(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) **World Intellectual Property Organization**
International Bureau

(43) **International Publication Date**
09 March 2023 (09.03.2023)

WIPO | PCT

(10) **International Publication Number**
WO 2023/034387 A1

(51) **International Patent Classification:**
A61B 5/053 (2006.01) *A61B 5/024* (2006.01)
A61B 5/0537 (2021.01) *A61B 5/00* (2006.01)

(21) **International Application Number:**
PCT/US2022/042165

(22) **International Filing Date:**
31 August 2022 (31.08.2022)

(25) **Filing Language:** English

(26) **Publication Language:** English

(30) **Priority Data:**
63/239,085 31 August 2021 (31.08.2021) US

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(81) **Designated States** *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CV, CZ, DE, DJ, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IQ, IR, IS, IT, JM, JO, JP, KE, KG, KH, KN, KP, KR, KW, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, ST, SV, SY, TH,

(54) **Title:** BIO ELECTRIC IMPEDANCE MONITORS, ELECTRODE ARRAYS AND METHOD OF USE

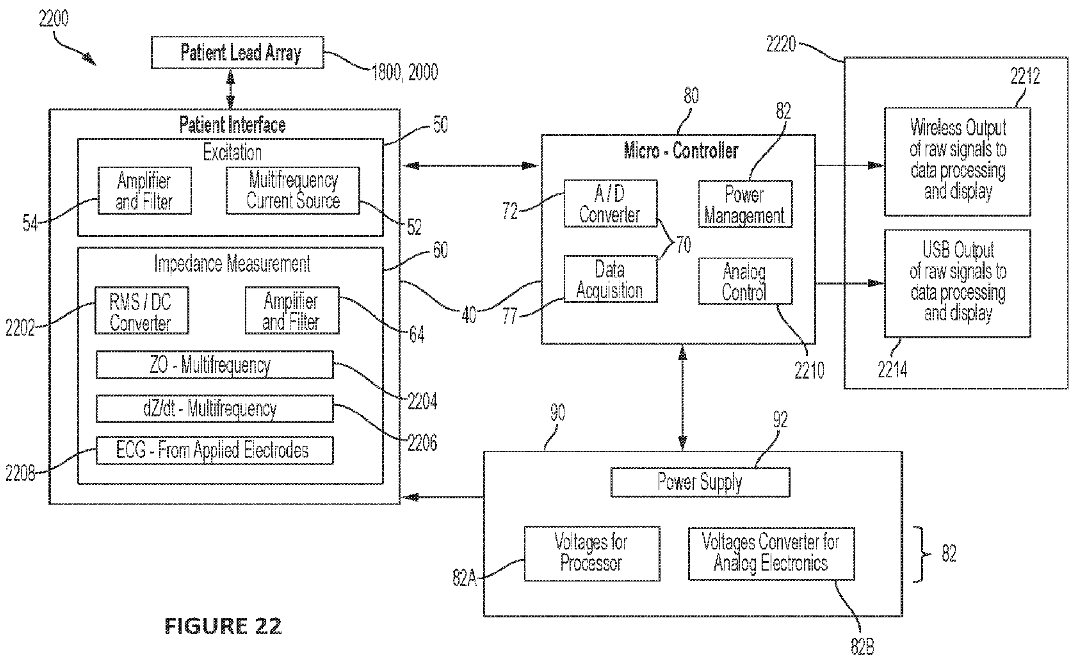

FIGURE 22

(57) **Abstract:** A portable bioelectric impedance monitor and methods using the monitor can measure and monitor extracellular fluid levels and/or cardiac signals. The monitor may include a tetrapolar electrode array lead with four electrodes arranged sequentially and axially along the lead, and circuitry coupled with the at least four electrodes configured to measure bioelectric impedance extracellular fluid and/or cardiac signals in a human subject at various frequencies. The electrodes are adhered to a human subject/patient on the patient's torso or one of the patient's limbs. One embodiment includes a Tetrapolar Analog Front End Patient Interface circuit configured to convert two electrode operation of a commercial Impedance Converter, Network Analyzer into a tetrapolar operation for excitation and impedance measurement of the human subject.

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[Continued on next page]

TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, WS, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, ST, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:

— *with international search report (Art. 21(3))*

BIO ELECTRIC IMPEDANCE MONITORS, ELECTRODE ARRAYS AND METHOD OF USE

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RELATED APPLICATIONS

[0001] This application claims the benefit under 35 USC 119(e) to U.S. Provisional Application No. 63/239,085, filed August 31, 2021, and which is incorporated herein by reference.

FIELD

[0002] The disclosure relates to a bioelectric device for measuring impedance.

BACKGROUND

[0003] It is known in the art to measure human impedance to monitor levels of intrathoracic fluids, such as blood. In particular, it is known to use an impedance monitor to measure human thoracic impedance, along with electrocardiogram (EKG) signals, as indicative of blood flow and heart performance characteristics, as described in U.S. Patent No. 5,443,073 (Wang et al.), the subject matter of which is incorporated by reference herein in its entirety. A portable device for non-invasive thoracic impedance measurement for the determination of Stroke Volume (SV) and Cardiac Output (CO) is described in U.S. Patent No. 7,474,918. The relatively small and simple, portable, non-invasive device for bioelectric impedance measurement described in U.S. Patent No. 7,474,918 was superior to numerous prior invasive and non-invasive thoracic impedance measurement devices and methods detailed in that patent that is also incorporated herein by reference.

[0004] It is further known that certain medical conditions, such as congestive heart failure (CHF) or renal disease, correlate qualitatively with the level and variation of the level of intrathoracic fluids.

[0005] It would further be useful to be able to monitor levels of tissue hydration in a human subject in real/near real time, in particular, extracellular fluid (ECF) levels, to gauge the subject's response to various interventions, for example, kidney dialysis.

BRIEF DESCRIPTION OF THE DRAWINGS

[0006] The foregoing summary, as well as the following detailed description of the disclosure, will be better understood when read in conjunction with the appended drawings. For the purpose of illustrations, there are shown in the drawings embodiments which are presently preferred. It should be understood, however, that the disclosure is not limited to the precise arrangements and instrumentalities shown. In the drawings:

[0007] Figure 1 depicts a first embodiment of a fluid impedance monitor connected to a user for thoracic impedance measurement;

[0008] Figure 2 is a front perspective view of the front face of a base unit of the monitor of Figure l;

[0009] Figure 3A is a perspective view of an electrode array assembly of the impedance monitor of Figure 1 used on a first preferred embodiment;

[0010] Figure 3B is a plan view of a first side of an electrode pad assembly of the electrode array assembly of Figure 3A;

[0011] Figure 3C is a plan view of a second side of an electrode pad assembly of the electrode array assembly of Figure 3A;

[0012] Figure 3D is a plan view showing the electrode pad assembly of Figure 3B separated from first and second electrodes of the electrode array assembly of Figure 3A;

[0013] Figure 3E is a perspective view of components of an electrode array assembly in accordance with a second preferred embodiment;

[0014] Figure 4 is a block diagram of the major circuit components of the existing prior art thoracic impedance monitor base unit;

[0015] Figure 5 is a diagram of steps of a method of monitoring thoracic fluid level of a person;

[0016] Figures 6A-6D are flow diagrams describing in detail operation of the existing base unit;

[0017] Figure 7 is a functional block diagram of the circuit components of the base unit for Extracellular Fluid (ECF) patient monitoring;

[0018] Figure 8 is a functional block diagram of the circuit components of the base unit further modified from Figure 7 to selectively provide for Extracellular Fluid (ECF) or conventional thoracic impedance patient monitoring, as desired;

[0019] Figure 9 is a functional block diagram of the circuit components of the base unit further modified from Figure 8 to incorporate an Impedance Converter and Network circuit operating through a Tetrapolar Front End Patient Interface;

[0020] Figure 10 is a functional block diagram of the components of the Impedance Converter and Network circuit used with Tetrapolar Front End Patient Interface of Figure 9;

[0021] Figure 11 is a functional block diagram of the components of the Tetrapolar Front End Patient Interface of Figure 9 used with the Impedance Converter and Network circuit of Figs. 9 and 10;

[0022] Figure 12 illustrates one possible use of any of the subject systems to measure Extracellular Fluid in a human subject;

[0023] Figures 13-16 are detailed circuit diagrams for an embodiment of the Tetrapolar Analog Front End Patient Interface of Figure 11;

[0024] Figures 17A and 17B are a front view and back view, respectively of a second embodiment of the fluid impedance monitor;

[0025] Figure 18 illustrates a first implementation of a single electrode lead array for the monitor of Figures 17A and 17B;

[0026] Figures 19A and 19B illustrate more details of the single electrode lead array shown in Figure 18;

[0027] Figure 20 illustrates an implementation of a dual electrode lead array for the monitor of Figures 17A and 17B;

[0028] Figures 21A and 21B illustrate more details of the dual electrode lead array shown in Figure 20;

[0029] Figure 21C illustrates the points on a cardiac cycle whose timing may be measured using the ESG signals in this second embodiment of the fluid monitor;

[0030] Figure 22 illustrates a circuit board of the monitor housed in the monitor housing shown in Figures 17A-17B;

[0031] Figure 23 illustrates an output signal generator circuit that may be used for the second embodiment of the fluid monitor;

[0032] Figure 24 illustrates an input signal conditioning circuit that may be used for the second embodiment of the fluid monitor;

[0033] Figure 25 illustrates a differentiator circuit that may be used for the second embodiment of the fluid monitor;

[0034] Figure 26 is a flowchart of a method for measuring impedance of a patient;

[0035] Figure 27 illustrates more details of the method for measuring impedance of a patient;

[0036] Figure 28 is a flowchart of a method for determining cardiac characteristics based on bioimpedance;

[0037] Figure 29 is a flowchart of a method for estimating heart rate;

[0038] Figure 30 is a flowchart of a method for determining cardio cycles;

[0039] Figure 31 is a flowchart of a method for constructing a vector for each cardio cycle; and

[0040] Figure 32 is a flowchart of a method for determining stroke volume.

DETAILED DESCRIPTION OF ONE OR MORE EMBODIMENTS

[0041] According to one aspect of the disclosure, a device to monitor tissue hydration of a human subject comprises: at least four electrodes capable of being physically adhered and electrically coupled to the human subject; and circuitry coupled to four of the at least four electrodes to measure a bioelectric tissue impedance of the patient at a frequency of less than fifteen kilohertz. (< 15kHz) or 115 kHz. Furthermore, the device and method may measure ECG data from the heart of a patient and used that ECG data as discussed below.

[0042] In one embodiment, the device may facilitate a method for monitoring tissue hydration of a patient using at least four electrodes using multiple frequencies. The method may be used to monitor low frequencies, less than 15 kHz, to determine extracellular hydration and/or high frequencies, such as greater than 15 kHz up to 115 kHz, to determine extracellular and intracellular hydration.

[0043] In another embodiment, a method of using an electrode array and operably coupled impedance measuring device that comprises an electrode array for use with a physiological electronic monitor used to monitor electrical characteristics of a user's body may be provided. The method using a linear electrode array lead including at least first, second, third, and fourth electrodes arranged sequentially and axially along the linear electrode array lead (examples of which are shown in Figures 18-21), applying a sinusoidal current from the impedance measuring device to the first and fourth electrodes, detecting a differential electrical potential between the second and third electrodes with the impedance measuring device and determining the impedance of the user from the detected differential electrical potential. In another embodiment, a method is disclosed to calculate a numerical value of determined impedance on the impedance measuring device, wherein the detecting process differentially amplifies and low pass filters voltages from the second and third electrodes and wherein the calculating process includes sampling the differentially amplified and low pass filtered voltage from the second and third electrodes at predetermined intervals for a number of times, adding the sampled voltages to generate a sum, dividing the sum by the number of times to provide an averaged voltage value; scaling the averaged voltage value and combining the scaled averaged voltage value with a predetermined offset value to generate a numerical value of the differential impedance. In another embodiment, the connecting of the electrodes to the patient with an electrode pad includes adhering the electrode array to the user so as to operatively couple each of the first, second, third and fourth electrode pads to the user.

[0044] In another embodiment, a method for processing a Bioimpedance signal of a patient to derive heart rate, heart stroke volume, and cardiac output signals for the patient is provided. The method digitally filters and phase corrects the Bioimpedance signal to remove gain-phase-frequency distortions, estimates heart rate using a power spectrum of the Bioimpedance signal and an auto-convolution function of the said power spectrum, suppresses breath waves to remove undesired power spectra components and generate a Bioimpedance signal of restored shape, determines one or more cardio cycles of the restored Bioimpedance signal and determines effective left ventricular ejection time (ELVET) using check points within said cardio cycles; and discarding at least some of said cardio cycles which exhibit interference artifacts. The method may also locate points on a time-derivative Bioimpedance curve for the Bioimpedance signal; and select the points which most accurately reflect cardiac events.

[0045] In another aspect, a method of estimating heart rate is provided that calculates a power spectrum of a Bioimpedance signal (from a Bioimpedance monitor device), multiplies the power spectrum by a selected amplitude-frequency function to differentiate the signal and suppress breath harmonics, auto convolutes the resulting power spectrum according to a formula and determines a maximum amplitude value of auto convolution in a predefined frequency range as an estimation of heart rate.

[0046] In another aspect, a method of determining cardio cycles is provided that filters a Bioimpedance signal from a Bioimpedance monitor to emphasize fronts (a beginning of each cardio cycle) of cardio cycles, calculates a time-amplitude envelope of the cardio cycles by analyzing the first five harmonics of the powers spectrum of said Bioimpedance signal after filtration, selects the cardio cycle fronts by comparison with said calculated time-amplitude envelope and rejects erroneously-detected fronts. The method for discarding cardio cycles exhibiting interference artifacts may also detect time and amplitude relations referencing check points within individuals of a plurality of cardio cycles, compare the time and amplitude relations between individuals of a plurality of cardio cycles and further examine selected cardio cycles which exhibit the presence of artifact according to a plurality of comparison criteria.

[0047] In another aspect, a method of constructing a multi-dimensional vector for each selected cardio cycle is provided by comparing the multi-dimensional vector with such vectors for other cardio cycles and rejecting the cardio cycles with vectors having no neighboring vectors of other cardio cycles. In another aspect, a method of determining effective left ventricular ejection time (ELVET) is provided that filters the Bioimpedance signal (from a Bioimpedance monitor device) and suppresses breath waves therein, detects a cardio cycle, calculates the time derivative of the Bioimpedance signal, determines the maximum value of the time derivative, determines effective ejection start time, determines effective ejection end time and calculates effective left ventricular ejection time (ELVET) as change in time between effective ejection start time and end time.

[0048] In another aspect, a method of determining stroke volume is provided that, using the Bioimpedance monitor, determines specific blood resistivity (P), measures a distance L between two Bioimpedance electrodes applied to the patient, determines a base thoracic impedance Z, determines effective left ventricular ejection time (ELVET) and calculates stroke volume SV according to the equation where K is a novel scale factor related to body composition of the patient. A method of determining cardiac output as a product of stroke volume and heart rate is also disclosed.

[0049] A process for monitoring human subjects such as medical patients is disclosed that applies electrodes to points on the body of the subject/patient, passes an alternating current of very low amperage between a first pair of the electrodes, measures voltages (V) of the body through a second pair of the electrodes located on the subject/patient between the first pair, calculates an average impedance value Zo that may be determined by first measuring the thoracic voltage (V) of the body and then calculating an average thoracic (base) impedance value, Zo, based on the measured current (I) and the thoracic voltage (V) and displays the average impedance value for comparison with baseline values previously established preferably when the subject/patient was in a known, stable condition, to determine if differences are within established tolerances. The disclosed methods may be performed using a fluid monitor that has a battery powered, portable base unit which performs all the necessary functions. For example, the device and system disclosed in U.S. Patent No. 7,474,918 (thoracic impedance monitor) may be used.

[0050] In another aspect, a method is disclosed for operating the aforesaid device to monitor extracellular fluid status of a human subject in which the method connects four of the electrodes in a linear arrangement to the skin of the human subject; generates an oscillating voltage signal having a frequency or multiple frequencies between less than 15 kHz and up to 115 kHz; removes a direct current (DC) bias from the oscillating voltage signal; converts the oscillating voltage signal into an oscillating current having a frequency of between less than 15 kHz and up to 115 kHz; passes the oscillating current through the human subject between a first pair of the electrodes; samples voltages from the human subject through a second pair of electrodes positioned between the first pair of electrodes on the human subject; generates a differential voltage signal from sampled voltages; converts the differential voltage signal into an alternating current; adds, to the alternating current, a constant bias equivalent to the DC bias removed from the oscillating voltage signal to provide a current output; and determines from the current output one or more biometric impedance values for the human subject.

[0051] In yet another aspect, a method is disclosed that monitors the extracellular fluid status of a human subject that adheres, to skin of the human subject, four spaced apart electrodes in a linear array; passes an oscillating current having a frequency of between less than 15 kHz and up to 115 kHz through the patient between an outermost pair of the four electrodes; senses voltage levels from the human subject through an innermost pair of the four electrodes; calculates a bioelectric impedance value for the human subject from the sensed voltage levels; and outputs the calculated biometric impedance value to a human interface device.

[0052] A second embodiment of the monitor device monitors tissue hydration (like the first embodiment) and hemodynamics variables of a human subject using at least four electrodes capable of being physically adhered and electrically coupled to the human subject and circuitry coupled to four of the at least four electrodes to measure a bioelectric tissue impedance of the patient at a frequency of less than two hundred kilohertz.

[0053] The process for monitoring human subjects such as medical patients is disclosed that applies electrodes to points on the body of the subject/patient, passes an alternating current of very low amperage between a first pair of the electrodes, measures voltages (V) of the body through a second pair of the electrodes located on the subject/patient between the first pair, calculates an average impedance value Zo based on the applied current (I) and measured voltages (V) and displays the average impedance value for comparison with baseline values previously established preferably when the subject/patient was in a known, stable condition, to determine if differences are within established tolerances

[0054] The EKG signal typically displays electro cardio events as perturbations referred to as waves. The heartbeat is most clearly reflected in the EKG signal as an R wavepeak between a pair of adjoining Q and S wave valleys. A process for monitoring human subjects such as medical patients is disclosed that applies electrodes to points on the body of the subject/patient and interfaces to a signal conditioning circuit for ECG biopotential measurement , designed to extract, amplify, and filter small bio-potential signals in the presence of noisy conditions, such as those created by motion or remote electrode placement. This design allows for an ultralow power analog-to-digital converter (ADC) or an embedded microcontroller to acquire the output signal.

<u>First Embodiment of Monitor Device</u>

[0055] Figure 1 depicts an embodiment of a monitoring system or "monitor" 10 that is connected to, by electrodes, a user U. The system 10 measures thoracic impedance using a patient interface that includes a four electrode array assembly 100 which is coupled to a small, hand transportable base unit 20.

[0056] Figure 2 depicts a front panel 24 of the base unit 20. The base unit 20 is relatively lightweight and is contained in a relatively small (i.e. hand transportable) housing 22. Preferably, the base unit 20 is provided with a handle 36 to facilitate transport. The base unit 20 contains several user interfaces in addition to a connector port 26 for receiving a connection end of the electrode array assembly 100. These interfaces preferably include a three digit display 30 (e.g. formed by three, seven segment LED's) which preferably digitally display impedance as xx.x ohms, a start switch 28 to start the system 10, a low battery alert light 32, and a cable disconnect alert light 34. Preferably the base unit 20 also contains a beeper 86 (see FIGURE 4) or other sound generator for signaling purposes. Alternatively, additional alert lights (not illustrated) could be substituted for the beeper 86.

[0057] The base unit 20 is preferably configured to perform all necessary steps to measure, determine and display the patient's base impedance after the start switch 28 is actuated. However, the system 10 does not provide any patient diagnostic parameters. That is, it provides only a measurement of impedance over a predetermined fixed length of the patient's body. This value can be compared with other impedance values for the patient or against limit values and used as a relative measure of patient's "dryness" or "level of hydration". An analogy will be a blood pressure instrument which displays patient's systolic and diastolic blood pressure, but does not diagnose if a patient has hypertension or not. The information provided by system 10 will be evaluated along with various other parameters by health care or other professional to identify the use of the information for their specific purpose.

[0058] The base unit 20 will provide the following outputs. The three digit LED display 30 preferably will display impedance value as xx.x. During measurement, a rotating/flickering pattern can be displayed to indicate the measurement is in progress. To ensure that the user U records ONLY the impedance values, the system software preferably will not display any numerical values other than impedance value. This means that there should be no countdown timers and no error or diagnostic codes expressed as numerical values. The base unit 20 will also indicate an error condition (by the beeper or flashing lights) in the event it detects that it could not perform a valid impedance measurement or that the impedance value was outside of a predetermined measurement range (such as, for example, 5 to 55 ohms). If the electrode array assembly is disconnected from the system cable disconnect alert light 34 will illuminate.

[0059] The base unit 20 may activate the low battery indication light 32 in the event it detects that the battery voltage is below a level that will allow for reliable impedance measurement. In the event of a low battery voltage condition, the base unit 20 may blink this LED 32, for example at a rate of once every 10 (+/-0.5) seconds for a period of 30 (+/-2) sec. If the battery voltage drops below 5.25 volts, but remains above 4.75 volts, the impedance results will be displayed along with blinking-battery condition LED 32 to indicate that the battery power is getting low but still acceptable. If the battery voltage drops below 4.75 volts, both LEDs 32, 34 can be made to blink to indicate that the battery voltage is low and accurate results could not be displayed. Preferably a micro-controller 80 in the base unit 20 will continue to operate below 4.75 volts, even though an accurate measurement cannot be made, to warn the user of the condition of the unit.

[0060] The base unit 20 can be configured to provide various beeper alerts to the user. Preferably the base unit 20 beeps to indicate that the measurement is completed and the displayed value should be recorded. The beeper 86 may further be activated to indicate other, different conditions or steps, for example, when the base unit 20 is initially activated, while the unit is initializing, while the power supply is stabilizing, while measurements are being taken and/or before the unit shuts itself off The beeper 86 can also be activated in the event a successful measurement was not accomplished or an error condition was detected. Different beep patterns may be used for different conditions including different states of the base unit 20.

[0061] The base unit 20 was configured to perform all necessary steps to measure, determine and display the patient's base thoracic impedance after the start switch 28 is actuated. However, the system 10 did not provide any patient diagnostic parameters. That is, it provided only a measurement of impedance over a predetermined fixed length of the patient's body. This value can be compared with other impedance values for the patient or against limit values. The information provided by system 10 would be evaluated along with various other parameters by health care or other professional to identify the use of the information for their specific purpose.

[0062] The base unit 20 provided the following outputs. The three digit LED display 30 preferably displayed impedance value as xx.x. During measurement, a rotating/flickering pattern was displayed to indicate the measurement is in progress. To ensure that the user U records ONLY the impedance values, the system software preferably did not display any numerical values other than impedance value. This means that there were no countdown timers and no error or diagnostic codes expressed as numerical values. The base unit 20 would also indicate an error condition (by the beeper or flashing lights) in the event it detects that it could not perform a valid impedance measurement or that the impedance value was outside of a predetermined measurement range (such as, for example, 5 to 55 ohms). If the electrode array assembly was disconnected from the system cable disconnect alert light 34 would illuminate.

[0063] The base unit 20 would activate the low battery indication light 32 in the event it detected that the battery voltage is below a level that will allow for reliable impedance measurement. In the event of a low battery voltage condition, the base unit 20 might blink this LED 32, for example at a rate of once every 10 (+/- 0.5) seconds for a period of 30 (+/-

2) sec. If the battery voltage dropped below 5.25 volts, but remains above 4.75 volts, the impedance results would still be displayed along with blinking battery condition LED 32 to indicate that the battery power was getting low but still acceptable. If the battery voltage dropped below 4.75 volts, both LEDs 32, 34 would be made to blink to indicate that the battery voltage was low and accurate results could not be displayed. Preferably a microcontroller 80 in the base unit 20 would continue to operate below 4.75 volts, even though an accurate measurement could not be made, to warn the user of the condition of the unit.

[0064] The base unit 20 could be configured to provide various beeper alerts to the user. Preferably the base unit 20 beeped to indicate that the measurement is completed and the displayed value should be recorded. The beeper 86 could further be activated to indicate other, different conditions or steps, for example, when the base unit 20 was initially activated, while the unit was initializing, while the power supply was stabilizing, while measurements were being taken and/or before the unit shut itself off The beeper 86 could also be activated in the event a successful measurement was not accomplished or an error condition was detected. It was suggested that different beep patterns could be used for different conditions including different states of the base unit 20.

[0065] Referring to Figures 3A-3D, a first preferred embodiment of the electrode array assembly 100 included a single, linear electrode array lead 110 having a first end 112 and a second end 114. An electrical connector 116 is provided at the first end 112. Electrical connector 116 operatively connects to connector port 26. First through fourth electrodes 120, 122, 124, and 126 are arranged axially and spaced along the length of the lead 110. As discussed further below, preferably first and fourth electrodes 120, 126 are current sources, while preferably second and third electrodes 122, 124 measure electrical potential. Because the electrodes 120-126 are fixed along the lead 110, their spacing relative to one another is also fixed and predetermined, with the first and second electrodes 120, 122 being spaced a first pre-determined distance DI, and the third and fourth electrodes 124, 126 being spaced an equal pre-determined distance D2. The pre-determined distances DI, D2 were preferably about five centimeters or about two inches.

[0066] Preferably, identical first and second electrode pad assemblies 140 were releasably connected to the electrodes 120-126. The preferred electrode pad assemblies included an overlapped arrow-shaped body member 142 into which were mounted a first electrode pad 146 and a second electrode pad 150. The body member 142 had a first side 142a, and the electrode pads 146, 150 were exposed on this first side 142a. On a second side 142b of the body member, male snap elements 152, rigidly connected to the electrode pads 146, 150, are exposed. The male snap elements 152 were adapted to releasably connect with complementary female snap elements 128 provided in the electrodes 120-126 on the lead 110. Any other conventional structure used for coupling electrode pads to such cardio leads could also be used.

[0067] Preferably, the body member 142 was pre-coated during manufacture with a contact adhesive on the first side 142a. A removable, adhesive protective film 144 was preferably provided. Preferably, the electrode pads 146, 150 were coated with an electrically conductive hydrogel which acted along with the contact adhesive and allowed the electrode pads 146, 150 to releasably adhere to the user's skin. The electrodes 120-126 and electrode pads 146, 150 incorporated into the electrode array assembly 110 were off-the-shelf commercially available components.

[0068] Referring to Figure 3E, a second embodiment electrode array assembly 100' was generally similar to the first embodiment electrode array assembly 100, with the exception that a second embodiment electrode array lead 11 O' was substantially shorter, and a connection cord 130 was provided to connect the electrode array lead 110' to the base unit 20. The connection cord 130 had a first end 132, a second end 134, a first connector 136 at the first end 132 configured to mate with array lead connector 116, and a second connector 138 at the second end 134 configured to mate with base unit connector port 26. Note that electrode pad assemblies 140 are omitted from the illustration of Figure 3E, but conventional conductive pads were used as part of the second embodiment electrode array assembly 100'.

[0069] Each of the array leads 110, 110' was flexible along its length. While the spacing between the first and second electrodes 120, 122 and between the third and fourth electrodes 124, 126 with the electrodes 120-126 operatively connected to a user was preferably the same for all users, given the flexibility of the array lead 110, 110', the spacing between the second electrode 122 and the third electrode 124 could be adjusted to accommodate users of various sizes. That is, for a user having a long sternum, with the electrodes 120-126 placed as indicated above, the electrode array lead 110, 11 0' will be more fully extended between the second and third electrodes 122, 124 than would be the case for a user having a shorter sternum and also having the electrodes 120-126 placed as indicated above.

[0070] Figure 4 illustrates an example of the hardware and circuitry 40 of the base unit 20 that may include signal generating circuitry 50, voltage detection circuitry 60 and impedance calculation circuitry 70. The impedance calculation circuitry 70 includes an analog/digital converter 72, data acquisition circuitry 74, and data analysis and storage circuitry 76. Along with power management circuitry 82, the impedance calculation circuitry 70 is provided by a micro-controller 80.

[0071] The signal generating circuitry 50 generates the stable excitation current (I). A current source subcircuit 52 includes a constant current source (not depicted) and clock oscillator (not depicted) to supply a current of about 2 mA or less, preferably a.98.+-.0.0l mA, at a 100.+-. l 0 kHz and 5+/- 5 kHz (frequency preferably to the first and fourth electrodes 120, 126 through an isolation transformer 54, the connection cord 130 and electrode array lead 110. The current source subcircuit 52 is configured to output a current of less than 4 mA under all conditions including equipment component failure. The wave form of the current may be sinusoidal with less than ten percent total harmonic distortion. Voltage values across two of the four electrodes, preferably the second and third electrodes 122, 124, are passed through isolation transformer 62 to an amplifier and low pass filter subcircuit 64. The low pass filter subcircuit 64 functions to remove extraneous electrical interference from ambient sources, for example, home appliances operating on standard residential 60 Hz current. A preferred cut-off frequency of the low pass filter subcircuit 64 is about 50 Hz. The base unit 20 measures voltage developed across detection electrodes 122, 124 when the excitation current source is energized. The voltage level will be between about 18 millivolts and 104 millivolts (to provide an anticipated range of impedance measurement of about 10 ohms to 50 ohms, at the 2 mA current).

[0072] Micro-controller 80, which might be a PIC 16F873 device, controls generation of the excitation current and receives the filtered voltage analog signal from the amplifier and low pass filter 64 at the input of analog to digital converter 72. In one embodiment, the injected current is not generated for a short period of time (e.g. fifteen to thirty seconds) after the start switch 28 is actuated to allow the user to settle into a quiescent state. The current may be then injected for a predetermined period, e.g. thirty seconds, to perform the measurement. Voltage values sampled from the A/D converter 72 are received by the data acquisition circuitry 74 of the micro-controller 80 at a rate of about five samples per second for all or most of the thirty second period. Data analysis and storage circuitry 76 of micro-controller 80 sums the counts generated by the A/D converter 72, divides sum by the total number of samples taken to provide an average voltage value which is converted into an impedance value. The algorithm used for generating impedance in tenths of ohms is: averaged A/D counts*Gain+Offset, where in the preferred circuit the Gain is 0.6112 and the Offset is 1.1074. Gain and Offset are based on the electronics design and operating range and are used for all base units 20. Each system 10 is calibrated to match the use of these numbers. The data analysis circuitry 76 also controls the various displays 30, 32, and 34. The power management circuitry 82 controls the generation and distribution of power in the base unit circuitry 40 to control operation of the system 10. Specific functions of the power management circuitry 82 include a first function 82a of providing power to the processor; a second function 82b of providing power to the A/D converter, and a third function 82c of monitoring the input voltage. A power supply 90 may be provided by conventional dry-cell batteries (not shown) or by an external power adapter (not shown) connected to a conventional 120 V outlet.

[0073] The base unit 20 may be provided with a serial port 84 to work with logic level signals. The timing for the serial data can be similar to RS232 signal or other conventional data transfer format. The base unit 20 would preferably be provided with a serial port, for example one configured to operate at 9600 baud, with 8 bit data, 1 Start bit, 1 Stop bit and no parity bit format. An external level translator may be necessary to interface the base unit to a PC or a PALM device. Upon receipt of a specific command, the base 20 unit would be configured to transmit the information related to all or a subset (e.g. the last ten) of the readings of the impedance measurement. This information may also include the date and time of measurement, impedance value, and/or the serial number of the unit.

[0074] With reference to Figure 5, a method of monitoring thoracic fluid level of a person included a first step 210 of providing the thoracic impedance monitor 10, as described herein. In a second step 220, the user obtained a measurement of their thoracic impedance. To accomplish this second step 220, in a third step 230, the user connected the first through fourth electrodes 120-126, via electrode pads 146, 150, to the users' body, as described above.

[0075] With the electrodes 120-126 in place, in a fourth step 240, the user initiated operation of the impedance monitor IO by actuating the start switch 28. The user was to remain "relatively" still for the length of the measurement period. The system 10 injected the relatively high frequency (e.g. about l00KHz) very low amperage (about 2 or less mA) current into the user and took voltage readings from the second and third electrodes 122, 124 for a period of time (e.g. about thirty seconds), calculated the average thoracic (base) impedance and then displayed the average value, preferably for a predetermined period (e.g. fifteen seconds to two minutes). In particular, activation of the start switch 28 initiated a series of steps 242-314. For brevity, the reader is referred to Figs. 6A-6D, which describe in detail the series of steps 242-314. In short, assuming proper functioning of the impedance monitor 10, activation of the start switch 28 culminated in display of the user's thoracic impedance (measured in ohms) on the base unit display 30. Once the reading was obtained, in a fifth step 320, it was desirable that the user log the reading into a record of impedance measurements taken over time.

[0076] Preferably, the user need use the system 10 only once a day for thoracic impedance but might take it more than once a day if needed or desired. The total time required for a test was brief, approximately five minutes. Preferably, to improve the ability to compare measurements, the measurements were to be taken at the same time of day (thoracic impedance measurements typically vary over the course of a day, as eating, drinking, and other activities affect thoracic fluid levels). More preferably, the test was performed daily before the user ate his or her first meal of the day. The test might be taken more often, for example, to monitor the effects of medication (e.g. diuretics) or exercise.

[0077] It has been found that the basic thoracic impedance monitoring device described above could be modified and used in different ways to better monitor relative fluid levels in human patient tissues. More particularly, it has been found that a relative hydration status of a human subject such as a patient can be based on the impedance values (Z) reported in ohms over different ranges of frequency measurements. Extracellular Fluid ("ECF"), sometimes referred to as Extracellular Water ("ECW"), is the fluid which surrounds cellular membranes in human tissue. Intracellular Fluid ("ICF"), sometimes referred to as Intracellular Water ("ICW"), is the fluid trapped in the cellular membranes forming human tissue. The ECF/ECW and ICF/ICW are predominately electrical resistive entities, whereas the cellular membrane, due to its lipid layer, has an isolating (capacitive) behavior. It has been found that the behavior of an injected current will be different for "low" and "high" frequencies. Low frequency currents only flow around the cells through the ECF/ECW, whereas high frequency currents will also pass through the cell membrane and the ICF/ICW. Thoracic impedance measurement is therefore a measure of the two. "Low Frequency" is hereinafter used to refer to a bioelectric impedance measuring current of a sufficiently low frequency magnitude as to flow only or essentially only through Extracellular Fluid component in the tissue of a human subject. A "Low Frequency" impedance measuring current may be less than 15kHz (<15kHz), preferably less than l0kHz and, more preferably, only about 5kHz. "High frequency" is hereinafter used to refer to an impedance measuring current of a sufficiently high frequency magnitude as to flow through or essentially through both the Intracellular (ICF) and Extracellular (ECF) fluid components in the tissue of a human subject. A "High Frequency" impedance measuring current therefore above 15 kHz (>15 kHz) and even above 50 kHz (>50 kHz) and more typically about l00kHz like that of the described U.S. Patent No. 7,474,918 device. The clinical benefit resides in the serial determination of these ECF/ECW impedance values as the patient undergoes therapeutic interventions as a gauge of the relative changes in the EC fluid volumes, for example, during dialysis treatment.

[0078] Referring to Figure 7, a modified system 410 with modified Patient Interface has been substituted for that of Figure 4 to simplify the design and to enable measurement of ECF/ECW generated impedance values. Apart from the Patient Lead Array, which remains the same, the other components of the system would again be housed in a base unit 420. In particular, a 5kHz signal source 452 has been substituted for the original 100 kHz signal source 52 and appropriate operational amplifiers with appropriate filter(s) 454, 464 have been substituted for the original isolation transformers 54, 62 and amplifier/filter 64. In addition, an RMS to DC converter IC chip 462 has been provided as a detector for front end impedance measurement to reduce the computational load on the main micro-controller 80. There has been slight revisions to characteristics of the power supplies 482a, 482b reflective of the changes to the patient interface circuitry. No changes to the software of microcontroller 80 were need to continue to drive the display 30 to output impedance values in decimal form or to operate the sound output/beeper 86.

[0079] Figure 8 represents a further modification of the Figure 7 system (or alternate revision of the original Figure 4 device). The Figure 8 system 510 permits either conventional thoracic impedance measurements with a l00kHz signal (current) source 52 and circuitry or ECF/W impedance measurements with a low frequency (5kHz) signal (current) source 452 and related excitation and measurement circuitry. Again, apart from the Patient Lead Array 100, the other components would be housed in a base unit 520. CMOS analog switch circuitry 558 was provided to control a DPDT switch via the micro-controller 580 for user selection of the current source. Existing coding of the micro-controller was modified to permit storage and use of two sets of calibration gain and offset figures, with the appropriate figures being used automatically based upon the current source selected by the user.

[0080] Figure 9 represents a further improvement to the Low Frequency system of Figure 7. Again, apart from the Patient Lead Array, the components of this system 610 are housed in a base unit 620. Here, a commercial circuit, an Analog Devices AD5933 1 MPSP, 12Bit Impedance Converter, Network Analyzer 650 is provided as a current generator, voltage receiver and impedance data processor. A functional block diagram of AD5933 Impedance Converter, Network Analyzer 650 is set forth in Figure 10. The AD5933 circuit 650 has an output or transmit stage generating and providing at VOUT, an excitation signal at a particular frequency to an external impedance (i.e. the patient/subject) to be measured. It has an input or receive stage that samples at VIN, the excitation signal after it has been passed through the external impedance. The input stage comprises a current-to-voltage amplifier, followed by a programmable gain amplifier, anti-aliasing (low pass) filter, and an analog to digital converter (ADC). Output of the ADC is passed to an on-board D SP engine with discrete Fourier transform algorithm which outputs calculated real (E) and imaginary (I) data-words. These words are passed from the AD5933 circuit to the microprocessor controller 680 for calculation of impedance (Z) values.

[0081] The AD5933 circuit 650 is used in combination with a Tetrapolar Analog Front End Patient Interface 660, a functional block diagram of which is presented in Figure 11. As can be seen, the Tetrapolar Analog Front End Patient Interface 660 is connected across the VIN, VOUT, RFB connection points of the AD5933 circuit 650. Tetrapolar Analog Front End Patient Interface 660 converts the bipolar impedance operation of the AD5933 circuit 650 circuit into tetrapolar operation. The AD5933 circuit 650 is configured to operate as a two electrode impedance measurement device and this fact limits severely the range of application of usage, e.g. applications of spectral characterization are basically discarded since the impedance measurement obtained will also contain the electrode polarization impedance as well as the electrode-skin impedance. Another important limitation of AD5933 circuit 650 is a safety issue. The voltage output VOUT contains a DC level component, a DC bias. This imbalance produces a DC voltage across the electrodes and the body, introducing DC current into the body of any human subject on which it might be used, which can be a health hazard for the subject. The AD5933 circuit 650 is a voltage-driven measurement system that does not itself provide any control over the injected current. This is a separate safety hazard issue since the injected current can be larger than recognized limits, for example, the limits set by the International Electrotechnical Commission standard IEC-60601 for electrical medical equipment. For these many reasons, the AD5933 circuit 650 is itself unsuitable for human bioelectric impedance or tetrapolar impedance determination.

[0082] The four terminal, Tetrapolar Analog Front End Patient Interface 660 provides an interface between the AD5933 circuit 650, and the human subject. As such, it must have the proper input and output stages to interconnect to each of them.

[0083] The four terminal, Tetrapolar Analog Front End Patient Interface 660 may be considered as a combination of two voltage-to-current converters, one in the direction from AD5933 circuit 650 to the human subject's body and another from the human subject's body to AD5933 circuit 650. Since AD5933 circuit 650 applies voltage at its VOUT output and expects a current flowing into its VIN input, the four terminal Tetrapolar Analog Front End Patient Interface 660 interfaces with AD5933 circuit 650 has a voltage input and a current output. The current source output generates the current resulting from the ratio of VOUT and the impedance of the body, which is the current expected by AD5933 circuit 650 at the VIN input. At the body side, the four terminal Tetrapolar Analog Front End 660 provides a current source as output while the input is a differential voltage measurement channel. The current source excites the human subject with an alternating current. In this case, an output current of 900 pA rms has been selected to fully comply with IEC-60601 for electrical safety, but that level is only currently preferred and is neither fixed nor required for measurement purposes.

[0084] The operation of the four terminal Tetrapolar Analog Front End Patient Interface 660 can be described as follows. The AC voltage output (VOUT) of the AD5933 circuit 650 is passed to the input of the first voltage to current converter, which includes at least a Highpass filter (HPF) providing first order filtering at 500 Hz for bias removal and 60 Hz suppression. It may also be passed through a Low-pass filter (LPF) for second order filtering at 1000 Hz. The HPF or the combined HPF /LPF may be replaced by other types or notch or Band-pass filter (BPF). The filtered AC voltage (Vac) drives a voltage-controlled current source (VCCS) of the first voltage to current converter, which injects an AC current (+I or lout) into the body of the human subject. I+/Iout is directly proportional to the Vac, the filtered VOUT. The AC current I+/Iout causes a voltage drop across the body of the human subject, which is sensed by the second voltage to current converter. Since the voltage drop at the body of the human subject drives the second voltage to current converter, it generates an AC current proportional to the voltage drop in the body of the human subject. Finally, a DC component is added to the generated AC current. This added DC component is equivalent to the DC bias originally removed from VOUT. The resulting alternating current is fed to the VIN and RFB connections of the AD5933 circuit 650.

[0085] Figure 12 illustrates the use of any of the systems 410,510,610 with a human subject. The electrode pad assemblies 140 of the single, linear electrode array lead 110 are adhered to the patient U. A suggested spacing between the inner (voltage) electrodes is about ten centimeters, but different lead arrangements and different electrode spacings might be used. The electrodes are applied linearly to the patient U but may be applied to either the torso of the patient as in Figure 1 or to a limb. In Figure 12, the electrodes are applied to a human subject's leg, where they might be used to monitor ECF fluid level changes in a patient undergoing dialysis. In some embodiments, multiple body segments may be measured for their relative contribution to the ICF and ECF to the hydration of the tissue/body.

[0086] Thereafter, the unit 410, 510, 620 generates and feeds a Low Frequency, low amperage current between the outer two electrodes 120, 126 and takes voltage measurements across the inner pair of electrodes 122, 124. An impedance value is calculated by the unit 410, 510, 610 and uploaded to the display 30. Individual measurements may be taken at spaced time intervals and displayed or series of measurements may be made and combined in various ways, for example, averaged non-overlapping or overlapping serial blocks of measurements. The real time/near real time reaction of the patient/subject to a procedure such as dialysis can be monitored by observing the changes in measured impedance values on the display.

[0087] It will be appreciated that measurement of ECF/ECW differs from thoracic impedance measurement for cardiopulmonary purposes by (1) the use of a Low Frequency signal and (2) the ability to locate the electrodes anywhere on the torso or any of the limbs of the human subject. Limb location is actually preferred for certain applications such as ECF monitoring of dialysis patients as illustrated by Figure 12.

[0088] Figures 13-16 are detailed diagrams of components for a suggested Tetrapolar Analog Front End Patient Interface 660. Figure 13 provides details of the filtering subcircuit associated with the first voltage to current converter. The signal from the VOUT terminal of the AD5933 circuit 650 is passed through a High Pass Filter and then a Low Pass Filter that provides a filtered voltage output, Vac passed to the voltage-controlled current source (VCCS) of the first voltage to current converter, details of which are depicted in Figure 14. Also depicted in Figure 14 is an optional, isolated Current Sensing output, which monitors the magnitude of the ac current passing through the human subject for safety considerations. Voltages are obtained from the human subject through the V+, V- electrodes. R_Body represents the resistance of the body between the V+, V- electrodes. U2 and U3 in this figure and US in Figure 15 are transconductance amplifiers providing current outputs.

[0089] Figure 15 depicts details of the second voltage to current converter. The V+, V voltages are combined in the U4 amplifier and the differential voltage output passed to the US voltage to current converter, which converts the differential voltage into a current and adds a bias equal to that stripped out of the voltage signal at the AD5933 VOUT terminal. The resulting current (Current Out) is fed to the VIN and RFB terminals of the AD5933 circuit 650.

[0090] Figure 16 provides details of a Current Sensor RMS Detector connected across the output side of the Tl transformer in the Figure 14. As configured, it provides an RMS Buffered Voltage output that can be tapped by a monitoring circuit such as an input of the device controller 80. It also powers an Over Current Alarm in the form of a light source diode D5.

<u>Second Embodiment of Fluid Monitor Device and Method</u>

[0091] Now, a second embodiment of the monitor device and method are disclosed in which the hydrations signals at multiple frequencies may be determined and cardiac signals may also be measured and monitored. The monitoring of the cardiac signals facilitates various diagnostic and measuring methods including a method for monitoring tissue hydration of a patient using at least four electrodes using multiple frequencies, a method for monitoring low frequencies, less than 15 kHz analysis to determine extracellular hydration, a method for monitoring High frequency, greater than 15 kHz analysis to determine extracellular and intracellular hydration, a method of using an electrode array and operably coupled impedance measuring device to determine the impedance of the user, a method to calculate a numerical value of determined impedance on the impedance measuring device, a method for measuring a user's fluid level, a method for processing a Bioimpedance signal of a patient for derivation of heart rate, heart stroke volume, and cardiac output, a method of determining the effective left ventricular ejection time (ELVET), a method of estimating heart rate, a method of determining cardio cycles, a method of discarding cardio cycles exhibiting interference artifacts, a method of constructing a multi-dimensional vector for each selected cardio cycle, a method of determining stroke volume and a method of determining cardiac output as a product of stroke volume and heart rate. Each of these methods and their processes that may be implemented using the second embodiment of the fluid monitor device are described below in more detail.

[0092] The cardiac monitoring (based on the measured ECG voltages) may include both the determination of heart rate (HR) from electrocardiogram (EKG) signals and the determination of heart stroke volume (SV) from thoracic impedance signals, from which cardiac output (CO) can be estimated. The heart rate can be determined in a number of ways.

The phonocardiogram is considered among the most accurate methods of determining heart rate. However, due to the practical difficulties in using it, the phonocardiogram method is generally not employed for any continuous or long-term monitoring. Thus, heart rate is most typically determined by the electrocardiogram (EKG). The analog EKG signal typically displays electro-cardial events as perturbations referred to as waves. The heartbeat is most clearly reflected in the EKG signal as an R wavepeak between a pair of adjoining Q and S wave valleys. The basic and commonly used method of automatically identifying the QRS wave pulses in point is the threshold method in which the rate of voltage change between consecutive data points of the EKG signal is monitored and compared with a threshold value. Slopes exceeding the threshold value are deemed to be associated with a portion of the QRS pulse. While this method commonly detects the interval between consecutive R waves successfully more than eighty percent of the time, it typically has difficulty in dealing with sources of irregular signal components such as pacemakers, muscle noise, 60 Hz interference as well as nearby T or P waves which may also be associated with significant slope changes.

[0093] Hemodynamic monitoring of the heart can provide very valuable physiological information regarding the functional state of the myocardium, which is intimately related to its mechanical behavior. The quantitative measurement of blood flow, or the cardiac output (CO), is one of the most useful parameters in assessing cardiac capability, but it is also one of the most difficult to measure. It cannot be accomplished with the electrocardiogram (EKG) which does not reflect the real pumping action of the heart. Both invasive and non-invasive methods are available for measurement of cardiac output. The invasive methods are considered the most accurate. The risks associated with them are often an unacceptable trade-off, for they require direct access to the arterial circulation. In addition, invasive methods are not suitable for repetitive measurements and normally cannot be performed outside a hospital. Furthermore, invasive methods are very demanding in terms of time consumption and the need for skilled personnel. Impedance Cardiography has been found to be one non-invasive method with the potential for monitoring the mechanical activity of the heart with virtually no risk. It can be conveniently handled by nursing and non-technical staff. It can usually tolerate moderate patient movement and can be left unattended for continuous and long-term monitoring .U.S. Pat. No. 3,340,867, now RE 30,101, to Kubicek et al. discloses an impedance plethysmograph which employs four electrodes, two around the neck and two around the torso of a patient, to provide an impedance difference signal from the two center electrodes. The outermost pair of electrodes applies a small magnitude, high frequency alternating current to the patient while the inner pair of electrodes were used to sense voltage levels on the patient above and below the patient's heart. The impedances of the patient at each of the inner pair of electrodes could be determined from the sensed voltages and known applied currents. According to Kubicek et al., stroke volume (SV) is related to impedance Z as follows: where R is blood resistivity, L is the distance between the inner voltage sensing electrodes, Z sub o is the mean thoracic impedance determined from the inner, voltage sensing electrodes, VET is the ventricular ejection time and dZ/dt sub min is the maximum negative slope change of the time-differentiated impedance signal, which is the time-differentiated difference between the impedances determined at the center two electrodes. The above equation is referred to as Kubicek's equation. Cardiac output per minute is stroke volume time's heart rate in beats per minute. The Kubicek equation is based upon a parallel column model of the thorax in which it is assumed: (1) the thorax is a cylinder, consisting of two electrically conducting tissue paths, also of cylindrical form with uniform cross-sectional areas and homogenous conducting materials, the first path being the blood with a relatively low resistivity and the second path being all other tissues with relatively high resistivity's; (2) the relationship between the maximum impedance change and the stroke volume during the cardiac cycle is linear; (3) impedance measurements of the individual specific tissue volumes are not very useful in developing the model (the parallel columns model relies on the intact thoracic measurements); and (4) at 100 kHz frequency, a physiologically safe frequency, the relative thoracic impedance changes are at a maximum, the effects of polarization are negligible, and the reactive component of impedance is small, especially when compared to the real component, so that the reactance could be ignored in determining impedance with only a small error.

[0094] Yet another model and equation for relating impedance and stroke volume has been proposed by Sramek. According to Sramek, stroke volume (SV) is related to impedance Z as follows: where H is the patient's height. The Sramek equation is based upon a frustoconical model of the thorax. The Sramek model illustrates some improvement and accuracy over the Kubicek model but its major assumptions are still similar to those of the Kubicek model. Despite its advantages, impedance cardiography has not been well accepted by clinicians for three primary reasons: (1) poor correlation of the methods and models with the results of the more accepted invasive techniques; (2) still a relatively high dependence on skilled technical operators; and (3) still a discomfort to and/or disturbance of patients associated with the use and application of band electrodes. It is believed that poor correlation, the primary reason, can be traced back to a single source, namely the continuing inability to relate impedance cardiography and its mathematical model directly to cardiac physiology.

[0095] The following are definitions and abbreviations of some of the terms used frequently herein: Heart Rate (HR): the number of times the heart contracts each minute. Ventricular Ejection Time (VET): the time interval of the opening and closing of aortic value during which there is movement of blood out of a ventricle. Stroke Volume (SV): the volume of blood pumped out by a ventricle (in particular the left ventricle) with each contraction of the heart. Cardiac Output (CO): the amount of blood pumped out of the heart into the systemic circulation each minute. Ejection Fraction (EF): the ratio SV/EDV, which is the percentage of blood in a ventricle ejected with each contraction; it is directly related to the strength of the heart with <50% considered abnormal. End Diastolic Volume (EDV): the volume of blood that fills the ventricle before ejection. It would be desirable to determine heart rate more accurately than can be determined using the cardiogram threshold method currently employed. It further would be desirable to provide non-invasive, cardio graphic impedance monitoring to estimate stroke volumes, cardiac outputs and related cardiac function parameters which correlate more closely with the stroke volumes, cardiac outputs and the like determined by means of recognized, accepted invasive procedures, but which does not require of operators the technical skills required by current impedance cardiograph systems, and does minimize discomfort to the patient on which the system is used, thereby permitting relatively long-term monitoring of the patient's condition. The device and method disclosed herein may use the above methods to perform hemodynamic monitoring of the patient using the multiple frequency impedance and ECG voltage measurements.

[0096] Figures 17A and 17B are a front view and back view, respectively of a second embodiment of a fluid impedance monitor 1700 that has a tablet computer form factor. The monitor device 1700 may have a power button 1702 to turn on/off the device, a handle 1704 for easy carrying of the monitor device and a large display 1706 (with the tablet computer form factor) that can display a lot of medical data clearly to the operator of the monitor device. The display may or may not be a touchscreen. In one embodiment, the device 1700 may be built around and based on an existing tablet computer.

[0097] The monitor device 1700 may also have a speaker for playing sound to the operator, a set of status lights 1708 that show the status of the monitor device, such as for example including power on/off, wireless network connection. The monitor device 1700 also may have a navigation controller 1710 that allows the operator to control the functioning and operations of the fluid monitor device 1700. In the example shown in Figure 17A, the navigation controller may be a well-known D-pad controller. The monitor device 1700 performs the same hydration measurement and monitoring methods as described above for the first embodiment and also measure/monitor cardiac signals and perform various methods described below using the hydration data and the cardiac data.

Embodiments of Electrode Lead Array for Second Embodiment

[0098] Figure 18 illustrates a first implementation a single electrode lead array 1800 for the monitor of Figures 17A and 17B when placed on a patient and Figure 19A 19B illustrates more details of the single electrode lead array 1800 shown in Figure 18. The electrode lead array 1800 may rest on the patient as shown in 18 and may be made of a suitable material. The electrode lead array 1800 may have a first electrode 1802, a second electrode 1804, a third electrode 1806 and a fourth electrode 1808 that are axially aligned, spaced apart, down the body of the patient. The electrode may also have a connector at an end that allows a wire/cord to be connected/disconnected to/from the electrode lead array to delivery signals to the patient and receive signals from the body of the patient.

[0099] The electrode lead array 1800 and the electrodes 1802-1808 may be adhered to the body of the patient to deliver signals (voltages and/or currents) to the patient and measure signals of the patient (including hydration signals and cardiac signals. In one embodiment, the first and fourth electrodes 1802, 1808 are current sources delivering current to the patient and being located at opposite ends of the electrode lead array 1800, while the second and third electrodes 1804, 1806 measure electrical potential from the body of the patient and are preferably located adjacent to each current source. For example, a sinusoidal current may be applied from the impedance measuring device 1700 to the first and fourth electrodes and the monitor device may detect a differential electrical potential between the second and third electrodes to determine the impedance of the user from the detected differential electrical potential. The electrodes 1802-1808 may be fixed along the electrode lead array 1800 and their spacing relative to one another is also fixed and predetermined, with the first and second electrodes 1802, 1804 being spaced a first pre-determined distance D1 apart, and the third and fourth electrodes 1806, 1808 being spaced an equal pre-determined distance D2. The pre-determined distances DI, D2 may be about five centimeters or about two inches.

[00100] The side of the electrode lead array 1800 that will rest against the body of the patient may be pre-coated during manufacture with a contact adhesive. A removable, adhesive protective film may be provided that is removed when the electrode lead array 1800 is being adhered to the patient. Preferably, the electrodes 1802-1808 may be coated with an electrically conductive hydrogel which acts, along with the contact adhesive, and allows the electrode lead array 1800 to releasably adhere to the user's skin.

[00101] As shown in Figure 19A and 19B, the electrode lead array 1800 may have a connector 1900 that releasably connects to an electrical lead 1902 that has the electrodes 1802-1808 formed thereon. As shown in Figure 19B, each electrode 1802-1808 has its one wire that connects the electrode to the connector 1900. In one embodiment, the electrode lead array 1800 may have the one or more electrodes 1802-1808 formed on a conductive trace pad layer 1904 that is adhered to/formed on a flexible layer 1906 that may be a Mylar material.

[00102] Figure 20 illustrates an implementation of a dual electrode lead array 2000 for use with the monitor of Figures 17A and 17B and Figure 21A 21B illustrates more details of the dual electrode lead array 2000 shown in Figure 20. The dual electrode lead array 2000 is two of the single electrode lead arrays 1800 placed and adhered to each side of the patient body as shown in Figure 20. Each side of the dual electrode lead array 2000 has a same set of four axially oriented electrodes 1802-1808 that has the same spacing and construction as described above. As shown in Figure 21A, the dual electrode lead array 2000 has the same connector 1900 (that electrically connects the lead array to the monitor device as is known), but the connector accepts two leads 1902A, 1902B which one lead for each set of four electrodes. As shown in Figure 21B, each side of the dual electrode lead array 2000 may be made out of similar material as the single electrode lead array.

[00103] Both the single and dual electrode lead array 1800, 2000 function to apply the high frequency current, measure the resulting voltage changes and measure the differential ECG voltages. The single electrode lead array 1800 may be used with the second embodiment of the fluid monitor 1700 to perform hydration/fluid measuring and monitoring. The dual electrode lead array 2000 may be used when hydration and cardiac signals (differential ECG voltages) are being monitored and measured.

[00104] The dual electrode lead array 2000 is used for the ECG to provide the triangulation and capture ECG voltages with enough ECG signal fidelity to satisfy the signal processing requirements for fiducial landmark recognition as shown in Figure 21C to determine the Z, Q, R, S, J, ST, T and new E points in the ECG signal for a patient . There are cases where because of differences in anatomy and disease state where a bilateral configuration produces the ECG fidelity needed with a 3 lead method. The Z point is the isoelectric baseline of the ECG, while the Q wave is any negative deflection that precedes an R wave. The Q wave represents the normal left-to-right depolarization of the intraventricular septum. The R wave is the first upward deflection after the P wave and the R wave represents early ventricular depolarization. The S wave is the first downward deflection of the QRS complex that occurs after the R wave. The J point denotes the junction of the QRS complex and the ST segment on the electrocardiogram (ECG), marking the end of depolarization and beginning of repolarization. The ST segment encompasses the region between the end of ventricular depolarization and beginning of ventricular repolarization on the ECG. In other words, it corresponds to the area from the end of the QRS complex to the beginning of the T wave. The T wave on an electrocardiogram (ECG) represents typically ventricular repolarization The E point represents the end of the T wave.

[00105] The time relationships are defined as the following to be an acceptable signal.

B>Q;
C>R;
X>T;
ST>B;
X>E;
X>ST;
B>R;

(X-C)>(C-B);
B is after ST segment;
B is before R;
B to C time greater than C to X time;
C is before R;
X is too close to T;
X is too far from E; and
X is before ST.

[00106] Figure 22 illustrates a circuit board 2200 of the monitor 1700 housed in the monitor housing shown in Figures 17A-17B that is capable of measuring/monitoring both fluid signals (hydration) and cardiac signals of the patient when the dual electrode lead array 2000 is adhered to the patient (an example of the positions of the dual electrode lead array 2000 is shown in Figure 20). The circuit board may have an excitation portion 50 that includes a multifrequency current source/signal generation circuits 52 and an amplifier and filter portion 54. The signal generating circuitry 50 generates a stable excitation current (I) using the current source subcircuit 52 includes a known constant current source (not depicted) and clock oscillator (not depicted) to supply a current of about 1 mA or less, preferably a.98.+/-0.01 mA, at a 100+/- .10 kHz and 5+/-1kHz preferably to the first and fourth electrodes 1802, 1808 through an isolation transformer that may be part of the amplifier and filter 54 through a connection cord to the patient electrode lead 1800, 2000. The current source subcircuit 52 is configured to output a current of less than 4 mA under all conditions including equipment component failure. The wave form of the current may be sinusoidal with less than ten percent total harmonic distortion. Voltage values across two of the four electrodes, such as the second and third electrodes 804, 806 are passed through an amplifier and filter portion 64 that includes an isolation transformer connected to an amplifier and low pass filter subcircuit. The low pass filter subcircuit functions to remove extraneous electrical interference from ambient sources, for example, home appliances operating on standard residential 60 Hz current. An example of a cut-off frequency of the low pass filter sub circuit may be about 50 Hz. The base unit 1700 measures voltage developed across detection electrodes 1804, 1806 when the excitation current source is energized. The measured voltage level will be between about 18 millivolts and 104 millivolts (to provide an anticipated range of impedance measurement of about 10 ohms to 50 ohms, at the 2 mA current).

[00107] The circuit 2200 further comprises a micro-controller 80 that is connected to the patient interface (including the excitation circuits 50 and the impedance measurement portion 60) and a power supply 90. The microcontroller 80 may generate a wireless output 2212 of the raw signals for data processing and display by a fluid monitor system 2220 that may be coupled to the monitor device 1700 as shown. The fluid monitor system 2220 may be a computer system having a plurality of lines of computer code/instructions that are executed by a processor of the fluid monitor computer system 2220 to perform the data processing and display of the patient monitoring data. The microcontroller 80 also may generate a USB output 2214 of the raw signals for data processing and display by another computer system. The USB output 2214 may be output to a tablet computer or other computer such as the fluid monitor system 2220. The microcontroller may be any known microcontroller, such as a PIC 16F87 device, that controls generation of the excitation current and receives the filtered voltage analog signal from the amplifier and low pass filter 64 at the input of an RMS to DC converter 2202

[00108] The injected current may not generated for a short period of time (e.g. fifteen to thirty seconds) after the start switch 28 is actuated to allow the user to settle into a quiescent state. For example, the current may be injected into the patient for a predetermined period, e.g. thirty seconds, to perform the measurement. Voltage values sampled and A.D converted 72 and received by the data acquisition circuitry 77 of the micro-controller 80 at a rate, such as for example, of about five Hundred samples per second for all or most of the thirty second period. Data analysis and storage circuitry of micro-controller 80 sums the counts generated by the A/D converter 72, divides sum by the total number of samples taken to provide an average voltage value which is converted into an impedance value. The algorithm used for generating impedance in tenths of ohms is: averaged A/D counts*Gain+Offset, where in the preferred circuit the Gain is 0.6112 and the Offset is 1.1074. Gain and Offset are based on the electronics design and operating range and are used for all base units 20.

[00109] Although not shown in Figure 22, the circuit board 2200, the data analysis circuitry also may control various displays 30, 32, and 34. The power supply 90 may have power management circuitry 82 that controls the generation and distribution of power in the base unit circuitry 2200 to control operation of the system. Specific functions of the power management circuitry 82 include a first function 82A of providing power to the processor and a second function 82B of providing power to the analog electronics. A power supply 90 may be provided by conventional dry-cell batteries (not shown) or by an external power adapter (not shown) connected to a conventional AC outlet.

[00110] The monitor device 1700 may be provided with a USB port that provides tablet computer or PC connection) to work with logic level signals. There may also be a USB or serial port and the timing for the serial data can be similar to RS232 signal or other conventional data transfer format.

[00111] The monitor device 1700 also may have an output Stimulus Signal Source Generation circuitry 2300, an example implementation of which is shown in Figure 23. The stimulus output signal source is generated by a voltage sine wave 52 circuit whose frequency is controlled by the MCU 80. The sine wave can be set to any frequency between 0 Hz to 12.5 MHz with high precision. This waveform is filtered to remove any DC bias prior to being amplified and then converted to a constant current waveform. The conversion to a current waveform is performed by an MCU controlled resistance value in calibration prior to use on the patient. The monitor device 1700 also may have an input signal conditioning circuitry 2400 with an example implementation shown in Figure 24. The patient probe return signal is modified via a low pass filter and amplifier circuit with set gain characteristics. The resulting signal is a primary sine wave voltage that is converted into its root-mean-square (RMS) equivalent voltage. The RMS DC value is low pass filtered to ensure no high frequency artifacts or system noise remain in the signal. The RMS DC value is amplified again with fixed gain to ensure sufficient resolution during conversion to digital format and output as ZOUT. The analog ZOUT signal is then converted to digital format by a high precision analog-to-digital (ADC) converter. The resulting digital values are used by the software to generate hemodynamic data. The ZOUT signal is also sent to a Differentiator Circuit 2500 for further processing.

[00112] The differentiation circuitry 2500 of the monitor device 1700 (an example implementation of which is shown in Figure 25) may transform the ZOUT signal to produce the time varying slope of the RMS return signal, dZ/dt. The resulting time differential signal is amplified with variable gain that is controlled by the MCU. The resulting signal (dZ/dt) is then amplified with fixed gain to ensure sufficient resolution during conversion to digital format. This analog signal is then converted to digital format by a high precision analog-to-digital (ADC) converter. The resulting digital values are used by the software to calculate the hemodynamic parameters of the subject.

[00113] In these embodiments, the cardiac signal generation differs in that multiple frequencies of current are being used (for the impedance measurement). Furthermore, unlike typical system in which the well known ECG signals are used to determine a point in the cardiac cycle (timing) to start the spectral analysis of the dZ/dt waveform, the measured ECG signals are used for the timing, but also to determine the points Z,Q,R,S,ST,T and an new point we call E, which is the end of the ECG T wave which must coincide with the X point on the dZ/dt in order to be processed for the stroke volume calculation.

[00114] Figure 26 is a flowchart of a method 2600 for measuring impedance of a patient using an electrode array and operably coupled impedance measuring device. An electrode array for use with a physiological electronic monitor used to monitor electrical characteristics of a user's body may be adhered (2602) to the patient. The method using a linear electrode array lead including at least first, second, third, and fourth electrodes arranged sequentially and axially along the linear electrode array lead (examples of which are shown in Figures 18-21). The method then applies a sinusoidal current (2604) from the impedance measuring device to the first and fourth electrodes (outer electrodes of the linear electrode array lead(s). The method may then detect a differential electrical potential (2606) between the second and third electrodes (middle electrodes of the linear electrode array lead(s)) with the impedance measuring device. The impedance measuring device may process the signals received at the middle electrodes. The method may then determine the impedance (2608) of the user from the detected differential electrical potential.

[00115] Figure 27 illustrates more details of a method 2700 for measuring impedance of a patient to calculate a numerical value of determined impedance on the impedance measuring device. The method differentially amplifies and low pass filters (2702) voltages from the second and third electrodes (middle electrodes). The method may then calculating the impedances by sampling (2704) the differentially amplified and low pass filtered voltage from the second and third electrodes at predetermined intervals for a number of times and adding the sampled voltages to generate a sum (2706). The method then divides the sum by the number of times to provide an averaged voltage value (2708) and scales (2710) the averaged voltage value and combining the scaled averaged voltage value with a predetermined offset value to generate a numerical value of the differential impedance.

[00116] Figure 28 is a flowchart of a method 2800 for determining cardiac characteristics based on bioimpedance including heart rate, heart stroke volume, and cardiac output signals for the patient. The method digitally filters and phase corrects (2802) the bioimpedance signal to remove gain-phase-frequency distortions and estimates (2804) heart rate using a power spectrum of the bioimpedance signal and an auto-convolution function of the power spectrum. The method suppresses breath waves (2806) to remove undesired power spectra components and generate a bioimpedance signal of restored shape. The method then determines (2802) one or more cardio cycles of the restored bioimpedance signal and determines effective left ventricular ejection time (ELVET) using check points within said cardio cycles. The method then discards (2810) at least some of said cardio cycles which exhibit interference artifacts. The method may also locate points on a time-derivative Bioimpedance curve for the Bioimpedance signal; and select the points which most accurately reflect cardiac events.

[00117] Figure 29 is a flowchart of a method 2900 for estimating heart rate. The method calculates (2902) a power spectrum of a bioimpedance signal (from a bioimpedance monitor device) and multiplies (2904) the power spectrum by a selected amplitude-frequency function to differentiate the signal. The method then suppresses breath harmonics and auto convolutes the resulting power spectrum according to a formula (2906) and determines (2908) a maximum amplitude value of auto convolution in a predefined frequency range as an estimation of heart rate.

[00118] Figure 30 is a flowchart of a method 3000 for determining cardio cycles. The method filters (3002) a bioimpedance signal from a bioimpedance monitor to emphasize fronts (a beginning of each cardio cycle) of cardio cycles and calculates (3004) a time-amplitude envelope of the cardio cycles by analyzing the first five harmonics of the powers spectrum of the bioimpedance signal after filtration. The method then selects (3006) the cardio cycle fronts by comparison with said calculated time-amplitude envelope and rejects erroneously-detected fronts. In another embodiment, the method for discarding cardio cycles exhibiting interference artifacts may also detect time and amplitude relations referencing check points within individuals of a plurality of cardio cycles, compare the time and amplitude relations between individuals of a plurality of cardio cycles and further examine selected cardio cycles which exhibit the presence of artifact according to a plurality of comparison criteria. In another aspect, a method for constructing a vector for each cardio cycle and in particular a multi-dimensional vector for each selected cardio cycle. The method compares the multi-dimensional vector with such vectors for other cardio cycles and rejecting the cardio cycles with vectors having no neighboring vectors of other cardio cycles.

[00119] Figure 31 is a flowchart of a method 3100 of determining effective left ventricular ejection time (ELVET). The method filters (3102) the bioimpedance signal (from a bioimpedance monitor device) and suppresses (3104) breath waves in the filtered signals. The method detects (3106) a cardio cycle, calculates (3108) the time derivative of the bioimpedance signal and determines (3110) the maximum value of the time derivative. The method then determines (3112) effective ejection start time, determines (3114) effective ejection end time and calculates (3116) effective left ventricular ejection time (ELVET) as change in time between effective ejection start time and end time.

[00120] Figure 32 is a flowchart of a method 3200 for determining stroke volume. The method, using the bioimpedance monitor, determines (3202) specific blood resistivity (P) and measures (3204) a distance L between two bioimpedance electrodes applied to the patient and determines (3206) a base thoracic impedance Z. The method determines (3208) an effective left ventricular ejection time (ELVET) and calculates stroke volume SV according to the equation where K is a novel scale factor related to body composition of the patient. A method of determining cardiac output as a product of stroke volume and heart rate is also disclosed.

[00121] The foregoing description, for purpose of explanation, has been with reference to specific embodiments. However, the illustrative discussions above are not intended to be exhaustive or to limit the disclosure to the precise forms disclosed. Many modifications and variations are possible in view of the above teachings. The embodiments were chosen and described in order to best explain the principles of the disclosure and its practical applications, to thereby enable others skilled in the art to best utilize the disclosure and various embodiments with various modifications as are suited to the particular use contemplated.

[00122] The system and method disclosed herein may be implemented via one or more components, systems, servers, appliances, other subcomponents, or distributed between such elements. When implemented as a system, such systems may include and/or involve, inter alia, components such as software modules, general-purpose CPU, RAM, etc. found in general-purpose computers,. In implementations where the innovations reside on a server, such a server may include or involve components such as CPU, RAM, etc., such as those found in general-purpose computers.

[00123] Additionally, the system and method herein may be achieved via implementations with disparate or entirely different software, hardware and/or firmware components, beyond that set forth above. With regard to such other components (e.g., software, processing components, etc.) and/or computer-readable media associated with or embodying the present inventions, for example, aspects of the innovations herein may be implemented consistent with numerous general purpose or special purpose computing systems or configurations. Various exemplary computing systems, environments, and/or configurations that may be suitable for use with the innovations herein may include, but are not limited to: software or other components within or embodied on personal computers, servers or server computing devices such as routing/connectivity components, hand-held or laptop devices, multiprocessor systems, microprocessor-based systems, set top boxes, consumer electronic devices, network PCs, other existing computer platforms, distributed computing environments that include one or more of the above systems or devices, etc.

[00124] In some instances, aspects of the system and method may be achieved via or performed by logic and/or logic instructions including program modules, executed in association with such components or circuitry, for example. In general, program modules may include routines, programs, objects, components, data structures, etc. that perform particular tasks or implement particular instructions herein. The inventions may also be practiced in the context of distributed software, computer, or circuit settings where circuitry is connected via communication buses, circuitry or links. In distributed settings, control/instructions may occur from both local and remote computer storage media including memory storage devices.

[00125] The software, circuitry and components herein may also include and/or utilize one or more type of computer readable media. Computer readable media can be any available media that is resident on, associable with, or can be accessed by such circuits and/or computing components. By way of example, and not limitation, computer readable media may comprise computer storage media and communication media. Computer storage media includes volatile and nonvolatile, removable and non-removable media implemented in any method or technology for storage of information such as computer readable instructions, data structures, program modules or other data. Computer storage media includes, but is not limited to, RAM, ROM, EEPROM, flash memory or other memory technology, CD-ROM, digital versatile disks (DVD) or other optical storage, magnetic tape, magnetic disk storage or other magnetic storage devices, or any other medium which can be used to store the desired information and can accessed by computing component. Communication media may comprise computer readable instructions, data structures, program modules and/or other components. Further, communication media may include wired media such as a wired network or direct-wired connection, however no media of any such type herein includes transitory media. Combinations of the any of the above are also included within the scope of computer readable media.

[00126] In the present description, the terms component, module, device, etc. may refer to any type of logical or functional software elements, circuits, blocks and/or processes that may be implemented in a variety of ways. For example, the functions of various circuits and/or blocks can be combined with one another into any other number of modules. Each module may even be implemented as a software program stored on a tangible memory (e.g., random access memory, read only memory, CD-ROM memory, hard disk drive, etc.) to be read by a central processing unit to implement the functions of the innovations herein. Or, the modules can comprise programming instructions transmitted to a general-purpose computer or to processing/graphics hardware via a transmission carrier wave. Also, the modules can be implemented as hardware logic circuitry implementing the functions encompassed by the innovations herein. Finally, the modules can be implemented using special purpose instructions (SIMD instructions), field programmable logic arrays or any mix thereof which provides the desired level performance and cost.

[00127] As disclosed herein, features consistent with the disclosure may be implemented via computer-hardware, software, and/or firmware. For example, the systems and methods disclosed herein may be embodied in various forms including, for example, a data processor, such as a computer that also includes a database, digital electronic circuitry, firmware, software, or in combinations of them. Further, while some of the disclosed implementations describe specific hardware components, systems and methods consistent with the innovations herein may be implemented with any combination of hardware, software and/or firmware. Moreover, the above-noted features and other aspects and principles of the innovations herein may be implemented in various environments. Such environments and related applications may be specially constructed for performing the various routines, processes and/or operations according to the invention or they may include a general-purpose computer or computing platform selectively activated or reconfigured by code to provide the necessary functionality. The processes disclosed herein are not inherently related to any particular computer, network, architecture, environment, or other apparatus, and may be implemented by a suitable combination of hardware, software, and/or firmware. For example, various general-purpose machines may be used with programs written in accordance with teachings of the invention, or it may be more convenient to construct a specialized apparatus or system to perform the required methods and techniques.

[00128] Aspects of the method and system described herein, such as the logic, may also be implemented as functionality programmed into any of a variety of circuitry, including programmable logic devices ("PLDs"), such as field programmable gate arrays ("FPGAs"), programmable array logic ("PAL") devices, electrically programmable logic and memory devices and standard cell-based devices, as well as application specific integrated circuits. Some other possibilities for implementing aspects include: memory devices, microcontrollers with memory (such as EEPROM), embedded microprocessors, firmware, software, etc. Furthermore, aspects may be embodied in microprocessors having software-based circuit emulation, discrete logic (sequential and combinatorial), custom devices, fuzzy (neural) logic, quantum devices, and hybrids of any of the above device types. The underlying device technologies may be provided in a variety of component types, e.g., metal-oxide semiconductor field-effect transistor ("MOSFET") technologies like complementary metal-oxide semiconductor ("CMOS"), bipolar technologies like emitter-coupled logic ("ECL"), polymer technologies (e.g., silicon-conjugated polymer and metal-conjugated polymer-metal structures), mixed analog and digital, and so on.

[00129] It should also be noted that the various logic and/or functions disclosed herein may be enabled using any number of combinations of hardware, firmware, and/or as data and/or instructions embodied in various machine-readable or computer-readable media, in terms of their behavioral, register transfer, logic component, and/or other characteristics. Computer-readable media in which such formatted data and/or instructions may be embodied include, but are not limited to, non-volatile storage media in various forms (e.g., optical, magnetic or semiconductor storage media) though again does not include transitory media. Unless the context clearly requires otherwise, throughout the description, the words "comprise," "comprising," and the like are to be construed in an inclusive sense as opposed to an exclusive or exhaustive sense; that is to say, in a sense of "including, but not limited to." Words using the singular or plural number also include the plural or singular number respectively. Additionally, the words "herein," "hereunder," "above," "below," and words of similar import refer to this application as a whole and not to any particular portions of this application. When the word "or" is used in reference to a list of two or more items, that word covers all of the following interpretations of the word: any of the items in the list, all of the items in the list and any combination of the items in the list.

[00130] Although certain presently preferred implementations of the invention have been specifically described herein, it will be apparent to those skilled in the art to which the invention pertains that variations and modifications of the various implementations shown and described herein may be made without departing from the spirit and scope of the invention. Accordingly, it is intended that the invention be limited only to the extent required by the applicable rules of law.

[00131] While the foregoing has been with reference to a particular embodiment of the disclosure, it will be appreciated by those skilled in the art that changes in this embodiment may be made without departing from the principles and spirit of the disclosure, the scope of which is defined by the appended claims.

What is claimed is:

1. A method for measuring patient characteristics, the method comprising:

generating, by a current source in a fluid monitor device, two excitation signals each having a different frequency;

delivering, by an electrode lead array having multiple electrodes configured to adhere to a patient, the two excitation signals;

receiving, at the fluid monitor device through the electrode lead array, two bioimpedance signals from the patient based on the two excitation signals; and

monitoring, by the fluid monitor device, a hydration state and a cardiac state that are generated from the two bioimpedance signals.

2. The method of claim 1, wherein monitoring the hydration state further comprises monitoring the excitation signal having a lower frequency to determine extracellular hydration, wherein the lower frequency is less than 15 kHz.

3. The method of claim 2, wherein monitoring the hydration state further comprises monitoring the excitation signal having a higher frequency to determine extracellular and intracellular hydration, wherein the higher frequency is greater than 15 kHz.

4. The method of claim 1, wherein the electrode lead array has four axially aligned electrodes and wherein delivering the two excitation signals and receiving the two bioimpedance signals further comprises delivering the two excitation signals using a first electrode and a fourth electrode of the axially aligned electrodes and receiving the two bioimpedance signals using a second and third electrodes of the axially aligned electrodes.

5. The method of claim 1, wherein the cardiac state is one of a heart rate, a heart stroke volume, a cardiac output and a cardiac cycles signal.

6. The method of claim 4, wherein generating the two excitation signals further comprising generating two sinusoidal excitation signals and wherein receiving the two bioimpedance signals further comprises detecting a differential electrical potential between the second and third electrodes and determining, by the fluid monitor device, the bioimpedance of the patient from the detected differential electrical potential.

7. The method of claim 6, wherein detecting the differential electrical potential further comprises differentially amplifying and low pass filtering voltages from the second and third electrodes, and wherein the determining the bioimpedance further comprises sampling the differentially amplified and low pass filtered voltage from the second and third electrodes at predetermined intervals for a number of times, adding the sampled voltages to generate a sum, dividing the sum by the number of times to provide an averaged voltage value; scaling the averaged voltage value and combining the scaled averaged voltage value with a predetermined offset value to generate a numerical value of the differential impedance.

8. The method of claim 1, wherein monitoring the cardiac state further comprises digitally filtering and phase correcting the bioimpedance signal to remove gain-phase-frequency distortions; estimating a heart rate using a power spectrum of the bioimpedance signal and an auto-convolution function of the said power spectrum; suppressing breath waves to remove undesired power spectra components to generate a bioimpedance signal of restored shape; determining cardio cycles of said restored Bioimpedance signal; determining effective left ventricular ejection time (ELVET) using check points within said cardio cycles and discarding at least some of said cardio cycles which exhibit interference artifacts.

9. The method of claim 1, wherein monitoring the cardiac state further comprises determining an effective left ventricular ejection time (ELVET) that further comprises locating points on a time-derivative bioimpedance curve for the bioimpedance signal and selecting the located points which most accurately reflect cardiac events.

10. The method of claim 1, wherein monitoring the cardiac state further comprises estimating heart rate by calculating a power spectrum of the bioimpedance signal, multiplying the power spectrum by a selected amplitude-frequency function to differentiate the signal and suppress breath harmonics; auto convoluting the resulting power spectrum and determining a maximum amplitude value of auto convolution in a predefined frequency range as an estimation of heart rate.

11. The method of claim 1, wherein monitoring the cardiac state further comprises determining cardio cycles by filtering the bioimpedance signal to emphasize fronts of cardio cycles; calculating a time-amplitude envelope of the cardio cycles by analyzing the first five harmonics of a powers spectrum of the bioimpedance signal after filtration; selecting cardio cycle

fronts by comparison with said calculated time-amplitude envelope; and rejecting erroneously-detected fronts.

12. The method of claim 11, wherein determining cardio cycles further comprises discarding cardio cycles exhibiting interference artifacts by detecting time and amplitude relations referencing check points within individuals of the cardio cycles; comparing the time and amplitude relations between individuals of the cardio cycles and examining selected cardio cycles which exhibit the presence of artifact according to a plurality of comparison criteria.

13. The method of claim 11, wherein determining cardio cycles further comprises constructing a multi-dimensional vector for each selected cardio cycle; comparing said multi-dimensional vector with such vectors for other cardio cycles and rejecting the cardio cycles with vectors having no neighboring vectors of other cardio cycles.

14. The method of claim 1, wherein monitoring the cardiac state further comprises determining effective left ventricular ejection time (ELVET) that comprises filtering the bioimpedance signal and suppressing breath waves; detecting a cardio cycle; calculating a time derivative of the bioimpedance signal; determining a maximum value of the time derivative; determining an effective ejection start time; determining an effective ejection end time and calculating an effective left ventricular ejection time as change in time between effective ejection start time and end time.

15. The method of claim 1, wherein monitoring the cardiac state further comprises determining a stroke volume by determining specific blood resistivity (P); measuring a distance L between two electrodes that receive the bioimpedance patient signals; determining a base thoracic impedance Z.; determining effective left ventricular ejection time (ELVET); and calculating stroke volume (SV) according to an equation where K is a novel scale factor related to body composition of the patient.

16. The method of claim 1, wherein monitoring the cardiac state further comprises determining cardiac output as a product of stroke volume and heart rate.

17. A monitoring device, comprising:

an electrode lead array configured to be adhered to a patient that provides two excitation signals to the patient and receives bioimpedance patient signals in response to excitation signals;

a monitor device, couplable to the electrode lead array, having a processor connected to a patient interface wherein the patient interface generates the two excitation signals and receives the bioimpedance patient signals;

the processor having a plurality of lines of instructions that configure the processor to:

generate a hydration signal for the patient based on the bioimpedance patient signals;

generate a cardiac signal for the patient based on the bioimpedance patient signals; and

monitor a hydration state and a cardiac state of the patient based on the hydration signal and the cardiac signal.

18. The monitoring device of claim 17 further comprising a second electrode lead array configured to be adhered to a patient that injects the two excitation signals into the patient and receives the bioimpedance patient signals in response to excitiation signals, wherein the electrode lead array and the second electrode lead array are positionable lengthwise along each side of a chest of the patient.

19. The monitoring device of claim 17, wherein the two excitation signals are a 5kHz sinusoidal signal and a 100 kHz sinusoidal signal

20. The monitoring device of claim 18, wherein each electrode lead array further comprises four axially aligned electrodes, wherein the excitation signals are provided using a first electrode and a fourth electrode of the four axially aligned electrodes and the bioimpedance patient signals are received using a second electrode and a third electrode of the four axially aligned electrodes.

21. The monitoring device of claim 17, wherein the monitor device is battery powered.

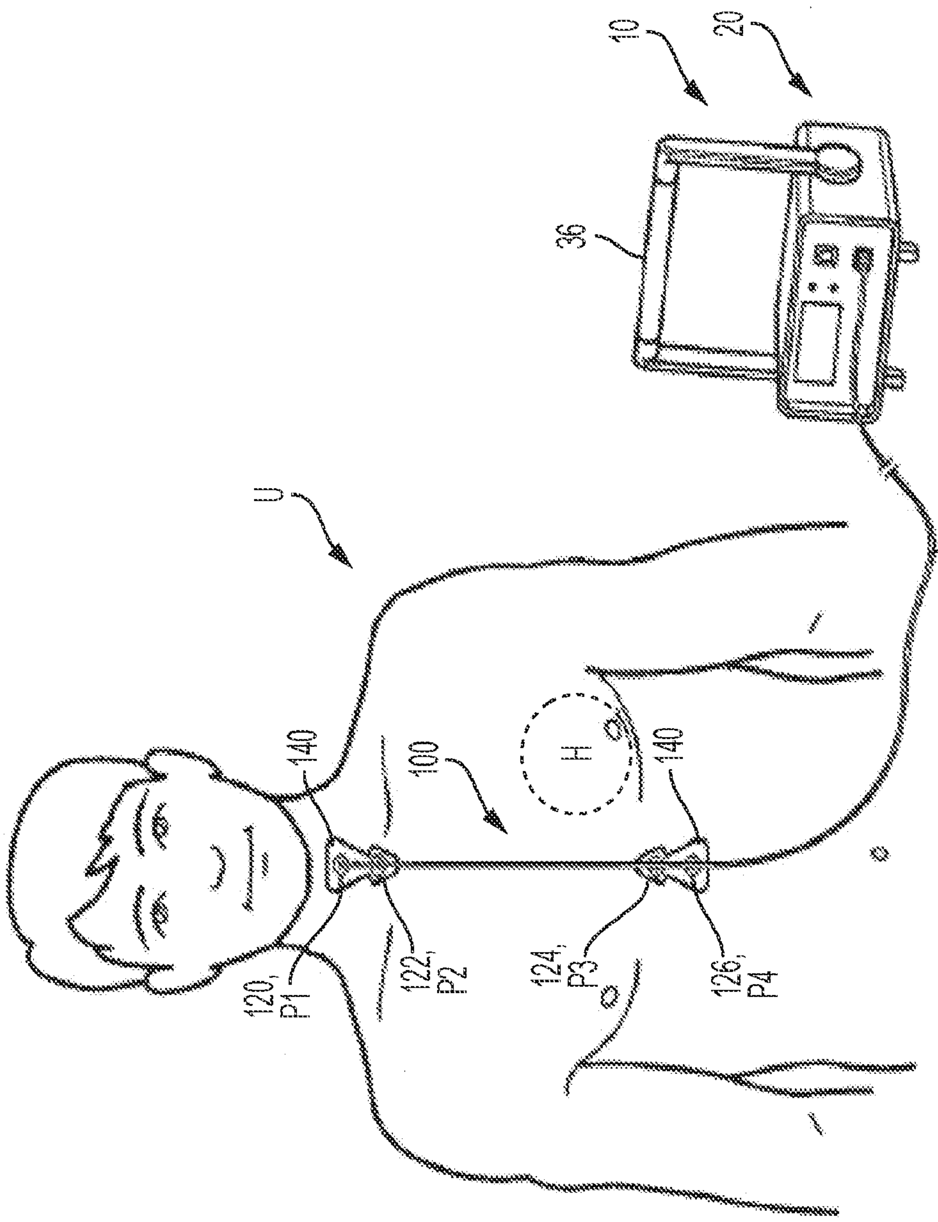


FIGURE 1

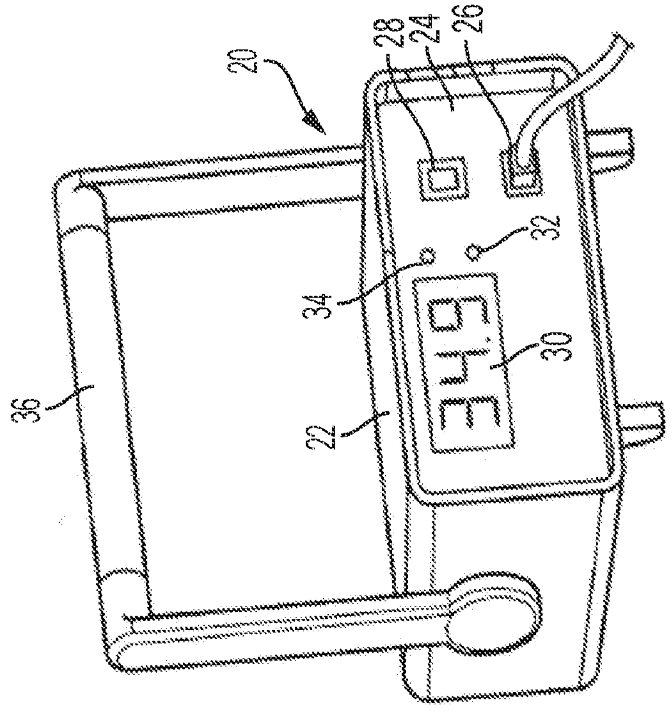


FIGURE 2

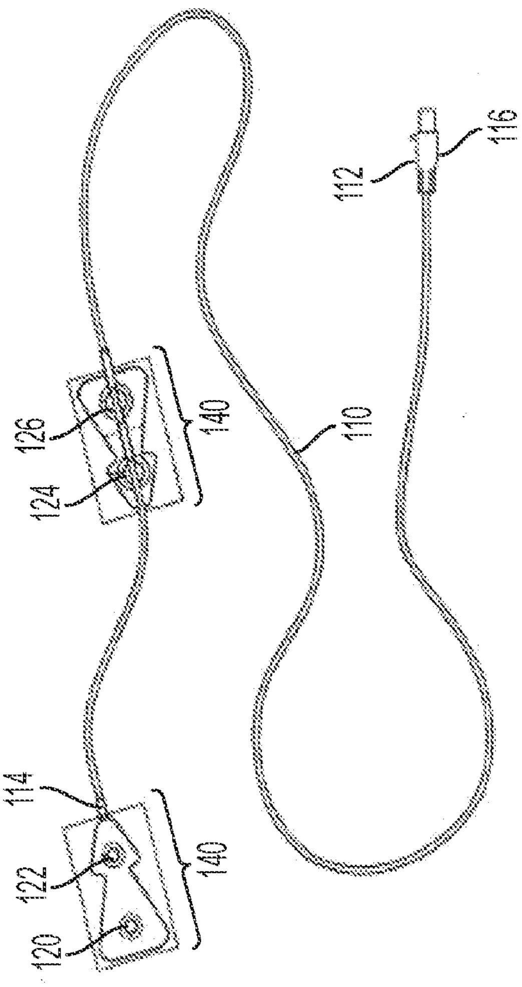


FIGURE 3A

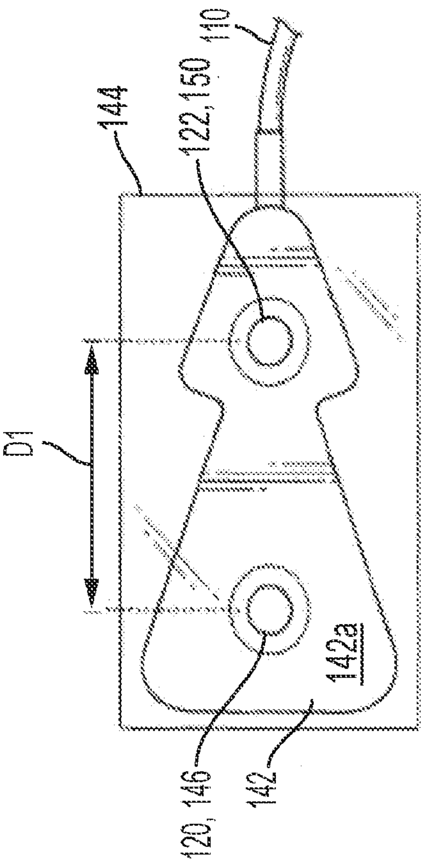


FIGURE 3B

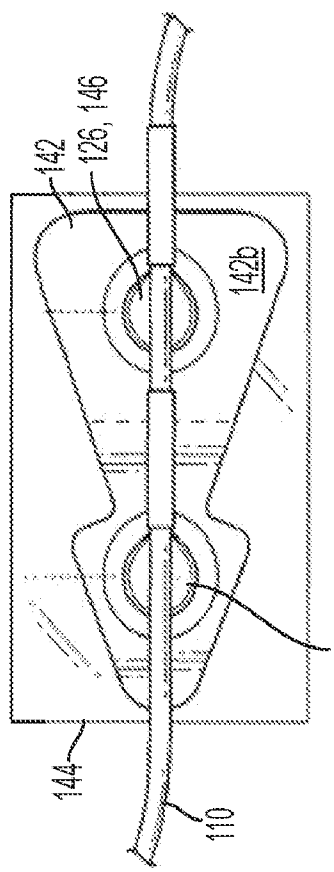


FIGURE 3C

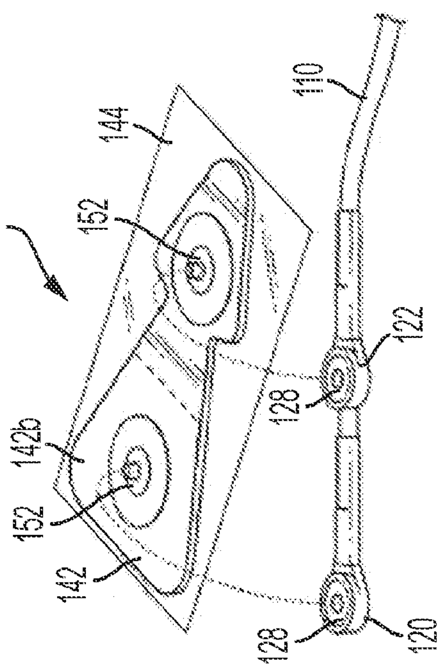


FIGURE 3D

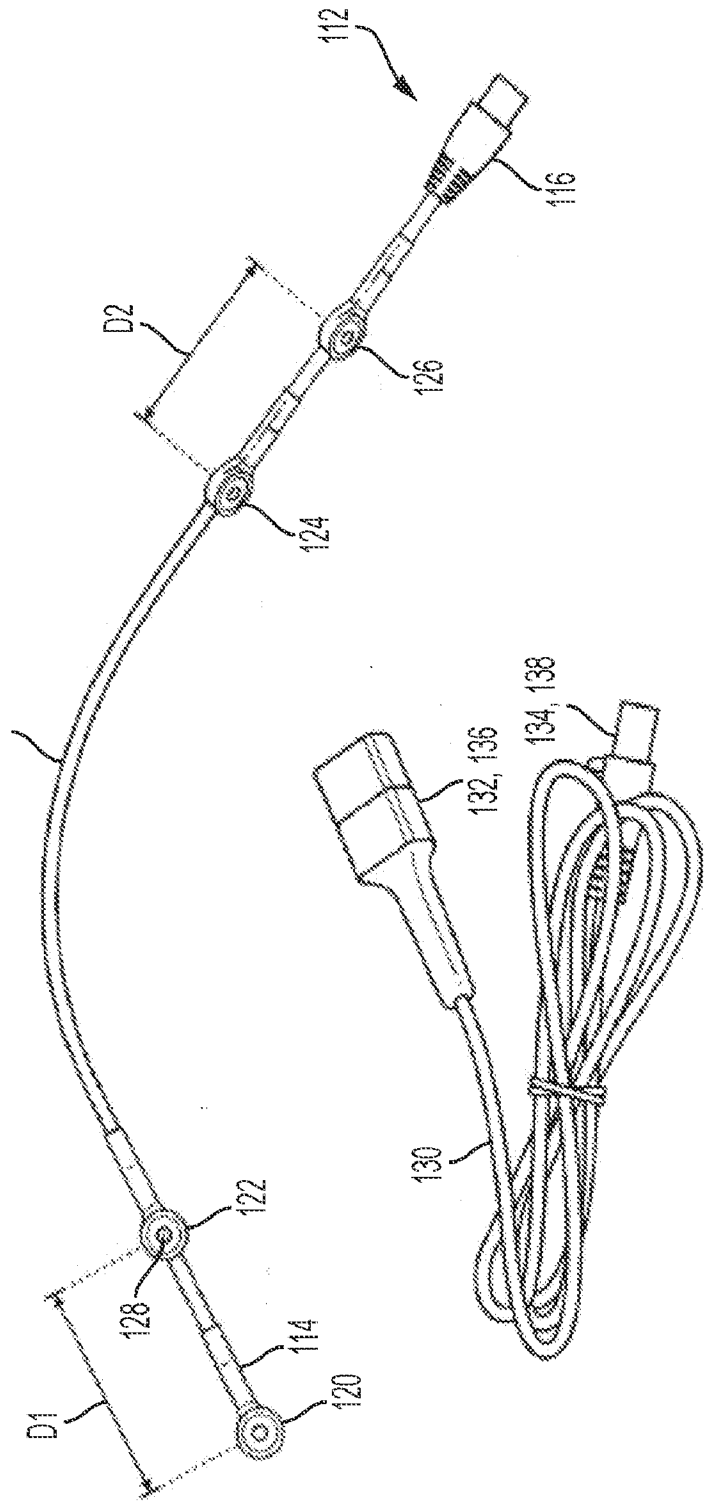


FIGURE 3E

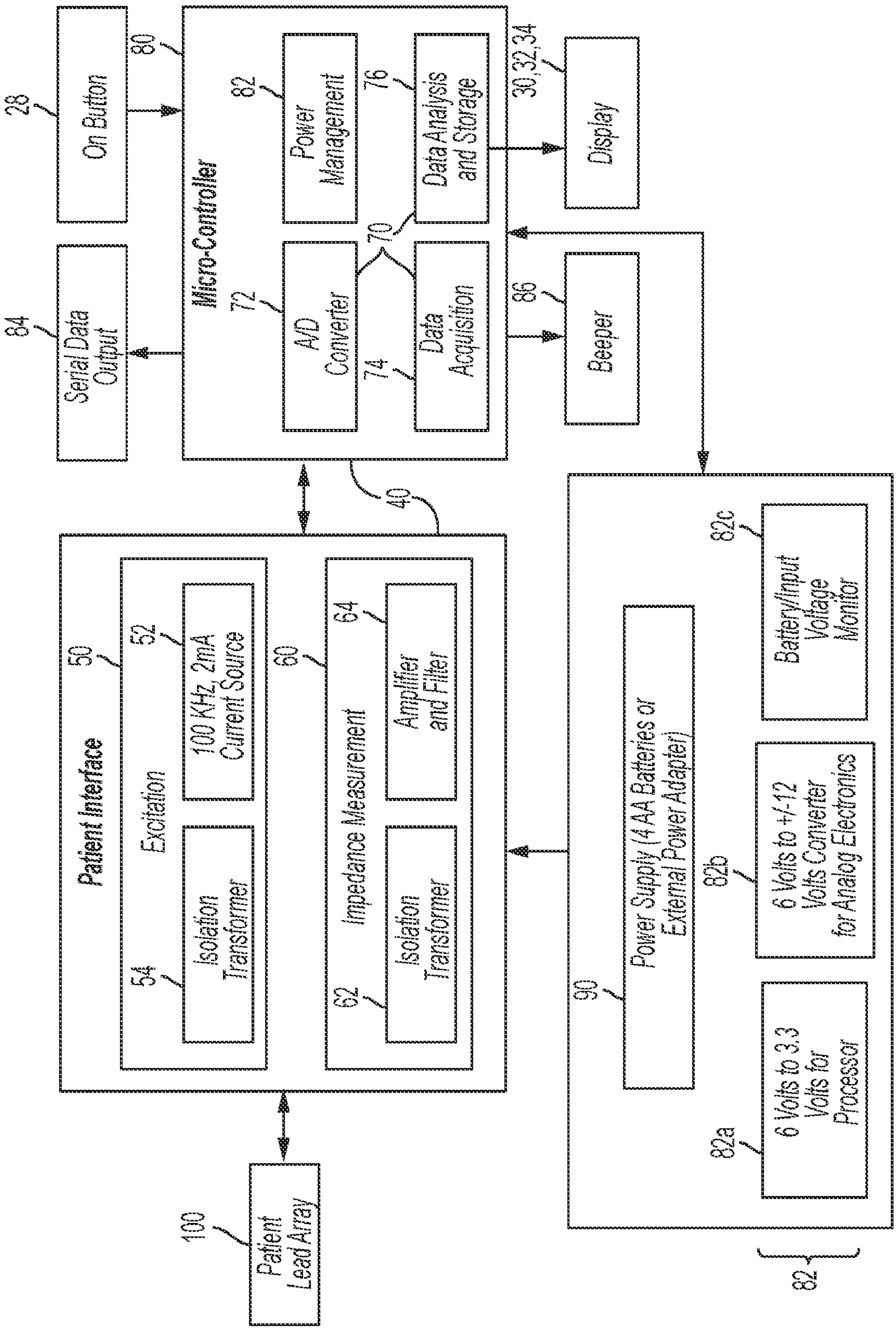

28
On Button
80
84
Serial Data Output
Micro-Controller
82
Power Management
76
Data Analysis and Storage
70
72
A/D Converter
74
Data Acquisition
30,32,34
Display
86
Beeper
40
Patient Interface
50
Excitation
52
100 KHz, 2mA Current Source
54
Isolation Transformer
60
Impedance Measurement
64
Amplifier and Filter
62
Isolation Transformer
90
Power Supply (4 AA Batteries or External Power Adapter)
82c
Battery/Input Voltage Monitor
82b
6 Volts to +/-12 Volts Converter for Analog Electronics
82a
6 Volts to 3.3 Volts for Processor
82
100
Patient Lead Array

FIGURE 4

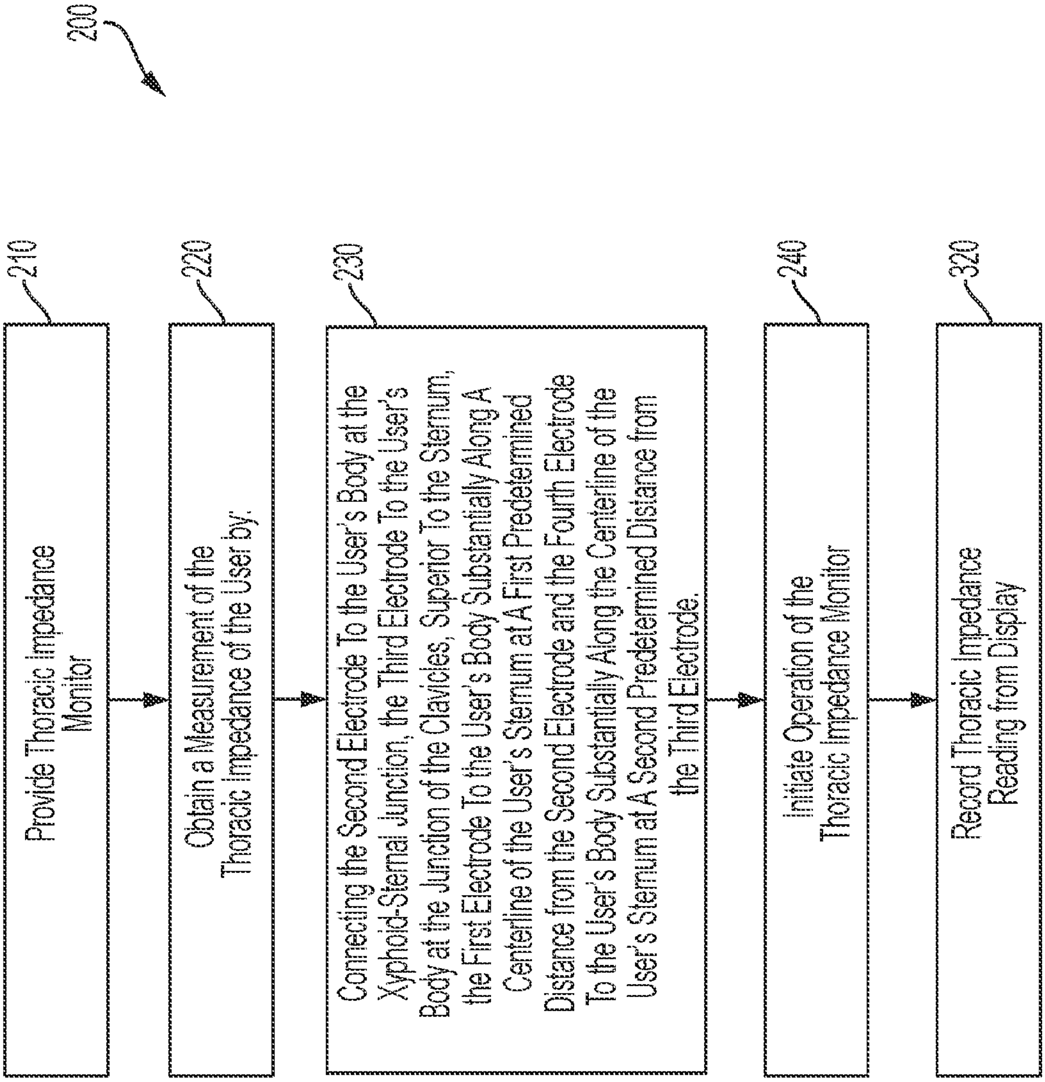

200
210
Provide Thoracic Impedance Monitor
220
Obtain a Measurement of the Thoracic Impedance of the User by:
230
Connecting the Second Electrode To the User's Body at the Xyphoid-Sternal Junction, the Third Electrode To the User's Body at the Junction of the Clavicles, Superior To the Sternum, the First Electrode To the User's Body Substantially Along A Centerline of the User's Sternum at A First Predetermined Distance from the Second Electrode and the Fourth Electrode To the User's Body Substantially Along the Centerline of the User's Sternum at A Second Predetermined Distance from the Third Electrode.
240
Initiate Operation of the Thoracic Impedance Monitor
320
Record Thoracic Impedance Reading from Display

FIGURE 5

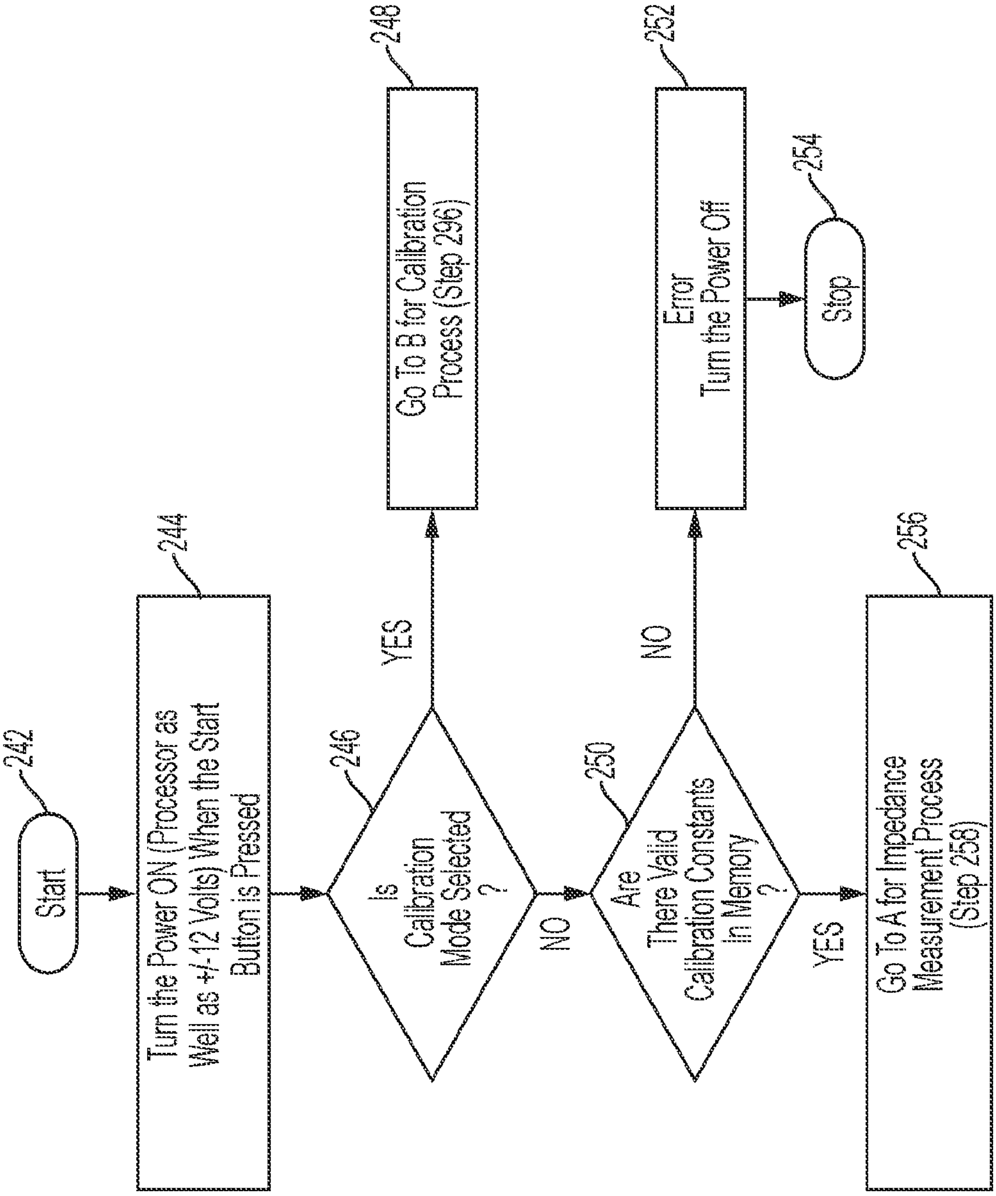

242
Start
244
Turn the Power ON (Processor as Well as +/-12 Volts) When the Start Button is Pressed
246
Is Calibration Mode Selected ?
YES
248
Go To B for Calibration Process (Step 296)
NO
250
Are There Valid Calibration Constants In Memory ?
NO
252
Error Turn the Power Off
254
Stop
YES
256
Go To A for Impedance Measurement Process (Step 258)

FIGURE 6A

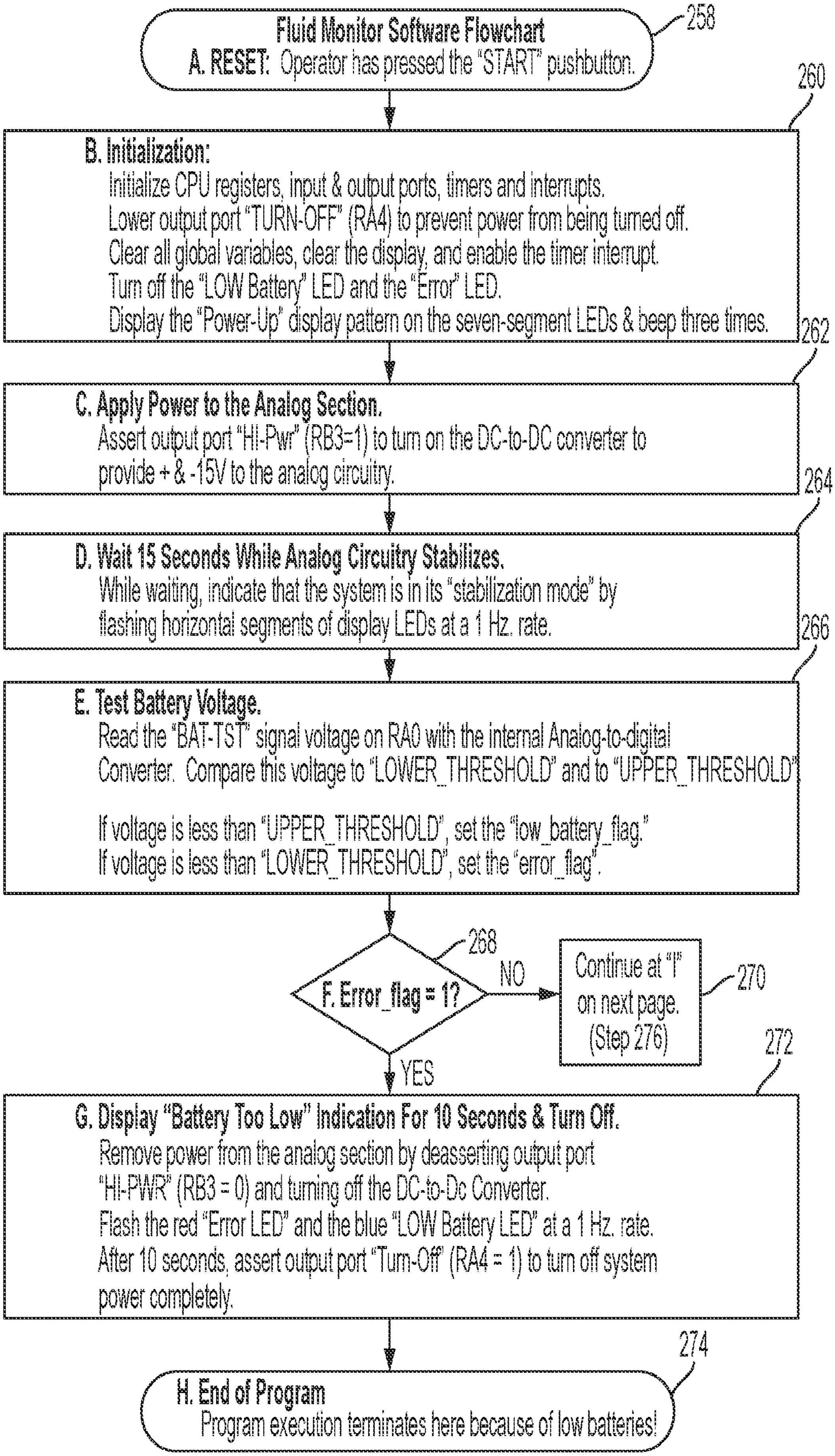

Fluid Monitor Software Flowchart
A. RESET: Operator has pressed the "START" pushbutton.
258
260
B. Initialization:
Initialize CPU registers, input & output ports, timers and interrupts.
Lower output port "TURN-OFF" (RA4) to prevent power from being turned off.
Clear all global variables, clear the display, and enable the timer interrupt.
Turn off the "LOW Battery" LED and the "Error" LED.
Display the "Power-Up" display pattern on the seven-segment LEDs & beep three times.
262
C. Apply Power to the Analog Section.
Assert output port "HI-Pwr" (RB3=1) to turn on the DC-to-DC converter to provide + & -15V to the analog circuitry.
264
D. Wait 15 Seconds While Analog Circuitry Stabilizes.
While waiting, indicate that the system is in its "stabilization mode" by flashing horizontal segments of display LEDs at a 1 Hz. rate.
266
E. Test Battery Voltage.
Read the "BAT-TST" signal voltage on RA0 with the internal Analog-to-digital Converter. Compare this voltage to "LOWER_THRESHOLD" and to "UPPER_THRESHOLD".
If voltage is less than "UPPER_THRESHOLD", set the "low_battery_flag."
If voltage is less than "LOWER_THRESHOLD", set the "error_flag".
268
F. Error_flag = 1?
NO
Continue at "I" on next page. (Step 276)
270
YES
272
G. Display "Battery Too Low" Indication For 10 Seconds & Turn Off.
Remove power from the analog section by deasserting output port "HI-PWR" (RB3 = 0) and turning off the DC-to-Dc Converter.
Flash the red "Error LED" and the blue "LOW Battery LED" at a 1 Hz. rate.
After 10 seconds, assert output port "Turn-Off" (RA4 = 1) to turn off system power completely.
274
H. End of Program
Program execution terminates here because of low batteries!

FIGURE 6B

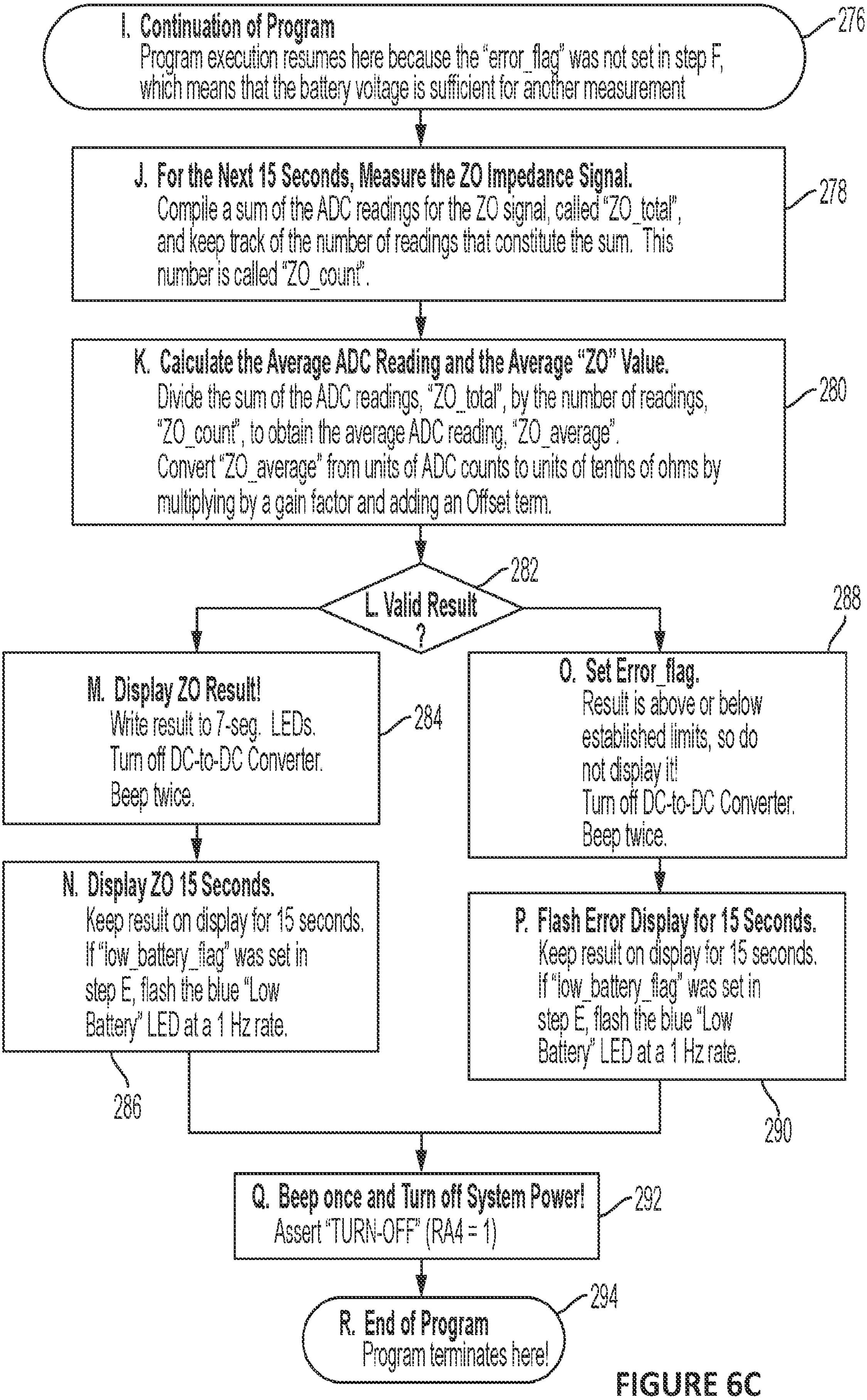

I. Continuation of Program
Program execution resumes here because the "error_flag" was not set in step F, which means that the battery voltage is sufficient for another measurement
276
J. For the Next 15 Seconds, Measure the ZO Impedance Signal.
Compile a sum of the ADC readings for the ZO signal, called "ZO_total", and keep track of the number of readings that constitute the sum. This number is called "ZO_count".
278
K. Calculate the Average ADC Reading and the Average "ZO" Value.
Divide the sum of the ADC readings, "ZO_total", by the number of readings, "ZO_count", to obtain the average ADC reading, "ZO_average".
Convert "ZO_average" from units of ADC counts to units of tenths of ohms by multiplying by a gain factor and adding an Offset term.
280
L. Valid Result ?
282
M. Display ZO Result!
Write result to 7-seg. LEDs.
Turn off DC-to-DC Converter.
Beep twice.
284
O. Set Error_flag.
Result is above or below established limits, so do not display it!
Turn off DC-to-DC Converter.
Beep twice.
288
N. Display ZO 15 Seconds.
Keep result on display for 15 seconds.
If "low_battery_flag" was set in step E, flash the blue "Low Battery" LED at a 1 Hz rate.
286
P. Flash Error Display for 15 Seconds.
Keep result on display for 15 seconds.
If "low_battery_flag" was set in step E, flash the blue "Low Battery" LED at a 1 Hz rate.
290
Q. Beep once and Turn off System Power!
Assert "TURN-OFF" (RA4 = 1)
292
R. End of Program
Program terminates here!
294

FIGURE 6C

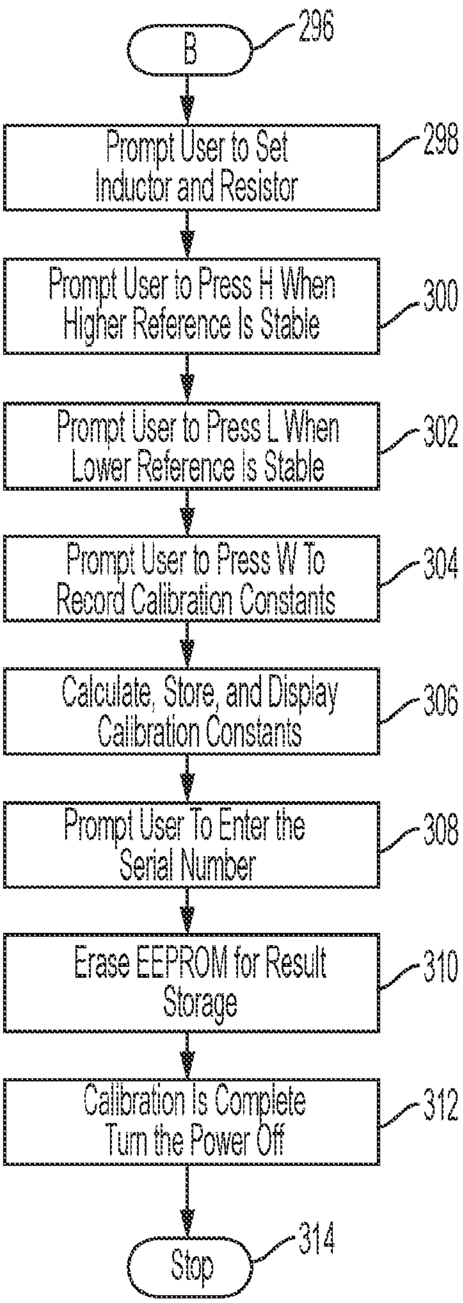

B
296
Prompt User to Set Inductor and Resistor
298
Prompt User to Press H When Higher Reference Is Stable
300
Prompt User to Press L When Lower Reference Is Stable
302
Prompt User to Press W To Record Calibration Constants
304
Calculate, Store, and Display Calibration Constants
306
Prompt User To Enter the Serial Number
308
Erase EEPROM for Result Storage
310
Calibration Is Complete Turn the Power Off
312
Stop
314

FIGURE 6D

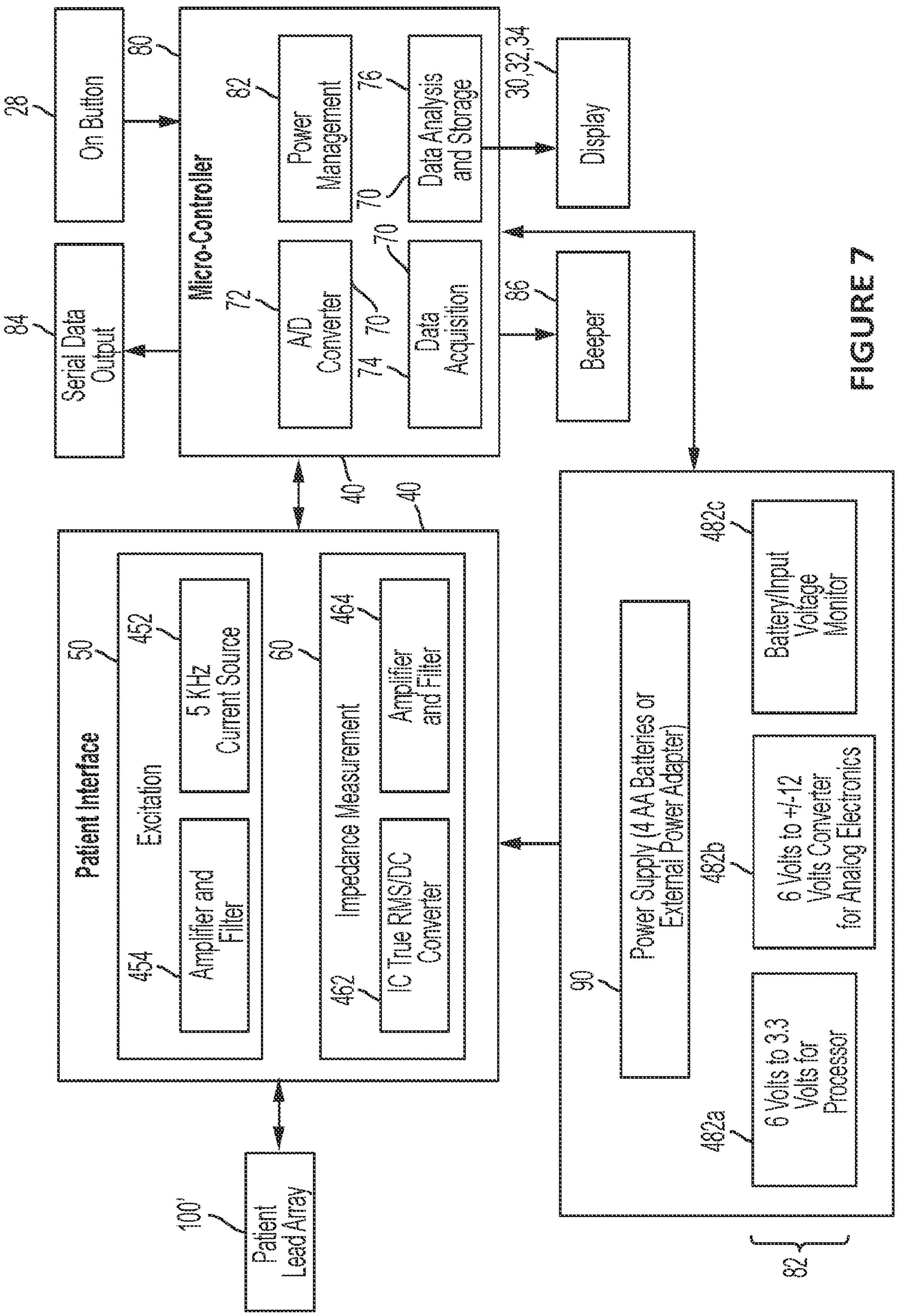
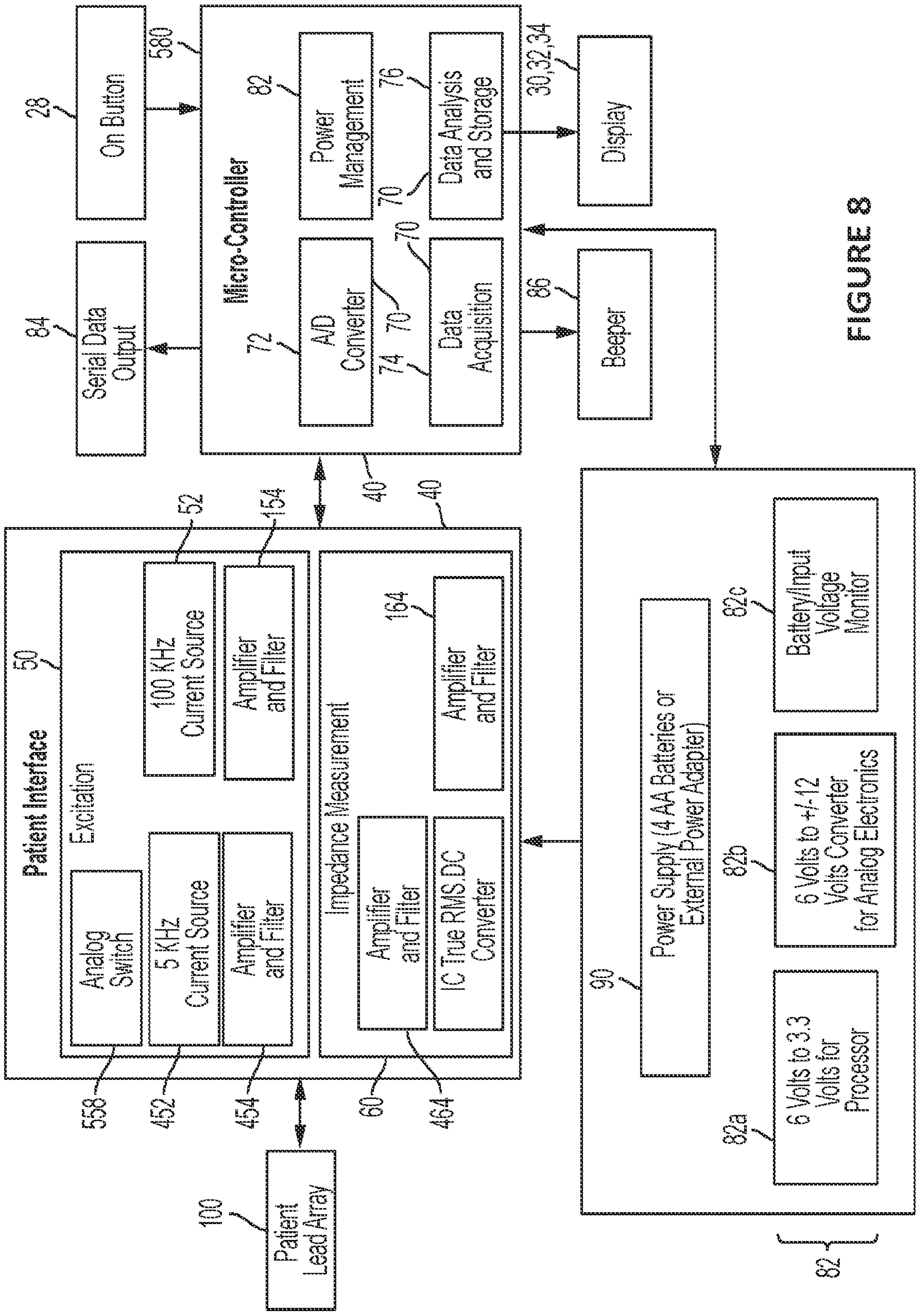

28
On Button
84
Serial Data Output
80
Micro-Controller
82
Power Management
72
A/D Converter
76
Data Analysis and Storage
70
74
Data Acquisition
30,32,34
Display
86
Beeper
40
Patient Interface
50
Excitation
452
5 KHz Current Source
454
Amplifier and Filter
60
Impedance Measurement
464
Amplifier and Filter
462
IC True RMS/DC Converter
100'
Patient Lead Array
90
Power Supply (4 AA Batteries or External Power Adapter)
482a
6 Volts to 3.3 Volts for Processor
482b
6 Volts to +/-12 Volts Converter for Analog Electronics
482c
Battery/Input Voltage Monitor
82

FIGURE 7



28
On Button
580
84
Serial Data Output
Micro-Controller
82
Power Management
76
Data Analysis and Storage
70
72
A/D Converter
74
Data Acquisition
30,32,34
Display
86
Beeper
40
52
154
50
Patient Interface
Excitation
100 KHz Current Source
Amplifier and Filter
Analog Switch
5 KHz Current Source
Amplifier and Filter
Impedance Measurement
164
Amplifier and Filter
IC True RMS DC Converter
558
452
454
60
464
100
Patient Lead Array
90
Power Supply (4 AA Batteries or External Power Adapter)
82c
Battery/Input Voltage Monitor
82b
6 Volts to +/-12 Volts Converter for Analog Electronics
82a
6 Volts to 3.3 Volts for Processor
82

FIGURE 8

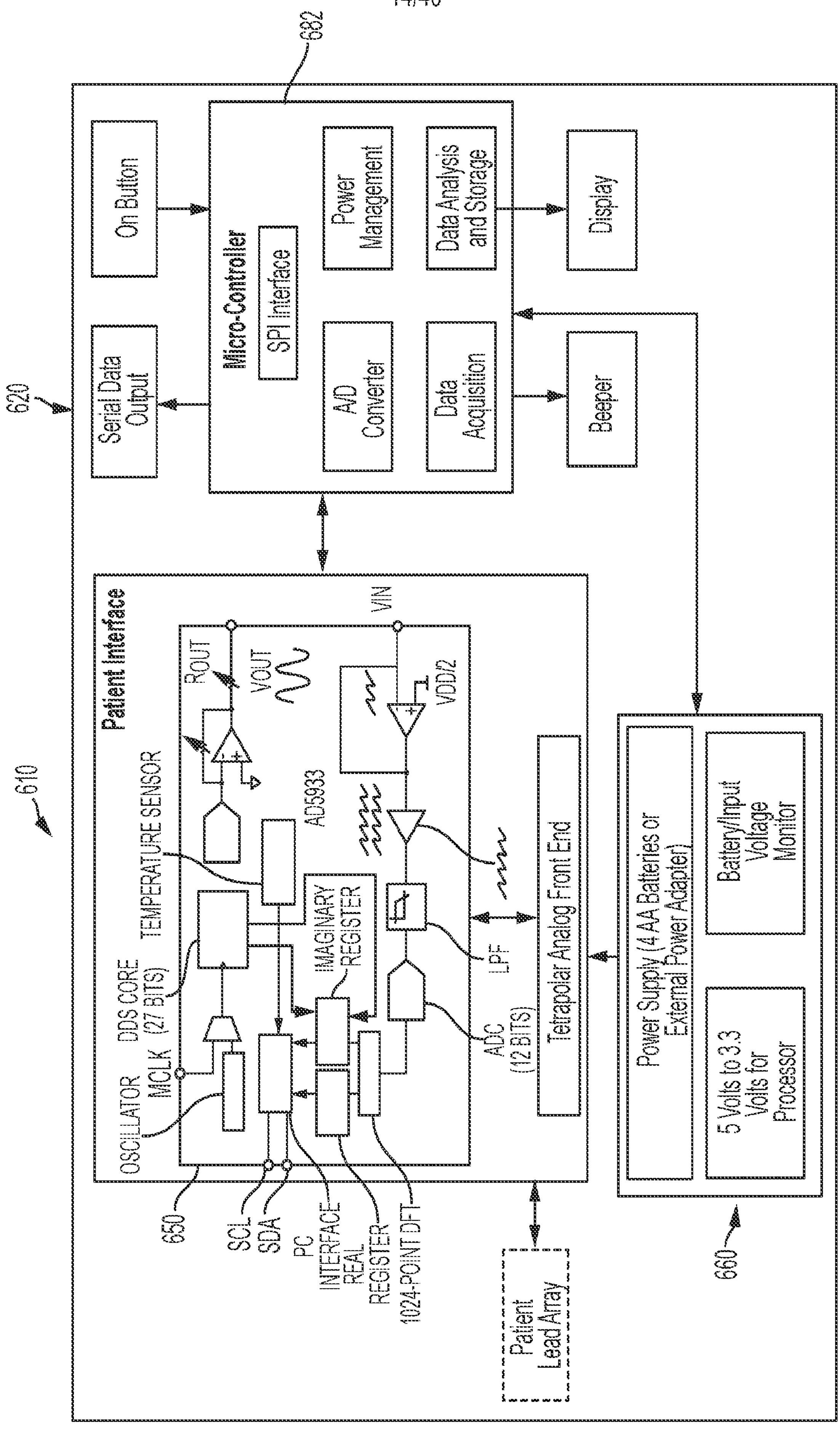

682
620
610
On Button
Serial Data Output
Micro-Controller
SPI Interface
Power Management
Data Analysis and Storage
A/D Converter
Data Acquisition
Display
Beeper
Patient Interface
VIN
ROUT
VOUT
VDD/2
TEMPERATURE SENSOR
AD5933
DDS CORE (27 BITS)
IMAGINARY REGISTER
OSCILLATOR
MCLK
LPF
ADC (12 BITS)
Tetrapolar Analog Front End
650
SCL
SDA
PC INTERFACE
REAL REGISTER
1024-POINT DFT
Patient Lead Array
Power Supply (4 AA Batteries or External Power Adapter)
Battery/Input Voltage Monitor
5 Volts to 3.3 Volts for Processor
660

FIGURE 9

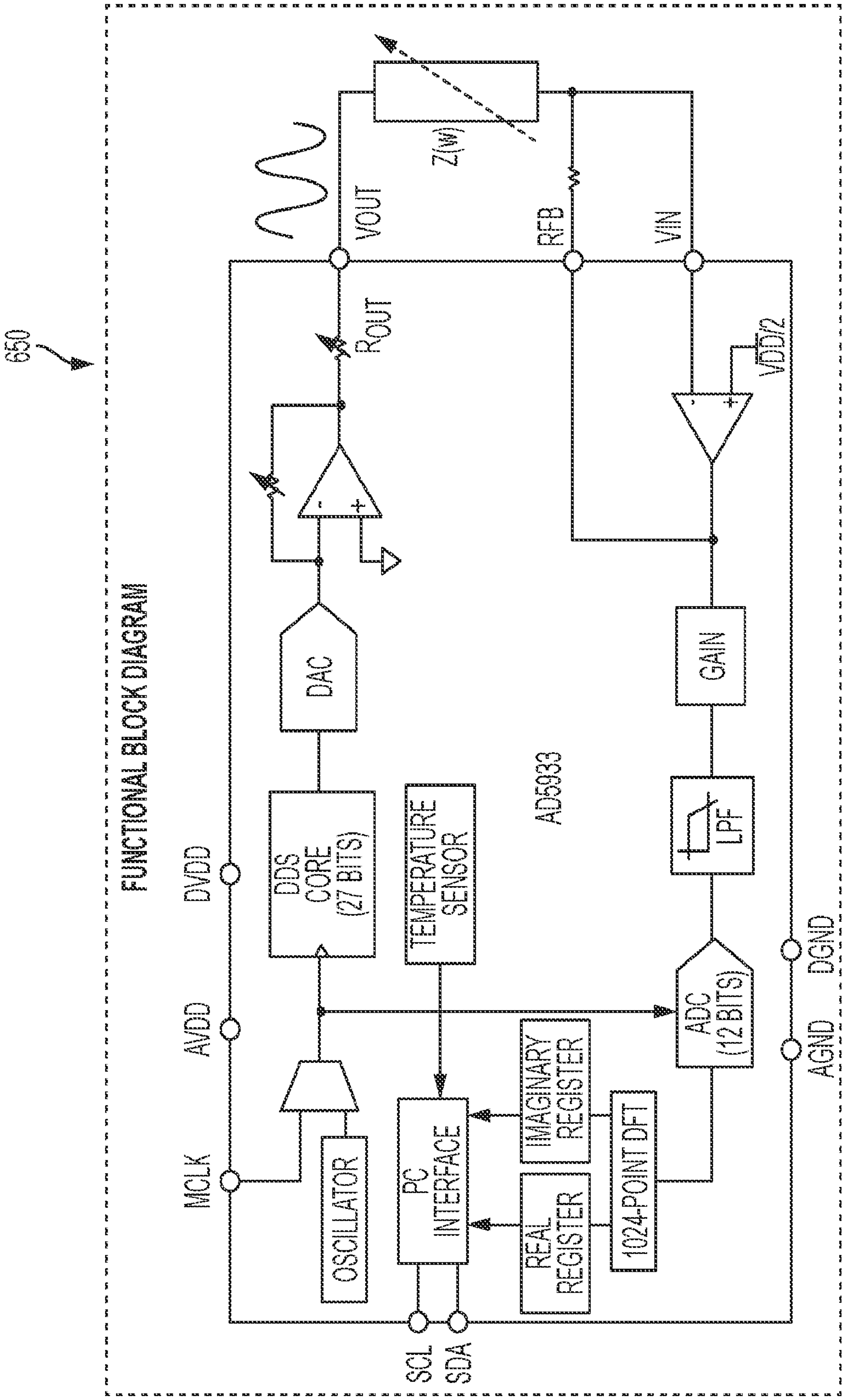

650
FUNCTIONAL BLOCK DIAGRAM
MCLK
AVDD
DVDD
OSCILLATOR
DDS
CORE
(27 BITS)
DAC
TEMPERATURE
SENSOR
AD5933
PC
INTERFACE
REAL
REGISTER
IMAGINARY
REGISTER
1024-POINT DFT
ADC
(12 BITS)
LPF
GAIN
ROUT
VDD/2
VOUT
RFB
VIN
Z(w)
SCL
SDA
AGND
DGND

FIGURE 10

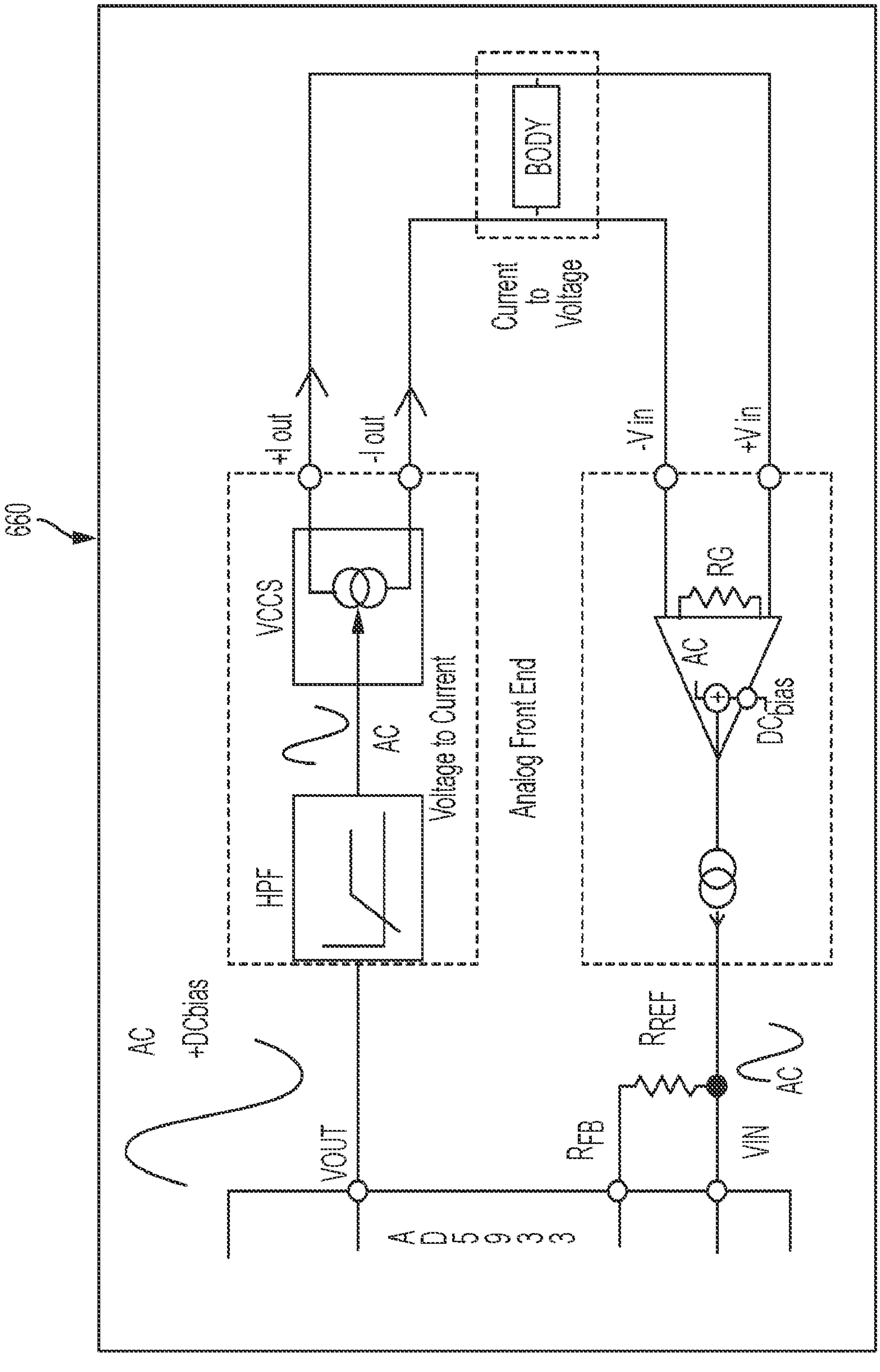

660
BODY
Current
to
Voltage
+I out
-I out
-V in
+V in
VCCS
RG
AC
DCbias
Voltage to Current
Analog Front End
HPF
AC
AC
+DCbias
VOUT
RFB
RREF
VIN
AC
AD5933

FIGURE 11

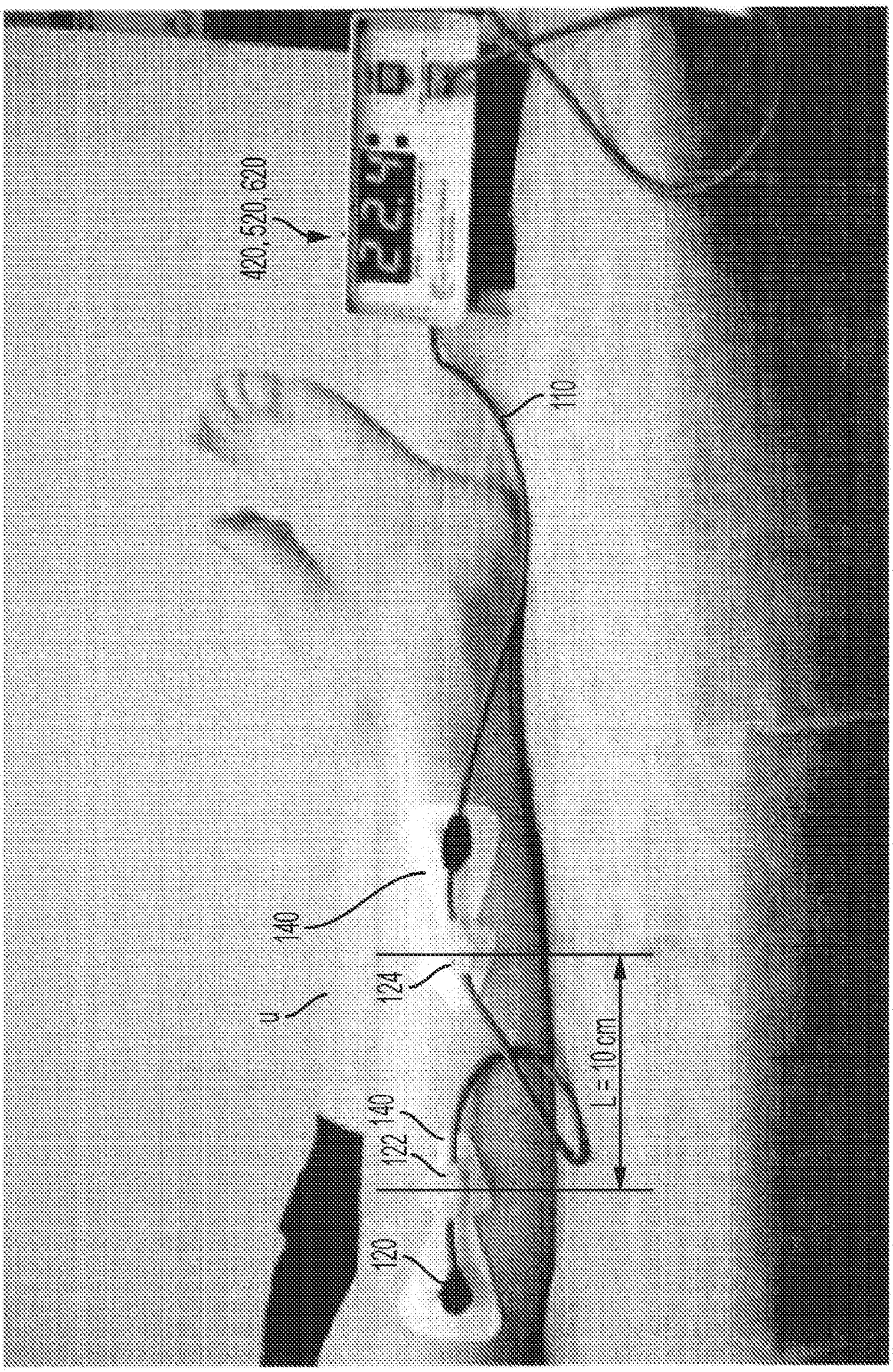


FIGURE 12

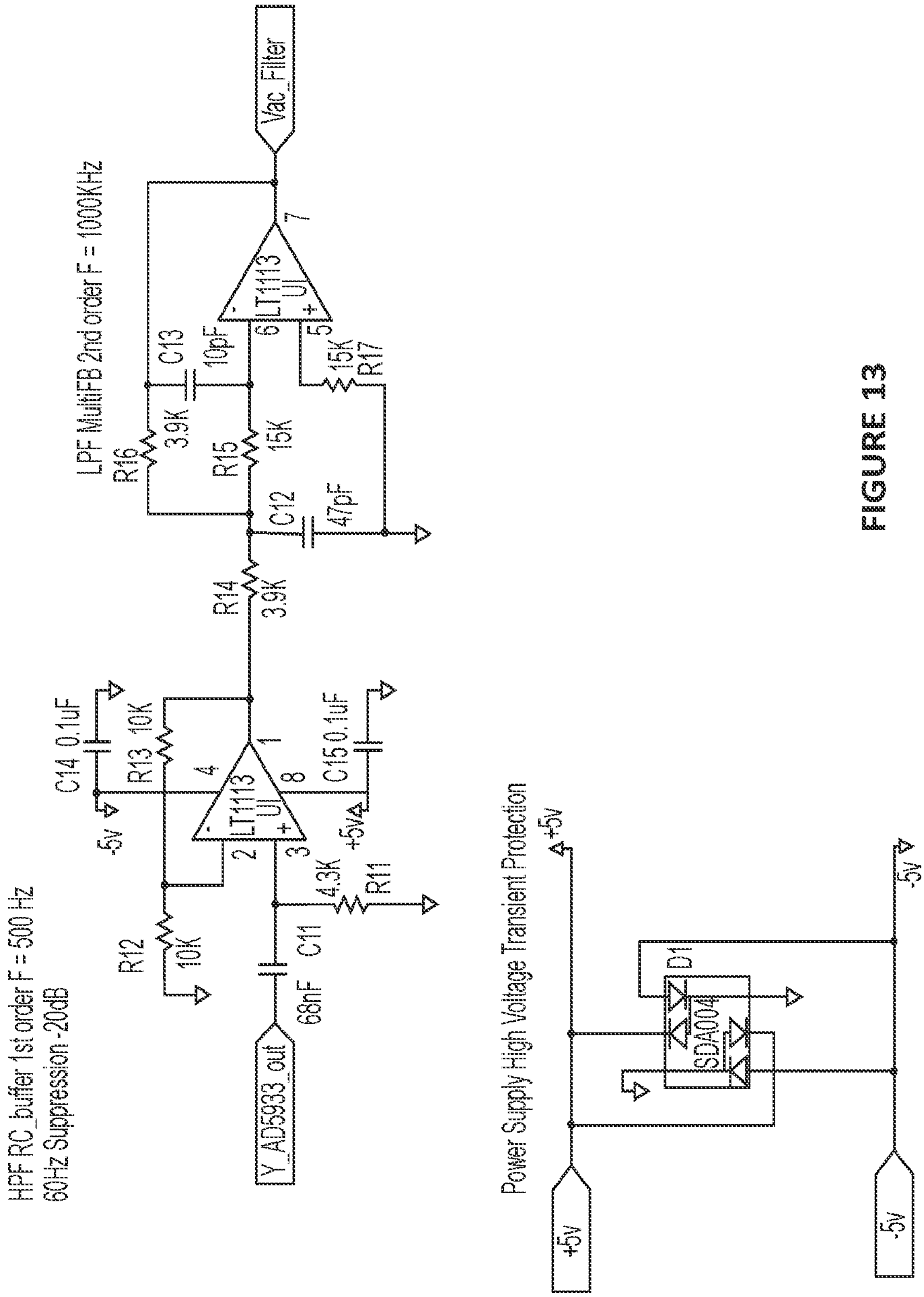
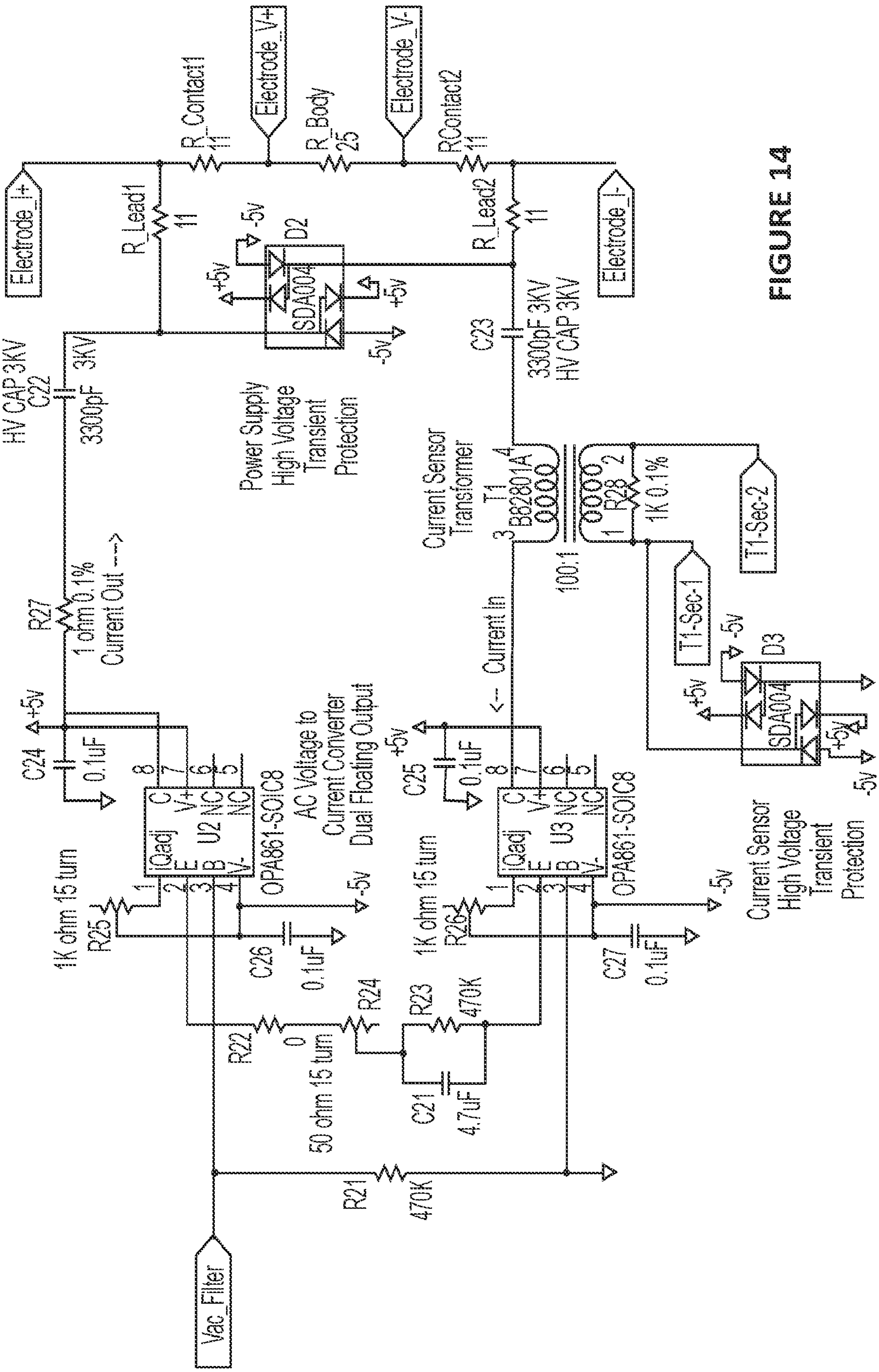

HPF RC_buffer 1st order F = 500 Hz
60Hz Suppression -20dB
LPF MultiFB 2nd order F = 1000KHz
Vac_Filter
Y_AD5933_out
LT1113
UI
R12
10K
R13
10K
C14
0.1uF
C15
0.1uF
C11
68nF
R11
4.3K
-5v
+5v
R14
3.9K
C12
47pF
R15
15K
R16
3.9K
C13
10pF
R17
15K
Power Supply High Voltage Transient Protection
D1
SDA004
+5v
-5v

FIGURE 13



Electrode_I+
Electrode_I-
Electrode_V+
Electrode_V-
R_Contact1
11
R_Body
25
RContact2
11
R_Lead1
11
R_Lead2
11
D2
SDA004
+5v
-5v
Power Supply
High Voltage
Transient
Protection
HV CAP 3KV
C22
3300pF 3KV
C23
3300pF 3KV
HV CAP 3KV
Current Sensor
Transformer
T1
B82801A
100:1
R28
1K 0.1%
T1-Sec-1
T1-Sec-2
D3
SDA004
Current Sensor
High Voltage
Transient
Protection
R27
1 ohm 0.1%
Current Out --->
<--- Current In
C24
0.1uF
C25
0.1uF
IQadj
E
B
V-
C
V+
NC
U2
U3
OPA861-SOIC8
AC Voltage to
Current Converter
Dual Floating Output
R25
1K ohm 15 turn
R26
1K ohm 15 turn
C26
0.1uF
C27
0.1uF
R22
0
R24
50 ohm 15 turn
R23
470K
C21
4.7uF
R21
470K
Vac_Filter

FIGURE 14

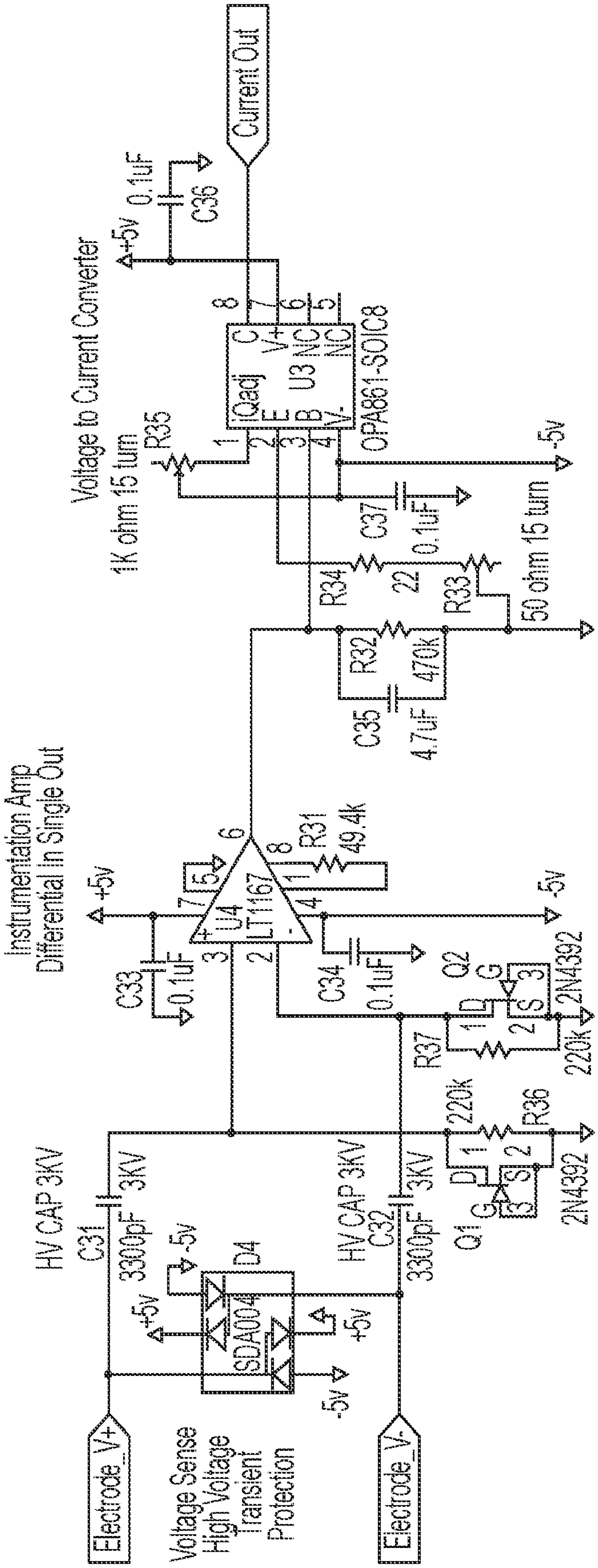

Current Out
0.1uF
C36
+5v
Voltage to Current Converter
1K ohm 15 turn
R35
IQadj
E
B
V-
C
V+
NC
NC
U3
OPA861-SOIC8
C37
0.1uF
-5v
R34
22
R33
50 ohm 15 turn
R32
470k
C35
4.7uF
Instrumentation Amp
Differential In Single Out
U4
LT1167
R31
49.4k
C33
0.1uF
C34
0.1uF
Q2
2N4392
R37
220k
R36
220k
Q1
2N4392
HV CAP 3KV
C31
3300pF
3KV
HV CAP 3KV
C32
3300pF
3KV
D4
SDA004
Electrode_V+
Electrode_V-
Voltage Sense
High Voltage
Transient
Protection

FIGURE 15

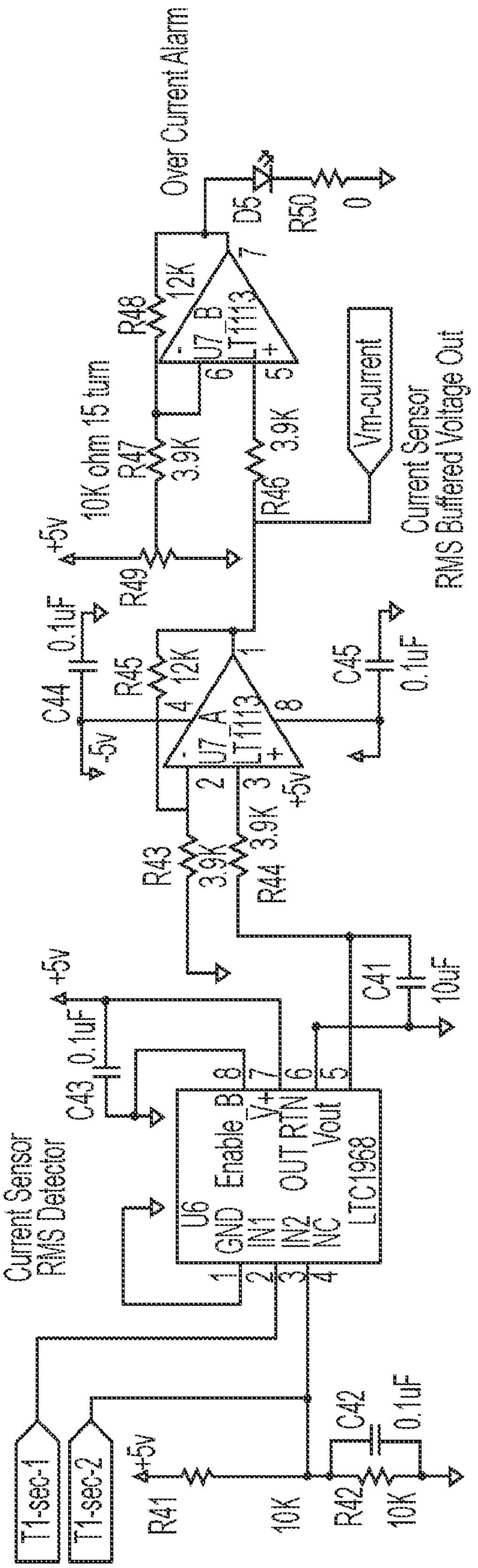

Over Current Alarm
D5
R50
0
R48
12K
U7_B
LT1113
7
6
5
10K ohm 15 turn
R47
3.9K
R46
3.9K
Vm-current
Current Sensor
RMS Buffered Voltage Out
+5v
R49
C44
0.1uF
R45
12K
U7_A
LT1113
-5v
4
1
2
3
8
+5v
C45
0.1uF
R43
3.9K
R44
3.9K
C41
10uF
Current Sensor
RMS Detector
C43
0.1uF
U6
GND
IN1
IN2
NC
Enable_B
V+
OUT RTN
Vout
LTC1968
T1-sec-1
T1-sec-2
R41
+5v
10K
R42
10K
C42
0.1uF

FIGURE 16

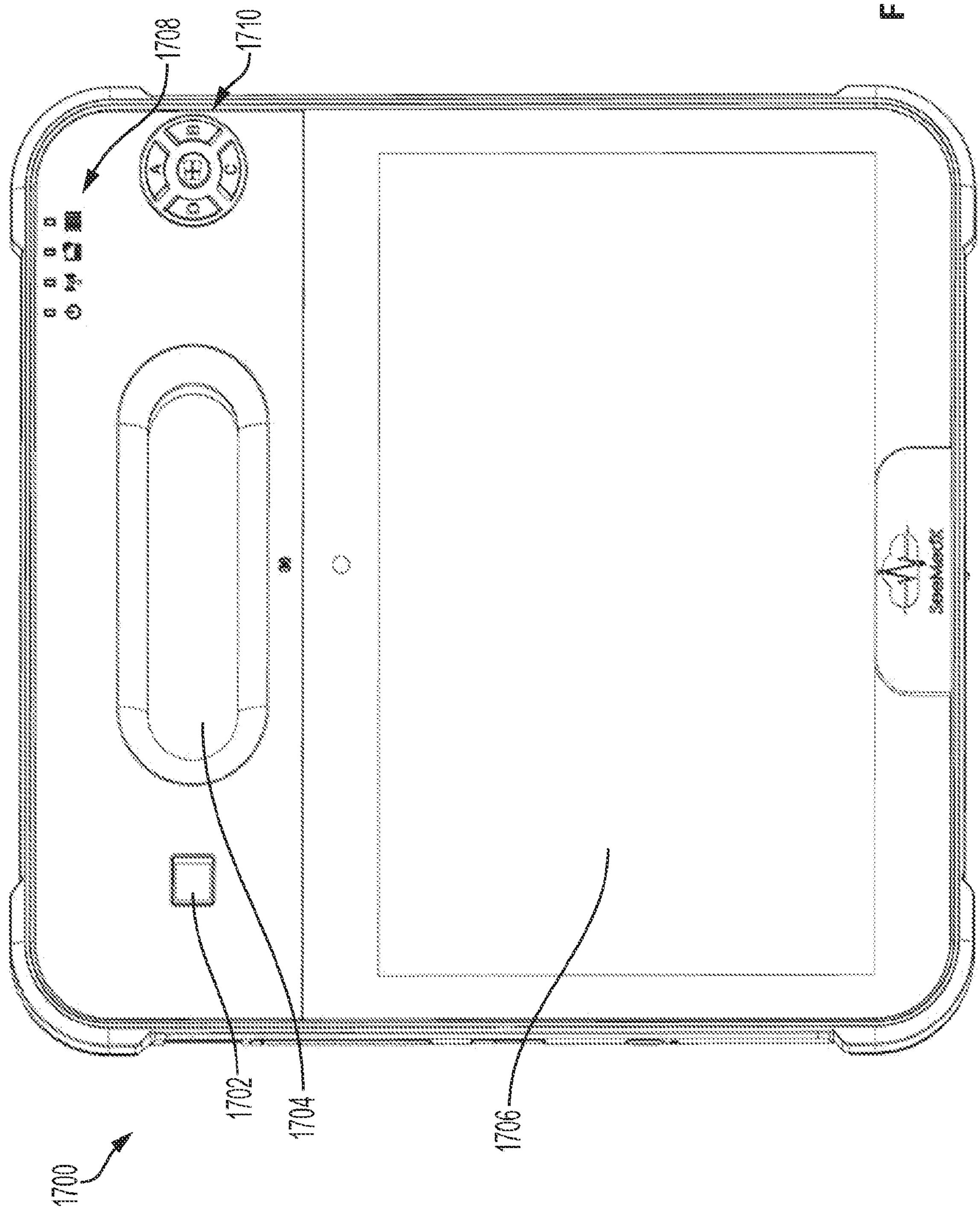


FIGURE 17A

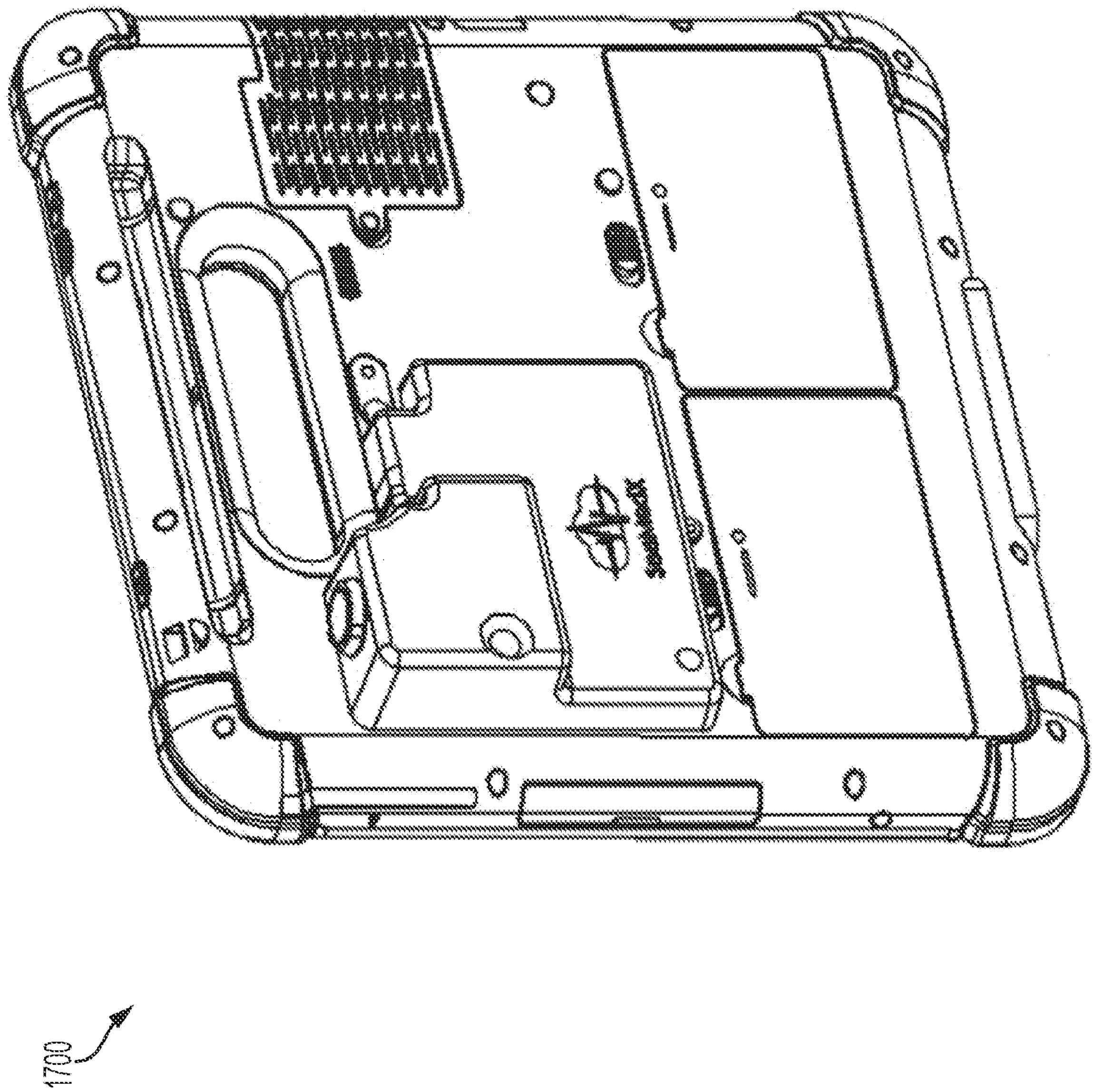


FIGURE 17B

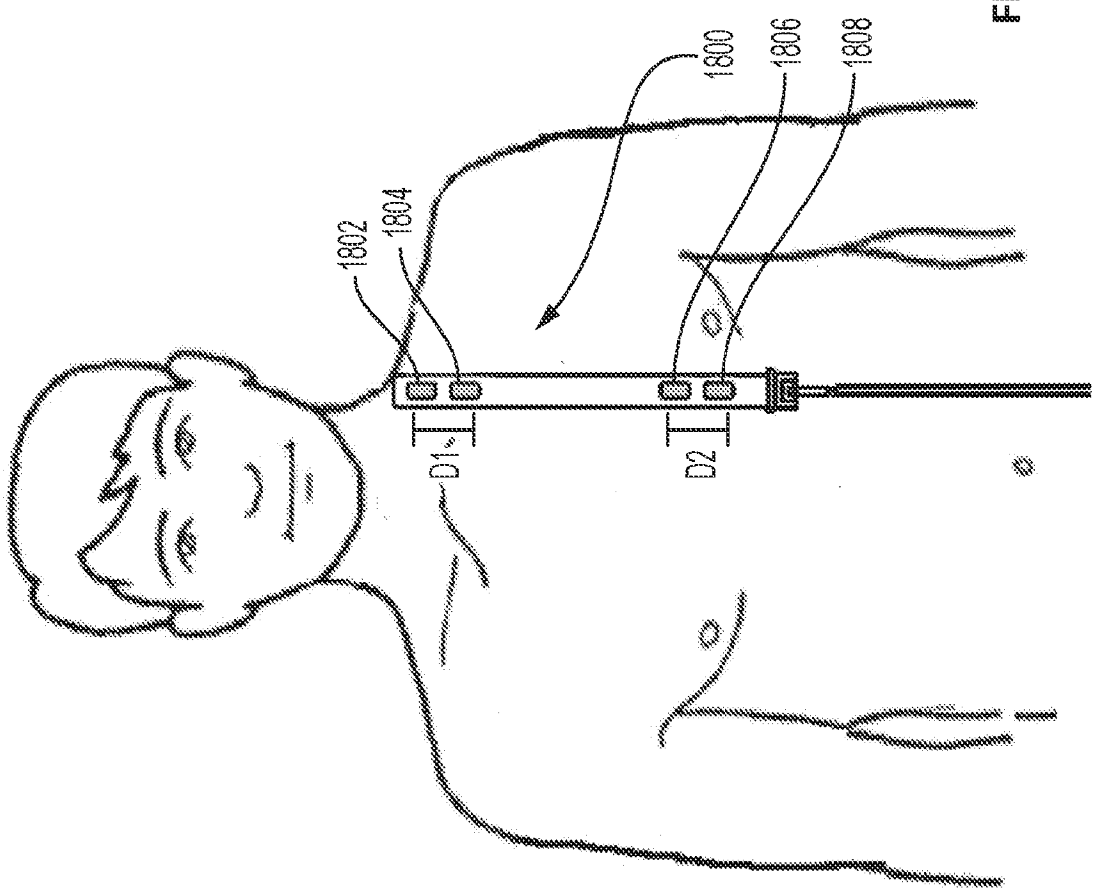


FIGURE 18

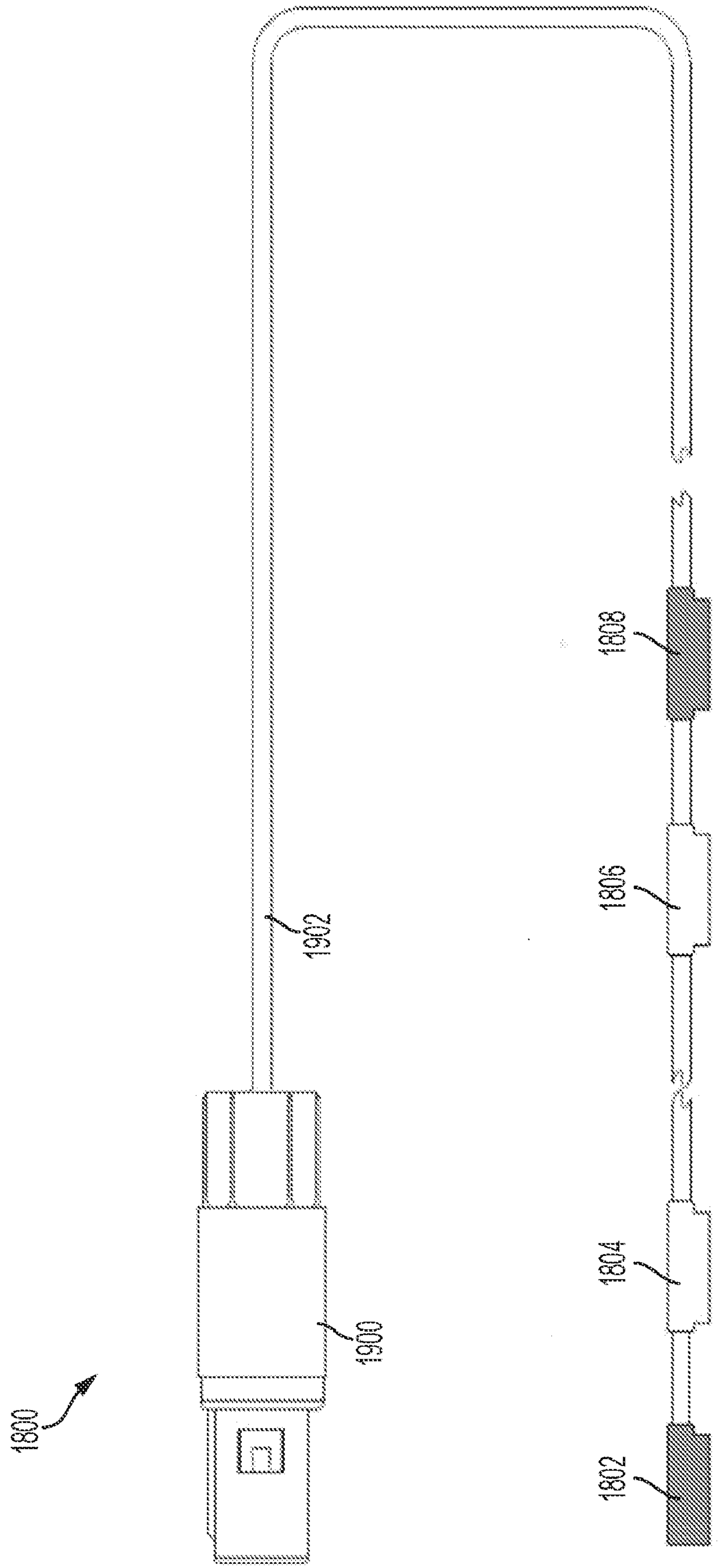


FIGURE 19A

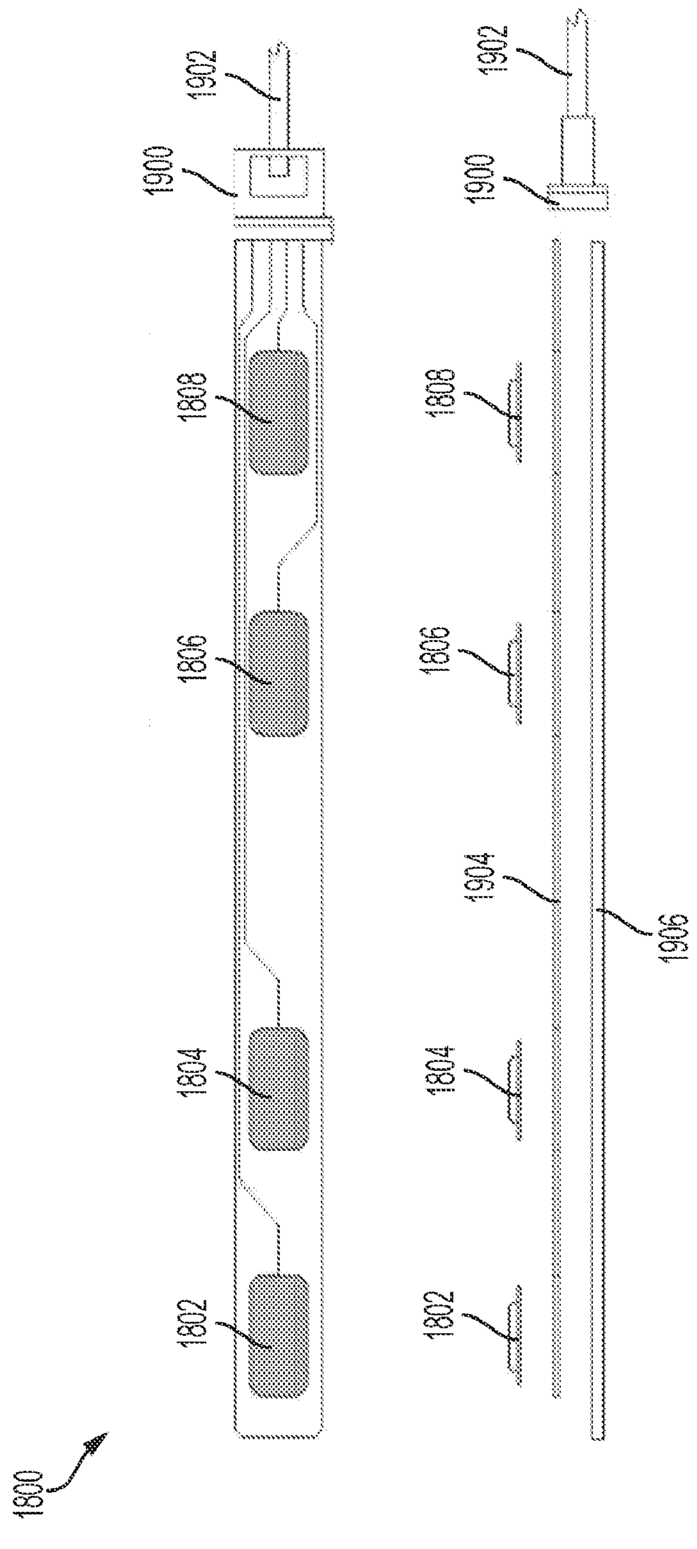

1902
1900
1808
1806
1804
1802
1902
1900
1808
1806
1904
1906
1804
1802
1800

FIGURE 19B

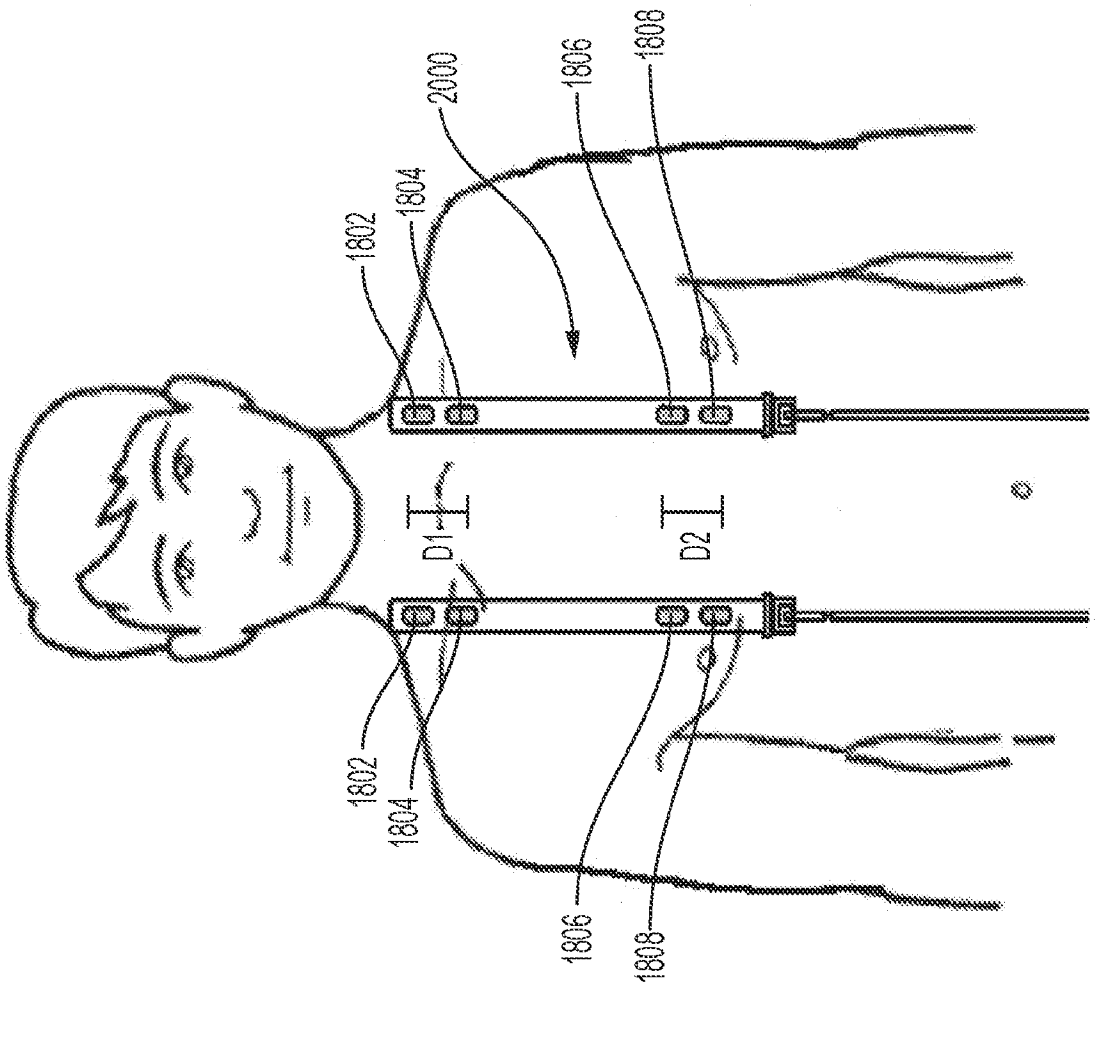


FIGURE 20

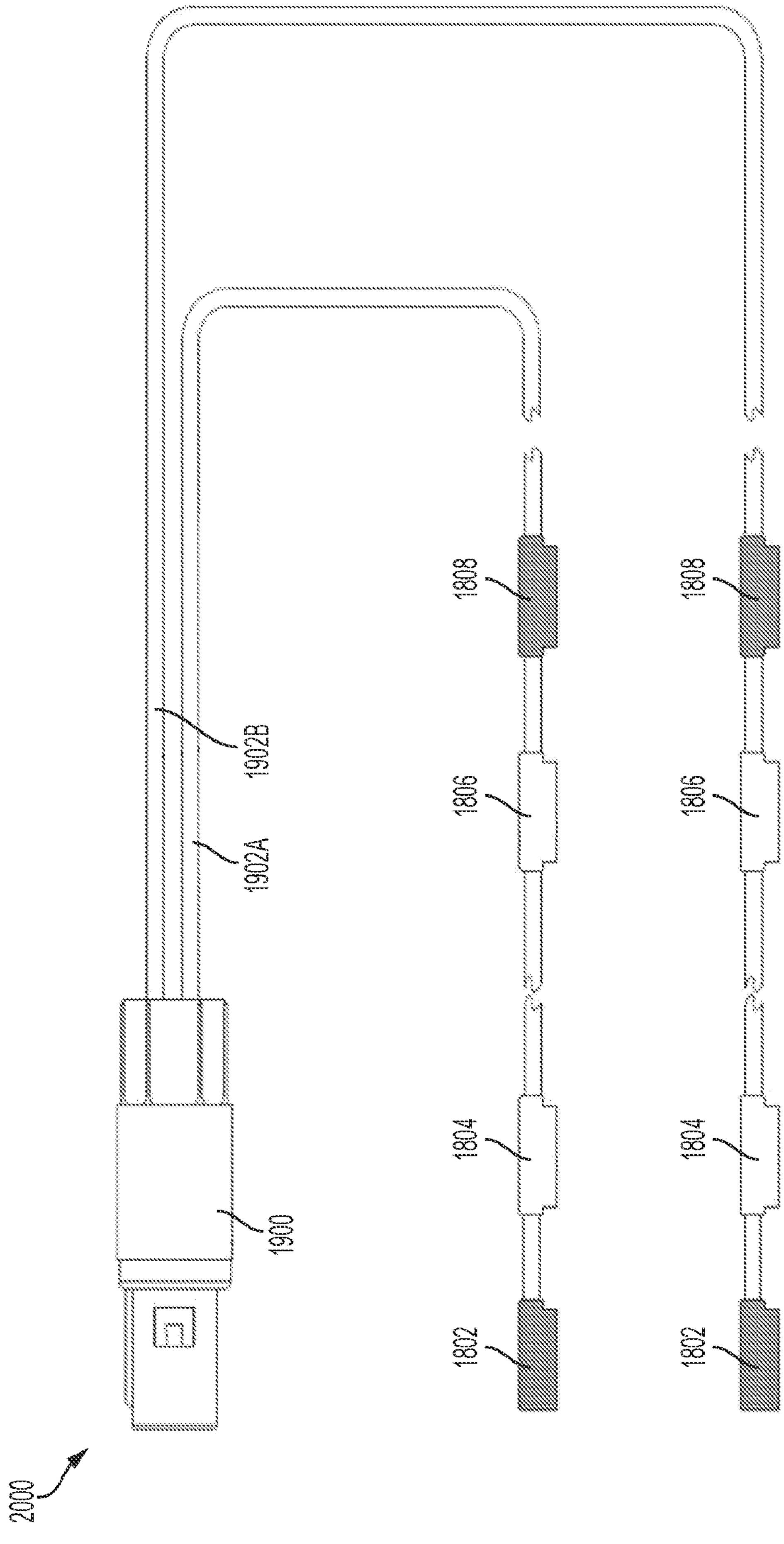

1808
1808
1902B
1806
1806
1902A
1804
1804
1900
1802
1802
2000

FIGURE 21A

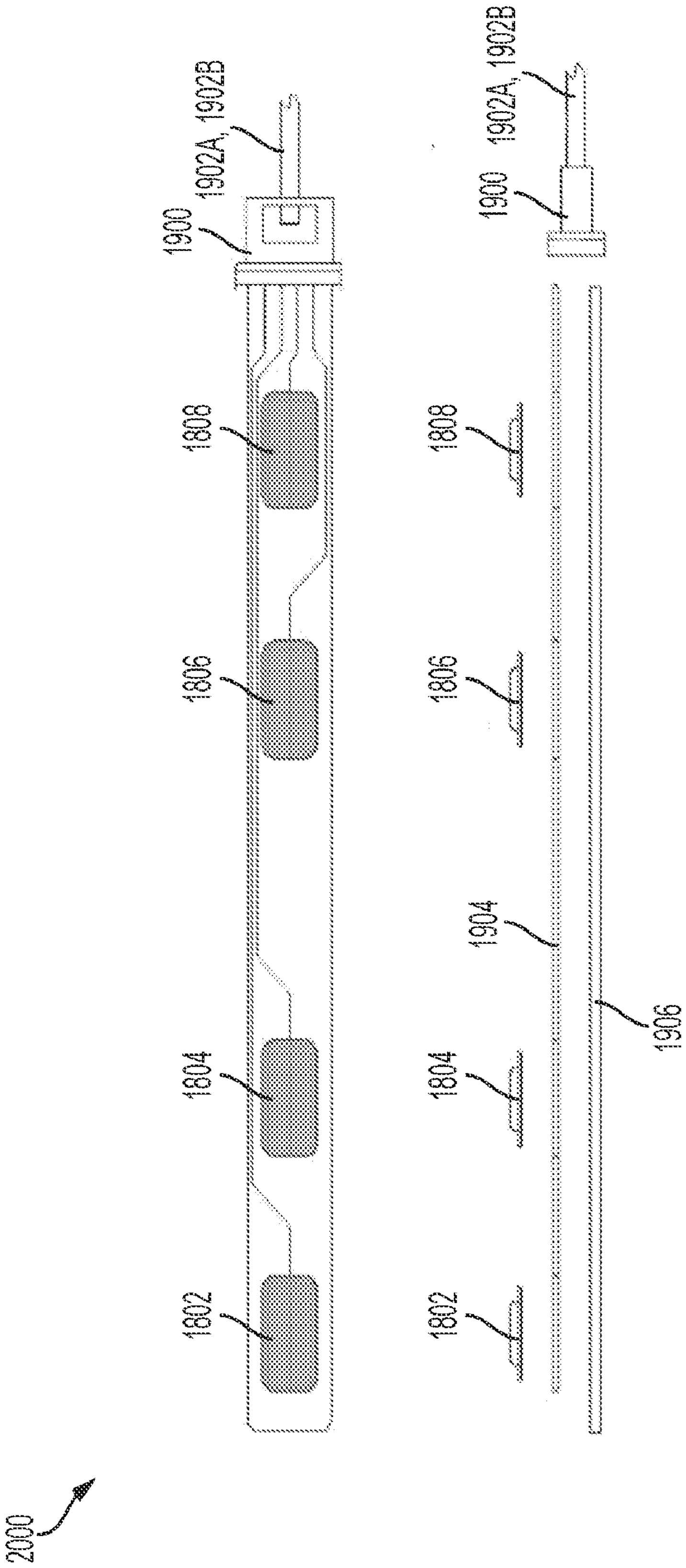

1902A, 1902B
1900
1808
1806
1804
1802
1902A, 1902B
1900
1808
1806
1904
1906
1804
1802
2000

FIGURE 21B

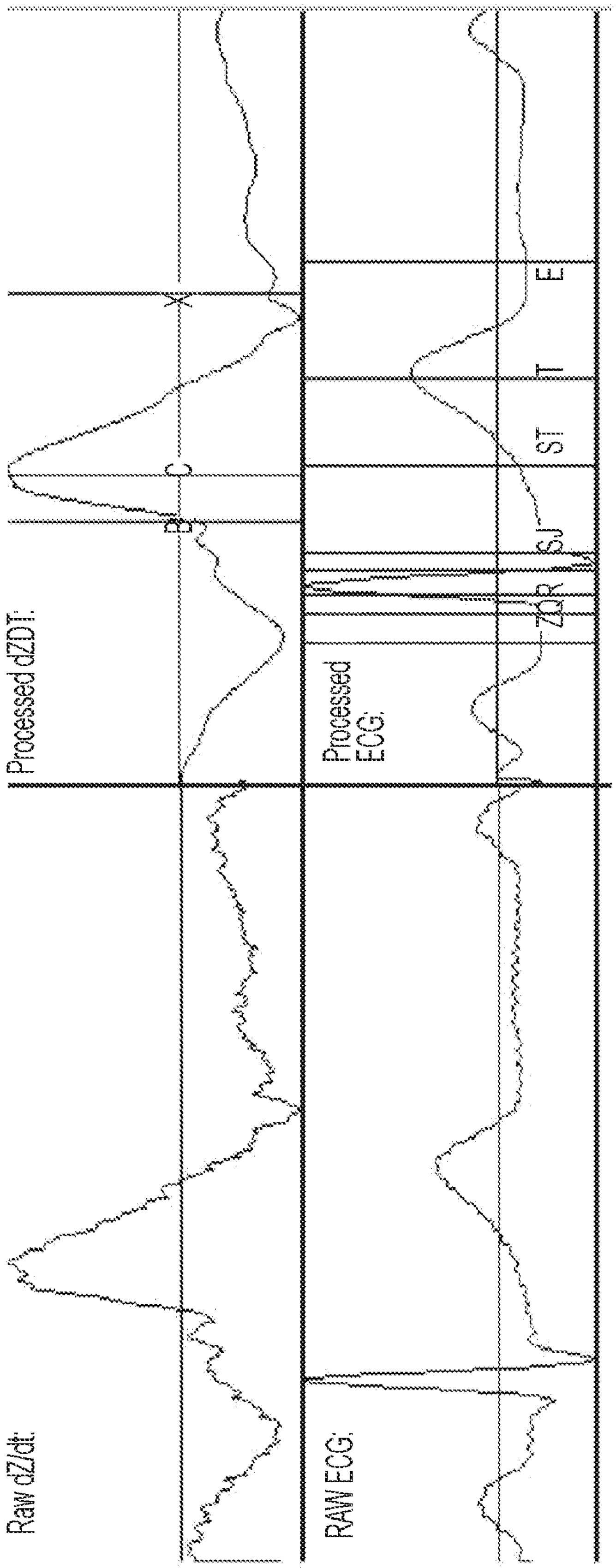

Raw dZ/dt:
Processed dZDT:
RAW ECG:
Processed ECG:
B
C
X
Z
Q
R
SJ
ST
T
E

FIGURE 21C

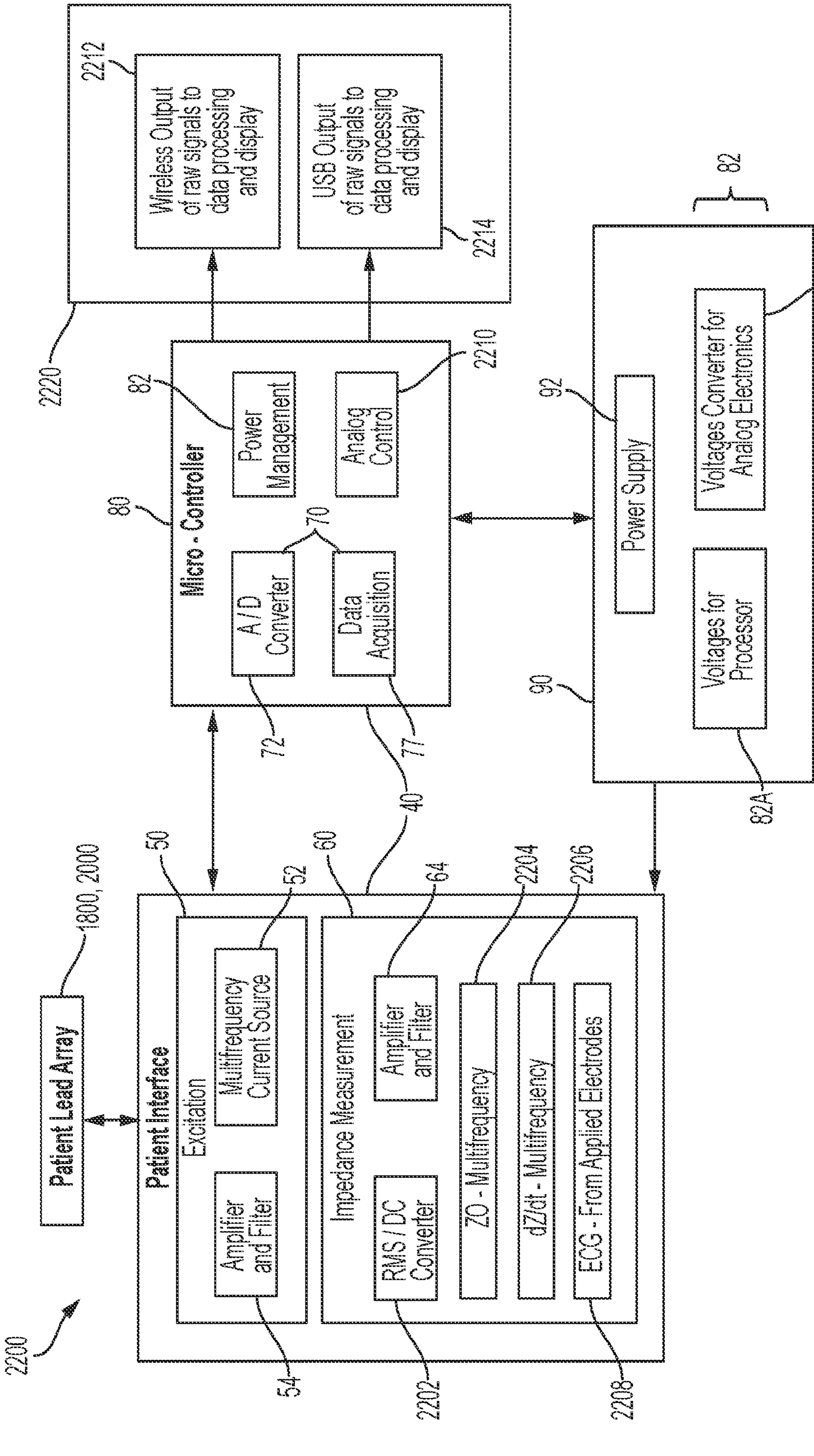

2200
Patient Lead Array
1800, 2000
Patient Interface
Excitation
50
Amplifier and Filter
54
Multifrequency Current Source
52
Impedance Measurement
60
RMS / DC Converter
2202
Amplifier and Filter
64
ZO - Multifrequency
2204
dZ/dt - Multifrequency
2206
ECG - From Applied Electrodes
2208
40
Micro - Controller
80
A / D Converter
72
Data Acquisition
77
70
Power Management
82
Analog Control
2210
2220
Wireless Output of raw signals to data processing and display
2212
USB Output of raw signals to data processing and display
2214
90
Power Supply
92
Voltages for Processor
82A
Voltages Converter for Analog Electronics
82B
82

FIGURE 22

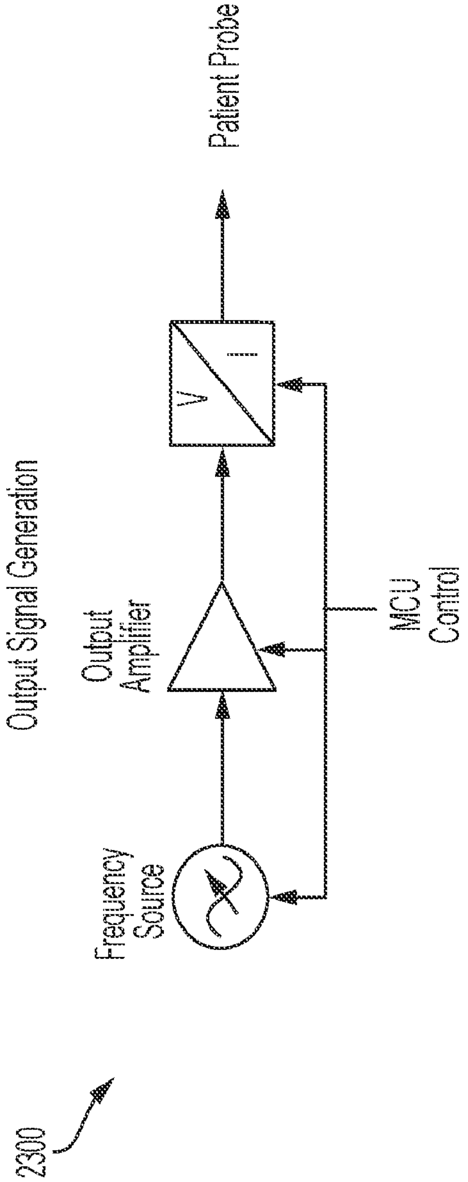


FIGURE 23

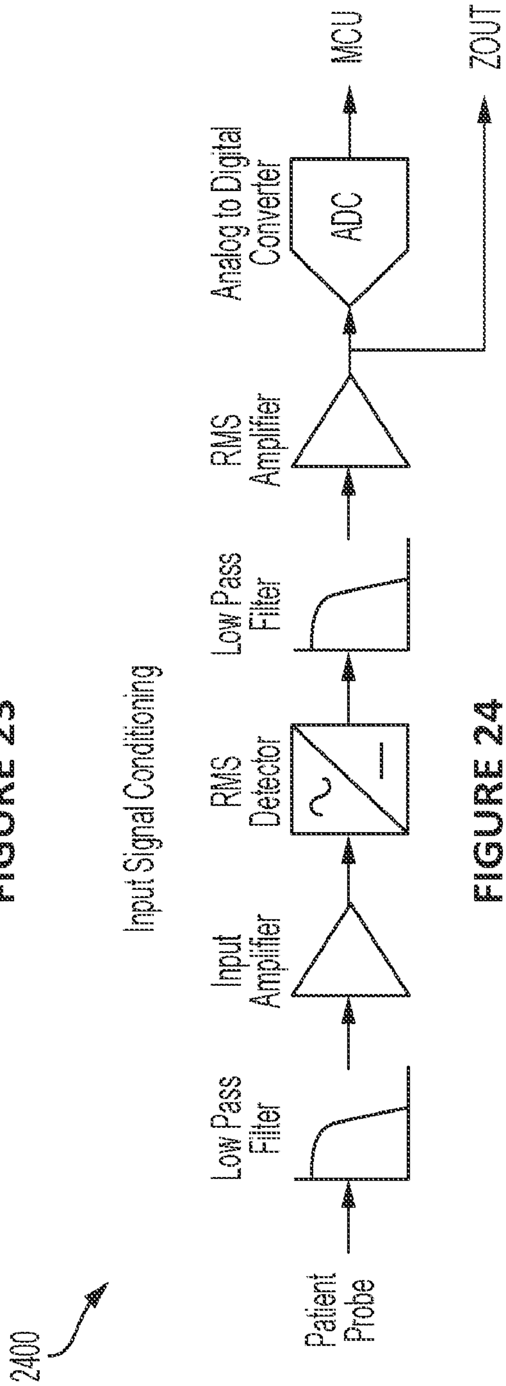


FIGURE 24

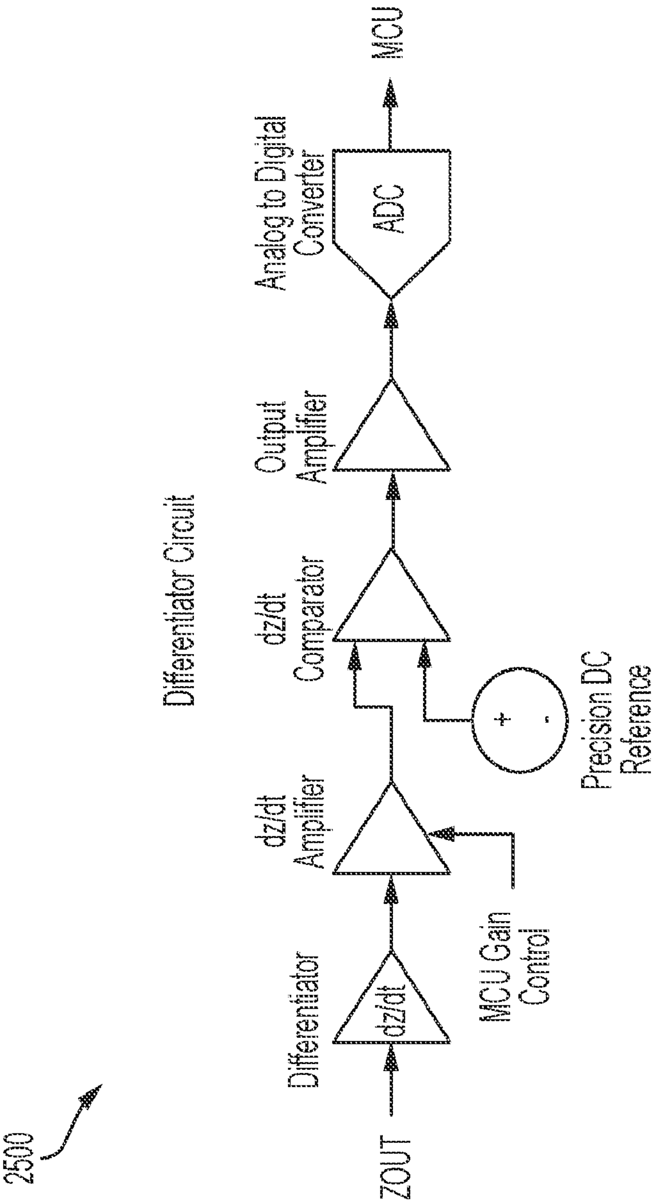

Differentiator Circuit
2500
ZOUT
Differentiator
dz/dt
dz/dt
Amplifier
MCU Gain
Control
dz/dt
Comparator
+
-
Precision DC
Reference
Output
Amplifier
Analog to Digital
Converter
ADC
MCU

FIGURE 25

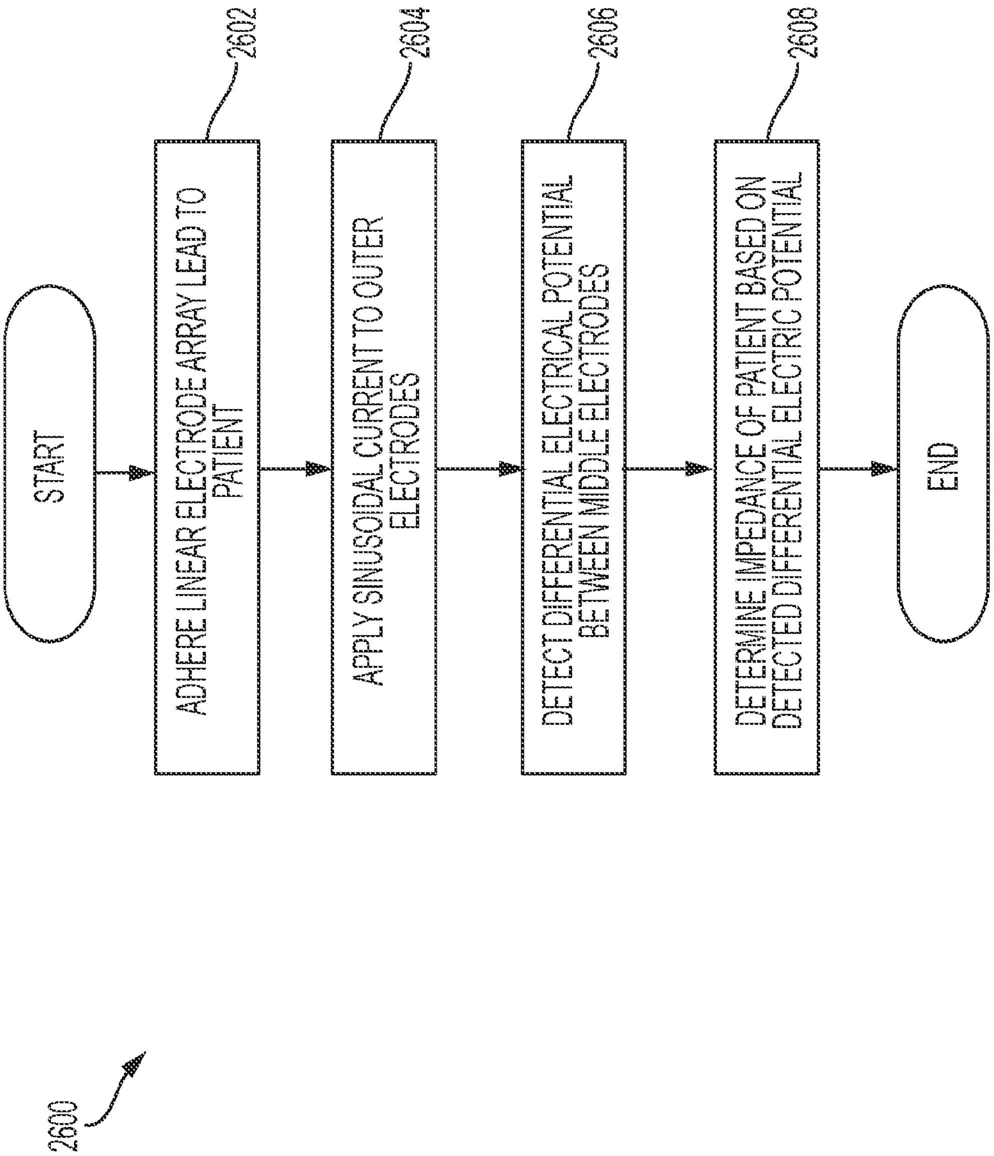

START
ADHERE LINEAR ELECTRODE ARRAY LEAD TO PATIENT
2602
APPLY SINUSOIDAL CURRENT TO OUTER ELECTRODES
2604
DETECT DIFFERENTIAL ELECTRICAL POTENTIAL BETWEEN MIDDLE ELECTRODES
2606
DETERMINE IMPEDANCE OF PATIENT BASED ON DETECTED DIFFERENTIAL ELECTRIC POTENTIAL
2608
END
2600

FIGURE 26

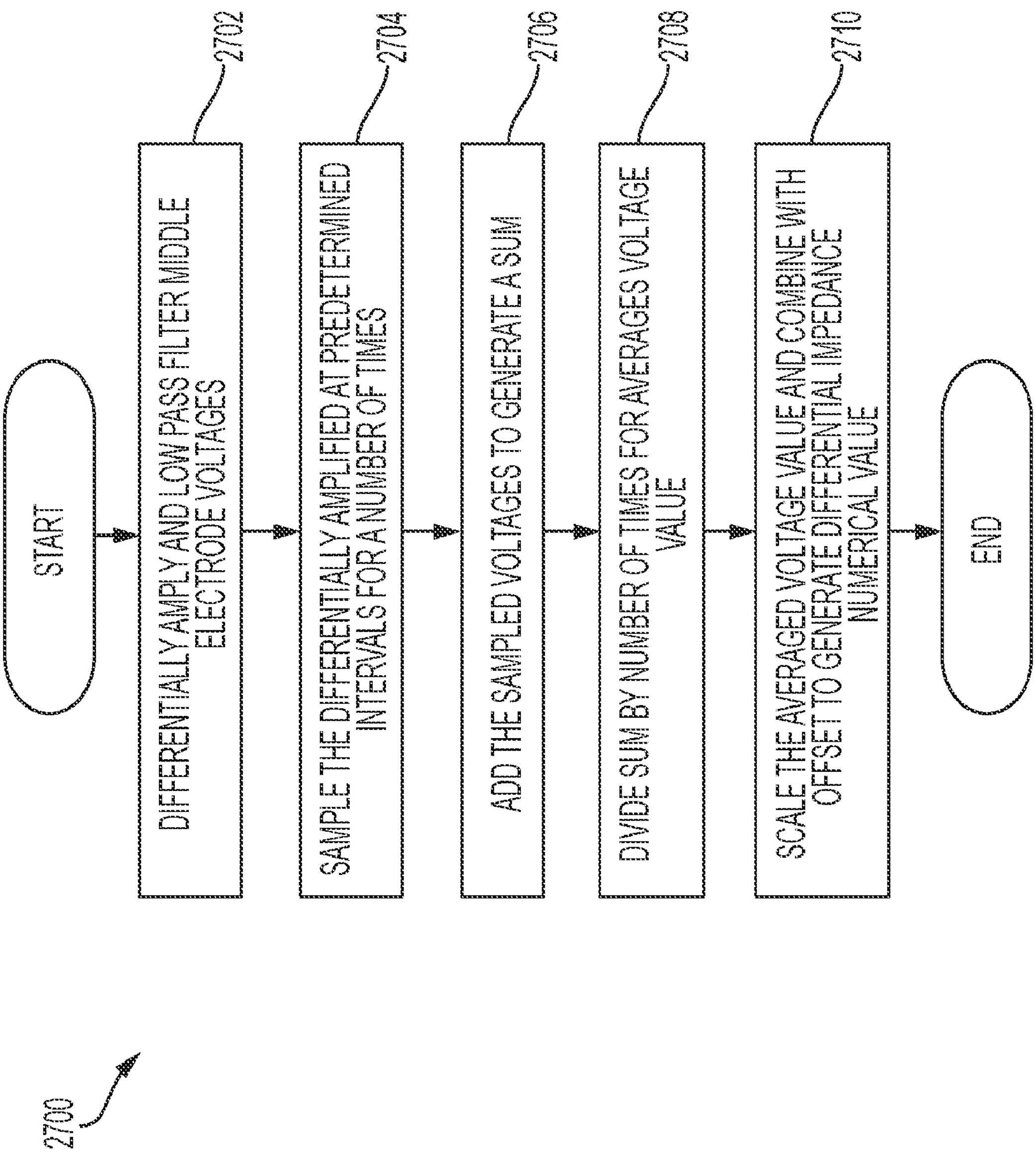

START
DIFFERENTIALLY AMPLY AND LOW PASS FILTER MIDDLE ELECTRODE VOLTAGES
2702
SAMPLE THE DIFFERENTIALLY AMPLIFIED AT PREDETERMINED INTERVALS FOR A NUMBER OF TIMES
2704
ADD THE SAMPLED VOLTAGES TO GENERATE A SUM
2706
DIVIDE SUM BY NUMBER OF TIMES FOR AVERAGES VOLTAGE VALUE
2708
SCALE THE AVERAGED VOLTAGE VALUE AND COMBINE WITH OFFSET TO GENERATE DIFFERENTIAL IMPEDANCE NUMERICAL VALUE
2710
END
2700

FIGURE 27

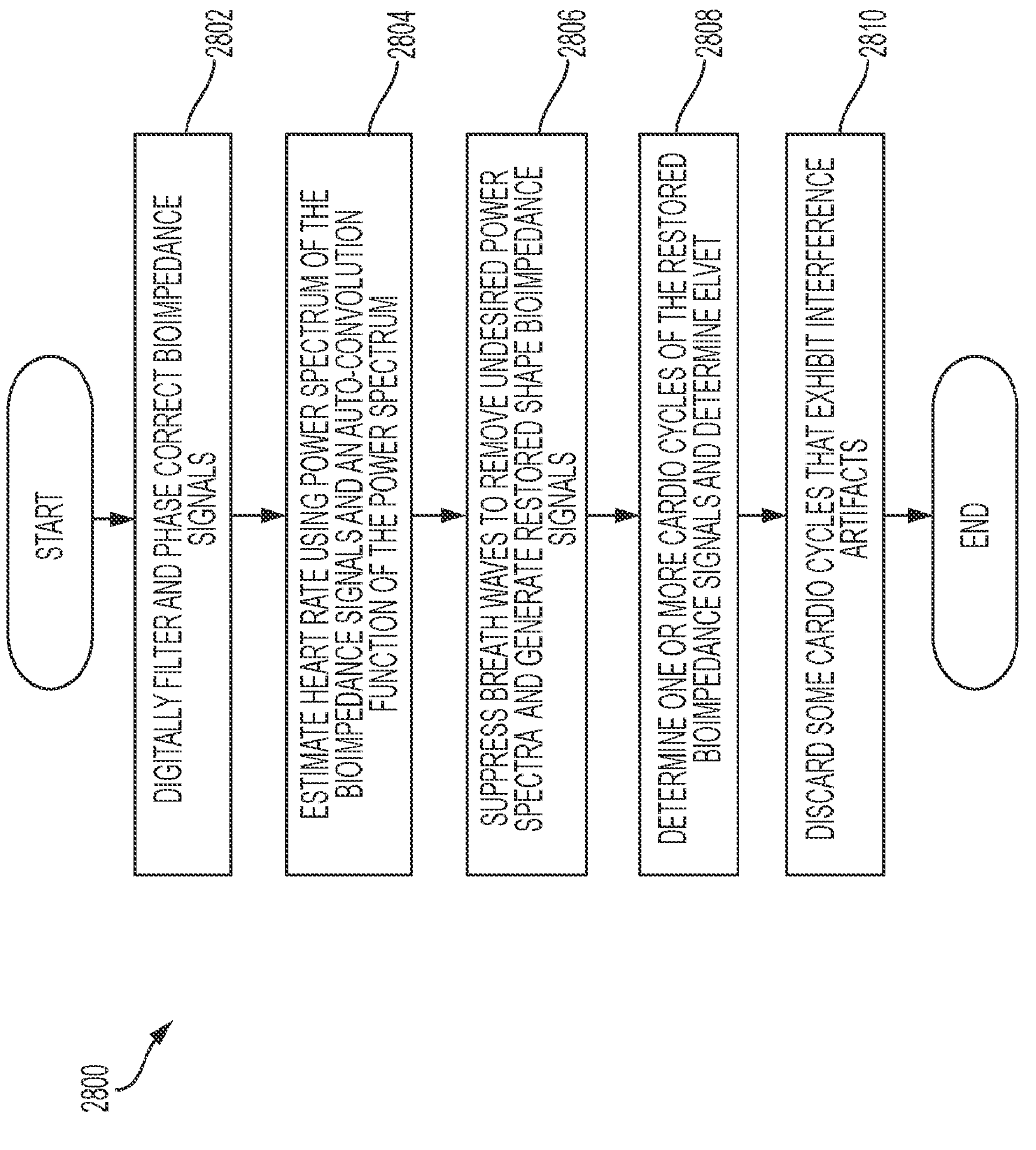

2800
START
DIGITALLY FILTER AND PHASE CORRECT BIOIMPEDANCE SIGNALS
2802
ESTIMATE HEART RATE USING POWER SPECTRUM OF THE BIOIMPEDANCE SIGNALS AND AN AUTO-CONVOLUTION FUNCTION OF THE POWER SPECTRUM
2804
SUPPRESS BREATH WAVES TO REMOVE UNDESIRED POWER SPECTRA AND GENERATE RESTORED SHAPE BIOIMPEDANCE SIGNALS
2806
DETERMINE ONE OR MORE CARDIO CYCLES OF THE RESTORED BIOIMPEDANCE SIGNALS AND DETERMINE ELVET
2808
DISCARD SOME CARDIO CYCLES THAT EXHIBIT INTERFERENCE ARTIFACTS
2810
END

FIGURE 28

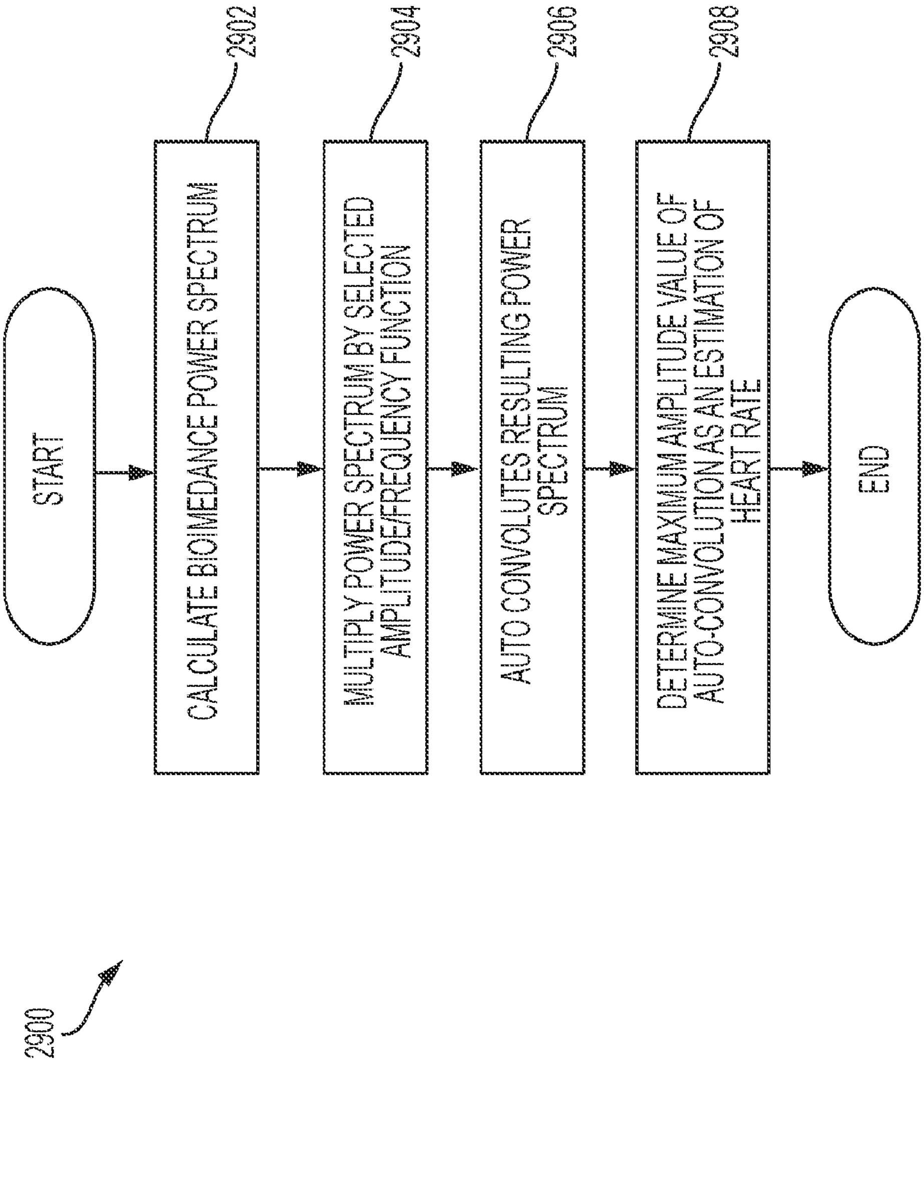

START
CALCULATE BIOIMEDANCE POWER SPECTRUM
MULTIPLY POWER SPECTRUM BY SELECTED AMPLITUDE/FREQUENCY FUNCTION
AUTO CONVOLUTES RESULTING POWER SPECTRUM
DETERMINE MAXIMUM AMPLITUDE VALUE OF AUTO-CONVOLUTION AS AN ESTIMATION OF HEART RATE
END
2902
2904
2906
2908
2900

FIGURE 29

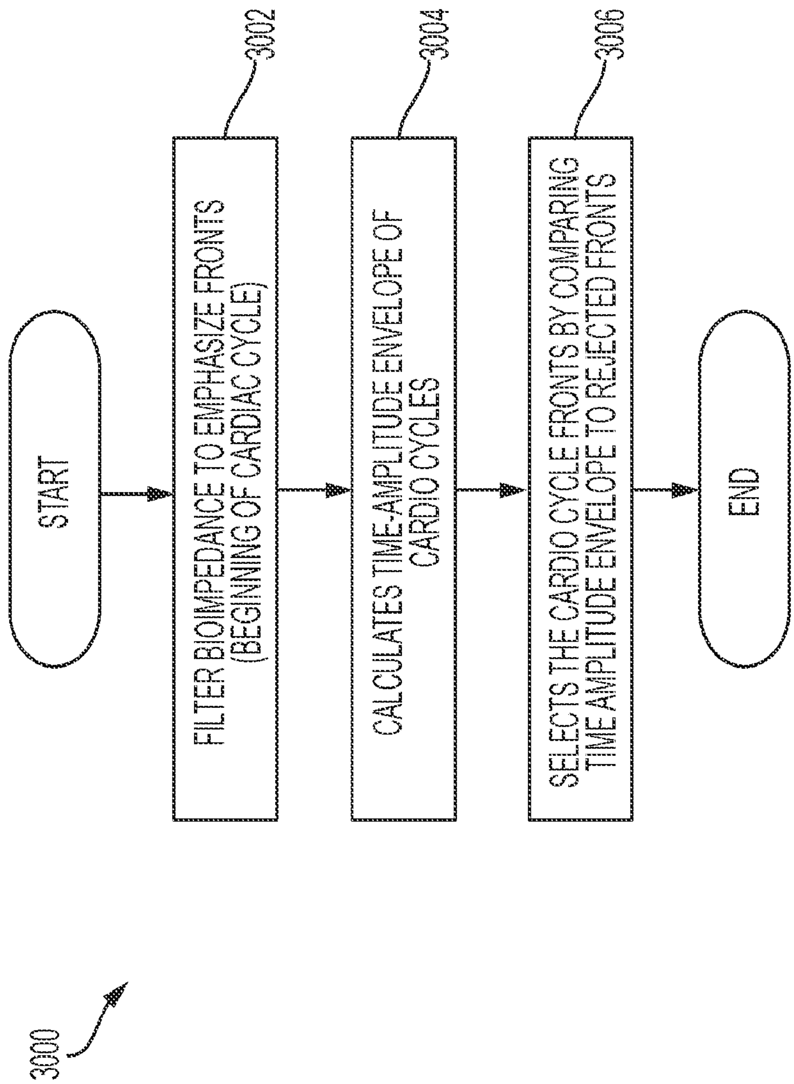

START
FILTER BIOIMPEDANCE TO EMPHASIZE FRONTS (BEGINNING OF CARDIAC CYCLE)
3002
CALCULATES TIME-AMPLITUDE ENVELOPE OF CARDIO CYCLES
3004
SELECTS THE CARDIO CYCLE FRONTS BY COMPARING TIME AMPLITUDE ENVELOPE TO REJECTED FRONTS
3006
END
3000

FIGURE 30

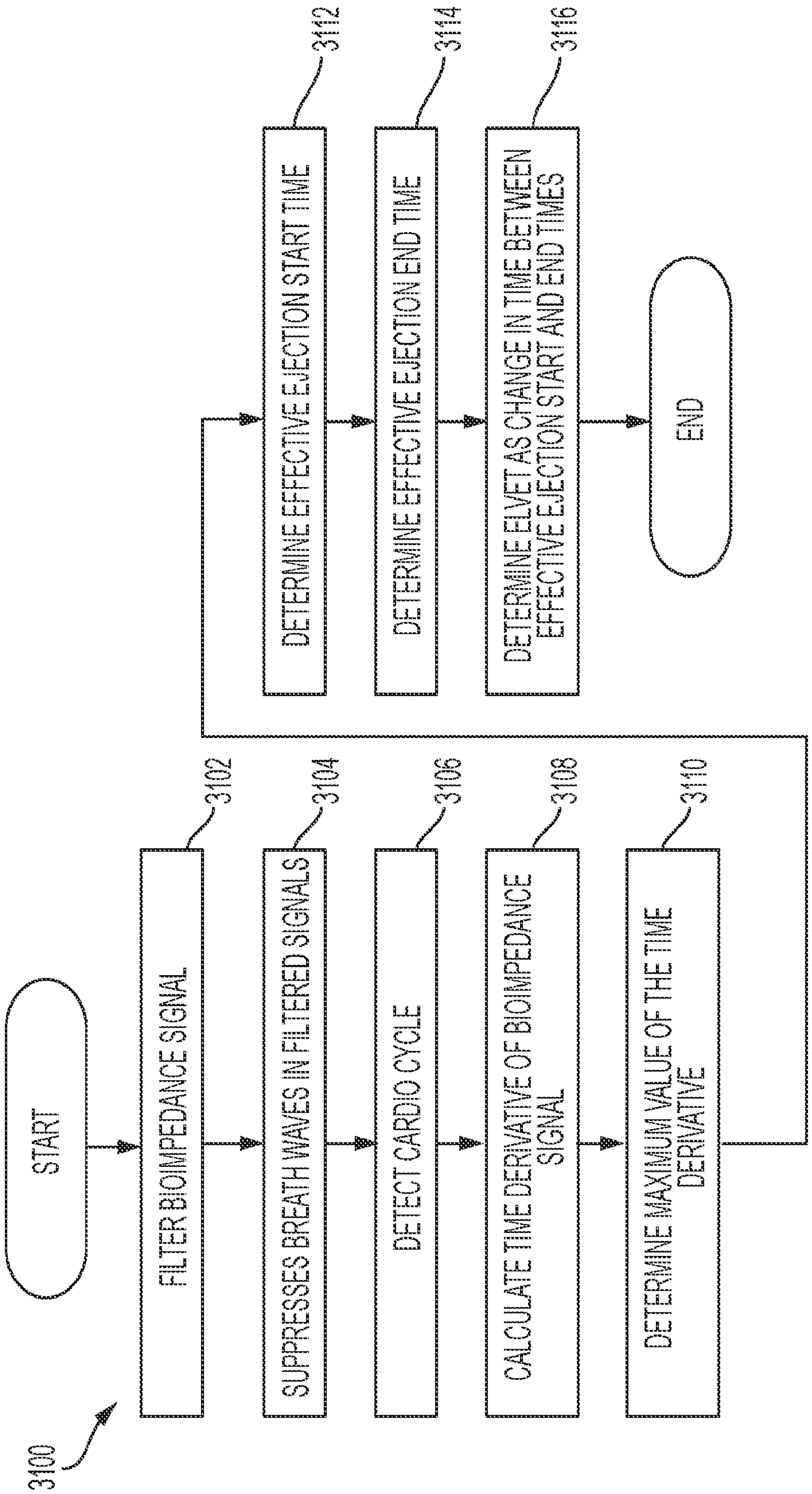

3100
START
FILTER BIOIMPEDANCE SIGNAL
3102
SUPPRESSES BREATH WAVES IN FILTERED SIGNALS
3104
DETECT CARDIO CYCLE
3106
CALCULATE TIME DERIVATIVE OF BIOIMPEDANCE SIGNAL
3108
DETERMINE MAXIMUM VALUE OF THE TIME DERIVATIVE
3110
DETERMINE EFFECTIVE EJECTION START TIME
3112
DETERMINE EFFECTIVE EJECTION END TIME
3114
DETERMINE ELVET AS CHANGE IN TIME BETWEEN EFFECTIVE EJECTION START AND END TIMES
3116
END

FIGURE 31

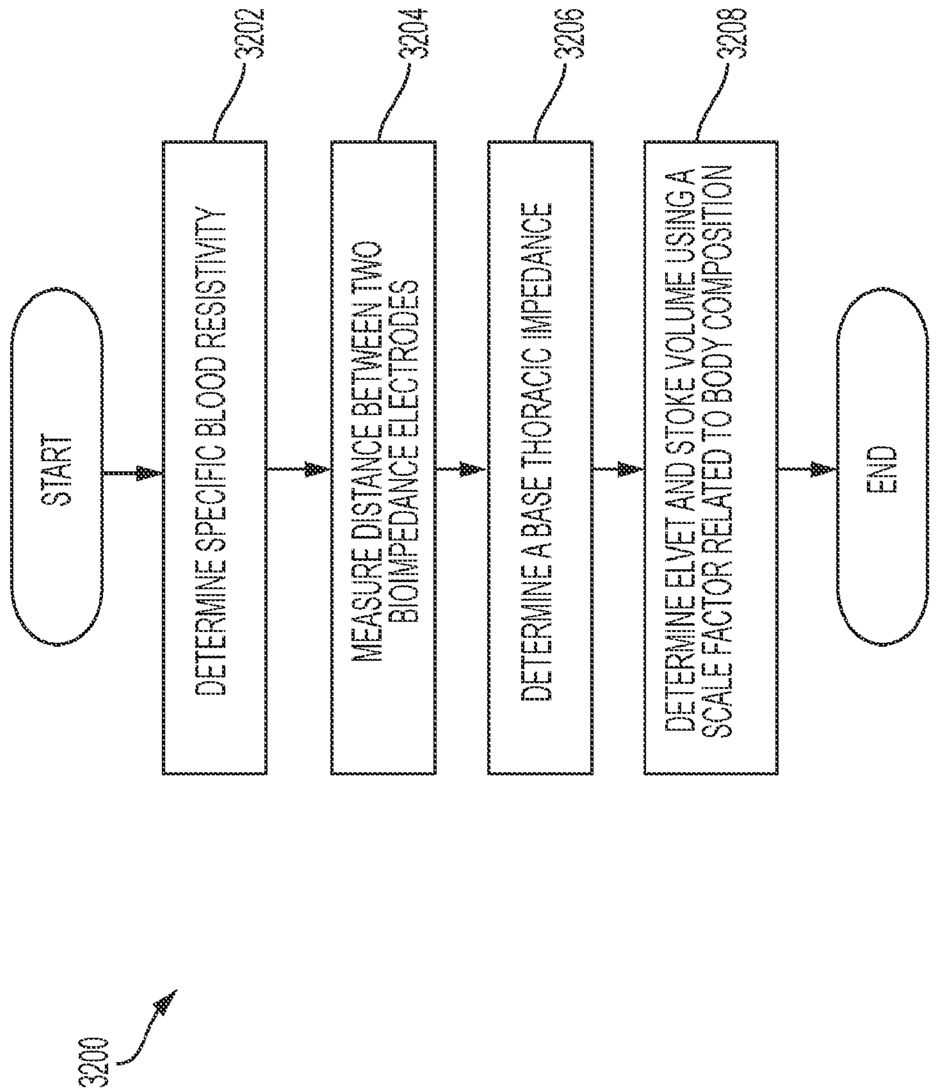

START
DETERMINE SPECIFIC BLOOD RESISTIVITY
3202
MEASURE DISTANCE BETWEEN TWO BIOIMPEDANCE ELECTRODES
3204
DETERMINE A BASE THORACIC IMPEDANCE
3206
DETERMINE ELVET AND STOKE VOLUME USING A SCALE FACTOR RELATED TO BODY COMPOSITION
3208
END
3200

FIGURE 32

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US2022/042165

A. CLASSIFICATION OF SUBJECT MATTER

A61B 5/053(2006.01)i; **A61B 5/0537**(2021.01)i; **A61B 5/024**(2006.01)i; **A61B 5/00**(2006.01)i

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

A61B 5/053(2006.01); A61B 5/00(2006.01); A61B 5/05(2006.01); A61B 5/1477(2006.01)

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Korean utility models and applications for utility models
Japanese utility models and applications for utility models

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

eKOMPASS(KIPO internal) & Keywords: electrode, impedance, current, frequency, hydration, cardiac, state

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
	US 2016-0066812 A1 (CHENG et al.) 10 March 2016 (2016-03-10)	
X	abstract; paragraphs [26], [72]; figure 4	1,5,16,17
Y		2-4,6,18-21
A		7-15
	WO 2009-036369 A1 (CORVENTIS, INC. et al.) 19 March 2009 (2009-03-19)	
Y	paragraphs [68], [69], [85], [142]; claims 22, 23, 34; figure 1A	2-4,6,18-21
	US 2017-0172484 A1 (UNIVERSITY OF CINCINNATI et al.) 22 June 2017 (2017-06-22)	
A	whole document	1-21
	US 8406865 B2 (MCKENNA) 26 March 2013 (2013-03-26)	
A	whole document	1-21

☑ Further documents are listed in the continuation of Box C. ☑ See patent family annex.

* Special categories of cited documents:
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"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed
"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
16 December 2022	**16 December 2022**

Name and mailing address of the ISA/KR	Authorized officer
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C. DOCUMENTS CONSIDERED TO BE RELEVANT

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