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(54) **THORACIC IMPEDANCE DETECTION WITH BLOOD RESISTIVITY COMPENSATION**

(75) Inventors: **Jeffrey E. Stahmann**, Ramsey, MN (US); **John D. Hatlestad**, Maplewood, MN (US); **Boyce Moon**, Ham Lake, MN (US)

(73) Assignee: **Cardiac Pacemakers, Inc.**, St. Paul, MN (US)

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See application file for complete search history.

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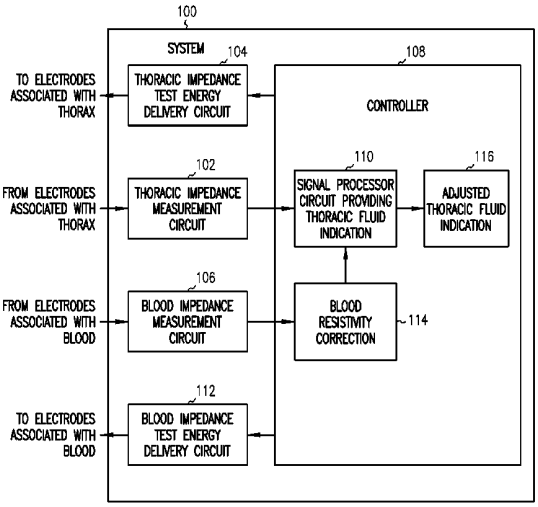
Primary Examiner—Robert L Nasser
(74) *Attorney, Agent, or Firm*—Schwegman, Lundberg & Woessner, P.A.

(57) **ABSTRACT**

This document discusses, among other things, a cardiac rhythm management device or other implantable medical device that uses thoracic impedance to determine how much fluid is present in the thorax, such as for detecting or predicting congestive heart failure, pulmonary edema, pleural effusion, hypotension, or the like. The thoracic fluid amount determined from the thoracic impedance is compensated for changes in blood resistivity, which may result from changes in hematocrit level or other factors. The blood-resistivity-compensated thoracic fluid amount can be stored in the device or transmitted to an external device for storage or display. The blood-resistivity-compensated thoracic fluid amount can also be used to adjust a cardiac pacing, cardiac resynchronization, or other cardiac rhythm management or other therapy to the patient. This document also discusses applications of the devices and methods for predicting or indicating anemia.

30 Claims, 4 Drawing Sheets



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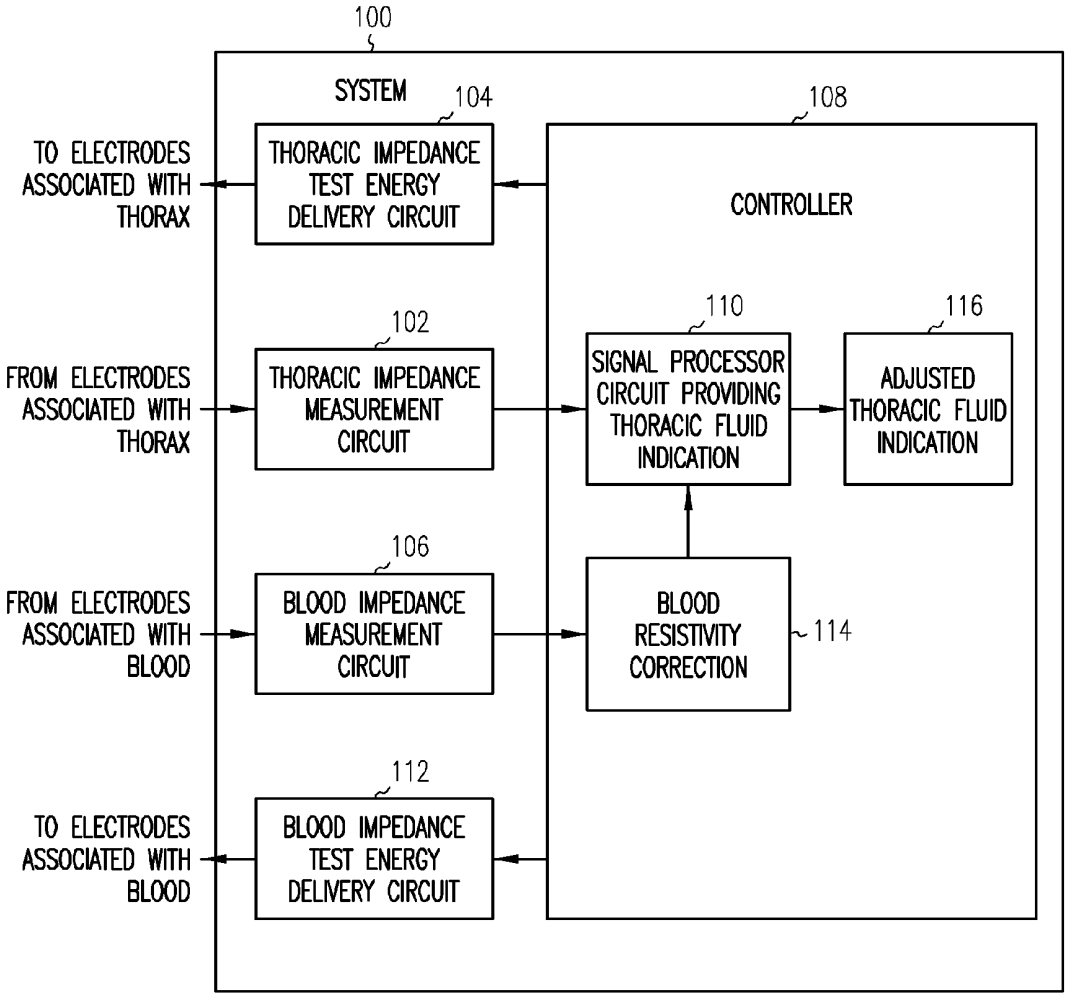
100
SYSTEM
104
108
TO ELECTRODES ASSOCIATED WITH THORAX
THORACIC IMPEDANCE TEST ENERGY DELIVERY CIRCUIT
CONTROLLER
110
116
102
FROM ELECTRODES ASSOCIATED WITH THORAX
THORACIC IMPEDANCE MEASUREMENT CIRCUIT
SIGNAL PROCESSOR CIRCUIT PROVIDING THORACIC FLUID INDICATION
ADJUSTED THORACIC FLUID INDICATION
106
FROM ELECTRODES ASSOCIATED WITH BLOOD
BLOOD IMPEDANCE MEASUREMENT CIRCUIT
BLOOD RESISTIVITY CORRECTION
114
112
TO ELECTRODES ASSOCIATED WITH BLOOD
BLOOD IMPEDANCE TEST ENERGY DELIVERY CIRCUIT

FIG. 1

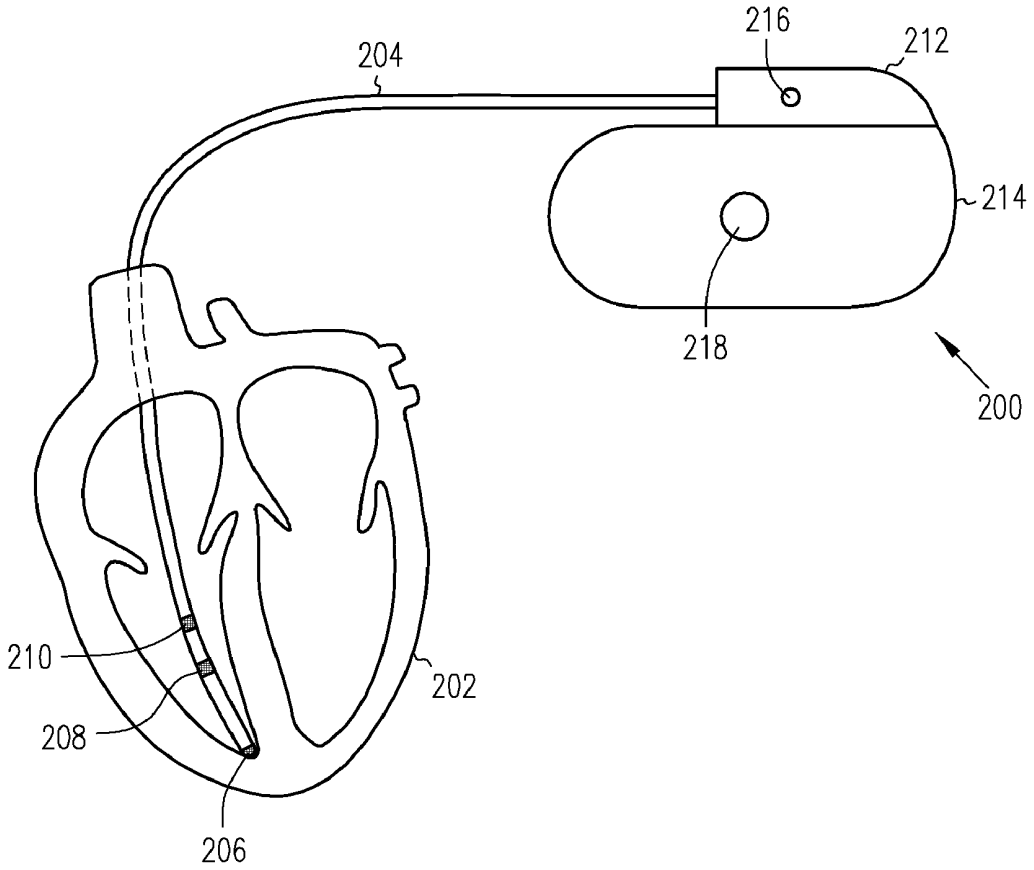

FIG. 2

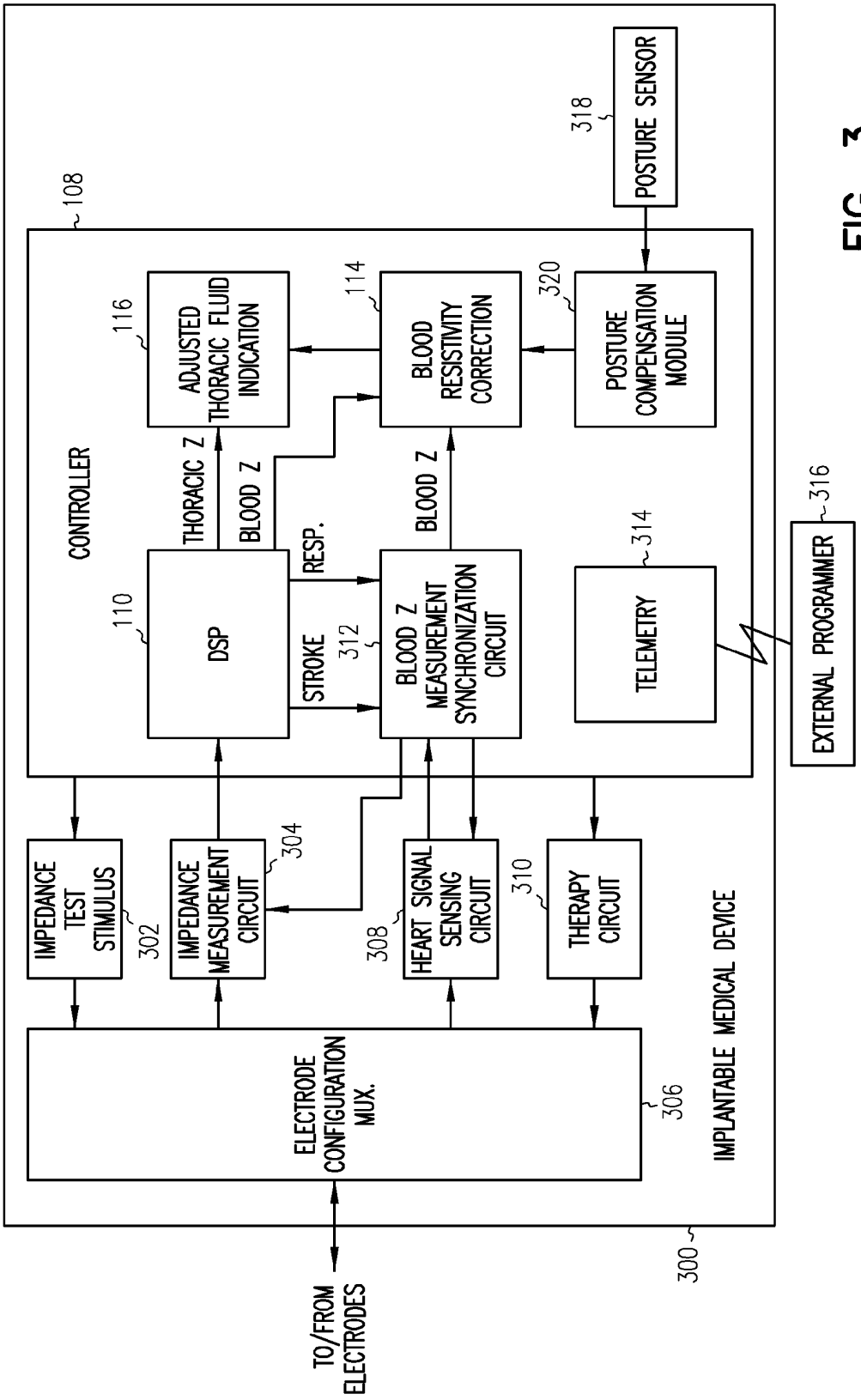
300
IMPLANTABLE MEDICAL DEVICE
TO/FROM ELECTRODES
ELECTRODE CONFIGURATION MUX.
306
IMPEDANCE TEST STIMULUS
302
IMPEDANCE MEASUREMENT CIRCUIT
304
HEART SIGNAL SENSING CIRCUIT
308
THERAPY CIRCUIT
310
CONTROLLER
108
DSP
110
THORACIC Z
BLOOD Z
RESP.
STROKE
312
BLOOD Z MEASUREMENT SYNCHRONIZATION CIRCUIT
BLOOD Z
ADJUSTED THORACIC FLUID INDICATION
116
BLOOD RESISTIVITY CORRECTION
114
POSTURE COMPENSATION MODULE
320
POSTURE SENSOR
318
TELEMETRY
314
EXTERNAL PROGRAMMER
316

FIG. 3

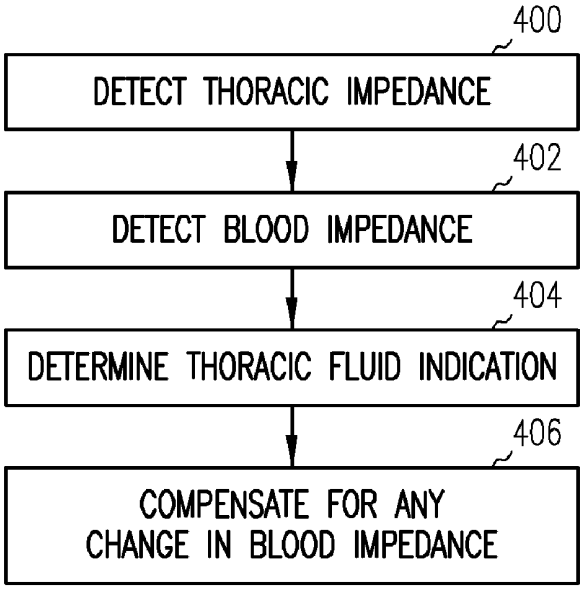
400
DETECT THORACIC IMPEDANCE
402
DETECT BLOOD IMPEDANCE
404
DETERMINE THORACIC FLUID INDICATION
406
COMPENSATE FOR ANY CHANGE IN BLOOD IMPEDANCE

FIG. 4

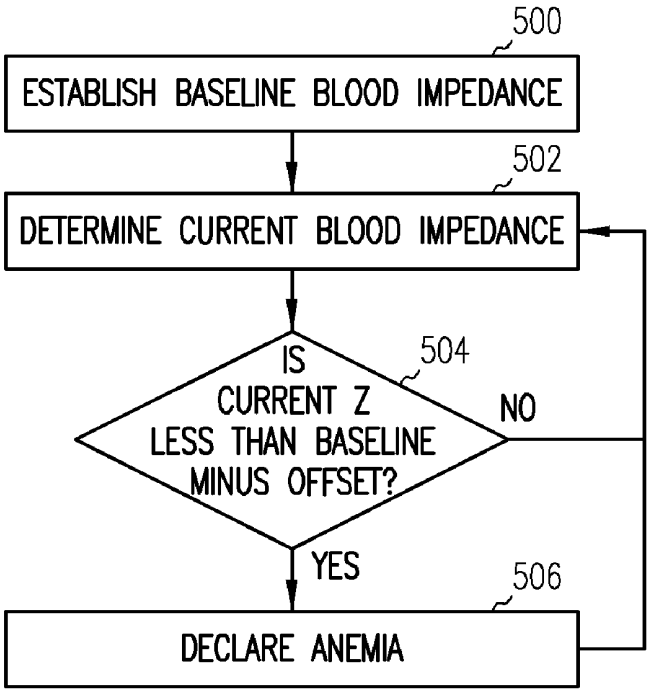
500
ESTABLISH BASELINE BLOOD IMPEDANCE
502
DETERMINE CURRENT BLOOD IMPEDANCE
504
IS CURRENT Z LESS THAN BASELINE MINUS OFFSET?
NO
YES
506
DECLARE ANEMIA

FIG. 5

THORACIC IMPEDANCE DETECTION WITH BLOOD RESISTIVITY COMPENSATION

CROSS REFERENCE TO RELATED APPLICATION

This application is a continuation of U.S. application Ser. No. 12/139,948, now U.S. Pat. 7,672,718, filed Jun. 16, 2008, which is a continuation of U.S. application No. 10/921,503, filed Aug. 19, 2004, now U.S. Pat. No. 7,387,610, the specifications of which are herein incorporated by reference.

TECHNICAL FIELD

This document pertains generally to implantable medical devices and more particularly, but not by way of limitation, to congestive heart failure (CHF) thoracic fluid detection and other thoracic impedance systems, devices, or methods that compensate or correct for changes in blood resistivity.

BACKGROUND

Variations in how much fluid is present in a person's thorax can take various forms and can have different causes. Eating salty foods can result in retaining excessive fluid in the thorax and elsewhere. Posture changes can also affect the amount of thoracic fluid. For example, moving from supine to standing can shift intravascular fluid away from the thorax toward the lower extremities.

Another example is pulmonary edema, which results in buildup of extravascular fluid in the lungs. In pulmonary edema, fluid accumulates in extracellular spaces, such as the spaces between lung tissue cells. One cause of pulmonary edema is congestive heart failure (CHF), which is also sometimes referred to as "chronic heart failure," or as "heart failure." CHF can be conceptualized as an enlarged weakened portion of heart muscle. The impaired heart muscle results in poor cardiac output of blood. As a result of such poor blood circulation, blood tends to pool in blood vessels in the lungs. This intravascular fluid buildup, in turn, results in the extravascular fluid buildup mentioned above. In sum, pulmonary edema can be one important condition associated with CHF.

Yet another example of thoracic fluid accumulation is pleural effusion, which is the buildup of extravascular fluid in the space between the lungs and the rib cage. Pleural effusion can also result from CHF because, as discussed above, intravascular fluid buildup can result in the extravascular interstitial fluid buildup. The extravascular fluid buildup of pulmonary edema can, in turn, result in the extravascular fluid buildup of pleural effusion.

CHF may also activate several physiological compensatory mechanisms. Such compensatory mechanisms are aimed at correcting the reduced cardiac output. For example, the heart muscle may stretch to increase its contractile power. Heart muscle mass may also increase. This is referred to as "hypertrophy." The ventricle may also change its shape as another compensatory response. In another example, a neuroendocrine response may provide an adrenergic increase in heart rate and contraction force. The Renin-Angiotensin-Aldosterone-System (RAAS) may be activated to induce vasoconstriction, fluid retention, and redistribution of blood flow. Although the neuro-endocrine response is compensatory, it may overload the cardiovascular system. This may result in myocardial damage, and may exacerbate CHF.

Diagnosing CHF may involve physical examination, electrocardiogram (ECG), blood tests, chest radiography, or echocardiography. Managing a CHF patient is challenging.

CHF may require potent drugs. Moreover, treatment may be thwarted by the compensatory mechanisms, which may recompensate for the presence of the medical treatment. Therefore, treating CHF involves a delicate balance to properly manage the patient's hemodynamic status in a state of proper compensation to avoid further degeneration.

However, this delicate balance between compensation and effective CHF treatment is easily upset, even by seemingly benign factors, such as common medication (e.g., aspirin), physiological factors, excitement, or gradual progression of the disease. This may plunge the patient into a decompensation crisis, which requires immediate corrective action so as to prevent the deterioration of the patient's condition which, if left unchecked, can lead to death. In sum, accurately monitoring the symptoms of CHF, such as thoracic fluid accumulation, is very useful for avoiding such a decompensation crisis and properly managing the CHF patient in a state of relative well-being.

BRIEF DESCRIPTION OF THE DRAWINGS

In the drawings, which are not necessarily drawn to scale, like numerals describe substantially similar components throughout the several views. Like numerals having different letter suffixes represent different instances of substantially similar components. The drawings illustrate generally, by way of example, but not by way of limitation, various embodiments discussed in the present document.

FIG. **1** is a block diagram illustrating generally one example of a system that provides a thoracic fluid amount indication that is adjusted to compensate for a change in blood resistivity, if any.

FIG. **2** is a schematic illustration of one example in which portions of the system are implemented in an implantable cardiac rhythm management (CRM) or other implantable medical device (IMD).

FIG. **3** is a block diagram illustrating generally another example in which portions of the system are implemented in an implantable CRM or other IMD.

FIG. **4** is a flow chart illustrating generally one example of a method of providing a thoracic fluid amount indication that is compensated for any changes in blood resistivity.

FIG. **5** is a flow chart illustrating generally one example of a method of detecting anemia using a blood impedance measurement performed by an implantable medical device.

DETAILED DESCRIPTION

The following detailed description includes references to the accompanying drawings, which form a part of the detailed description. The drawings show, by way of illustration, specific embodiments in which the invention may be practiced. These embodiments, which are also referred to herein as "examples," are described in enough detail to enable those skilled in the art to practice the invention. The embodiments may be combined, other embodiments may be utilized, or structural, logical and electrical changes may be made without departing from the scope of the present invention. The following detailed description is, therefore, not to be taken in a limiting sense, and the scope of the present invention is defined by the appended claims and their equivalents.

In this document, the terms "a" or "an" are used, as is common in patent documents, to include one or more than one. In this document, the term "or" is used to refer to a nonexclusive or, unless otherwise indicated. Furthermore, all publications, patents, and patent documents referred to in this

document are incorporated by reference herein in their entirety, as though individually incorporated by reference. In the event of inconsistent usages between this document and those documents so incorporated by reference, the usage in the incorporated reference(s) should be considered supplementary to that of this document; for irreconcilable inconsistencies, the usage in this document controls.

In this document, the term intravascular includes the term intracardiac.

In this document, the term cardiovascular includes an association with either the heart or a blood vessel.

In this document, the term "thorax" refers to a human subject's body other than the subject's head, arms, and legs.

FIG. **1** is a block diagram illustrating generally one example of a system **100** that provides an indication of the amount of fluid in the thorax ("thoracic fluid indication") that is adjusted to compensate for a change in blood resistivity, if any. In this example, the system **100** includes a thoracic impedance measurement circuit **102**. The thoracic impedance measurement circuit **102** receives at least one electrical signal from electrodes associated with a patient's thorax. This electrical signal is typically received in response to a test energy applied to the thorax, such as by a thoracic impedance test energy delivery circuit **104**.

One illustrative example of some electrode configurations and circuits for performing thoracic impedance measurements is described in Hartley et al. U.S. Pat. No. 6,076,015 entitled RATE ADAPTIVE CARDIAC RHYTHM MANAGEMENT DEVICE USING TRANSTHORACIC IMPEDANCE, which is assigned to Cardiac Pacemakers, Inc., and which is incorporated herein by reference in its entirety, including its description of performing thoracic impedance measurements. The Hartley et al. U.S. Pat. No. 6,076,015 uses thoracic impedance to obtain a respiration signal. By contrast, the present patent application uses thoracic impedance to obtain a thoracic fluid status signal. Therefore, the signal of interest in the present patent application would be deemed noise in the Hartley et al. U.S. Pat. No. 6,076,015, and vice-versa. However, both thoracic fluid status and respiration are obtainable using the thoracic impedance detection techniques described in the Hartley et al. U.S. Pat. No. 6,076,015. The present thoracic fluid status signal of interest is obtained from a lower frequency (i.e., a "near-DC") portion of the thoracic impedance signal rather than the frequencies of the respiration signal described in the Hartley et al. U.S. Pat. No. 6,076,015. In this document, the "near-DC" component of the thoracic impedance signal refers to the frequencies below which respiration and cardiac contractions significantly influence the thoracic impedance signal. This near-DC component of the thoracic impedance signal, therefore, typically refers to signal frequencies below a cutoff frequency having a value of about 0.1 Hz, such as at signal frequencies between about 5×10^{-7} Hz and 0.05 Hz, because the cardiac stroke and respiration components of the thoracic impedance signal lie at higher frequencies. Fluid accumulation in the thorax corresponds to a decrease in the near-DC thoracic impedance. Conversely, fluid depletion in the thorax corresponds to an increase in the near-DC thoracic impedance. As discussed above, fluid accumulation may result from, among other things, pulmonary edema or pleural effusion, both of which may result from CHF.

In the example of FIG. **1**, the system **100** also includes a controller **108**. The controller **108** is typically a microprocessor or any other circuit that is capable of sequencing through various control states such as, for example, by using a digital microprocessor having executable instructions stored in an associated instruction memory circuit, a microsequencer, or a state machine. In this example, the controller **108** includes a digital signal processor (DSP) circuit **110**. The digital signal processor circuit **110** performs any digital filtering or other signal processing needed to extract from the thoracic impedance signal a near-DC desired thoracic fluid amount signal. The digital signal processor circuit **110**, therefore, may implement one or more filter circuits, and such filter circuits may be implemented as a sequence of executable instructions, rather than by dedicated filtering hardware.

However, the present inventors have recognized that the near-DC thoracic impedance signal is typically also affected by confounding factors other than the amount of fluid present in the thorax. One such confounding factor is any change in blood resistivity. Blood resistivity changes as a function of hematocrit in the blood. The hematocrit (Ht) or packed cell volume (PCV) is the proportion of blood that is occupied by red blood cells. It is typically between 0.35 and 0.52, and is slightly higher on average in males than in females. For example, when a patient is dehydrated, there will be less fluid in the patient's blood. Therefore, the patient's hematocrit level will increase, that is, the patient's blood will include a higher percentage of other components, such as insulative red blood cells. This will increase the blood resistivity, which, in turn, will affect the thoracic impedance signal even through it is not necessarily associated with the extravascular fluid accumulation of pulmonary edema or pleural effusion. Other factors that are believed to possibly influence blood resistivity include the patient's electrolyte level, certain medications in the blood, proteins in the blood, or blood gas concentrations.

As an illustrative example, the above change in hematocrit percentage from 35% to 52% may correspond to a change in resistivity from about 140 Ω·cm to about 200 Ω·cm. Such changes in blood resistivity will influence the near-DC thoracic impedance measurement. This will confound an extravascular thoracic fluid amount determination using the near-DC thoracic impedance measurement, unless the extravascular thoracic fluid amount determination is corrected for such variations in blood resistivity, if any. Measurement of variations in blood resistivity is typically affected by the frequency of the excitation signal that are used. At higher excitation frequencies, blood cells typically become more resistive.

Accordingly, the system in FIG. **1** illustrates a blood resistivity measurement circuit **106**. The blood impedance measurement circuit **106** receives a blood impedance measurement from electrodes that are associated with blood (and preferably blood in the thorax) such as in response to a delivery of test energy by a blood impedance test energy delivery circuit **112**. In one example, the blood impedance measurement circuit **106** and the blood impedance test energy delivery circuit **112** are configured similar to the thoracic impedance measurement circuit **102** and the thoracic impedance test energy delivery circuit **104**, respectively, as discussed above, except for being connected to different electrodes. Using the blood impedance measurement, the controller **108** executes a sequence of instructions to compute a blood resistivity correction **114**. The blood resistivity correction **114** is applied to the thoracic fluid indication that is output by the digital signal processor circuit **110**. This yields an adjusted thoracic fluid amount indication **116**.

In FIG. **1**, the thoracic impedance test energy delivery circuit **104** is illustrated separately from the blood impedance test energy delivery circuit **112** to assist the reader's conceptualization. In practice, these circuits, or portions thereof, may be combined. The combined circuit may be coupled to different electrodes for delivering the thoracic impedance test energy than for delivering the blood impedance test energy.

Similarly, in FIG. **1**, the thoracic impedance measurement circuit **102** is illustrated separately from the blood impedance test energy delivery circuit **112** to assist the reader's conceptualization. In practice, these circuits, or portions thereof, may be combined. The combined circuit may be coupled to different electrodes for measuring the responsive voltages for the thoracic and blood impedance measurements, as discussed below.

FIG. **2** is a schematic illustration of one example in which portions of the system **100** are implemented in an implantable cardiac rhythm management (CRM) or other implantable medical device (IMD) **200**. In this example, the IMD **200** is coupled to a heart **202** using at least one leadwire, such as a multielectrode leadwire **204**. In this example, the leadwire **204** includes a tip electrode **206**, a distal ring electrode **208**, and a proximal ring electrode **210**, each of which is disposed in the right ventricle of the heart **202**. In this example, each of the tip electrode **206**, the distal ring electrode **208**, and the proximal ring electrode **210** is independently electrically connected to a corresponding separate electrically conductive terminal within an insulating header **212**. The header **212** is affixed to a housing **214** carrying electronic components of the IMD **200**. In this example, the header **212** includes a header electrode **216**, and the housing **214** includes a housing electrode **218**.

In one example, thoracic impedance is sensed by delivering a test current between: (1) at least one of the ring electrodes **208** or **210**; and (2) the housing electrode **218**, and a resulting responsive voltage is measured across the tip electrode **206** and the header electrode **216**. Because the IMD **200** is typically pectorally implanted at some distance away from the heart **202**, this electrode configuration injects the test current over a substantial portion (but typically not the entire portion) of the patient's thorax, such that when the resulting voltage measurement is divided by the test current magnitude, it yields an indication of thoracic impedance. Using different electrodes for delivering the current and for measuring the responsive voltage reduces the component of the measured impedance signal that results from ohmic losses in the leadwires to the test current delivery electrodes. While such a "four-point" probe is useful, it is not required. In other examples, a "three-point probe" (having three electrodes, with one electrode used for both test current delivery and responsive voltage measurement), or a "two-point probe" (having two electrodes, each electrode used for both test current delivery and responsive voltage measurement) are used. Moreover, other electrode combinations could alternatively be used to implement a four-point probe. The above four-point probe description provides an illustrative example of one suitable four-point probe configuration.

In one example, blood impedance is sensed by delivering a test current between: (1) one of the distal ring electrode **208** or the proximal ring electrode **210**; and (2) the housing electrode **218**. A resulting responsive voltage is measured between: (1) the other of the distal ring electrode **208** or the proximal ring electrode **210**; and (2) the tip electrode **206**. In this example, although the test current is injected across a substantial portion of the patient's thorax, as discussed above, the responsive voltage signal of interest is measured across electrodes within the same chamber of the patient's heart (or, alternatively, within the same blood vessel). Therefore, when the responsive voltage measurement is divided by the test current magnitude, it yields an indication of the blood impedance in the heart chamber rather than the thoracic impedance. The measured blood impedance is used to compensate the measured thoracic impedance for changes in the blood impedance.

FIG. **3** is a block diagram illustrating generally another example in which portions of the system **100** are implemented in an implantable CRM or other IMD **300**. The example of FIG. **3** includes an impedance test stimulus circuit **302** that, together with an impedance measurement circuit **304**, provides thoracic and blood impedance measurements. In response to one or more control signals from the controller **108**, an electrode configuration multiplexer **306** couples these circuits to the appropriate electrodes for the particular thoracic or blood impedance measurement. In this example, the multiplexer **306** is also coupled to a heart signal sensing circuit **308**, which includes sense amplifier or other circuits for detecting from particular electrodes intrinsic electrical heart signals that include electrical depolarizations corresponding to heart contractions. The multiplexer **306** is also coupled to a therapy circuit **310**, such as a pulse delivery circuit for delivering pacing, cardioversion, or defibrillation energy to particular electrodes in response to one or more control signals received from the controller **108**.

In the example of FIG. **3**, the DSP circuit **110** processes the thoracic impedance measurements from the impedance measurement circuit **304**. The DSP circuit **110** extracts a cardiac stroke signal or a respiration signal from the thoracic impedance signal, such as by using techniques described in the above-incorporated Hartley et al. U.S. Pat. No. 6,076,015. One or both of the extracted cardiac stroke or respiration signals is provided to a blood impedance measurement synchronization circuit **312**. The synchronization circuit **312** includes one or more peak-detector, level-detector, or zero-cross detector circuits to synchronize the blood impedance measurement to the same sample point of a cardiac contraction cycle or a respiration cycle. This reduces the effect of variations in one or both of these cycles on the blood impedance measurement. Similarly, the measurements can be taken under the same conditions with respect to posture or circadian cycle to reduce those effects on the blood impedance measurement. Posture can be detected using an accelerometer or other posture sensor; circadian cycle can be ascertained from a time-of-day indication provided by a clock circuit within the controller **108**. Alternatively, the cardiac cycle information is extracted from the heart signal sensing circuit **308**, either by itself or in combination with information from the controller **108** about when pacing or other stimulus pulses that evoke a responsive heart contraction are issued. The controller **108** computes an adjusted thoracic fluid indication **116** from the measured thoracic impedance. The adjusted thoracic fluid indication **116** is compensated for blood resistivity variations using the blood resistivity correction **114** obtained using the measured blood impedance. In a further example, the implantable medical device **300** includes a telemetry circuit **314** that communicates one of the blood-resistivity-compensated thoracic impedance or the blood resistivity and thoracic impedance measurements to an external programmer **316** or the like for further processing, storage, or display.

FIG. **4** is a flow chart illustrating generally one example of a method of providing a thoracic fluid amount indication that is compensated for any changes in blood resistivity. At **400**, a thoracic impedance is detected. This may be accomplished in a number of different ways. In one illustrative example, such as described in Hartley et al. U.S. Pat. No. 6,075,015, it includes injecting a four-phase carrier signal, such as between a housing electrode **218** and a ring electrode **208**. In this example, the first and third phases are +320 microampere pulses that are 20 microseconds long. The second and fourth phases are −320 microampere pulses that are 20 microseconds long. The four phases are repeated at 50 millisecond intervals to provide a carrier test current signal from which a

responsive voltage can be measured. However, as discussed elsewhere in this document, because blood resistivity varies with excitation frequency, a different excitation frequency may also be used.

The Hartley et al. U.S. Pat. No. 6,075,015 describes an exciter circuit for delivering such a test current stimulus (however, the present system can alternatively use other suitable circuits, including an arbitrary waveform generator that is capable of operating at different frequencies or of mixing different frequencies to generate an arbitrary waveform). It also describes a signal processing circuit for measuring a responsive voltage between a housing electrode **216** and a tip electrode **206**. In one example, the signal processing circuit includes a preamplifier, demodulator, and bandpass filter for extracting the thoracic impedance data from the carrier signal, before conversion into digital form by an A/D converter. Further processing is performed digitally, and is performed differently in the present system **100** than in the Hartley et al. U.S. Pat. No. 6,075,015.

For example, the Hartley et al. U.S. Pat. No. 6,075,015 includes a bandpass filter that receives the output of the A/D converter. The purpose of the highpass portion of the bandpass filter is to attenuate the near-DC portion of the thoracic impedance signal, which is the signal of interest to the present system **100**. Therefore, the present system **100** eliminates the highpass filter. The cutoff frequency of the remaining lowpass filter is selected to pass the near-DC portion of the thoracic impedance signal and attenuate higher frequency portions of the thoracic impedance signal, including the respiration and cardiac stroke components of the thoracic impedance signal. In one example, a programmable cutoff frequency lowpass filter is used. In another example, an adaptive cutoff frequency lowpass filter is used, such that the cutoff frequency is moved to a higher frequency for higher values of heart rate and respiration frequency, and the cutoff frequency is moved to a lower frequency for lower values of heart rate and respiration frequency.

At **402** of FIG. **4**, blood impedance is detected and measured. There are a number of ways in which this can be done. In one example, the blood impedance measurement is performed in the same manner as the thoracic impedance measurement, except that measurement of the responsive voltage is across two electrodes that are both typically located in the same heart chamber or same blood vessel, such as between (1) one of the distal ring electrodes **208** or the proximal ring electrode **210**; and (2) the other of the distal ring electrode **208** or the proximal ring electrode **210**. Because the blood impedance is to be used to correct a thoracic fluid indication, it is typically detected and measured at or near the thorax. Alternatively, however, even an external blood impedance measurement could be used, if desired. In one example, the blood impedance is sampled under appropriate other conditions (e.g., at a like point in different cardiac cycles, at a like point in different respiration cycles, etc.).

At **404**, a thoracic fluid amount indication is determined. There are a number of ways in which this can be done. In one example, the thoracic fluid amount indication is given by the value of the near-DC thoracic impedance signal, which may be averaged or otherwise filtered, if desired. In another example, a baseline value of this averaged or otherwise filtered near-DC thoracic impedance signal is obtained from the patient, and the thoracic fluid amount indication is given by the difference of the near-DC thoracic fluid impedance value (with the same or different averaging or filtering) from this baseline value.

At **406**, the thoracic fluid amount indication obtained from the near-DC thoracic impedance is adjusted to compensate for changes in blood resistivity. In one example, the adjusted thoracic fluid amount indication is given by: $TFA_{adj}=TFA_{raw}\cdot(\rho_{Blood,\,current})\div(\rho_{Blood,\,baseline})$. In this equation, TFA_{adj} is the adjusted value of the thoracic fluid amount, $(\rho_{Blood,\,baseline})$ is the baseline value of the blood resistivity, and $(\rho_{Blood,\,current})$ is the current value of the blood resistivity. In the present case, since the same electrodes are used for both the baseline and current blood resistance measurements, the resistivity ratio $(\rho_{Blood,\,current})\div(\rho_{Blood,\,baseline})$ is given by the corresponding ratio of the blood resistances, i.e., $(Z_{Blood,\,current})\div(Z_{Blood,\,baseline})$.

In a further example, such as where the implantable medical device **300** optionally includes a posture sensor or detector **318**, a separate baseline impedance or resistivity is provided for different postures, since posture affects thoracic impedance measurements. In one example, a separate baseline impedance or resistivity is stored for upright postures (e.g., sitting or standing) than for recumbent postures (e.g., supine, prone, left lateral decubitus, right lateral decubitus). In a further example, a separate baseline impedance or resistivity is stored for one or more of the different subtypes of upright or recumbent postures. In compensating the thoracic fluid amount indication, the posture compensation module **320** compensates a particular resistivity measurement by using a baseline resistivity that corresponds to the then-current posture indicated by the postures detector **318**. One example of a suitable posture detector **318** is a commercially available two-axis accelerometer, such as Model No. ADXL202E, manufactured by Analog Devices, Inc. of Norwood, Mass., USA.

The compensated thoracic fluid amount indication can be stored in the implantable medical device **300** or transmitted to the external device **316**. Moreover, in one example, the implantable medical device **300** or external device **316** is capable of storing a history of the values of the thoracic fluid amount indication to assist the physician in managing the CHF state of the patient. In one example, the external device **316** is capable of displaying a graph, histogram or other chart of such thoracic fluid amount values.

In a further example, the implantable medical device **300** or the external device **316** determines whether heart failure decompensation, pulmonary edema, or pleural effusion is present, such as by comparing an increase in the blood-resistivity-compensated thoracic fluid amount indication to a corresponding first threshold value to deem one or more of these conditions to be present.

In yet a further example, the implantable medical device **300** or the external device **316** predicts whether heart failure decompensation, pulmonary edema, or pleural effusion is likely to become present in the future, such as by comparing an increase in the blood-resistivity-compensated thoracic fluid amount indication to a corresponding second threshold value to deem one or more of these conditions to be likely in the future. The second threshold used for the condition prediction may be different from the first threshold used for the condition detection. In one example, the second threshold value reflects a smaller increase in the thoracic fluid amount indication than the first threshold value.

In yet a further example, the implantable medical device **300** adjusts a therapy to the patient using the thoracic fluid amount indication. In one example, an increase in the thoracic fluid amount indication triggers an increase in a rate at which pacing pulses are delivered to the heart. In another example, a change in the thoracic fluid amount indication results in altering another programmable parameter of cardiac pacing or cardiac resynchronization therapy, such as, for example, atrioventricular (AV) delay, particular cardiac stimulation

sites, interventricular delay, or intraventricular delay. In a further example, a change in the thoracic fluid amount indication triggers the providing of a warning or other indication to the patient to adjust a medication level (for example, a diuretic).

Another application for the present systems, devices, and methods is in anemia detection. Anemia is a pathological condition that is often present in CHF patients. Diagnosing and treating anemia will improve a patient's cardiac function. Therefore, there is a need to detect anemia in CHF patients, for example, to communicate a diagnosis regarding the anemia status to a CHF patient's health care provider.

FIG. **5** is a flow chart illustrating generally one example of an anemia detection method. At **500**, a baseline blood impedance is established. In one example, this includes obtaining one or more near-DC blood impedance measurements (typically taken within the same blood vessel or heart chamber) such as described above with respect to **402** of FIG. **4**. In one example, the baseline blood impedance is established by computing a central tendency (e.g., average, median, low-pass filtered, etc.) value of a series of such measured blood impedances over a desired time interval (for example, one month).

At **502**, a current blood impedance is measured. This near-DC blood impedance measurement is typically performed in the same manner and location as described above for establishing the baseline. At **504**, the current blood impedance is compared to the baseline blood impedance. As described above, a higher percentage of red blood cells tends to increase blood impedance. Therefore, when the current blood impedance falls far enough below the baseline blood impedance, then anemia may be indicated. Therefore, in one example, if the current blood impedance falls below the baseline blood impedance by at least an offset threshold value, then anemia is declared to be present at **506**. In one example, the offset value is a fixed or programmable percentage of the baseline blood impedance (e.g., 5%, 10%, 20%, etc.). The offset value is typically set to prevent normal physiological variations in blood impedance from triggering an anemia detection. In another example, such as by choosing a different threshold value, the comparison predicts that anemia is likely to occur (e.g., if the blood impedance falls at least 10% below its baseline value, then future anemia is predicted; if the blood impedance then falls at least 20% below its baseline value, then present anemia is declared).

In one further example, if anemia is predicted or declared present, that information is telemetered or otherwise communicated to the patient's health care provider from the implantable medical device, such as by providing such information to an external device for storage or display. In one example, such communication takes place the next time that the implantable medical device is interrogated by a programmer or other external interface. In another example, such the implantable medical device itself initiates a telemetric or other communication of such information to an external device. In yet a further example, an anemia warning is provided to the patient, either directly by the implantable medical device (e.g., an audible warning), or via an external interface device.

In some examples, a method includes detecting a first blood resistivity at a first time using at least one implanted intravascular or intracardiac electrode in a patient. In an example, a second blood resistivity is detected at a second time using the at least one implanted intravascular or intracardiac electrode. In an example, the second blood resistivity is compared to the first blood resistivity. If the second blood resistivity is less than the first blood resistivity by at least a threshold value, then, in some examples, an indication or prediction of anemia is declared to be present in the patient. In an example, the first blood resistivity and the second blood resistivity are detected in the same blood vessel or heart chamber. In an example, the first blood resistivity and the second blood resistivity are detected under like conditions with respect to at least one of a cardiac cycle, a respiratory cycle, a posture, and a circadian cycle. In some examples, an indication of anemia predicted or anemia present is communicated to an external device.

In various examples, a system includes an implantable medical device. In an example, the implantable medical device includes first and second electrodes located within the same blood vessel or heart chamber. The implantable medical device, in an example, includes a blood resistivity measurement circuit configured to provide a blood resistivity signal obtained between the first and second electrodes at different first and second times. In an example, a controller is coupled to the blood resistivity measurement circuit and is configured to determine an indication or prediction of anemia by comparing a current blood resistivity to a stored baseline blood resistivity value. The blood resistivity measurement circuit, in some examples, includes a test current circuit to deliver a test current to first and second implantable electrodes. In further examples, the blood resistivity measurement circuit includes a voltage measurement circuit to measure a resulting voltage between third and fourth implantable electrodes. In an example, the system includes a heart signal sensing circuit configured to synchronize the first and second times to like portions of different cardiac cycles. In another example, the system includes a thoracic impedance measurement circuit configured to synchronize the first and second times to like portions of different respiration cycles. In still another example, the system includes a posture sensor and a posture compensation module configured to store different baseline blood resistivities corresponding to different postures obtained from the posture sensor.

As discussed above, measurement of variations in blood resistivity is typically affected by the frequency of the excitation signal that are used. At higher excitation frequencies, blood cells typically become more resistive. Therefore, to yield a more sensitive measurement of anemia, it may be desirable to use a higher excitation frequency than would be used for detecting thoracic impedance, and for correcting the resulting thoracic impedance measurements for changes in blood resistivity. Alternatively, such as for measuring thoracic fluid status, if a change in blood resistivity exceeds a certain threshold value then, in one example, the system automatically switches to a lower excitation frequency that is affected less by changes in blood resistivity.

Although much of the above discussion has emphasized correcting near-DC thoracic impedance measurements to account for changes in blood resistivity, hematocrit-related blood resistivity changes may also affect higher frequency components of the thoracic impedance signal (e.g., respiration components, cardiac stroke components, etc.), somewhat analogous to the way in which patient posture can affect such higher frequency components of the thoracic impedance signal. Aspects of the present blood resistivity measurement and correction techniques may also be used for correcting such higher-frequency components of the thoracic impedance signal. However, the effects of blood resistivity changes at higher frequency may be nonlinear, making correction with a single multiplicative correction factor difficult or impossible. Therefore, a nonlinear correction function may be used, if needed. Such a nonlinear correction function may be empirically determined. In one example, the nonlinear correction function may be implemented as a lookup table. Moreover,

the higher-frequency components of the thoracic impedance signal may be used to infer thoracic fluid status even much of the above discussion focused on particular examples that extract a thoracic fluid status signal from the near-DC component of the thoracic impedance signal.

It is to be understood that the above description is intended to be illustrative, and not restrictive. For example, the above-described embodiments (and/or aspects thereof) may be used in combination with each other. Many other embodiments will be apparent to those of skill in the art upon reviewing the above description. The scope of the invention should, therefore, be determined with reference to the appended claims, along with the full scope of equivalents to which such claims are entitled. In the appended claims, the terms "including" and "in which" are used as the plain-English equivalents of the respective terms "comprising" and "wherein." Also, in the following claims, the terms "including" and "comprising" are open-ended, that is, a system, device, article, or process that includes elements in addition to those listed after such a term in a claim are still deemed to fall within the scope of that claim. Moreover, in this document and in the following claims, the terms "first," "second," and "third," etc. are used merely as labels, and are not intended to impose numerical requirements on their objects.

What is claimed is:

1. A method comprising:

obtaining a first blood resistivity quantity during a first time using at least one implanted intravascular or intracardiac electrode in a subject;

obtaining a second blood resistivity quantity during a second time using the at least one implanted intravascular or intracardiac electrode;

comparing, using a controller, the second blood resistivity quantity to the first blood resistivity quantity; and

declaring an indication or prediction of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least a first amount.

2. The method of claim **1**, wherein obtaining the first blood resistivity quantity and obtaining the second blood resistivity quantity are performed in the same blood vessel or heart chamber.

3. The method of claim **1**, wherein obtaining the first blood resistivity quantity and obtaining the second blood resistivity quantity are performed under a like condition with respect to at least one of a cardiac cycle, a respiratory cycle, a posture, and a circadian cycle.

4. The method of claim **1**, comprising communicating an indication of anemia predicted or anemia present to an external device.

5. The method of claim **1**, wherein declaring an indication or prediction of anemia includes:

declaring an indication of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least a second amount; and

declaring a prediction of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least the first amount and by less than the second amount.

6. The method of claim **1**, comprising communicating information regarding the indication or prediction of anemia.

7. The method of claim **1**, comprising displaying information regarding the indication or prediction of anemia.

8. The method of claim **7**, wherein displaying the information includes displaying the information using an external device.

9. The method of claim **1**, comprising providing an anemia warning with the indication or prediction of anemia.

10. The method of claim **1**, wherein at least one of:

obtaining the first blood resistivity quantity includes detecting a first blood resistivity measurement during the first time; or

obtaining the second blood resistivity quantity includes detecting a second blood resistivity measurement during the second time.

11. The method of claim **1**, wherein at least one of:

obtaining the first blood resistivity quantity includes combining two or more first blood resistivity measurements acquired during the first time; or

obtaining the second blood resistivity quantity includes combining two or more second blood resistivity measurements acquired during the second time.

12. The method of claim **1**, wherein at least one of:

obtaining the first blood resistivity quantity includes averaging two or more first blood resistivity measurements acquired during the first time; or

obtaining the second blood resistivity quantity includes averaging two or more second blood resistivity measurements acquired during the second time.

13. An implantable medical device comprising:

a blood resistivity measurement circuit configured to provide a blood resistivity signal using first and second electrodes;

a controller coupled to the blood resistivity measurement circuit, the controller configured to determine an indication or prediction of anemia by comparing a second blood resistivity quantity to an earlier first blood resistivity quantity; and

a communication circuit coupled to the controller, the communication circuit configured to communicate the indication or prediction of anemia to an external interface device.

14. The implantable medical device of claim **13**, wherein the first and second electrodes are located within the same blood vessel or heart chamber.

15. The implantable medical device of claim **13**, wherein the first and second electrodes are included on a leadwire.

16. The implantable medical device of claim **15**, wherein the first electrode includes a ring electrode and the second electrode includes a tip electrode.

17. The implantable medical device of claim **15**, comprising:

a third electrode included on a housing of the implantable medical device; and

a fourth electrode included on the leadwire;

a test current circuit configured to deliver a test current between the third and fourth electrodes; and

a voltage measurement circuit configured to measure a resulting voltage between the first and second electrodes.

18. The implantable medical device of claim **17**, wherein the fourth electrode includes a ring electrode.

19. The implantable medical device of claim **13**, comprising a heart signal sensing circuit configured to synchronize the first and second times to like portions of different cardiac cycles.

20. The implantable medical device of claim **13**, comprising a posture sensor and a posture compensation module configured to store different first blood resistivity quantities corresponding to different postures obtained from the posture sensor.

21. The implantable medical device of claim 13, comprising a thoracic impedance measurement circuit configured to synchronize the first and second times to like portions of different respiration cycles.

22. The implantable medical device of claim 13, wherein at least one of:

the first blood resistivity quantity includes a first blood resistivity measurement detected during the first time; or

the second blood resistivity quantity includes a second blood resistivity measurement detected during the second time.

23. The implantable medical device of claim 13, wherein at least one of:

the first blood resistivity quantity includes a combination of two or more first blood resistivity measurements acquired during the first time; or

the second blood resistivity quantity includes a combination of two or more second blood resistivity measurements acquired during the second time.

24. The implantable medical device of claim 13, wherein at least one of:

the first blood resistivity quantity includes an average of two or more first blood resistivity measurements acquired during the first time; or

the second blood resistivity quantity includes an average of two or more second blood resistivity measurements acquired during the second time.

25. A system comprising:

means for obtaining a first blood resistivity quantity during a first time using at least one implanted intravascular or intracardiac electrode in a subject;

means for obtaining a second blood resistivity quantity during a second time using the at least one implanted intravascular or intracardiac electrode;

means for comparing the second blood resistivity quantity to the first blood resistivity quantity; and

means for declaring an indication or prediction of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least a first amount.

26. The system of claim 25, wherein the means for declaring the indication or prediction of anemia includes:

means for declaring an indication of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least a second amount; and

means for declaring a prediction of anemia to be present in the patient when the second blood resistivity quantity is less than the first blood resistivity quantity by at least the first amount and by less than the second amount.

27. The system of claim 25, comprising means for communicating information regarding the indication or prediction of anemia.

28. The system of claim 25, wherein at least one of:

the means for obtaining the first blood resistivity quantity includes means for detecting a first blood resistivity measurement during the first time; or

the means for obtaining the second blood resistivity quantity includes means for detecting a second blood resistivity measurement during the second time.

29. The system of claim 25, wherein at least one of:

the means for obtaining the first blood resistivity quantity includes means for combining two or more first blood resistivity measurements acquired during the first time; or

the means for obtaining the second blood resistivity quantity includes means for combining two or more second blood resistivity measurements acquired during the second time.

30. The system of claim 25, wherein at least one of:

the means for obtaining the first blood resistivity quantity includes means for averaging two or more first blood resistivity measurements acquired during the first time; or

the means for obtaining the second blood resistivity quantity includes means for averaging two or more second blood resistivity measurements acquired during the second time.

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专利名称(译)	胸阻抗检测与血电阻率补偿		
公开(公告)号	US7881781	公开(公告)日	2011-02-01
申请号	US12/628384	申请日	2009-12-01
[标]申请(专利权)人(译)	STAHMANN JEFFREYê 哈雷斯特约翰 MOON BOYCE		
申请(专利权)人(译)	STAHMANN JEFFREYê 哈雷斯特约翰 MOON BOYCE		
当前申请(专利权)人(译)	心脏起搏器，INC.		
[标]发明人	STAHMANN JEFFREY E HATLESTAD JOHN D MOON BOYCE		
发明人	STAHMANN, JEFFREY E. HATLESTAD, JOHN D. MOON, BOYCE		
IPC分类号	A61B5/00		
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摘要(译)

本文件讨论了心律管理装置或其他植入式医疗装置，其使用胸阻抗来确定胸腔中存在多少液体，例如用于检测或预测充血性心力衰竭，肺水肿，胸腔积液，低血压。等。由胸阻抗确定的胸液量被补偿血电阻率的变化，这可能由血细胞比容水平或其他因素的变化引起。血液电阻率补偿的胸液量可以存储在装置中或传输到外部装置以进行存储或显示。血液电阻率补偿的胸腔液量还可用于调节心脏起搏，心脏再同步或其他心律管理或对患者的其他治疗。该文献还讨论了用于预测或指示贫血的装置和方法的应用。