by electrodes or leads includes components that are the result of many physiological processes and sources that are often interdependent. For example, an ECG signal measured at the surface of the body may include signals emanating from sources such as the atria, ventricles, electrical noise (*e.g.*, from sources outside the body), noise from muscular electrical activity, signals resulting from pathologies such as conduction defects, scars in tissue from myocardial infarction, ischemia resulting from reduced blood flow to a region of the heart, and other representative examples. In these and other contexts, various embodiments of the present disclosure are directed to addressing challenges relative to the analysis of ECG signals, such as those that benefit from independent analysis for denoising and evaluation of cardiac function.

The terms “quasi-periodic”, “signal wave”, “feature point”, “parameter”, and “event” are used in connection with the discussion of various embodiments as follows. The term quasi-periodic refers to a periodic signal with a period with a cycle length that varies with time, and a signal wave is a particular portion or aspect of a period of a physiological signal. For example, an ECG signal may include signal waves referred to as P, Q, R, S, T and U waves. Another example signal wave is the QRS complex portion of an ECG (*e.g.*, the portion of an ECG signal corresponding to the depolarization of the left and right ventricles). In an arterial pressure signal, the dicrotic notch is a signal wave. Other examples of signal waves include changes in temperature and motor activity that occur during an estrous cycle of a mammal, Korotkov sounds that occur in conjunction with the cardiac cycle when performing a cuff-based non-invasive blood pressure measurement, or changes in tissue impedance that correspond to a respiratory cycle.

Regarding the term “feature point,” many physiological signals can be characterized as having features and parameters. A feature point is an identified point within a physiological signal, which may be useful for characterizing the signal and related

characteristics of the signal's source. For instance, in heart-related signals, such as ECG, arterial pressure, and blood flow, most cardiac cycles have a feature point or points of interest. Examples include a point corresponding to the onset of the Q-wave (*e.g.*, Q-wave onset) or offset of the T-wave (*e.g.*, T-wave offset) in an ECG or systole in an arterial blood pressure signal. Each of these feature points can be described by time of occurrence and amplitude, and consecutive feature points can be combined to form a feature signal.

With further respect to the term "feature point," it is sometimes useful to combine feature points over a predetermined period of time, referred to as a feature signal, to compute a parameter. For example, systole feature points can be combined to compute a mean systolic pressure. Computing a parameter can have the effect of reducing or eliminating short-term physiological fluctuations (*e.g.*, changes with respiration) in the feature signal that are not of interest to the user.

Regarding the term "event," physiological signals may include information that can be used for identifying the onset and offset of an event, or simply the fact that an event occurred. For example, when monitoring the ECG of a subject it may be useful to know that an arrhythmic event such as ventricular tachycardia or atrial fibrillation has occurred. In additional embodiments, information is combined from multiple signals to compute features and parameters. For example, the QA interval (the time difference between Q-wave onset and the upstroke of an arterial pressure wave) can be used as a surrogate for cardiac contractility, and employs both a pressure and an ECG signal from a subject. QA interval feature points are computed for a cardiac cycle, can be used to create a feature signal, and averaged over a predetermined period of time to create a QA parameter.

The term "time-spectral distribution" refers to the distribution of spectral content over time within a cycle of a quasi-periodic signal. For example, cycles of an ECG signal can be identified and can be divided into time segments wherein the time segments contain a characteristic spectral energy that is relatively consistent. This concept can also be referred to as time-spectral distribution of subcomponents derived from decomposition of an input signal. In some embodiments spectral energy can be characterized by the subcomponents present within a time segment. In some embodiments the term "spatially selective filtering" is used to describe removing subcomponents within a time segment based upon time-spectral distribution.

Regarding the terms referring to a "desired physiological signal," such a signal refers to a signal that may contain useful information in the context of the application. The desired signal is to be extracted from a captured signal that may also contain noise. This desired

signal may correspond, for example, to a physiological signal within captured input signals that include the physiological signal and noise. In this context, the noise may also include a physiological signal that is not the desired physiological signal. For instance, where a desired physiological signal is an ECG signal generated by a subject's heart tissue, other physiological signals included with an input signal, such as a respiratory signal and EMG of the subjects skeletal muscle, are not desirably extracted.

A "captured signal" is a signal that is sensed and recorded (*e.g.*, digitized), and may also be conditioned. Conditioning of the signal may involve amplification and the application of a conventional filter to remove much of the noise that is outside the bandwidth of the signal. Following digitization, additional filtering may be applied to further remove noise from outside the signal bandwidth, typically using linear filtering techniques such as a finite impulse response filter.

A "denoised" signal is a captured signal that has been processed to remove noise, such as by removing undesirable signal components or subcomponents having spectral content within a bandwidth of a selected (*e.g.*, target, or desired signal), or in-band noise. Alternately, a denoised signal can be thought of as the desired signal that has been extracted from a captured signal. Denoising can be useful for rendering clarity to the desired signal as a result of suppression of undesired signal components (*e.g.*, noise), hence making the desired signal more suitable for analysis.

In accordance with various embodiments, a region of interest is identified for a quasi-periodic physiological signal as follows. The physiological signal is decomposed into subcomponents, and a subset of the subcomponents is selected based upon a degree of overlap of spectral energy (*e.g.*, one half or more), in at least one of the subcomponents, with expected spectral energy of the region of interest (*e.g.*, with a known/expected energy of a particular characteristic in the physiological signal). In some implementations, each subcomponent is selected based upon a degree of overlap of spectral energy of that subcomponent with the expected spectral energy. At least two of the subcomponents in the subset are combined, and the combined subcomponents are compared to a threshold. The location of the region of interest is identified based upon the comparison. Such an approach may, for example, be carried out in the context of Figure 24, in which the subset of subcomponents is selected at block 2420, the at least two subcomponents are combined at block 2430, and the output from block 2430 is compared to the threshold and used to identify the location of the region of interest. Such approaches may also be carried out in the context of Figure 29, also discussed further below.

The aforesaid approach to identifying a region of interest is applicable to a variety of physiological signals. For instance, such an approach may involve identifying the location of the QRS complex in an ECG, using an expected spectral energy of the QRS complex. Other such signals may include, for example, blood pressure, photoplethysmography or electrical impedance signals with respective regions of interest pertaining to cardiac stroke, dicrotic notch, systole or diastole. In the following discussion, reference is made to various example embodiments involving the identification of the QRS complex in an ECG, by way of example. However, it should be understood that the embodiments discussed in this context may be implemented for identifying different types of regions of interest in a variety of different signals.

The subcomponents are combined in one or more of a variety of manners, to suit particular applications. In some embodiments, the subcomponents are combined by computing a point-wise product of the subcomponents. In other embodiments, the subcomponents are combined by computing a linear combination of the subcomponents. In still another embodiment, the subcomponents are combined by cross-correlating the subcomponents. In yet other embodiments, two or more of these approaches (point-wise product, linear combination, and cross-correlation) are used in conjunction with one another in combining subcomponents.

In some embodiments, the subcomponents are decomposed by generating time-synchronized subcomponents (*e.g.*, for ECG subcomponents). This approach may include, for example, applying one or more of a non-orthogonal wavelet transform, an undecimated wavelet transform, a stationary wavelet transform, and a shift-invariant wavelet transform to an ECG.

The threshold discussed above is based upon one or more of a variety of characteristics useful for identifying a subcomponent as pertaining to a particular region of a signal. This estimate may, for example, be made by estimating the level of noise energy using one of variance, zero crossings, and amplitudes of peaks and valleys in the ECG. In some embodiments, the threshold is computed based upon an estimated level of noise energy in an isoelectric portion of an ECG. In this context, the isoelectric portion of the ECG describes the region between the end of the T-wave and the onset of the P-wave in a cardiac cycle. Isoelectric also has a connotation that refers to a flat line in an electrical measurement (*e.g.*, which is not descriptive when the isoelectric portion of the ECG is noisy).

In some embodiments, the threshold is computed by first selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to

expected noise spectral energy in the ECG. The portion of subcomponent spectral energy attributable to noise spectral energy may, for example, be a portion having at least one-half of the total energy of the subcomponent. At least two of the noise subcomponents are combined, and the threshold is set based upon the combination of the at least two of the noise subcomponents. Selecting the subset of noise subcomponents may, for example, be carried out by selecting a subset of noise subcomponents using a portion of subcomponent spectral energy attributable to expected noise spectral energy in a portion of the ECG that excludes the QRS complex. In some embodiments, combining at least two of the noise subcomponents includes combining noise subcomponents of a portion of the ECG outside the QRS complex. In other embodiments, combining at least two of the noise subcomponents includes computing one of a point-wise product of the at least two of the noise subcomponents, a linear combination (*e.g.*, weighted sum) of the at least two of the noise subcomponents, and a cross-correlation of the at least two of the noise subcomponents.

In accordance with other example embodiments of the present disclosure, in-band noise of a physiological signal is reduced using a technique involving a multi-domain filtering-type of approach, referred to in connection with various embodiments as MDF, as may be implemented with one or more circuits, devices or systems. Resulting (output) signals are thus denoised, in the context that at least some noise components in the physiological signal have been removed, relative to the resulting signal (*e.g.*, as re-generated from components of the physiological signal, in a different domain). These denoised signals facilitate the use of computer-type circuits to automatically evaluate the signals for identifying characteristics therein with a high degree of reliability.

In one embodiment, signals captured in a first domain are decomposed into subcomponents in a second domain. To remove in-band noise from the captured physiological signal, the subcomponents in the second domain are processed based upon their time-spectral distribution using, for example, spatially selective filtering or principal component analysis to identify those subcomponents that are primarily associated with noise. Those subcomponents that are identified as primarily associated with noise are removed and the residual subcomponents (those not having been removed) are combined to reconstruct a denoised signal in the first domain. A subcomponent occurring in a time window within a cycle of the quasi-periodic signal is said to be associated with a signal if it contains frequencies within a band that has previously been characterized as being present in the signal. A subcomponent within a time segment is considered to be primarily associated with noise if the energy associated with the signal is attributed primarily (*e.g.*, more than

half) to noise. In some embodiments, a larger tolerance for noise is set, to permit signals with larger noise content to be processed without distorting the signal (*e.g.*, where a desired signal is expected to include a substantial portion of noise). Under such conditions, processing of the detected signal can be carried out with the understanding that noise forms a majority of the power in the subcomponent. In some implementations, a signal within a time segment is considered to be primarily associated with noise when at least one-half of the signal's energy within that time segment is noise energy (*i.e.*, energy not affiliated with the desired signal). Likewise, a subcomponent within a time segment is considered to be primarily associated with the desired signal if at least one-half (*e.g.*, 50%) of its energy is within the band of frequencies characterized as being present in the desired signal.

In the context of various embodiments, references to the removal of subcomponents may not involve any removal or modification of the subcomponents, but rather involve a selective combination of those subcomponents that have been determined to be desirable. For example, for a physiological signal having various subcomponents, certain components can be identified as likely representing components of a signal corresponding to a particular physiological characteristic that is to be analyzed. These identified (desirable) components can be selectively combined, leaving behind other subcomponents. In this context, undesirable subcomponents are not necessarily removed, but rather, have not been used when forming a recombined signal corresponding to a received physiological signal.

The subcomponents that result from decomposition in MDSP embodiments are also used in other aspects of physiologic signal processing, for various embodiments. An assumption of MDSP is that a signal wave in a multisource physiological signal can be represented using a small number of subcomponents that contain most of the energy of the signal wave. For example, in an ECG decomposed into 4 levels using a discrete wavelet transform, a group of 3 subcomponents may contain most of the energy found in the P-wave of an ECG while most of the energy of a QRS complex may be contained in 4 subcomponents. A subcomponent or its time segment is said to be primarily associated with the desired signal within that time segment if at least one-half of the energy of the subcomponent is signal energy. In various implementations, a subcomponent may be associated with more than one signal wave.

Subcomponents or their time segments associated with a signal wave can be identified, isolated, and used to construct a signal wave independent of other signal waves in a multisource signal. For example, subcomponents associated with a P-wave of an ECG can be identified within the second domain independent of ventricular electrical activity, and

used to reconstruct a denoised signal representative of atrial activity. Another group of subcomponents is associated with the T-wave of an ECG, and can be used in the extraction of repolarization activity. Signal source extraction of the signal subcomponents in a multisource signal can lead to more accurate analysis and evaluation of certain physiological functions. Another embodiment involving signal source extraction is directed to the extraction of fetal ECGs from ECG signals captured from a pregnant female. Yet another example embodiment is directed to extracting an oxygen saturation signal from a photoplethysmography signal.

Subcomponents can also be used to compute a denoised emphasis signal that exaggerates the features of particular signal waves to facilitate the identification and detection of feature points. For example, subcomponents associated with the T-wave of an ECG can be identified and used to compute a denoised emphasis signal that exaggerates the T-wave relative to other ECG signal waves to facilitate accurate and consistent detection of T-wave offset.

In some embodiments, physiological signal subcomponents are used to compute a dynamic signal-to-noise ratio (dSNR) that represents the ratio of signal energy relative to noise energy on a sample-by-sample basis, or longer period of time such as for one or more cardiac cycles of an ECG. The dSNR can be used for computing a validity signal for assessing the accuracy and reliability of information derived from the captured physiological signal. In some embodiments, the dSNR is used to directly assess the accuracy and reliability of derived information. If dSNR for a cardiac cycle or a portion of a cardiac cycle is low, for example, certain information derived from the ECG for that cardiac cycle may not be accurate and, if it is very low, a cardiac cycle may be uninterpretable.

MDF denoising and other aspects of one or more MDSP-based embodiments are directed to facilitating the efficient compression of physiological signals, which can be used to reduce the volume of data corresponding to sensed signals. Reducing the amount of data (*e.g.*, by eliminating or omitting noise) serves to help the efficient storage of data, and also to reduce the amount of data needed in communications which can be helpful, for example, to simplify wireless transmission protocol. For example, various MDSP-based embodiments involve compressing ECG signals at compression rates of 15:1 to 20:1 (*e.g.*, relative to an original physiological signal prior to denoising) with minimal signal distortion. An MDSP approach is used to achieve accurate identification of cardiac cycles of a denoised ECG, and significantly reduce in-band noise. These identification and denoising approaches are used in connection with one or more of a variety of compression algorithms, in accordance with

various embodiments, together with identified redundancies in adjacent cardiac cycles to achieve efficient compression.

In some embodiments, an MDF-based approach is used to remove noise from peripheral nerve activity (PNA) signals and facilitate the accurate quantification of PNA. In these and/or other embodiments, an MDSP-based approach is used across a broad range of physiological signal processing applications with related processing without necessarily relying upon assumptions regarding the independence of signal and noise sources, and/or based upon an assumption (for processing) that physiological signals being processed are quasi-periodic signals.

In the following discussion, reference is made to various documents or other references listed near the end of this patent document, by numerals within square brackets. The information in the documents/references to which these numerals refer may be implemented in connection with one or more example embodiments.

Various embodiments are directed to processing signals from ambulatory subjects under conditions in which noise is common and measurements are often obtained using a limited number of electrodes or sensors. Such signals often include aspects of signals emanating from multiple sources that are generally interdependent. For example, heart activity in an atrial chamber usually initiates activity in a ventricular chamber. Statistically, this situation is characterized by mutual dependency of signal sources contaminated by noise. This violates a central assumption of independence of sources that is fundamental to successful application of many current techniques used to separate signal source and noise in multisource signals, such as independent component analysis (ICA) [1]. Various embodiments are directed to processing related signals, without necessarily assuming such independence, to obtain a denoised signal for a desired signal (*i.e.*, a signal corresponding to a characteristic to be detected).

The problem of extracting source signals from the multisource physiological signals contaminated with noise can be expressed mathematically as:

$$x(t) = As(t) + n(t) \quad (1)$$

where $s(t) = (s_1(t), \dots, s_p(t))$ is a vector of source signals that are mixed together by an unknown mixing matrix A of size $M \times P$, where M is a number of observed signals and P is a number of source signals and additive noise $n(t) = (n_1(t), \dots, n_p(t))$. The extraction of source signals is achieved by estimating the inverse of a mixing matrix A and computing the

denoised and separated signals $s(t) = (s_1(t), \dots, s_P(t))$ from the observed signals $x(t) = (x_1(t), \dots, x_M(t))$ according to the formula $s(t) = A^{-1}x(t) - A^{-1}n(t)$. When the number of leads M is less than the number of sources P , which is most often the case in physiologic monitoring, the mixing matrix $A_{M \times P}$ is underdetermined. In addition, the sources that make up multisource physiological signals are often not independent. In this situation the matrix $A_{M \times P}$ is not invertible, rendering it difficult to estimate an inverse of the $M \times P$ mixing matrix and recover unmixed sources from mixed observations, as is the case when using ICA techniques.

Turning now to the figures, Figure 16 shows an example system implementation involving a specific aspect of the implementation of various embodiments, as applicable to circuitry, related programming and corresponding data that relates to electrodes, signal lines and processing circuitry for controlling and evaluating ECG signals obtained via electrodes. While various aspects of Figure 16 are discussed in greater detail later in this document, reference to this figure may be made in connection with the following (and above) discussion in considering functionality of such a system (e.g., via modules implemented in code and used by such a system to process data signals and both generate and evaluate denoised signals).

Various example embodiments are directed to an apparatus, and other embodiments are directed to a system, which carry out the one or more of the various approaches discussed herein. Still other embodiments are directed to non-transitory computer-readable medium having instructions stored thereupon that, when executed by a computer-type processor, carry out the respective steps discussed herein. In this context, another example embodiment is directed to an apparatus including a computer-based circuit configured and arranged with a plurality of modules for identifying a region of interest in a physiological signal, such as by identifying a QRS complex in an electrocardiogram (ECG).

Figure 29 shows one such example circuit 2900, having modules as follows. A decomposing module 2910 decomposes the physiological signal into subcomponents. A selection module 2920 selects a subset of the subcomponents based upon a degree of overlap of spectral energy, in at least one of the subcomponents, with expected spectral energy of the region of interest. A combiner module 2930 combines at least two of the subcomponents in the subset, and a comparator module 2940 compares the combined subcomponents to a threshold. An identifier module 2950 identifies the location of the region of interest in the physiological signal based on the comparing. In some embodiments, the apparatus also includes a threshold module 2935 that computes the threshold based upon an estimated level

of noise energy. In some implementations involving the detection of a QRS complex, the threshold module 2935 computes the threshold based upon an estimated level of noise energy in an isoelectric portion of an ECG. One or more of these modules may, for example, carry out one or more of the other steps discussed above as related to each
5 respective module's function (*e.g.*, threshold determination, subset selection, subcomponent combination, and signal decomposition).

Figures 1 and 2 show signal denoising and signal source extraction, respectively, as carried out in a three-step process in accordance with another example embodiment of the present disclosure. Processes 102 and 202 in Figures 1 and 2, respectively, involve
10 decomposing an input signal 101 and 201 into subcomponents in a second domain. Decomposition processes 102 and 202 are performed using one of a variety of transforms that provide accurate signal reconstruction. Such transforms may involve using, for example, a discrete cosine transform, a wavelet related transform, a Karhunen-Loeve transform, a short-time Fourier transform, a Gabor transform, or a filter bank, as may be
15 implemented by signal-processing circuitry. In one embodiment decomposition is performed in a manner that preserves time relationship between the subcomponents. Examples of such methods are: non-orthogonal wavelet transform, undecimated wavelet transform, shift-invariant or stationary wavelet transform. In Figures 1 and 2, the set of subcomponents resulting from the decomposition processes 102 and 202, respectively is referred to as
20 D2sub. The dimension of the first domain is defined by the number of observed, or captured, signal channels. The dimension of the second domain is defined by the number of channels multiplied by the number of subcomponents in each channel. The second step involves the identification of signal and noise subcomponents in the second domain followed by removal of unwanted subcomponents, including noise.

25 Various embodiments herein refer to identifying subcomponents based upon their time-spectral distribution. Such identification may involve, for example, identifying subcomponents based upon their relationship with one another as a function of time. For example, a time-based separation of the subcomponents in a particular signal can be used to identify a characteristic to which the subcomponents pertain, as relative to an expected
30 distribution for the particular signal. In the context of a cardiac signal, a time-based position can be used to identify portions of a cardiac cycle.

Referring to Figure 1, subcomponents of the set D2sub that are primarily associated with noise are identified and removed to create a subset D2s of residual denoised subcomponents, to denoise a desired signal. Referring to Figure 2, if signal source

extraction is performed, such as when an atrial activity signal is extracted from a surface ECG, the subcomponents associated with the signal sources to be extracted are identified and the other subcomponents are removed to create a subset D2soi containing energy primarily associated with the desired signal. In certain embodiments involving both

5 denoising and signal source extraction, subcomponents to be removed are identified based upon their time-spectral distribution using, for example, Spatially Selective Filtering (SSF) or Periodic Component Analysis (πCA). Decision processes 103 and 203 in Figures 1 and 2 determine if the signal is processed at block 104/204 or at block 105/205, depending upon whether the dimension of the first domain is greater than 1 or equal to one, respectively.

10 If the dimension of the first domain is greater than 1, then SSF, πCA , Principal Component Analysis (PCA), and Independent Component Analysis (ICA) can be used in combination (*e.g.*, as in blocks 104 and 204). If the dimension of the first domain is equal to 1, then SSF or πCA can be used to evaluate the time-spectral distribution of the subcomponents, as in 105 and 205. Once the subcomponents primarily associated with noise

15 are identified they are effectively removed by setting them to zero. Following removal of the subcomponents associated with noise, the subcomponents are said to be denoised. In the case of signal source extraction, those subcomponents associated with noise and those not associated with a signal source to be extracted are identified and effectively removed by setting them to zero.

20 In process 106, the denoised signal subcomponents, referred to as subset D2s, are reconstructed in the first domain to provide denoised output signal 107 by performing the inverse of the transform used to decompose the signal. If signal source extraction is performed only those subcomponents not removed and identified as associated with the desired signal sources, referred to as subcomponent subset D2soi in process 206, are

25 reconstructed into the extracted desired signal 207 in Figure 2.

Various embodiments directed to MDSP approaches use a mathematical representation of physiological signals as a linear combination of basis functions. Each of these basis functions is chosen to fit a prominent feature, or signal wave, of a signal source. For example, a basis function might represent a type of QRS complex contained in an ECG

30 signal. In one embodiment, decomposition is achieved by representing the observed signals as a linear combination of basis functions,

$$s_i(t) = \sum_{k=1}^K \varphi_k(t) d_{ik}, i = 1:P, \quad x_i(t) = \sum_{k=1}^K \varphi_k(t) c_{ik}, i = 1:M \quad (2)$$

where d_{ik} and c_{ik} are decomposition coefficients or subcomponents. In some embodiments, the elements ϕ_k may be mutually dependent and form an overcomplete set referred to as a “dictionary” D . The elements are selected to provide a sparse representation of the signal sources, meaning that most of the decomposition coefficients d_{ik} and c_{ik} in equation (2) are close to or equal to zero. A combination of basis functions or elements of the dictionary forms the second domain.

Higher sparsity provides for greater statistical independence of signal source and noise subcomponents. This property of sparse decomposition is used to facilitate effective identification and extraction of signal source subcomponents and removal of noise. Sparse decomposition, for example, can result in a concentration of signal energy in a relatively small number of large decomposition coefficients while noise is spread out across many basis functions and represented by small decomposition coefficients as described in equation (2).

The sparsity criteria can be used to choose a dictionary that provides for sparse decomposition of the signal sources and hence improved signal source extraction and denoising. In certain embodiments it may be advantageous to choose an over-complete dictionary D in order to increase the degree of sparsity of the decomposition. The examples of dictionaries that are relevant to this embodiment include eigenfunctions [2], wavelet related transforms, including wavelet packets, shift-invariant or stationary wavelets [3] and a synchrosqueezed wavelet transform [29], cosine transform [4], Fourier basis [5] or Gabor basis and their combinations [6]. A custom dictionary that achieves a higher degree of sparsity can also be designed by a number of methods for finding basis functions that represent building blocks of the signal [7]. The examples of these methods are matching pursuit [5], orthogonal matching pursuit [8], basis pursuit [9], K-SVD [10], bounded error subset selection [11] or orthogonal subspace pursuit. In general, the dictionary design methods use a representative set of training signals, to find a dictionary that leads to the best representation for each signal wave in this set, under strict sparsity constraints:

$$\min_{D,c} \{ \|x - Dc\|_F^2 \} \quad \text{subject to } \|c\|_0 \leq T$$

where $\|\cdot\|_0$ is a l^0 norm, counting the nonzero entries of a vector, T is a constraint on the maximum number of non-zero elements allowed, and $\|\cdot\|_F^2$ is a Frobenius norm. For example, a representative set of training signals can be a set of ECG recordings with a broad range of morphologies that include normal sinus rhythm, conduction abnormalities,

pathologies, arrhythmias for a particular species. The set should include a sample of each morphology that the algorithm may experience during analysis of an ECG recording of the species.

In one embodiment the custom dictionary is built using the following iterative
 5 procedure. A vector is selected from the training set and is represented as a sparse linear combination of initial dictionary elements. This can be accomplished, for example, by finding elements that are maximally correlated with the vector. This subset of elements form a subspace that is decorrelated from the rest of the training set (*e.g.*, by using singular value decomposition (SVD)). The decorrelated subset is added to the dictionary and the elements
 10 of the training set that lie in this subspace are removed from the training set. The process is repeated until an error threshold is reached or an allowed number of iterations is exhausted.

In another embodiment decomposition into the second domain is achieved by applying an adaptive filter that automatically selects the frequency bands, or subbands of wavelet filters or a filter bank, in which subcomponents of the signal sources are most
 15 independent. Then each subcomponent $x_i(t), i = 1:M$, can be represented as a sum of subband signals

$$x_i(t) = x_{i,1}(t) + \dots + x_{i,K}(t) \quad (3)$$

where K is the number of chosen subbands. The filter coefficients can be adaptively tuned to reduce and/or minimize the mutual information and increase independence between the filter
 20 outputs as described by Zhang, *et al.* [12]

In an embodiment involving a wavelet-related transform decomposition, the first decomposition step for an observed signal with M channels into a wavelet-related dictionary D can be formally described as iterative multiplication of the observed signal by an operator matrix W^j with each iteration step j making up a decomposition level. In this embodiment,
 25 the operator matrix W^j is a Toeplitz matrix with the entries formed by a wavelet impulse response. The size of the matrix operator W^j is N x N where N is the number of samples of the observed signal. N is greater than both the number of sources P and the number of observed signals M. Wavelets may be used to generate subcomponents having characteristics of both sparse and subband decomposition. In some implementations, a
 30 wavelet decomposition is computed via iterative convolution with wavelet filters or filter banks as described by Vaidyanathan [13].

Referring again to Figures 1 and 2, another example embodiment may be implemented at step 2, for an MDSP-based approach. The statistical independence of at

least some subcomponents related to signal sources and noise is leveraged to identify those subcomponents associated with signal sources and those associated with noise. For example, in case of wavelet decomposition, the equation (1) becomes

$$W_x^j = A_j W_s^j + W_n^j, j = 1 : NL + 1 \quad (4)$$

- 5 where NL is the number of decomposition levels or subbands. The mixing matrix A is indexed by j, denoting that the sources are mixed differently in each subband according to their spectral distribution.

The subcomponents associated with signal sources can be identified based on time-spectral distribution of subcomponents using one of several techniques, many of which
10 involve finding the mixing matrices A_j and then inverting them to calculate the signal source subcomponents. Examples of techniques that are applicable to identification of subcomponents associated with noise and subcomponents associated with signal sources (*i.e.*, the desired signal) include principal component analysis (PCA), independent component analysis (ICA), periodic component analysis (πCA) techniques and spatially
15 selective filtering as shown in processes 104 and 105 of Figure 1 and processes 204 and 205 of Figure 2. When the dimension of the first domain is > 1 , these techniques can be used separately or in combination to achieve acceptable performance.

In one embodiment, spatially selective filtering (SSF) is employed to identify subcomponents associated with the desired signal and is used in combination with PCA or
20 ICA in processes 104 and 204 if the input signal consists of two or more channels. In one embodiment, if the input signal is a single channel signal (dimension of the first domain = 1), then only SSF is employed in processes 105 and 205 to identify subcomponents associated with noise.

In another embodiment, periodic component analysis (πCA) is employed to identify
25 subcomponents associated with the desired signal and is used in combination with PCA or ICA in processes 104 and 204 if the input signal consists of two or more channels. In one embodiment, if the input signal is a single channel signal (dimension of the first domain = 1), then πCA is employed in processes 105 and 205 to discriminate between subcomponents associated with a desired signal and subcomponents associated with noise.

30 When using the PCA technique to identify subcomponents associated with signal sources, an inverse of the mixing matrix A_j is estimated using singular value decomposition or eigenvalue decomposition of the covariance matrix of signal subcomponents. This PCA approach involves rotating and scaling the data in order to orthogonalize independent

subcomponents. The orthogonalized subcomponents with low signal power are often associated with noise and can be removed to achieve denoising. In some embodiments, PCA is used as a preliminary step prior to applying ICA technique.

In various embodiments involving the use of an ICA technique to identify
 5 subcomponents associated with signal sources, an inverse of the mixing matrix A_j is estimated [1,14] as a solution of an optimization problem that maximizes independence between the signal sources. For example, ICA techniques can use either higher-order statistics of signal components or information-theoretic criteria to maximize independence. Information theoretic criteria that can be applied include maximization of negentropy or its
 10 approximation, minimization of mutual information, maximum likelihood estimation, maximum a posteriori probability, or expectation-maximization of Gaussian mixture models of sources. These solutions can be approximated via efficient numerical methods [1], such as FastICA and JADE algorithms. The Figure 3a provides a scatter plot image of subcomponents in the second domain prior to the application of PCA and ICA in the
 15 denoising process. Figure 3b demonstrates an example result of denoising and subcomponent separation achieved by applying a combination of PCA and ICA techniques. Note that application of PCA and ICA has achieved a decorrelation of the subcomponents and alignment along the coordinate axis 302 of the scatter plot, indicating a large degree of independence has been achieved between the subcomponents (*e.g.*, relative to alignment in
 20 with coordinate axis 301 Figure 3a). This high degree of independence between subcomponents that occurs as a result of the denoising process facilitates efficient removal of noise and signal source extraction.

In embodiments involving the use of πCA technique to identify signal sources, an inverse of the mixing matrix A_j is estimated as a solution of an optimization problem that
 25 separates subcomponents based on their periodicity or quasi-periodicity [15]. Instead of diagonalizing and inverting the covariance matrix as is done with PCA, the πCA technique jointly diagonalizes the covariance matrix and an autocorrelation matrix. The autocorrelation matrix is calculated as an average of autocorrelation matrices computed over time lags corresponding to period lengths.

30 In one embodiment, a quasi-periodic signal can be phase-wrapped by mapping the period length to a linear phase $\varphi(t)$ ranging from $-\pi$ to $+\pi$ assigned to each sample. The autocorrelation matrix can then be calculated in the polar coordinates in which cardiac cycles are phase aligned. The technique involves extracting most quasi-periodic subcomponents corresponding to a desired physiologic signal. The technique is efficient at identifying signal

source and noise subcomponents but relies on accurate detection of cycles such as cardiac or respiratory cycles. It can be used in combination with spatially selective filtering (SSF), a technique that facilitates better cycle detection.

In another embodiment, referring to Figure 2, signal sources are identified and
5 separated in step 2, processes 204 and 205 as discussed above, using spatially selective filtering. Spatially selective filtering [16,17] techniques detect signal-related features and pass them across the subcomponents while blocking features inherent to noise. The technique relies on the differences of noise and signal distributions between subcomponents.

In one embodiment, signal decomposition is carried such that signal energy is
10 concentrated in a small number of higher amplitude subcomponent coefficients while noise is spread out across many decomposition levels and is represented by small coefficients. Techniques similar to wavelet thresholding [18] can be used to remove noise, but their use may result in a degradation of signal morphology.

In another embodiment, spatially selective filtering (SSF) is used with quasi-periodic
15 physiological signals in which the frequency content of the signal varies predictably over the course of a cycle. For example, when the physiological signal is an ECG, SSF can be implemented to identify the QRS complex via the confinement of noise subcomponents to decomposition levels that represent high frequencies, with the QRS complex having high amplitude across subcomponents representing both high and low frequencies. Using such
20 aspects, a fiducial point in a cycle of a quasi-periodic signal is identified by computing a time function indicative of the time-spectral distribution of subcomponents, and the time function is compared to a threshold to identify aspects of the signal pertaining to the QRS complex. In one embodiment the time function is a point-wise product of subcomponents selected to comprise at least a majority of QRS complex energy. In some embodiments, the
25 point-wise product is normalized to match the energy of a subcomponent. In another embodiment the time function is a cross-correlation between the subcomponents selected to comprise at least a majority of QRS complex energy. In another embodiment the time function is a linear combination of the subcomponents selected to comprise at least a majority of QRS complex energy.

30 In some embodiments, it can be useful to apply the transform in a manner that preserves the time relationship between subcomponents in order to facilitate computation of the time function. Examples of such methods are: non-orthogonal waveform, undecimated wavelet transform, shift-invariant or stationary wavelet transform, and undecimated filter bank. To identify a fiducial point in a cycle, the computed time function is compared to a

threshold to identify locations of peaks and valleys that occur simultaneously across multiple subcomponents. For example, when the physiological signal is an ECG, large peaks and valleys of the time function may correspond to a QRS wave. In some embodiments, all of the subcomponents within a window surrounding the QRS complex are preserved and the
5 subcomponents outside of the QRS window that correspond to higher frequencies are zeroed out.

In one embodiment, aspects of MDSP can be useful for identifying a QRS complex of an ECG, such as for deriving heart rate. In this embodiment the input ECG signal is decomposed from a first domain into subcomponents in a second domain and a time function
10 representing a time-spectral distribution of selected subcomponents is computed and compared to a predetermined threshold. In some embodiments, the time function is computed by combining subcomponents selected to comprise at least a majority of the energy of the QRS complex. For the purpose of computing the time function, subcomponents can be combined using a point-wise product, a linear combination, or a
15 cross-correlation of the selected subcomponents.

In accordance with these approaches and related embodiments involving detecting QRS complexes in an ECG, it has been recognized/discovered that the existence of relatively high energy in a wide frequency band of the QRS results in high amplitude in most of the subcomponents, and that this can be used to identify appropriate subcomponents that are less
20 likely to include noise for use in recognizing the QRS complex. The time function computed from selected subcomponents is used to enhance the relationship between subcomponents in a way that emphasizes the QRS complex above noise and narrower bandwidth waves. For example, T-waves will be de-emphasized in the time function because they contain mostly low-frequencies which will provide a safeguard against
25 mistakenly detecting a tall T-wave as a QRS complex. In some embodiments, the discrimination between QRS features and other ECG waves in this time function can be sufficiently high that a pre-computed threshold can be used instead of commonly applied adaptive threshold. This can result in increased computational efficiency of a QRS detector, a particularly important attribute for QRS detection algorithms implemented in implanted
30 devices such as neural stimulators, pacemakers, rhythm monitors, and defibrillators as well as non-implanted battery powered devices such as event recorders and mobile cardiac telemetry.

In some embodiments of the present invention, referring to Figure 25, a noise floor that defines the lowest amplitude QRS complex that can be detected is set based upon noise

energy measured on an ongoing basis, thus adjusting the noise floor to suit a particular application. When the QRS amplitude falls below the noise floor, it is not detected. Using this ongoing noise floor determination approach, instances in which QRS complex detection that could be missed with a too-high fixed noise floor can be addressed, as can instances in which false positive QRS detection would result via episodes of high-amplitude noise in an ECG.

One embodiment uses the measured noise level to provide dynamic (*e.g.*, sample-by-sample) adjustment of a QRS detection threshold to improve the accuracy of QRS detection. Hence, when brief episodes of high-amplitude noise occur, the noise floor will be automatically adjusted to reduce the risk of false QRS detections. In some embodiments, the segments with high-amplitude noise can be detected and marked as segments containing unreliable detections. The dynamic noise floor can facilitate discrimination between QRS complexes and noise for reliable QRS detection.

Referring to Figure 25, an embodiment for detecting the QRS complex of an ECG is described. Input signal at 2501 is decomposed in process 2502 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 it is useful to preserve time synchronization of the subcomponents derived in process 2502. In one embodiment, when a wavelet-related transform is employed, time synchronization of subcomponents is preserved using techniques such as non-orthogonal wavelet, undecimated wavelet transform, or stationary wavelet transform. In alternate embodiments, an approach resulting in non-decimated subcomponents can be used to preserve time synchronization of subcomponents when non-wavelet transforms are employed.

If the dimension of the input signal is > 1 (*e.g.*, a multi-lead ECG), in some embodiments, process 2504 is used to remove some of the noise using at least one of PCA or ICA. Subcomponents that are primarily associated with the energy of the QRS complex are identified in process 2505 using one of spatially selective filtering or π CA as described herein. In one embodiment, at least one-half of the energy of each subcomponent in the identified subset, D2qrs, is attributed to the QRS complex (*e.g.*, the energy is primarily associated with the QRS complex). In one embodiment, the subcomponents of D2qrs are selected based upon *a priori* knowledge of the frequency content of the QRS complexes in the recording. In some embodiments, subcomponents of D2qrs are combined in 2506 to form a time (emphasis) function $F_{qrs}(t)$ that emphasizes the QRS complex of the ECG. In some embodiments, a single subcomponent is selected from which to compute the emphasis

function. In some embodiments the emphasis function is formed by computing a point-wise product of the selected subcomponents. In another embodiment, the time function is formed by computing a linear combination of the selected subcomponents. In yet another embodiment, the emphasis function is created in 2506 by computing a cross-correlation

5 between the selected subcomponents. In some embodiments the likely location of the QRS complex is identified in process 2507 by comparing the emphasis function computed in 2506 to a predetermined threshold. Once a likely location of a QRS complex is identified, the associated cardiac cycle is divided into two windows; a first window, W_{qrs} , surrounding the QRS complex and a second window, W_{nqrs} , which includes the remainder of the cardiac

10 cycle. A subset D_{2n} consisting of subcomponents that are primarily associated with noise energy is created in process 2508. A noise floor time function, $F_{nf}(t)$, is computed on a sample-by-sample basis in process 2509 by combining subcomponents in D_{2n} located within time window W_{nqrs} . In one embodiment, $F_{nf}(t)$ is computed as the absolute value of the linear combination of the subcomponents D_{2n} . In another embodiment, $F_{nf}(t)$ is computed

15 by performing inverse transform of subcomponents D_{2n} and computing absolute value of the result. An adaptive threshold function, $F_{thr}(t)$ is computed in process 2510 using $F_{nf}(t)$ and information on the likely location of the QRS as detected in 2507. In some embodiments, $F_{thr}(t)$ could be computed using $F_{nf}(t)$ as a lower limit and using an adaptive algorithm upon detection of a QRS complex. $F_{qrs}(t)$ is compared to $F_{thr}(t)$ to identify the

20 actual location of QRS complexes in process 2511.

Figure 26, by way of example, shows a single channel ECG input signal 2601 and the result of processing according to steps 2502, 2505, and 2506 to create QRS emphasis function $F_{qrs}(t)$ 2605. Likely QRS locations are detected and time windows W_{qrs} 2603 (designated by line with short dashes) and W_{nqrs} 2606 are identified in process 2507. Noise

25 floor function $F_{nf}(t)$ 2604 (designated by line with long dashes) is computed in 2509 by combining the subcomponents D_{2n} located within time window W_{nqrs} identified in process 2508. In one embodiment, the noise floor function is updated on a sample-by-sample basis. In one embodiment, the noise floor is determined by computing the sum of the squares of the subcomponents D_{2n} .

30 By way of example Figure 27 shows a single channel ECG input signal 2701 and QRS emphasis function $F_{qrs}(t)$ 2702 that results from processing according to steps 2502, 2505, and 2506. Adaptive threshold function $F_{thr}(t)$ 2702 is computed as in process 2510. $F_{thr}(t)$ 2702 adapts to the level of noise present in the signal and is computed using $F_{nf}(t)$ as in 2509 and $F_{qrs}(t)$. $F_{thr}(t)$ 2703 is equal to $F_{nf}(t)$ until $F_{qrs}(t)$ is greater than the

predetermined threshold used in 2507 to detect likely QRS location. At the likely QRS location, $F_{thr}(t)$ follows a QRS pattern until it once again drops to the noise floor. Once $F_{qrs}(t)$ reaches this predetermined threshold, $F_{thr}(t)$ increases as $F_{qrs}(t)$ increases. If $F_{qrs}(t)$ does not increase for a predetermined refractory time, $F_{thr}(t)$ drops to a predetermined value and then decays exponentially to $F_{nf}(t)$. In this embodiment, $F_{thr}(t)$ is always greater than or equal to noise floor $F_{nf}(t)$. QRS detection locations 2704 are identified in process 2511 by comparing $F_{qrs}(t)$ to adaptive threshold function $F_{thr}(t)$ and then examining input signal 2701 using techniques known in the art.

In alternate embodiments, the need for time synchronization of subcomponents can be relaxed. In these embodiments a time function can be computed from a single subcomponent and that time function can be evaluated for the presence of a QRS complex. If the presence of a QRS complex is indicated, then its presence could be confirmed by evaluating time functions computed from other subcomponents.

In one embodiment, processes 2508, 2509, and 2510 are used to create a QRS detection threshold that adapts to the noise level in input signal 2501. An adaptive threshold that is updated on a point-by-point basis is used to improve QRS detection accuracy for noisy ECG recordings. In an alternate embodiment, the likely location of a QRS complex is identified in process 2507, and a template match is then performed on the likely QRS complex to confirm that presence of a QRS complex.

Each subcomponent resulting from the decomposition of a physiological signal such as an ECG is said to have an associated spectral energy. A specific wave (*e.g.*, QRS complex) or region of interest (*e.g.*, the portion of a cardiac cycle extending from QRS offset to QRS onset or from T-wave onset to T-wave offset) in the physiological signal is said to have an expected spectral energy. The expected spectral energy of the wave or region of interest is the spectral energy density function that is characteristic of the wave or region of interest when computed for a relatively noise-free recording. In this context, a relatively noise-free recording refers to a recording in which the energy of noise components is less than about 5% of the total energy of the recording.

Referring to Figure 28, 2801 represents the expected spectral energy (*i.e.*, spectral energy density function) of a wave or region of interest in a physiological signal, while 2803 is an example of expected noise spectral energy (*i.e.*, spectral energy density function of noise present on the physiological signal). Like expected spectral energy of the wave or region of interest, expected noise spectral energy is based upon empirical data on noise

content of the physiological signal and upon consideration of sources that may contribute to noise.

By way of example, envelope 2802 represents the spectral energy (*i.e.*, spectral energy density function) of a subcomponent (referred to as subcomponent A) resulting from decomposition of the physiological signal. Note that envelope 2802 overlaps both noise energy, represented by the darker shaded area at the bottom of the envelope, and energy of the wave or region of interest. The total energy of subcomponent A is gauged by the area under envelope 2802. Roughly 10% of the total energy is attributable to noise, as represented by the darker shaded area 2805 toward the base of the envelope, and the remainder of the area is attributable to energy of the wave or region of interest. In this example, the degree of overlap of the spectral energy of subcomponent A with the expected spectral energy of the wave or region of interest, as represented by envelope 2801, is therefore about 90%.

Envelope 2804 represents the spectral energy of a second subcomponent (referred to as subcomponent B) resulting from decomposition of the physiological signal. Like 2802, envelope 2804 overlaps both noise energy and energy of the wave or region of interest. However, the portion of the total energy attributable to noise in subcomponent B as represented by the lighter shaded area 2806, is more than 50%. In some embodiments, therefore, subcomponent B could be useful when computing a threshold for the purpose of identifying a feature point such as the QRS of an ECG.

Referring to Figure 20, a cardiac cycle of an ECG consisting of a desired signal generated by heart tissue and noise is partitioned into time windows 2002 and 2003: 2002 containing the QRS complex and 2003 spanning the remainder of the cardiac cycle. The subcomponents associated with both low and high frequencies, represented by region 2004, are preserved within the time window 2002 containing the identified QRS complex. The low-frequency subcomponents associated with the desired signal, present in window 2003 and represented by region 2005, are preserved and subcomponents associated with high frequencies are removed. In another embodiment, subcomponents associated with low-frequencies, represented by region 2007, are preserved throughout the cardiac cycle and subcomponents associated with high frequencies are preserved in time window 2002 represented by region 2006.

In one embodiment time window 2002 is a rectangular-shaped window that transitions the preservation of high-frequency components at the window boundaries as a step function, as in Figure 20(b) and 20(c). In another embodiment, the leading and/or

trailing edges of the time window surrounding the QRS complex transitions the degree of preservation of high frequency components gradually over time to reduce the potential for signal distortion near the beginning and end of the window. In one embodiment, window 2008 transitions in time as a linear function and the shape is trapezoidal, as in Figure 20(d).

5 In Figure 20(d), time span 2008 designates the base of the trapezoidal window and generally corresponds to window 2002. In another embodiment, the preservation of high frequency components around the QRS complex transitions as a monotonic region of a cosine or other function that provides a gradual transition from low to high and high to low at the leading and trailing edge, respectively, of the window. Region 2009 generally corresponds to
10 windows 2003 and 2005, the portion of the cardiac cycle outside QRS window 2002, where low frequency content (*e.g.*, subcomponents associated P and T-waves) is preserved and high frequency content is rejected.

Use of SSF is not limited to quasi-periodic signals. For example, in another embodiment, SSF is used to remove EMG artifact from non-quasiperiodic PNA signals. SSF
15 is used to detect high amplitude EMG artifact as large peaks present in multiple subcomponents simultaneously. A time window containing artifact can be identified by comparing a point-wise product of subcomponents or a cross-correlation of subcomponents to a threshold. The EMG noise can then be removed by zeroing most or all subcomponents in the time window.

20 In some embodiments the spatially selective filtering is combined with one or more of PCA, ICA, or πCA . For example, the subcomponents that were identified as noise at the PCA stage can be further filtered to remove only some segments that have spectral and temporal characteristics of noise. In another example, the subcomponents that were identified as signals at the PCA stage can be further filtered to remove additional time
25 segments from subcomponents that have spectral and temporal characteristics of noise.

The choice of a technique for identification of subcomponents for denoising or signal source extraction depends upon signal and noise characteristics of the physiologic signal, time-spectral distribution of its subcomponents and the signal acquisition and processing environment. For example, it may be useful to process multi-channel signals with PCA and
30 ICA techniques and their combinations with spatially selective filtering. When processing ECG signals obtained from an implanted single lead device, spatially selective filtering or πCA may be applied, as in blocks 105 and 205 in Figures 1 and 2, respectively.

In some embodiments, step three as shown in processes 106 and 206 of Figures 1 and 2, respectively, involves reconstructing the subcomponents remaining following denoising

and/or signal source extraction, $s(t) = (s_1(t), \dots, s_p(t))$, in the first domain. This is accomplished by an inverse transform of extracted signal sources or denoised signal sources. If wavelet-based decomposition is employed, the reconstruction step can be formally described as iterative multiplication by an inverse of the operator matrix W^{-1} with each
 5 iteration step j corresponding to a reconstruction level.

Referring to Figure 4, other example embodiments involve a circuit and approach for computing a dynamic signal-to-noise ratio (dSNR) in a manner that can be updated on a sample-by-sample basis. dSNR can be used to assess the accuracy and reliability of information derived from the physiological signal and, in some embodiments, can be
 10 combined with other signal information such as ECG arrhythmic event information to compute a validity metric to assess the validity of information extracted from a signal. If dSNR for a cardiac cycle is low, for example, information derived from the ECG for that cardiac cycle may not be accurate.

A dynamic signal-to-noise ratio (dSNR) can be computed as the ratio of the energies
 15 in signal and noise subcomponents. In one embodiment, referring to Figure 4, an input signal 401 in a first domain is decomposed in process 402 into subcomponents, referred to as set D2sub, in a second domain. At decision point 403, the signal processing flow proceeds according to the dimension of the first domain. If the dimension of the first domain equals 1, then subcomponents primarily associated with noise, referred to as set D2n, are identified in
 20 process 405 using SSF. In another embodiment, the subcomponents primarily associated with noise are identified in process 405 using π CA. Subcomponents of set D2n are combined to compute an estimate of noise energy and residual subcomponents, referred to as set D2s, are combined to compute an estimate of signal energy in process 407. The SNR is then computed in process 408 to create output dSNR value 409 according to formula:

$$25 \quad SNR_{dB} = 10 \log_{10} \left(\frac{P_{signal}}{P_{noise}} \right) = 20 \log_{10} \left(\frac{A_{signal}}{A_{noise}} \right)$$

Where P_{signal} and P_{noise} are respective signal and noise energy and A_{signal} and A_{noise} are respective signal and noise amplitude.

If the dimension of the first domain is larger than one, a PCA or ICA technique can be used in process 404 for initial subcomponent denoising. In some embodiments, PCA is
 30 applied prior to ICA. The noise subcomponents extracted at this initial denoising step can be discarded and the residual noise and desired signal subcomponents can be identified using SSF or π CA in process 406. In process 407, the noise and desired signal subcomponents

D2n and D2s, respectively, are used to calculate an estimate of noise energy (Pn) and energy of the desired signal (Ps). The SNR is then computed in process 408 (e.g., as described above). In alternate embodiments measurement of signal energy Ps and noise energy Pn can be used to compute other metrics indicative of signal quality such as a difference between Ps and Pn and a ratio of Ps/Pn.

In other embodiments, a signal-to-noise ratio is estimated using conventional approaches following denoising of the signal using an MDSP embodiment. In one embodiment, the noise is measured between signal waves (e.g., between T-wave and P-wave of an ECG) by computing the peak amplitude and density of zero crossings or variability of signal in a segment between signal waves. In other embodiments, a signal-to-noise ratio is estimated by computing a spectral distribution of the denoised signal. For example, in an ECG signal, peaks in the spectral distribution are evaluated to determine the relative power in the spectrum that occurs within and outside of the normal range of the QRS complex, T-wave, and P-wave.

In one embodiment dSNR is updated for each cardiac cycle. Alternative implementations may update dSNR more or less often. For example, the dSNR is updated on a sample-by-sample basis in accordance with embodiments described in Figure 4. It may be useful in some embodiments to compute a value of dSNR for a window of two to ten cardiac cycles and use this value in calculation of the validity metric for all cardiac cycles within the window. Figure 5 provides an example of a dynamic confidence signal (dCS), also referred to as a validity metric in some embodiments, derived from dSNR computed for a rabbit ECG signal 501. dCS is shown in the dashed line 505 in the top plot, with the bottom plot showing automatic marking of QRS complexes (squares as 504) and detected VT (circles as 503), in accordance with another example embodiment. The dynamic confidence signal 505 (dashed) is updated for each cardiac cycle. The dCS is updated every cardiac cycle and the value of dCS reflects changes in signal amplitude relative to the level of noise in the recording. Despite the low signal-to-noise ratio, all QRS complexes (as indicated by the squares 504 marking bottom ECG trace) and non-sustained episodes of ventricular tachycardia (marked by circles 503 marking bottom ECG trace) are detected. These signals can be output and used to provide an indication upon which a degree of confidence in automated signal event detection can be based.

In various embodiments, MDSP-based approaches as discussed herein are used to denoise and extract information from a physiologic signal acquired in a low SNR environment. For example, a collar or spring-loaded clip placed on the neck of an animal

with embedded ECG electrodes can be used to collect ECG signals in animals non-invasively. While such an approach may result in a low SNR, denoising approaches such as discussed herein can be used to glean meaningful data from the ECG signals. Another embodiment is directed to placing a collar or spring-loaded clip on the neck of an animal, or
5 a collar placed around the tail or limb, which can house light emitting diode transmitters and light receiving sensors to collect photoplethysmography signals. In one embodiment, the collar or clip includes two pairs of light transmitters and receivers placed at different locations on the neck of the animal in order to achieve redundancy and improved noise suppression. In another embodiment a cage floor having ECG sensors is used to collect ECG
10 signals from the animal feet. These applications are exemplary of many applications characterized by low SNR, for which denoising approaches as discussed can be used to render the applications viable for analysis.

In the following examples, performance aspects are discussed as may be relevant to various embodiments is illustrated and analyzed, many of which involve the analysis of ECG
15 and PNA signals that can be challenging to carry out using other approaches. In many of the examples discussed here, the signal sources are contaminated with noise and the number of sources is larger than the number of observed signals. Although embodiments of the present disclosure can be used with a broad range of physiological signals, exemplary performance is illustrated with ECG and PNA signals that are problematic for traditional signal
20 processing approaches because of contamination with in-band noise that could not be substantially reduced without distorting signal morphology. Such embodiments are applicable to the use of MDSP-based approaches as discussed herein, to achieve certain performance-related conditions or characteristics, which can be measured or relatively characterized in manners as discussed herein. Accordingly, various embodiments are
25 directed to approaches to achieving such things as levels or degrees of noise reduction, signal quality, cardiac (or other signal) event detection accuracy and more, as facilitated using aspects of the disclosure as described herein.

Various MDSP-based embodiments are directed to denoising an ECG recording which includes noise and a desired signal originating within heart tissue. It is useful to
30 perform such denoising without distorting the morphology of the desired ECG signal. Quality of signal reconstruction (QSR) is a metric commonly employed to assess the ability of a noise removal technique to preserve morphology of the desired signal in signals. QSR is defined as the mean squared error between the original signal x_{cl} and denoised signal x_{den} calculated sample-by-sample as a percentage of the original signal variance.

$$QSR = 100\% * (1 - \frac{\sum_i (x_{cl}^i - x_{den}^i)^2}{\sum_i (x_{cl}^i)^2})$$

QSR of a denoised signal would be close to 100% if the original recording is contaminated with minimal noise and if distortion introduced into the desired signal by denoising was very small.

5 Figure 6 shows plots characterizing an example MDF-based embodiment directed to removing noise from input 3-lead ECG signals 601, 602, and 603, as may be implemented (by way of example) using a PhysioNet [19] database. Results of ICA and PCA applied directly to the signal are also shown for comparison. The ECG recordings of Figure 6 represent signals with portions that are relatively noise free (601) and portions that are noisy
10 (602 and 603), demonstrating the removal of noise with an MDF-type approach as discussed herein while also preserving morphology. In such a short recording, the character of the denoising technique, whether MDF, ICA, or PCA, is relatively consistent throughout the duration. Evaluating QSR for the portion which is relatively noise free, therefore, provides an indication of a technique's ability to preserve morphology while a visual inspection of the
15 noisy portion provides for a qualitative assessment of the ability of a particular technique to suppress noise. For the relatively noise free portion, the MDF-based approach shows QSR values of 98%, 99%, and 92% for the top 610, middle 611, and bottom 612 traces, respectively, indicating that distortion was negligible. The PCA approach shows QSR values of 83%, 68%, and 71%, for the top 604, middle 605, and bottom 606 traces, respectively,
20 indicating significant distortion of signal content. The ICA approach shows QSR values of 2%, 1%, and 4%, for the top 607, middle 608, and bottom 609 traces, respectively, indicating very significant distortion of signal content. Accordingly, various MDSP-based embodiments are directed to processing signals to address challenges relating to such QSR values that may be insufficient, if processed otherwise.

25 Figure 7 shows characteristic results of denoising using an MDF-type approach (in accordance with an example/experimental-type embodiment), Butterworth bandpass filtering (BPF) with a pass-band of 1 to 60 Hz, and PCA. A relatively noise-free ECG from the PhysioNet Long-Term ST database (record s30661) is corrupted with increasing levels of band-limited (.05 to 70Hz) white noise, and processed in accordance with an example
30 embodiment. The denoising results are quantified measuring QSR and input and output signal SNR according to the formula:

$$SNR_{dB} = 20 \log_{10} \left(\frac{\sigma_{signal}}{\sigma_{noise}} \right)$$

where σ_{signal} and σ_{noise} are respective clean-signal and noise standard deviations. Note that SNR is computed differently than SNR_A (EC57 standard approach) described elsewhere herein.

5 Referring to the plots 710 and 711 of Figure 7, SNR and QSR versus input signal SNR achieved by MDF, PCA, and BPF are illustrated. Plot 710 illustrates the SNR improvement by comparing input SNR (on x-axis) to denoised signal SNR (on y-axis). Plot 711 illustrates the corresponding QSR for the same range of input SNR. For example, as illustrated in plot 710, an input signal with 4 dB SNR is denoised with an MDF-based
10 approach, a 9 dB SNR improvement (13 dB SNR for denoised output signal vs. 4 dB input SNR) is achieved. Referring to the plot 711, 95% of the original clean signal content is preserved following MDF denoising of an input signal with 4 dB SNR. With increasing input SNR, QSR performance for the MDF-based embodiment quickly approaches 100% with an approximately linear denoising characteristic as measured by SNR on a logarithmic
15 scale. Similar processing with a two-lead ECG generates similar results observed for an MDF-based embodiment. As shown in the Figure 7 the PCA and BPF results can be improved upon using various embodiments of MDSP as discussed herein.

Referring to the ECG tracings 707, 708, and 709 of Figure 7, these three tracings illustrate an input noise-free signal 707; signal 708 corrupted with band-limited white noise
20 to reduce SNR to 4 dB, and signal 709 denoised with an MDF-based approach. Adding band-limited white noise to achieve a 4 dB SNR renders the T and P waves indiscernible, as indicated in the middle tracing, while application of MDF restores the P and T waves.

Figure 8 shows another example embodiment, involving the use of MDF to suppress noise while preserving morphology for a single-channel ECG signal as implemented, for
25 example, with an ECG processing circuit and evaluation system. Plot 801 is an input ECG recording corrupted with noise, and plot 802 shows the result of applying MDF filtering. The noise present in the input signal has similar characteristics (frequency content and amplitude) as QRS complexes which would result in QRS detection errors without denoising. In some applications, the application of MDF is used to avoid potential QRS
30 detection errors, and can remove most of the noise while preserving morphology of QRS, P, and T waves, allowing for high accuracy detection of all ECG features, including QRS complex, P, and T-waves.

The embodiments shown in Figures 6, 7, and 8 characterize an example use of MDF to suppress in-band noise while substantially preserving signal morphology, in accordance with various embodiments. This combination of attributes is useful for a multitude of applications, including clinical diagnosis and research involving the measurement and analysis of ECG recordings and other physiological signals. Waveform morphology is preserved for a variety of applications, such as for detecting abnormalities in cardiac function from an ECG, evaluating respiratory function from a respiratory signal, evaluating a signal from a photoplethysmography sensor for measuring oxygen saturation, evaluating sleep stages from an EEG, and other applications. For example, preserved QRS morphology is used in diagnosing bundle branch block or ventricular hypertrophy, T-wave morphology is used in measuring repolarization abnormalities in clinical care and drug toxicity studies, and ST segment changes are used when diagnosing ischemia, electrolyte imbalance, and Brugada syndrome. In another example, the ability to preserve P-wave and QRS complex morphology facilitates the analysis of time correspondence of P-wave and QRS complex to diagnose AV block.

In another embodiment, an MDSP-based approach is used for detection of the QRS complex of a fetal ECG. In this embodiment, an ECG is recorded by placing sensing leads on the surface of the skin of the mother, typically in the lower abdomen. In one embodiment, individual sources that make up a fetal ECG are separated from the remaining subcomponents in the second domain by applying ICA signal source extraction techniques in step 2 (processes 104 and 204) of Figures 1 and 2, respectively. In another embodiment the fetal ECG is extracted from the other subcomponents by spatially selective filtering, periodic component analysis, or their combination (*e.g.*, in step 2 as shown in Figures 1 and 2 above). In this embodiment, the undesired sources, such as maternal ECG, are treated as noise and removed, leaving the denoised fetal ECG.

In another embodiment, an MDSP-based technique is used for measuring a degree of synchronization of uterine contractions of a pregnant female for predicting and detecting labor. In this embodiment, an electrohysterogram (EHG) is collected from the female abdomen. In one embodiment, an MDF-based approach as discussed herein is used to remove ECG artifacts from the EHG signal. In one embodiment, the degree of synchronization of contractions in denoised EHG signal is measured by amplitude correlation via linear or nonlinear regression and the frequency relationship measured as coherence, or the amplitude and phase synchronization in the time-frequency domain. In another embodiment, a cross wavelet coherence function is used to measure amplitude

correlation between two contraction bursts. In another embodiment, an envelope of a multi-channel EHG signal is calculated using a Hilbert transform or low-pass filtering of a rectified EHG signal to measure amplitude of a contraction wave and its time-spectral synchronization.

5 In other embodiments, a MDSP approach as discussed herein is used for detecting atrial arrhythmias in an ECG. In these embodiments, atrial activity can be extracted by applying ICA, SSF, PCA, π CA, or their combination, as part of step 2 (processes 204 and 205) as shown in Figure 2. When used for detecting atrial arrhythmias, step 2 involves removing subcomponents associated with ventricular activity and noise.

10 In one embodiment, when processing a single channel ECG to evaluate atrial activity, process 205 employs SSF to remove ventricular activity and noise. In another embodiment, when processing a single channel ECG, process 205 employs π CA to remove ventricular activity and noise. In some embodiment SSF or π CA can be used to identify the boundaries of a window containing the QRS complex and T-wave of a cardiac cycle so that
15 the contents of the window containing the QRS complex and T-wave can be removed. In some embodiments, identification of the window further involves the use of emphasis signals in a manner similar to that described in Figure 12 and associated text.

In one embodiment, when evaluating atrial activity in an ECG signal containing more than a single channel either SSF or π CA can be used in combination with PCA or ICA to
20 identify ventricular activity and noise and to identify a window containing the QRS complex and T-wave of a cardiac cycle. As for single channel ECG, some embodiments of multichannel ECG may employ the use of emphasis signals in a manner similar to that described in Figure 12 and associated text to identify the boundaries of a window containing the QRS complex and T-wave. Once the boundaries of the window containing the QRS
25 complex and T-wave are identified, they can be removed to facilitate evaluation of atrial activity.

In Figure 9, the results of atrial activity extraction are illustrated on an atrial flutter recording from the PhysioNet database. Plots 901, 902, and 903 show ECG recorded from surface leads. Plot 904 is a recording from an intracardiac catheter located near an atrial free
30 wall. The intracardiac recording is shown as an exemplary benchmark of atrial activity. Plot 905 shows the atrial activity separated from the surface ECG recordings using an MDSP-based embodiment. Note that the locations of P-waves on the plot 905 of extracted atrial activity matches locations of atrial depolarizations recorded from intracardiac lead shown in 904. The gaps in atrial activity coincide with ventricular depolarizations (*i.e.*, QRS

complexes) and repolarizations (T-waves). Analysis of the separated atrial activity can significantly improve the accuracy of atrial flutter detection.

In Figure 10, the results of atrial activity extraction are illustrated on an atrial fibrillation recording from the PhysioNet database. During atrial fibrillation the P-waves
5 often degenerate into more fractionated and variable f-waves which are barely visible in surface recording traces 1001, 1002, and 1003 shown in Figure 10. Despite that, the atrial activity signal 1005 extracted using an MDSP embodiment clearly shows the f-waves that coincide with the atrial depolarization trace 1004 recorded using an intracardiac lead. The only gaps are when atrial activity coincides with ventricular depolarizations (*i.e.*, QRS
10 complexes) and repolarizations (T-waves).

Figures 9 and 10 demonstrate that various MDSP-based embodiments are capable of extracting atrial activity from ECG recordings. There are a number of ways that extracted atrial activity can be used to improve the accuracy of atrial flutter and fibrillation detection and discrimination. For example, spectral analysis of the extracted atrial activity signal can
15 be used to quantify P-wave regularity and frequency and discriminate between atrial fibrillation and flutter. In one embodiment, atrial rate can be estimated by spectral analysis. In another embodiment, extracted subcomponents corresponding to atrial activity can be analyzed in the second domain to estimate the atrial rate. In another embodiment, a P-wave similarity measure is used to detect and discriminate between atrial fibrillation and flutter. In
20 another embodiment, zero crossings can be used to estimate atrial rate in the segments where atrial activity is present. The analysis of atrial activity can be combined with analysis of cardiac cycle variability and regularity as well as PR interval variability to further improve the accuracy of atrial fibrillation and flutter detection. In one embodiment, the validity of a detected atrial arrhythmia event is determined by comparing the dynamic confidence signal
25 (dCS) to a threshold. In one embodiment, dCS is computed using dSNR. In another embodiment, dCS is computed using a combination of dSNR and one or more of a rhythm characteristic such as presence of bigeminy, presence of trigeminy, and presence of PVCs. False positive detections of an atrial arrhythmia event during noisy ECG segments can thereby be reduced.

30 In other embodiments an MDSP-based approach is used for separating ventricular depolarization and repolarization activity. This may be useful for assessing repolarization activity for the risk stratification of sudden cardiac death. Once repolarization activity is extracted, it can be used to assess T-wave alternans and morphology, ST elevation in

ischemia, and other cardiac abnormalities that can be useful for assessing cardiovascular risk.

Another embodiment is directed to using MDF for removing high amplitude noise from PNA signals. In particular, this is useful when removing high amplitude noise from PNA recordings such as those from the vagal nerve, sympathetic nerves, or motor nerves. In this embodiment, the high amplitude noise or artifact is detected by identifying undesired peaks in subcomponents that do not overlap the band of frequencies associated with desired signal. In Figure 11 exemplary performance of an MDF embodiment is illustrated on a single channel PNA input signal 1101 with high amplitude EMG noise. Plot 1101 is a PNA recording from a rat renal nerve while plot 1102 is the denoised signal following application of MDF. The MDF embodiment extracts the signal source from a very low SNR acquired signal while preserving PNA information. It removes nearly all noise and artifacts from the recording including myoelectric artifacts (EMG), electrical, and other noises while preserving information regarding the underlying PNA. An MDF embodiment could be used to extend the life of preparations where PNA is recorded by improving the ability to extract accurate information from PNA signals with low SNR. In PNA recording preparations for research in animal models, it is also common to dose the subject with a drug that shuts down PNA so that baseline noise can be measured and subsequently subtracted from the neural signal. This procedure can be dangerous for the subject and is labor intensive and cumbersome for the researcher. The MDF embodiment removes background noise automatically and eliminates the need for such an intervention.

In connection with another example, embodiment, PNA is quantified from recordings such as discussed above, via the computation of integrated sympathetic nerve activity. In one embodiment for computing a PNA envelope, an orthogonal component of the denoised PNA signal is computed using a transform such as a Hilbert transform, or similar transform. An envelope is computed as the square root of the sum of the squared denoised PNA signal and its orthogonal denoised component. This provides an accurate representation of neural activity without the phase delays inherent in conventional approaches. The neural activity represented by the signal envelope has a much lower frequency content compared to the raw PNA signal and thus substantially reduces the bandwidth and sampling rate requirement of a system for measuring PNA activity.

Monitoring devices that transmit raw PNA signals as may be used in accordance with this or other embodiments, such as that available from Telemetry Research, Auckland, NZ, employs a sampling rate of about 8,000 Hz. By employing this embodiment to calculate an

accurate PNA envelope, the sampling rate can be reduced to 100Hz or less, resulting in a reduction of transmitted bandwidth of a factor of 80 and a reduction in current drain of a wireless transmitter used to transmit data from an ambulatory subject.

In another embodiment, an MDSP-based approach is used for removing noise and extracting signal sources from signals acquired as a result of programmed periodic stimulation, such as auditory brainstem response and in peripheral nerve stimulation therapies where evoked response is analyzed to titrate therapy or peripheral nerve recruitment. These signals are often characterized by low SNR and a limited number of observed channels. These signals are also characterized by a consistent response following a consistent stimulation regimen. These signals can be segmented in the time domain based upon knowledge of timing of stimulation. Segmentation in the time domain allows for the creation of the equivalent of multiple channels from the observed signal, hence increasing the number of dimensions in the first domain. Following decomposition into the second domain, one or more MDSP-based embodiments previously described can be applied for denoising and signal source extraction. In one embodiment, segmentation of the signal according to timing of stimulation facilitates the use of principal component analysis (PCA) following decomposition to remove noise. Further application of SSF following PCA can further suppress noise. For auditory brainstem response, an MDSP-based embodiment can be used to improve the accuracy of intracranial pressure estimation, such as with a system similar to that described in US Patent Publication No. 2008/0200832, and U.S. Patent No. 6,589,189.

In the case of a peripheral nerve stimulation therapy, evoked responses can be analyzed to titrate therapy targeted at specific nerve fibers or monitor neuropathy progression. For example, in vagal nerve stimulation applied for cardiovascular therapy, the stimulation protocols are optimized to selectively recruit efferent smaller fibers that control heart function and block stimulation of efferent larger fibers and afferent fibers that could invoke pain or coughing reflex. Such approaches may be implemented in accordance with that described in U.S. Patent Publication No. 2008/0065158. A device employed for neural stimulation may incorporate a feedback control of stimulation by observing parameters of an evoked response. In this type of neural stimulation device, an MDSP-based embodiment could be applied to assess evoked response resulting from nerve stimulation targeted at a specific nerve fiber type. In some implementations, an MDF-based embodiment is used to achieve an accurate assessment of the response of particular nerve fibers to changes in programmed stimulation parameters

such as timing, frequency, pulse width, pulse repetition, duty cycle and amplitude of stimulation in order to appropriately affect therapy.

In another embodiment, an MDSP-based embodiment is used to monitor neuropathy progression by measuring changes in nerve conduction velocity of small diameter axons.

5 The changes in these axons can serve as an earlier marker of neuropathy development. An MDSP-based embodiment that involves segmenting the evoked response signal and denoising as described above is used to remove background noise and for measurement of evoked response amplitude and time. In many embodiments, this approach is used to facilitate the detection of changes in axons that can otherwise be challenging to detect due to
10 low amplitude and phase cancellation of evoked response potentials.

In another embodiment, MDSP is used for detecting feature points of a physiological signal containing signal waves. For example, the physiological signal can be an ECG signal and the signal waves are the P, Q, R, S and T waves of the ECG. In this example, it may be useful to detect feature points including Q-wave onset, P-wave onset, and T-wave offset.

15 Referring to Figure 12, a feature point detection embodiment involves a processing circuit and related modules that use one or more selected denoised signal subcomponents to compute an emphasis signal that emphasizes a signal wave or feature point of interest, computed as a linear or non-linear combination of selected subcomponents that are primarily associated with the signal wave of interest. In another embodiment, the emphasis signal is
20 computed by performing the inverse of the transform used for decomposing the physiological signal on the selected subcomponents. In some embodiments it may be useful to differentiate the signal following the inverse transform.

Referring to Figure 12, subcomponents are generated by decomposing input signal 1200 in process 1205 using an MDSP embodiment, as described herein. The subcomponents
25 that are primarily associated with the signal waves are selected for the emphasis signal and are denoised in process 1206 using an MDSP embodiment. The emphasis signal computed in process 1207 may include frequency subbands (*i.e.*, spectral content of the selected subcomponents) matching the spectral energy of particular signal waves of interest or a subset of basis functions tuned to the signal wave or feature point of interest and its
30 variations across a range of normal, perturbed, and pathological conditions. In another embodiment wavelet related subcomponents can be used to compute the emphasis signal in process 1207. The specific subcomponents used depend upon the decomposition technique used, sampling rate, and the species from which the physiologic signal was recorded. Example emphasis signals of an ECG are shown in 1202, 1204, 1211, and 1213 of Figure

12, along with corresponding ECG waveforms 1201, 1203, 1210, and 1212. The vertical dashed lines in Figure 12 show the point of feature point detection in each ECG waveform and corresponding emphasis signal.

The transition points of the emphasis signal are evaluated in process 1208 to detect
5 feature points, shown by example in 1214 and 1215, of the physiologic signal to create output 1209 representing the time of the feature point. In one embodiment, the pattern of significant peaks, valleys, and zero crossings within the emphasis signal are used to detect feature points. In another embodiment, the feature points are detected by applying a threshold to the emphasis signal. In yet another embodiment, the feature points are detected
10 by applying pattern or template matching to the emphasis signal.

The following illustrative example demonstrates exemplary performance of an embodiment involving the detection of QRS complexes in an ECG. For QRS detection, an MDSP-based approach is used to compute an emphasis signal from a combination of denoised subcomponents that are algorithmically selected to include, for example,
15 complexes that are wide, narrow, premature, fractionated, biphasic, monophasic, fibrillatory, tachycardic, or complexes where morphology has been modified as a result of a pharmacological agent. Figure 13 provides an illustration of an ECG signal 1301 that is processed in accordance with such an approach, to address problems that may relate to the presence of bigeminy, tall T-waves, and low QRS amplitude, in addition to rapid changes in
20 QRS amplitude. An emphasis signal computed as described in this MDSP embodiment facilitates the detection of all QRS complexes, as shown by the circles as in 1302 of Figure 13, with no false detection on T-waves. When tested on the MIT BIH Arrhythmia database, a QRS detection accuracy of 99.8% sensitivity and 99.8% positive predictive value can be achieved on single lead ECGs. Feature detection can be enhanced by using and analyzing
25 multiple lead recordings, using an MDSP-based approach to leverage redundancies between the leads in a manner that is more efficient at removing noise.

In another example embodiment, an MDSP approach is used for extracting a respiration signal from an ECG, blood flow, photoplethysmography, thoracic impedance, or an arterial blood pressure signal. Figure 14 shows one such implementation, in which a
30 respiration signal is extracted using one or more selected denoised signal subcomponents to compute an emphasis signal that is associated with a respiratory pattern. In some embodiments, the subcomponents are filtered with a low-pass filter to extract the low-frequency respiration signal.

In other embodiments, the emphasis signal is combined with a heart rate signal to improve the accuracy of computed respiratory parameters. For example, canines have pronounced respiratory sinus arrhythmia which is characterized by heart rate changes that correlate to respiration; these characteristics can be used in connection with these
 5 embodiments, for extracting, processing or otherwise using canine respiration signals.

Referring again to Figure 14, subcomponents resulting from decomposition in process 1402 of input signal 1401 are used to compute an emphasis signal, using an MDSP approach such as described in connection with one or more example embodiments herein. Subcomponents are denoised in process 1403 using an MDSP embodiment as described
 10 herein. Subcomponents that are primarily associated with the desired signal (*e.g.*, respiration signal) are selected and combined to create a respiratory emphasis signal in process 1404, also using an MDSP embodiment as discussed herein. In some embodiments, the selected subcomponents contain a majority of the energy of the respiration signal. The emphasis signal may include frequency subbands matching the spectral energy of a respiratory signal
 15 or a subset of basis functions tuned to the respiratory signal and its variations across a range of normal, perturbed, and pathological conditions. The emphasis signal is processed in 1405 using zero crossings or spectral analysis to compute respiratory parameters.

In another embodiment (*e.g.*, relative to Figure 14), wavelet related subcomponents are used to compute an emphasis signal. The specific subcomponents that are used are
 20 selected relative to one or more of the decomposition techniques used, sampling rate, and the species from which the physiologic signal was recorded.

In some embodiments, the noise level of an ECG is measured using an embodiment described herein, and zero crossings that occur too frequently during noisy segments are discarded (or simply not used). In another embodiment, the emphasis signal can be low-pass
 25 filtered prior to measurement of respiration rate and tidal volume. The tidal volume can be extracted by measuring peak and valley amplitude between valid zero crossings. In another embodiment, the tidal volume is computed as a function of area under the curve between valid zero crossings.

Figure 15 illustrates another example embodiment as directed to the processing of a
 30 primate ECG. Plot 1501 is a noisy input ECG, plot 1502 is an ECG denoised with a MDF-based approach as described herein, and plot 1503 is an extracted respiration emphasis signal. The residual noise level is estimated and is shown on the bottom plot as a dotted line. Estimating residual noise power using an MDSP embodiment (*e.g.*, such as described in Figure 4 and related text) can be useful for assessing the validity of the respiration signal.

Zero crossings can be discarded or otherwise not used if the noise exceeds a predetermined threshold.

An additional MDSP embodiment is directed to detecting events in a physiological signal by combining aspects of feature point detection and signal source extraction as discussed herein, to detect cardiac abnormalities and arrhythmias of ventricular and atrial origin, such as ventricular fibrillation, tachycardia, bradycardia, atrial fibrillation and flutter, AV block and others. In one embodiment, the intervals between QRS complexes are computed to detect rate abnormalities indicative of tachycardia or bradycardia. In another embodiment, a transition to tachycardia and QRS complex morphology are evaluated to discriminate between sinus-, supra-ventricular or life-threatening ventricular tachycardia. In another embodiment, the ventricular rhythm statistics and separated atrial activity rate are evaluated to detect atrial fibrillation and flutter.

Another embodiment involving an MDSP-based approach is used to achieve efficient compression of quasi-periodic signals such as ECG signals, such as by suppressing noise while preserving signal morphology and providing accurate QRS detection. In one embodiment, arrhythmic events are detected and ECG traces corresponding to these events are compressed to reduce the volume of data required for storage and transmission to communicate the signal to a location remote from the monitored subject. Referring to Figure 16, an ECG signal sensed by electrodes 1601 is conditioned by signal conditioning circuits 1602. The digitized signal 1609 is decomposed into subcomponents in process 1610 and the resulting subcomponents are denoised (*i.e.*, subcomponents primarily associated with noise are removed) in process 1611 using MDSP techniques described herein. An emphasis signal is computed and QRS complexes are detected in process 1612 using an MDSP embodiment described herein. Cardiac events are detected in process 1613 using predefined thresholds for heart rate and morphology. The denoised ECG traces containing the detected arrhythmic events are reconstructed in process 1614. The ECG traces are segmented into cardiac cycles and are aligned using a feature point of the QRS complex in time to form an image in process 1615.

The two-dimensional (2D) image plot thus formed includes consecutive cardiac cycles in one dimension and the ECG signal of each cardiac cycle in the other dimension. An illustration of the 2D image is shown as a 3D plot by way of example in Figure 17. Plot 1701 shows a 3D plot of sequential cardiac cycles with cardiac cycle length equalization (*e.g.*, sequential RR intervals are padded by a constant value at the end of the cycle to ensure that all cardiac cycle lengths are equal). The redundancy between adjacent beats results in more

efficient compression. However, due to physiologic factors such as respiration or rhythm disturbances the adjacent beat redundancy might be low. In the 3D ECG plot 1702, cardiac cycles are sorted by length (*e.g.*, RR interval), resulting in smoother beat transitions that lead to more efficient compression. Accordingly, the image can be efficiently compressed by leveraging redundancies between adjacent cardiac cycles. In addition, redundancies across adjacent subbands or wavelet scales can be utilized by wavelet or cosine transforms of the image. Examples of techniques used in process 1616 that could be utilized to achieve efficient compression of the 2D image include transform, subband or wavelet based encoding techniques such as embedded zerotree wavelet (EZW) [20], set partitioning in hierarchical trees (SPIHT) [21], modified SPIHT [22] and embedded block coding with optimal truncation (EBCOT) encoding algorithms [23]. Compression ratios on the order of 15:1 to 20:1 with less than 5% distortion can be achieved using this technique when performed in conjunction with MDSP denoising of ECG signals.

In another embodiment, a quasi-periodic signal is compressed by phase wrapping cardiac cycles and converting the source signals into a polar or cylindrical system of coordinates (38). Then the signal can be efficiently represented by a 3 dimensional plot of phase-aligned cardiac cycles and compressed.

In another embodiment a denoised ECG is compressed by computing a template QRS complex and subtracting it from detected normal QRS complexes. The residual signal has lower frequency content and can be compressed by a lossy compression technique that can include subsampling and quantization. The decompression technique makes use of lossless coding of the QRS template and QRS complex locations as well as lossy coding of the residual signal to reconstruct the ECG signal with small amount of distortion.

Other MDSP-based embodiments are used to compress blood pressure, pulse oximetry signals, respiration, heart sounds, and other quasi-periodic signals. In order to achieve high levels of compression with any of these embodiments without introducing significant signal distortion, accurate QRS or cardiac cycle detection and effective noise suppression (denoising) are used. In some implementations, creating the intermediate representation of the signal that leverages redundancy between the cycles by sorting cycles by length, as in plot 1702 of Figure 17, is used to achieve a high compression rate. Accurate cycle detection is useful for mitigating the introduction of noise and artifacts into the reconstructed signal following compression and decompression.

Another embodiment is directed to an MDSP-based approach used to compress PNA signals and other non-quasi periodic signals, where the denoising provided by MDSP leads

to a sparse signal as demonstrated in plot 1102 of Figure 11. Note that the denoised signal is almost always near zero in the absence of a neural spike. The sparse signal is compressed using one or more compression schemes, such as direct time-domain coding or transform based coding. In general, a compression scheme for a particular physiological signal is
5 selected based upon the ability of a given scheme to leverage redundancies in the signal.

Referring to Figure 18, another example embodiment is directed to using an MDSP-based approach to evaluate ECG strips captured by an ambulatory monitoring device 1801 with arrhythmic event detection capability. The captured ECG strips are forwarded to a data review system 1804 and evaluated using an MDSP-based algorithm 1800 implemented on a
10 computing device in data review system 1804. Algorithm 1800 is used to evaluate the captured ECG strips 1805 and assign a classification to each strip. Each ECG strip is assigned one of four classifications using an MDSP-based approach for arrhythmia detection and validity, the classifications including: A) arrhythmia is present, B) no arrhythmia present, C) strip is uninterpretable, or D) uncertain (*e.g.*, the ECG strip cannot be placed with
15 confidence in classification A, B, or C). If an ECG is classified as D, an automatic indicator can be triggered to suggest human review to determine if an arrhythmia is present. For ECG strips falling in classifications B and C, no further review is needed. Depending upon the nature of the arrhythmia and the clinical care process, ECG strips falling in classification A may be reviewed by a person to assign a suggested diagnosis prior to forwarding to a
20 decision maker. Using this approach and considering a relatively low percentage of the ECG strips evaluated requiring review, labor and costs associated with providing review services can be substantially reduced. In addition, the quality of review services can be improved, since an accurate computer-based algorithm can provide better consistency due to elimination of subjectivity.

25 The system shown in Figure 18 includes ambulatory monitoring devices 1801, or subject devices, that are worn by patients being evaluated for a heart rhythm disorder or for research. Ambulatory monitoring device 1801 includes a computing circuit configured with an algorithm to evaluate the ECG signal from the patient and, if an arrhythmia is detected, capture an ECG strip containing the arrhythmia in memory. Such captured ECG strips may,
30 for example, be one to five minutes in duration and stored in memory for later wireless communication to a base station 1802 (*e.g.*, located in the patient's home), and from the base station 1802 to a data review system 1804 via telecom or data network 1803. The data review system 1804 may be located at a center where the received ECG strips can be reviewed, if necessary, to verify the presence of an arrhythmia or to suggest a diagnosis. The

information derived from the ECGs is then packaged into a report which is forwarded to a researcher or clinician for use in decision making.

The results contained in a report may be provided to physicians, clinics, hospitals, or to organizations engaged in drug safety research, and can be delivered via a service provider system/review center that processes the resulting signals using data review system 1804, possibly in combination with trained healthcare personnel. Such a review center may provide services to a large number of clinics and physicians, or it may be housed within a clinic or a research facility and provide service to one or a small number of clinics or research groups or businesses.

10 In some embodiments, subject devices forward full-disclosure ECG recordings, and in yet other embodiments ECG strips of 10 second to 5 minutes are captured at regular intervals for analysis at the review center without regard to the content or nature of the ECG signal. In some embodiments, the subject device from which results are communicated as above is implanted in a patient. One type of implantable device is the Reveal XT from
15 Medtronic of Minneapolis, Minnesota.

Flow chart 1800 in Figure 18 shows an example embodiment directed to processing and evaluation of ECG strips received by data review system 1804. A captured ECG strip 1805 received at data review system 1804 is evaluated for the presence or absence of arrhythmias using an MDSP-based embodiment as described herein. Criteria 1806 input
20 from a care provider or other decision maker is used by process 1807 to determine if an arrhythmia is present. Examples of criteria 1806 include a heart rate above which a rhythm is considered to be a tachycardia, a heart rate below which a rhythm is considered to be a bradycardia, and a minimum duration atrial fibrillation (AF) episode required for an occurrence of AF to be reported as an arrhythmia. A dynamic signal-to-noise ratio (dSNR) is
25 computed for the ECG strip in process 1808 as described herein. A validity metric (VM) may be additionally computed using the dSNR and signal morphology in process 1808. Signal morphology, for example, can be implemented by comparing QRS width to a threshold. If the threshold is exceeded, the QRS is considered to be a wide QRS and is likely to be associated with an arrhythmia. Additional approaches to computing VM may include
30 evaluating the presence of P-waves and their one-to-one correspondence with ventricular activity. VM is compared to a Validity Threshold VT1 in decision point 1809. If VM exceeds VT1, the result is considered valid and captured ECG strip 1805 is classified "A" if process 1807 detected an arrhythmia and "B" if process 1807 did not detect an arrhythmia. If VM does not exceed VT1 at decision point 1809, then VM is compared to a threshold VT2

at decision point 1810 ($VT2 < VT1$). If VM exceeds VT2, then the result is determined to be uncertain and the ECG strip is classified as "D" to indicate that review by a person should be carried out to determine whether an arrhythmia is present in the ECG strip. If VM does not exceed VT2, then the ECG strip is classified as uninterpretable ("C") and is considered
5 uninterpretable because the noise level is too high to be evaluated by either human or by automated algorithm.

In some implementations, classifications A, B, and C are assigned with a degree of certainty, designated by the validity metrics VM1 and VM2, sufficiently high that any error in classification can be tolerated by the user (*e.g.*, 90% likelihood). In some embodiments,
10 the threshold of certainty of the classification, VM1 and VM2, is determined by the user. Captured ECGs assigned classification D (uncertain) generally include segments for which a computing circuit configured with executable software to carry out an MDSP-based processing approach is unable to make a determination of the rhythm as being classified as A, B, or C with a sufficiently high degree of certainty. Segments with classification D may,
15 for example, contain a level of noise such that the denoised signal is not interpretable by the algorithm, but may be interpretable by a technician, or it may contain an unusual morphology that the software was unable to recognize.

The embodiments described herein for analyzing physiologic signals may be implemented in a variety of platforms that include a logic circuit or computer, with reference
20 made herein to a logic circuit or computing circuit being applicable to a variety of such circuits operating in accordance with one or more embodiments discussed herein. In one embodiment, a microprocessor (such as Pentium or Core microprocessors available from Intel of Santa Clara, CA) in a personal computer running an operating system such as Windows (available from Microsoft of Seattle, WA) or the Unix standard (set by The Open
25 Group of San Francisco, CA) is used to execute programming to carry out MDSP-based functions as discussed herein. In another embodiment, a microcontroller suitable for low-power applications, such as the MSP 430 available from Texas Instruments of Dallas, TX, is implemented to carry out MDSP-based functions as discussed herein. When a review center as described herein is involved, the choice of microprocessor employed for implementation
30 of algorithm 1800 may be based upon simplicity of implementation with a relatively low degree of concern about power consumption. Many implementations involving an MDSP-based approach in an ambulatory monitoring device employ a microcontroller (such as the MSP 430 above, or alternately an ARM Cortex M series core such as that available in the ST Micro STM32) to reduce/minimize power consumption. In other embodiments where power

consumption and size are of high priority, implementation in a silicon-based state machine using a hardware description language such as VHDL may be useful.

In some embodiments, referring to Figures 16 and 19, aspects of the present disclosure are implemented using a battery powered or passively-powered device (*e.g.*, via
5 radio frequency power) that is worn by or implanted within a human or animal subject. Depending upon the application and specific design requirements, referring to Figure 18, various aspects of MDSP-based embodiments discussed herein may be partitioned between implementation within subject device 1801 and implementation within the data review system 1804.

10 Referring to Figure 19, an apparatus for improving the signal-to-noise ratio of a physiological signal is shown, in accordance with another example embodiment. While referencing an ECG signal, the apparatus shown in Figure 19 may be implemented with other signals such as blood pressure, respiration, photoplethysmography, blood glucose, blood flow, heart sounds, PNA, EMG, and EEG. In this example embodiment, ECG is
15 sensed using either surface or implanted electrodes 1901. Signal conditioning circuits 1902 receives the signal from sensing electrodes 1901 and is conditioned to amplify and filter the signal to remove much of the noise outside the bandwidth of the ECG signal. Analog-to-digital conversion (ADC) is accomplished by an ADC on board a Texas Instruments MSP430 microcontroller 1903 (shown by way of example, and implementable with other
20 processors).

Referring to the right side of Figure 19, digitized signal 1906 is processed by computer instructions executed by a 16-bit RISC MCU of the MSP430 1903. The digitized signal 1906 is decomposed into subcomponents in a second domain in process 1907. The dimension of the first domain is equal to the number of sensed ECG signals. Decomposition
25 in process 1907 is performed using one of a discrete cosine transform, a wavelet related transform, a Karhunen-Loeve transform, a short-time Fourier transform, a Gabor transform, and a filter bank. A subset of subcomponents containing primarily desired signal energy is identified using MDSP techniques described herein (*e.g.*, one or more of SSF, PCA, ICA, and πCA) in process 1908. In process 1909, a denoised signal is reconstructed from the
30 subcomponents identified as primarily associated with desired signal energy by performing the inverse of the transform used to decompose the signal. It should be noted that the concepts described herein can also be applied to other MDSP-based embodiments beyond denoising, including feature point detection, event detection, and computing a dynamic

signal-to-noise ratio to evaluate the accuracy of information extracted from a physiological signal.

Referring back to Figure 16, another example embodiment involves using an MDSP-based approach for implementing a battery-powered apparatus capable of wirelessly communicating physiological signals. Referring to Figure 18, an apparatus described in Figure 16 is one embodiment of subject device 1801. The example refers specifically to ECG signals, but a similar embodiment can be used for other quasi-periodic physiological signals such as arterial blood pressure, respiration, blood oxygen saturation derived from PPG, heart sounds, and blood flow for example. In this example embodiment, ECG is sensed using either surface or implanted electrodes 1601. Signal conditioning circuits 1602 receive the signal from electrodes 1601 and amplify and filter the signal to remove much of the noise outside the bandwidth of the ECG signal. Analog-to-digital conversion (ADC) of the conditioned signal is accomplished by an ADC on board a Texas Instruments MSP430 microcontroller 1603.

Referring to the right side of Figure 16, digitized signal 1609 is processed by computer instructions executed by the 16-bit RISC MCU of MSP430 1603. The MCU is in communication with offboard memory 1605 which may be used to provide additional data storage and for storage of computer instructions. The MCU is additionally in communication with wireless transmitter 1604 that can send compressed data to wireless receiver 1606 located remote from subject device 1801 of Figure 18. Wireless receiver 1606 is further in communication with a logic circuit or computer that is configured to execute instructions to decompress the compressed signal in process 1607. In some embodiments the wireless transmitter 1604 and receiver 1606 may each be wireless transceivers capable of both sending and receiving data. An example of a wireless transceiver that may be used in connection with this embodiment is the CC2540 low-energy Bluetooth chip available from Texas Instruments.

Referring to the data flow diagram on the right side of Figure 16, digitized signal 1609 is decomposed in process 1610 into subcomponents in a second domain. The dimension of the first domain is equal to the number of sensed ECG signals. Decomposition in process 1610 is performed using one of a discrete cosine transform, a wavelet related transform, a Karhunen-Loeve transform, a Fourier transform, a Gabor transform, and a filter bank. The subcomponents are denoised in process 1611 using MDSP techniques (e.g., one or more of SSF, PCA, ICA, and π CA) and a QRS emphasis signal is computed in process 1612 as a linear combination of a subset of denoised subcomponents containing QRS signal wave

energy. The emphasis signal is evaluated to detect each QRS complex, as described herein, and a feature signal containing a series of feature points indicating the R-R interval of consecutive cardiac cycles is constructed in process 1613. The feature signal and morphology of the denoised ECG are evaluated in process 1613, as described herein, to
5 detect arrhythmic events, such as bradycardia and tachycardia, and the denoised ECG strips containing arrhythmic events reconstructed in process 1614. The denoised ECG strips to be transmitted are compressed using processes 1615 and 1616. In process 1615, ECG strips are segmented by cardiac cycle and adjacent strips are aligned using a feature point of the QRS complex to form an image of a 3D plot similar to that shown in plot 1701 of Figure 17. The
10 image is subsequently encoded in process 1616 by, for example, applying transform encoding, subband encoding, or wavelet based encoding, after which the encoded image is saved in memory for later wireless transmission or it may be sent immediately.

In an alternate embodiment, instead of forming the image by aligning adjacent strips, the strips are arranged by cardiac cycle length, as in plot 1702 of Figure 17. This results in
15 additional redundancies in strips adjacent to each other in the image and hence results in more efficient compression.

In another embodiment a template QRS complex is computed and subtracted from detected normal QRS complexes. The residual signal is compressed by a lossy compression technique and the QRS template and QRS complex locations are compressed using lossless
20 coding.

In order to meet power consumption requirements for implementation within a subject device for denoising and wireless communication, computer-executable instructions for carrying out an MDSP-based approach as discussed herein, stored as embedded code within the subject device, are configured to facilitate low-power implementation. In one
25 embodiment, the computer instructions are optimized using integer or fixed point arithmetic and lifting or B-spline wavelet implementation [24] of a signal decomposition transform in order to reduce and/or minimize the number of clock cycles or machine states required. In such an embodiment, a portion of the computations required to analyze physiologic signals may be implemented within the subject device while others may be implemented in the data
30 review system. In another embodiment, the subject device captures, denoises, and compresses the ECG of the subject and information is extracted from the ECG recording off-line in the data review system. The data review system may include a review function that facilitates human review of ECGs that were classified as uncertain by the algorithm.

Other embodiments are directed to a computer-based system or logic circuit, such as a computer operating using one or more processor circuits, which operate using executable modules. Each module (*e.g.*, dedicated circuitry, or software-based modules implemented with a processing circuit) is executable by a computer circuit to carry out one or more
5 functions or processes as described herein. For instance, one such module may carry out MDSP computations upon input data such as ECG data, and transform the data into a denoised output. This transformation may involve, for example, the use of a software module that, when executed by a computer, carries out the processes as shown in Figures 1 and 2 herein. Various other embodiments are directed to similar transformations, which may
10 involve taking and processing one or more inputs to generate a transformed output useful for one or more of a variety of purposes, such as for detecting physiological conditions. Accordingly, the embodiments discussed herein may be carried out using such a system and/or computer circuit and related executable modules.

In one embodiment, and referring again to Figure 19, a denoising function is
15 implemented within a computerized apparatus configured to execute programming instructions. A digitized physiological signal 1906 is input to the computerized apparatus. The device computes a denoised output signal from the digitized input signal resulting in an improvement in SNR.

In one implementation, an input physiological signal sensed by ECG electrodes 1901
20 is a relatively noise-free single lead ECG recording from a human being or other mammal with a resting heart rate less than 150 BPM, contaminated with band-limited (0.5 to 100 Hz) white noise. Following denoising in processes 1907, 1908, and 1909 using an MDSP embodiment, the SNR is improved by at least 5 decibels and the mean QRS amplitude for any 10 second interval of the recording is preserved within +/- 10% of the mean QRS
25 amplitude of the input signal for the same 10 second interval.

In this embodiment, the input ECG can be described as residing in a first domain having a dimension equal to the number of leads (*e.g.*, channels) in the recording. For example, a recording consisting of a lead set (*e.g.*, Leads I, II, and III), would have three dimensions. Referring to Figure 19, the digitized input ECG 1906 is decomposed in process
30 1907 into a second domain. Decomposition into the second domain results in generation of subcomponents that represent the information contained in the ECG signal. The dimension of the second domain is defined as the number of subcomponents representing each lead of the ECG multiplied by the number of leads. In one embodiment, decomposition performed in process 1907 is accomplished using a transform that largely achieves independence of the

subcomponents. Independence of the subcomponents combined with qualification of the frequency content of each subcomponent facilitates a more precise identification of the association of a subcomponent or group of subcomponents with an aspect (*e.g.*, T-wave) of the ECG signal.

5 In another embodiment, the subcomponents in the second domain are examined in process 1908, as described herein, based on their time-spectral distribution, to identify noise and other aspects of the ECG. In one embodiment, the time-spectral distribution is examined using spatially selective filtering, whereby aspects or waves of the ECG associated with wider frequency band, such as the QRS complex, are identified and preserved across the
10 subcomponents. In this embodiment, subcomponents associated with high frequencies (*e.g.*, above 30 Hz for ECG recordings from a human being) are preserved for a time window corresponding to the QRS complex, but are zeroed out in (or not used for) the time window corresponding to the remainder of the cardiac cycle. This allows the morphology of the QRS complex to be preserved, while removing the high frequency noise from the remainder of the
15 cardiac cycle, where the primary signal content corresponds to lower frequencies. When the input signal is a multi-lead ECG, principal component analysis and independent component analysis may be used in addition to spatially selective filtering or π CA to further enhance the independence of subcomponents from noise.

In another example embodiment, referring to Figure 20, spatially selective filtering is
20 used to remove noise from an ECG. Figure 20a (top) shows a cardiac cycle of an ECG waveform showing a P-wave, QRS complex 2001, and T-wave, as processed in connection with this embodiment. The QRS complex is detected and, referring to Figure 20b, the time spanning the beginning of Q to end of S defines time window 2002. The remainder of the cardiac cycle is defined as time window 2003. Time window 2002, containing the QRS
25 complex, includes a wide range of frequencies ranging from low-frequencies to high-frequencies. For the ECG of a healthy human, this frequency range may span from 0.05 to 100 Hz. In order to preserve the morphology of the QRS complex, subcomponents in this frequency range are preserved and considered to be associated with a desired signal wave.

In one embodiment, the beginning and end of time window 2002 are respectively
30 defined by the onset of the Q-wave and the offset of the S-wave. In other embodiments, the beginning and end of time window 2002 are shifted somewhat earlier or later. Because the signal waves occurring in time window 2003 (*e.g.*, P-wave, T-wave) contain lower frequency components, the subcomponents comprising the low-frequencies, represented by

region 2005, are preserved and the subcomponents comprising high-frequencies are removed since they are primarily associated with noise.

In one embodiment the time window 2002 is a rectangular window, in another embodiment the time window 2002 decays gradually to reduce signal distortion nearby the transition points. Figure 20d illustrates a three-dimensional view of the time-frequency trapezoidal window 2008 with vertical axis showing window amplitude. Window 2008 decays gradually to enable a graded attenuation of high frequencies (*i.e.*, low subcomponent scales) near the edges of window 2008. The content of ECG signal at low frequencies of a cardiac cycle is preserved as shown by window 2009. Figure 20b is a two-dimensional projection of Figure 20d on the time-frequency plane. In one embodiment, the top of trapezoidal window 2008 extends from Q-wave onset to S-wave offset and the base is expanded by a predetermined time toward the P- and T-waves.

In an alternate embodiment as shown in Figure 20c, subcomponents associated with higher frequencies are removed in time windows 2002 and 2003 while retaining low-frequency components represented by region 2007. Those containing the higher frequencies are then added back during the time spanning time window 2002, as represented by region 2006, to restore the higher frequency components of the QRS complex. In some embodiments, this approach is carried out to mitigate or avoid morphology distortion.

In one embodiment, and referring to Figure 21, an input ECG signal 2101 is decomposed into subcomponents, as described herein, in process 2102. Transforms useful for decomposition in 2101 include discrete cosine transform, wavelet related transform, Karhunen-Loeve transform, short-time Fourier transform, Gabor transform, and filter bank. The ECG is segmented into cardiac cycles in process 2104. In one embodiment, this is accomplished by identifying a defining feature that provides a reliable fiducial point for identifying a cardiac cycle. In one embodiment, the defining feature is the QRS complex of a cardiac cycle. The QRS complex can be identified in 2103 using one of a number of techniques including thresholding, adaptive thresholding, and spatially selective filtering. An embodiment of spatially selective filtering (SSF) for identifying the QRS complex can be used in processes 2102a and 2103. SSF can be implemented by computing a time function that measures time-spectral distribution of subcomponents and comparing the time function to a threshold function. In one embodiment the time function $P_{pw}(t)$ is a point-wise product of selected subcomponents as described in process 2102a. In one embodiment the selected subcomponents comprise at least a majority of QRS complex energy. In one embodiment, the selection of a subcomponent is based upon the criteria that it is primarily associated with

the energy of the QRS complex. In one embodiment of process 2102a, a pointwise product is computed as:

$$P_{pw}(t) = (a_1, a_2, \dots, a_N) \times (b_1, b_2, \dots, b_N) = (a_1 \times b_1, a_2 \times b_2, \dots, a_N \times b_N).$$

In another embodiment, the point-wise product is normalized to match the energy of a subcomponent. In another embodiment the time function $P_{pw}(t)$ is a cross-correlation between the selected subcomponents. In another embodiment the time function $P_{pw}(t)$ is a linear combination of selected subcomponents. The computed time function $P_{pw}(t)$ is compared to a threshold function $T_{pw}(t)$ at each time t in process 2103. A defining feature of a cardiac cycle is detected when $P_{pw}(t) > T_{pw}(t)$. In one embodiment $T_{pw}(t)$ is a predefined constant-value function that is determined by the statistics of the ECG signal during the first few cardiac cycles of the ECG recording being processed. For example, the threshold may be defined as a percentage of the maximum amplitude during the first few cardiac cycles. In another embodiment $T_{pw}(t)$ is a predefined function computed using, for example, low-pass filtering of a subcomponent. In another embodiment $T_{pw}(t)$ is an adaptive function that decays toward a predefined floor following a predefined refractory interval after detecting the defining feature of the cardiac cycle (*i.e.*, QRS complex).

Following identification of the QRS complex, the ECG is segmented by cardiac cycle in process 2104 using the QRS complex as a fiducial point. In one embodiment, each cardiac cycle is partitioned into two time windows in process 2104. Referring to Figure 20, first time window 2002 begins at the onset of the Q-wave of the cardiac cycle and ends at the offset of the S-wave. The second time window 2003 begins at the offset of the S-wave and ends at the onset (*i.e.*, start of window 2002) of the next cardiac cycle. In other embodiments, the start and end times of windows 2002 and 2003 can differ somewhat from the above relative to the location of feature points of the QRS complex and some degree of overlap in the windows is also acceptable. In other embodiments, the cardiac cycle is partitioned into more than two time windows. For example, in an alternate embodiment a first time window includes the QRS complex, a second time window includes the T-wave, and a third time window includes the remainder of the cardiac cycle. The set of subcomponents present within time window 2002 are referred to as QRSsub and the set of subcomponents present in time window 2003 are referred to as PTsub.

In process 2105, subcomponents of the set PTsub are evaluated to identify subcomponent subset PTsubnn that overlaps the spectral content of (are primarily associated with) a characteristic frequency band of the desired signal present in window 2003, FBpt, of

the representative ECG signal. The value of a subcomponent represents the energy of the input ECG signal contained in a narrow range of frequencies.

Process 2105 relies on knowledge of the characteristic frequency band, FBpt, of the desired ECG signal in window 2003. FBpt is a frequency band that contains most of the
5 energy of the desired signal in window 2003. FBpt is pre-identified using a database of representative ECG recordings for a species as described in process 2108.

In process 2106, subcomponents of the set QRSSub are evaluated to identify subcomponent subset QRSSubnn that overlaps the spectral content of (are associated with) a characteristic frequency band, FBqrs, of the representative ECG signal present in window
10 2002 of Figure 20. In one embodiment, the input ECG signal is preprocessed to remove out-of-band noise in a signal conditioning circuit such as 1902 in Figure 19, by a digital filter implemented in a computing or logic circuit such as may be implemented in microcontroller 1903, or a combination thereof. In one embodiment, where the energy of out-of-band noise in the input signal is negligible, the energy of the QRS subcomponents QRSSub and
15 QRSSubnn substantially overlap.

Process 2106 relies on knowledge of the characteristic frequency band, FBqrs, of the desired ECG signal in window 2002. Frequency band FBqrs contains most of the energy of the QRS complex of the desired ECG signal. FBqrs is pre-identified using a database of representative ECG recordings for a species as described in process 2108.

20 The identification of the characteristic frequency bands for a time window is performed by evaluating a database of representative ECG recordings for a species (input 2107). The database of representative ECG recordings is selected to represent a broad scope of ECG morphologies, heart rate, anomalies, noise characteristics, and pathologies. In the process 2108, the spectral content of the QRS complex is characterized and the characteristic
25 frequency band of the QRS complex FBqrs is identified using spectral analysis techniques. The time window outside of the QRS complex in a cardiac cycle is identified and the characteristic frequency band of the cardiac cycle outside of the QRS complex, FBpt, is identified using spectral analysis techniques. In some embodiments the characteristic frequency band is typically only determined once for the ECG sampled for a given species.
30 Once FBpt and FBqrs are determined for a given species, they are used as parameters by the processes 2105 and 2106 and it is not necessary to recompute them.

In the process 2111 the identified target subcomponents contained in subset PTsubnn and subset QRSSubnn are combined and subjected to the inverse of the transform used in process 2102 to construct a denoised ECG signal.

Figure 22 shows a system 2200 for computing a denoised ECG signal from an input signal including a desired ECG signal and noise, according to another example embodiment. The system includes a logic circuit 2210 and a memory circuit 2212. The memory circuit 2212 stores instructions that, when executed by the logic circuit, carry out the following steps. For illustration, Figure 22 represents steps as carried out in the logic circuit 2210 with various blocks; however, various approaches for carrying out the steps may be implemented in connection with other embodiments. Moreover, the logic circuit 2210 may include two or more logic circuits, such as two or more processors that carry out different aspects of the respective steps. In addition, the various steps shown in Figure 22 and described here may be implemented using one or more embodiments as discussed above, in connection with the other figures and otherwise.

Referring again to Figure 22, an input ECG signal 2205, in a first domain, is received at a communications input port 2214, and is decomposed at block 2220 into subcomponents 2225 in a second domain. QRS window-based identification is carried out at block 2230, at which a location of the QRS complex of a cardiac cycle in the ECG signal is identified. A first time window in the cardiac cycle that includes the QRS complex is identified, along with at least one time window in the cardiac cycle that does not include the QRS complex. The first time window and the at least one time window span the duration of the cardiac cycle.

At block 2240, and for each of the identified time windows, target subcomponents 2245 are identified as subcomponents that contain more desired ECG signal energy than noise energy. For example, when a particular subcomponent exhibits more energy associated with a desired ECG signal than energy not associated with such a signal, that subcomponent may be identified as a target subcomponent. The target subcomponents 2245 are used at block 2250 to construct a denoised physiological signal 2255.

The denoised ECG signal is then output via a communications output port 2216. This output may, for example, involve a wired or wireless communication, and may be carried out in accordance with one or more embodiments as described herein. In some implementations, some or all of the logic circuit 2210 is included as part of an implantable device, and carries out some or all of the steps in the implantable device, for generating the denoised physiological signal 2255. Using this approach, and as consistent with the above, the logic circuit 2210 can be implemented to significantly reduce the size of the denoised and compressed physiological signal 2255 (*e.g.*, as relative to approaches that do not denoise

in this manner), and facilitate the communication of the denoised and compressed physiological signal using less data and, correspondingly, lower power.

Those skilled in the art will appreciate that various alternative logic circuits or computing arrangements, including one or more processors and a memory arrangement
5 configured with program code, would be suitable for carrying out the approaches as discussed herein, including those discussed in connection with Figure 22 above, along with data structures for organizing the required data. Such computer code can be encoded in a processor executable format and may be stored on and/or provided via a variety of computer-readable storage media or delivery channels such as magnetic or optical disks or
10 tapes, electronic storage devices, or as application services over a network. With specific reference to Figure 22, the logic circuit 2210 (and memory 2212, where appropriate) may be implemented with separate components on a circuit board or may be implemented internally within an integrated circuit. When implemented internally within an integrated circuit, the logic circuit 2210 can be implemented as a microcontroller.

15 The architecture of the logic circuits, processors and computer type circuits as described herein depends on implementation requirements as would be recognized by those skilled in the art. In this context, these components may be one or more general purpose processors, or a combination of one or more general purpose processors and suitable co-processors, or one or more specialized processors (*e.g.*, RISC, CISC, and pipelined).

20 Referring again to Figure 22, the memory circuit 2212 may include multiple levels of cache memory and a main memory, and local and/or remote persistent storage such as provided by magnetic disks, flash, EPROM, or other non-volatile data storage. The memory circuit 2212 may be read or read/write capable. The logic circuit 2210 may store instructions (*e.g.*, software) in the memory circuit 2210, read data from and stores data to the
25 memory circuit 2212, and communicate with external devices through the input/output ports 2214 and 2216. These functions may be synchronized by a clock signal generator that may be part of the logic circuit 2210. The resources of the logic circuit 2210 may be managed by either an operating system, or a hardware control unit.

Referring to process 1907 in Figure 19, example transforms that can be used for
30 decomposing the signal into a second domain while enhancing, or maximizing, the independence of the subcomponents include a discrete cosine transform, a wavelet related transform, a Karhunen-Loeve transform, a Fourier transform, a Gabor transform, or a filter bank.

In another embodiment, referring to Figure 16, a denoising function and a data compression function are implemented within a computerized apparatus (*e.g.*, a logic circuit) configured to execute programming instructions. In this embodiment, denoising and compression of an ECG results in a reduction in bit rate required to retain the information in the signal. By compressing the ECG signal prior to wireless transmission, fewer bits can be used to achieve data transmission, and hence the power consumed in the transmission (*e.g.*, a telemetry link) is reduced. This approach may also be implemented when storing ECG data (or other signal data) in memory, to reduce the data storage space required.

In a more particular example embodiment, denoising and compression involves reducing the bit rate of a signal by a factor of 15:1 to 20:1, relative to the bit rate of input signal. For instance, such a compression factor can be achieved with a 16-bit digitized single lead ECG having a bit rate of 4,000 bits per second and sampled at 250 Hz, and with band-limited (0.5 to 100 Hz) noise corresponding to a SNR of 4 dB (computed as

$$SNR_{dB} = 10 \log_{10} \left(\frac{P_{signal}}{P_{noise}} \right) \quad \text{where } P_{signal} \text{ and } P_{noise} \text{ are respective signal and noise energy.}$$

The compressed signal is communicated to a receiving device where the compressed ECG signal is reconstructed. The denoising approach facilitates reconstructing QRS complex, from the compressed signal, having mean amplitude for any 10-second interval therein that differs from the mean amplitude of the input ECG by 10% or less. In one embodiment, communication is performed using a wireless link such as a Bluetooth transceiver. In some embodiments, events in the ECG, such as arrhythmias, are detected and compression and transmission of the ECG is triggered based on the presence of a detected event.

In this example embodiment, the input ECG is processed by the computerized apparatus to first denoise the signal, then detect cardiac cycles, segment the ECG by cardiac cycles and align them in time according to an identifiable fiduciary (*e.g.*, a QRS peak), and form an image consisting of aligned cardiac cycles. The image is then encoded using a lossy compression technique such as transform encoding, subband encoding, or wavelet based encoding. The encoded image contains nearly all of the information in the denoised ECG, but does so with about 15-fold fewer bits. The encoded image is reconstructed at a receiver using the inverse of the transform used to encode the image, and ECG segments are reconnected to form a denoised version of the input signal. In one embodiment, denoising prior to compression is achieved in the same manner as described herein for the apparatus used for denoising.

In another embodiment, cardiac cycles are detected by computing a QRS emphasis signal and evaluating peaks, valleys, or slopes to identify the QRS complex. In another embodiment, the QRS complex is detected and a template QRS complex that is representative of the average complex is subsequently computed. The template is then subtracted from each QRS complex, creating a time series consisting of the resulting difference. The time series of residuals is subsequently encoded using a lossy compression technique such as transform encoding, subband encoding, or wavelet based encoding. The template QRS is encoded using a lossless transform such as Huffman encoding. This approach may be useful, for example, in connection with an embodiment in which a relatively simpler and less computationally intense compression technique is appropriate (e.g., where QRS morphology is not expected to change rapidly). Accordingly, the computational intensity of denoising and resulting signal quality/size can be weighed as a tradeoff, for implementation in various embodiments.

Another example embodiment is directed to a system and approach for denoising, that facilitates an improvement in signal-to-noise ratio (SNR_A) of between about 14 and 26 dB while preserving signal morphology for a controlled application using a signal-to-noise computation as follows. Considering a noise free single-channel 30-minute digitized human ECG recording sampled at 250 Hz with average heart rate of about 83 BPM, the signal is contaminated with controlled levels of additive band-limited (0.5 to 100 Hz) white noise. Testing can be carried out using, for example, multiple copies of the 30-minute ECG strip, each contaminated with a known and different noise level. Each strip is processed using an MDSP-type denoising approach, and signal-to-noise ratio improvement and morphology preservation can be quantified. Signal-to-noise ratio is computed as described in the ANSI/AAMI EC57:1998 standard [27, 28], and is referred to as SNR_A .

Morphology preservation is computed as:

$$QSR = 100\% \left(1 - \frac{\sum_{i=1}^N (ECG_{clean} - ECG_{denoised})^2}{\sum_{i=1}^N (ECG_{clean})^2} \right)$$

30

in which ECG_{clean} is the input ECG signal prior to imputing white noise and $ECG_{denoised}$ is the ECG signal after denoising/filtering. This approach quantifies preservation of signal morphology through a point-by-point comparison of the denoised signal to the clean ECG, prior to imputing noise. SNR_A improvement is calculated as SNR_A of the output signal less

35 SNR_A of the input signal.

Figure 23a shows SNR_A improvement and QSR for MDSP denoising of ECG recordings with controlled levels of imputed band-limited white noise, in accordance with a particular embodiment. An example ECG trace is shown in Figure 23b before (top) and after (bottom) for a signal processed using an MDSP-type denoising system/approach.

5 Figure 23b shows an example ECG recording according to a controlled application with 0 dB SNR_A as a result of imputed band-limited white noise (top) and post-MDSP denoising (bottom), in accordance with another example embodiment. The input trace shown at the top of Figure 23b is imputed with band-limited white noise to provide $SNR_A =$ 0 dB. In the bottom ECG trace, SNR_A has been improved by about 26 dB as a result of
10 denoising using an MDSP-based system/approach as discussed herein. SNR_A improvements ranging from 14 to 26 dB for input SNR_A ranging from 0 to 40 dB can be realized, with a higher degree of SNR_A improvement for lower input SNR_A . QSR ranges from about 97.5 to 99.8%, indicating negligible distortion of ECG morphology. Accordingly, various circuits, systems and approaches as discussed herein may be configured for operation in accordance
15 with Figures 23a and 23b.

In a more particular implementation, a system and/or method for denoising a signal is configured for implementation in denoising a single-channel human ECG-based signal sampled at 250 Hz and contaminated with 0.5 to 100 Hz band-limited white noise between about 0 to 40 dB SNR_A . Using approaches as discussed herein, a denoised physiological
20 signal is constructed by, for an input signal having a 40 dB SNR_A , constructing a denoised physiological signal having a quality of signal reconstruction (QSR) of at least 99% and a SNR_A that is at least 10 dB greater than the SNR_A of the input signal. For an input signal having a 0 dB SNR_A , a denoised physiological signal having a QSR of at least 95% and a SNR_A that is at least 22 dB greater than the SNR_A of the input signal is constructed.

25 Figure 24 shows a system 2400 and related flow functions for computing a denoised physiological signal from an input signal including a desired physiological signal and noise signal, in accordance with another example embodiment of the present invention. At block/module 2410, an input signal is decomposed from a first domain into subcomponents of the input signal in a second domain. These subcomponents are passed to block/module
30 2420, via which target subcomponents of the input signal that are associated with the desired physiological signal are identified, based upon a time-spectral distribution of the subcomponents. The target subcomponents are provided for block/module 2430, which constructs a denoised physiological signal in the first domain from at least one of the identified target subcomponents. The respective blocks/modules may be implemented using,

for example, a processing circuit executing modules to process input signals/data and generate output signals/data as shown. In some implementations, the system 2400 includes components such as shown in Figure 16 and/or otherwise, such as ECG leads coupled to provide the input signal, and output communications devices for communicating the denoised signal and/or otherwise processing the denoised signal.

For general information regarding a variety of fields that may relate to one or more embodiments of the present disclosure, and for specific information regarding the application of one or more such embodiments, reference may be made to the following documents. Various ones of these references are further cited above via corresponding numerals, and may be implemented as such.

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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 disclosure without strictly following the exemplary embodiments and applications illustrated and described herein. Such modifications and changes may include, for
5 example, using one or more aspects described in the above references and/or applying one or more embodiments thereto, or combining embodiments. Other changes may include using subcomponents having energy that is more associated with noise than with a desired signal, such as for applications in which a lower quality or resolution may be in order. These and other modifications do not depart from the scope of the present disclosure,
10 including that set forth in the following claims.

CLAIMS:

1. A method for identifying a QRS complex in an electrocardiogram (ECG), the method comprising:
 - decomposing the ECG into subcomponents;
 - selecting a subset of the subcomponents based upon a degree of overlap of spectral energy, in at least one of the subcomponents, with expected spectral energy of the QRS complex of the ECG;
 - combining at least two of the subcomponents in the subset;
 - comparing the combined subcomponents to a threshold; and
 - identifying the location of the QRS complex in the ECG based on the comparing, and outputting a signal indicative of the identified location of the QRS complex, relative to other aspects of the ECG.
2. The method of claim 1, further comprising computing the threshold based upon an estimated level of noise energy in an isoelectric portion of the ECG.
3. The method of claim 2, further comprising estimating the level of noise energy using one of variance, zero crossings, and amplitudes of peaks and valleys in the ECG.
4. The method of claim 1, further including computing the threshold by:
 - selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral energy in the ECG;
 - combining at least two of the noise subcomponents; and
 - setting the threshold based upon the combination of the at least two of the noise subcomponents.
5. The method of claim 4, wherein selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral

energy in the ECG includes selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral energy in a portion of the ECG that excludes the QRS complex.

6. The method of claim 4 wherein combining at least two of the noise subcomponents includes combining noise subcomponents of a portion of the ECG outside the QRS complex.
7. The method of claim 4, wherein the portion of subcomponent spectral energy attributable to noise spectral energy is at least one-half of the total energy of the subcomponent.
8. The method of claim 4, wherein combining at least two of the noise subcomponents includes computing one of a point-wise product of the at least two of the noise subcomponents, a linear combination of the at least two of the noise subcomponents, and a cross-correlation of the at least two of the noise subcomponents.
9. The method of any one of claims 1 to 8, wherein combining includes computing one of a point-wise product of the at least two of the subcomponents, a linear combination of the at least two of the subcomponents, and a cross-correlation of the at least two of the subcomponents.
10. The method of any one of claims 1 to 8, wherein decomposing the ECG into subcomponents includes generating time-synchronized subcomponents.
11. The method of claim 10, wherein generating time-synchronized subcomponents includes applying a transform to the ECG, the transform including one of: a non-orthogonal wavelet transform, an undecimated wavelet transform, a stationary wavelet transform, and a shift-invariant wavelet transform.

12. The method of any one of claims 1 to 8, wherein the degree of overlap of the spectral energy, of the at least one of the subcomponents, with the spectral energy of the QRS complex of the ECG, is at least one-half.

13. The method of any one of claims 1 to 8, wherein selecting the subset of the subcomponents includes selecting a subcomponent in the subset of subcomponents based upon a degree of overlap of spectral energy, in the subcomponent, with expected spectral energy of the QRS complex of the ECG.

14. An apparatus comprising:

a computer-based circuit configured and arranged with a plurality of modules for identifying a QRS complex in an electrocardiogram (ECG), the modules including:

a decomposing module configured and arranged to decompose the ECG into subcomponents;

a selection module configured and arranged to select a subset of the subcomponents based upon a degree of overlap of spectral energy, in at least one of the subcomponents, with expected spectral energy of the QRS complex of the ECG;

a combiner module configured and arranged to combine at least two of the subcomponents in the subset;

a comparator module configured and arranged to compare the combined subcomponents to a threshold; and

an identifier module configured and arranged to identify the location of the QRS complex in the ECG based on the comparing, and to output a signal indicative of the identified location of the QRS complex, relative to other aspects of the ECG.

15. The apparatus of claim 14, further comprising a threshold module configured and arranged to compute the threshold based upon an estimated level of noise energy in an isoelectric portion of the ECG.

16. The apparatus of claim 15, wherein the threshold module is configured and arranged to estimate the level of noise energy using one of variance, zero crossings, and amplitudes of peaks and valleys in the ECG.

17. The apparatus of claim 14, further comprising a threshold module configured and arranged to compute the threshold by:

- selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral energy in the ECG;
- combining at least two of the noise subcomponents; and
- setting the threshold based upon the combination of the at least two of the noise subcomponents.

18. The apparatus of claim 17, wherein the selection module is configured and arranged to select the subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral energy in the ECG by selecting a subset of noise subcomponents based upon a portion of subcomponent spectral energy attributable to expected noise spectral energy in a portion of the ECG that excludes the QRS complex.

19. The apparatus of claim 17 wherein the threshold module is configured and arranged to combine at least two of the noise subcomponents by combining noise subcomponents of a portion of the ECG outside the QRS complex.

20. The apparatus of claim 17, wherein the threshold module is configured and arranged to select the subset of noise subcomponent based upon a portion of

subcomponent spectral energy attributable to noise spectral energy that is at least one-half of the total energy of the subcomponent.

21. The apparatus of claim 17, wherein the threshold module is configured and arranged to combine the at least two of the noise subcomponents by computing one of a point-wise product of the at least two of the noise subcomponents, a linear combination of the at least two of the noise subcomponents, and a cross-correlation of the at least two of the noise subcomponents.

22. The apparatus of any one of claims 14 to 21, wherein the combiner module is configured and arranged to combine the at least two of the subcomponents by computing one of a point-wise product of the at least two of the subcomponents, a linear combination of the at least two of the subcomponents, and a cross-correlation of the at least two of the subcomponents.

23. The apparatus of any one of claims 14 to 21, wherein the decomposing module is configured and arranged to decompose the ECG into subcomponents by generating time-synchronized subcomponents.

24. The apparatus of claim 23, wherein the decomposing module is configured and arranged to generate the time-synchronized subcomponents by applying a transform to the ECG, the transform including one of: a non-orthogonal wavelet transform, an undecimated wavelet transform, a stationary wavelet transform, and a shift-invariant wavelet transform.

25. The apparatus of any one of claims 14 to 21, wherein the selection module is configured and arranged to select the subset based upon a degree of overlap of the spectral energy that is at least one-half.

26. The apparatus of any one of claims 14 to 21, wherein the selection module is configured and arranged to select the subset by selecting each of the subcomponents in the subset based upon a degree of overlap of spectral energy, in the subcomponent, with expected spectral energy of the QRS complex of the ECG.

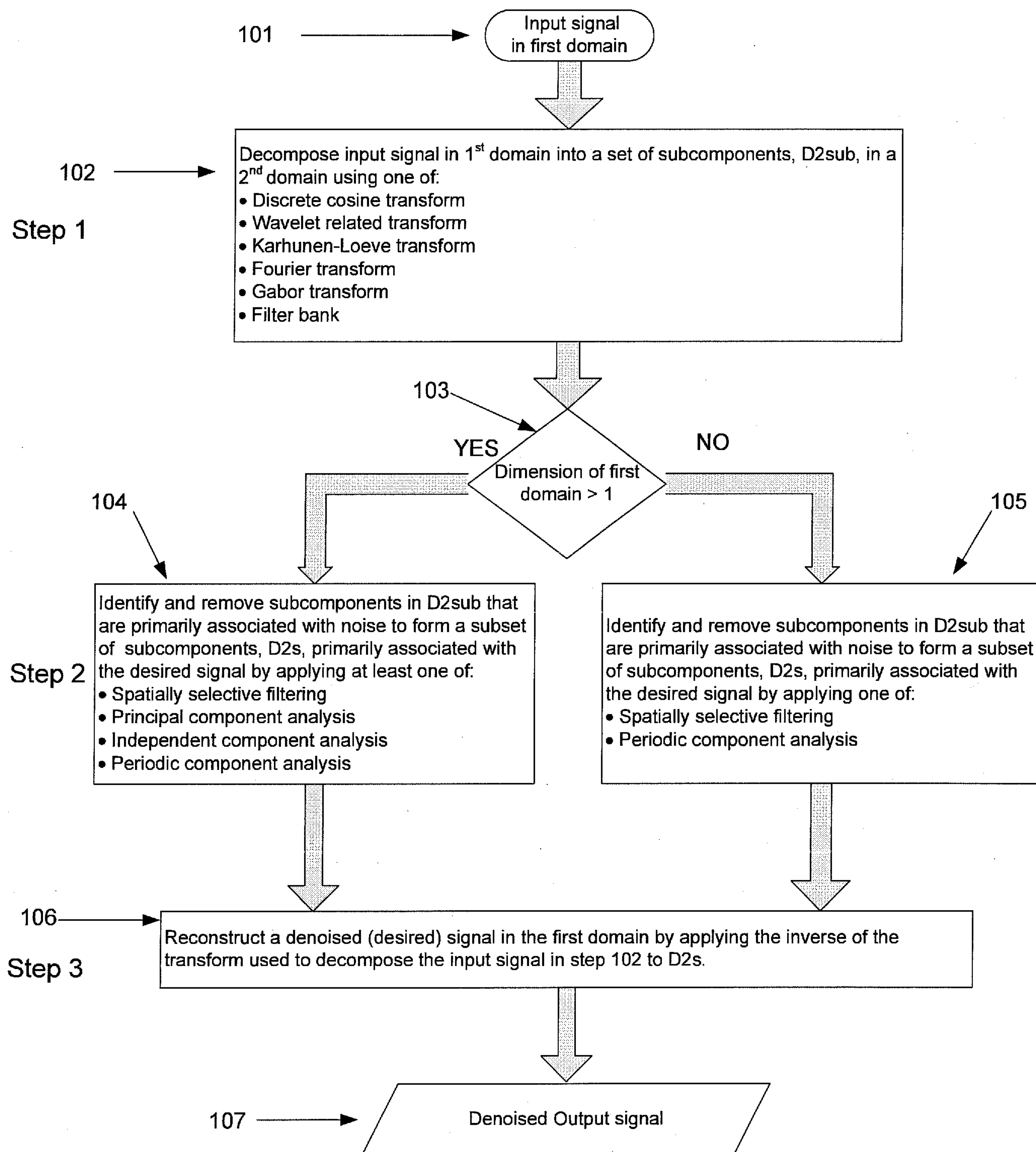
Figure 1

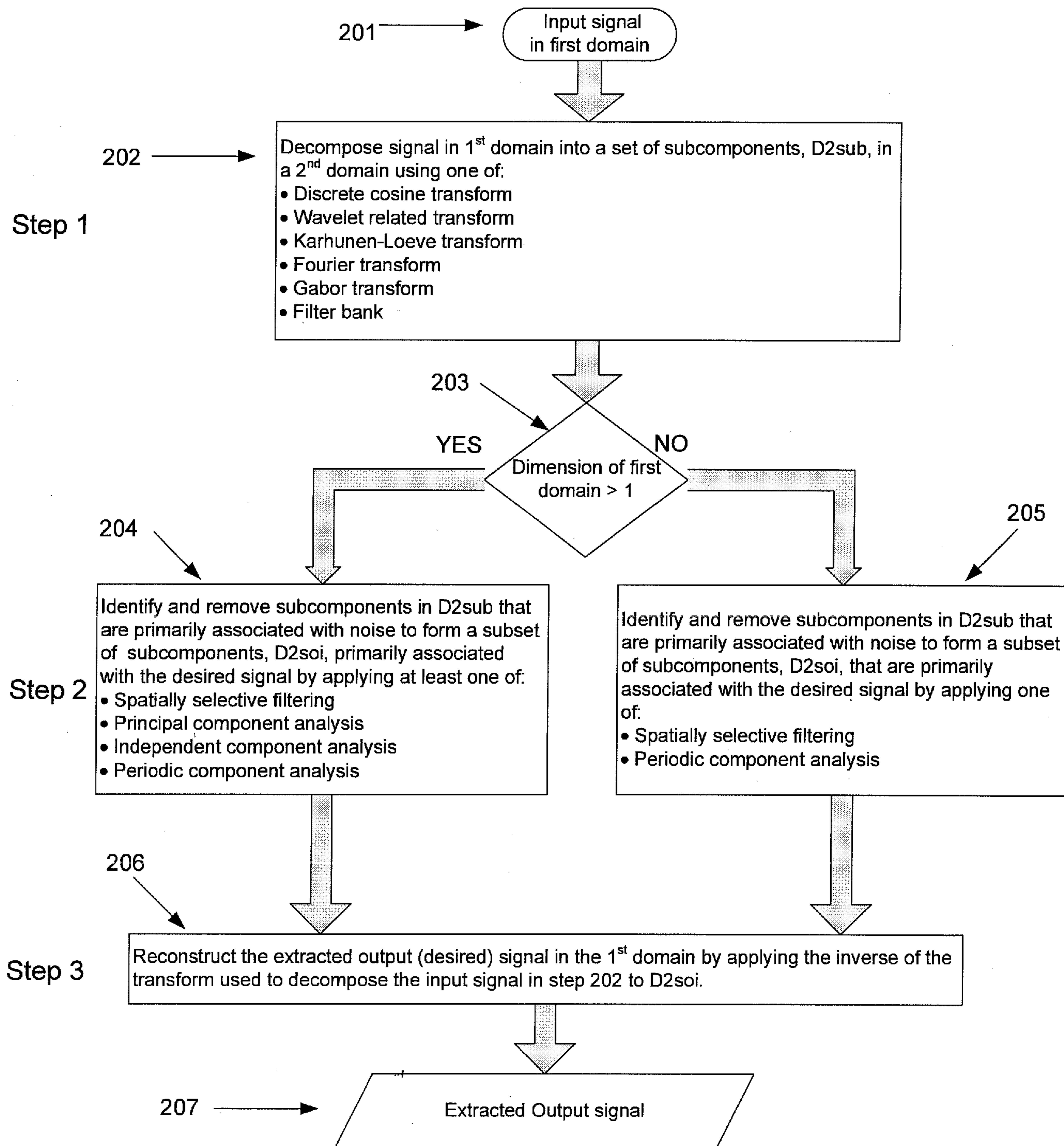
Figure 2

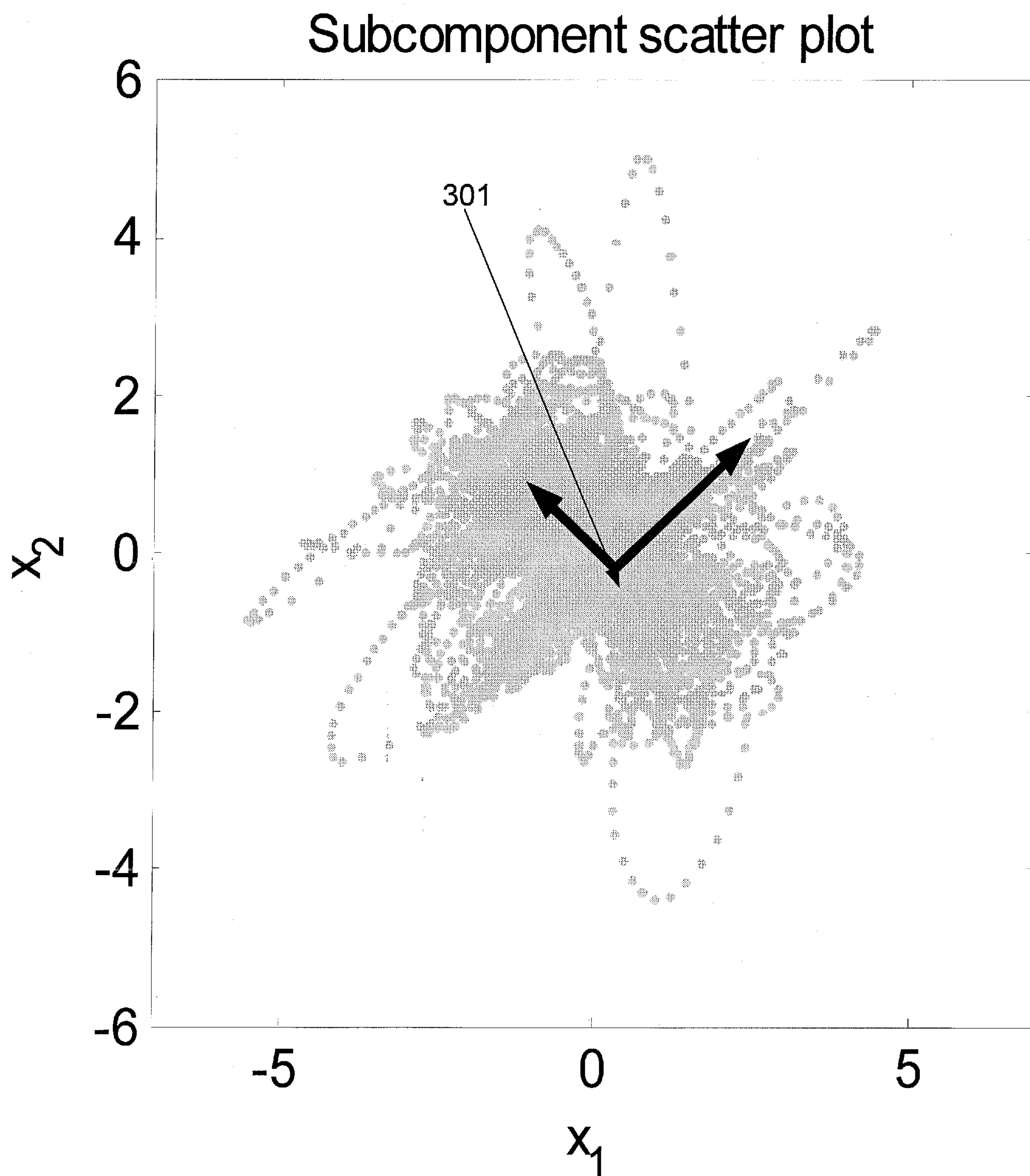
Figure 3a

Figure 3b

Post PCA+ICA scatter plot

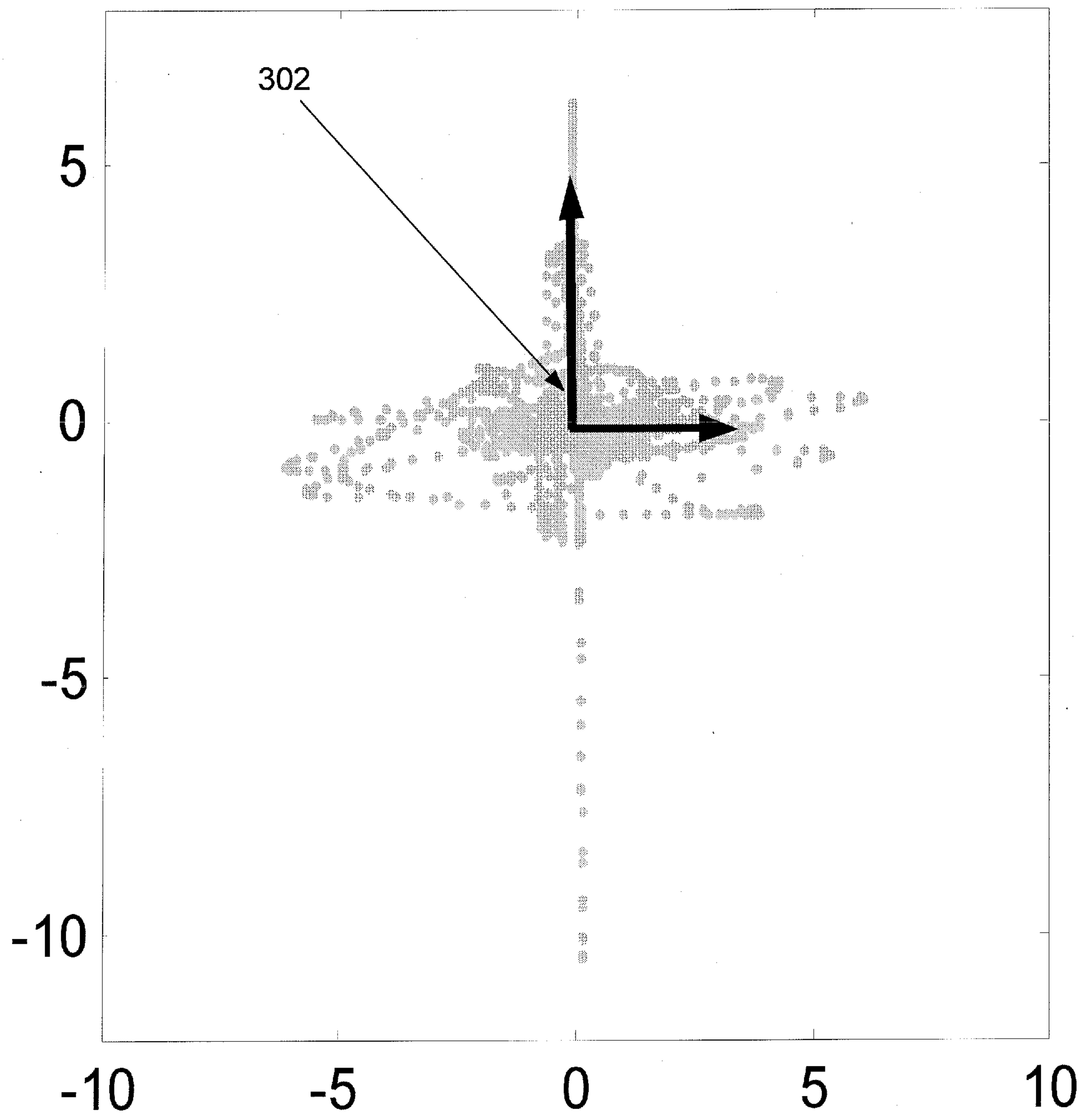


Figure 4

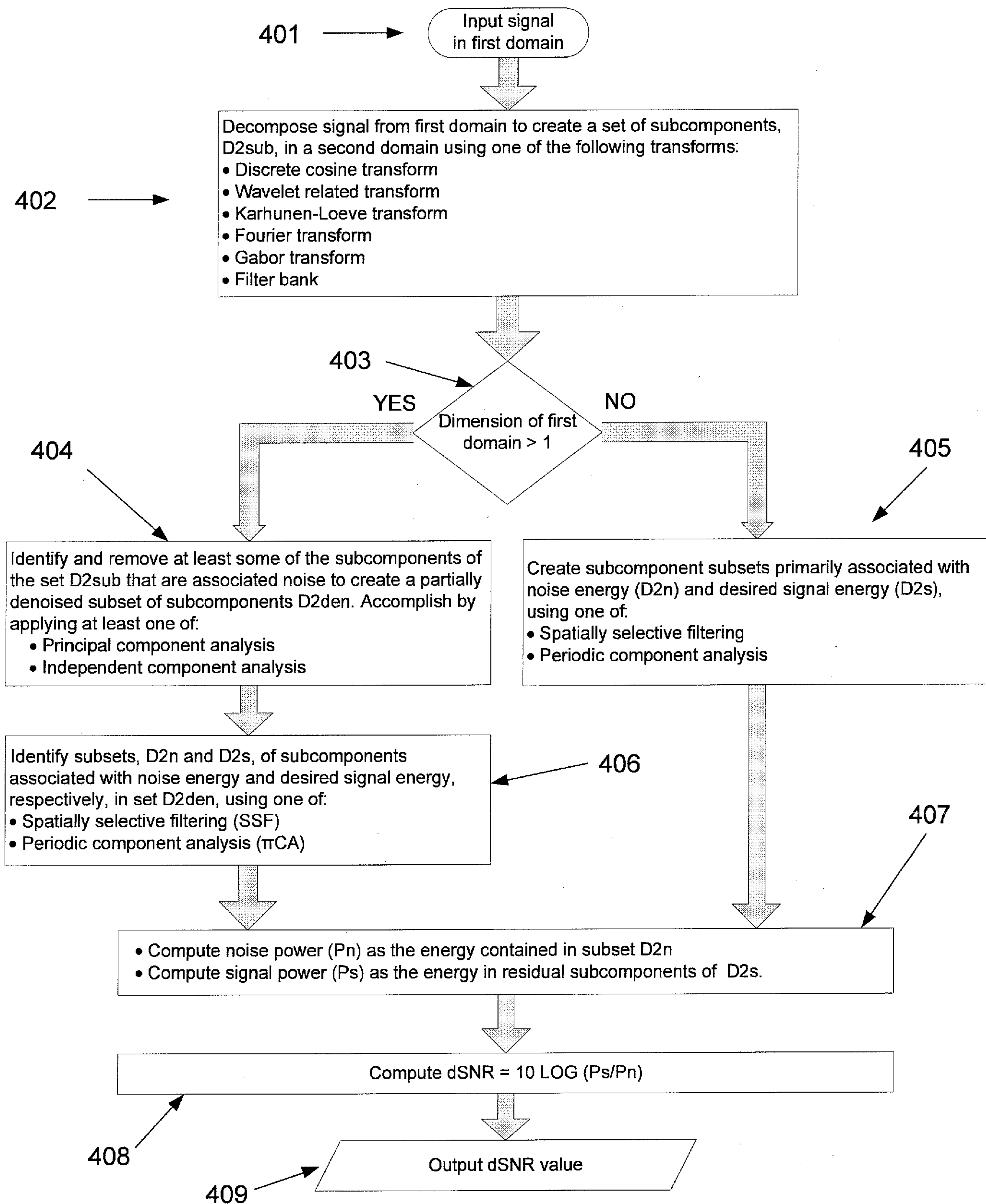


Figure 5

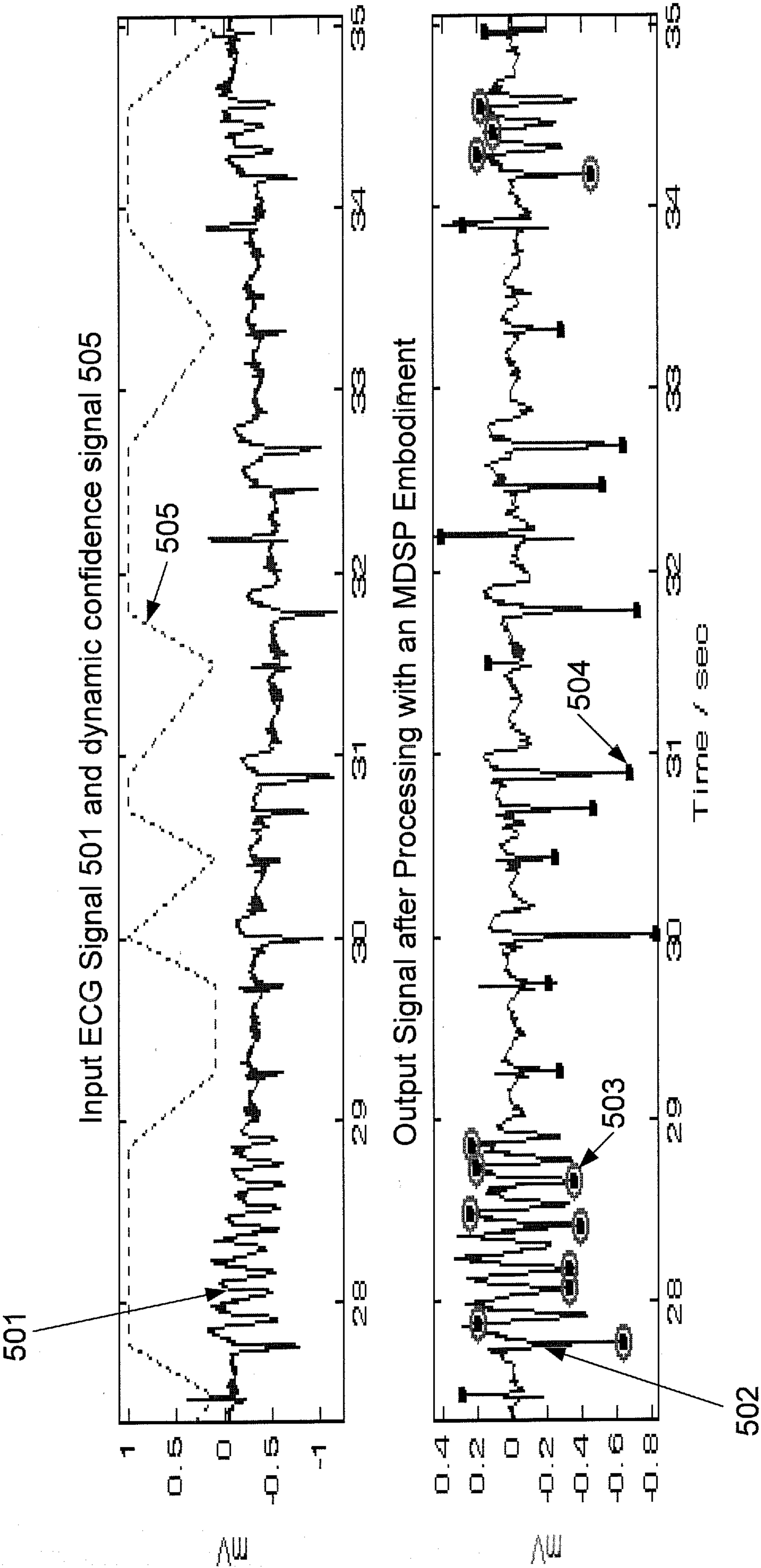
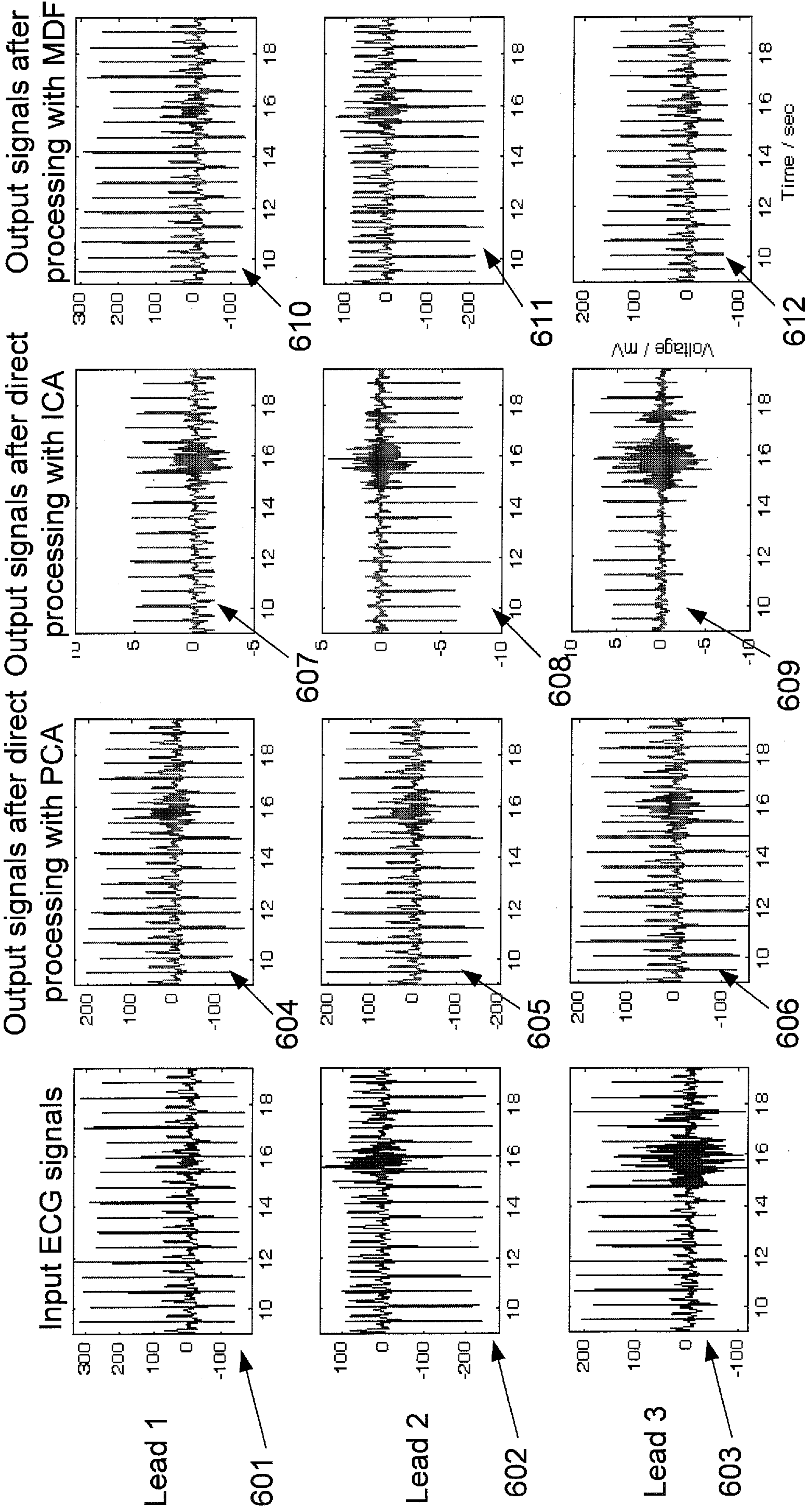


Figure 6



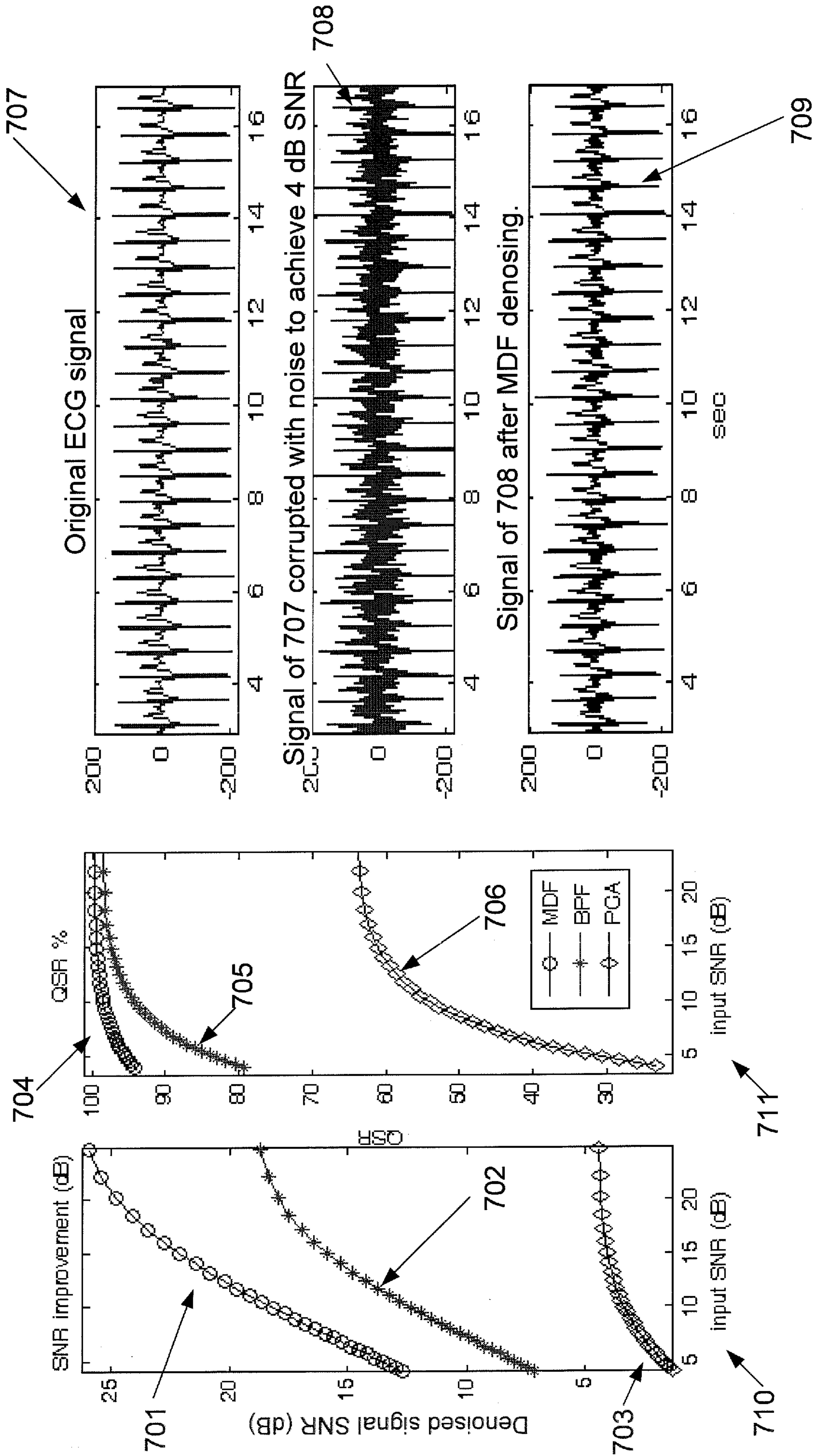
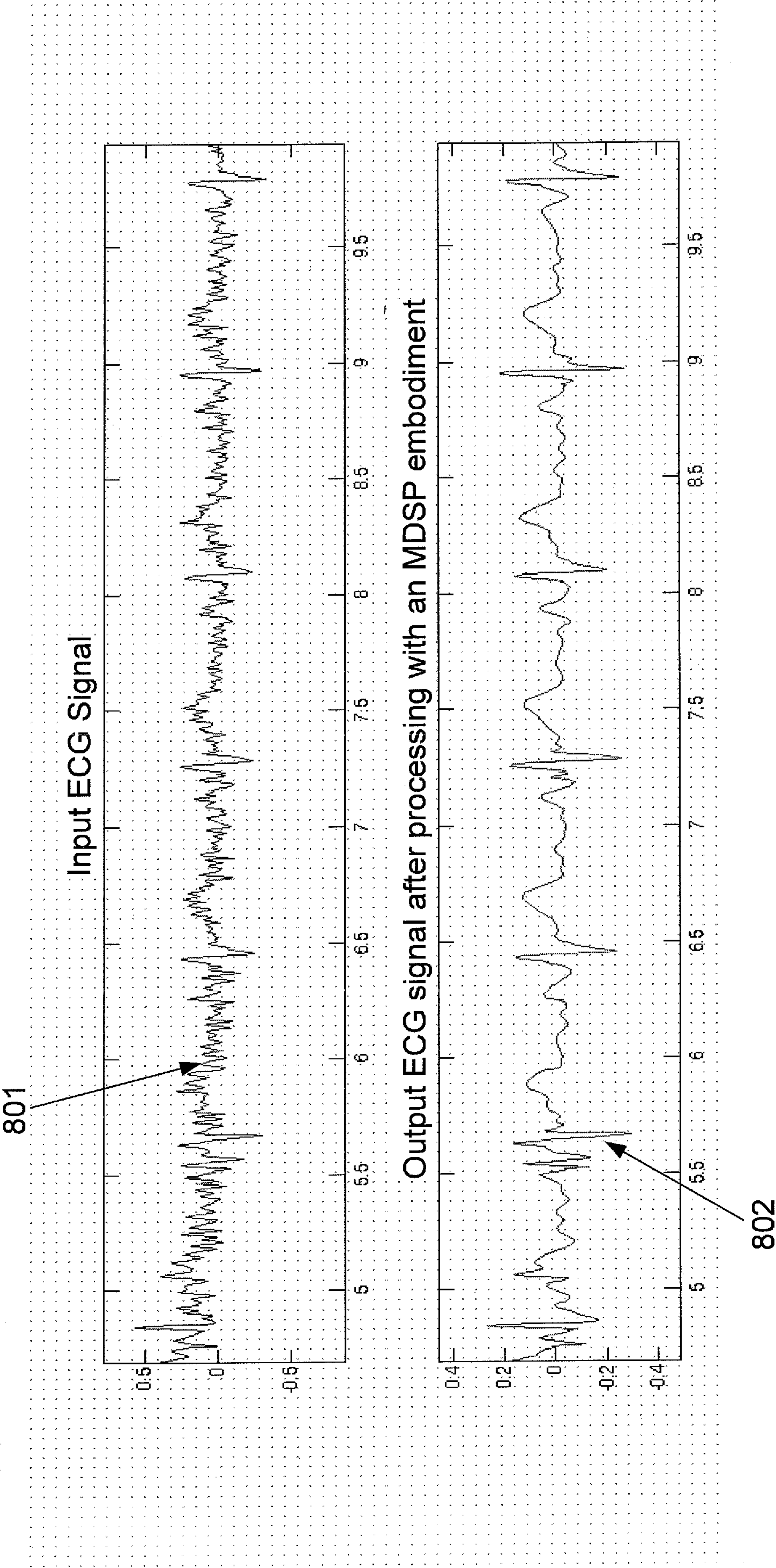


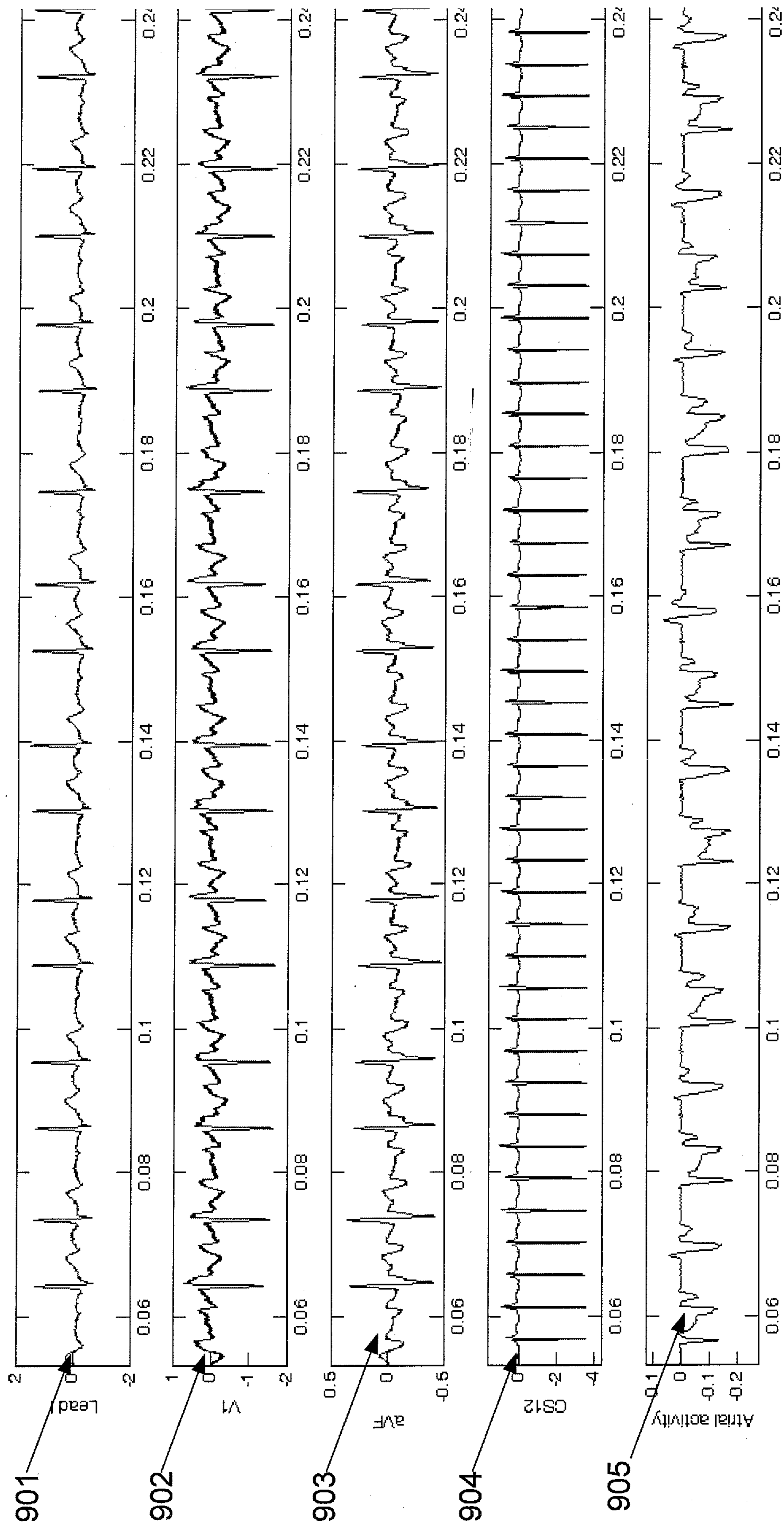
Figure 7

Figure 8



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Figure 9



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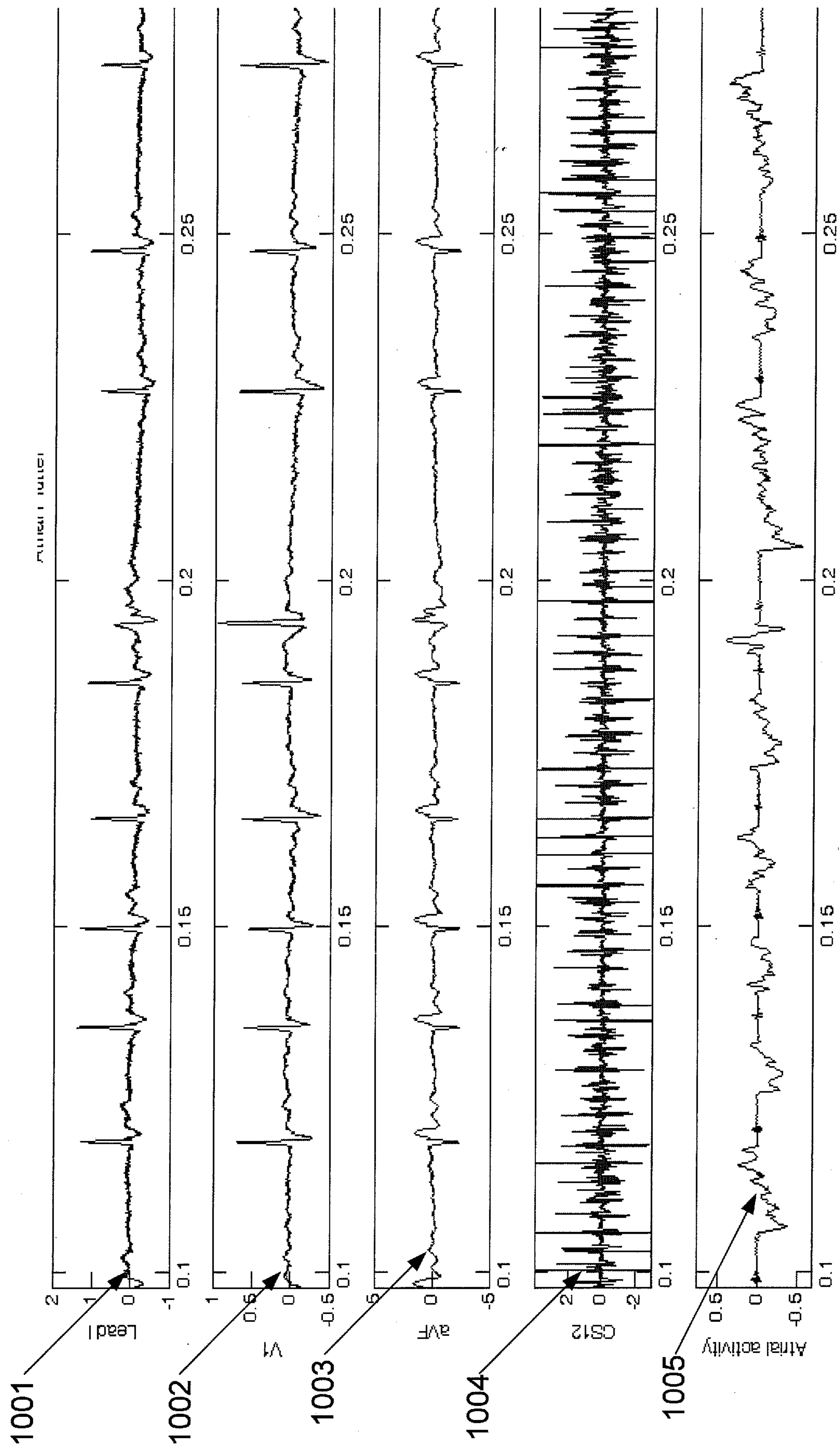
Figure 10

Figure 11

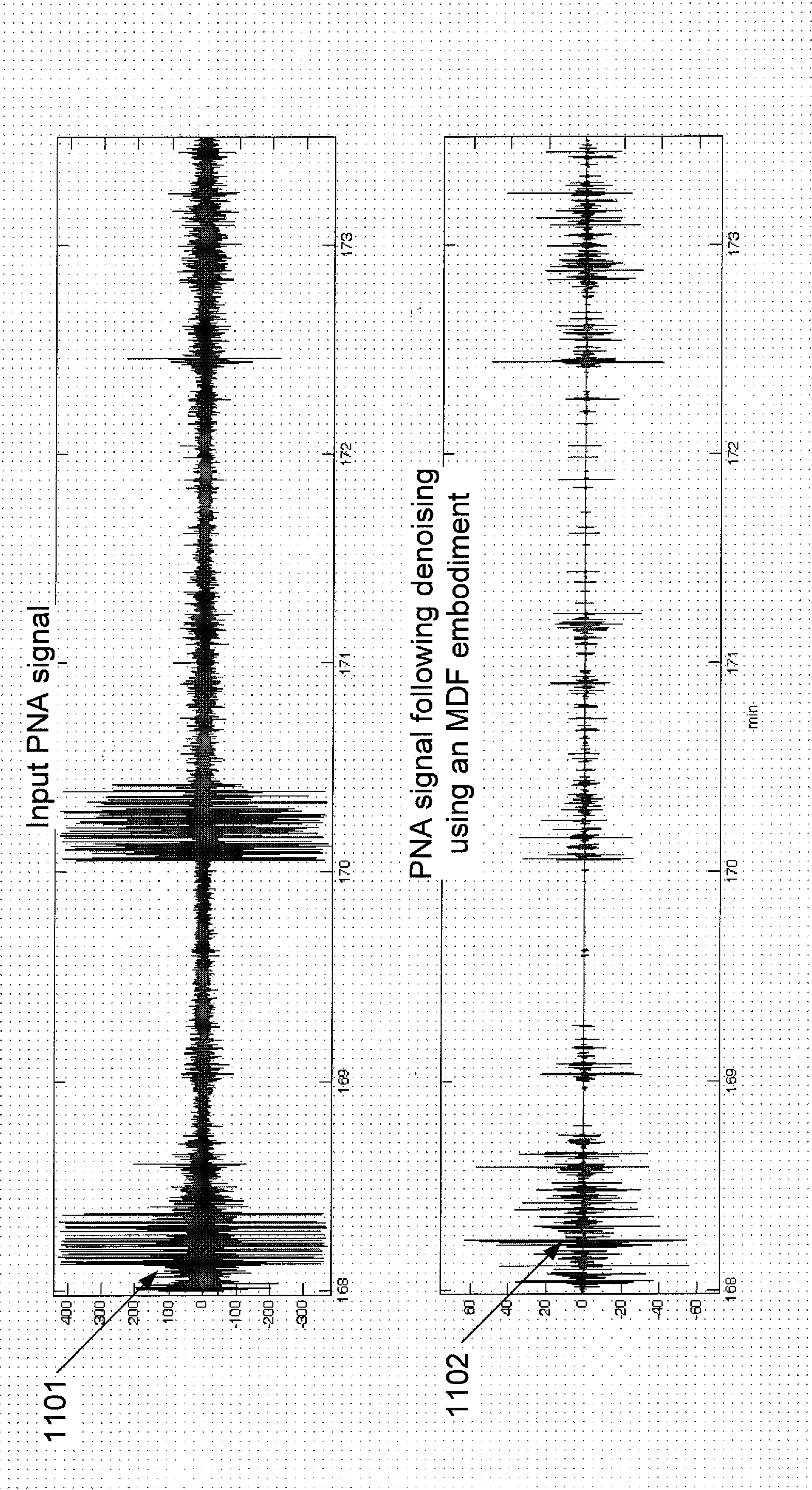
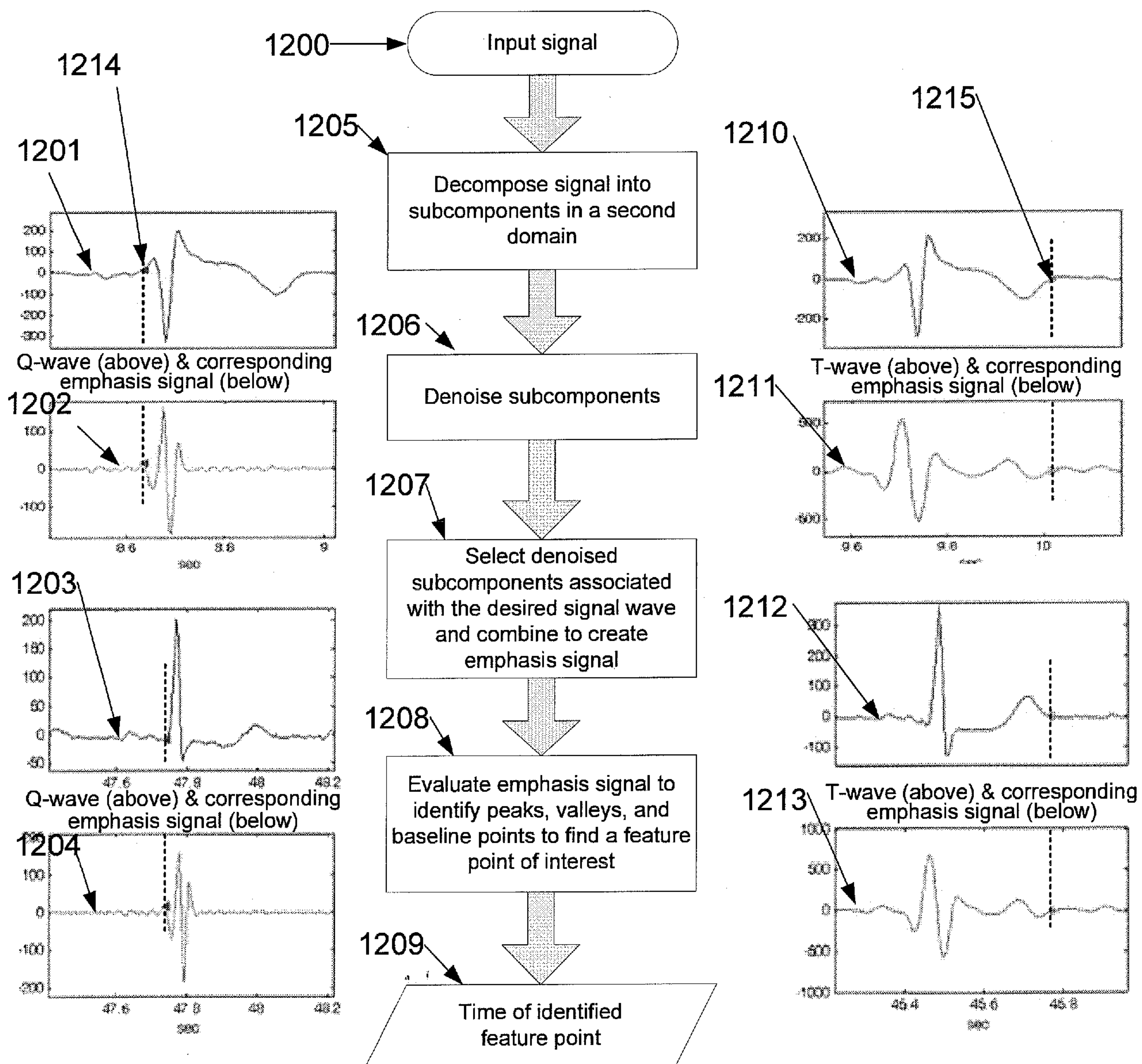


Figure 12

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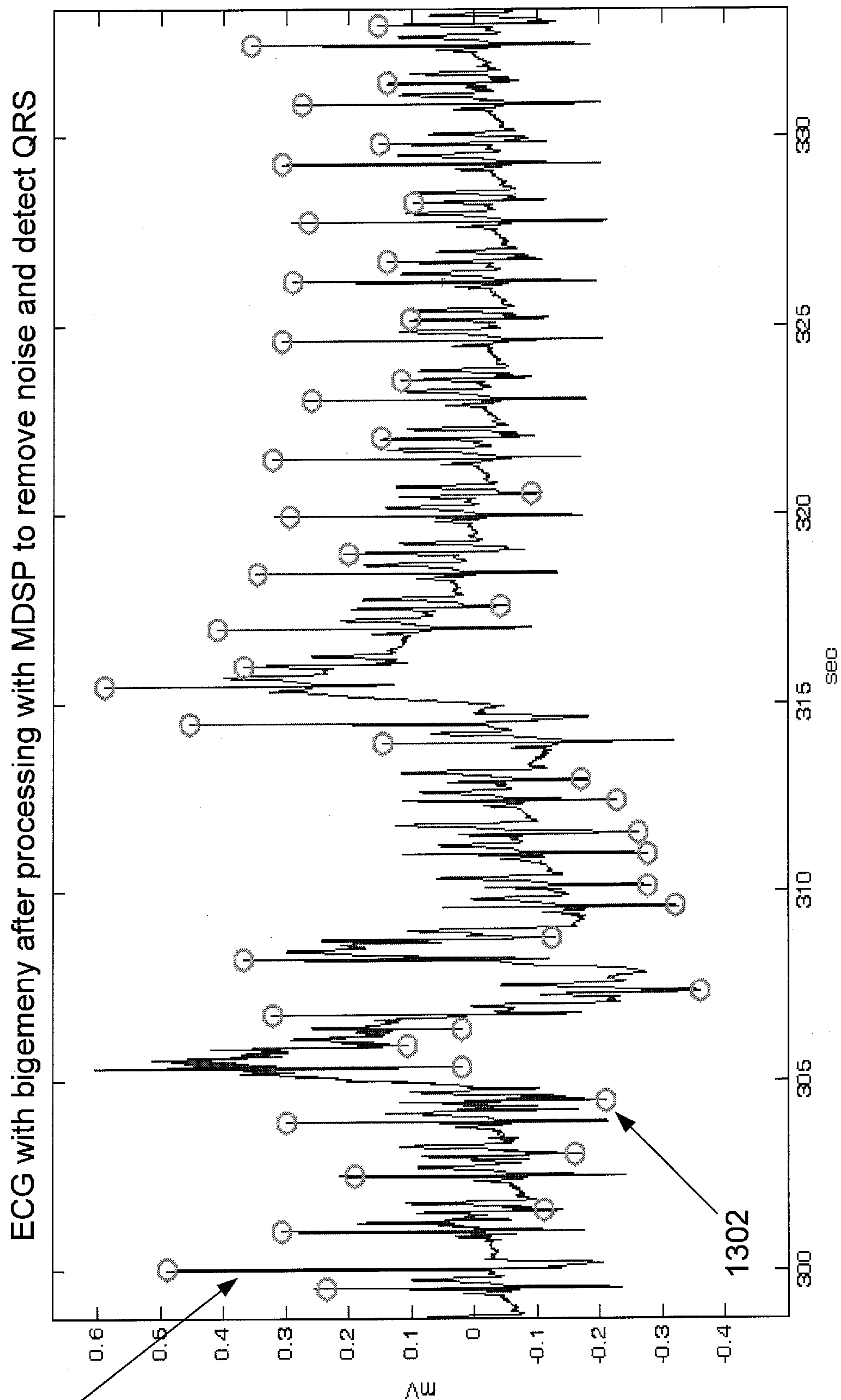


Figure 13

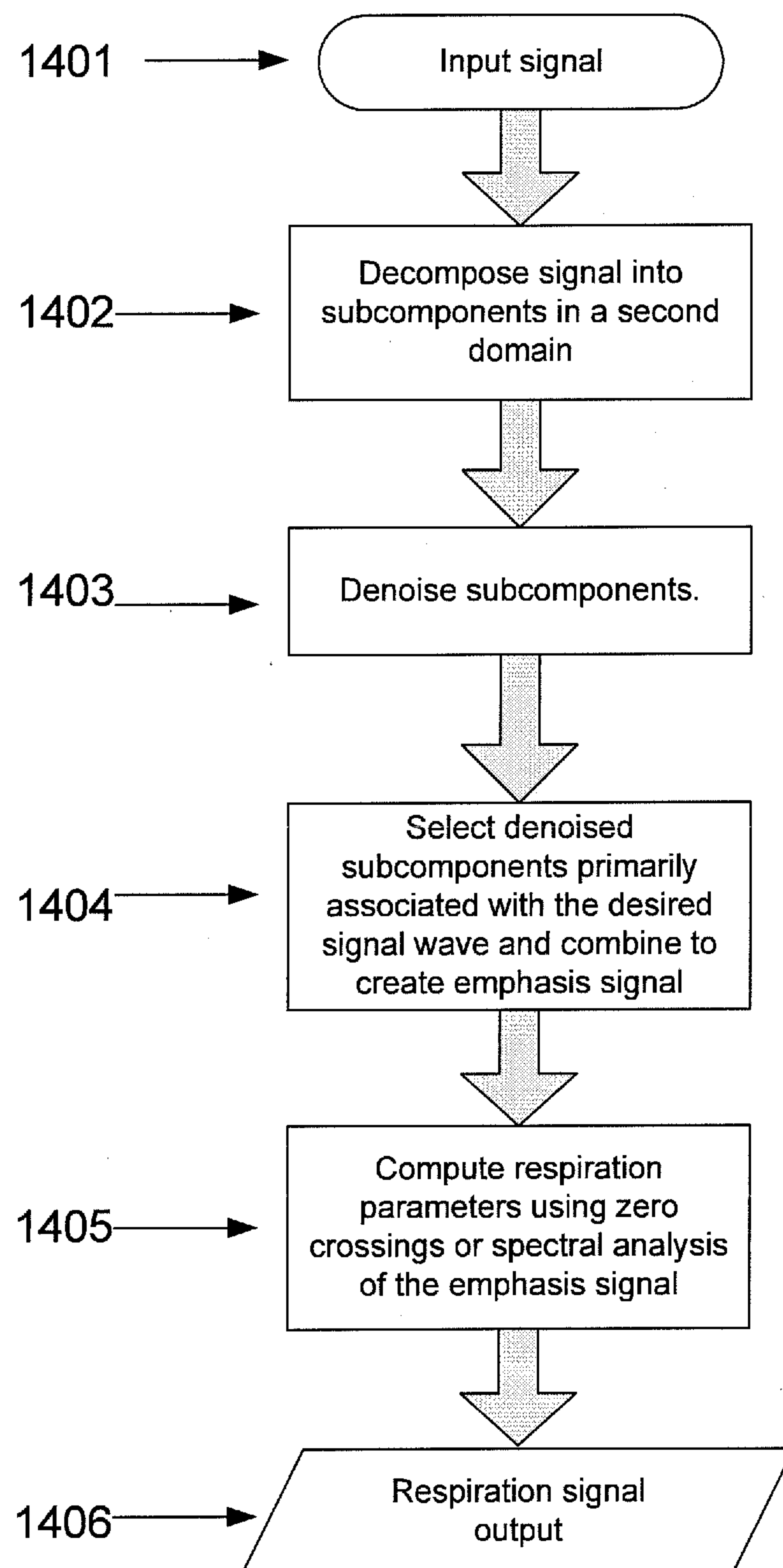
Figure 14

Figure 15

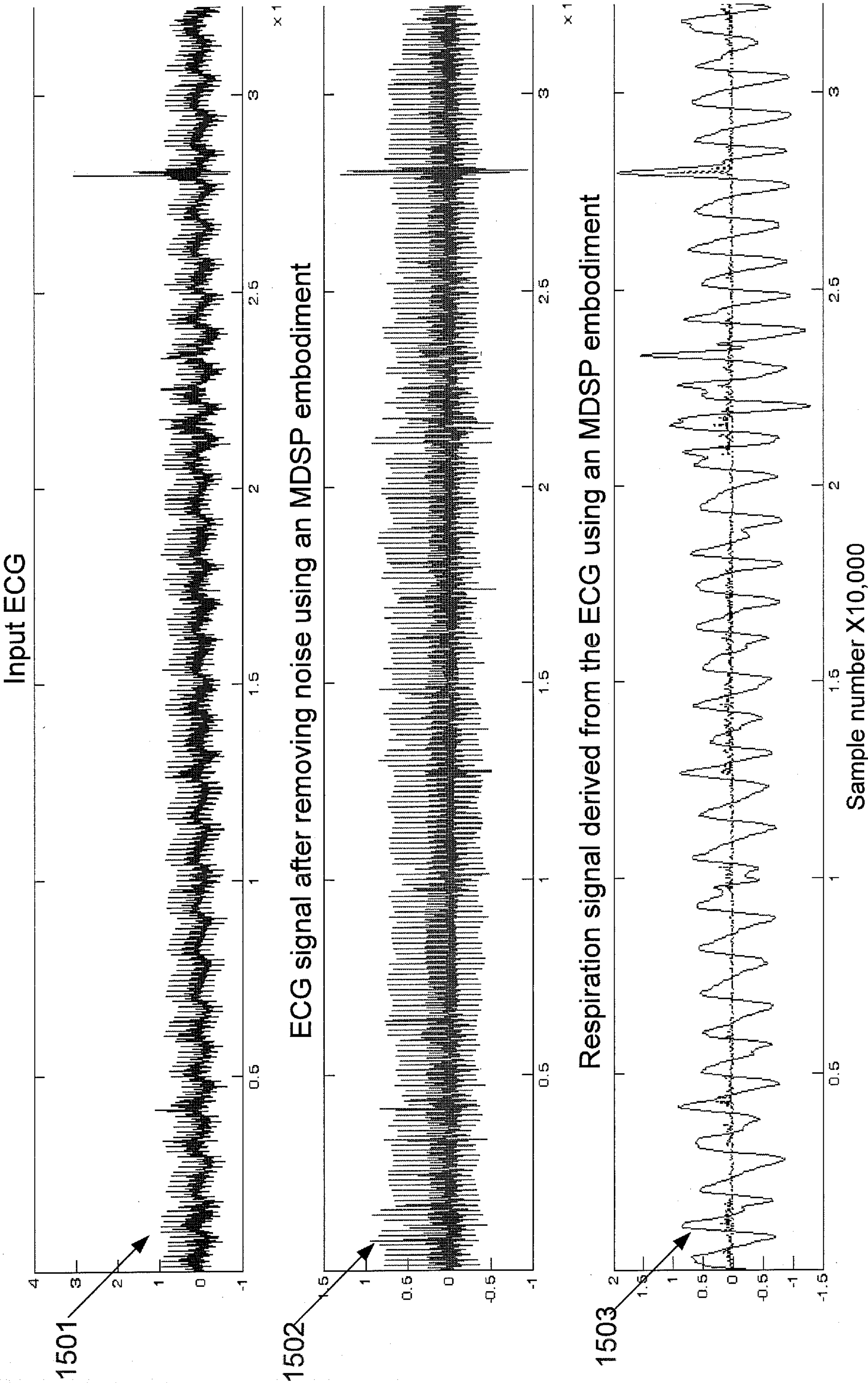


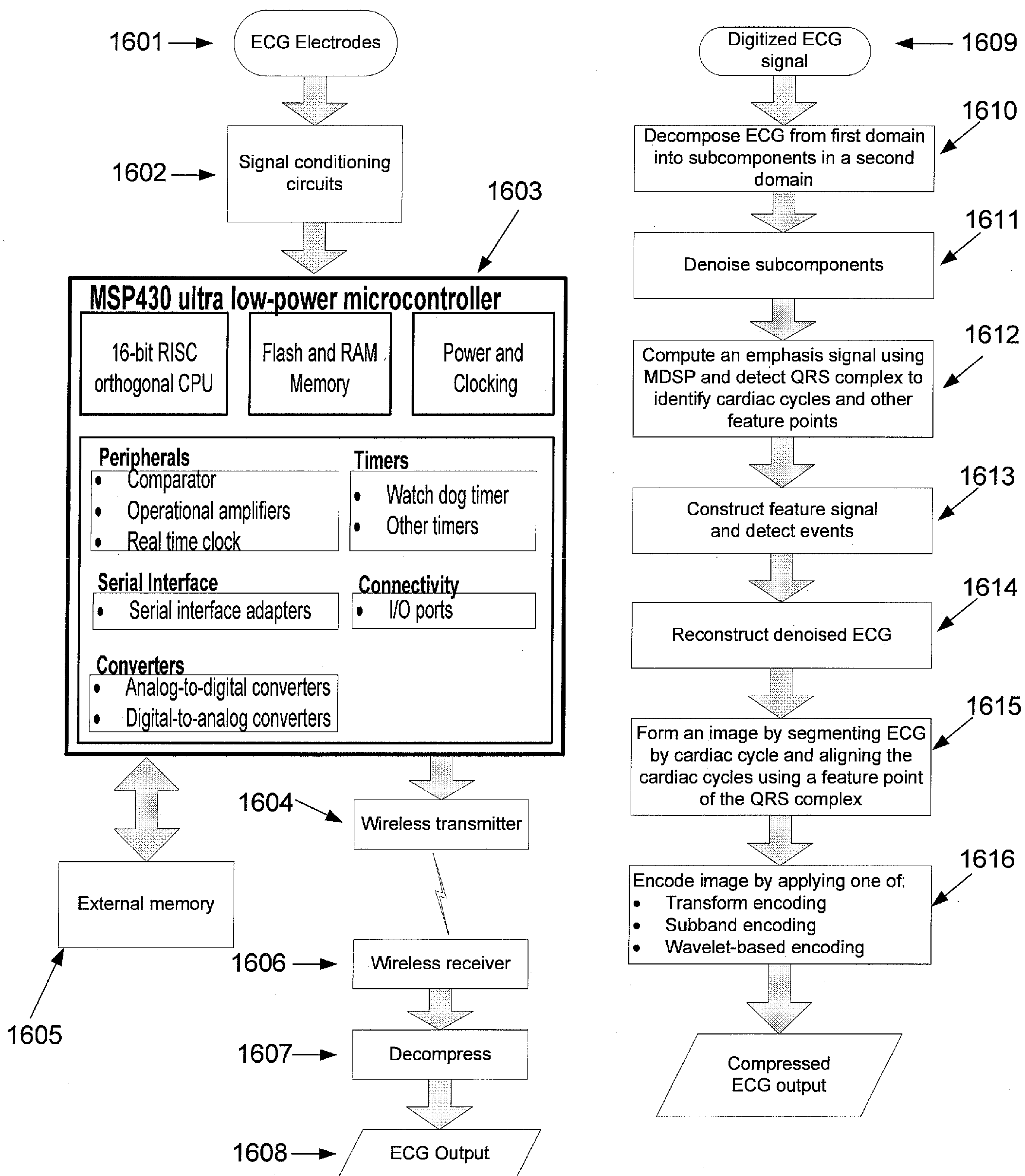
Figure 16

Figure 17

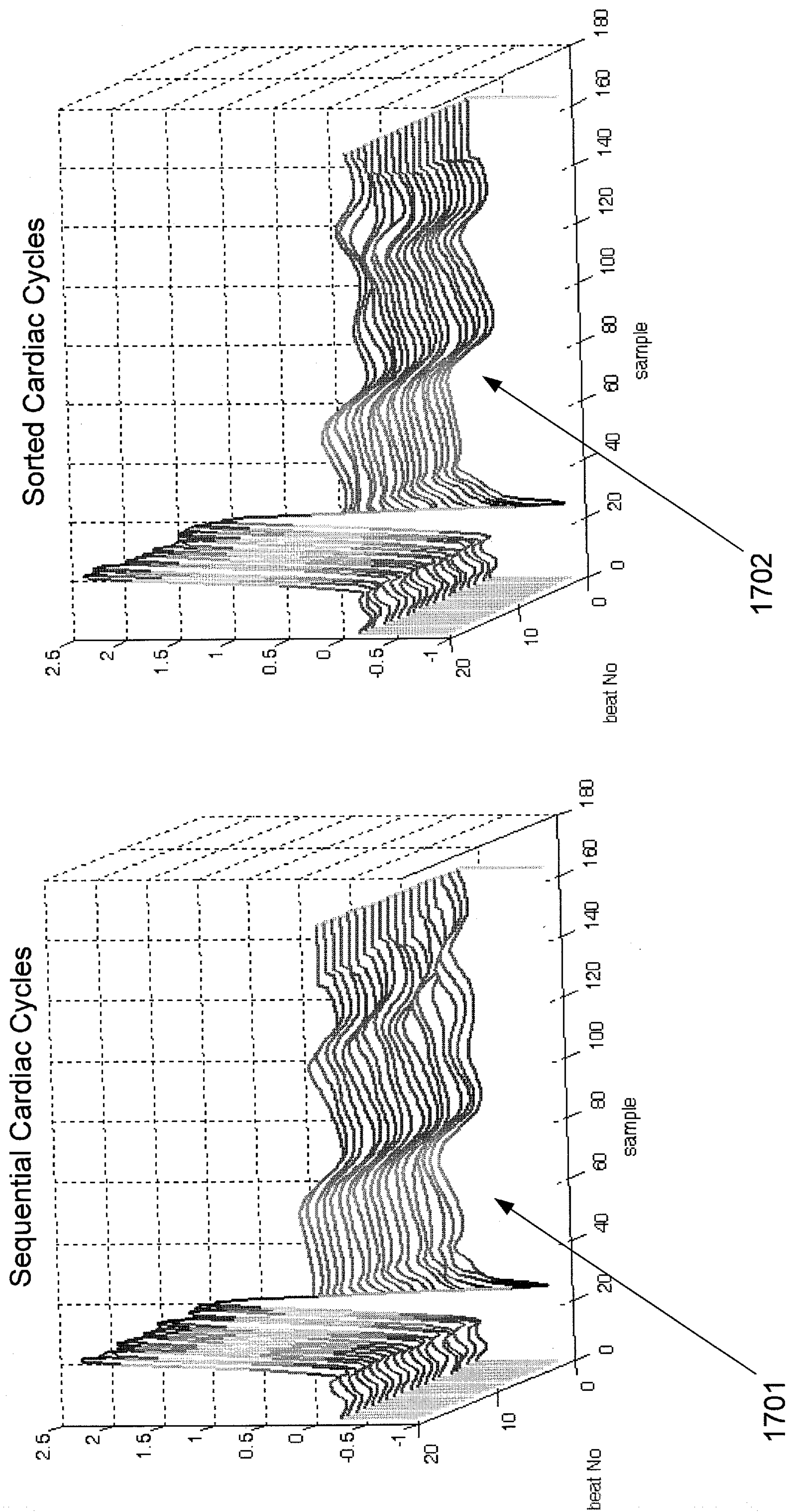


Figure 18

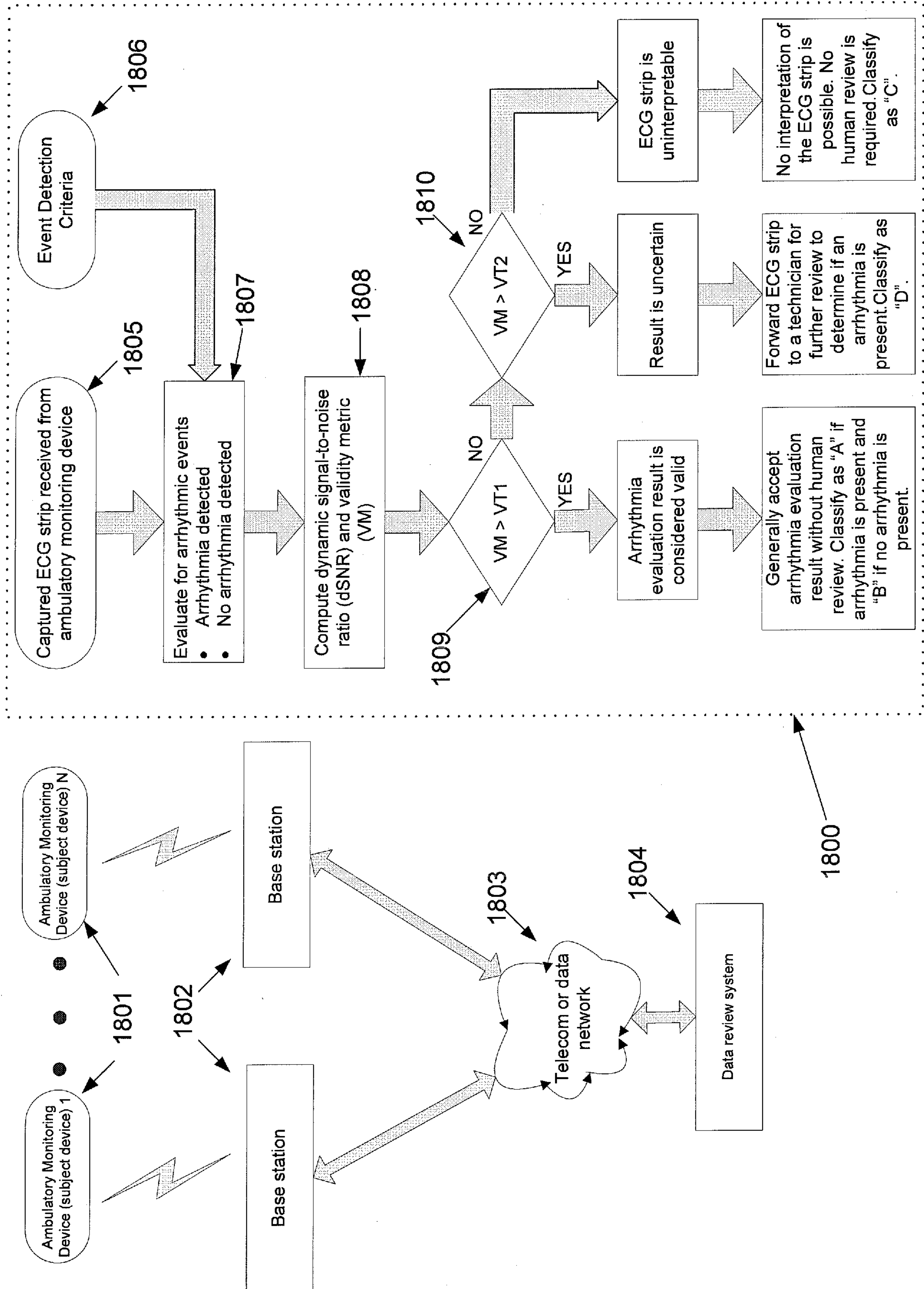


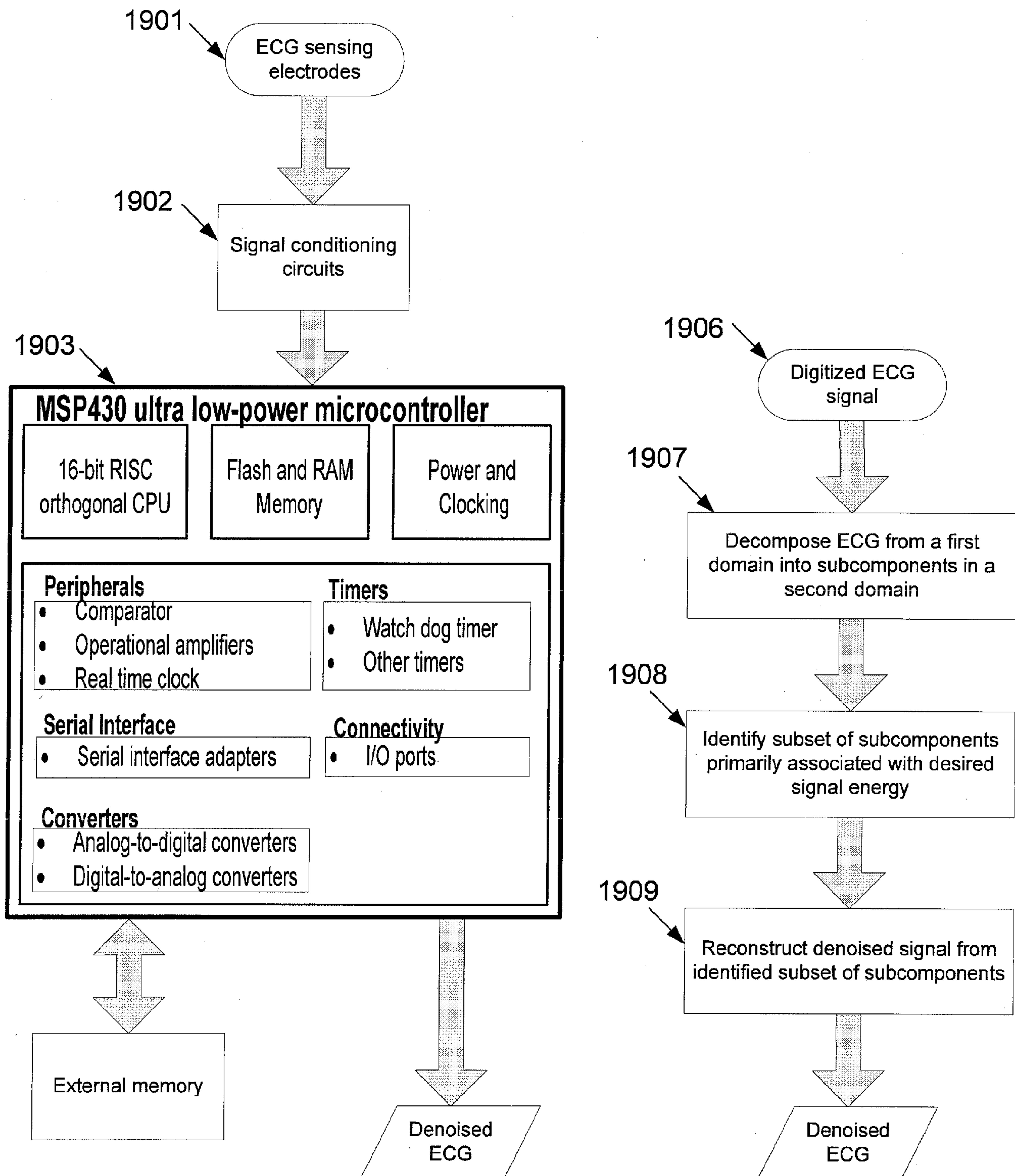
Figure 19

Figure 20

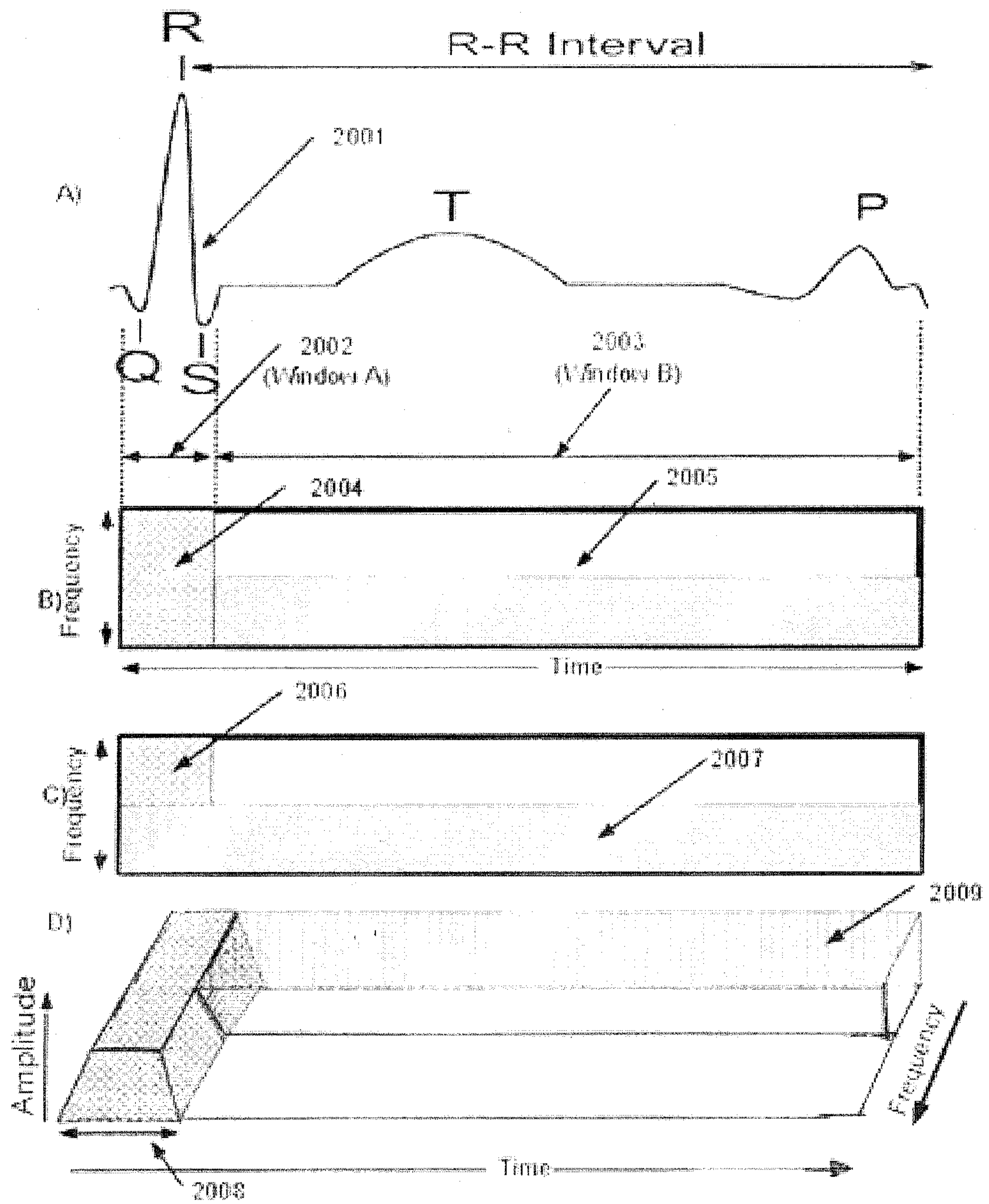
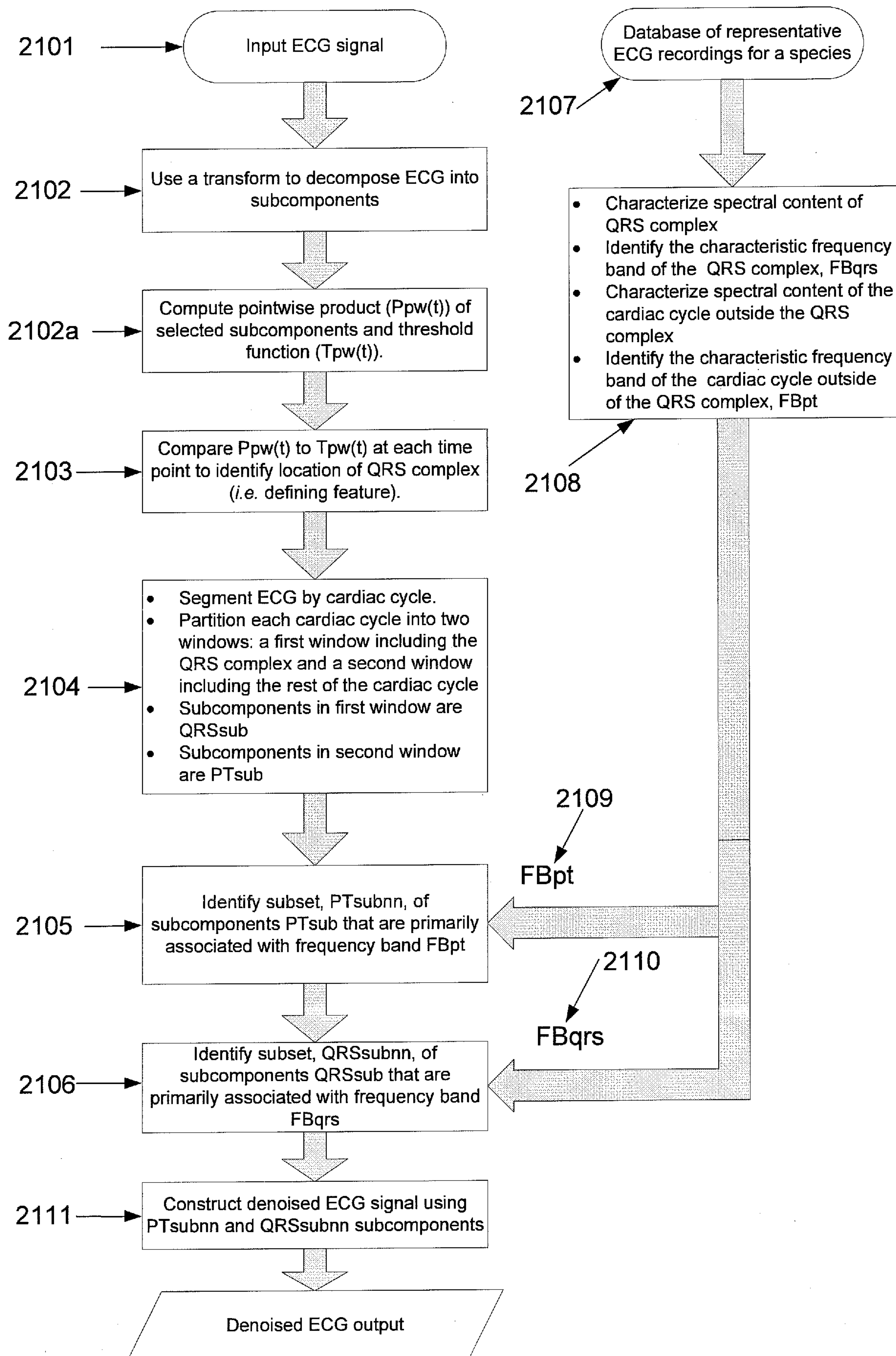


Figure 21

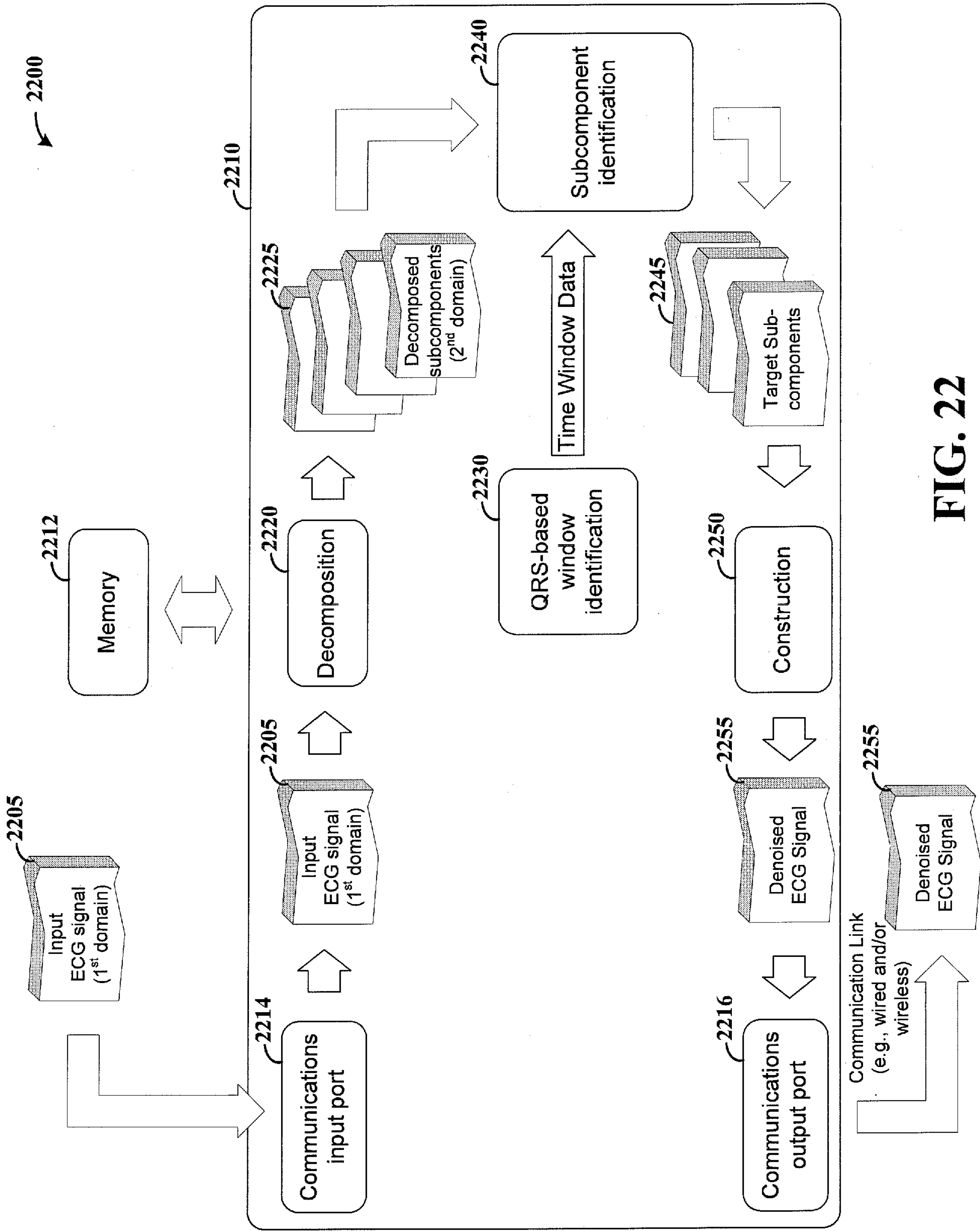


FIG. 22

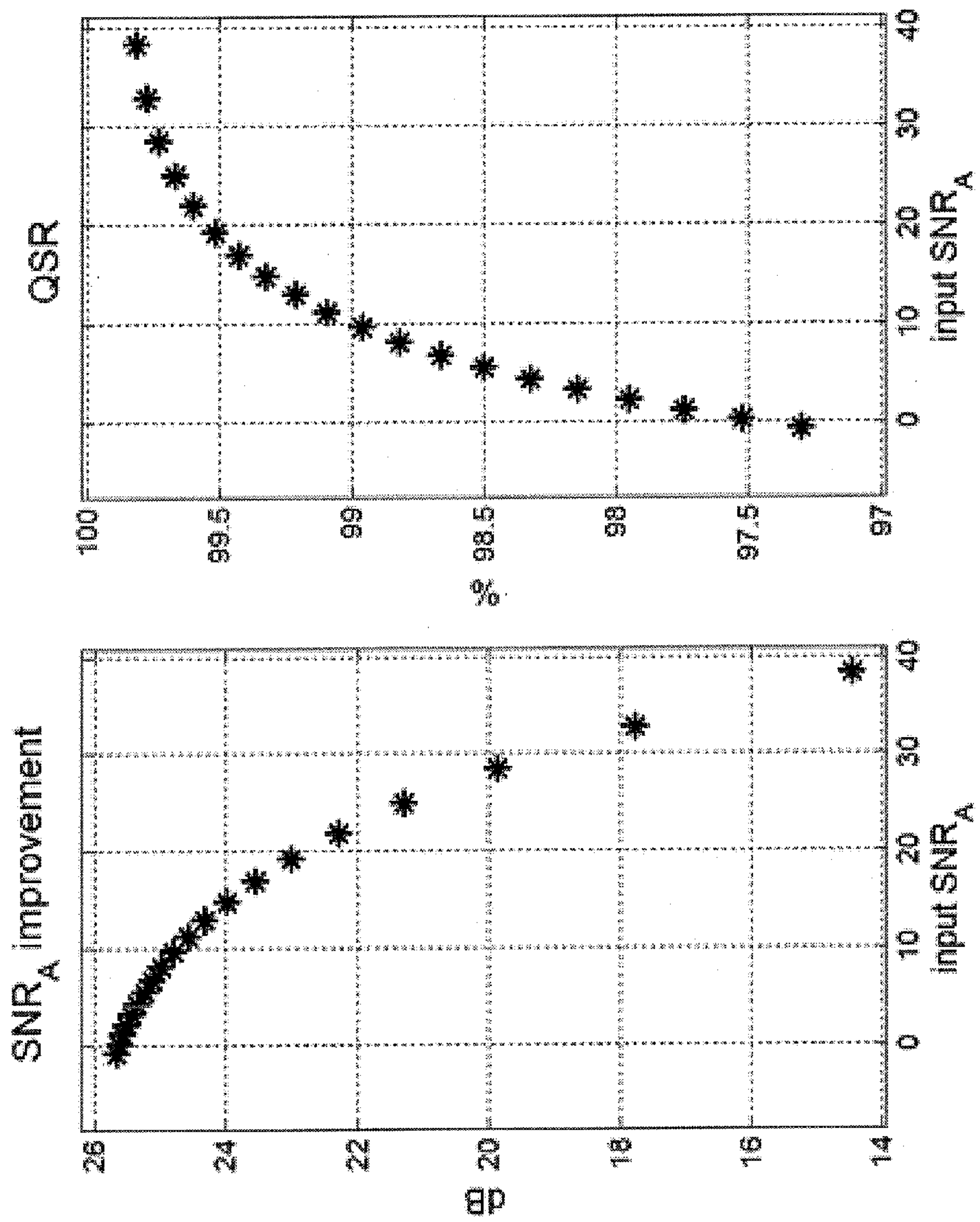
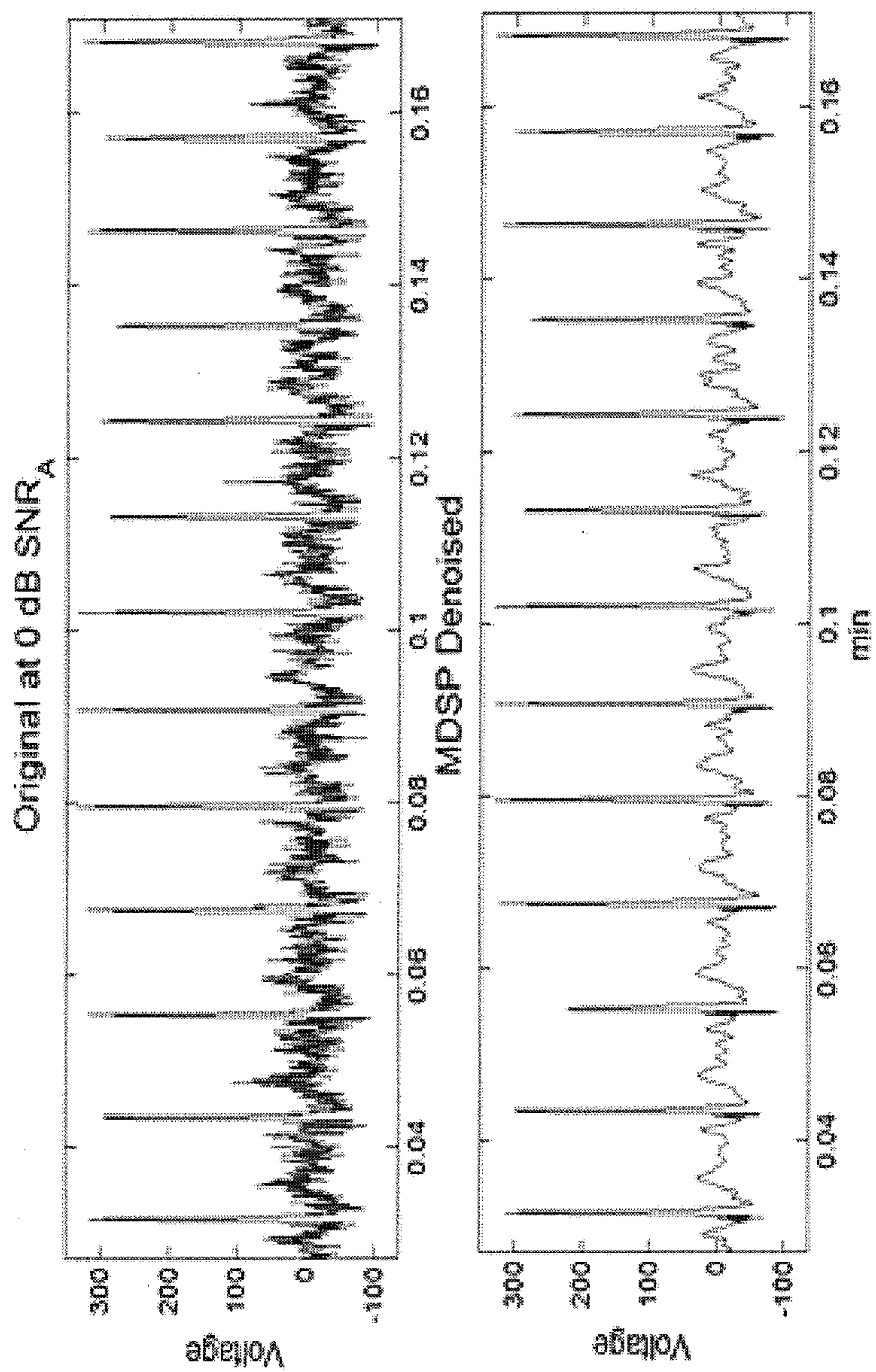


FIG. 23a

**FIG. 23b**

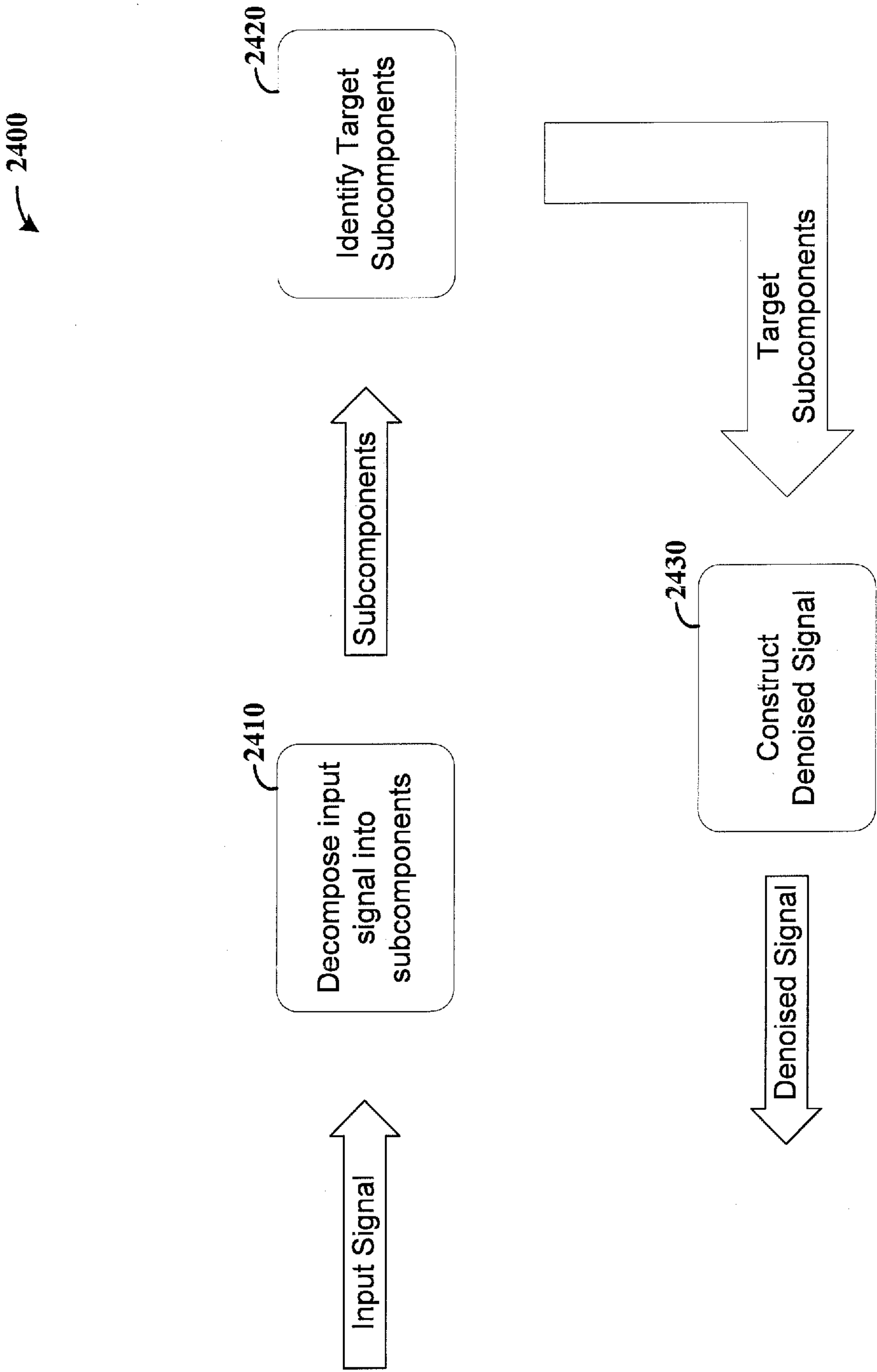


FIG. 24

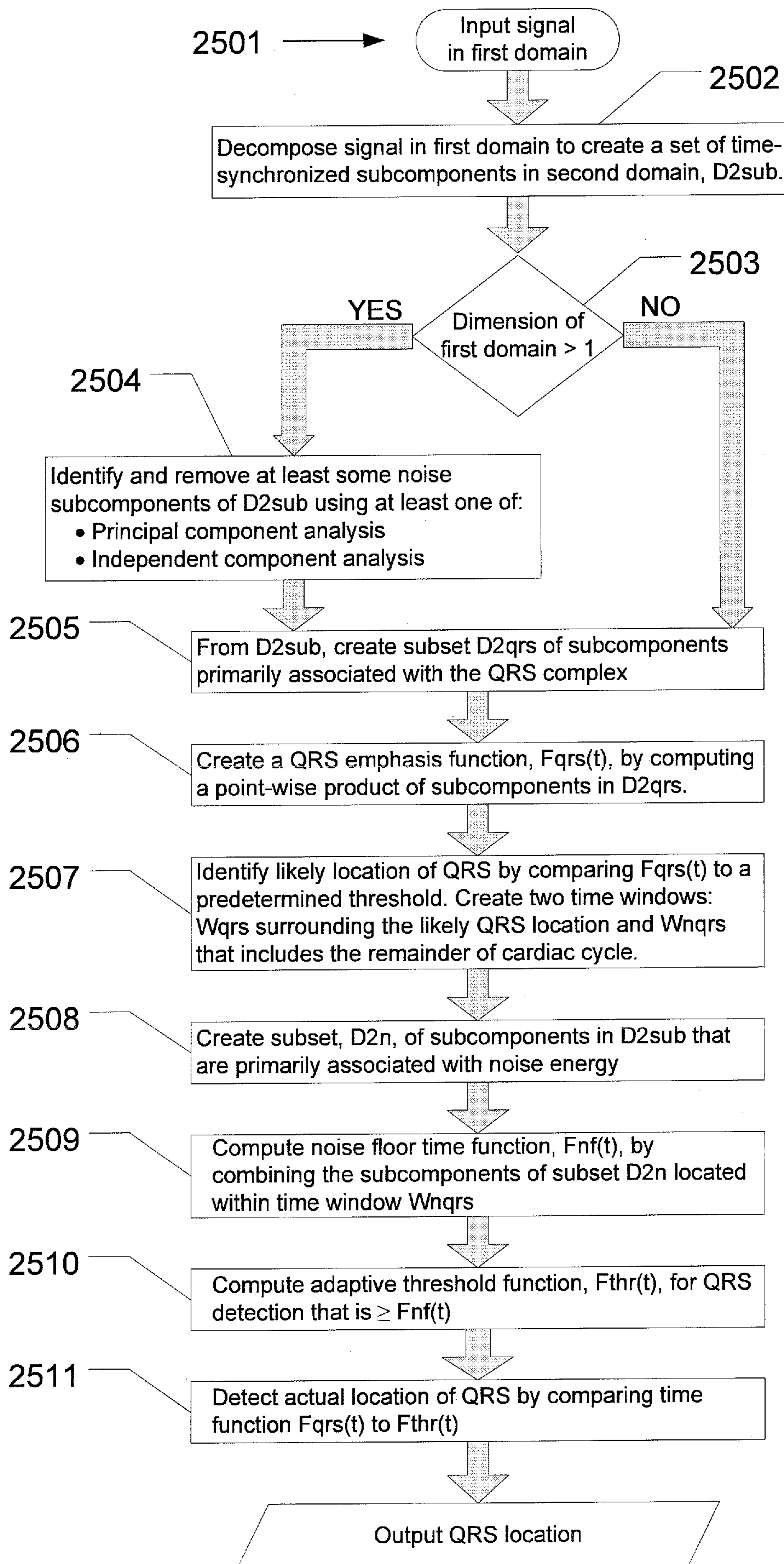
Figure 25

Figure 26

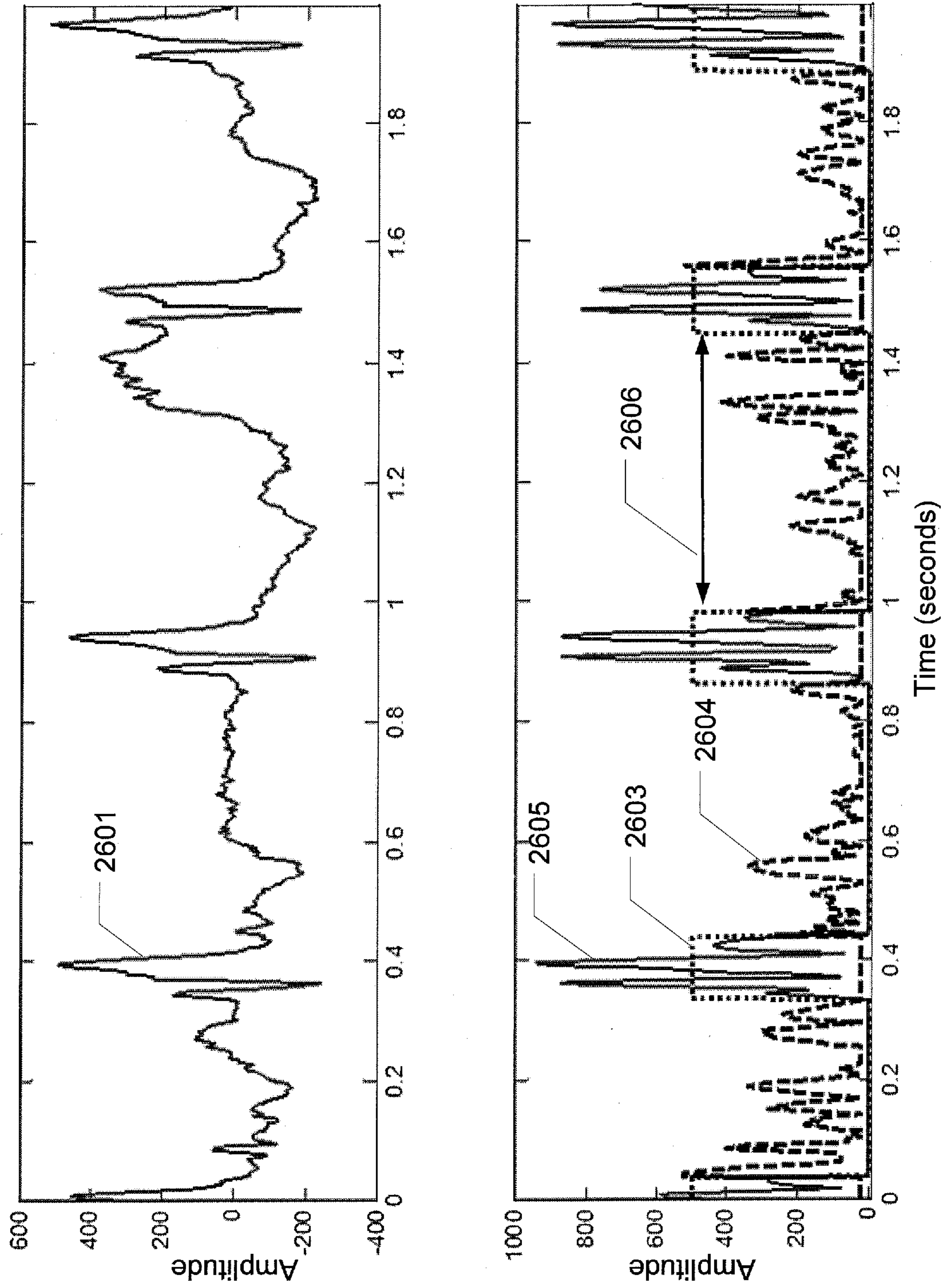


Figure 27

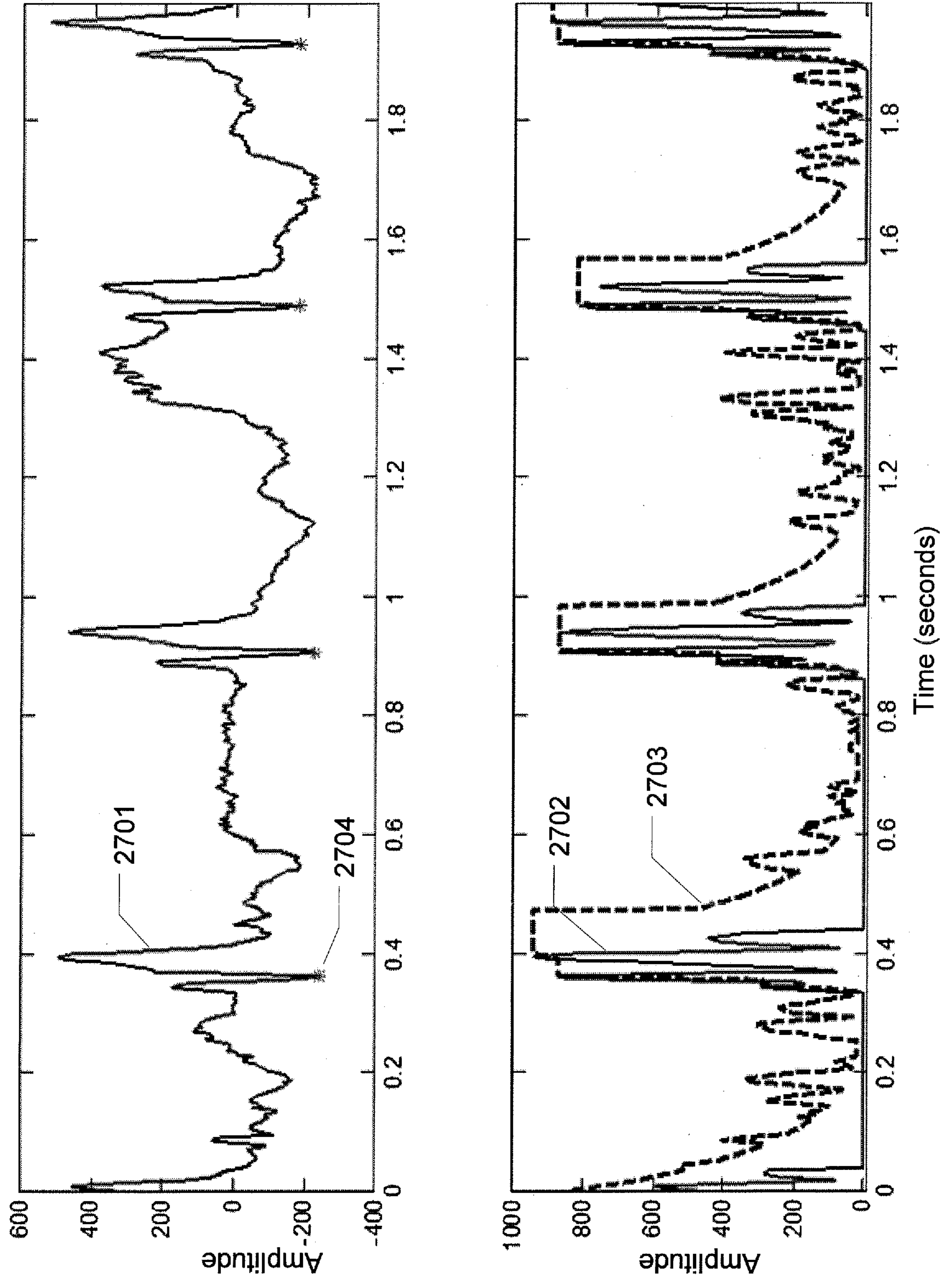
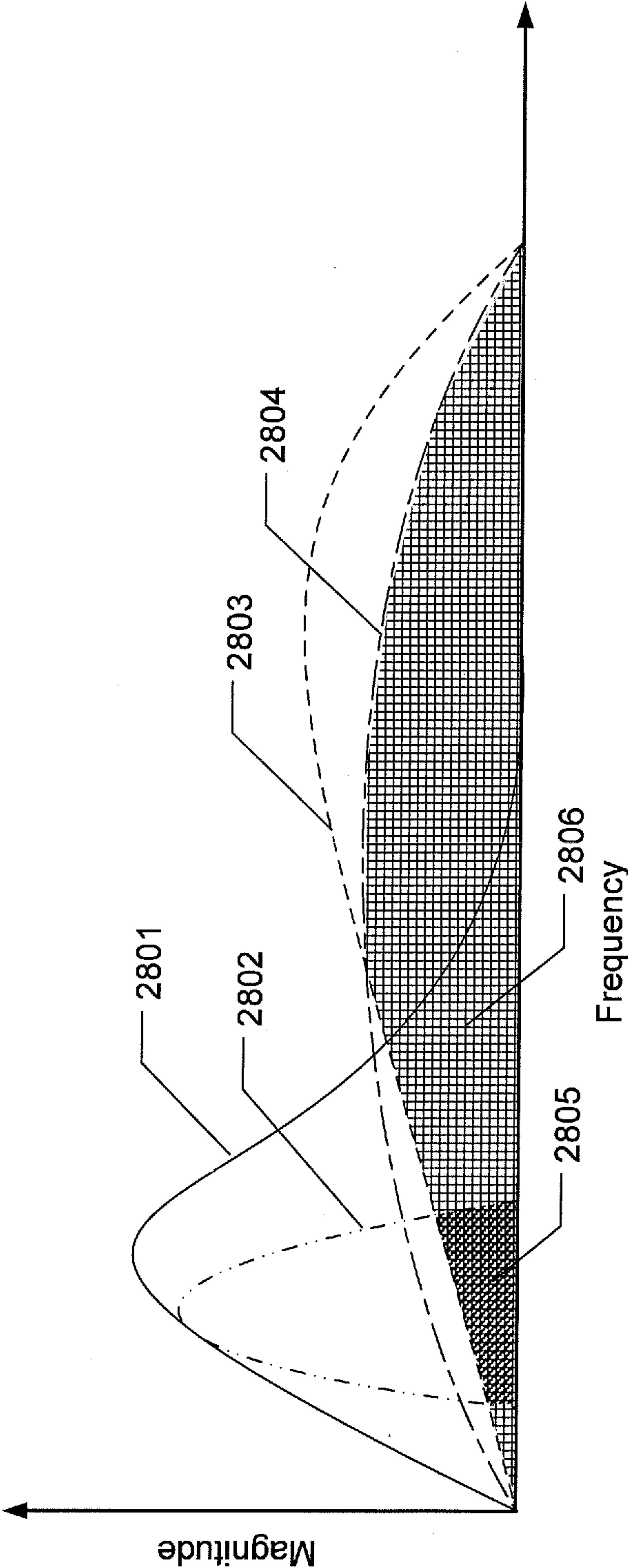
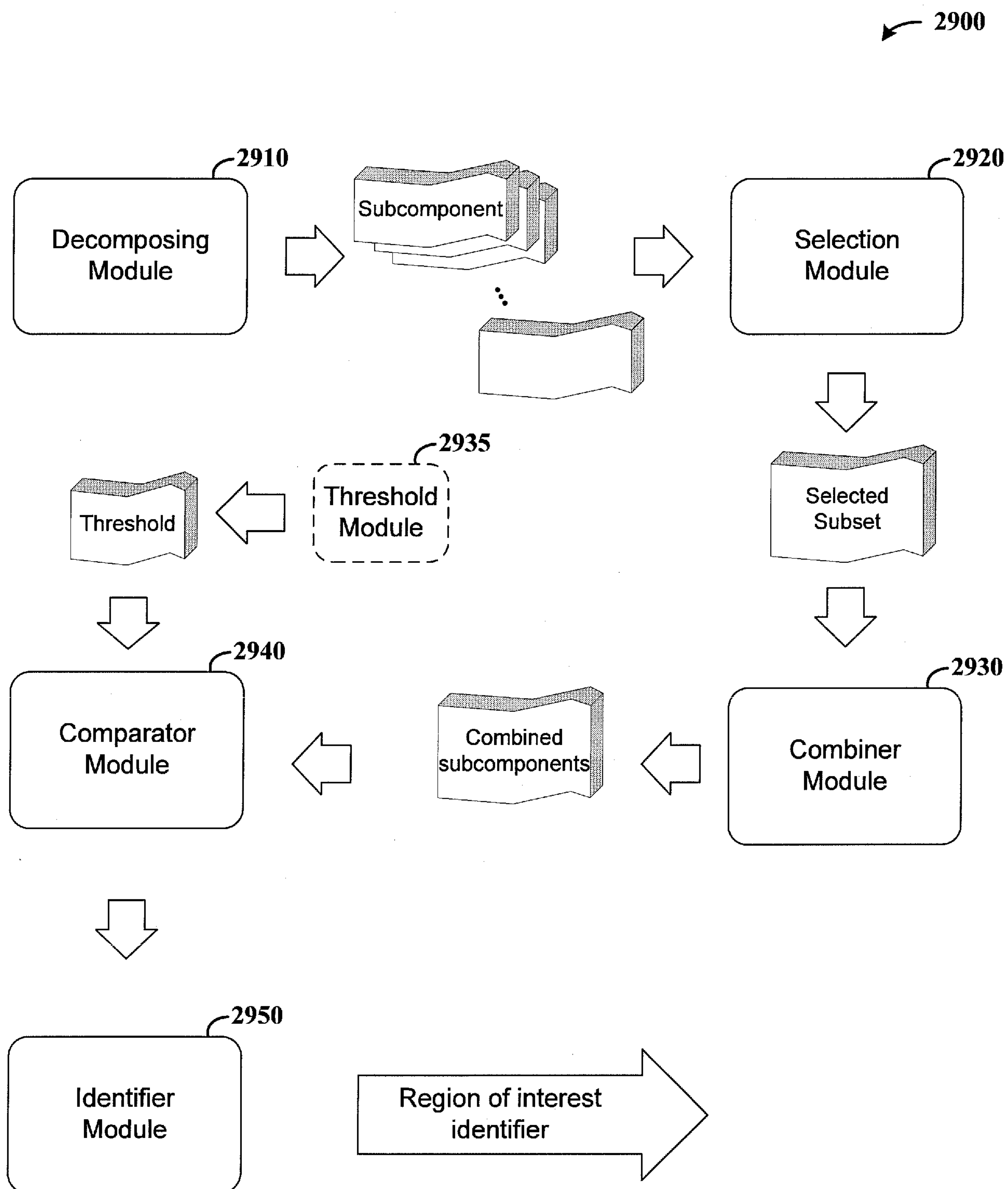


Figure 28



**FIG. 29**

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