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(54) **METHOD AND SYSTEM FOR DETERMINING VENTRICULAR FAR FIELD CONTRIBUTION IN ATRIAL ELECTROGRAMS**

VERFAHREN UND SYSTEM ZUR BESTIMMUNG DES VENTRIKULÄREN FERNFELDBEITRAGS IN EINEM VORHOF-EKG

PROCÉDÉ ET SYSTÈME POUR DÉTERMINER LA CONTRIBUTION DE CHAMP LOINTAIN VENTRICULAIRE DANS DES AURICULOGRAMMES

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- **LENIS GUSTAVO ET AL: "Orthogonal component analysis to remove ventricular far field in non periodic sustained atrial flutter", COMPUTING IN CARDIOLOGY 2013, N/A, 6 September 2015 (2015-09-06), pages 669-672, XP032865238, ISSN: 2325-8861, DOI: 10.1109/CIC.2015.7410999 ISBN: 978-1-4799-0884-4 [retrieved on 2016-02-17]**

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Description

Technical Field

[0001] The present invention generally relates to electronic data processing of clinical measurement data, and more particularly, relates to methods, computer program products and systems for Determining Ventricular Far Field (VFF) contribution in Atrial Electrograms based on clinical sensor data.

Background

[0002] For the diagnosis of a patient's cardiac arrhythmias medical heart specialists record the electrical activity of the patient's heart with catheters having sensors (electrodes) to measure the electrical potential inside the heart. The resulting sensor signal is called an electrogram. Such electrograms as well as the location (position) of the respective sensors during the measurement can be recorded by using so-called electro-anatomical mapping systems (EAMS) which are well known in the art. The analysis of electrograms plays an important role in supporting a medically trained person in the diagnosis of potential heart diseases for finally taking appropriate therapeutic measures.

[0003] The contraction of the heart is triggered by an electrical pulse which is running in the form of a depolarization wave over the heart muscle and generates an electrical field in its environment. The field strength decreases with the distance r. While atrial tissue typically has a thickness of just a few millimeters, ventricular tissue typically has a thickness of more than a centimeter. For this reason, the electrical field generated in the ventricles of the heart is also measurable in the atria where it is called the Ventricular Far Field (VFF). Further, the VFF can also be measured on the body surface of the patient, leading to the well-known electrocardiogram (ECG) potential. Such an approach is described in the European patent application EP3192441 where, after ECG acquisition, ventricular far field detection is performed by a spatial averaging method, a temporal averaging method, a system identification method or principal component analysis.

[0004] Measurement technology requires to measure the electrical potential in relation to a reference potential. Inside the human body there is no absolute and constant reference potential available. Therefore, in clinical measurements, an artificial reference potential is generated as known by a skilled person. For example, this can be achieved by averaging the electrical potentials of the patient's right arm, left arm and left leg, thus providing the reference signal Wilson Central Terminal (WCT). The electrodes (sensors) in the inner of the heart measure the potential in relation to this virtual "zero" potential as a so-called unipolar electrogram (cf. Tedrow, Stevenson, Recording and Interpreting Unipolar Electrograms to Guide Catheter Ablation, Heart Rhythm, ISSN 1556-3871, Vol. 8, Issue 5).

[0005] In unipolar electrograms primarily the information about the excitation of heart areas in close proximity to the catheter is useful for diagnostic purpose. For example, for diagnosis of atrial flutter the propagation of the electrical excitation in the patient's atria is analyzed. The respective signals reflect the atrial activity in respective electrograms. However, such electrograms are always superimposed by noise signals which make the diagnostic analysis of the data more difficult. The European patent application EP3033993 discloses such a system where far field reduction is carried out in a cardiac electrogram by extracting unipolar beats of an intracardiac electrogram that occur within a predetermined time interval that includes QRS peaks.

[0006] Prior art approaches typically subtract the signals of two neighboring electrodes to eliminate disturbing noise. However, the resulting bipolar electrogram cannot be accurately assigned to a physical location in the patient's heart anymore resulting in a spatial inaccuracy. Further, it is not possible to precisely derive the local atrial excitation from the morphology from the bipolar electrograms because its shape and amplitude significantly depend on the angle between the measurement electrode and the direction of the excitation wave. Therefore, diagnostic indicators depending on shape or amplitude can produce misleading results, and thus the original unipolar electrograms are more suitable for diagnostic purposes.

[0007] Other prior art approaches observe recorded electrograms over longer time intervals (> 5 seconds) to learn the VFF component and then try to separate the VFF components from the atrial activity components by using statistical methods. Multiple statistical methods (e.g., Principal Component Analysis (PCA), Template Matching and Subtraction (TMS) or Periodic Component Analysis (PiCA)) may be used. For example, the paper "Orthogonal Component Analysis to Remove Ventricular Far Field in Non Periodic Sustained Atrial Flutter" by Gustavo Lenis et al. published in 2015 Computing in Cardiology Conference (CinC) suggests the use of Orthogonal Component Analysis to learn how the clean, not corrupted AA are distributed in the first components of a PCA and to extrapolate that knowledge to reconstruct the original AA from the corrupted EGM. However, the temporal coupling between ventricles and atria is critical with regards to the selection of the appropriate method. If this is not recognized in a correct manner, the method is not working. Moreover, the catheter has to be stable at each measuring location during the learning time interval. As modern medical devices can collect approximately 10000 data points in high density mode (corresponding to several hundred locations of the catheter inside the heart) the measuring procedure becomes very long and increases the health risk for the patient. Further, the required learning phase does not allow to apply such prior art approaches in near-real-time. Some modern catheters, like the one disclosed in US patent application 20170100049, combine ultrasonic sensors with electrode sensors to determine the density of dipoles in the cardiac wall. However, they fail to determine and eliminate the disturbing influence of the Ventricular Far Field.

Summary

[0008] Therefore, there is a need to provide an improved method for determining the VFF component in electrograms with improved accuracy without additional risks for the patient. This technical problem is solved by the features of a computer system, a computer-implemented method and a computer program product as disclosed in the independent claims. Once the VFF component is determined it can be removed from the electrograms which allows more accurate interpretation of the sensor data leading to improved diagnostics.

[0009] The invention is defined in the independent claims. In one embodiment, a computer system is provided for determining Ventricular Far Field contribution in Atrial Electrograms of a patient. The computer system includes an interface module to receive a plurality of electrical signals generated by a plurality of sensors. The sensors are the electrodes of a catheter as known in the art. The plurality of electrical signals relate to a plurality of locations in an atrium of the patient. In other words, the electrodes of the catheter record the signal at different locations within the atrium while the catheter is moved by a medically trained person. The respective signal data is finally received by the interface module.

[0010] The system further includes a reference module to determine a reference signal reflecting electrical excitation of the patient's ventricles. For example, in one embodiment, receiving the reference signal may originate from one or more electrocardiogram sensors measuring at least the R wave of the patient's ventricular electrical activity. In an alternative embodiment, the reference signal may be received from a coronary sinus catheter sensor. In yet another alternative embodiment the reference signal may be computed by blind source separation or analysis of periodicity from the electrical signal data recorded by the plurality of sensors (electrodes). In yet another alternative embodiment, the reference signal may be determined from information about the ventricular contraction obtained by using optical techniques like laser interferometry, pulse oximetry or near-infrared spectroscopy.

[0011] Further, the computer system includes a data processing module configured to select from the plurality of the received electrical signals such electrical signals which are recorded under one or more of the following conditions by verifying each of the three conditions:

- the respective signals are recorded at locations inside the atrium where the respective sensor has no contact to the atrial tissue,
- the respective signals are recorded, irrespective of the sensor location, during time intervals where the respective (adjacent) part of the atrium shows no electrical activity,
- the respective signals are recorded, irrespective of the sensor location, during time intervals which comprise a plurality of heart beat intervals and are subject to subsequent spatial smoothing.

[0012] In other words, signals which are suitable for determining the VFF contribution in the atrium are such signals that were either recorded while the catheter electrodes were entirely surrounded by blood (no atrial tissue contact) or, even in case of atrial tissue contact, were recorded when the atrium showed no activity. Of course, also signals recorded while no atrial tissue contact occurred and no atrial activity occurred are suitable signals for the further analysis. In one embodiment, sensor signals may be recorded at a particular region over a longer period which includes multiple heart beat intervals. In this case, subsequent spatial smoothing of such signals can be used to eliminate the disturbing influence of atrial activity even if the signals were recorded at locations where the sensor had contact to the atrial tissue or was located close enough to the atrial tissue so that atrial activity still had an impact of the sensor signals.

[0013] The data processing module further determines a spatio-temporal distribution of the VFF inside the atrium by approximating the spatio-temporal distribution based on signal data of the selected signals by using an approximation model. Different approximation models can be used. For example, a linear spatial model, a non-linear spatial model, a temporal model, or a look-up table may be used. As described in more detail in the detailed description, the approximation model is computed on the basis of the selected signal data and allows for interpolation and extrapolation of the Ventricular Far Field contribution at locations where no sensor data has been recorded. Approximating can be implemented in different ways. In one embodiment, polynomial approximation may be used. In an alternative embodiment, approximation with a dipole source model may be used. In yet another alternative embodiment, the approximation is performed with spatio-temporal linear combination of the recorded signal data. In yet another alternative embodiment, the approximation is performed with Radial Basis Functions. In yet another alternative embodiment, the approximation is performed with using Look-Up Tables, which contain the recorded signal data of the VFF component at the respective measurement position. In yet another alternative embodiment, the approximation is performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table. In yet another alternative embodiment, spatio-temporal smoothing may be applied to the signal data to smoothen the recorded data. A person skilled in the art may use other approximation methods which are appropriate in the described context.

[0014] The spatio-temporal distribution represents the VFF contribution at each respective measuring location of the sensors recording the sensor data for the respective electrograms. In other words, for each electrogram which represents the measurement signal data at a particular location at a particular point in time within the patient's atrium the contribution of the VFF at this location can now be derived from the spatio-temporal distribution. The alternative approximation methods allow to determine the VFF contributions at the respective locations in near-real-time with high accuracy.

[0015] A near-real-time system response, as used herein, means that a computation for approximating the spatio-temporal distribution in response to the received sensor data is only delayed by the time delay introduced, by automated data processing or network transmission, between the occurrence of an event (e.g., receipt of measurement data) and the use of the processed data (e.g., use of the processed data in a diagnostic activity by the medically trained person.) For example, a near-real-time display depicts an event or situation as it existed at the current time minus the processing time, as nearly the time of the live event. (Near-) real-time methods are not part of the claimed subject-matter.

[0016] With the above method near-real-time evaluation of the sensor data provided by the electrode sensors of the catheter can be performed to determine the VFF contribution in an area as defined through the sensor locations of the respective sensors. As a consequence, for each heartbeat interval (ventricular electrical excitation period) the VFF contribution can be derived for a region within the atrium which is reached by the various sensors while the catheter moves through the atrium. Example standard catheters have 64 electrode sensors covering 64 measurement locations in parallel.

[0017] It may be desirable to generate a map of the VFF contribution for the entire atrium. However, it is not possible to scan the entire inner of the atrium within a single heartbeat interval. For this purpose, in one embodiment, the system may receive signals which include signal data recorded over multiple ventricular electrical excitation periods. For example, such data may be provided from an external storage device which records the sensor data while the catheter is being moved through the patient's atrium. It is also possible, that the computer system itself includes such a storage device and records the data being received through the interface component. In this embodiment, the data processing module is further configured to synchronize the selected signal data with the measured electrical excitation events using the reference module. The determination of the spatio-temporal distribution of the Ventricular Far Field is then based on the synchronized signal data. Synchronizing the selected signal data with the measured electrical excitation events, as used herein, means that the electrical excitation event of each heartbeat interval defines the time reference for the signal of the respective ventricular electrical excitation period. That is, all recorded signal data can be projected to the same virtual heartbeat interval referenced to the respective measured electrical excitation event. The approximation method can then be applied to all signal data in the virtual heart beat interval which allows to estimate the spatio-temporal distribution by taking into account all sensor locations which are associated with sensor data of the selected signals, even when being recorded during different ventricular electrical excitation periods.

[0018] In one embodiment, the data processing module further has a model generator to generate a plurality of models wherein each model relates to a particular section within the atrium in combination with a particular time point in relation to the electrical excitation event and reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point. For example, the particular sections within the atrium can be formed by splitting the volume of the atrium by planes (layers) which are in parallel to the plane forming the boundary between atria and ventricles. The split can be done by using 2 or more planes which can also have another orientation in space. Also more complex forms of spatial models can be applied. Using an appropriate spatial separation into models, the accuracy of the individual distribution mode can be improved and avoids so called over-fitting. Some models (e.g., polynomic models) may not be able to model the entire atrium correctly in a single model. However, in many cases it is possible to correctly model a part of the atrium (e.g., the upper / lower half). In such cases, such sub-models can be trained and used separately.

[0019] In one embodiment, the VFF contribution measurement system is part of a diagnosis support system which preprocesses recorded electrogram data before providing the data for further medical diagnostics to a medically trained person or to a computer system trained for atrial disease diagnosis support. In this embodiment, the data processing module further has an Atrial Electrogram improvement module (AEIM) to subtract the determined contribution of the Ventricular Far Field at a particular location inside the atrium from the Atrial Electrogram represented by an electrical signal generated by one of the sensors at the particular location. In other words, for each sensor location where sensor data have been recorded and where the VFF contribution can be determined with the herein disclosed method, the AEIM can eliminate the influence of the VFF contribution in the electrogram. As a result, atrial disease caused signals showing electrical activity of the atrium become much better "visible" in the corrected unipolar electrogram especially where such atrial disease cause signals overlap with the disturbing VFF contribution (in the neighborhood of the R wave). Typically, such atrial electrical activity shows an active interval when electrical activity occurs, and a resting interval when no electrical activity occurs. Electrical activity hereby is a consequence of the depolarization of cardiac tissue close to the sensor. No electrical activity occurs when neighboring tissue is in rest and does not depolarize. Thus the electrical activity reflects changes of the transmembrane voltage of adjacent myocardial cells. For example, diagnostic analysis of atrial flutter types can be significantly facilitated by the disclosed signal correction.

[0020] In one embodiment, a computer-implemented method is provided for determining Ventricular Far Field contribution in Atrial Electrograms of a patient. The computer-implemented method can be executed by the modules of the computer system disclosed herein. A computer program with computer readable instructions implementing said modules can be loaded into a memory of the computer system and can be executed by one or more processors of the computer system to cause the computer system to perform the computer-implemented method. In other words, the computer program implements the functions of the respective modules which, in operation, run the computer-implemented method. The method includes: receiving a plurality of electrical signals measured by a plurality of sensors wherein the plurality of electrical signals relate to a plurality of locations in an atrium of the patient; determining a reference signal measuring the electrical excitation of the patient's ventricles; selecting from the plurality of the received electrical signals such electrical signals which are recorded under one or more of the following conditions: conditions by verifying each of the three conditions:

- the respective signals are recorded at locations inside the atrium where the respective sensor has no contact to the atrial tissue,
- the respective signals are recorded, irrespective of the sensor location, during time intervals where the respective part of the atrium shows no electrical activity,
- - the respective signals are recorded, irrespective of the sensor location, during time intervals which comprise a plurality of heart beat intervals and are subject to subsequent spatial smoothing; and

determining a spatio-temporal distribution of the Ventricular Far Field inside the atrium by approximating the spatio-temporal distribution based on signal data of the selected signals by using an approximation model wherein the approximation model is computed on the basis of the selected signal data and allows for interpolation and extrapolation of the Ventricular Far Field contribution at locations where no sensor data has been recorded.

[0021] For example, approximating may use any one of the following methods: polynomial approximation, approximation with a dipole source model, approximation with a spatio-temporal linear combination of the recorded signal data, approximation performed with Radial Basis Functions, approximation using Look-Up Tables, and approximation performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table.

[0022] For example, determining a reference signal may be performed by any one of the following methods:

- receiving the reference signal from one or more electrocardiogram sensors measuring at least the R wave of the patient's ventricular electrical activity,

- receiving the reference signal from a coronary sinus catheter sensor, computing the reference signal from the recorded electrical signal data by blind source separation or analysis of periodicity, and

- determining information about the ventricular contraction by using optical techniques like laser interferometry, pulse oximetry or near-infrared spectroscopy.

[0023] In one embodiment, where the plurality of received signals comprises signal data recorded over multiple ventricular electrical excitation periods, the method further includes: synchronizing the selected signal data with the measured electrical excitation events, and determining the spatio-temporal distribution is based on the synchronized signal data.

[0024] In one embodiment, determining the spatio-temporal distribution of the Ventricular Far Field further includes: generating a plurality of models wherein each model relates to a particular section within the atrium in combination with a particular time point in relation to the electrical excitation event and reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point.

[0025] In one embodiment, the method further includes: subtracting the contribution of the Ventricular Far Field at a particular location inside the atrium from the Atrial Electrogram represented by electrical signal data measured by a particular sensor at the particular location.

[0026] Further aspects of the invention will be realized and attained by means of the elements and combinations particularly depicted in the appended claims. It is to be understood that both, the foregoing general description and the following detailed description are exemplary and explanatory only and are not restrictive of the invention as described.

Brief Description of the Drawings

[0027]

FIG. 1 is a simplified block diagram illustrating an embodiment of a computer system for determining Ventricular Far Field contribution in Atrial Electrograms of a patient;

FIG. 2 is a simplified flowchart of a computer-implemented method determining Ventricular Far Field contribution in Atrial Electrograms of a patient according to an embodiment of the invention;

FIG. 3A is a schematic illustration of signal contributions in a unipolar atrial electrogram before and after correction;

FIG. 3B shows a simulation of signal contributions in a unipolar electrogram;

FIG. 4 shows a heart catheter with sensor electrodes;

FIG. 5 illustrates an atrium with a catheter inside the atrium and some respective sensor electrodes;

FIGs. 6A, 6B show traces of sensor electrodes of a heart catheter being moved through an atrium;

FIG. 7 shows an example of synchronized selected signals as captured by an embodiment of the computer system;

FIG. 8 illustrates a spatio-temporal distribution of the Ventricular Far Field contribution at a particular point in time in an atrium;

FIG. 9A shows an example of recorded reference signals and selected electrogram signals;

FIG. 9B shows an example of the corrected electrograms according to an embodiment;

FIG. 10 illustrates different performances of different approximation methods for determining the VFF contribution;

FIG. 11 illustrates an example of a temporal approximation model with four basis vectors; and

FIG. 12 is a diagram that shows an example of a generic computer device and a generic mobile computer device which may be used with the techniques described herein.

Detailed Description

[0028] FIG. 1 is a simplified block diagram illustrating an embodiment of a computer system 100 for determining Ventricular Far Field contribution in Atrial Electrograms of a patient. FIG. 2 is a simplified flowchart of a computer-implemented method 1000 for determining Ventricular Far Field contribution in Atrial Electrograms according to an embodiment of the invention. The functions of system 100 of FIG. 1 are discussed in the context of the method steps of method 1000 which are performed by the respective system components (modules) of system 100. Therefore, the following description refers to reference numbers of the FIGs. 1 and 2.

[0029] The system 100 includes an interface component 110 configured to receive 1100 data 240 from one or more external data sources 200. The external data sources may be sensors S1 to Sn providing real time data about the electric activation of a patient's atrium or it may be a data storage device 210 which provides historic (previously recorded or simulated) sensor data about the electric activation of the patient's atria. In the example of FIG. 1, the interface module 110 receives 1100 the electrical signals Fn (F2, F3, F4) measured by respective sensors (e.g., S2 to Sn). Thereby, the measured (recorded) electrical signals F2 to F4 relate to a plurality of locations in the atrium of the patient. In other words, the sensor data F2 is recorded at a particular location which is different from the recording locations of sensor data F3 and F4. Typically, sensors S1 to Sn are electrodes of a multi-polar mapping catheter 201 as illustrated in FIG. 4 (cf. electrodes S1 to S10). Such catheters are which are widely used for electro-anatomical mapping systems EAMS. Examples of such sensors are described in detail in the references: "A multi-purpose spiral high-density mapping catheter: initial clinical experience in complex atrial arrhythmias; (by Jones, D. G.; McCready, J. W.; Kaba, R. A.; Ahsan, S. Y.; Lyne, J. C.; Wang, J.; Segal, O. R.; Markides, V.; Lambiase, P. D.; Wong, T.; Chow, A. W. C.; 2011; Journal of Interventional Cardiac Electrophysiology: an International Journal of Arrhythmias and Pacing; 31: 225-235)", and "Rapid high resolution electroanatomical mapping: evaluation of a new system in a canine atrial linear lesion model; (by Nakagawa, H.; Ikeda, A.; Sharma, T.; Lazzara, R.; Jackman, W. M.; 2012; Circulation. Arrhythmia and Electrophysiology; 5; 417-424)".

[0030] Turning back to FIG. 1, the computer system 100 further has a reference module 120 to determine 1200 a reference signal RS (e.g. I, II, CS) reflecting electrical excitation of the patient's ventricles. In the example of FIG. 1, reference signals I, II are reference signals received from one or more electrocardiogram sensors ECG measuring at least the R wave of the patient's ventricular electrical activity. The R wave reflects the electrical excitation event in the patient's ventricles which typically triggers the contraction of the heart. Alternative embodiments may use other reference signals RS if appropriate. For example, the reference signal may be received from a coronary sinus catheter sensor CS. In another alternative embodiment, the reference signal RS may be encoded in the recorded electrograms (e.g., F2 to F4) and can be computed from the recorded electrical signal data by blind source separation or analysis of periodicity. In yet another alternative embodiment, the reference signal may be determined based on data obtained by respective sensors (not shown) about the ventricular contraction obtained by using optical techniques like laser interferometry, pulse oximetry or near-infrared spectroscopy. The excitation events of the reference signal provide a time reference for the later analysis of the received electrograms. In the example of FIG. 1, the ECG signals I, II show a good indication of the ventricular electrical excitation event and are therefore suitable to be determined as reference signals. The CS signal in the example is less appropriate. In this example, the effect of the excitation event on the sensor signals F2, F3, F4 can be clearly distinguished in the electrograms from other signal contributions. Therefore, the blind source separation or analysis of periodicity methods based on the recorded signals may also be appropriate to determine the reference signal by computation.

[0031] Further, the computer system has a data processing module 130 with a signal selector module 131 to select 1300 from the plurality of the received electrical signals such electrical signals which are recorded at locations inside the atrium where the respective sensor has no contact to the atrial tissue, and/or which are recorded, irrespective of the sensor location, during time intervals where the atrium shows no electrical activity. Both conditions correspond to situations where no atrial activity is measured in the respective electrograms. If the first condition is fulfilled (i.e. there is no contact of the electrodes with the atrial tissue) the sensor electrodes of the catheter are surrounded by blood and no atrial electrical activity is measured. If the second condition (time interval with no electrical activity of the atrium) is fulfilled, atrial tissue contact becomes irrelevant because no electrical activity can be measured as long as no electrical activity occurs. As the atrial activity contribution in the selected signals is negligible they basically measure the VFF contribution at the respective sensor locations. Separating data according to time intervals can be done automatically by analyzing the respective time intervals with regards to signal data indicating atrial electrical activity. Separating signals in relation to locations with or without atrial tissue contact may be achieved by, for example, using a catheter which can measure the contact pressure. In such embodiments, a contact pressure magnitude greater zero indicates tissue contact. In an alternative embodiment, the distance of respective electrodes and the virtual heart anatomy may be measured. In other words, the position of the catheter and its sensor electrodes can be determined for example, by using a Coronary Sinus catheter sensor (CSCS) as a position reference signal source to determine the current position of the electrodes within the atrium while sensor data is recorded. This position can then be marked in a virtual spatial model of the atrium and the distance to the surface elements of the spatial model can be determined by appropriate distance algorithms. A distance greater zero indicates no tissue contact. Position reference, as used herein, means that the position of each further sensor is known relative to the position of the reference sensor. Often, the CSCS is used as the position reference. Typically, the locations of a few sensors in the heart are determined via magnet coils mounted at the tip of the catheter and underneath the patient whereas the other sensors are located using impedance and the principle of potential divider. Once the position of the reference sensor is determined the positions of other sensors are known relative to the position of the reference sensor. If the patient moves, the relative positions can be determined because the movement of the reference sensor is detected. It is also possible to determine the movement of the patient by other means. For example, the movement may be detected by a camera system or by ultrasonic sensors. The detected movement can then be used to re-compute the sensor locations by compensating the movement accordingly. In these implementations, the position reference function of the reference sensor is optional.

[0032] Further, the data processor 130 has a VFF approximator module 132 to determine 1400 a spatio-temporal distribution of the Ventricular Far Field inside the atrium by approximating the spatio-temporal distribution based on signal data of the selected signals. The approximation method can be a polynomial approximation, an approximation with a dipole source model, an approximation with spatio-temporal linear combination of the recorded signal data, approximation performed with Radial Basis Functions, approximation using Look-Up Tables, or approximation performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table.

[0033] An example of polynomial approximation is to use a polynomial that describes the value of the VFF potential in dependency of the spatial coordinates (like Cartesian coordinate system with coordinates x, y, z or the Polar coordinate system spatial coordinates r, φ, □) for each point in time. The degree of the polynomial can be chosen to minimize the residual of approximation for the measured signal data. A least-squares-fit can be used to estimate the coefficients of the polynomial.

[0034] An example of a linear polynomial model (second order) can be expressed by the following formula:

$$\begin{pmatrix} \phi_1 \\ \phi_2 \\ \cdots \\ \phi_M \end{pmatrix} = \begin{pmatrix} 1 & x_1 & y_1 & z_1 & x_1y_1 & x_1z_1 & y_1z_1 & x_1^2 & y_1^2 & z_1^2 \\ 1 & x_2 & y_2 & z_2 & x_2y_2 & x_2z_2 & y_2z_2 & x_2^2 & y_2^2 & z_2^2 \\ \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots & \cdots \\ 1 & x_M & y_M & z_M & x_My_M & x_Mz_M & y_Mz_M & x_M^2 & y_M^2 & z_M^2 \end{pmatrix} \cdot \begin{pmatrix} c_0 \\ c_x \\ c_y \\ c_z \\ c_{xy} \\ c_{xz} \\ c_{yz} \\ c_{x2} \\ c_{y2} \\ c_{z2} \end{pmatrix}$$

with subscripts 1 to M indicating the respective measurements; x, y, z are the spatial coordinates, and cx are the coefficients of model component X. φ are the resulting potentials at the measurement positions. The coefficients are determined during model generation.

[0035] An example of a dipole source model approach is to place multiple dipoles in space and adapt their strength to compute a potential field which approximates the measured signal data. The dipoles may be located in the atria, in the ventricles, and/or surrounding region. If for example the location of dipoles should reflect the position of the ventricles, this position can either be known from intracardiac mapping using an EAMS or by additional imaging techniques like computed tomography or magnetic resonance imaging or based on a statistical model of cardiac shape. The dipoles of the dipole source model may be located in the convex hull of the cardiac chambers or at the tissue blood boundary or in the tissue or other locations. The strength of each dipole in each spatial direction can be adapted so that the resulting potential field approaches the measured signal data. Approximation can be done by using a least-squares-approximation. Alternatively, the method of Tikhonov regularization can be applied to constrain the strength of individual dipoles and to prevent over-fitting. Additional dipoles may be located in the atrium, generating a model to explain both local atrial and VFF components. An example of dipole source model can be expressed by the following formula:

$$\Phi(\vec{r}) = \frac{1}{4\pi\kappa}\left[\frac{\left(\vec{r}-\vec{r_p}\right)}{\left|\vec{r}-\vec{r_p}\right|^3}\right] \cdot J_i\left(\vec{r_p}\right)$$

with r being the measurement position in space, r_P being the position of the dipole P in space, J being the impressed current density of dipole P, and κ being the conductivity.

[0036] An example of a spatio-temporal linear combination is the so-called principal component analysis, which can be used to remove random noise from the signals and focus on dominant signal components. Another spatio-temporal linear combination is the weighted average of selected signals for each point in time. By computing the approximation model on the basis of the selected signal data, interpolation and extrapolation of the VFF contribution is possible. Therefore, there is no need for sensor data recorded at locations with atrial tissue contact. Rather, the VFF contribution at such locations can be extrapolated from the determined spatio-temporal distribution.

[0037] Turning briefly to FIG. 11, this figure illustrates an example of a temporal model TM1 which is generated as a basis system reflecting the temporal VFF course. This basis system can be based on signal segments associated with electrodes that are far enough away from the atrial wall so that the signals are not disturbed by the atrial component, and analyzed using the so-called principal component analysis. Each basis vector has a duration of the time interval in which the VFF contribution should be removed (e.g., a 250 ms time interval associated with the R wave). From signal theory, the first basis vector describes the template VFF signal, and together with the second and a few more basis vectors, all VFF signals anywhere in the atrium can be described by an appropriately chosen linear combination of the basis vectors. Advantageously, the number of basis vectors is in the range of 1 to 10. The example of FIG. 11 shows a temporal model using four basis vectors BV (1, 2, 3, 4). Measured signals, synchronized to the VFF window, can now be approximated by this temporal basis system TM1 using a Least Squares approach. The obtained estimate for the VFF contribution can subsequently be subtracted from the measurement to get a corrected signal showing the atrial signal without the disturbance caused by the VFF contribution. In other words, the signal components orthogonal to the temporal VFF basis are considered to represent the pure atrial signal.

[0038] Turning back to FIG. 1, in one embodiment, the system 100 has a visualizer component 140 which is configured to render a graphic representation 231 of the spatio-temporal distribution. The graphic representation can be displayed to a medically trained person for diagnosis support through a display device 220 which is communicatively coupled with the computer system 100 through the interface 110.

[0039] In one embodiment, the system 100 has a synchronizer component 133 which is used in situations where the plurality of received signals includes signal data recorded over multiple ventricular electrical excitation periods. This is typically the case when the catheter is moved to multiple measurement locations within the patient's atrium for recording respective sensor data. It is to be noted that while recording sensor data the catheter is not moved. In this case, the selected signal data is synchronized 1350 with the measured electrical excitation events. In other words, a particular electrical excitation event of the heart serves as the time reference for the recorded signals which are recorded during the respective heart beat interval. The signal data recorded over multiple heartbeat intervals can thus be projected into one interval where the respective excitation event defines the start point of the interval. In this embodiment the spatio-temporal distribution is determined based on the synchronized signal data. The VFF contribution can thus be determined 1400 by using all selected measurement data recorded over a period of time during which the catheter was moved through the patient's atrium. By using the approximator 132 functions for interpolation and extrapolation a continuous model of the VFF contribution may be generated for the entire inner of the atrium.

[0040] In one embodiment, the system 100 may further include a model generator 135 to generate a plurality of models wherein each model relates to a particular section within the atrium in combination with a particular time point in relation to the electrical excitation event. Such a model reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point. For example, the model sections may correspond to virtual layers within the atrium.

[0041] In one embodiment, the system 100 has an Atrial Electrogram improvement module 134 also referred to as VFF corrector 134. The VFF corrector subtracts 1500 the contribution of the Ventricular Far Field at a particular location inside the atrium from the Atrial Electrogram represented by an electrical signal generated by one of the sensors S1 to Sn at the particular location. As a result, the VFF contribution is removed from the respective Atrial Electrograms and the visibility of the signals of interest for medical diagnosis is improved in the Atrial Electrograms.

[0042] FIGs. 3A and 3B illustrate the function of the VFF corrector. FIG. 3A is a schematic illustration of signal contributions in a unipolar atrial electrogram (UAE) before and after correction. In the left part of the figure (left to the dotted vertical line), signals are shown which occur during the period where the atrium is active. In the right part of the figure, signals are shown during the period where the ventriculum is active. For each signal, the momentary signal value corresponds to a voltage which can be measured by the respective sensors. The signal AA shows the UAE contribution of the atrial activity of an atrium. The shown signal curve is a typical curve which is expected at a particular sensor location. The signal PN illustrates the powerline noise (e.g., 50 Hz) which superimposes the AA signal. The ECG signal illustrates the ventricular activity of the heart with the so-called P, Q, R, and S waves. The QRS waves are the root cause of the Ventricular Far Field VFF which can be observed in the atrium and which also superimposes the UAEs. The EGMm signal schematically illustrates a measured UAE with signal contributions from the AA, PN and VFF signals. A goal is to eliminate the VFF contribution from EGMm as illustrated in the corrected UAE for the signal AGMc.

[0043] FIB. 3B shows simulated signals over a time period of 600 ms which have a more realistic signal shape (Voltage) than the schematic signal illustrations in FIG. 3A. The Atrial signal AA corresponds to the atrial activity of the atrium. The Ventricular component VA shows the VFF contribution to the UAE (Unipolar EGM). It is to be noted that the VFF contribution can result in a negative Voltage peak (cf. FIG. 3A) or in positive Voltage peaks dependent on how the voltage is measured. The Powerline hum PN corresponds to the PN noise curve in FIG. 3A. In addition, HF noise HFN is present in a real UAE. The EGMs signal illustrates the aggregate UAE which results from the signal contributions of the signals above. In cases where the VFF contribution interferes with the AA signal the AA signal information becomes invisible in the measured EGM signal because the VFF contribution dominates the aggregate signal in a way that it becomes meaningless for diagnostic analysis purposes.

[0044] FIG. 4 shows a heart catheter 201 with sensor electrodes S1 to S10. Sensor electrodes S1 to S10 are only illustrated for two (B1, B2) of the catheter branches B1 to B7. However, each branch is equipped with a plurality of sensor electrodes. Such catheter sensors are typically used inside the atrium and moved through the atrium to measure the atrial activity by making a contact between the sensors S1 to S10 and the atrial tissue of the patient. It is clear that typically only a subset of the sensors can make contact with the atrial tissue at a given point in time. When the catheter 201 is moved through the patient’s atrium, at least a subset of the sensor electrodes is always entirely surrounded by blood (i.e. such electrodes which have no contact with the atrial tissue). It can also happen that no sensor electrode has a contact with the atrial tissue at a given point in time. In normal EGM analysis the measurement values provided by the sensors S1 to S10 without having contact with the atrial tissue are discarded as they do not carry useful information with regards to the atrial activity. However, such measurement data is used by the disclosed method for determining the VFF contribution at respective locations.

[0045] FIG. 5 illustrates, at a particular point in time, an atrium 500 with a catheter inside the atrium and some respective sensor electrodes Sn. The catheter is placed in such a way that the sensor electrodes Sn of three branches Ba, Bb, Be are in contact with the atrial tissue whereas the sensor electrodes mounted on the neighboring or opposite branches are surrounded by blood and are not shown in the figure.

[0046] FIGs. 6A and 6B show traces of an electrode sensor of a heart catheter being moved through an atrium 500. In FIG. 6A, each dot on a trace represents a sensor location at which measurement data was recorded and synchronized. For example, location LSa is a location close to the upper bound (roof) of the atrium 500. LSb is a location close to lower bound (valve plane) of the atrium and LSc is a location closer towards the center of the atrium 500. In the example, locations of measurement data outside the atrium 500 (for example, the trace portions which are located left from the atrium boundary) result from particular conditions during the measurement. In those cases the catheter was pushed against the atrial tissue (for example to get as many electrodes as possible into contact with the atrial tissue). The pressure which was applied during the measurement caused some stretch of the atrial tissue so that the recorded measurement data appear outside the atrial boundaries which were extended to the respective measurement locations during the measurement. However, after the measurement data was recorded the atrial tissue retracted to its normal position once the pressure was released. At each point in time the catheter can simultaneously record signal data provided by the electrode sensors mounted on the various branches. That is, signal data being recorded simultaneously reflect signal data at different locations but during the same heart beat interval. However, the number of locations in FIG. 6A is recorded over a longer time period (e.g. 20 minutes). That is, most measurement data is recorded at different points in time which also belong to different heart beat intervals.

[0047] A catheter typically has a plurality of electrode sensors (cf. FIG. 4). FIG. 6B illustrates traces of one electrode sensor using a grey scale to illustrate the measurement time points for the respective sensor locations. The catheter positions at the beginning and at the end of the measurement are visualized by the catheter shape representations 201b and 201e, respectively, to provide an impression of the size of the area which can be covered by the electrodes of the catheter at any given point in time. In other words, for this area the sensor electrodes record measurement data for different sensor locations (i.e. the locations of the respective electrodes) in parallel. That is, for this area measurement data can be simultaneously recorded in relation to the same heartbeat interval. In the example, the first sensor location LS1 is illustrated as a black bullet which indicates that the measurement has been made during the first minute of the measurement period. The measurement data recorded at the second sensor location LS2 was recorded approximately after nine minutes, and the measurement data recorded at the third sensor location LS3 was recorded approximately after 20 minutes towards the end of the measurement period.

[0048] FIG. 7 shows an example of synchronized selected signals 600 as captured by an embodiment of the computer system. The figure illustrates measurement data recorded at a plurality of different sensor locations (e.g., the sensor locations of the traces in FIG.s 6A or 6B) and for different heart beat intervals. Thereby, each curve of the measurement data (e.g., curve 601, 602, 603) relates to a particular sensor electrode signal recorded a respective location during a respective heartbeat interval. The synchronizer of the computer system can use a reference signal (e.g., the peak of the R wave) of each heartbeat interval to define the time point 0 for the respective heartbeat interval. All measurement data can thus be superposed into the same time interval starting with the common sync time point 0. The synchronized selected signals in FIG. 7 can be handled in the same way like the sensor signals of the sensor electrodes being recorded simultaneously during the same heartbeat interval. That is, the synchronized selected signals correspond to a snapshot of the atrium where all measurement data at the respective locations were recorded during the same heartbeat interval. The basic assumption thereby is that the signal behavior at each location is substantially the same after the sync point t_0 for each heartbeat interval.

[0049] Whereas the real simultaneous measurement data of the electrodes of the catheter during a single heartbeat interval (while the catheter is at a particular location) allow to compute the VFF contribution for the atrium section covered by the electrodes during this heartbeat interval in near-real-time, the synchronized selected signal data allows to determine the VFF contribution after the end of the measurement period (i.e. offline) for the entire section covered by all sensor locations used during the measurement period (e.g., 10 to 30 min).

[0050] FIG. 8 illustrates a spatio-temporal distribution 800 of the Ventricular Far Field contribution VFFc at a particular point in time for one layer 501 of the atrium. The VFF contribution can be determined for multiple layers of the entire atrium based on synchronized selected signals from a plurality of sensor locations which are reached by moving the catheter trough the atrium during a longer measurement period. Sensor locations in the proximity of the selected layer contribute to the VFFc computation of this layer.

[0051] However, the proposed approximation methods can also be applied to the measurement data provided by the catheter electrodes during a single heartbeat interval. This allows to determine in near-real-time VFF contributions as a spatio-temporal distribution for the section of the atrium which is covered by the catheter electrodes during this heartbeat interval. In other words, the corrections of the respective UAEs can be performed in near-real-time for such subsections of the atrium while the catheter is being moved through the patient's atrium.

[0052] Turning back to the simulated selected signals scenario illustrated in FIG. 8, the amplitude of the VFF contribution VFFc at the respective locations of the layer 501 at a particular time point is visualized. Amplitude values around 0 mV are measured at the lower right part of the layer 501 in the figure, representing the right superior pulmonary vein. Higher amplitude values (> 2 mV in magnitude) are measured at the upper left and lower left layer sections in the figure, being located closer to the mitral valve. By using any one of the methods: polynomial approximation, approximation with a dipole source model, approximation with a spatio-temporal linear combination (e.g., principal component analysis), approximation performed with Radial Basis Functions, approximation using Look-Up Tables, and approximation performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table for approximating the spatio-temporal distribution 800 based on signal data of the selected signals a continuous spectrum of VFFc voltage values can be computed. Extrapolation allows to also compute values for the atrial tissue surface. A person skilled in the art may use other approximation methods which are appropriate in the described context.

[0053] In more general terms, a plurality of models can be generated wherein each model relates to a particular section (e.g., layer 501) within the atrium in combination with a particular time point in relation to the electrical excitation event. Such a model reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point.

[0054] FIG. 9A shows a real world example 9010 of recorded reference signals and selected electrogram signals. The signals I, II show the R-wave of the patient's ventricular activity defining the excitation events E1, E2 which can later on be used as reference signals for the synchronization of the unipolar atrial electrograms across multiple heartbeat intervals.

[0055] The CS signal corresponds to the signal of a coronary sinus sensor and is not relevant for this example. The signal G2 corresponds to a broken electrode of the catheter. The other signals (F*, G*, H*, B*, C*, D*) correspond to signals recorded by the electrode sensors of the respective branches of the catheter. The impact of the VFF in the

Unipolar Atrial Electrograms is clearly visible at the times the excitation events E1, E2 occur (~ 0.94 s, 1.7 s).

[0056] FIG. 9B shows a real world example of the corrected electrograms 9020 according to an embodiment. The corrected UAEs (F*c, G*c, H*c, B*c, C*c, D*c) are derived by subtracting the VFF contribution from the original UAEs in FIG. 9A once the VFF contribution has been determined in accordance with the method disclosed herein.

[0057] FIG. 10 illustrates different performances of different approximation methods for determining the VFF contribution. The horizontal axis shows the number of estimated Parameters (number of unknowns) which corresponds to the complexity of the model. The vertical axis shows the percentage of not perfectly estimated data points. Not perfectly estimated in this context means that the estimate is worse than some specified threshold.

[0058] The curves show the performance for two approximation methods: the Dipole method (AM3, AM4) versus the Polynomial method (AM1, AM2). The Dipole method (AM3, AM4) shows a better performance than the Polynomial method (AM1, AM2).

[0059] For the Polynomial method the curve AM1 is the result of an approximation using a single model for the entire atrium whereas the curve AM2 is the result for an approximation based on two sub-models which are separated by a plane parallel to the plane between the atria and ventricles (valves). For this method, the approach using two sub-models provides better performance.

[0060] For the Dipole method the curve AM4 is the result of an approximation using a single model for the entire atrium whereas the curve AM3 is the result for an approximation based on two sub-models which are separated by a plane parallel to the plane between the atria and ventricles (valves). For this method, the single model approach provides slightly better performance.

[0061] FIG. 12 is a diagram that shows an example of a generic computer device 900 and a generic mobile computer device 950, which may be used with the techniques described here. Computing device 900 is intended to represent various forms of digital computers, such as laptops, desktops, workstations, personal digital assistants, tablets, servers, blade servers, mainframes, and other appropriate computers. Generic computer device 900 may correspond to a computer system 100 as illustrated in FIG. 1. Computing device 950 is intended to represent various forms of mobile devices, such as personal digital assistants, cellular telephones, smart phones, and other similar computing devices. For example, computing device 950 may be used by a user as a front end to interact with the computer system 100. Computing device may, for example, include the display device 220 of FIG. 1. The components shown here, their connections and relationships, and their functions, are meant to be exemplary only, and are not meant to limit implementations of the inventions described and/or claimed in this document.

[0062] Computing device 900 includes a processor 902, memory 904, a storage device 906, a high-speed interface 908 connecting to memory 904 and high-speed expansion ports 910, and a low speed interface 912 connecting to low speed bus 914 and storage device 906. Each of the components 902, 904, 906, 908, 910, and 912, are interconnected using various busses, and may be mounted on a common motherboard or in other manners as appropriate. The processor 902 can process instructions for execution within the computing device 900, including instructions stored in the memory 904 or on the storage device 906 to display graphical information for a GUI on an external input/output device, such as display 916 coupled to high speed interface 908. In other implementations, multiple processing units and/or multiple buses may be used, as appropriate, along with multiple memories and types of memory. Also, multiple computing devices 900 may be connected, with each device providing portions of the necessary operations (e.g., as a server bank, a group of blade servers, or a processing device).

[0063] The memory 904 stores information within the computing device 900. In one implementation, the memory 904 is a volatile memory unit or units. In another implementation, the memory 904 is a non-volatile memory unit or units. The memory 904 may also be another form of computer-readable medium, such as a magnetic or optical disk.

[0064] The storage device 906 is capable of providing mass storage for the computing device 900. In one implementation, the storage device 906 may be or contain a computer-readable medium, such as a floppy disk device, a hard disk device, an optical disk device, or a tape device, a flash memory or other similar solid state memory device, or an array of devices, including devices in a storage area network or other configurations. A computer program product can be tangibly embodied in an information carrier. The computer program product may also contain instructions that, when executed, perform one or more methods, such as those described above. The information carrier is a computer- or machine-readable medium, such as the memory 904, the storage device 906, or memory on processor 902.

[0065] The high speed controller 908 manages bandwidth-intensive operations for the computing device 900, while the low speed controller 912 manages lower bandwidth-intensive operations. Such allocation of functions is exemplary only. In one implementation, the high-speed controller 908 is coupled to memory 904, display 916 (e.g., through a graphics processor or accelerator), and to high-speed expansion ports 910, which may accept various expansion cards (not shown). In the implementation, low-speed controller 912 is coupled to storage device 906 and low-speed expansion port 914. The low-speed expansion port, which may include various communication ports (e.g., USB, Bluetooth, Ethernet, wireless Ethernet) may be coupled to one or more input/output devices, such as a keyboard, a pointing device, a scanner, or a networking device such as a switch or router, e.g., through a network adapter.

[0066] The computing device 900 may be implemented in a number of different forms, as shown in the figure. For example, it may be implemented as a standard server 920, or multiple times in a group of such servers. It may also be implemented as part of a rack server system 924. In addition, it may be implemented in a personal computer such as a laptop computer 922. Alternatively, components from computing device 900 may be combined with other components in a mobile device (not shown), such as device 950. Each of such devices may contain one or more of computing device 900, 950, and an entire system may be made up of multiple computing devices 900, 950 communicating with each other.

[0067] Computing device 950 includes a processor 952, memory 964, an input/output device such as a display 954, a communication interface 966, and a transceiver 968, among other components. The device 950 may also be provided with a storage device, such as a microdrive or other device, to provide additional storage. Each of the components 950, 952, 964, 954, 966, and 968, are interconnected using various buses, and several of the components may be mounted on a common motherboard or in other manners as appropriate.

[0068] The processor 952 can execute instructions within the computing device 950, including instructions stored in the memory 964. The processor may be implemented as a chipset of chips that include separate and multiple analog and digital processing units. The processor may provide, for example, for coordination of the other components of the device 950, such as control of user interfaces, applications run by device 950, and wireless communication by device 950.

[0069] Processor 952 may communicate with a user through control interface 958 and display interface 956 coupled to a display 954. The display 954 may be, for example, a TFT LCD (Thin-Film-Transistor Liquid Crystal Display) or an OLED (Organic Light Emitting Diode) display, or other appropriate display technology. The display interface 956 may comprise appropriate circuitry for driving the display 954 to present graphical and other information to a user. The control interface 958 may receive commands from a user and convert them for submission to the processor 952. In addition, an external interface 962 may be provided in communication with processor 952, so as to enable near area communication of device 950 with other devices. External interface 962 may provide, for example, for wired communication in some implementations, or for wireless communication in other implementations, and multiple interfaces may also be used.

[0070] The memory 964 stores information within the computing device 950. The memory 964 can be implemented as one or more of a computer-readable medium or media, a volatile memory unit or units, or a non-volatile memory unit or units. Expansion memory 984 may also be provided and connected to device 950 through expansion interface 982, which may include, for example, a SIMM (Single In Line Memory Module) card interface. Such expansion memory 984 may provide extra storage space for device 950, or may also store applications or other information for device 950. Specifically, expansion memory 984 may include instructions to carry out or supplement the processes described above, and may include secure information also. Thus, for example, expansion memory 984 may act as a security module for device 950, and may be programmed with instructions that permit secure use of device 950. In addition, secure applications may be provided via the SIMM cards, along with additional information, such as placing the identifying information on the SIMM card in a non-hackable manner.

[0071] The memory may include, for example, flash memory and/or NVRAM memory, as discussed below. In one implementation, a computer program product is tangibly embodied in an information carrier. The computer program product contains instructions that, when executed, perform one or more methods, such as those described above. The information carrier is a computer- or machine-readable medium, such as the memory 964, expansion memory 984, or memory on processor 952, that may be received, for example, over transceiver 968 or external interface 962.

[0072] Device 950 may communicate wirelessly through communication interface 966, which may include digital signal processing circuitry where necessary. Communication interface 966 may provide for communications under various modes or protocols, such as GSM voice calls, SMS, EMS, or MMS messaging, CDMA, TDMA, PDC, WCDMA, CDMA2000, or GPRS, EDGE, UMTS, LTE, among others. Such communication may occur, for example, through radio-frequency transceiver 968. In addition, short-range communication may occur, such as using a Bluetooth, WiFi, or other such transceiver (not shown). In addition, GPS (Global Positioning System) receiver module 980 may provide additional navigation- and locationrelated wireless data to device 950, which may be used as appropriate by applications running on device 950.

[0073] Device 950 may also communicate audibly using audio codec 960, which may receive spoken information from a user and convert it to usable digital information. Audio codec 960 may likewise generate audible sound for a user, such as through a speaker, e.g., in a handset of device 950. Such sound may include sound from voice telephone calls, may include recorded sound (e.g., voice messages, music files, etc.) and may also include sound generated by applications operating on device 950.

[0074] The computing device 950 may be implemented in a number of different forms, as shown in the figure. For example, it may be implemented as a cellular telephone 980. It may also be implemented as part of a smart phone 982, personal digital assistant, or other similar mobile device.

[0075] Various implementations of the systems and techniques described here can be realized in digital electronic circuitry, integrated circuitry, specially designed ASICs (application specific integrated circuits), computer hardware, firmware, software, and/or combinations thereof. These various implementations can include implementation in one or more computer programs that are executable and/or interpretable on a programmable system including at least one programmable processor, which may be special or general purpose, coupled to receive data and instructions from, and to transmit data and instructions to, a storage system, at least one input device, and at least one output device.

[0076] These computer programs (also known as programs, software, software applications or code) include machine instructions for a programmable processor, and can be implemented in a high-level procedural and/or object-oriented programming language, and/or in assembly/machine language. As used herein, the terms "machine-readable medium" and "computer-readable medium" refer to any computer program product, apparatus and/or device (e.g., magnetic discs, optical disks, memory, Programmable Logic Devices (PLDs)) used to provide machine instructions and/or data to a programmable processor, including a machine-readable medium that receives machine instructions as a machine-readable signal. The term "machine-readable signal" refers to any signal used to provide machine instructions and/or data to a programmable processor.

[0077] To provide for interaction with a user, the systems and techniques described here can be implemented on a computer having a display device (e.g., a CRT (cathode ray tube) or LCD (liquid crystal display) monitor) for displaying information to the user and a keyboard and a pointing device (e.g., a mouse or a trackball) by which the user can provide input to the computer. Other kinds of devices can be used to provide for interaction with a user as well; for example, feedback provided to the user can be any form of sensory feedback (e.g., visual feedback, auditory feedback, or tactile feedback); and input from the user can be received in any form, including acoustic, speech, or tactile input.

[0078] The systems and techniques described here can be implemented in a computing device that includes a backend component (e.g., as a data server), or that includes a middleware component (e.g., an application server), or that includes a front end component (e.g., a client computer having a graphical user interface or a Web browser through which a user can interact with an implementation of the systems and techniques described here), or any combination of such backend, middleware, or frontend components. The components of the system can be interconnected by any form or medium of digital data communication (e.g., a communication network). Examples of communication networks include a local area network ("LAN"), a wireless local area network ("WLAN"), a wide area network ("WAN"), and the Internet.

[0079] The computing device can include clients and servers. A client and server are generally remote from each other and typically interact through a communication network. The relationship of client and server arises by virtue of computer programs running on the respective computers and having a client-server relationship to each other.

[0080] A number of embodiments have been described. Nevertheless, it will be understood that various modifications may be made without departing from the scope of the invention.

[0081] In addition, the logic flows depicted in the figures do not require the particular order shown, or sequential order, to achieve desirable results. In addition, other steps may be provided, or steps may be eliminated, from the described flows, and other components may be added to, or removed from, the described systems. Accordingly, other embodiments are within the scope of the following claims.

Claims

1. A computer-implemented method (1000) for determining Ventricular Far Field contribution in Atrial Electrograms of a patient, the method comprising:

receiving (1100) a plurality of previously recorded electrical signals (ECGm, Fm, Gm, Hm, Bm, Cm, Dm) measured by a plurality of sensors (S1 to Sn) wherein the plurality of electrical signals relates to a plurality of locations in an atrium (500) of the patient;
determining (1200) a reference signal (RS, I, II, CS) measuring the electrical excitation of the patient's ventricles;
selecting (1300) from the plurality of the received electrical signals such electrical signals which are recorded under one or more of the following conditions by verifying each of the three conditions:

- the respective signals are recorded at locations inside the atrium (500) where the respective sensor has no contact to the atrial tissue,
- the respective signals are recorded, irrespective of the sensor location, during time intervals where the respective part of the atrium shows no electrical activity,
- the respective signals are recorded, irrespective of the sensor location, during time intervals which comprise a plurality of heart beat intervals and are subject to subsequent spatial smoothing; and

determining (1400) a spatio-temporal distribution of the Ventricular Far Field inside the atrium by approximating the spatio-temporal distribution based on signal data of the selected signals by using an approximation model, wherein the approximation model is computed on the basis of the selected signal data and allows for interpolation and extrapolation of the Ventricular Far Field contribution at locations where no sensor data has been recorded.

2. The method of claim 1, wherein the approximation model is selected from any one of the following: a linear spatial model, a non-linear spatial model, a temporal model, and a look-up table.

3. The method of claim 1, wherein approximating uses any one of the following methods:
polynomial approximation, approximation with a dipole source model, approximation with a spatio-temporal linear combination of the recorded signal data, approximation performed with Radial Basis Functions, approximation using Look-Up Tables, and approximation performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table.

4. The method of any of the previous claims, wherein the plurality of received signals comprises signal data recorded over multiple ventricular electrical excitation periods, the method further comprising:
synchronizing (1350) the selected signal data with the measured electrical excitation events (E1, E2), wherein determining (1400) the spatio-temporal distribution is based on the synchronized signal data (601, 602, 603).

5. The method of any of the previous claims, wherein determining (1400) the spatio-temporal distribution of the Ventricular Far Field further comprises:
generating a plurality of models wherein each model relates to a particular section within the atrium in combination with a particular time point in relation to the electrical excitation event (E1, E2) and reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point.

6. The method of any of the previous claims, further comprising:
subtracting (1600) the contribution of the Ventricular Far Field at a particular location inside the atrium from the Atrial Electrogram represented by electrical signal data measured by a particular sensor (S1 to Sn) at the particular location (LSa, LSb, LSc, LS1, LS2, LS3).

7. The method of any of the previous claims, wherein determining (1200) a reference signal is performed by any one of the following methods:

   receiving the reference signal from one or more electrocardiogram sensors measuring at least the R wave of the patient's ventricular electrical activity,
   receiving the reference signal from a coronary sinus catheter sensor,
   computing the reference signal from the recorded electrical signal data by blind source separation or analysis of periodicity, and
   determining the reference signal based on information about the ventricular contraction obtained by using optical techniques like laser interferometry, pulse oximetry or near-infrared spectroscopy.

8. A computer program product for determining Ventricular Far Field contribution in Atrial Electrograms of a patient, the computer program product when loaded into a memory of a computing device and executed by at least one processor of the computing device executes the steps of the computer-implemented method according to any one of the previous claims.

9. A computer system (100) for determining Ventricular Far Field contribution in Atrial Electrograms of a patient, the computer system comprising:

   an interface module (110) configured to receive a plurality of electrical signals generated by a plurality of sensors wherein the plurality of electrical signals relate to a plurality of locations (LSa, LSb, LSc, LS1, LS2, LS3) in an atrium (500) of the patient;
   a reference module (120) configured to determine a reference signal (RS, I, II, CS) reflecting electrical excitation of the patient's ventricles;
   a data processing module (130) configured to select from the plurality of the received electrical signals such electrical signals which are recorded under one or more of the following conditions by verifying each of the three conditions:

   - the respective signals are recorded at locations inside the atrium (500) where the respective sensor has no contact to the atrial tissue,
   - the respective signals are recorded, irrespective of the sensor location, during time intervals where the respective part of the atrium shows no electrical activity"
   - the respective signals are recorded, irrespective of the sensor location, during time intervals which comprise a plurality of heart beat intervals and are subject to subsequent spatial smoothing; and

further configured to determine a spatio-temporal distribution of the Ventricular Far Field inside the atrium by approximating the spatio-temporal distribution (VFFc) based on signal data of the selected signals by using an approximation model, wherein the approximation model is computed on the basis of the selected signal data and allows for interpolation and extrapolation of the Ventricular Far Field contribution at locations where no sensor data has been recorded.

10. The system of claim 9, wherein the approximation model is selected from any one of the following: a linear spatial model, a non-linear spatial model, a temporal model, and a look-up table.

11. The system of claim 9, wherein approximating uses any one of the following methods:
polynomial approximation, approximation with a dipole source model, approximation with a spatio-temporal linear combination of the recorded signal data, approximation performed with Radial Basis Functions, approximation using Look-Up Tables, and approximation performed by transforming the signal data to a regular grid before providing it in form of a Look-Up Table.

12. The system of any of the claims 9 to 11, wherein the plurality of received signals comprise signal data recorded over multiple ventricular electrical excitation periods, the data processing module (130) further configured to:
synchronize the selected signal data with the measured electrical excitation events (E1, E2), wherein determination of the spatio-temporal distribution of the Ventricular Far Field is based on the synchronized signal data.

13. The system of any of the claims 9 to 12, wherein the data processing module has a model generator module (135) configured to:
generate a plurality of models wherein each model relates to a particular section within the atrium in combination with a particular time point in relation to the electrical excitation event and reflects a respective approximation of the Ventricular Far Field for the particular section at the particular time point.

14. The system of any one of the claims 9 to 13, wherein the data processing module has an Atrial Electrogram improvement module (134) configured to:
subtract the contribution of the Ventricular Far Field at a particular location inside the atrium from the Atrial Electrogram represented by an electrical signal generated by one of the sensors at the particular location.

15. The system of any one of the claims 9 to 14, wherein the reference module (120) is configured to determine one or more reference signals from the following list of potential reference signals:

the reference signal received from one or more electrocardiogram sensors measuring at least the R wave of the patient's ventricular electrical activity,
the reference signal received from a coronary sinus catheter sensor,
the reference signal computed from the recorded electrical signal data by blind source separation or analysis of periodicity, and
the reference signal determined based on information about the ventricular contraction obtained by using optical techniques like laser interferometry, pulse oximetry or near-infrared spectroscopy.

Patentansprüche

1. Computerimplementiertes Verfahren (1000) zum Bestimmen eines ventrikulären Fernfeldbeitrags in Herzvorhofelektrogrammen eines Patienten, das Verfahren umfassend:

Empfangen (1100) einer Vielzahl von zuvor aufgezeichneten elektrischen Signalen (ECGm, Fm, Gm, Hm, Bm, Cm, Dm), gemessen durch eine Vielzahl von Sensoren (S1 bis Sn), wobei sich die Vielzahl von elektrischen Signalen auf eine Vielzahl von Stellen in einem Vorhof (500) des Patienten beziehen;
Bestimmen (1200) eines Referenzsignals (RS, I, II, CS), das die elektrische Anregung der Ventrikel des Patienten misst;
Auswählen (1300) von solchen elektrischen Signalen aus der Vielzahl der empfangenen elektrischen Signale, die unter einer oder mehreren der folgenden Bedingungen aufgezeichnet werden, durch ein Verifizieren jeder der drei Bedingungen:

- die jeweiligen Signale werden an Stellen innerhalb des Vorhofs (500) aufgezeichnet, wobei der jeweilige

Sensor keinen Kontakt zu dem Herzvorhofgewebe aufweist,
- die jeweiligen Signale werden unabhängig von der Sensorstelle während Zeitintervallen aufgezeichnet, wobei der jeweilige Teil des Vorhofs keine elektrische Aktivität zeigt,
- die jeweiligen Signale werden unabhängig von der Sensorstelle während Zeitintervallen aufgezeichnet, die eine Vielzahl von Herzschlagintervallen umfassen und einer nachfolgenden räumlichen Glättung unterzogen werden; und

Bestimmen (1400) einer räumlich-zeitlichen Verteilung des ventrikulären Fernfeldes innerhalb des Vorhofs durch ein Annähern der räumlich-zeitlichen Verteilung basierend auf Signaldaten der ausgewählten Signale durch ein Verwenden eines Näherungsmodells, wobei das Näherungsmodell auf der Basis der ausgewählten Signaldaten berechnet wird und eine Interpolation und eine Extrapolation des ventrikulären Fernfeldbeitrags an Stellen ermöglicht, an denen keine Sensordaten aufgezeichnet worden sind.

2. Verfahren nach Anspruch 1, wobei das Näherungsmodell aus einem beliebigen der Folgenden ausgewählt ist: einem linearen räumlichen Modell, einem nichtlinearen räumlichen Modell, einem zeitlichen Modell und einer Nachschlagetabelle.

3. Verfahren nach Anspruch 1, wobei das Annähern ein beliebiges der folgenden Verfahren verwendet: Polynomnäherung, Näherung mit einem Dipolquellenmodell, Näherung mit einer räumlich-zeitlichen linearen Kombination der aufgezeichneten Signaldaten, Näherung, die mit radialen Basisfunktionen durchgeführt wird, Näherung unter Verwendung von Nachschlagetabellen, und Näherung, die durch ein Transformieren der Signaldaten in ein regelmäßiges Gitter durchgeführt wird, bevor sie in Form einer Nachschlagetabelle bereitgestellt werden.

4. Verfahren nach einem der vorstehenden Ansprüche, wobei die Vielzahl von empfangenen Signalen Signaldaten umfasst, die über mehrere ventrikuläre elektrische Anregungszeiträume aufgezeichnet werden, das Verfahren ferner umfassend:
Synchronisieren (1350) der ausgewählten Signaldaten mit den gemessenen elektrischen Anregungsereignissen (E1, E2), wobei das Bestimmen (1400) der räumlich-zeitlichen Verteilung auf den synchronisierten Signaldaten (601, 602, 603) basiert.

5. Verfahren nach einem der vorstehenden Ansprüche, wobei das Bestimmen (1400) der räumlich-zeitlichen Verteilung des ventrikulären Fernfeldes ferner umfasst:
Erzeugen einer Vielzahl von Modellen, wobei sich jedes Modell auf einen bestimmten Abschnitt innerhalb des Vorhofs in Kombination mit einem bestimmten Zeitpunkt in Bezug auf das elektrische Anregungsereignis (E1, E2) bezieht und eine jeweilige Näherung des ventrikulären Fernfeldes für den bestimmten Abschnitt zu dem bestimmten Zeitpunkt widerspiegelt.

6. Verfahren nach einem der vorstehenden Ansprüche, ferner umfassend:
Subtrahieren (1600) des Beitrags des ventrikulären Fernfeldes an einer bestimmten Stelle innerhalb des Vorhofs aus dem Herzvorhofelektrogramm, das durch elektrische Signaldaten dargestellt wird, die durch einen bestimmten Sensor (S1 bis Sn) an der bestimmten Stelle (LSa, LSb, LSc, LS1, LS2, LS3) gemessen werden.

7. Verfahren nach einem der vorstehenden Ansprüche, wobei das Bestimmen (1200) eines Referenzsignals durch ein beliebiges der folgenden Verfahren durchgeführt wird:

Empfangen des Referenzsignals von einem oder mehreren Elektrokardiogrammsensoren, die mindestens die R-Welle der ventrikulären elektrischen Aktivität des Patienten messen,
Empfangen des Referenzsignals von einem Koronarsinuskathetersensor,
Berechnen des Referenzsignals von den aufgezeichneten elektrischen Signaldaten durch Blindquellentrennung oder Analyse der Periodizität, und
Bestimmen des Referenzsignals basierend auf Informationen über die ventrikuläre Kontraktion, die durch Verwenden von optischen Techniken wie Laserinterferometrie, Pulsoximetrie oder Nahinfrarot-Spektroskopie erhalten werden.

8. Computerprogrammprodukt, zum Bestimmen eines ventrikulären Fernfeldbeitrags in Herzvorhofelektrogrammen eines Patienten, wobei das Computerprogrammprodukt, wenn es in einen Speicher einer Rechenvorrichtung geladen wird und durch mindestens einen Prozessor der Rechenvorrichtung ausgeführt wird, die Schritte des computerimplementierten Verfahrens nach einem der vorstehenden Ansprüche ausführt.

9. Computersystem (100) zum Bestimmen eines ventrikulären Fernfeldbeitrags in Herzvorhofelektrogrammen eines Patienten, das Computersystem umfassend:

ein Schnittstellenmodul (110), das konfiguriert ist, um eine Vielzahl von elektrischen Signalen zu empfangen, die durch eine Vielzahl von Sensoren erzeugt werden, wobei sich die Vielzahl von elektrischen Signalen auf eine Vielzahl von Stellen (LSa, LSb, LSc, LS1, LS2, LS3) in einem Vorhof (500) des Patienten beziehen;
ein Referenzmodul (120), das konfiguriert ist, um ein Referenzsignal (RS, I, II, CS) zu bestimmen, das eine elektrische Anregung der Ventrikel des Patienten widerspiegelt;
ein Datenverarbeitungsmodul (130), das konfiguriert ist, um solche elektrischen Signale aus der Vielzahl der empfangenen elektrischen Signale auszuwählen, die unter einer oder mehreren der folgenden Bedingungen aufgezeichnet werden, durch das Verifizieren jeder der drei Bedingungen:

- die jeweiligen Signale werden an Stellen innerhalb des Vorhofs (500) aufgezeichnet, wobei der jeweilige Sensor keinen Kontakt zu dem Herzvorhofgewebe aufweist,
- die jeweiligen Signale werden unabhängig von der Sensorstelle während Zeitintervallen aufgezeichnet, wobei der jeweilige Teil des Vorhofs keine elektrische Aktivität zeigt,
- die jeweiligen Signale werden unabhängig von der Sensorstelle während Zeitintervallen aufgezeichnet, die eine Vielzahl von Herzschlagintervallen umfassen und einer nachfolgenden räumlichen Glättung unterzogen werden; und

ferner konfiguriert ist, um eine räumlich-zeitliche Verteilung des ventrikulären Fernfeldes innerhalb des Vorhofs durch das Annähern der räumlich-zeitlichen Verteilung (VFFc) basierend auf Signaldaten der ausgewählten Signale durch Verwenden eines Näherungsmodells zu bestimmen, wobei das Näherungsmodell auf der Basis der ausgewählten Signaldaten berechnet wird und die Interpolation und die Extrapolation des ventrikulären Fernfeldbeitrags an Stellen ermöglicht, an denen keine Sensordaten aufgezeichnet worden sind.

10. System nach Anspruch 9, wobei das Näherungsmodell aus einem beliebigen der Folgenden ausgewählt ist: einem linearen räumlichen Modell, einem nichtlinearen räumlichen Modell, einem zeitlichen Modell und einer Nachschlagetabelle.

11. System nach Anspruch 9, wobei das Annähern eine beliebiges der folgenden Verfahren verwendet: Polynomnäherung, Näherung mit einem Dipolquellenmodell, Näherung mit einer räumlich-zeitlichen linearen Kombination der aufgezeichneten Signaldaten, Näherung, die mit radialen Basisfunktionen durchgeführt wird, Näherung unter Verwendung von Nachschlagetabellen, und Näherung, die durch das Transformieren der Signaldaten in ein regelmäßiges Gitter durchgeführt wird, bevor sie in Form einer Nachschlagetabelle bereitgestellt werden.

12. System nach einem der Ansprüche 9 bis 11, wobei die Vielzahl von empfangenen Signalen Signaldaten umfassen, die über mehrere ventrikuläre elektrische Anregungszeiträume aufgezeichnet werden, wobei das Datenverarbeitungsmodul (130) ferner konfiguriert ist zum:
Synchronisieren der ausgewählten Signaldaten mit den gemessenen elektrischen Anregungsereignissen (E1, E2), wobei die Bestimmung der räumlich-zeitlichen Verteilung des ventrikulären Fernfeldes auf den synchronisierten Signaldaten basiert.

13. System nach einem der Ansprüche 9 bis 12, wobei das Datenverarbeitungsmodul ein Modellgeneratormodul (135) aufweist, das konfiguriert ist zum:
Erzeugen einer Vielzahl von Modellen, wobei sich jedes Modell auf einen bestimmten Abschnitt innerhalb des Vorhofs in Kombination mit einem bestimmten Zeitpunkt in Bezug auf das elektrische Anregungsereignis bezieht und eine jeweilige Näherung des ventrikulären Fernfeldes für den bestimmten Abschnitt zu dem bestimmten Zeitpunkt widerspiegelt.

14. System nach einem der Ansprüche 9 bis 13, wobei das Datenverarbeitungsmodul ein Herzvorhofelektrogrammverbesserungsmodul (134) aufweist, das konfiguriert ist zum:
Subtrahieren des Beitrags des ventrikulären Fernfeldes an einer bestimmten Stelle innerhalb des Vorhofs von dem Herzvorhofelektrogramm, das durch ein elektrisches Signal dargestellt wird, das durch einen der Sensoren an der bestimmten Stelle erzeugt wird.

15. System nach einem der Ansprüche 9 bis 14, wobei das Referenzmodul (120) konfiguriert ist, um ein oder mehrere Referenzsignale aus der folgenden Liste potenzieller Referenzsignale zu bestimmen:

das Referenzsignal, das von einem oder mehreren Elektrokardiogrammsensoren empfangen wird, die mindestens die R-Welle der ventrikulären elektrischen Aktivität des Patienten messen,
das Referenzsignal, das von einem Koronarsinuskathetersensor empfangen wird,
das Referenzsignal, das aus den aufgezeichneten elektrischen Signaldaten durch eine Blindquellentrennung oder Analyse der Periodizität berechnet wird, und
das Referenzsignal, das basierend auf Informationen über die ventrikuläre Kontraktion bestimmt wird, die durch Verwenden von optischen Techniken wie Laserinterferometrie, Pulsoximetrie oder Nahinfrarot-Spektroskopie erhalten werden.

Revendications

1. Procédé mis en oeuvre par ordinateur (1000) permettant de déterminer une contribution de champ lointain ventriculaire dans des électrogrammes auriculaires d'un patient, le procédé comprenant :

   la réception (1100) d'une pluralité de signaux électriques enregistrés précédemment (ECGm, Fm, Gm, Hm, Bm, Cm, Dm) mesurés par une pluralité de capteurs (S1 à Sn), dans lequel la pluralité de signaux électriques se rapporte à une pluralité d'emplacements dans une oreillette (500) du patient ;
   la détermination (1200) d'un signal de référence (RS, I, II, CS) mesurant l'excitation électrique des ventricules du patient ;
   la sélection (1300) à partir de la pluralité des signaux électriques reçus de tels signaux électriques qui sont enregistrés sous une ou plusieurs des conditions suivantes en vérifiant chacune des trois conditions :

   - les signaux respectifs sont enregistrés à des emplacements à l'intérieur de l'oreillette (500) où le capteur respectif n'a pas de contact avec le tissu auriculaire,
   - les signaux respectifs sont enregistrés, indépendamment de l'emplacement du capteur, pendant des intervalles de temps où la partie respective de l'oreillette ne montre aucune activité électrique,
   - les signaux respectifs sont enregistrés, indépendamment de l'emplacement du capteur, pendant des intervalles de temps qui comprennent une pluralité d'intervalles de battement cardiaque et sont soumis à un lissage spatial ultérieur ; et

   la détermination (1400) d'une distribution spatio-temporelle du champ lointain ventriculaire à l'intérieur de l'oreillette par une approximation de la distribution spatio-temporelle sur la base des données de signal des signaux sélectionnés à l'aide d'un modèle d'approximation, dans lequel le modèle d'approximation est calculé sur la base des données de signal sélectionnées et permet une interpolation et une extrapolation de la contribution de champ lointain ventriculaire à des emplacements où aucune donnée de capteur n'a été enregistrée.

2. Procédé selon la revendication 1, dans lequel le modèle d'approximation est sélectionné parmi l'un quelconque des modèles suivants : un modèle spatial linéaire, un modèle spatial non linéaire, un modèle temporel et une table de recherche.

3. Procédé selon la revendication 1, dans lequel l'approximation utilise l'une quelconque des procédés suivants : approximation polynomiale, approximation avec un modèle de source dipolaire, approximation avec une combinaison linéaire spatio-temporelle des données de signal enregistrées, approximation effectuée avec des fonctions de base radiales, approximation à l'aide de tables de recherche, et approximation effectuée en transformant les données de signal en une grille régulière avant de les fournir sous forme d'une table de recherche.

4. Procédé selon l'une quelconque des revendications précédentes, dans lequel la pluralité de signaux reçus comprend des données de signal enregistrées sur plusieurs périodes d'excitation électrique ventriculaire, le procédé comprenant en outre :
   la synchronisation (1350) des données de signal sélectionnées avec les événements d'excitation électrique mesurés (E1, E2), dans lequel la détermination (1400) de la distribution spatio-temporelle est basée sur les données de signal synchronisées (601, 602, 603).

5. Procédé selon l'une quelconque des revendications précédentes, dans lequel la détermination (1400) de la distribution spatio-temporelle du champ lointain ventriculaire comprend en outre :
   la génération d'une pluralité de modèles, dans lequel chaque modèle se rapporte à une section particulière à l'intérieur de l'oreillette en combinaison avec un point temporel particulier par rapport à l'événement d'excitation

électrique (E1, E2) et reflète une approximation respective du champ lointain ventriculaire pour la section particulière au point temporel particulier.

6. Procédé selon l'une quelconque des revendications précédentes, comprenant en outre : la soustraction (1600) de la contribution du champ lointain ventriculaire à un emplacement particulier à l'intérieur de l'oreillette à partir de l'électrogramme auriculaire représenté par des données de signal électrique mesurées par un capteur particulier (S1 à Sn) à l'emplacement particulier (LSa, LSb, LSc, LS1, LS2, LS3).

7. Procédé selon l'une quelconque des revendications précédentes, dans lequel la détermination (1200) d'un signal de référence est effectuée par l'une quelconque des procédés suivants :

la réception du signal de référence à partir d'un ou de plusieurs capteurs électrocardiogrammes mesurant au moins l'onde R de l'activité électrique ventriculaire du patient,
la réception du signal de référence à partir d'un capteur de cathéter de sinus coronaire,
le calcul du signal de référence à partir des données de signal électrique enregistrées par séparation aveugle de sources ou par analyse de périodicité, et
la détermination du signal de référence sur la base d'informations concernant la contraction ventriculaire obtenue à l'aide de techniques optiques comme interférométrie laser, oxymétrie de pouls ou spectroscopie à infrarouge proche.

8. Produit programme informatique pour déterminer la contribution de champ lointain ventriculaire dans des électrogrammes auriculaires d'un patient, le produit programme informatique, lorsqu'il est chargé dans la mémoire d'un dispositif informatique et exécuté par au moins un processeur du dispositif informatique, exécute les étapes du procédé mis en oeuvre par ordinateur selon l'une quelconque des revendications précédentes.

9. Système informatique (100) pour déterminer une contribution de champ lointain ventriculaire dans des électrogrammes auriculaires d'un patient, le système informatique comprenant :

un module d'interface (110) configuré pour recevoir une pluralité de signaux électriques générés par une pluralité de capteurs, dans lequel la pluralité de signaux électriques se rapportent à une pluralité d'emplacements (LSa, LSb, LSc, LS1, LS2, LS3) dans une oreillette (500) du patient ;
un module de référence (120) configuré pour déterminer un signal de référence (RS, I, II, CS) reflétant une excitation électrique des ventricules du patient ;
un module de traitement de données (130) configuré pour sélectionner parmi la pluralité des signaux électriques reçus de tels signaux électriques qui sont enregistrés sous une ou plusieurs des conditions suivantes en vérifiant chacune des trois conditions :

- les signaux respectifs sont enregistrés à des emplacements à l'intérieur de l'oreillette (500) où le capteur respectif n'a pas de contact avec le tissu auriculaire,
- les signaux respectifs sont enregistrés, indépendamment de l'emplacement du capteur, pendant des intervalles de temps où la partie respective de l'oreillette ne montre aucune activité électrique,
- les signaux respectifs sont enregistrés, indépendamment de l'emplacement du capteur, pendant des intervalles de temps qui comprennent une pluralité d'intervalles de battement cardiaque et sont soumis à un lissage spatial ultérieur ; et

configuré en outre pour déterminer une distribution spatio-temporelle du champ lointain ventriculaire à l'intérieur de l'oreillette par une approximation de la distribution spatio-temporelle (VFFc) sur la base des données de signal des signaux sélectionnés à l'aide d'un modèle d'approximation, dans lequel le modèle d'approximation est calculé sur la base des données de signal sélectionnées et permet une interpolation et une extrapolation de la contribution de champ lointain ventriculaire à des emplacements où aucune donnée de capteur n'a été enregistrée.

10. Système selon la revendication 9, dans lequel le modèle d'approximation est sélectionné parmi l'un quelconque des modèles suivants : un modèle spatial linéaire, un modèle spatial non linéaire, un modèle temporel et une table de recherche.

11. Système selon la revendication 9, dans lequel l'approximation utilise l'une quelconque des procédés suivants : approximation polynomiale, approximation avec un modèle de source dipolaire, approximation avec une combinaison linéaire spatio-temporelle des données de signal enregistrées, approximation effectuée avec des fonctions de base radiales, approximation à l'aide de tables de recherche, et approximation effectuée en transformant les données de signal en une grille régulière avant de les fournir sous forme d'une table de recherche.

12. Système selon l'une quelconque des revendications 9 à 11, dans lequel la pluralité de signaux reçus comprend des données de signal enregistrées sur de multiples périodes d'excitation électrique ventriculaire, le module de traitement de données (130) étant en outre configuré pour :
la synchronisation des données de signal sélectionnées avec les événements d'excitation électrique mesurés (E1, E2), dans lequel la détermination de la distribution spatio-temporelle du champ lointain ventriculaire est basée sur les données de signal synchronisées.

13. Système selon l'une quelconque des revendications 9 à 12, dans lequel le module de traitement de données a un module générateur de modèle (135) configuré pour :
générer une pluralité de modèles, dans lequel chaque modèle se rapporte à une section particulière à l'intérieur de l'oreillette en combinaison avec un point temporel particulier par rapport à l'événement d'excitation électrique et reflète une approximation respective du champ lointain ventriculaire pour la section particulière au point temporel particulier.

14. Système selon l'une quelconque des revendications 9 à 13, dans lequel le module de traitement de données a un module d'amélioration d'électrogramme auriculaire (134) configuré pour :
soustraire la contribution du champ lointain ventriculaire à un emplacement particulier à l'intérieur de l'oreillette à partir de l'électrogramme auriculaire représenté par un signal électrique généré par l'un des capteurs à l'emplacement particulier.

15. Système selon l'une quelconque des revendications 9 à 14, dans lequel le module de référence (120) est configuré pour déterminer un ou plusieurs signaux de référence à partir de la liste suivante de signaux de référence potentiels :

le signal de référence reçu à partir d'un ou de plusieurs capteurs électrocardiogrammes mesurant au moins l'onde R de l'activité électrique ventriculaire du patient,
le signal de référence reçu à partir d'un capteur de cathéter de sinus coronaire,
le signal de référence calculé à partir des données de signal électrique enregistrées par séparation aveugle de sources ou par analyse de périodicité, et
le signal de référence déterminé sur la base d'informations concernant la contraction ventriculaire obtenue à l'aide de techniques optiques comme interférométrie laser, oxymétrie de pouls ou spectroscopie à infrarouge proche.

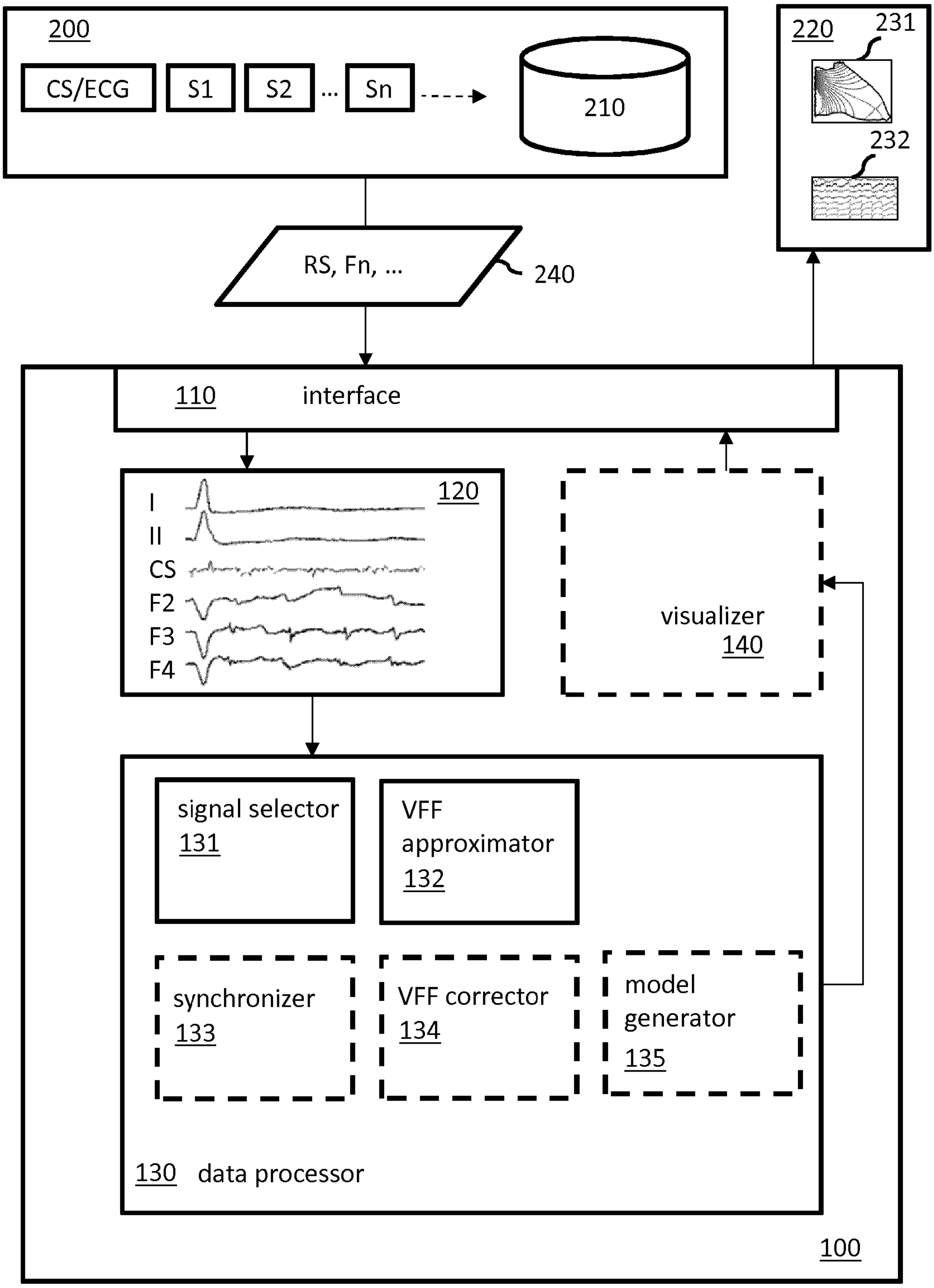
200
CS/ECG
S1
S2
...
Sn
210
220
231
232
RS, Fn, ...
240
110
interface
120
I
II
CS
F2
F3
F4
visualizer
140
signal selector
131
VFF
approximator
132
synchronizer
133
VFF corrector
134
model
generator
135
130
data processor
100

FIG. 1

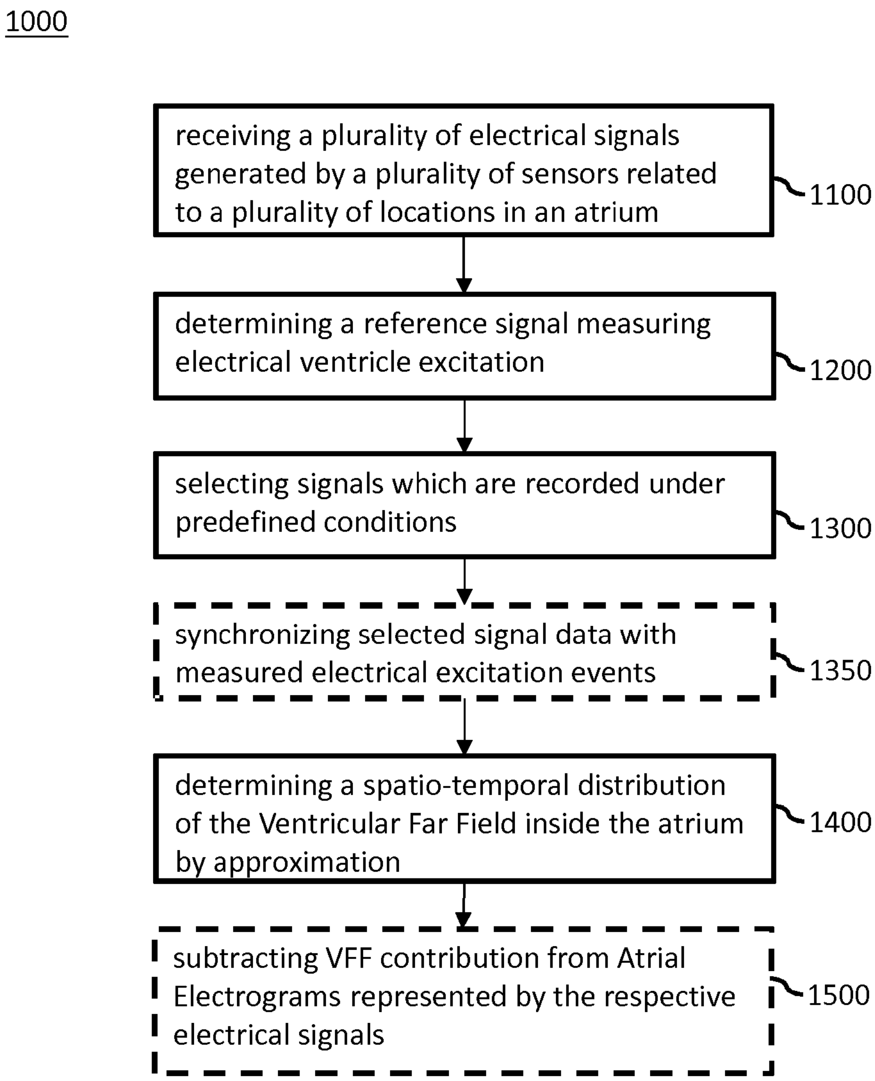
1000
receiving a plurality of electrical signals generated by a plurality of sensors related to a plurality of locations in an atrium
1100
determining a reference signal measuring electrical ventricle excitation
1200
selecting signals which are recorded under predefined conditions
1300
synchronizing selected signal data with measured electrical excitation events
1350
determining a spatio-temporal distribution of the Ventricular Far Field inside the atrium by approximation
1400
subtracting VFF contribution from Atrial Electrograms represented by the respective electrical signals
1500

FIG. 2

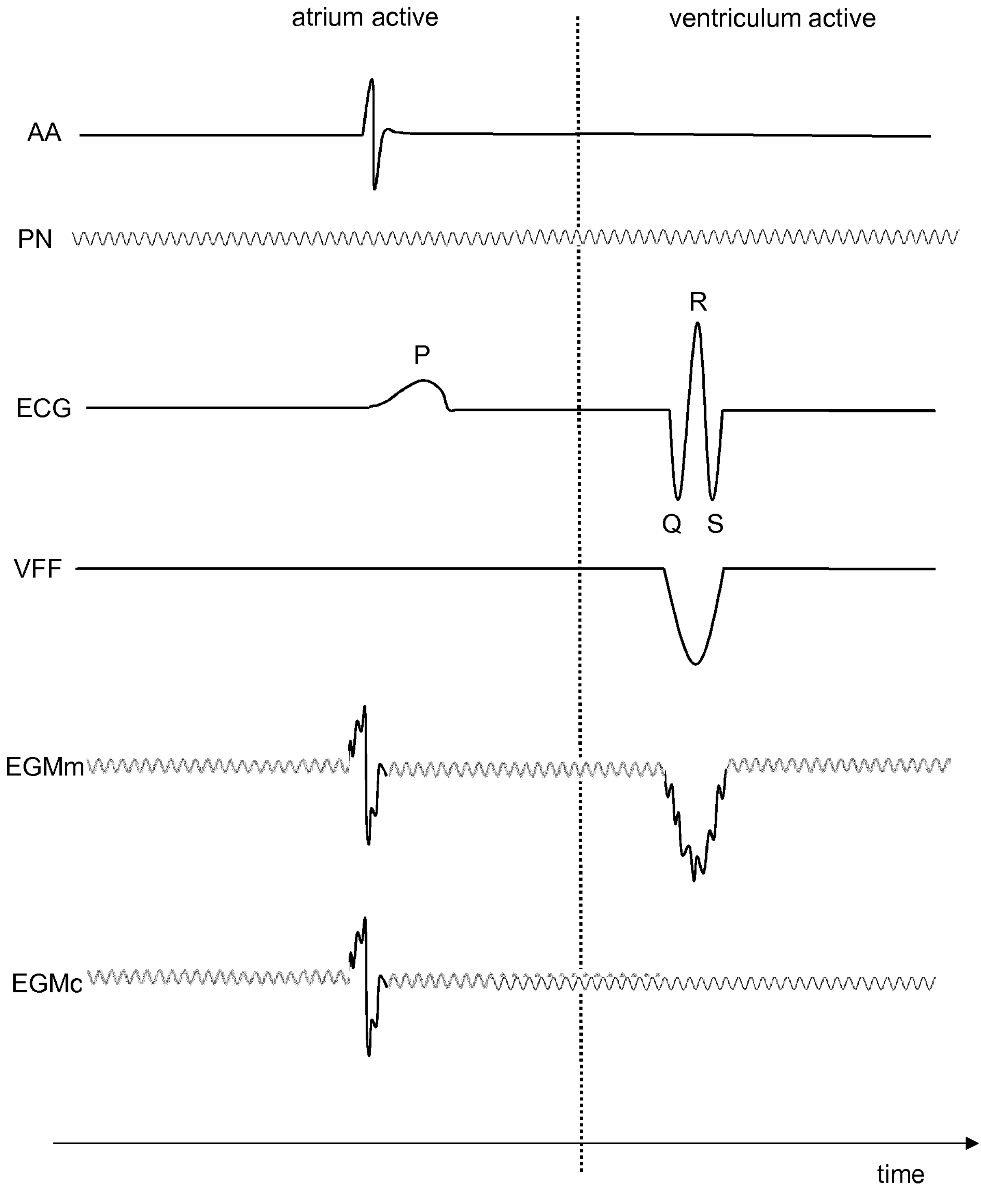
atrium active
ventriculum active
AA
PN
R
P
ECG
Q
S
VFF
EGMm
EGMc
time

FIG. 3A

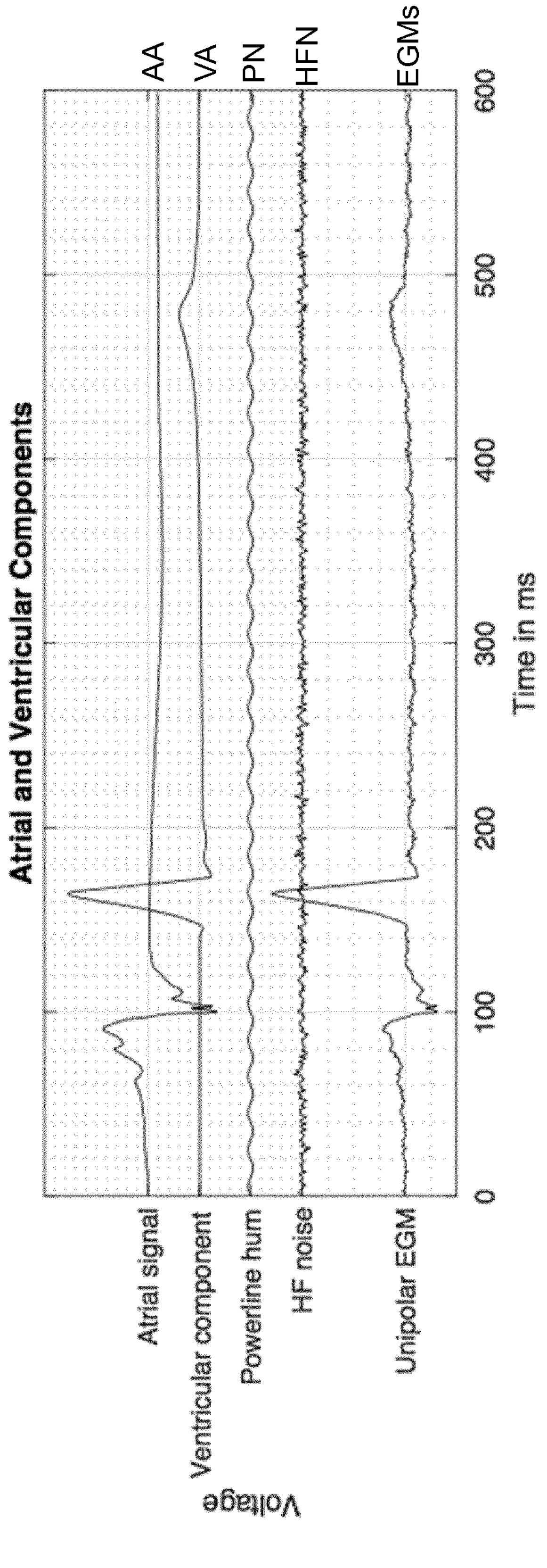
Atrial and Ventricular Components
Voltage
Atrial signal
Ventricular component
Powerline hum
HF noise
Unipolar EGM
AA
VA
PN
HFN
EGMs
0
100
200
300
400
500
600
Time in ms

FIG. 3B

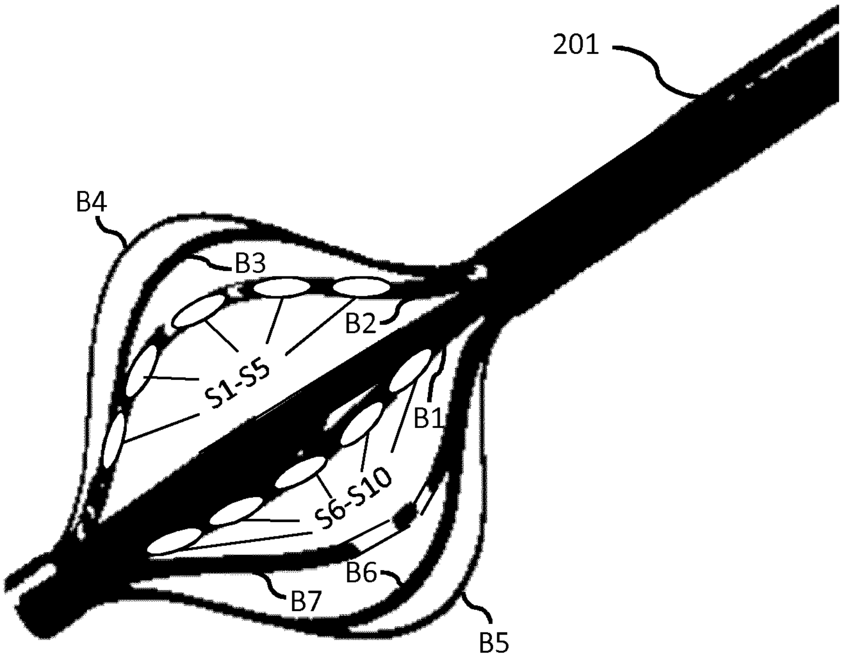

FIG. 4

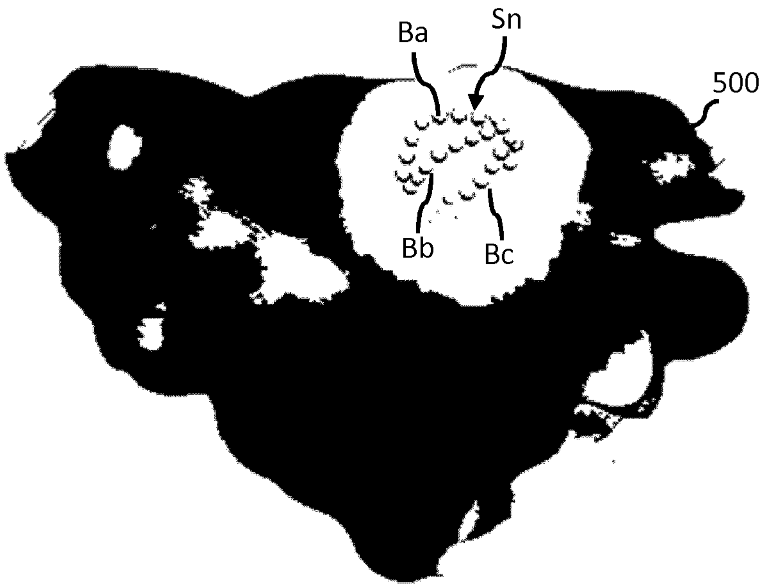

FIG. 5

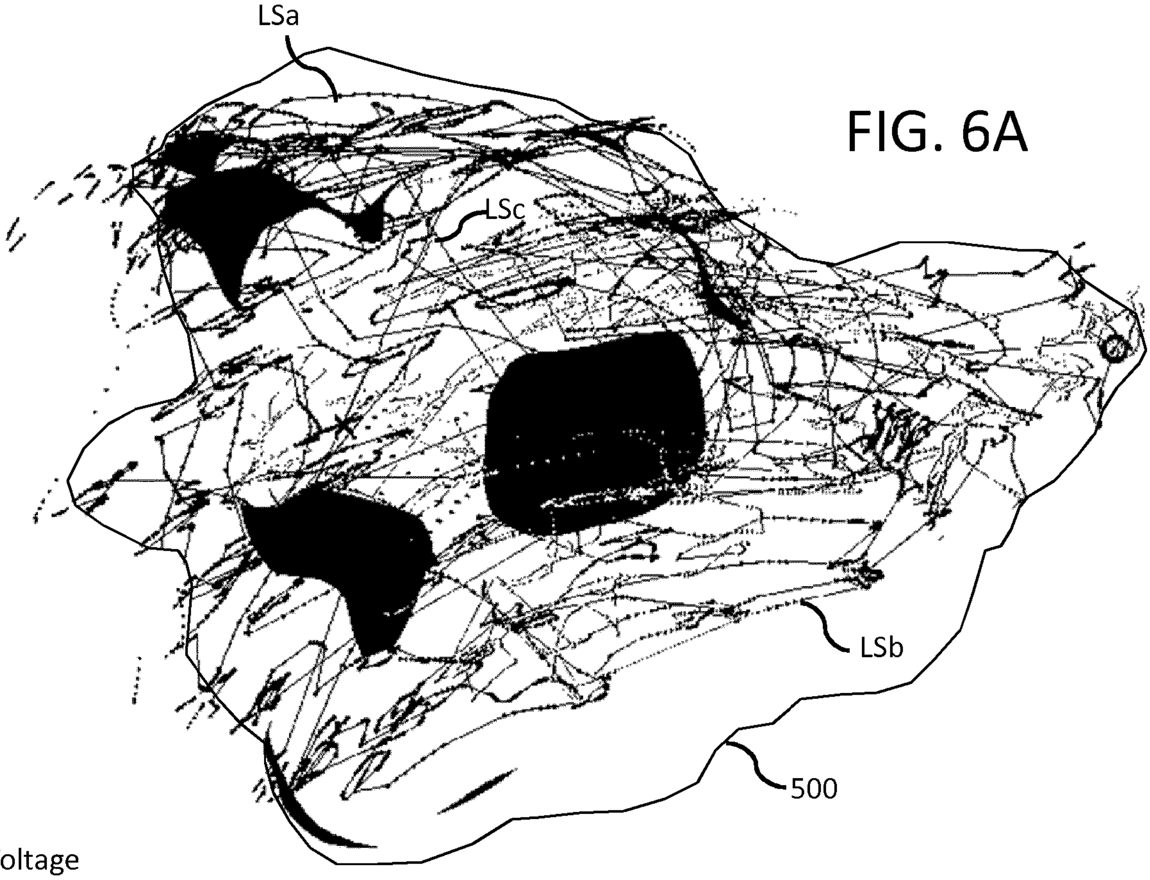
LSa
LSc
LSb
500

FIG. 6A

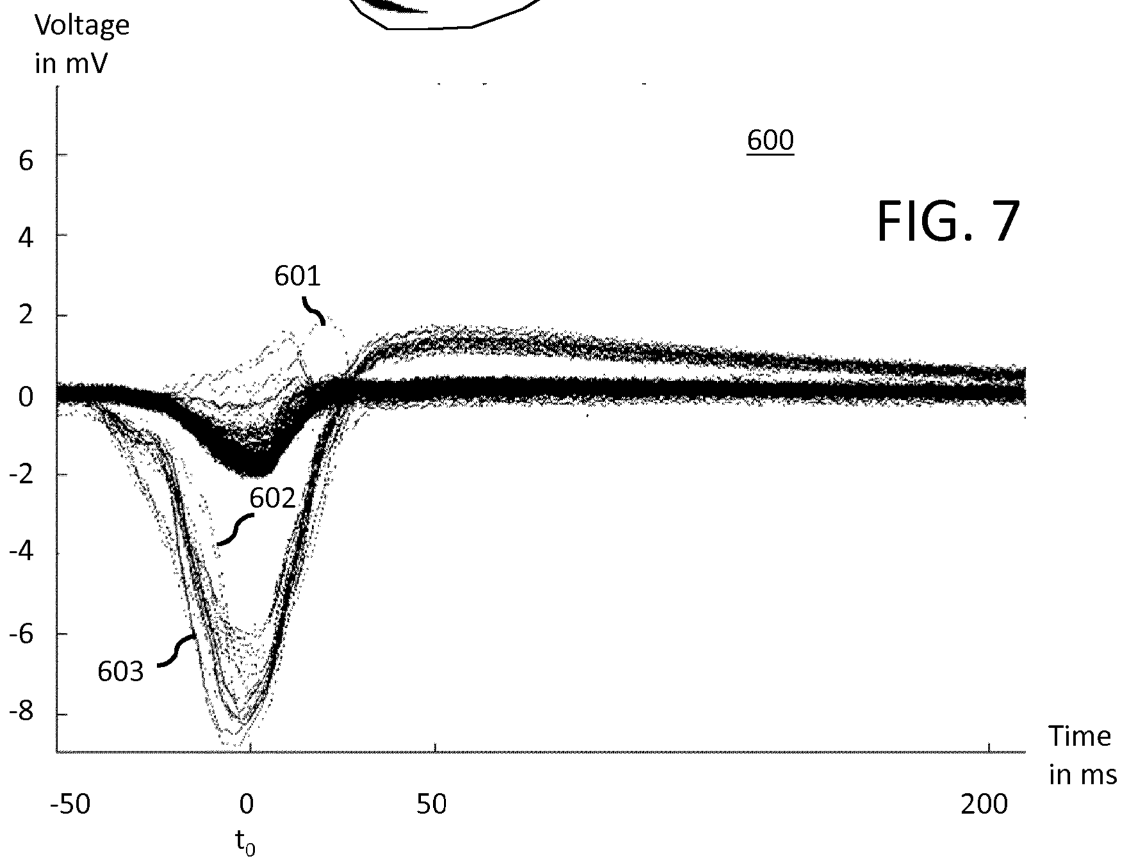
Voltage
in mV
6
4
2
0
-2
-4
-6
-8
600
601
602
603
Time
in ms
-50
0
50
200
t0

FIG. 7

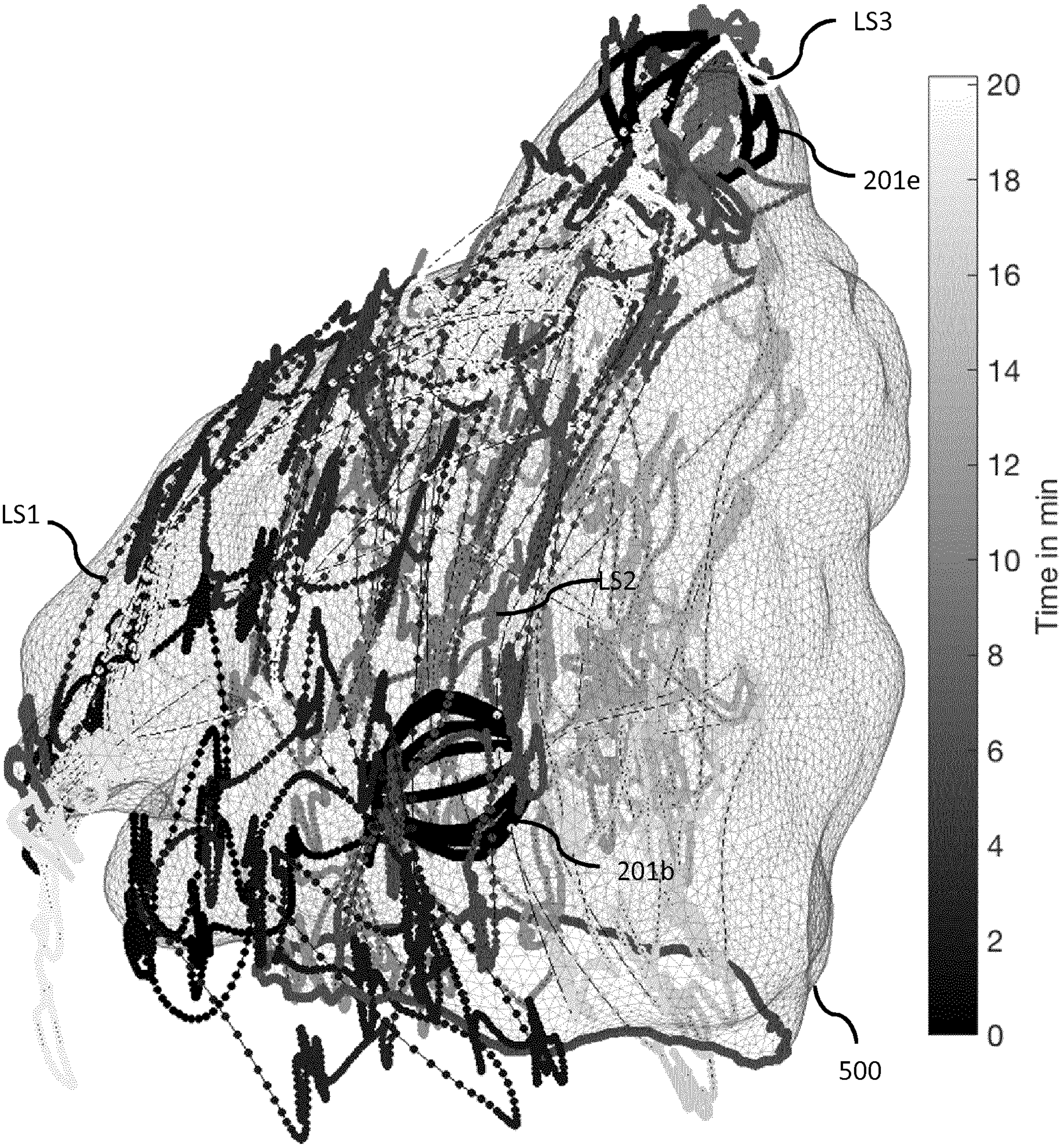

FIG. 6B

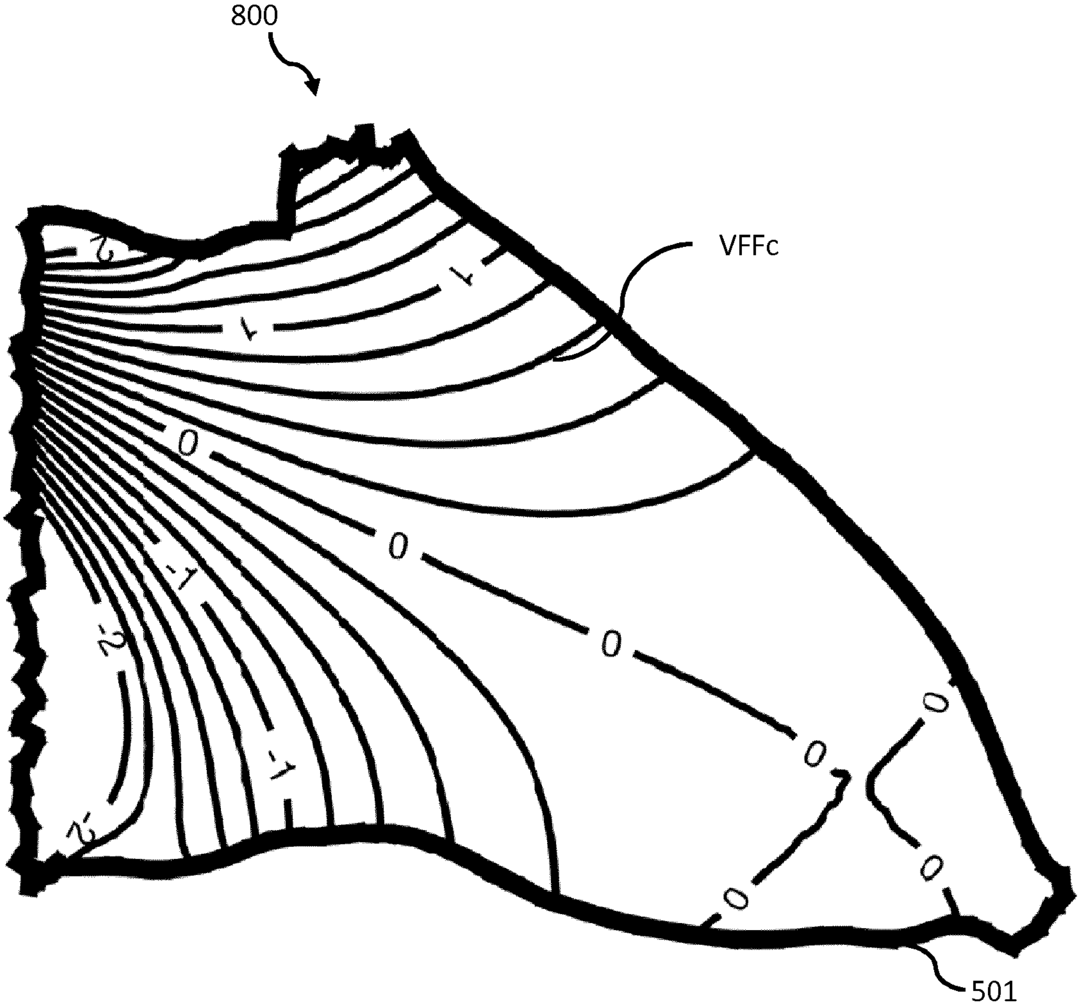

FIG. 8

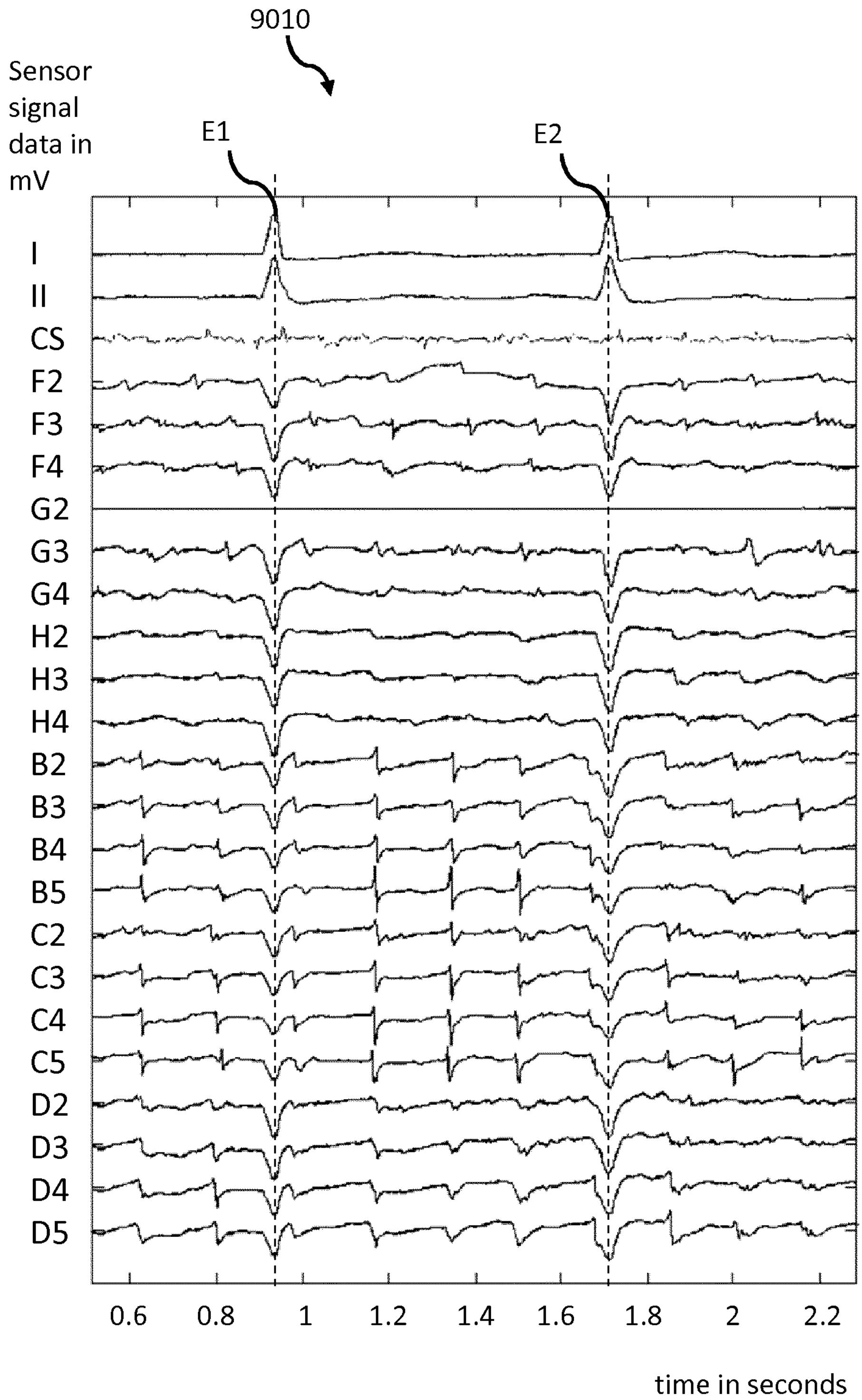
9010
Sensor signal data in mV
E1
E2
I
II
CS
F2
F3
F4
G2
G3
G4
H2
H3
H4
B2
B3
B4
B5
C2
C3
C4
C5
D2
D3
D4
D5
0.6
0.8
1
1.2
1.4
1.6
1.8
2
2.2
time in seconds

FIG. 9A

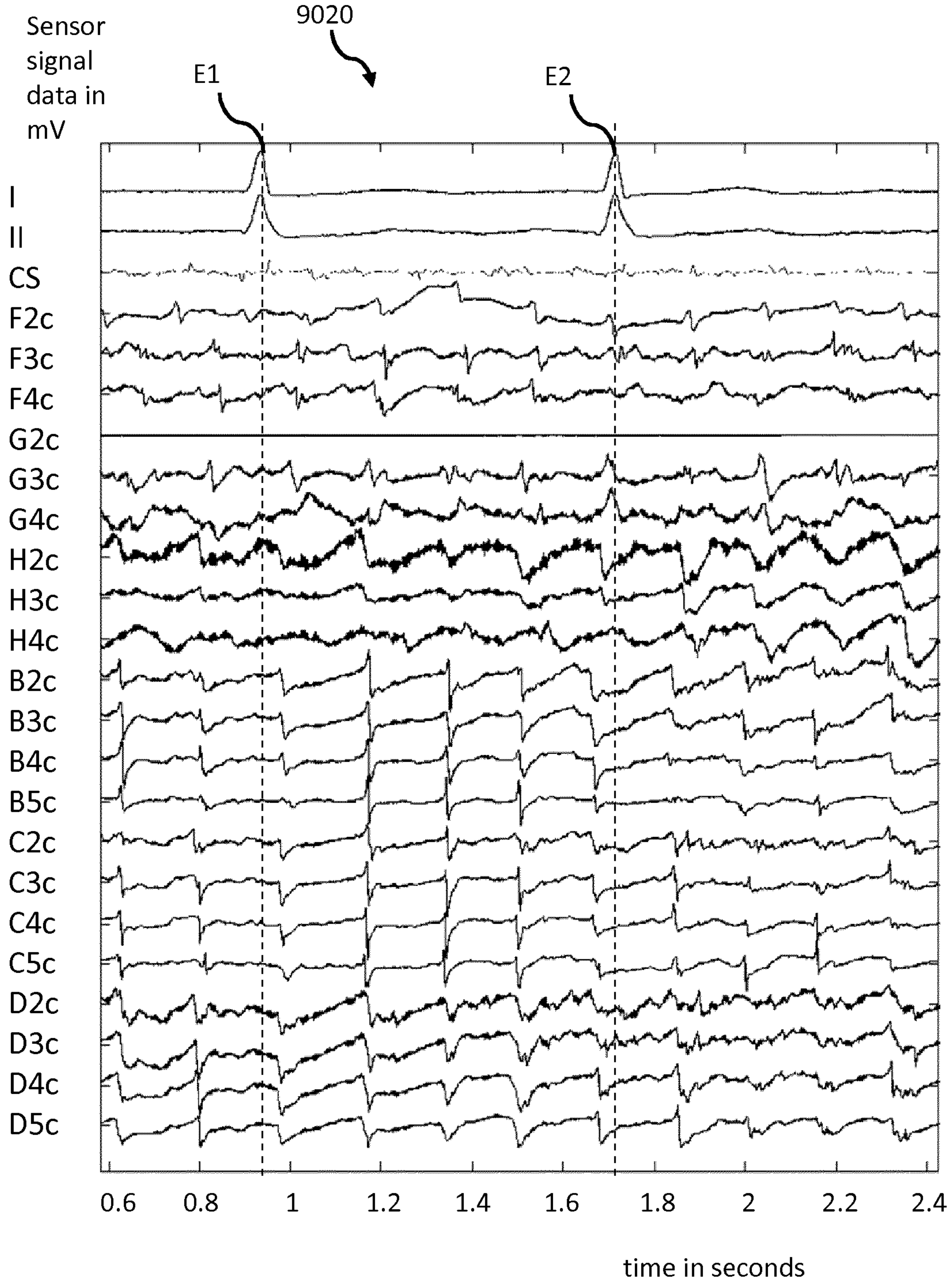
Sensor signal data in mV
9020
E1
E2
I
II
CS
F2c
F3c
F4c
G2c
G3c
G4c
H2c
H3c
H4c
B2c
B3c
B4c
B5c
C2c
C3c
C4c
C5c
D2c
D3c
D4c
D5c
0.6
0.8
1
1.2
1.4
1.6
1.8
2
2.2
2.4
time in seconds

FIG. 9B

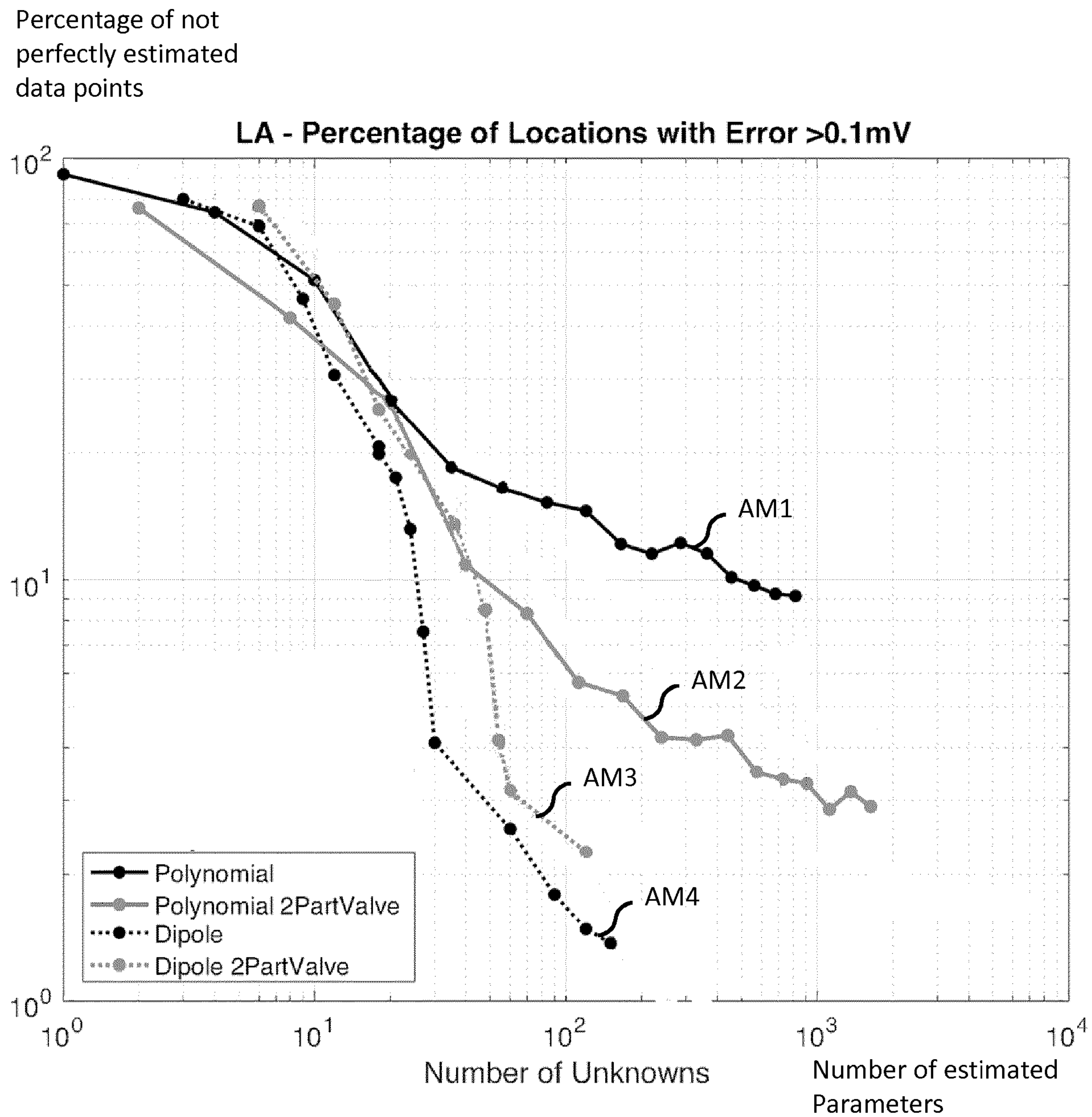
Percentage of not perfectly estimated data points
LA - Percentage of Locations with Error >0.1mV
10^2
10^1
10^0
10^0
10^1
10^2
10^3
10^4
AM1
AM2
AM3
AM4
Polynomial
Polynomial 2PartValve
Dipole
Dipole 2PartValve
Number of Unknowns
Number of estimated Parameters

FIG. 10

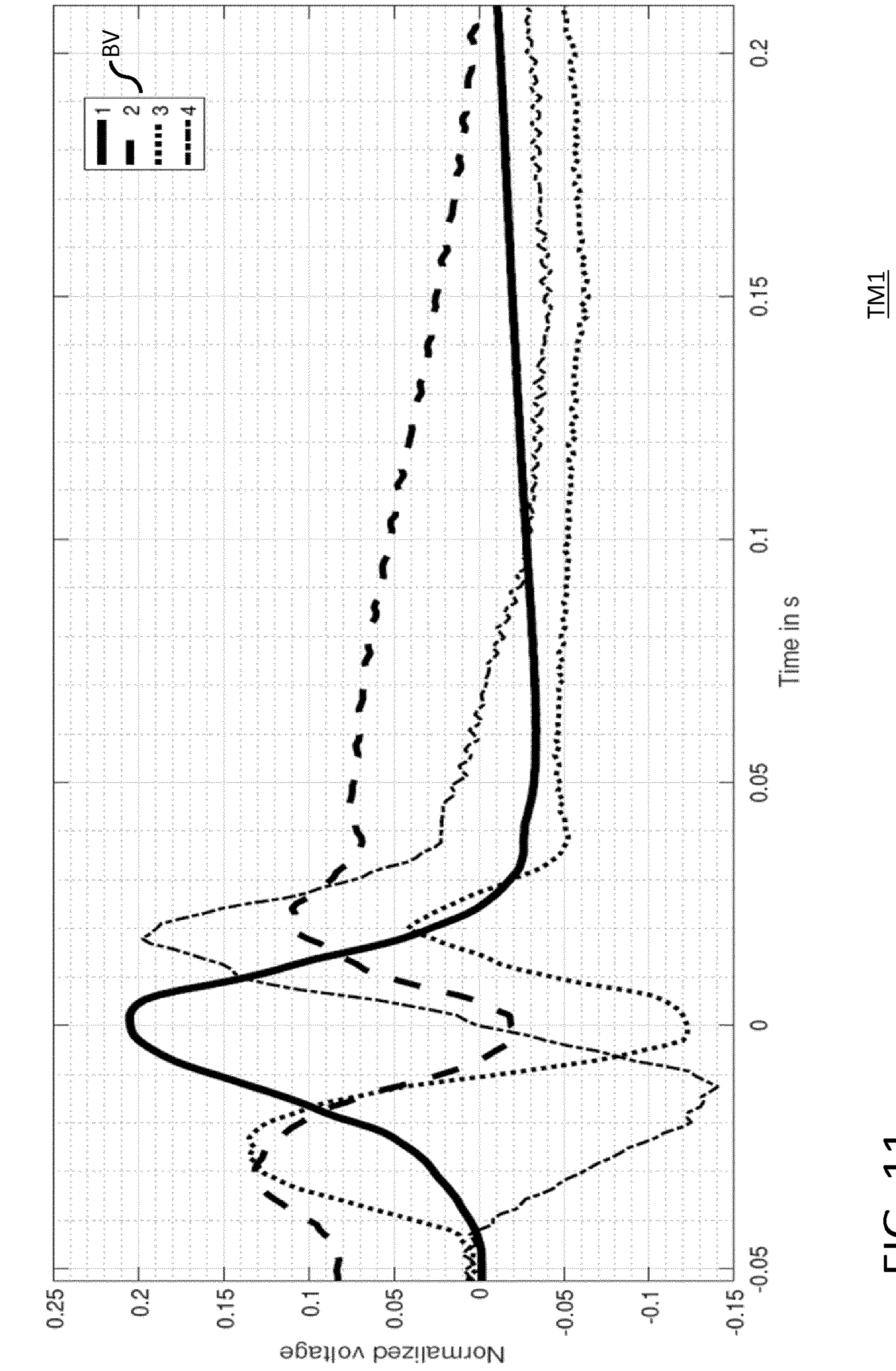
Normalized voltage
0.25
0.2
0.15
0.1
0.05
0
-0.05
-0.1
-0.15
-0.05
0
0.05
0.1
0.15
0.2
Time in s
1
2
3
4
BV
TM1

FIG. 11

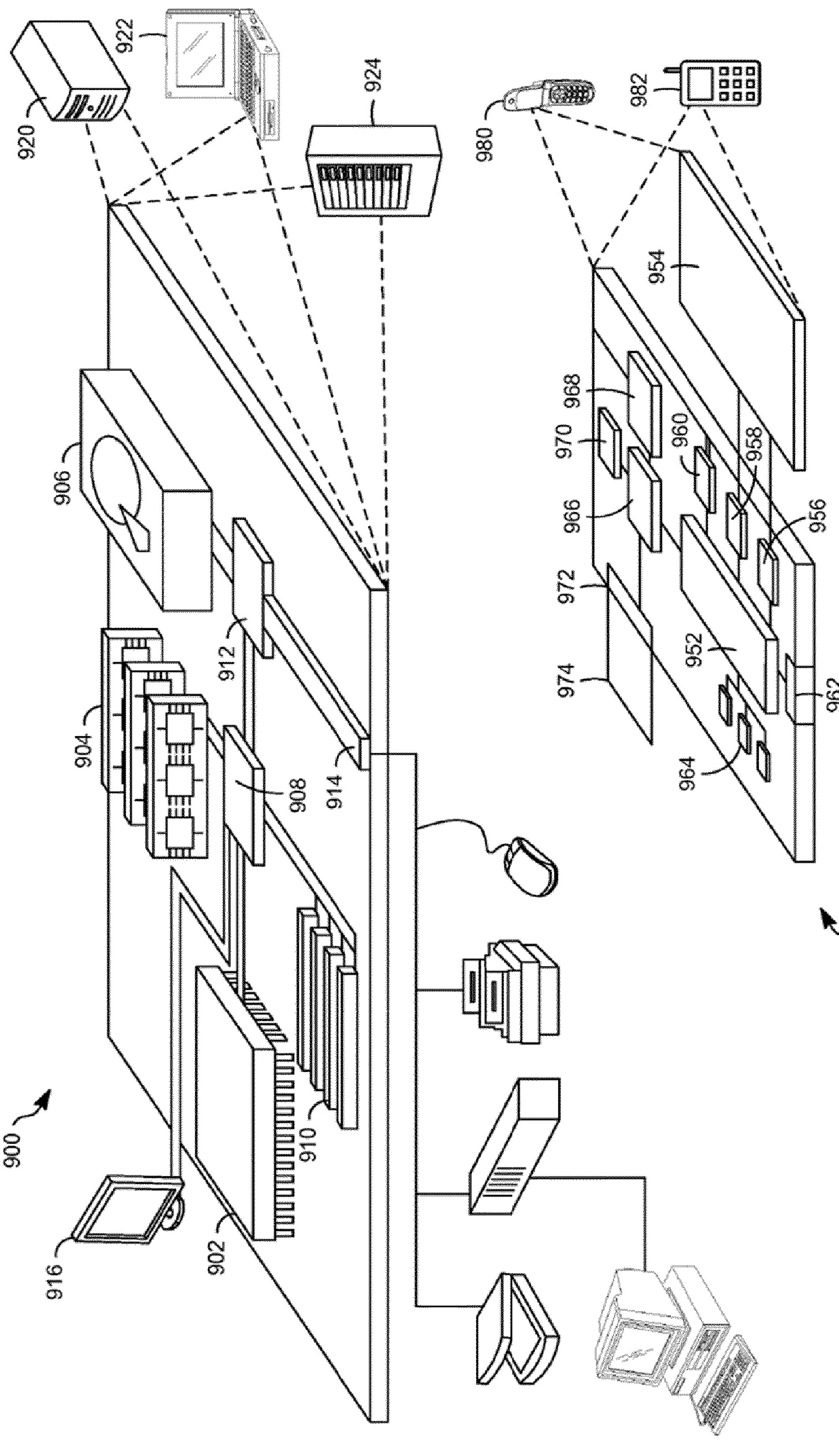
900
902
904
906
908
910
912
914
916
920
922
924
950
952
954
956
958
960
962
964
966
968
970
972
974
980
982

FIG. 12

REFERENCES CITED IN THE DESCRIPTION

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- US 20170100049 A **[0007]**

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- **GUSTAVO LENIS et al.** Orthogonal Component Analysis to Remove Ventricular Far Field in Non Periodic Sustained Atrial Flutter. *Computing in Cardiology Conference (CinC),* 2015 **[0007]**
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