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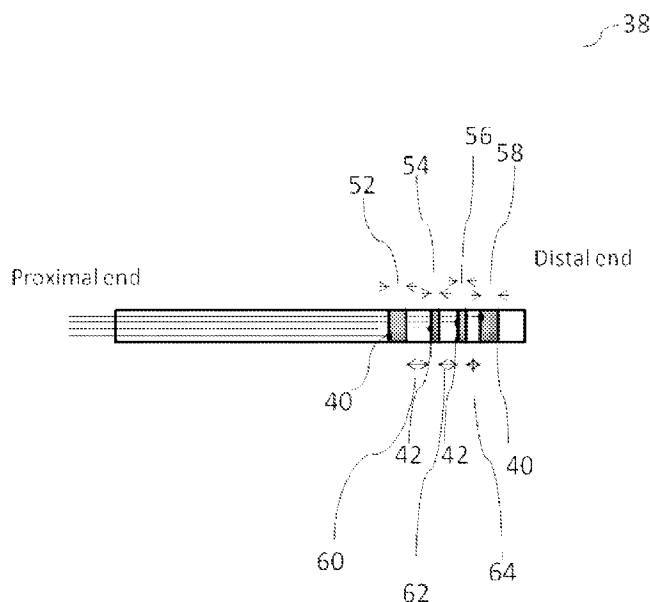


FIG. 4

(57) Abstract: Systems and methods for measurements of anatomical parameters in a region of interest in a subject are provided. A diagnostic device comprises one or more pairs of electrodes for transmitting an electrical signal in the volume of interest and for receiving the voltage signals from the region of interest. A method of using electrical measurements, such as voltage or current, across a range of frequencies derived from data representative of the voltage signals is also provided. These measurements may be used to determine parameters of one or more anatomical features of the subject.



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- as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))
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SYSTEMS AND METHODS FOR MEASUREMENTS OF ANATOMICAL PARAMETERS

The present application claims the benefit of priority to Patent Application Serial No. 1637/CHE/2010 filed in India on June 13, 2010.

FIELD OF THE INVENTION

[0001] The invention relates to the field of medical diagnostic devices and more specifically to a system and method for measuring anatomical parameters of region of interests of internal organs other than blood vessels.

BACKGROUND

[0002] In order to investigate the health of internal organs, it is important to be able to measure certain internal characteristics or parameters that can provide leads for specific diseases and ailments, so that appropriate treatment can be advised and conducted.

[0003] Traditional methods for measurements of anatomical parameters like bony components of different joints to locate degenerative changes in the joints, or detecting abnormal conditions in body organs such as stomach, bladder, lungs etc or detecting internal organ characteristics to determine a subsequent corrective action by stimulating a nerve or any sub component of the organ are not accurate and suffer from several deficiencies such as being static measurements like the X Rays. Traditional measurement systems are also very expensive. These methods also do not provide continuous monitoring that may be needed for certain treatments like incontinence, blood sugar control etc.

[0004] For example, it is widely accepted that congestive heart failure (CHF) is the most common admitting diagnosis among the Medicare patient population, placing a large social and financial burden upon the health care system (>\$40B/yr, 10pts/ 1000 over age 65). A particularly vexing problem is the high rate of readmission following a hospitalization for heart failure (~25% to 30% 30-day readmission rate). Since heart failure admissions account for 80% of heart failure-related costs, there is a great need for effective means of disease management that effectively reduce heart failure-related hospitalizations.

[0005] Impedance cardiography or ICG and electrical cardiometry (ECG) typically involve placement of four dual disposable sensors on the neck and chest. In these techniques electrical and impedance changes in the thorax, a region of interest in the heart, which are used to measure and calculate hemodynamic parameters. ICG measures the baseline impedance (resistance) to this input current. As is known, that due to the heart movement, the blood volume and velocity in the aorta change and ICG measures the corresponding change in impedance and uses this to calculate hemodynamic parameters. ECG uses electrical velocimetry (EV) for its calculations of parameters. EV is based on the fact that the conductivity of the blood in the aorta changes during the cardiac cycle.

[0006] Similarly other organs such as bladder, brain etc require measurements for monitoring the respective health of these organs. For example, in case of incontinence, bladder fluid or urine volume is an important measurement. Currently for incontinence related diagnosis and treatment, ultrasound imaging is used where sound waves are used to visualize the kidneys, ureters, bladder, and urethra. Another technique is cystoscopy where a thin tube with a tiny camera is inserted in the urethra and used to see the inside of the urethra and bladder.

[0007] For brain related disease diagnosis and therapy like hypertension, brain tumour, stroke and trauma etc, the intracranial pressure needs to be determined. Currently treatment is based on external symptoms and MRI (magnetic resonance imaging).

[0008] Therefore, a need exists for improved systems and methods for measurements of anatomical parameters in the region of interests.

[0009] SUMMARY

[0010] In accordance with an aspect of the invention, a diagnostic device is disclosed. The diagnostic device comprises a diagnostic element that includes a first set of electrodes to receive a pre-determined electrical input signal from an excitation and measurement device and to transmit it to a region of interest, a second set of electrodes to receive a plurality of voltage signals from the region of interest and transmit it to a measurement device. The measurement device receives the voltage signals and employs a processor that uses data represented by the voltage signals to measure parameters such as fluid index for a thoracic fluid, an instantaneous volume for chambers of a heart over a cardiac cycle, urine volume in a bladder, or level of cerebrospinal fluid in a brain. In an

exemplary embodiment, the first electrode comprises a first pair of electrodes and the second electrode comprises a second pair of electrodes.

[0011]

[0012] Other goals and advantages of the invention will be further appreciated and understood when considered in conjunction with the following description and accompanying drawings. While the following description may contain specific details describing particular embodiments of the invention, this should not be construed as limitations to the scope of the invention but rather as an exemplification of preferable embodiments. For each aspect of the invention, many variations are possible as suggested herein that are known to those of ordinary skill in the art. A variety of changes and modifications can be made within the scope of the invention without departing from the spirit thereof.

BRIEF DESCRIPTION OF THE DRAWINGS

[0013] The novel features of the invention are set forth with particularity in the appended claims. A better understanding of the features and advantages of the present invention will be obtained by reference to the following detailed description that sets forth illustrative embodiments, in which the principles of the invention are utilized, and the accompanying drawings of which:

[0014] FIG. 1 is a graphical representation showing the magnitude of specific impedance for various tissue types over a range of frequencies;

[0015] FIG. 2 is a graphical representation showing phase of specific impedance for various tissue types over a range of frequencies;

[0016] FIG. 3 is a diagrammatic representation of an exemplary embodiment of a diagnostic device;

[0017] FIG. 4 is a diagrammatic representation of a diagnostic element of the diagnostic device having spaced apart electrodes at pre-determined positions according to one aspect of an exemplary embodiment;

[0018] FIG. 5 is a diagrammatic representation of an exemplary embodiment of a diagnostic device;

[0019] FIG. 6 is a diagrammatic representation of another exemplary embodiment of a diagnostic device;

[0020] FIG. 7 is a diagrammatic representation of an exemplary embodiment of the diagnostic device showing associated electronics;

[0021] FIG. 8 is a diagrammatic representation of an exemplary embodiment of the diagnostic device used in conjunction with other medical devices;

[0022] FIG. 9 is a flowchart representation of exemplary method steps for using the diagnostic device;

[0023] FIG. 10 is a diagrammatic representation of a 2-port network with port voltages and port currents;

[0024] FIG. 11 is a diagrammatic representation of an exemplary embodiment with a multi port network at a distal end and the excitation and measurement entity at a proximal end;

[0025] FIG. 12 is a diagrammatic representation of another exemplary embodiment with a multi port network at a distal end and the excitation and measurement entity at a proximal end;

[0026] FIG. 13 is a diagrammatic representation of an exemplary embodiment for use in measuring electrical response from a region of interest;

[0027] FIG. 14 is a diagrammatic representation for another exemplary embodiment with a different configuration for obtaining the measurements from a region of interest;

[0028] FIG. 15 is a diagrammatic representation of a multi terminal embodiment used for modeling the system of FIG. 13 and FIG. 14;

[0029] FIG. 16 is a diagrammatic representation of a multi port network that can use the assumptions of the embodiment of FIG. 15;

[0030] FIG. 17 is a diagrammatic representation of a multi port network that can use the method of the invention where 6 degrees of freedom are presented in the load network;

[0031] FIG. 18 is a diagrammatic representation of an embodiment with an exemplary 3-port passive network 6 complex impedances;

[0032] FIG. 19 is a diagrammatic representation of another embodiment with an exemplary 3-port network;

[0033] FIG. 20 is a flowchart for the exemplary method steps of for calibration and de-embedding;

[0034] FIG. 21 is a flowchart representation of exemplary steps in the method for multiple frequency excitation according to an exemplary method;

[0035] FIG. 22 is a graphical representation that shows examples of current values that may be provided to a heart over a range of frequencies;

[0036] FIG. 23 is a block diagrammatic representation of a system for multiple frequency excitation;

[0037] FIG. 24 shows an exemplary implementation of a pseudo random binary sequence;

[0038] FIG. 25a shows the exemplary pseudo random binary sequence in time domain;

[0039] FIG. 25b shows a zoomed in portion of the exemplary pseudo random binary sequence in time domain;

[0040] FIG. 26 shows the power spectral density of the exemplary pseudo random binary sequence;

[0041] FIG. 27 shows the phase plot of the exemplary pseudo random binary sequence;

[0042] FIG. 28 shows an exemplary implementation for orthogonal frequency division multiplexed (OFDM) sequence using IFFT;

[0043] FIG. 29 shows a time domain signal for the OFDM sequence of FIG. 28;

[0044] FIG. 30 shows the OFDM Frequency Response for the implementation of FIG. 28; and

[0045] FIG. 31 shows another exemplary implementation for generating a customized multi-frequency sequence involving additions of multiple sinusoids with appropriate phases.

DETAILED DESCRIPTION

[0046] As used herein and in the claims, the singular forms "a," "an," and "the" include the plural reference unless the context clearly indicates otherwise.

[0047] The devices, systems, and methods described herein combine precise physical measurement and tissue characterization at a smaller footprint and at lower cost compared to other standard diagnostic techniques such as, without limitation, impedance cardiography (ICG) for parameters related to the health of a heart. The techniques described herein can further uncover more anatomical details than some other diagnostic approaches and provide several advantages in a variety of uses.

[0048] "Parameter" or "Dimension" or "Anatomical Parameter" are used herein interchangeable and include, without limitation, parameters or dimensions related to an internal organ characteristic other than blood vessels or vascular bodily lumen that involve the circulatory system. These include cross sectional area, diameter, radius, major/minor axis for the region of interest, flow of fluid, level of fluid, pressure of fluid in an organ and any derivatives thereof. It includes for heart, hemodynamic parameters, such as cardiac output, stroke volume, systemic vascular resistance, thoracic fluid content, acceleration index, and systolic time ratio, fluid index etc. The region of interest referred herein includes internal organ regions other than blood vessels. methods described herein may be applied for any specific type of treatment or diagnostic applications for various anatomical features of a subject. Aspects of the disclosure can be applied as stand-alone systems or methods, or as part of a greater diagnostic or therapeutic device or procedure. It shall be understood that aspects of the disclosure can be appreciated individually, collectively, or in combination with each other. Features described in one or more embodiments can be incorporated into other embodiments unless the disclosure specifically says otherwise.

[0049] An electrical network as referred herein is an interconnection of electrical elements such as resistors, inductors, capacitors, conductor wires, voltage sources, current sources and switches.

[0050] A terminal is the point at which a conductor from an electrical component, device or network comes to an end and provides a point of connection to external circuits. A terminal may simply be the end of a wire or it may be fitted with a connector or fastener. In network analysis, terminal means a point at which connections can be made to a network in theory and does not necessarily refer to any real physical object.

[0051] An electrical connector is an electro-mechanical device for joining electrical circuits as an interface using a mechanical assembly. The connection may be temporary, as for portable equipment, or may require a tool for assembly and removal, or may be a permanent electrical joint between two wires or devices.

[0052] As used herein electrical measurements include measurable independent, semi-independent, and dependent electrical quantities including for example voltage by the means of voltmeter (or using oscilloscope, including pulse forms), electric current by the means of ammeter, electrical resistance, conductance, susceptance and electrical conductance by the means of ohmmeter, magnetic flux and magnetic field by means of a Halls sensor, electrical charge by the means of electrometer, electrical power by the means of electricity meter, electrical power spectrum by the means of spectrum analyzer.

[0053] Electrical impedance as referred herein is defined as vector sum of electrical resistance and electrical reactance. Inductance is defined as frequency proportionality coefficient for reactance, and capacitance defined as reciprocal frequency proportional coefficient for reactance.

[0054] Voltage between any two points as generally referred herein is the electrical potential difference between the two points and is also referred herein as voltage difference or voltage drop.

[0055] As used herein, the phrase "peak to rms ratio" (PAR) means the value obtained for a waveform by the division of peak amplitude of the waveform by the root mean square value for the waveform. It is a dimensionless number generally expressed as a ratio of a positive rational number to one. It is also known in the art as "crest factor," peak to average ratio, or by other similar terms, known to those of ordinary skill in the art. PAR values for a variety of standard waveforms are generally known. PAR values may be obtained from theoretical calculations, or it may be measured using some PAR meters for specific situations.

[0056] As used herein, the phrase "Signal to noise ratio" (often abbreviated SNR or S/N) means the ratio of signal power to the noise power associated with the signal. The noise power is considered to corrupt the signal power. Hence, SNR is a measure to quantify how much a signal has been corrupted by noise. Ideally, a good SNR should have a ratio much higher than 1:1.

[0057] While preferable embodiments of the invention have been shown and described herein, it will be obvious to those skilled in the art that such embodiments are provided by way of example only. Numerous variations, changes, and substitutions will now occur to those skilled in the art without departing from the invention. It should be understood that various alternatives to the embodiments of the invention described herein may be employed in practicing the invention.

[0058] The invention provides systems and methods for measurements of anatomical parameters in region of interests of heart, bladder, brain, stomach and other organs. Various aspects of the invention described herein may be applied to any of the particular applications set forth below or for any other types of treatment or diagnostic applications for various anatomical features of a subject. The invention may be applied as a stand alone system or method, or as part of a diagnostic or therapeutic device or procedure. It shall be understood that different aspects of the invention can be appreciated individually, collectively, or in combination with each other.

[0059] The methods herein exploits the feature that the various bodily elements such as bodily fluid, tissue wall, fatty tissue, calcified tissue, etc. when excited by an electrical impulse or signal give rise to distinct voltage or frequency signatures that can be measured to determine different parameters. For example, FIG. 1 is a graphical representation 10 of impedance magnitude 12 for various tissue types over a range of frequencies 14. Impedance magnitude (absolute value of V_{in}/I_{in} measured in dB) versus frequency (Hz) is provided for aorta 16, blood 18, and fat (average infiltrated) 20. V_{in} represents voltage and I_{in} represents current. The plots of impedance magnitude (absolute value of V_{in}/I_{in} measured in dB) for blood, tissue (aortic vessel) and fat shown indicate that when an excitation (e.g., a sinusoidal current (AC), or any other waveform) at different frequencies is applied in series across the region of interest (1 cubic millimeter, for example) shown by area 21, the impedance magnitude varies depending on the type of bodily material that occupies that volume.

[0060] FIG. 2 is a graphical representation 24 of an example of impedance phase 26 (in degrees) for various tissue types over a range of frequencies 28. Line 30 represents the impedance

phase (angle of V_{in}/I_{in} measured in degrees) of tissue (e.g. aortic vessel) across a frequency range of 100 Hz to 100 MHz; line 32 represents impedance phase (angle of V_{in}/I_{in} measured in degrees) of blood across a frequency range; line 34 represents impedance phase (angle of V_{in}/I_{in} measured in degrees) of fat across a frequency range. V_{in} represents voltage and I_{in} represents current. The plots of impedance phase (angle of V_{in}/I_{in} measured in degrees) for blood, tissue and fat shown indicate that when an excitation (e.g., a sinusoidal current (AC), or any other waveform as described elsewhere) at different frequencies is applied in series across the volume of interest (1 cubic millimeter, for example), the impedance phase depends on the type of bodily material that occupies that volume.

[0061] The impedance magnitude and/or the impedance phase may also be additive as would be known to one skilled in the art. Thus the impedance measurements taken over the range of frequencies can yield the dimensions of the various tissue types. Thus, the unit electrical properties may be converted into volumetric data of the environment, utilizing the uniqueness of the combination.

[0062] Now turning to FIG. 3 is a diagrammatic representation of an exemplary embodiment of a diagnostic device 36 that includes a diagnostic element 38 according to an aspect of the invention. In an exemplary embodiment, the diagnostic element 38 includes at least two spaced apart set of electrodes 40 and 42 at a distal end 44 configured to be placed proximal to the region of interest 46, wherein at least one of the set of electrodes is configured to receive an input current from an excitation and measuring device (also referred to as excitation and measurement device) 48, and another pair (or the same pair) of electrodes is configured to receive a plurality of voltage signals from the region of interest 46 and transmit the voltage signal to the excitation and measuring device 48 at the proximal end 50. The excitation and measurement device 48 receives and measures the plurality of voltage signals to calculate a voltage difference between the spaced apart electrodes. These voltage differences are a function of bodily parameters and can be distinguished as explained in FIG. 1 and FIG. 2. In the exemplary embodiment a pair of electrodes has been referred to for measuring the signals from the region of interest, however any number of electrodes may be employed for this purpose.

[0063] FIG. 4 is a diagrammatic representation of the diagnostic element 38 that includes the spaced apart electrodes 40 and 42 arranged at pre-determined positions that may be symmetric in

one example and non-symmetric in another example, indicated by the reference numerals 60 through 64 as shown in FIG. 3. It may be noted here that the asymmetric positioning i.e. when the electrodes are not spaced apart equidistantly, may yield better sensitivity of measurements according to the methods described herein. The size of the electrodes may be same or different as indicated by reference numerals 52-58. It would be understood by those skilled in the art that the size and spacing of electrodes are designed for optimal performance. In some embodiments, electrodes may be formed of a conductive material. For example, electrodes may include a metal, such as copper, silver, aluminum, gold, or any alloys, plating, or combinations thereof. Electrodes may include exposed portions of wires. Electrodes may include any electrically conductive material in electrical communication with electronics for providing and/or receiving an electrical signal and/or current. It may be noted that in some configurations, the electrodes to send the input current and the electrodes to transmit the attenuated signals may be pre-determined. Further it is possible to select more than one pair of electrodes to send the input current and similarly more than one pair may be selected to sense the voltage signals from the region of interest. In specific embodiments electrodes may be provided in close proximity to an anatomical feature. For example, electrodes may be provided in close proximity to heart or other organ, where the electrodes may contact the outside surface and/or inside surface of the organ.

[0064] The technology and methods described in the disclosure that employ the diagnostic device of FIG. 4 can be used to monitor important trends of a few parameters of interest such as thoracic fluid volumes, systolic and end diastolic volumes of cardiac chambers etc. and use this to provide an early detection and warning system of impending clinical deterioration

[0065] In one exemplary embodiment of the technology, the diagnostic device may be placed proximal to the heart region and is arrayed to send and receive low-amplitude, multi-frequency electrical current signals in multiple transthoracic vectors. As described earlier, different tissues (blood, fat, cardiac tissue, etc.) have different frequency responses and therefore a unique signature when traversed by electrical signals. By applying methods described herein the information on thoracic fluid may be derived (e.g. a fluid “index”). The trend in fluid index in any one given patient may be examined for signs of heart deterioration. For example, in the immediate post-hospital discharge setting, the device-measured fluid levels are monitored, perhaps considered to reflect the “optimal” volume (= euvolemic) status. If a significant deviation from this baseline is detected, a clinical alert may be triggered. The availability of other physiologic data such as left ventricle (LV)

end-diastolic volume estimates, heart rate, etc. may be used to further enhance the system accuracy and sensitivity of the measurements.

[0066] Another exemplary embodiment involves the use of the diagnostic device for patients suffering from incontinence. Incontinence affects several 10's of millions of people around the world. They have reduced control of a few specific abdominal muscles and it causes them to lose control when the bladder is semi full. The diagnostic device for such use may be positioned across the lower abdomen, over the bladder, and is arrayed to send and receive low-amplitude, multi-frequency electrical current signals in multiple transabdominal vectors. Different tissues (skin, blood, fat, etc.) and urine inside the bladder will have different frequency responses as described earlier and therefore a unique signature when traversed by electrical signals. By applying methods described herein the information on amount of urine in the bladder can be obtained and a alarm can be set to go off (vibratory) when the bladder is full enough but before the patient loses control and has an accident.

[0067] The exemplary embodiment of the diagnostic device may also be used to estimate the level of cerebrospinal (CS) fluid (and hence the cranial pressure) noninvasively. Similar to the application that measures the amount of urine in the bladder, it is possible to trend the amount (increase or decrease) of CS fluid in the cranial cavity. Since the cranial cavity is inelastic, an increase in the CS fluid volume directly leads to an increase in intracranial pressure, which is a parameter of critical clinical importance. By using the diagnostic device the voltages across different cranial vectors are measured yielding the change in the volume occupied by the CS fluid over time that can be monitored for increase or decrease of the cranial pressure.

[0068] It would be appreciated by those skilled in the art that these are some non-limiting use examples of the diagnostic device where it can be placed externally on a body part close to the non-vasculature regions of the organs whose parameters need to be observed or measured.

[0069] FIG. 5 is a diagrammatic representation of an exemplary embodiment of the diagnostic device in the form of a band 66 that includes a strap 68 having the electrodes 40 and 42 integrated into it with known techniques, and a clasp 70 for placing it in position in proximity to the organ for the region of interest.

[0070] For example the band 66 may be placed around a chest of a patient being diagnosed or treated for heart failure or related disease, and may be used for continuous monitoring. It may be placed in proximity of the abdomen for inconsistency related treatments as explained herein above, or around a head for a patient for brain related disease and treatment. Such a band may be placed on any other bodily part where the measurements are needed for specific parameters. The exemplary band may be sized according to the bodily part where it is to be placed and may include other features for holding the band in contact and in place, and for ease of patient wear. Other exemplary configurations include a separate transmitter and receiver adapted to be worn on the wrists. In another example, a bathroom scale-like device may be used that is connected to electrodes mounted on handles which will be grasped by the patient. In another example the diagnostic device may be provided as an adjunct to the patient's footwear.

[0071] In another exemplary embodiment the diagnostic device or a part thereof for example the diagnostic element may be placed in vivo. FIG. 6 is a diagrammatic representation 72 of an implantable defibrillator 74 where the diagnostic element 38 that is the electrodes are embedded in the leads of the defibrillator (residing in left ventricle and atrium and the coronary sinus or a subset of these locations). The various electrodes can be used to either excite or measure voltages within the heart. Using the methods described herein, it is possible to obtain instantaneous volumes of the four chambers of the heart over the entire cardiac cycle. This advantageously allows a clinician to obtain many critical cardiac parameters. The excitation and measurements of electrical voltages and currents can be initiated and maintained by the implanted defibrillator and the information can be telemetrically transmitted to points outside the body. For such use the diagnostic element can also be a stand alone implant or be placed inside a cardiac pacemaker or an implantable cardioverter defibrillator (ICD) and be used for home and hospital monitoring. This arrangement advantageously would not require any new leads or work flow implementation.

[0072] FIG. 7 is a diagrammatic representation of the diagnostic device 76 having the diagnostic element 38 coupled to the excitation and measurement device that is shown with associated circuitry. The excitation and measurement device is incorporated in a dongle 78 and a host computer like a personal computer (PC) 80. The dongle 78 includes an electronics board that comprises a signal conditioning modules 82 which send and receive signal to and from the electrodes (or at least few of them). Each signal conditioner may be coupled to a high precision circuit shown general by 84 (for non-limiting example: a 16 bit data acquisition [DAQ]

digital/analog circuit, or an 18bit DAQ), which converts a digital signal to an analog signal and is coupled to a level 1 signal processing unit 86. The signal may comprise any waveform known in the art. For example, the signal may comprise a sinusoidal waveform, square waveform, triangular waveform, saw tooth waveform, pulse waveform, or any combinations thereof. These data acquisition circuits digitize the output voltage measured by the excitation and measurement device, and the digitized signal may be processed first by a level 1 signal processing unit 86. It may be noted here that any discussion of a computer or host computer, or any specific type of network device may include, but is not limited to, a personal computer, server computer, or laptop computer; personal digital assistants (PDAs). In some embodiments, multiple devices or processors may be used. In some embodiments, various computers or processors may be specially programmed to perform one or more step or calculation or perform any algorithm, as described herein

[0073] This signal processing unit 86 can be split into multiple sections, some residing in hardware in the dongle and the rest on a host computer as shown in FIG. 7 by a level 2 signal processing unit 88. This splitting is not mandatory and in some embodiments, the signal processing units 86 and 88 may be incorporated entirely on the host computer, or the signal processing units 86 and 88 may be provided entirely on a dongle. In one exemplary embodiment, a first level of the signal processor (level 1 signal processing unit) may reduce the sheer volume of data making it amenable to be transferred into a PC where the rest of the processing is done. A level 1 or a first level signal processing unit may compress the output signal such that essential information is not lost, but noise is reduced in the data, thus reducing the size of the data packet (or processed digital signals) passed to a level 2 or second level signal processing unit. In one exemplary embodiment the level 1 signal processing unit may remove the effects of device resistance and coupling.

[0074] The level 2 signal processor may be part of a computer or part of the electronics board itself. This level 2 processor may execute an algorithm or a technique or a method to determine the dimensional aspects of interest (measurements, tissue characterizations, displays of the same for non-limiting example). The level 1 and level 2 processors may be contained in a single processor which carries out both functions of the separate level 1 and level 2 processors described. Also, at least one of the processors and/or conditioner is configured and/or programmed to remove the effects (at least in part, if not entirely) of device resistance and coupling.

[0075] FIG. 8 is a diagrammatic representation of the exemplary embodiment 98 showing the integration of the diagnostic device with other medical devices for enhanced measurements. The diagnostic device 38 that includes the excitation and measurement device 92 receives the voltage signals from at least one set of electrodes of the diagnostic device 38 and converts (and/or transform) them to measurements and/or other anatomical information for non-limiting example using a processing unit 94. The processing unit 94 may also be coupled to an ECG capture unit 98 and another imaging modality such as X Ray or ultrasound 100 for further processing. The results from the processing unit 94 can be overlaid on the outputs from the ECG capture unit and the imaging modality to provide more insights for diagnosis and treatment.

[0076] In some embodiments, the processing unit 94 outputs the measurements (may include only the measurements from the diagnostic device or enhanced measurements or results after combining data from other medical devices such as ECG or imaging modality data) that are displayed on a display device 96. The display device 96 shows the results in different forms, dimension values, graphical representation or visual representations overlaid on angiograms. The display device and the processor or part of the processor may be incorporated in a host computer.

[0077] In some embodiments, the display is a video display. Video displays may include devices upon which information may be displayed in a manner perceptible to a user, such as, for example, a computer monitor, cathode ray tube, liquid crystal display, light emitting diode display, touchpad or touch screen display, and/or other means known in the art for emitting a visually perceptible output. Further in some embodiments, the visual representation may be monochromatic, or may include color. In some embodiments, colors or shading may be indicative of the dimensions.

[0078] The dimensions or parameters may be automatically displayed by the processing unit onto the display unit. Alternatively, the dimensions may be displayed in response to a user input. Examples of user input may include, but are not limited to, a cursor over a portion of the display (which may be controlled by a pointing device such as a mouse, trackball, joystick, touchscreen, arrow keys, remote control), or a keyboard entry. In some embodiments, the dimensions are provided in proximity to a cursor, or other user input. For example, as a user positions a mouse cursor over a portion of the visual representation, the dimension at that portion may be revealed. In other embodiments, all dimensions may be displayed.

[0079] In the embodiments described herein the signal processing may be based on different techniques including but not limited to calculating the voltage difference based on spatial diversity of the at least two electrodes. In another example, the voltage difference may be calculated based on frequency diversity of the input current and the measured voltage signals. In yet another example, the voltage difference is based on tissue diversity or the characteristic of the bodily fluid of the region of interest. The voltage distribution obtained from the plurality of electrodes created by the injection of current across a pair of electrodes is unique as mentioned herein above and is describable by a set of equations that govern the electrical transport through the media surrounding the electrodes. These equations are solvable due to the diversity of frequencies of measurement (frequency diversity), diversity in the dimensions over which electrical transport is observed (spatial diversity). Any electrical characteristic may be measured and/or determined from the measurements, including but not limited to voltage distribution, impedance distribution across the region of interest.

[0080] For frequency diversity, electrical measurements may be performed over a range of frequencies. The magnitude and the phase of the response may be captured. Electrical measurements may be taken over any range of frequency values. Electrical measurements taken may include voltage and current measurements. In one embodiment, the magnitude of the voltage for a given magnitude of input current and also the phase of the voltage with respect to the phase of the input current may be measured. The magnitude and phase may be used in one technique to determine the parameters. These techniques are described in more details in reference to FIG. 21-31.

[0081] The methods described herein exploit the frequency diversity of the input signal, frequency diversity of the measured voltage signals and also in some embodiments the spatial diversity of the electrodes to uniquely determine anatomical parameters from the response observed by the electrodes. There are several methods used to perform the decomposition or transformation of the measured voltage signals. One method is to create an electrically equivalent 3-dimensional model of the near-field using lumped finite-elements and mathematically invert the unit elements to allow the model to recreate the observed set of voltages, and thereby calculating the required parameters. A network may be provided with finite element unit characteristics of the subject's tissue embedded. This may accommodate different types of tissues. For example, the tissue may include blood and fat. An electrical network with a first element (e.g., blood) multiplied by a first dimension may be provided. An electrical network with a second element (e.g., fat) multiplied by a second dimension may be provided. The networks may provide simulations, and the various

dimensions (e.g., first and second dimension) may be altered until the desired response is achieved. This may be provided for any number of dimensions corresponding to various tissue types (e.g., one, two, three, four, or more dimensions). When the desired response is achieved, the dimension values used to achieve the desired response may be the dimensions of the tissue.

[0082] Another method is to embed the electromagnetic simulation within an inverting algorithm, again yielding the dimensions of interest. This may utilize similar steps as described above. Yet another method is to pre-calculate the responses using an off-line set of simulations using a 3-D EM simulator and the algorithm may search this pre-calculated data-set to arrive at the most appropriate set of responses which matches the observed responses. For example, various tissues (e.g., blood, fat, aorta) may each have a corresponding dimension. The dimension value may be 0 or greater. The various dimension values may be varied, and calculations may be performed, to yield a result. The result may be stored in memory. For example, the results may be stored within a look-up table. The look-up table may be searched for results that match one or more measurement, and the set of dimensions used to arrive at that results may be adopted as the dimensions of the tissue. Points outside the finite instances in the look-up table may be accommodated by an interpolation.

[0083] It may be appreciated by those skilled in the art that electrical de-embedding would be needed to obtain the voltage-current distributions found at the distal end where the electrodes are located based solely on the actual measurements that are performed at the proximal end of the guide-wire or catheter. This may include taking into consideration material properties of the devices, or device components, such as the wires or electrodes. Measurements may be calibrated to take such variations into account to yield accurate and precise measurements. De-embedding may occur for systems with any number of terminals, e.g., 2 port, 4 port, or any other number. Electrical values (e.g., voltage, current) may be transformed between the distal end and the proximal end of the diagnostic element as described herein.

[0084] FIG. 10-FIG. 20 describe some more novel techniques for calibration and de-embedding of the measured voltage signals.

[0085] It would be appreciated by those skilled in the art that the devices, systems, and methods described herein provide precise physical measurement and tissue characterization at a smaller footprint and at lower cost as compared to other standard diagnostic techniques. The

techniques described herein can further uncover anatomical details than some other diagnostic approaches and provide several advantages in different use case scenarios

[0086] FIG. 9 is a flowchart representation 102 of exemplary steps for a method for measuring the anatomical parameters for a region of interest. The method includes a step 104 for providing at least two sets of spaced apart electrodes configured to be placed proximal to a region of interest, a step 106 for receiving an input current from an electrical excitation source across at least one pair of the spaced apart electrodes and transmitting the input current to the a region of interest, a step 108 for receiving a plurality of voltage signals from the at least one set of spaced apart electrodes, a step 110 for measuring the plurality of voltage signals as a function of voltage difference between at least one set of the spaced apart electrodes; and a step 112 for converting the voltage differences to one or more parameter measurements through the various techniques that have been described herein.

[0087] Now turning to FIG. 10-20, the description relates generally to estimating a remotely located multi port electrical network and more specifically calibration and de-embedding of the electrical signals to and from the multi port network. These techniques are useful in determining the measurements from the voltage signals received by the diagnostic element as described earlier.

[0088] To provide a general background that has been addressed for calibration and de-embedding the following explanation is provided. In several applications that involve measuring electrical current or voltages, the measurement and the excitation apparatus is at a physical distance from the sensors or the load across which these measurements are desired. Usually conductors connect the electrical source and measurement apparatus and the load. The electrical source, measurement apparatus, conductors, and load form an electrical network. The electrical network has been modeled using electrical parameters. There are many types of parameters used in the art for modeling:

[0089] Z parameters, also called the impedance parameters of a network, relate the voltage and currents of a multi-port network. As an example of a 2 port network 210, with reference to FIG. 10, the two voltages and two currents are related by through Z parameters as follows:

$$\begin{pmatrix} V_1 \\ V_2 \end{pmatrix} = \begin{pmatrix} Z_{11} & Z_{12} \\ Z_{21} & Z_{22} \end{pmatrix} \begin{pmatrix} I_1 \\ I_2 \end{pmatrix} \text{ where } Z_{11} = \frac{V_1}{I_1} \Big|_{I_2=0} \quad Z_{12} = \frac{V_1}{I_2} \Big|_{I_1=0} \quad Z_{21} = \frac{V_2}{I_1} \Big|_{I_2=0} \quad Z_{22} = \frac{V_2}{I_2} \Big|_{I_1=0}$$

For the general case of an N-port network, it can be stated that

$$Z_{nm} = \frac{V_n}{I_m}, \{I_k\} = 0, k \neq m$$

$$n, m, k = 1, \dots, N$$

[0090] Y parameters, also called as Admittance parameters of a network, relate the voltage and currents of a multi-port electrical network. As an example of a 2 port network, the 2 voltages and 2 currents are related by through Y parameters as follows

$$\begin{pmatrix} I_1 \\ I_2 \end{pmatrix} = \begin{pmatrix} Y_{11} & Y_{12} \\ Y_{21} & Y_{22} \end{pmatrix} \begin{pmatrix} V_1 \\ V_2 \end{pmatrix} \text{ where } Y_{11} = \frac{I_1}{V_1} \Big|_{V_2=0} \quad Y_{12} = \frac{I_1}{V_2} \Big|_{V_1=0} \quad Y_{21} = \frac{I_2}{V_1} \Big|_{V_2=0} \quad Y_{22} = \frac{I_2}{V_2} \Big|_{V_1=0}$$

[0091] The S parameters, also called the Scattering parameters of a network, relate the incident and reflected power waves. The relationship between the reflected, incident power waves and the S-parameter matrix is given by:

$$\begin{pmatrix} b_1 \\ b_2 \end{pmatrix} = \begin{pmatrix} S_{11} & S_{12} \\ S_{21} & S_{22} \end{pmatrix} \begin{pmatrix} a_1 \\ a_2 \end{pmatrix} \text{ where } a_n \text{ and } b_n \text{ are the incident and reflected waves, and are related to the port voltages and currents.}$$

[0092] H parameters, also called the Hybrid parameters relate the port voltages and currents in a different way. For a 2-port network we have

$$\begin{bmatrix} V_1 \\ I_2 \end{bmatrix} = \begin{bmatrix} h_{11} & h_{12} \\ h_{21} & h_{22} \end{bmatrix} \begin{bmatrix} I_1 \\ V_2 \end{bmatrix} \text{ where } h_{11} \stackrel{\text{def}}{=} \frac{V_1}{I_1} \Big|_{V_2=0} \quad h_{12} \stackrel{\text{def}}{=} \frac{V_1}{V_2} \Big|_{I_1=0} \quad h_{21} \stackrel{\text{def}}{=} \frac{I_2}{I_1} \Big|_{V_2=0} \quad h_{22} \stackrel{\text{def}}{=} \frac{I_2}{V_2} \Big|_{I_1=0}$$

[0093] G parameter, also called the inverse Hybrid parameters of a network, relate the voltages and current as follows:

$$\begin{bmatrix} I_1 \\ V_2 \end{bmatrix} = \begin{bmatrix} g_{11} & g_{12} \\ g_{21} & g_{22} \end{bmatrix} \begin{bmatrix} V_1 \\ I_2 \end{bmatrix} \text{ where } g_{11} \stackrel{\text{def}}{=} \frac{I_1}{V_1} \Big|_{I_2=0} \quad g_{12} \stackrel{\text{def}}{=} \frac{I_1}{I_2} \Big|_{V_1=0} \quad g_{21} \stackrel{\text{def}}{=} \frac{V_2}{V_1} \Big|_{I_2=0} \quad g_{22} \stackrel{\text{def}}{=} \frac{V_2}{I_2} \Big|_{V_1=0}$$

[0094] All the above formulations are related, and one set of parameters can be derived from another. This is well known and established in the art. The Z and Y parameter matrices are inverses of each other. The H and G parameter matrices are inverses of each other. The Y and S parameters are also related, and can be derived from each other. All the mentioned types of models

are electrically equivalent. The choice of implementation depends on convenience and specific needs of a problem.

[0095] In some of these electrical networks, measurements taken at a distant load need to account for the electrical losses and coupling, and compensate for any parasitic effects of electrical networks formed at the electrical source, measurement apparatus and the conductors. This problem has been dealt with extensively for a single load situated remotely and connected across a pair of conductors that connects to an excitation and measurement apparatus proximally. This problem is usually solved using a two port network formulation. It is a commonly used technique in high precision measurements and is popularly referred to as Port Extension. Nodal analysis, Mesh analysis, Superposition methods have been proposed to solve linear electrical networks. Transfer functions have also been proposed for two port networks.

[0096] However, few solutions exist when the load is not a simple single load but a distributed network with multiple ports forming a load network. Such systems have multiple conductor wires and multiple measurement entities.

[0097] Therefore there exists a need to accurately measure electrical properties across a distant multi port load network.

[0098] It may be noted here that in the novel techniques provided herein to resolve the above issues, the process of estimating the effects of electrical properties of an intervening multiport network is referred herein as calibration. The process of using the estimated properties of the network to compensate for the network and obtain the compensated measurement is referred as de-embedding.

[0099] Z-parameters (the elements of an impedance matrix or Z-matrix) referred herein are the impedance parameters for an electrical network. The Z-parameters are also known as the open circuit parameters. For determining the kth column of the Z matrix, all but the kth port are opened, current is injected on the kth port and the voltages are analyzed on all ports. The procedure is performed for all N ports ($k = 1$ to N) to obtain the entire Z matrix. Though the exemplary embodiments have been described using Z parameters, the methods and systems described herein are equally applicable to other such parameters as Y, S, H, G parameters.

[00100] A generic multi-port network referred herein includes ports 1 to N, where N is an integer depicting the total number of ports. For port n, where n is ranging from 1 to N, the associated input current through that port to the network is defined as I_n and the voltage across that port is defined as V_n .

[00101] For all ports the output voltages may be defined in terms of the Z-parameter matrix and the input currents by the following matrix equation:

$$[00102] \quad V=ZI$$

[00103] where Z is an $N \times N$ matrix the elements of which can be indexed using conventional matrix notation. In general the elements of the Z-parameter matrix are complex numbers and functions of frequency. For a one-port network, as will be clear to one skilled in the art, the Z-matrix reduces to a single element, that is the ordinary impedance measured between the two terminals.

[00104] For a three port network, an equivalent relationship between port voltages and currents is

$$I = Y*Z$$

where Y is an $N \times N$ matrix. Y is related to Z. Generally speaking, Y is the matrix inverse of Z. In some special circumstances, either Z or Y becomes non-invertible.

[00105] Now referring to the drawings, FIG.11 is a diagrammatic representation of an exemplary embodiment 212. The problem being solved by the instant disclosure specifically deals with estimation of an electrical network 218 of a distant zone (herein referred to as a load network) when it is excited by an electrical stimulus from the near end. The load network 218 situated on the distal end is connected to a set of stimulating and measuring devices 214 on the proximal end through a bunch of conductors 216 whose combined electrical property is fixed but unknown. The stimulus can be either an arbitrary current or, voltage from the stimulating device located at the proximal end while the measurements are in form of voltage measurements again at the proximal end. The voltage measurement is in general non-ideal (i.e. the voltage measurement devices draw non-zero finite currents from the network and hence loads the network). As would be appreciated by those skilled in the art, the solution to this problem described subsequently can be extended and

applied to any area of operation, where the electrical network to be estimated is situated at a remote location where in-situ excitation and measurements are not feasible.

[00106] It would be understood by those skilled in the art that for an n-port load network, there would be multiple conductor wires (up to n pairs) running down to the proximal end connecting to an excitation entity and at least to corresponding “n” measurement entities. An additional reference measurement is also performed across two arbitrary nodes in the circuit, such that it has independent information from the previous n measurements. The choice of the pair of arbitrary nodes in the circuit for reference measurement would play a significant role in determining the overall accuracy of the calibration and de-embedding.

[00107] An exemplary embodiment 220 is shown in FIG. 12 that embodies the electrical network of FIG. 11, with the excitation and measurement entity 48 having an excitation entity 233, a reference resistor RRef 234, and voltage measurement entities VM1 231 for measuring voltage drop across the resistance RRef and voltage measurement entities 236, 238, 240 shown as VM2, VM3, and VM4 for measuring proximally the distal voltages of the multiport load network 218 through a multiport inter connecting network 216. The working of such an embodiment is described in more detail in reference to FIG. 13 herein below.

[00108] A specific practical scenario is illustrated in FIG. 13, through the exemplary embodiment 220 which is used for measuring proximal voltages corresponding to distal voltages across four conductors connected to the distal end electrodes 40 and 42 at distal end 44 placed proximal to the region of interest. The four electrodes are connected to four long conductors 232 which gets terminated on a connector on the proximal end 50. Though four electrodes are shown for the exemplary embodiment, three or more electrodes can be used in different configurations needed for measurements and these are included in the scope of the method described herein. The connector is connected to a hardware which provides the stimulus across the two conductors connected to the extreme electrodes and also measures the three voltages across the three pair of conductors. The hardware includes an electrical source and a measurement device 48 having the excitation entity 233 and measurement entities 236, 238, 240. A fourth measurement via the measurement entity 231 is done across a reference resistor 234 which is in series with this network. The entire network is in variant across various load configurations at the distal end 44 but not known to start with and needs to be estimated through carefully chosen load configurations. The calibration method as described

herein estimates this network in order to correctly determine and de-embed the measurements for any arbitrary load network connected distally to it.

[00109] FIG. 14 is another exemplary embodiment 242 with a different configuration for obtaining the measurements. In this embodiment the fourth measurement via the measurement entity 231 (VM1) is done, where VM1 is placed in parallel with the excitation entity 233 to obtain the reference voltage across the excitation entity, while the other three measurements are obtained as mentioned in reference to FIG. 12. The other components in FIG. 14 are same as in embodiment of FIG. 13. It would be appreciated by those skilled in the art that there may be other alternate configurations for obtaining the measurements and the embodiments described in reference to FIG. 12, FIG. 13 and FIG. 14 are to be treated as non-limiting examples. In general any four independent measurements would suffice for estimation of distal load network.

[00110] The measurement entities (VM1, VM2, VM3 and VM4) shown as 231, 236, 238 and 240 in FIG. 12, FIG. 13 and FIG. 14 respectively are typically but not limited to, a set of front end buffers and amplifiers for signal conditioning and noise filtering followed by an analog to digital converter. The measurement entity may provide frequency dependent gain to the incident signal across it. In an ideal scenario, a voltage measurement unit should not draw any current from the network it is connected to, but in practice it is impossible to implement the same. However, as would be appreciated by those skilled in the art, the voltage measurement entity can be equivalently modeled, as a cascade of an equivalent parasitic network that accounts for the loading, filtering and other non-idealities followed by an ideal buffer and gain unit that does not draw any input current and only amplify the incident voltage by a fixed amount. Further, the parasitic network can be merged as a part of the in between catheter network and estimated jointly, as is described in more detail herein below.

[00111] FIG. 15 is a terminal representation 244 for the specific scenario of FIG. 12. It will be understood by those skilled in the art that a terminal, generally referred as T_k (V_k , I_k) represents a terminal k whose voltage with respect to an arbitrary ground, represented as GND 254 in FIG. 15 is V_k while the current entering the network through that terminal is I_k . In the current embodiment, the terminals are defined in the following manner. Terminal-0 (T_0), referred also as 246 is the terminal across which a voltage source or, a current source 48 is connected. The voltage measured on Terminal-0 with respect to an arbitrary GND is defined as V_0 , while the current entering the

network through T0 is defined as I0. Terminal-1A (T1A) represented by 248 is one of the differential terminals across which the first measurement is done. This terminal does not source or, sink any current to the network as these terminals are modeled as ideal measurement points. Terminal -1B represented by 250 pairs with Terminal -1A and behaves similarly to Terminal-1A. Terminal-2A, Terminal-2B are the set of differential terminals for the second measurement. Terminal-3A, Terminal-3B are the terminals for the third measurements, while Terminal-4A, Terminal-4B are the set of differential terminals for the fourth measurement. Together, the terminals 2A, 2B, 3A, 3B, 4A, 4B are shown by referral numeral 252 and represent the terminals for proximal voltages in FIG. 15. Each of these terminals don't source or, sink any current. The voltages on these terminals are all measured with reference to the same GND 254 referenced in the previous paragraph.

[00112] On the distal side, Terminal-5, Terminal-6, Terminal-7 and Terminal-8, referred by 256 corresponds to the four electrodes forming the multi port load network 218 that is connected to the measurement entities and excitation source via the multi port interconnecting network 216 as explained herein above. The voltages on these terminals are referred to as V5, V6, V7 and V8 referred also as distal voltages respectively, where all these measurements are done with respect to the same GND 254. The currents entering the network through these terminals are referred to as I5, I6, I7 and I8 respectively.

[00113] The network can be described completely using Z parameter representation as given below :

$$\mathbf{V1} = \mathbf{Z1} * \mathbf{I1} \quad (1)$$

where, $\mathbf{V1}$ and $\mathbf{I1}$ are given by the following matrices,

$$\begin{aligned} \mathbf{V1} &= [V_0 \ V_{1A} \ V_{1B} \ V_{2A} \ V_{2B} \ V_{3A} \ V_{3B} \ V_{4A} \ V_{4B} \ V_5 \ V_6 \ V_7 \ V_8]^T \\ \mathbf{I1} &= [I_0 \ I_5 \ I_6 \ I_7 \ I_8]^T \end{aligned} \quad (2)$$

$\mathbf{Z1}$ is the impedance matrix of the network relating the current vector $\mathbf{I1}$ to the voltage vector $\mathbf{V1}$.

In another embodiment, the voltages of node 1, node 2, node 3 and node 4 representing the distal end electrodes, are represented differentially as:

$$V_1 = V_{1A} - V_{1B}$$

$$V_2 = V_{2A} - V_{2B}$$

$$V_3 = V_{3A} - V_{3B}$$

$$V_4 = V_{4A} - V_{4B} \quad (3)$$

Equation (1) can be now re-written as:

$$\mathbf{V2} = \mathbf{Z2} * \mathbf{I2} \quad (4)$$

where, $\mathbf{V2}$ and $\mathbf{I2}$ are given by the following matrices,

$$\mathbf{V2} = [V_0 \ V_1 \ V_2 \ V_3 \ V_4 \ V_5 \ V_6 \ V_7 \ V_8]^T$$

$$\mathbf{I2} = [I_0 \ I_5 \ I_6 \ I_7 \ I_8]^T \quad (5)$$

$\mathbf{Z2}$ is the impedance matrix of the network relating the current vector $\mathbf{I2}$ to the voltage vector $\mathbf{V2}$.

[00114] In yet another exemplary embodiment 258 as shown in FIG. 16, true for a floating network on the distal side, a port representation on the distal end is shown instead of the terminal representation of FIG. 15 as described herein above. The port voltages P1, P2, P3, P4 and PL1, PL2, PL3 in this exemplary embodiment, are defined as differences between two neighboring terminal voltages, the voltage difference being depicted by reference numerals 260-272 respectively, while the port currents are defined as the current that goes in through one arm of the port and comes out of the network through another arm of the port.

[00115] Those skilled in the art would recognize the equivalence of the representation of FIG. 15 and FIG. 16 for a floating network on the distal side. It would require a few manipulations of rows and columns of the system of equations represented by Equation (4) to come to a new set of equations represented by Equation (6).

$$\mathbf{V} = \mathbf{ZI} \quad (6)$$

where, $\mathbf{V}$ and $\mathbf{I}$ are given by,

$$\mathbf{V} = [V_0 \ V_1 \ V_2 \ V_3 \ V_4 \ V_{L1} \ V_{L2} \ V_{L3}]^T$$

$$\mathbf{I} = [I_0 \ I_{L1} \ I_{L2} \ I_{L3}]^T \quad (7)$$

$\mathbf{Z}$ is the impedance matrix of the network relating the current vector $\mathbf{I}$ to the voltage vector $\mathbf{V}$.

[00116] The floating network system as described by equation 6 is explained in more detail herein below. One skilled in the art would be able to extend the following set of derivations for use cases where the distal network is not floating. In the network depicted by FIG 15., V_0 is the voltage applied to the network, I_0 is the current getting into the network. If the excitation is a perfect voltage source, V_0 is fixed to the value of the voltage source. Similarly, for a perfect current source excitation, I_0 is fixed to the value of the current for the current source. However in practice, an ideal voltage source or, a current source do not exist. It may be possible to measure the voltage V_0 or, current I_0 precisely without affecting the network appreciably. However, such measurements would involve intricate electronics especially when the frequency of excitation is high, and therefore increase the hardware complexity. Aspects of the present technique advantageously overcome this problem by deriving a method to identify the load network without the need of knowledge of the voltage V_0 or, current I_0 as explained herein below.

[00117] Since, voltage V_0 , is not needed, it is taken off from the first row from the system of equations defined in Equation (6). The new system of equations are written as:

$$\begin{aligned}
 V_1 &= Z_{10}I_0 + Z_{11}I_{L1} + Z_{12}I_{L2} + Z_{13}I_{L3} \\
 V_2 &= Z_{20}I_0 + Z_{21}I_{L1} + Z_{22}I_{L2} + Z_{23}I_{L3} \\
 V_3 &= Z_{30}I_0 + Z_{31}I_{L1} + Z_{32}I_{L2} + Z_{33}I_{L3} \\
 V_4 &= Z_{40}I_0 + Z_{41}I_{L1} + Z_{42}I_{L2} + Z_{43}I_{L3} \\
 V_{L1} &= Z_{50}I_0 + Z_{51}I_{L1} + Z_{52}I_{L2} + Z_{53}I_{L3} \\
 V_{L2} &= Z_{60}I_0 + Z_{61}I_{L1} + Z_{62}I_{L2} + Z_{63}I_{L3} \\
 V_{L3} &= Z_{70}I_0 + Z_{71}I_{L1} + Z_{72}I_{L2} + Z_{73}I_{L3}
 \end{aligned} \tag{8}$$

[00118] In the exemplary method, the four measured voltages are grouped in a vector V_M and similarly the load side voltages are grouped in the vector V_L . The load side currents are similarly grouped in vector I_L , as shown in the equations below:

$$\begin{aligned}
 \mathbf{V}_M &= [V_1 \ V_2 \ V_3 \ V_4]^T \\
 \mathbf{V}_L &= [V_{L1} \ V_{L2} \ V_{L3}]^T \\
 \mathbf{I}_L &= [I_{L1} \ I_{L2} \ I_{L3}]^T
 \end{aligned} \tag{9}$$

Now re-writing equation (8) using the nomenclature defined above:

$$\mathbf{V}_M = \mathbf{Z}_{M0}\mathbf{I}_0 + \mathbf{Z}_{ML}\mathbf{I}_L$$

$$\mathbf{V}_L = \mathbf{Z}_{L0}\mathbf{I}_0 + \mathbf{Z}_{LL}\mathbf{I}_L \quad (10)$$

where, $\mathbf{Z}_{M0}$, $\mathbf{Z}_{ML}$, $\mathbf{Z}_{L0}$ and $\mathbf{Z}_{LL}$ are sub-matrices of the impedance matrix ($\mathbf{Z}$) formed by the grouping of the $\mathbf{Z}$ -terms in Eqn (16).

[00119] As would be appreciated by those skilled in the art, the distal side (load side) is also terminated by an arbitrary network which can be modeled as a 3x3 admittance matrix $\mathbf{Y}$ related to the load side voltage vector $\mathbf{V}_L$ and current vector $\mathbf{I}_L$. For passive networks, the admittance matrix, $\mathbf{Y}$ would have 6 independent variables, whereas for a general active network the number of variables would be 9. For some specific scenarios (including that of the one discussed) the load network may have other constraints and the degrees of freedom is lower than 6. In a specific example of FIG. 13, the anatomical constraints while measuring the non vasculature or anatomical parameters, may drive the degrees of freedom of the $\mathbf{Y}$ parameters to 3 or, less.

[00120] Since the current vector $\mathbf{I}_L$ is shown entering the catheter network, a negative sign is used while representing the following load equation:

$$\mathbf{I}_L = -\mathbf{Y}\mathbf{V}_L \quad (11)$$

Using, Equation (11) in Equation (10) the following is derived:

$$\mathbf{V}_L = \mathbf{Z}_{L0}\mathbf{I}_0 + \mathbf{Z}_{LL}\mathbf{I}_L$$

$$\mathbf{V}_L = \mathbf{Z}_{L0}\mathbf{I}_0 - \mathbf{Z}_{LL}\mathbf{Y}\mathbf{V}_L$$

$$(\mathbf{I} + \mathbf{Z}_{LL}\mathbf{Y})\mathbf{V}_L = \mathbf{Z}_{L0}\mathbf{I}_0$$

$$\mathbf{V}_L = (\mathbf{I} + \mathbf{Z}_{LL}\mathbf{Y})^{-1}\mathbf{Z}_{L0}\mathbf{I}_0$$

$$\mathbf{V}_M = \mathbf{Z}_{M0}\mathbf{I}_0 - \mathbf{Z}_{ML}\mathbf{Y}\mathbf{V}_L$$

$$= (\mathbf{Z}_{M0} - \mathbf{Z}_{ML}\mathbf{Y}(\mathbf{I} + \mathbf{Z}_{LL}\mathbf{Y})^{-1}\mathbf{Z}_{L0})\mathbf{I}_0$$

$$\mathbf{V}_M/\mathbf{I}_0 = \mathbf{Z}_{M0} - \mathbf{Z}_{ML}\mathbf{Y}(\mathbf{I} + \mathbf{Z}_{LL}\mathbf{Y})^{-1}\mathbf{Z}_{L0} \quad (12)$$

[00121] Since, $\mathbf{I}_0$ is assumed to be unknown, to resolve a situation where the results would have a scale factor ambiguity, a ratio of two voltages is used instead of the absolute voltage.

Without a loss of generality, the voltage across the reference resistor of FIG. 11 is used, as the reference voltage, V_1 and all other voltages are measured as a ratio to the reference voltage.

$$V_M/V_1 = (Z_{M0} - Z_{ML}Y[I + Z_{LL}Y]^{-1}Z_{L0}) / (Z_{10} - Z_{1L}Y[I + Z_{LL}Y]^{-1}Z_{L0}); \text{ where } M = 2, 3, 4$$

$$= [(Z_{M0}/Z_{10}) - Z_{ML}Y[I + Z_{LL}Y]^{-1}(Z_{L0}/Z_{10})] / [1 - Z_{1L}Y[I + Z_{LL}Y]^{-1}(Z_{L0}/Z_{10})];$$

where $M = 2, 3, 4$

$$= \frac{\overline{Z_{M0}} - \overline{Z_{ML}Y[I + Z_{LL}Y]^{-1}Z_{L0}}}{1 - \overline{Z_{1L}Y[I + Z_{LL}Y]^{-1}Z_{L0}}}$$

where $M = 2, 3, 4$

(13)

where, $\overline{Z_{M0}}$ and $\overline{Z_{L0}}$ are normalized by Z_{10} , and Z_{10} is fixed to unity.

[00122] Thus these equations effectively model the effect of an arbitrary load network connected at a distal end to the measurements done at a proximal end.

[00123] In the formulation above, voltage ratios V_M/V_1 are used. This is because the exact value of V_0 (in the case of voltage excitation) or I_0 (in the case of current excitation) is not known precisely in normal practical situations. However, if these can be determined with enough precision, the calibration method can be formulated with absolute voltages rather than voltage ratios. As such, the invention envisages such alternate formulations where the voltages can be used in forms other than ratios such as absolute value, voltage differences, linear or non-linear combinations of the voltages.

[00124] The exemplary method as described herein uses the above system model for determining the actual voltage difference measurements for an arbitrary load network connected at the distal end through proximal measurements. The next step for the method is to identify the Z parameters of the connecting network along with measurement parasitics, herein referred to as the calibration step. Thereafter, a step of de-embedding is done wherein, the proximal measurements are mapped to (or, fitted to) the distal load network after due consideration for the Z parameters of the connecting network and measurement parasitics.

[00125] In the process of calibration as referred herein, the three voltage ratios with respect to the first voltage is measured for different combination of precisely known load networks connected

on the distal end. It may be noted that for a passive load network, in Equation (13), the number of unknown Z-parameters to be estimated is 23. The Z parameters are obtained using a suitable fitting utility that runs on the set of measured data. Since every configuration provides three voltages, it is necessary to have at least measurements from 8 independent configurations to obtain all the Z parameters. More number of configurations provides better noise immunity to the fitted values. The fitter routine starts with an arbitrary starting point and computes the estimated ratios of voltages across different known load configurations for Equation (13). The method then computes an error metric which is the Euclidian distance between the measured ratios and the estimated ratios. The fitter tries to minimize this error by adjusting the Z parameter values. It is possible for the solution to converge to alternate solutions. However, skilled persons in this art would recognize these challenges and come up with suitable techniques to circumvent them. This can be done by employing suitable optimization techniques. It may be noted that the fitted Z parameters are not the true Z parameters of the network but are a mathematical representation that fits the observation under the constraints of one pre-determined Z-parameter (any one of ZL0). Further, a few Z-parameters are normalized to Z10 and Z10 is fixed to unity, as was mentioned earlier.

[00126] Once the Z parameters have been estimated through the process of calibration, the connecting network can be used to identify any arbitrary load network at the distal end. In specific applications, such as but not limiting to the embodiment of FIG. 13 where a diagnostic device with four distal electrodes (connecting network) is placed proximal to region of interest and the load presented on the distal side is due to the finite conductivity of bodily fluid or, the finite conductivity of wall tissue, the degrees of freedom for the network is 3. The three voltage distributions across the three consecutive pairs of electrodes completely define the Z-parameters of the equivalent electrical network formed by the electrodes inside the region of interest. Similar applications such as measurement of a cross section of a pipe electrically through similar means would also have similar degrees of freedom. Once a measurement of three ratios are taken for an arbitrary load network (with Admittance Y with 3 degrees of freedom), a similar fitter routine can be used to find out the load network. In one example, the fitter routine is initialized by a starting value of Y, which is the best case estimate given by the user. The ratios are accordingly estimated (according to Equation 13) and an error metric is computed as the difference between the measured ratios and the estimated ratios. The error metric is then minimized by adjusting the Y parameters of the load network. The Y parameters representing the lowest error represent the optimal Y parameters for the load network.

[00127] It may be noted that since only three ratios are measured, this method is applicable to identification of networks which has no more than 3 degrees of freedom. As discussed, for an arbitrary network with three ports, the Y parameter can have 9 degrees of freedom. For passive networks, the degrees of freedom are typically 6. Identification of such networks can also be done using extension of the exemplary method. To identify a passive arbitrary load network (with 6 degrees of freedom), the calibration and de-embedding processes needs to be done for two independent interconnecting networks. In practice, it can also be achieved by taking two measurements, one with the actual interconnecting network and the other with a tweaked version of the same. During the calibration phase, precisely known loads are attached to the distal side of the connecting network and the proximal voltages are measured and while maintaining the same load, the connecting network is tweaked using a reversible mechanism (such as a relay 276 shorting the two centre ports 2 and 3 at the proximal end of the embodiment 274 of FIG. 17) and the new proximal voltages are measured.

[00128] The same procedure is then repeated for various load configurations. Using similar principles of the calibration phase, the Z parameters are estimated both for the parent connecting network as well as its tweaked version. Finally, an arbitrary passive load network is connected distal to the same connecting network. The proximal voltages are measured once with the original connecting network and a second time when the connecting network has been tweaked as before. A two sets of measurements are obtained and with the knowledge of the Z parameters of the connecting network and its tweaked version from the calibration phase, it would be possible to unravel all the 6 degrees of freedom of the load network. The method can be also be extended to unravel an arbitrary active three port network with 9 degrees of freedom, by performing measurements using three different connecting networks.

[00129] In an alternate embodiment, an n-port load network is represented by L independent ($L = n^2$) complex impedances. As would be appreciated by skilled persons in this art, the complex impedances bear equivalence with the Z-parameters of the same network. For a passive load network, the number of independent complex impedances would be P ($=n*(n-1)$), since the network would be symmetric. FIG. 18 represents an embodiment 278 with an exemplary 3-port passive network 280 with 6 complex impedances shown generally by reference numeral 282. Any other passive 3-port network topology can be reduced to an equivalent network 280 with the topology

shown in the embodiment 284 of FIG. 19 as well. Other components related to the excitation and measurement entity remain the same as described in earlier figures.

[00130] According to network theory, as would be well understood by those skilled in the art, for any network consisting of an ordered set of discrete impedances, the voltage across any two points (u, v) in the network can be represented as a product of the excitation voltage or, excitation current (ξ_0) and a ratio of sum of polynomials formed by all the impedances present in the network. The denominator polynomial is referred to as the characteristic polynomial of the network consisting of all the impedances in the network. The characteristic polynomial is independent of the points of measurements. Further, if some part of the network consists of distributed elements and other parts consist of discrete impedances, the voltage can still be represented as a product of ξ_0 and the ratio of sum of polynomials formed by all the discrete impedances present in the network, wherein the coefficients of the polynomial would capture the effects of the distributed elements.

[00131] If some of the discrete impedances are of interest, the polynomials can be regrouped into a polynomial of just the discrete impedances of interest. In this case, the coefficients of the regrouped polynomial would contain the effects of the other discrete impedances as well as the distributed elements of the network.

[00132] Referring to FIG.11, where the measurement network 48 and the connecting network 216 are fixed while the multi-port load network 218 is allowed to change through variations of L number of load impedances ($Z_1, Z_2, \dots, Z_L$), the voltage between any two points (u, v) in the network can be written as:

$$V(u, v) = \xi_0 \frac{b_0(u, v) + \sum_i b_{1i}(u, v)Z_i + \sum_i \sum_{j, i \neq j} b_{2ij}(u, v)Z_i Z_j + \dots + b_L(u, v)Z_i Z_j \dots Z_L}{1 + \sum_i a_{1i}Z_i + \sum_i \sum_{j, i \neq j} a_{2ij}Z_i Z_j + \dots + a_L Z_i Z_j \dots Z_L} \quad (14)$$

[00133] In general, each of the L number of load impedances contributes to the voltage distribution within the network. The contribution of fixed elements within the network is absorbed in the polynomial coefficients. The denominator is equivalent to the characteristic polynomial for the combined network (48, 216 and 218), and its coefficients (a 's) are fixed for the given network and depends on network 48 and 216.

[00134] In specific instances, where only the port's self-impedances are of significance, the entire n-port load network can be represented by n complex impedances. In this scenario, the Z-

parameter for the network would be a diagonal matrix with n diagonal terms. FIG. 18 describes an exemplary embodiment where the number of ports (n) is 3. For such a network, with three impedances (Z_1 , Z_2 and Z_3) on the distal side, the voltage measurements in the proximal side (e.g. V_1 , V_2 , V_3 , V_4) is given by:

$$V_i = \xi_0 \frac{b_0(i) + b_{11}(i)Z_1 + b_{12}(i)Z_2 + b_{13}(i)Z_3 + b_{212}(i)Z_1Z_2 + b_{223}(i)Z_2Z_3 + b_{231}(i)Z_3Z_1 + b_3(i)Z_1Z_2Z_3}{1 + a_{11}Z_1 + a_{12}Z_2 + a_{13}Z_3 + a_{212}Z_1Z_2 + a_{223}Z_2Z_3 + a_{231}Z_3Z_1 + a_3Z_1Z_2Z_3}, i = 1, 2, 3, 4 \quad (15)$$

[00135] Instead of the absolute measurements in the proximal end, one can also work on voltage ratios to avoid dependencies on the excitation voltage or, excitation current (ξ_0). Without loss of generality, the voltage across the reference resistor (V_1) is taken as reference and three ratios are constructed with respect to V_1 .

$$\frac{V_i}{V_1} = \frac{b_0(i) + b_{11}(i)Z_1 + b_{12}(i)Z_2 + b_{13}(i)Z_3 + b_{212}(i)Z_1Z_2 + b_{223}(i)Z_2Z_3 + b_{231}(i)Z_3Z_1 + b_3(i)Z_1Z_2Z_3}{b_0(1) + b_{11}(1)Z_1 + b_{12}(1)Z_2 + b_{13}(1)Z_3 + b_{212}(1)Z_1Z_2 + b_{223}(1)Z_2Z_3 + b_{231}(1)Z_3Z_1 + b_3(1)Z_1Z_2Z_3} \quad (16)$$

[00136] The properties of the measurement and the connecting networks are represented by the polynomial coefficients. For a network with n impedances and $(n+1)$ measurement entities, the number of independent polynomial coefficients would be $(n+1)*2n-1$. It may be noted that all the polynomial coefficients in Equation (16) can be scaled by the first term in the denominator, thereby reducing one unknown. The act of calibrating these networks would involve making proximal measurements with known impedances connected to the distal ports. The number of such independent measurements required would depend on the number of unknowns that need to be solved and the number of information per measurement. A fitter routine would then run on all of these measurement ratios, for known set of loads and estimate the polynomial coefficients.

[00137] Once the process of calibration is completed, and the polynomial coefficients are obtained, any arbitrary load connected across the distal ports in a similar configuration can be estimated. With an arbitrary load connected across the distal ports in a similar configuration, the proximal measurements are made and the ratios are computed with respect to the reference measurement. Next a fitter routine is invoked with the pre-determined polynomial coefficients and the ratios corresponding to the arbitrary load. The fitter routine may be initialized by the user with a starting value of the load impedances based on best guess. The fitter shall converge to a minimal residue on finding the optimal values for impedances which would match the ratio of measurements.

Convergence to alternate solutions are possible, however skilled persons in this art would be adept in avoiding such situations.

[00138] To estimate a generalized three port passive load network which can be modeled by six independent impedances, one would need to write the polynomial equations in Equation (14) with all six impedance present. Since the numbers of ratios measured are only three, the method needs to be extended for measurement of six impedances as discussed before. The method of calibration would involve making measurements with various combinations of load networks (comprised of all six impedances) for two independent interconnecting networks. The polynomial coefficients for both these networks would then be estimated using the individual sets of measured voltages and the knowledge of load impedances. Next, measurements would be made with arbitrary six impedance load networks, again with the same two independent interconnecting networks. The two sets of measurements along with the polynomial coefficients for both the networks would jointly be fitted by a fitter routine for estimating the six impedances. The method can similarly be extended to active networks where a nine impedance model needs to be estimated.

[00139] The above method, exemplified by a three port network with four proximal measurement entities can be easily extended to a general n -port network with $n+1$ proximal measurement entities on basis of Equation (14). The computation complexity grows exponentially with increasing number of load impedances in the network.

[00140] Thus the methods described herein can be extended to de-embed and evaluate a generalized n -port load network where there are $n + 1$ measurements performed concurrently.

[00141] The methods as described herein above are also depicted in the form of flowchart 286 of FIG. 20. The calibration technique for use in measurements from a remotely located multi port network, is shown by steps 288 to 300 of the flowchart, and includes a step 288 of providing an excitation and measurement entity for exciting the remotely located multi port network and for measuring a plurality of proximal voltages corresponding to the remotely located multi port network; a step 290 of providing a connecting network for connecting the excitation and measurement entity and the remotely located multi port network; a step 292 providing a plurality of known load networks coupled to the connecting network. The calibration technique further includes a step 294 for measuring a set of voltages corresponding to each load of the known load networks; and a step 296 for estimating electrical parameters corresponding to the measurement entity and the connecting

network by using a fitting utility across the set of voltages, where the electrical parameters are used for calibration. The method further includes a step 298 for providing a plurality of unknown load networks coupled to the connecting network and a step 300 for using the electrical parameters to de-embed the measurements from the remotely located multi port network.

[00142] The embodiments described herein have been illustrated through use of Z parameters as electrical parameters for modeling the electrical network. As would be appreciated by those skilled in the art, using the same principles, a similar formulation can also be made using Y parameters, S parameters, H parameters and G parameters since all models are equivalent ways of representing the electrical network. As such, it is to be understood that the embodiments described herein covers all such formulations.

[00143] Thus the technique described herein can be effectively used for determining actual voltages or voltage differences between the measuring electrodes or terminals of a remotely located multi port network.

[00144] The method as described herein above maybe incorporated as a tool that is used to determine the voltages or any other electrical response from a remotely located multi port network.

[00145] In a specific example, a system for de-embedding measured proximal voltages across conductors connected to at least three electrodes placed proximal to the region of interest is also disclosed. The system may include the embodiments of FIG. 11- FIG. 14 having an excitation and measurement entity for exciting the at least three electrodes and for measuring a plurality of proximal voltages corresponding to the at least three electrodes. The system also includes a connecting network in the form of two or more conductors for connecting the excitation and measurement entity and the at least three electrodes, where the at least three electrodes are at a distal end of the two or more conductors. A processor is added in the embodiments of FIG.11- FIG. 14 coupled to the excitation and measurement entities and the connecting network for estimating a plurality of electrical parameters as calibration parameters corresponding to the excitation and measurement entity and the connecting network, and for estimating actual voltages across the at least two pair of the at least three electrodes using the electrical parameters to de-embed the measured proximal voltages.

[00146] It would be appreciated by those skilled in the art that the embodiments described herein for example the embodiments of FIG. 11 and FIG. 12 (and the equivalent embodiments in other figures), pertain to compensating for the effects to both, the excitation and measurement entity 48 and the multi-port interconnection network 216. However, in some practical situations, it may be necessary to calibrate the effects of each of the entities separately, and during the process of de-embedding, the effects of both the entities will be combined. Further, the multi-port interconnection network 216 may include multiple parts or components. In this case, each part would be calibrated separately and the parameters can be combined together at the time of de-embedding. It is to be understood that this divided approach for calibration and de-embedding is also within the scope of the invention as described herein.

[00147] A method for measuring a plurality of actual voltage distributions across the ports of a remotely located multi port network, a tool and system incorporating the method are disclosed. The method includes a calibration and de-embedding step that involves providing an excitation and measurement entity for exciting the remotely located multi port network and for measuring a plurality of proximal voltages corresponding to the remotely located multi port network and providing a connecting network for connecting the excitation and measurement entity and the remotely located multi port network. A calibration method is used to provide a plurality of electrical parameters as calibration parameters corresponding to the measurement entity and the connecting network. The method includes exciting the remotely located multi port network with a known voltage and a known current; measuring proximal voltages across at least two pair of ports for the remotely located multiport network; and estimating actual voltage distribution across the at least two pair of ports using the electrical parameters to de-embed the proximal voltages.

[00148] Now turning to another aspect, the following description relates to generating multiple frequency excitation for the diagnostic element as mentioned earlier. There are some scenarios where it is important to determine the frequency dependent electrical properties of an electrically conducting medium. This medium may be composed of different types of substances that have different electrical properties. Analysis of these electrical properties may lead to inferences of other useful properties of the medium such as physical features, dimensions and composition. One such electrically conducting medium is biological tissue. In particular, this could be a region that consists of bodily fluid and other surrounding tissue. Different types of tissue have different

frequency dependent properties. In such an analysis, the region of interest is treated as an electrical network through which electrical current can traverse.

[00149] In order to obtain the electrical network equivalence of a region of interest, an electrical excitation (stimulus) is sent to the specific regions of interest and the responses are picked up at different pick up points and analyzed. Often, to completely unravel the anatomical characteristics of the region of interest, excitation of the specific regions of interest at multiple frequencies are provided and the responses are analyzed at each frequency. Prior arts in this area mostly focus on exciting the network with one frequency at a time and scanning through multiple frequencies within a small interval of time. Such techniques are common in Electrical Impedance Tomography. The responses at each frequency belonging to slightly different time instances are stitched together and analyzed. However, for applications in areas close to certain organs such as brain, lungs and heart the region of interest experiences an oscillatory change from one time instance to the next time instance. The approach of stitching data from multiple time instances fail as the network changes considerably with the internal organ movement. It is imperative that alternate methods that can perform broadband measurements in one time instance are needed to counter this.

[00150] One of the ways tried in prior art is to excite regions of interest at multiple frequencies is to create an excitation sequence by adding up multiple coherent sinusoids. One such method is described in the paper “Simultaneous multi-frequency bio-impedance measurement applying synchronised uniform or, non-uniform sampling”: A. Ronk et al, ICEBI 2007, IFMBE Proceedings 17. However there are issues with this approach. If the sinusoids are all added up in phase, then the peak to root mean square (rms) ratio (PAR) of the excitation signal is extremely high and increases as the number of constituent sinusoid increase. In fact, if N sinusoids are added in phase the PAR increases by a factor of square root of N . If the sinusoids are added up with random phase, the PAR would be anywhere between 1 to square root of N .

[00151] Typically, the peak value of the excitation is limited by the admissible peak currents in the region of interest where the excitation is applied, and the linearity and dynamic range of the electrical hardware associated with the excitation (transmitter) and the measurements (receiver). For a given peak value of the excitation signal, an increased PAR would limit the average energy (rms) of the excitation signal. This in turn would degrade the signal to noise ratio (SNR) of the received responses. Hence, it is important to maintain the PAR to close to its minimum value of unity.

[00152] As can be seen from the discussion herein, the focus has been to provide multiple frequency excitation to regions of interest at various time points, and subsequently stitch the data together for analysis, which suffers from the drawback of getting data at different states of heart beats. Attempting to excite regions of interest with multiple frequencies simultaneously using a sequence of multiple sinusoids may affect the PAR and provides signal with poor quality. Thus, there is a need to provide a method and system to excite regions of interest at multiple excitation frequencies simultaneously while still being able to obtain signals having good PAR.

[00153] The techniques described hereinbelow involve broad band excitation of a slow time varying electrical network for example the network involving a cardiac region where the excitation is done using a signal that simultaneously presents multiple frequencies. More specifically, two exemplary non-limiting efficient ways of multi-frequency excitations are described, the first one using a pseudo random sequence and the second using a method using orthogonal frequency division multiplexed (OFDM) sequence.

[00154] As noted herein, in one aspect the invention provides a method for exciting at least a set of electrodes configured to be placed in vivo proximal to a volume of interest in a vasculature. The description herein below relates to an estimation method of a broad band frequency response of any slow time varying electrical network by exciting it with a broad band electrical excitation and analysing its electrical response. The broad band excitation is done using a signal that simultaneously presents multiple frequencies. FIG. 21 shows the method of the invention, depicted by numeral 310, in the form of a flowchart having exemplary steps. The method comprises generating a multiple frequency sequence pulse having a predetermined peak to root-to-mean-square (rms) ratio that is close to unity, as shown in FIG. 21 and represented by numeral 312.

[00155] In one embodiment, an excitation with multiple frequencies and a good PAR i.e. PAR close to unity is constructed by generating a pseudo random sequence. Without being bound to any theory, it is known that a pseudo random sequence of length L and generated at a sampling of f_s would contain discrete un-aliased tones of frequency from 0 (which corresponds to a DC frequency) to $f_s/2$, in steps of f_s/L . The power at each frequency (except DC) is equi-distributed while the phase of the individual tones is uniformly spread over $-\pi$ to $+\pi$.

[00156] In practice, one of the ways of achieving the excitation would be using a digital-to-analog converter (generally abbreviated in the art as D/A or DAC) converter with low noise. D/A

having the above stated requirements are known in the art, and can be effectively used in the invention. The D/A sampling rate needs to be at least double the required maximum frequency of excitation. The basic shape of the D/A converter output is a rectangular pulse of width equal to the time difference between two consecutive samples. It would be understood by those skilled in the art that if the D/A converter that outputs a pseudo random sequence is sampled at a frequency (f_s) that is at least twice the desired maximum frequency of measurement (f_H), it would create a frequency shape that is the product of the frequency shape of the basic pseudo random sequence and the frequency shape of the rectangular pulse (i.e. a Sinc function with the first null at f_s).

[00157] A big advantage of an excitation based on pseudo random sequence with a basic rectangular shape is that its PAR is unity. This leads to maximising the rms signal power for a given peak amplitude of the signal. There are further advantages on the electrical hardware. The output of the D/A converter in this implementation has only two levels ($-A$ and A), where A is the amplitude of excitation. The linearity of the transmit chain is irrelevant since non-linearity only produces a gain error and offset error to the signal. The receive chain design is also simplified with a lower PAR since dynamic range and linearity requirements are less demanding. Another major advantage of such an excitation based on rectangular pulse shapes (of duration $t_s = 1/f_s$) is that the D/A can be excited with a single bit excitation, minimizing the digital noise associated with toggling multiple bits simultaneously. A minor fall back of the rectangular pulse shape based approach is the small droop at higher frequencies of interest due to the roll off of Sinc response (up to about 4 dB at $f_s/2$) which results in proportionate drop in SNR of the information for channel estimation. However this drop in SNR for channel estimation does not impact system performance. In alternate implementations, it may be possible to make the basic pulse shape as close to a Delta function, in which case, the frequency characteristics would be flat across frequency. However, this comes with an increased PAR. The D/A converter output needs to be filtered effectively to prevent out of band emissions outside the band of interest. The filtering may be accomplished using a passive or, an active analog filter with pass band at the region of interest. It would be understood by those skilled in the art that filtering would result in a small increase in PAR which is however not significant and PAR would still remain close to unity.

[00158] In another embodiment, the excitation sequence is constructed as a repetitive orthogonal frequency division multiplexed (OFDM) sequence. The OFDM sequence consists of equal amplitude of all frequencies starting from a low frequency of interest to a high frequency of

interest. The number of frequencies excited is proportional to the ratio of the high frequency (fH) to the low frequency (fL), while the spacing between frequencies is same as the lowest frequency (fL) of interest that is chosen. The duration of the basic OFDM sequence is inversely related to its lowest frequency. The PAR of the OFDM sequence can be made to a low value close to unity by suitable choice of phase for each frequency. An OFDM based sequence is a sum of several discrete tones whose number is a power of 2, and provides distinct advantage of implementing the processing circuitry in an efficient manner based on Fast Fourier Transform (FFT).

[00159] In yet another embodiment, the excitation sequence can be constructed as additions of multiple coherent sinusoids with a method that would minimize the overall PAR of the sequence. PAR minimization can be achieved by suitably adjusting the phase of each sinusoid. Such sequences can also be constructed by appropriately dropping out one or, more tones from the OFDM sequence. These sequences are particularly useful over a full-fledged OFDM sequence where the electrical hardware may not handle a large set of frequency information due to its limited capacity or, the non-linearity of the electrical hardware is too high and dictates the use of tones that have non-multiplicative relationship with each other, so that the non-linear effect of one or, more tones do not impact another tone.

[00160] It will be appreciated in the art that the admissible rms current into the body is a function of frequency for a single frequency excitation (FIG. 22). The admissible current levels are at a minimum of 10uA and go up linearly with the frequency beyond 1 KHz. Prior arts don't describe admissible current levels for multi-frequency excitations. FIG. 22 shows a graphical representation 316 of exemplary current values 318 that may be provided to a heart over a range of frequencies 320. For example, maximum permissible current through a heart (in μA) may vary over the range of frequencies. The maximum permissible current through a heart may also vary depending on whether the current is applied in an abnormal non-continuous manner, abnormal continuous manner, or normal continuous manner as shown at reference numerals 322, 324, 326. In some embodiments, it may be preferable to apply a current in a normal continuous manner in a range around a 100 KHz frequency. In some embodiments, the current may be applied between about 40 KHz (fL) and 10 MHz (fH). A safe zone may be used as indicated by reference numeral 328. One possible way of determining the value of rms current for an excitation based on pseudo-random sequence can be by matching the rms current of the composite signal to the corresponding admissible rms current for the lowest frequency.

[00161] The exemplary method then comprises applying the multiple frequency sequence pulse to excite across the set of electrodes of the diagnostic device. Turning back to FIG. 21, the step of applying the multiple frequency pulse is depicted by numeral 314. The excited set of electrodes then sends a pulse of electric current across the region of interest. Depending on the electrical properties of the region of interest, a voltage is developed across the electrodes. There will be one voltage corresponding to each excitation frequency from the multiple frequency pulse. Hence, one skilled in the art will realize the vast amount of information that can be simultaneously obtained using the method of the invention.

[00162] The set of voltages at multiple frequencies developed across the electrodes may then be detected using an appropriate measurement device that is capable of handling all of these signals simultaneously. Different tissues (or non-tissues found in the body) have different signature in voltage and current relationships as the frequency of excitation are varied. This may be explained in terms of vessel, blood, and fat herein for ease of description, however, these examples are not meant to limit the invention thereto, as other tissues and non-tissues have different signatures as well, and are contemplated herein. The measurement device may be configured to process the multiple sets of information sequentially, in parallel, or in groups to provide results. One exemplary method comprises of extracting the frequency domain information from the set of voltages using either a multi-bank filter set at appropriate frequencies or, using a Fourier Transform, building a set of equations based on the voltages at each frequency across the electrodes when the current is passed from at least a first electrode set into the bodily fluid, tissue and/or non-tissue, and received at second electrode set, solving these equations to determine the unknowns in such equations which can be transformed into dimensions of the given anatomical feature or features. In one example the equations are solved simultaneously. The method of the invention works on exploiting these diversities to solve out the dimensions of the region of interests that could uniquely yield the response observed the electrodes. There are several methods used to perform this decomposition or transformation. One method is to create an electrically equivalent 3-dimensional model of the near-field using lumped finite-elements and mathematically invert the unit elements to allow the model to recreate the observed set of voltages, and thereby calculating the dimensions of the region of interests. A network may be provided with finite element unit characteristics of the subject's tissue embedded. This may accommodate different types of tissues. For example, the tissue may include blood and fat. An electrical network with a first element (e.g., blood) associated by a first dimension

may be provided. An electrical network with a second element (e.g., fat) associated by a second dimension may be provided. The various dimensions (e.g., first and second dimension) may be altered until the desired response is achieved. This may be provided for any number of dimensions corresponding to various tissue types (e.g., one, two, three, four, or more dimensions). When the desired response is achieved, the dimension values used to achieve the desired response may be the dimensions of the tissue. For example, various tissues (e.g., blood, fat, aorta) may each have a corresponding dimension. The dimension value may be 0 or greater. The various dimension values may be varied, and calculations may be performed, to yield a result.

[00163] De-embedding may include taking into consideration material properties of the devices, or device components, such as the wires or electrodes. For example, an electrode may be at a distal end of a wire at the region of interest, and electronics to receive and process the signals may be provided at a proximal end of a wire. An electrical measurement taken by the distal electrode(s) is received by the electronics. However, a signal provided at one end of the wire may be altered by the time it reaches the other end of the wire due to material properties of the wire. This variation may be taken into account by using appropriate models based on the material characteristics, length of the wire, and other variables relevant to this situation, or performing measurements with known electrical loads at distal end and calibrating the effect of the in between electrical conductors.

[00164] The exemplary method, as noted herein, provides for the capability of making multiple frequency measurements at the same time. Further, all the measurements are made in the same phase of heart beat, such as in the systolic phase or diastolic phase. This overcomes the difficulty associated with overlaying multiple measurements made at different times to account for the phases of the heart. Some exemplary measurements made using the method described herein include, for example, but not limited to, dimension, nature of a specific region of interest like fat, stenosis, block, artery, bodily fluid pressure, bodily fluid flow rate, tissue characteristics, and the like, and combinations thereof.

[00165] The exemplary method may be administered effectively as a tool in the form of a software program product. Thus, in another aspect, the invention provides a tool that uses the method of the invention. The software may comprise algorithm steps to generate multiple frequency pulse as delineated herein. The software may then be configured to excite the set of electrodes with the multiple frequency pulse. Subsequently, the software may further be configured to receive the

multiple signals from the region of interest to be processed. Accordingly, the computing requirements for the effective execution of the software will become obvious to one skilled in the art. Further, the other components required for the functioning of the tool will also be obvious to one of ordinary skill in the art, and may include, for example, a display module such as a monitor having a suitable resolution, an input module like a keyboard and a mouse, and so on.

[00166] In yet another aspect, the invention provides a system that comprises the tool of the invention, which in turn is based on the exemplary method. The system is shown in block diagrammatic representation in FIG. 23, and depicted generally by numeral 330. The system comprises at least a set of electrodes 332 configured to be placed proximal to the region of interest. The set of electrodes is capable of being excited by a multiple excitation pulse. The multiple excitation pulse is made possible using pseudo random generator that involves using a suitable number of flipflops 334. The number of flipflops required depends on the complexity of the pulse to be generated, among other factors. The exact sequence to be executed by the pseudo random generator may be obtained using an input module 336. The input module may be configured to take manual inputs, or may be configured to automatically generate a sequence for the pseudo random generator to execute. As mentioned herein above, instead of a pseudo random sequence, an OFDM sequence may also be used with the associated electronics for generation of the OFDM sequence as would be known to one skilled in this art.

[00167] The multiple excitation pulse generated is then sent through a D/A converter 338. The system further comprises a filter 340. The filter may be a passive or an active filter, depending on various factors, such as, the necessity, the requirement of the situation, computing abilities, cost, and so on, and combinations thereof. In some specific applications, the requirement of the filter may not be necessary and the filter may not be a part of the system. In one specific embodiment, the filter comprises a passive multi-stage LC ladder network.

[00168] The system further comprises a processing device 342 to process the input for a pseudo random generator. The processing device may also be configured to send the multiple excitation pulse to the set of electrodes. The system may also comprise a communicating device (not shown in FIG. 23) to communicate the pseudo random generator to the set of electrodes. The communication between different components and modules may be achieved through any wired or

wireless means known to those skilled in the art, and the exact requirement may be arrived at without undue experimentation.

[00169] The system also comprises a detector module 344 to detect the voltages developed across the electrodes. The detected signals may then be fed into the processing device for further processing. The signals may give rise to a wealth of information related to the region of interest, which the processing device is configured to unravel based on inputs such as, but not limited to, the signal, the algorithm, the region of interest characteristics, and the like. Thus, the system of the invention may be used to make multiple simultaneous frequency measurements, without having to resort to stitching of data acquired at different time points which may introduce errors into the final measurement.

EXAMPLE

[00170] In a specific implementation, the excitation frequency band is chosen between 40KHz (fL) to 10MHz (fH) based on the electrical characteristics of bodily fluid, tissue and fats. A 16 bit D/A converter was chosen to operate at a sampling rate of f_s (= 20MHz). The chosen D/A converter accept offset binary sequence (0x0000 for the lowest value and 0xFFFF for the highest value). The Most Significant Byte of the converter is toggled according to the single bit pseudo random pattern, while the next bit was kept permanently at logic 1. All other bits were kept at logic 0. Hence the DAC inputs toggles between 0x4000 and 0xC000, depending on a 0 or a 1 from the pseudo random generator. The pseudo random generator resides on a back end entity and is comprised of a chain of 9 D-flipflops referred to as flops, to represent a 9-tap pseudo random sequence. The resultant sequence is a maximal length pseudo random sequence with length of $L = 511$ ($2^9 - 1$). The generator polynomial used to generate the sequence is

$$X^9 + X^4 + 1 = 0 \quad (17)$$

which would mean that the input of the last tap is an xor-ed output of the first and the fifth flops, as shown in FIG. 24. The flop outputs are all initialized to 1's to begin with (Reset condition). The tones present in the excitation sequence are multiples of fl, wherein:

$$fl = f_s/L = 20/511 \text{ MHz} = 39.14 \text{ KHz} \quad (18)$$

[00171] The D/A converter produced an output with frequencies spaced at 39.14 KHz. The output is passed through a bandpass filter whose pass band starts at a value lower than 39.14 KHz and ends above 10 MHz ensuring decent flatness over the entire band. In the specific implementation, the filter is designed using a passive multi-stage LC ladder network. Since the minimum frequency of the final composite signal is at 39.14 KHz, the signal rms value is maintained to be lower than 391 μ A. The choice of the sampling frequency and the tap length depends on the minimum and maximum frequencies of operation. As described before, the sampling frequency is at least twice the maximum desired frequency in the excitation, while the tap-length (L) is the nearest integer satisfying the relationship

$$L = [\log_2(f_s/f_{\min})] \quad (19)$$

[00172] FIG. 25a shows the time domain waveform of the 9-tap pseudo random binary sequence generated as described herein. The current waveform has amplitude and rms value of 391 μ a. FIG. 25b shows a zoomed in portion of the exemplary pseudo random binary sequence in time domain.

[00173] FIG. 26 shows the power spectral density of the same 9-tap pseudo random binary sequence generated.

[00174] FIG. 27 shows the plot between phase angle and frequency for the 9-tap pseudo random binary sequence.

[00175] In yet another implementation, as shown in FIG. 28 an OFDM sequence is constructed using N_{freq} (=256) discrete tones of equal amplitudes and each being at a random phase. The phase angles for each tone are adjusted so as to obtain the minimum PAR close to unity. The construction of the OFDM sequence can be done either simply by adding all the discrete tones together or, by performing a IFFT (Inverse Fast Fourier Transform) of a symmetric sequence of $2N_{\text{freq}}$ (=512) complex numbers, where the first 256 complex numbers relate to the amplitude and phase of the individual tones and the next set of 256 complex numbers are simply the complex conjugate of the first 256 arranged in the reverse order (FIG. 28). The resultant time domain signal is sampled at f_s (= 20 MHz) which is twice the largest frequency of interest (f_H). The lowest frequency in this sequence is f_L (= $f_s/512$ = 39.0625 KHz). The time domain OFDM sequence can also be produced at higher sampling rates using appropriate size of IFFT inputs. FIG. 29 shows such

an OFDM time domain sequence that has been sampled at 80 MHz where the IFFT is performed on 2048 inputs. The inputs corresponding to frequencies higher than 10 MHz is zeroed. A higher sampling rate eases the requirement on anti-aliased filtering while increasing the complexity of the hardware in the transmit side. FIG. 30 shows an exemplary OFDM frequency response for the implementation of FIG. 28.

[00176] In yet another embodiment as shown in FIG. 31, a customised sequence is created using multiple coherent sinusoids added with appropriate phase angles so as to minimize the PAR. The resultant sequence may bear the property where any given frequency is not harmonically related to any other frequency. The same can also be constructed in the OFDM framework described above, where one or, more IFFT inputs are nulled to remove a set of tones from the original sequence.

[00177] Thus to summarize the multiple frequency excitation, in one aspect, an exemplary method is provided for exciting at least a set of electrodes configured to be placed in vivo proximal to a region of interest. The method comprises generating a multiple frequency sequence pulse having a predetermined peak to root-to-mean-square (rms) ratio close to unity, using multiple methods; and applying the multiple frequency sequence pulse to excite the at least the set of electrodes.

[00178] It should be understood from the foregoing that, while particular implementations have been illustrated and described, various modifications can be made thereto and are contemplated herein. It is also not intended that the invention be limited by the specific examples provided within the specification. While the invention has been described with reference to the aforementioned specification, the descriptions and illustrations of the preferable embodiments herein are not meant to be construed in a limiting sense. Furthermore, it shall be understood that all aspects of the invention are not limited to the specific depictions, configurations or relative proportions set forth herein which depend upon a variety of conditions and variables. Various modifications in form and detail of the embodiments of the invention will be apparent to a person skilled in the art. It is therefore contemplated that the invention shall also cover any such modifications, variations and equivalents.

We Claim:

1. A diagnostic device for measuring a non vasculature parameter, the diagnostic device comprising:

a diagnostic element comprising at least two spaced apart sets of electrodes configured to be placed proximal to a region of interest; and

an excitation and measurement device electrically connected to the diagnostic element,

wherein at least a first set of electrodes from the at least two spaced apart sets of electrodes is configured to receive an input current from the excitation and measurement device, and at least a second set of electrodes from the at least two spaced apart sets of electrodes is configured to receive a plurality of voltage signals from the region of interest and transmit the plurality of voltage signals to the excitation and measurement device, wherein the excitation and measurement device calculates a voltage difference between the at least second set of electrodes based on the voltage signals, wherein the voltage difference is a function of a non vasculature parameter.

2. The diagnostic device of claim 1 wherein the voltage difference is based at least on a spatial diversity of the at least two spaced apart sets of electrodes, first set of electrodes and second set of electrodes, or based on frequency diversity of the input current and the plurality of voltage signals.

3. The diagnostic device of claim 1 wherein the voltage difference is based at least on a tissue diversity of the region of interest, or a fluid characteristic of the region of interest or combination thereof.

4. The diagnostic device of claim 1 wherein the non vasculature parameter is selected from a group consisting of a fluid index for a thoracic fluid, an instantaneous volume for chambers of a heart over a cardiac cycle, urine volume in a bladder, or level of cerebrospinal fluid in a brain.

5. The diagnostic device of claim 1 wherein the diagnostic element is adapted at least as a body mountable strap or a lead for an implantable device, or a mount on an external device capable to be in contact with a patient body.

6. The diagnostic device of claim 1 further comprises a display device to display the non vasculature parameter.
7. A heart condition monitoring device comprising the diagnostic device of claim 1.
8. A CRT device comprising the diagnostic device of claim 1.
9. A bladder monitoring device comprising the diagnostic device of claim 1.
10. A cranial pressure monitoring device comprising the diagnostic device of claim 1.

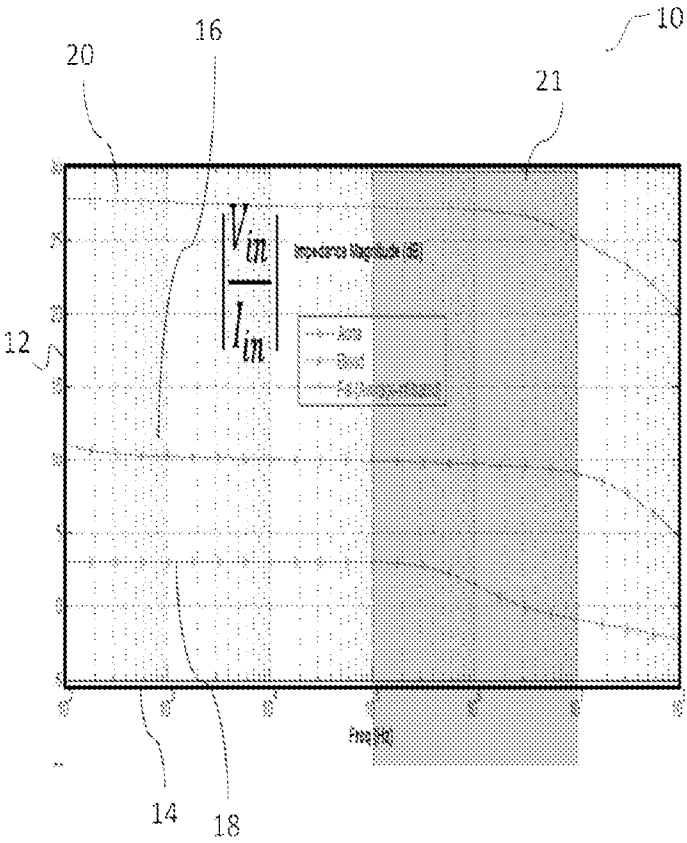


FIG. 1

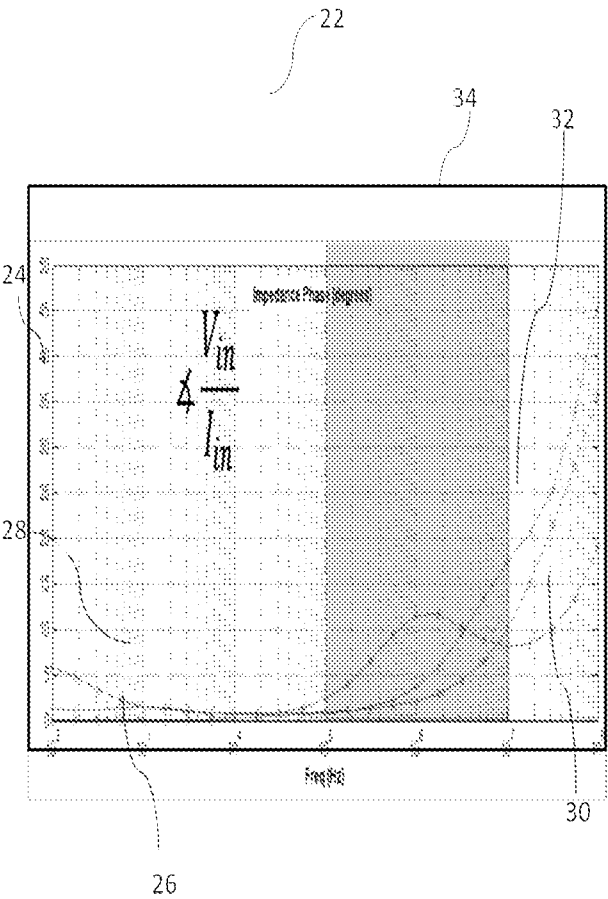


FIG. 2

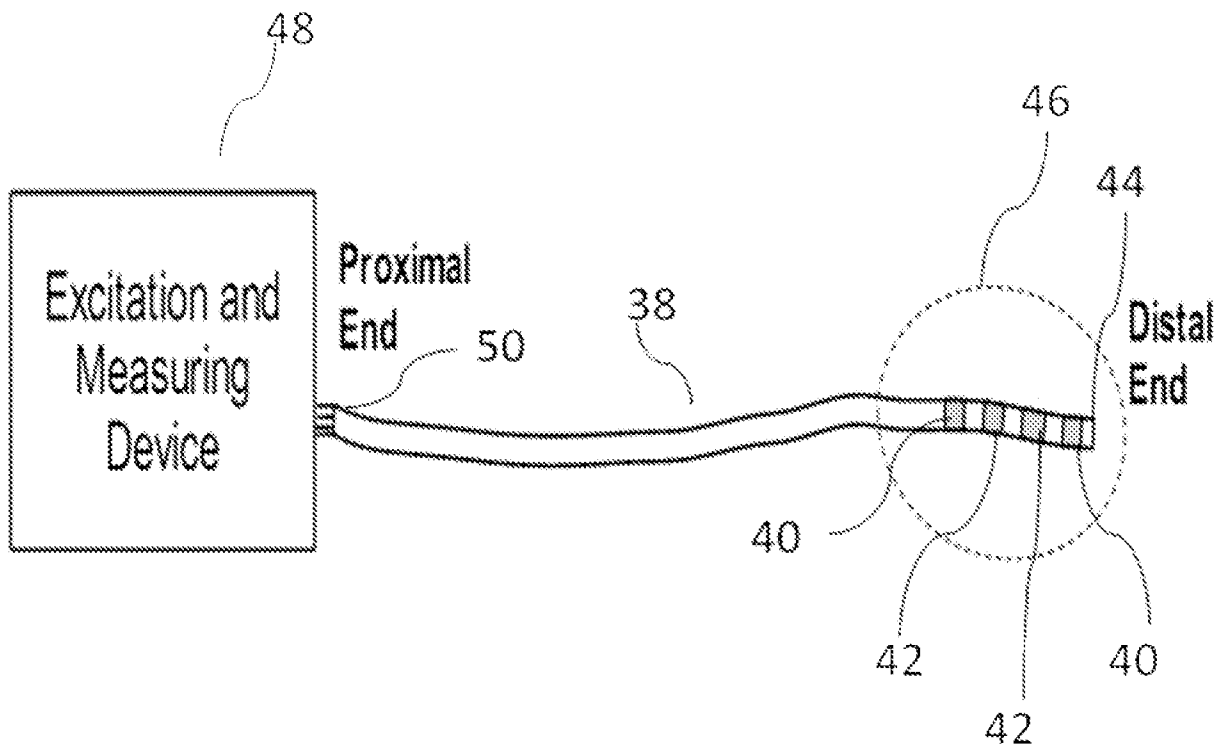


FIG. 3

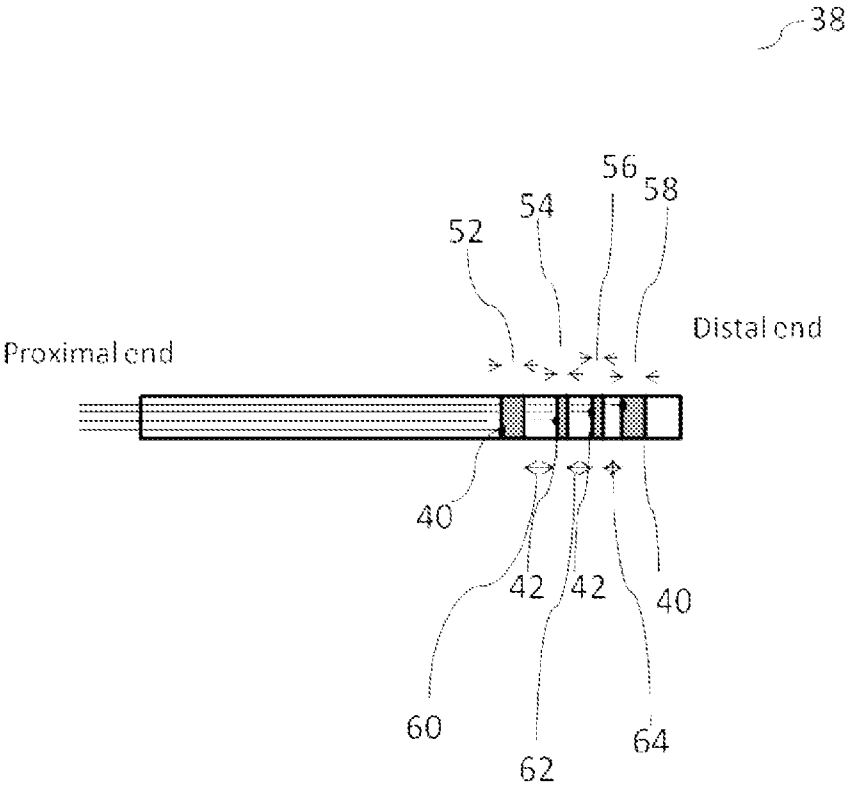


FIG. 4

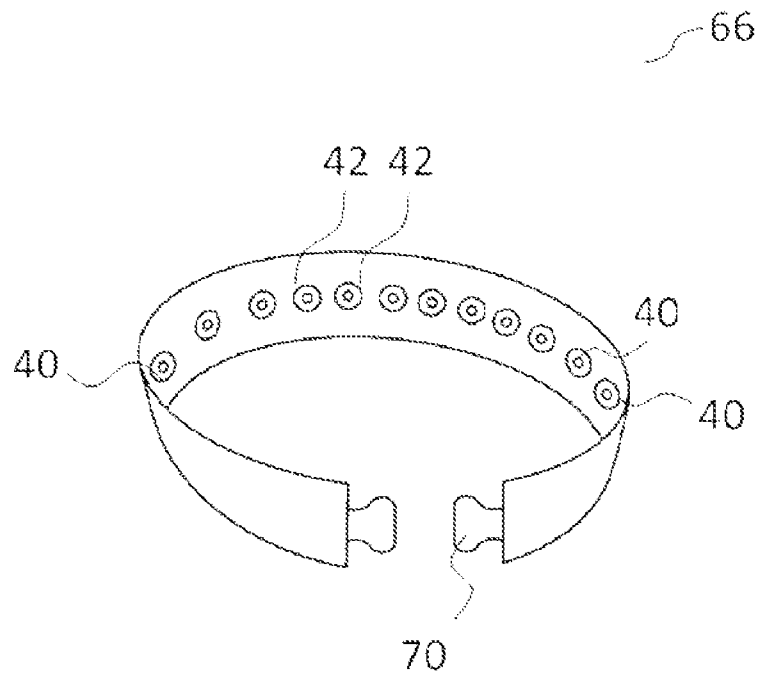


FIG. 5

72

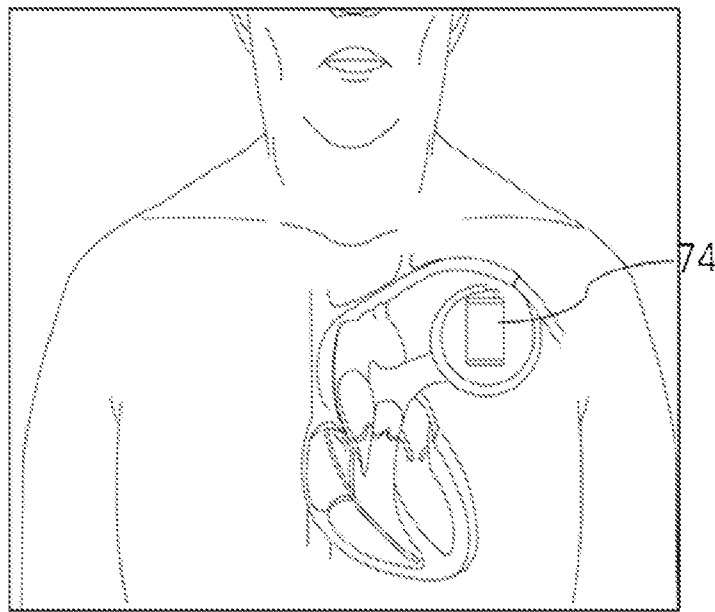


FIG. 6

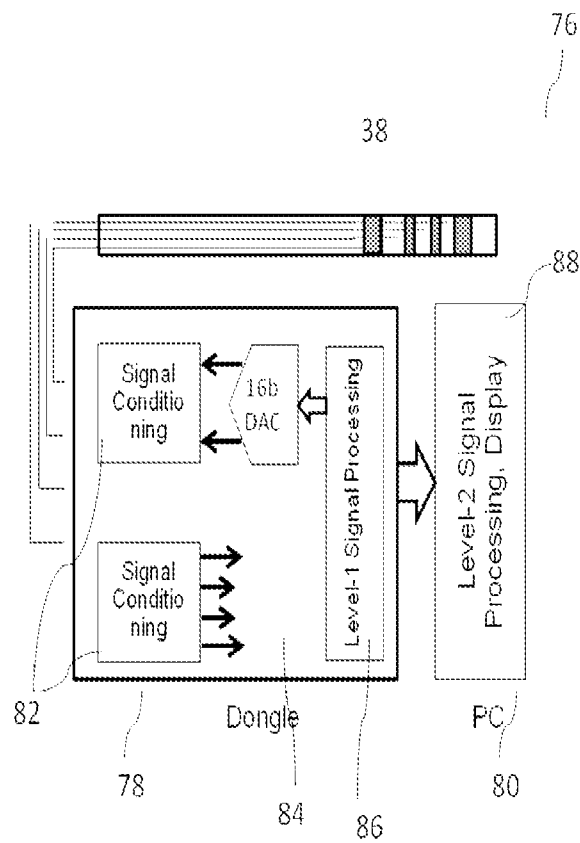


FIG. 7

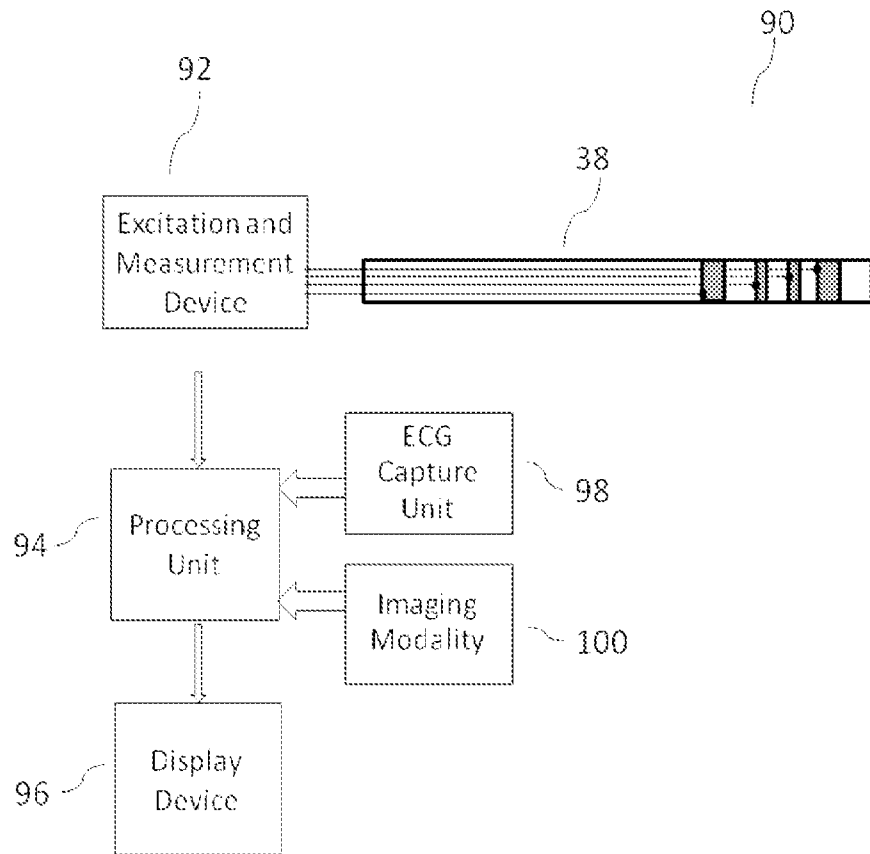


FIG. 8

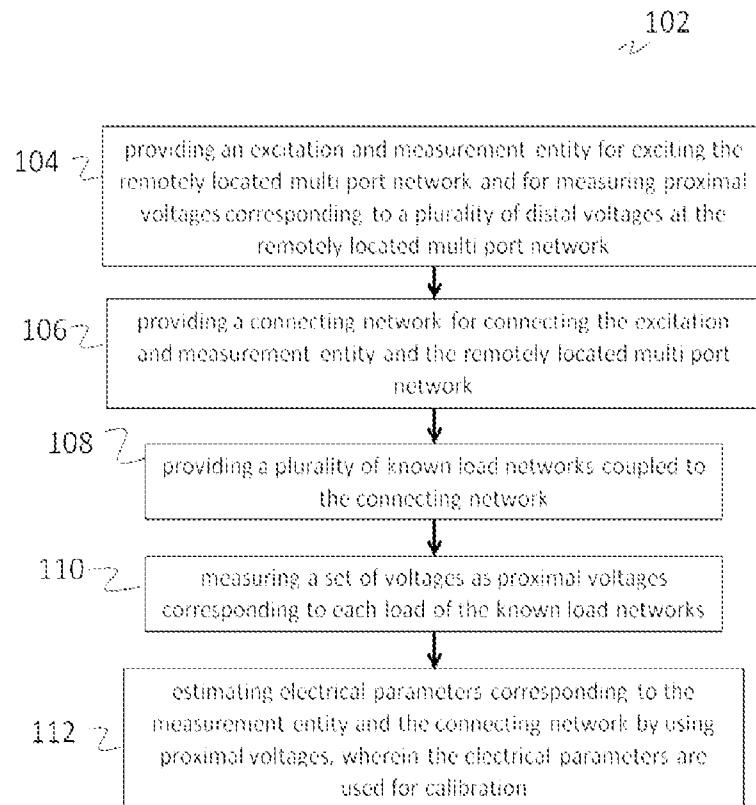


FIG. 9

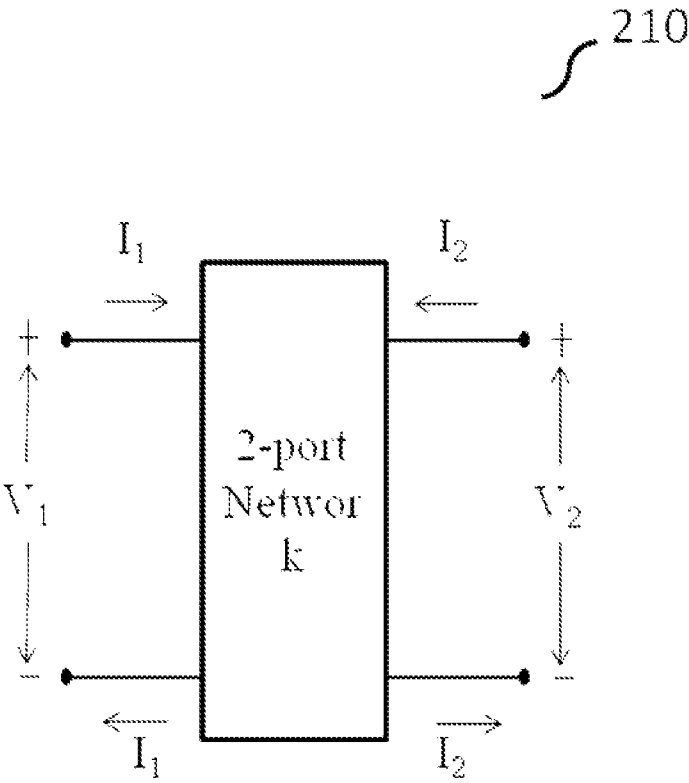
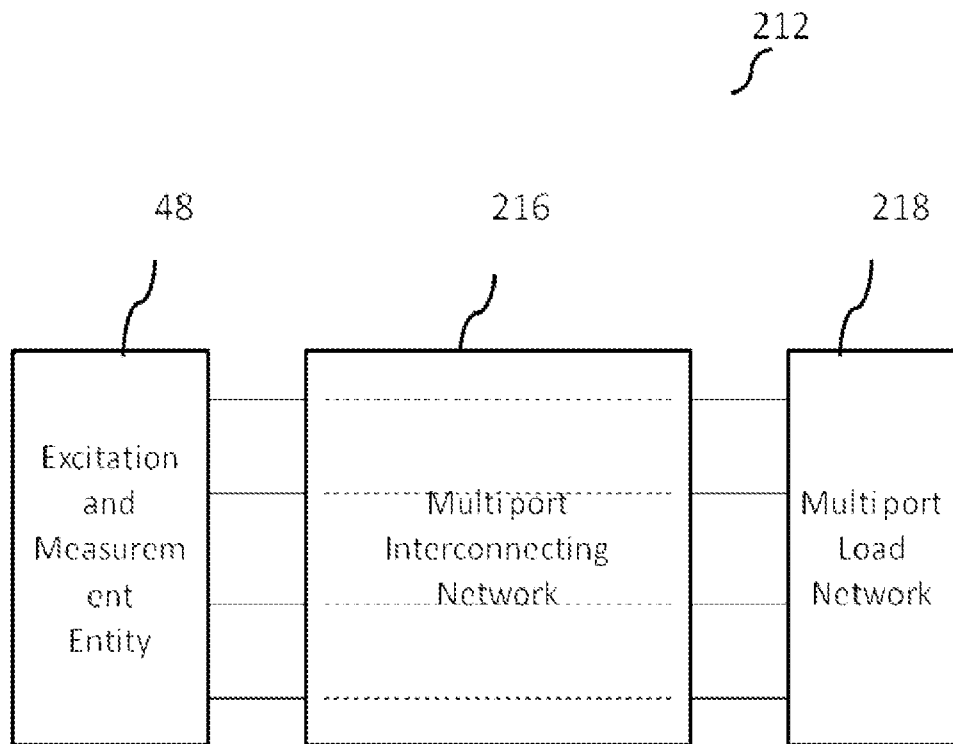


FIG. 10

**FIG. 11**

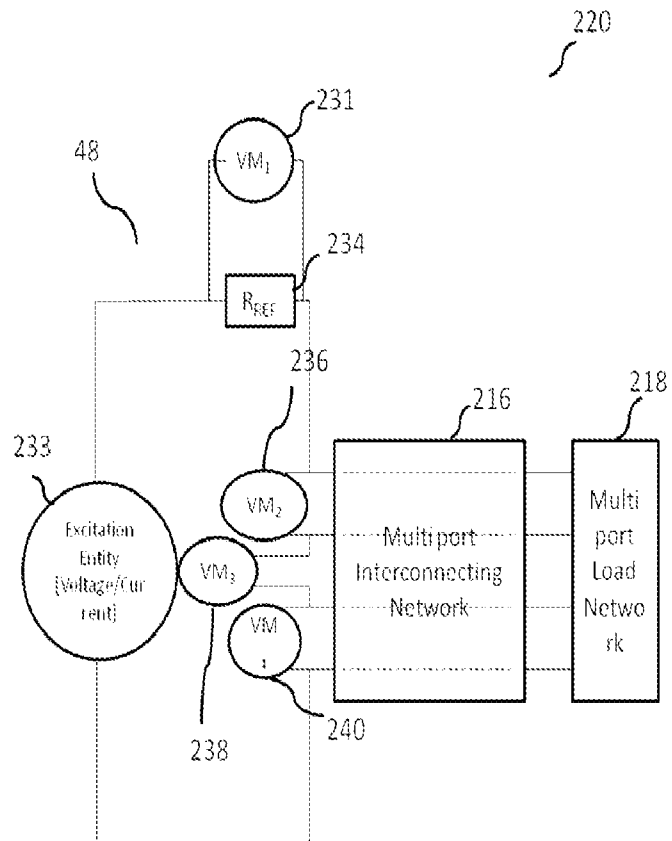


FIG. 12

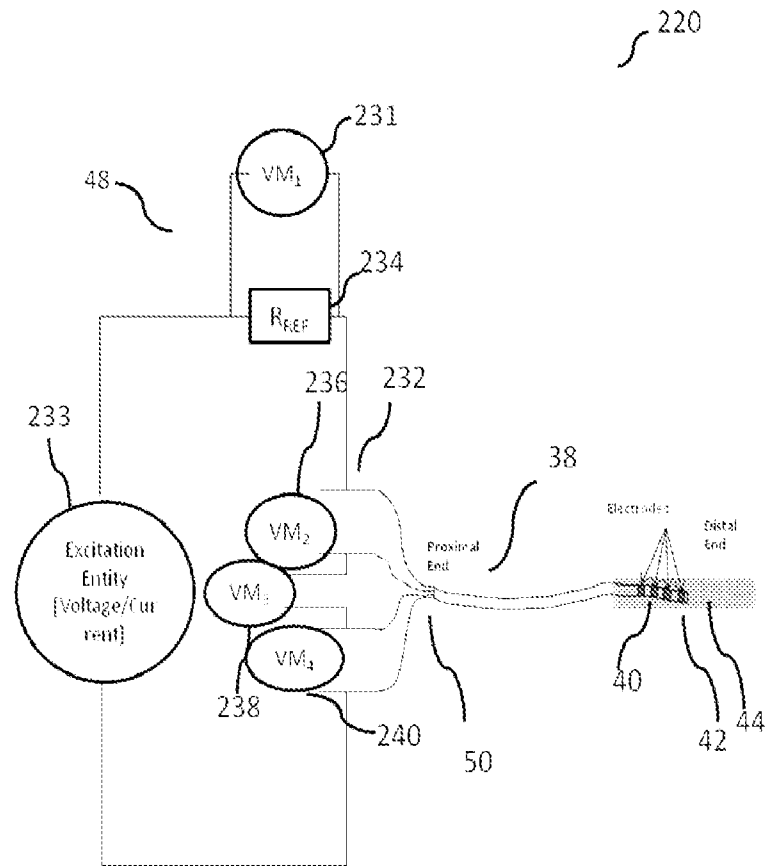


FIG. 13

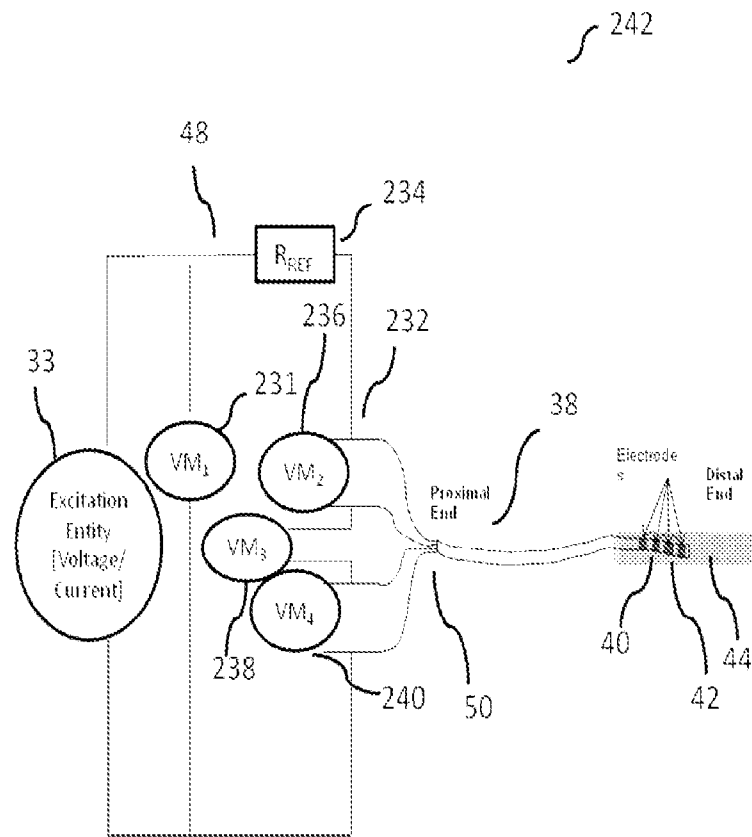


FIG. 14

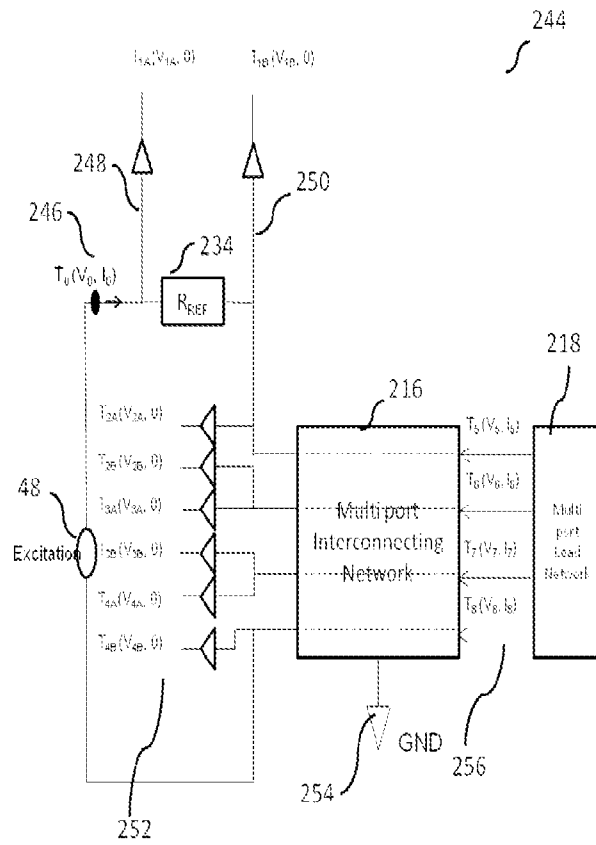


FIG. 15

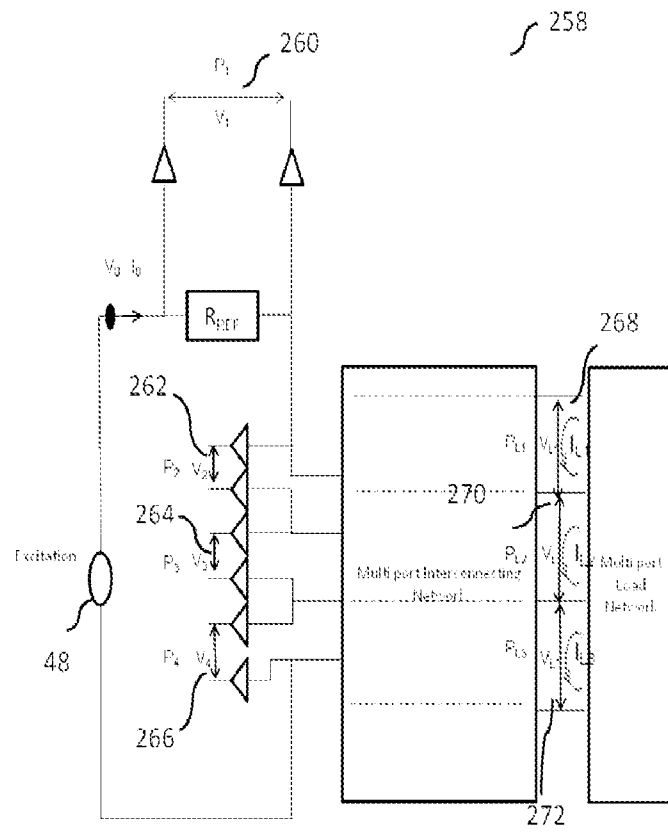


FIG. 16

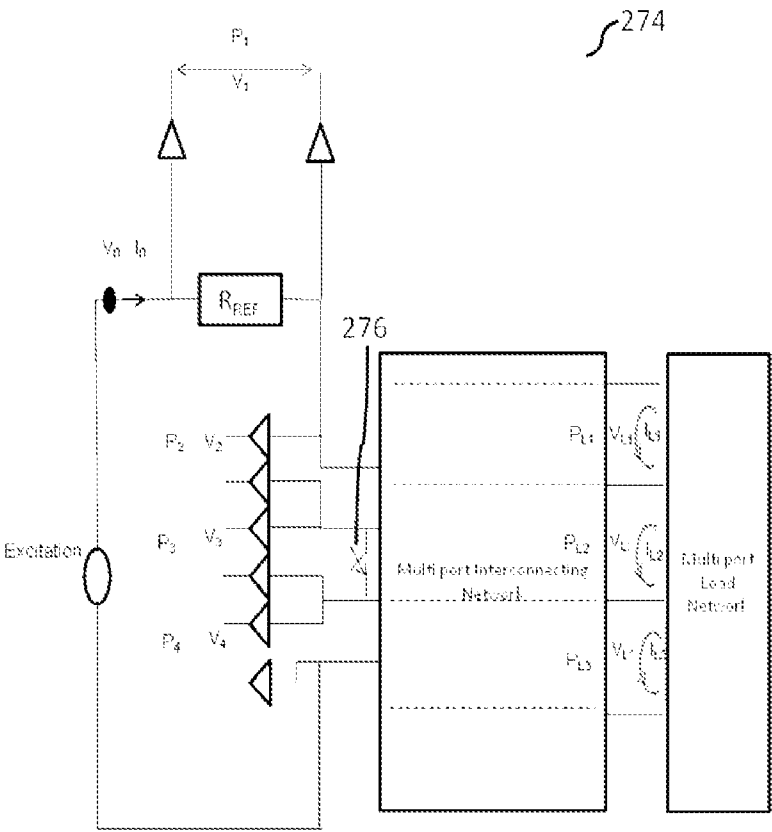


FIG. 17

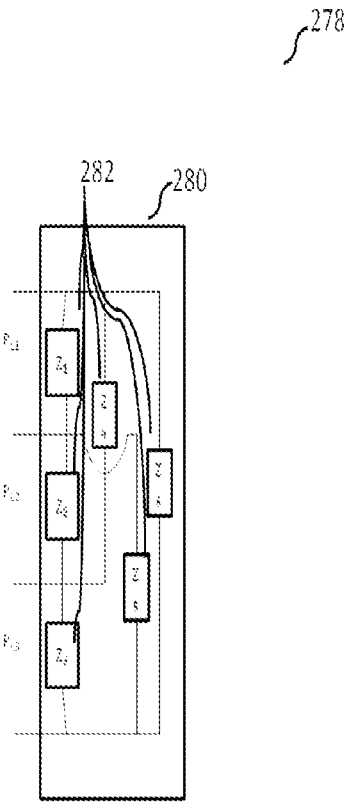


FIG. 18

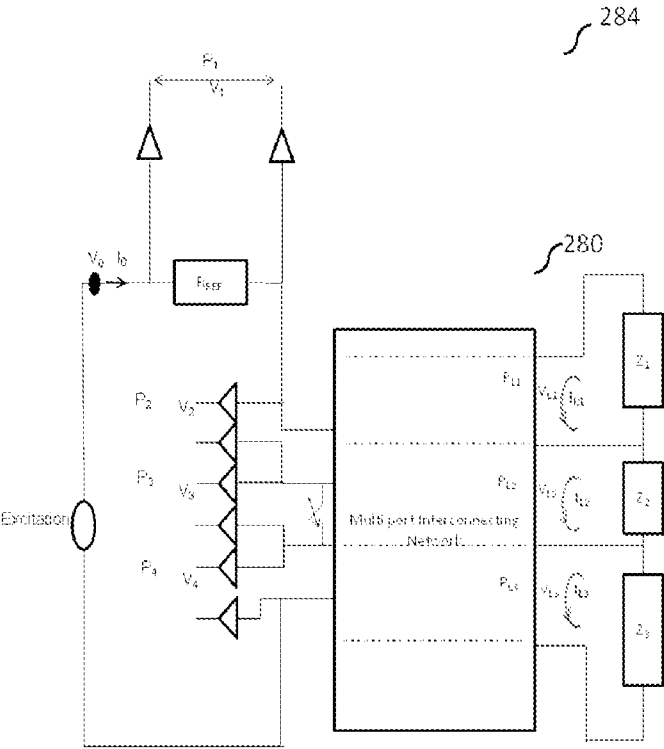


FIG. 19

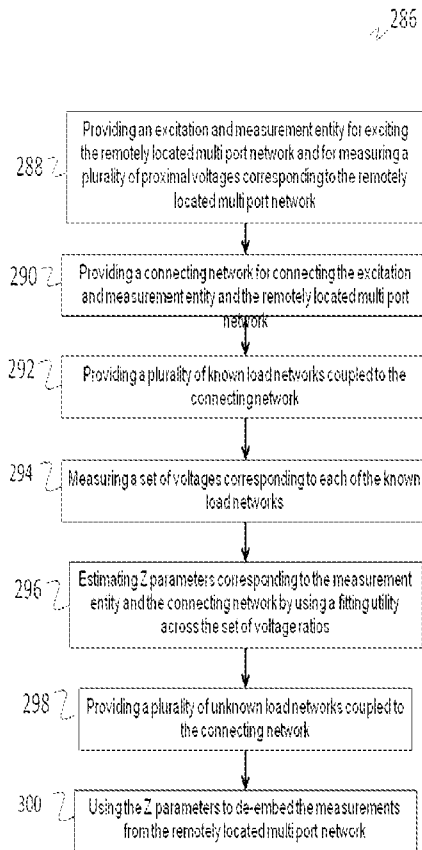


FIG. 20

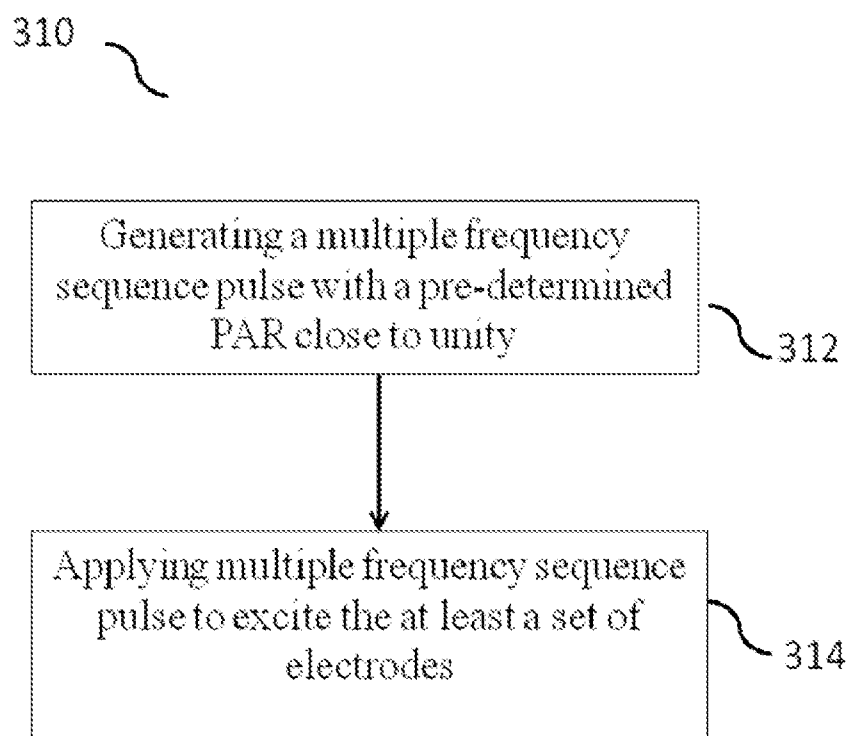


FIG. 21

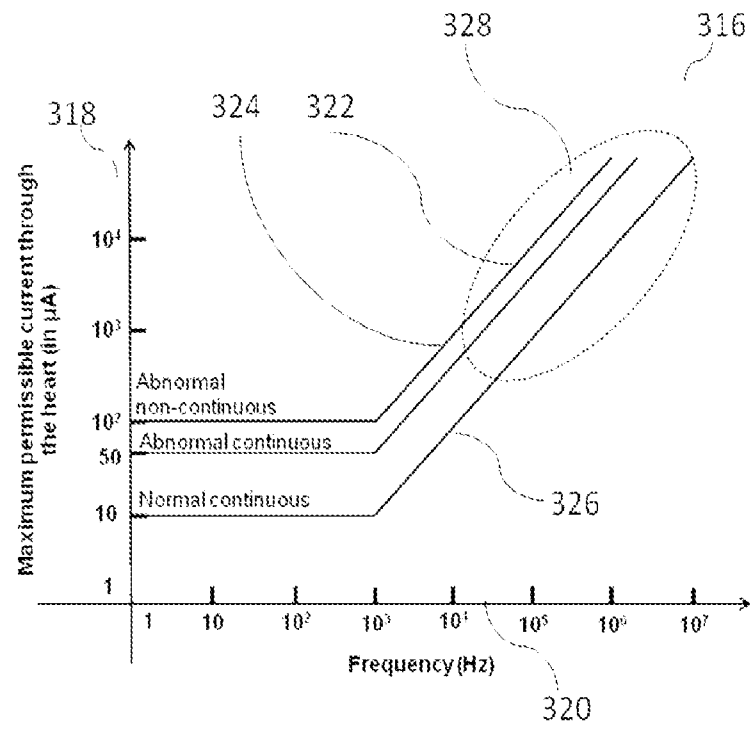


FIG. 22

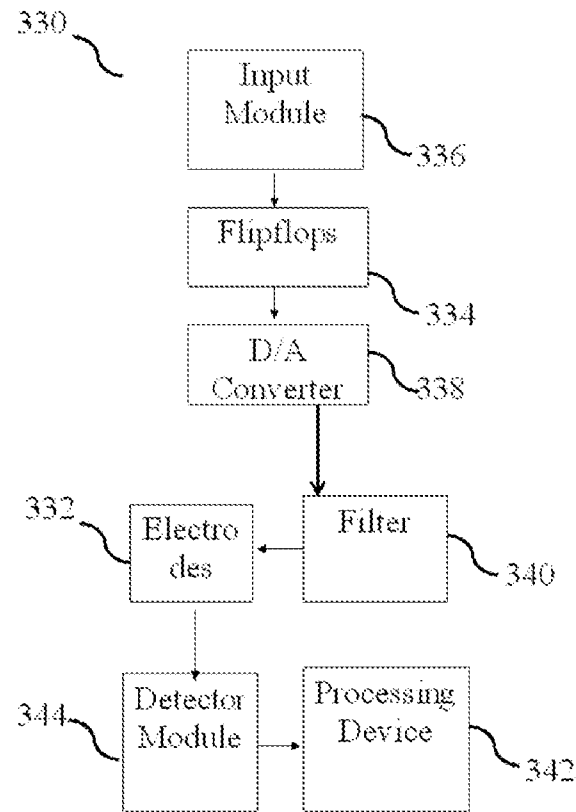


FIG. 23

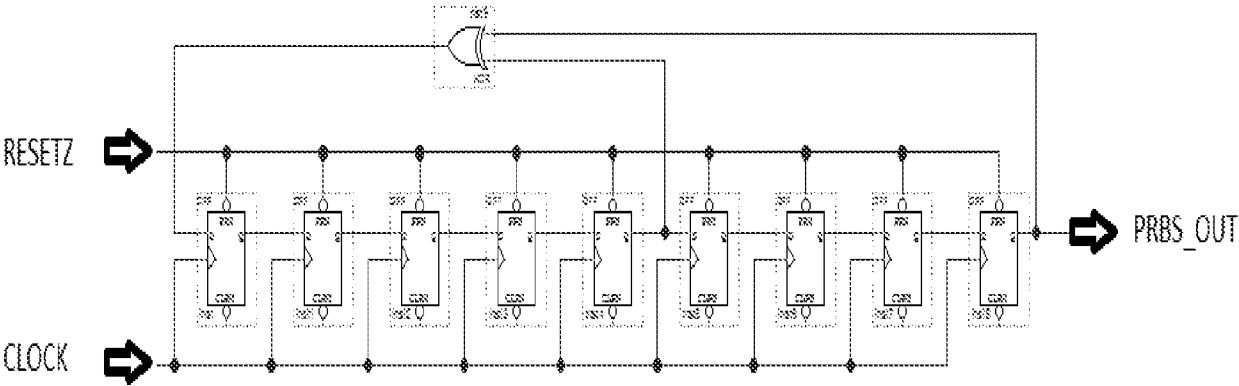
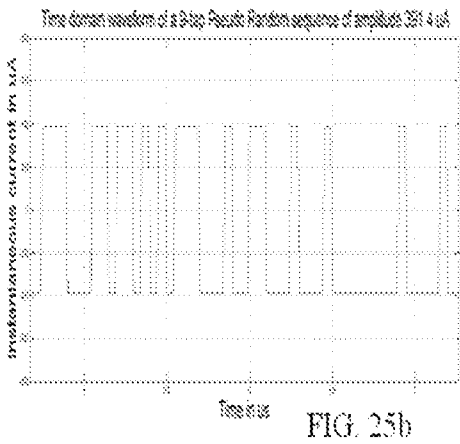
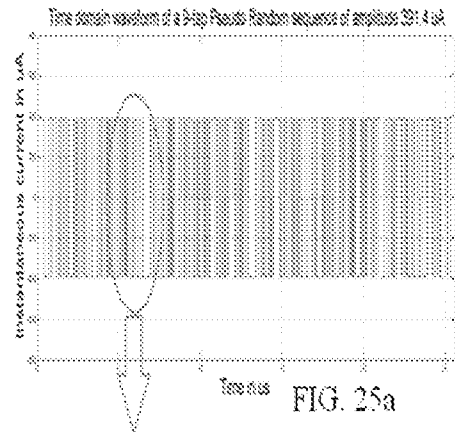


FIG. 24



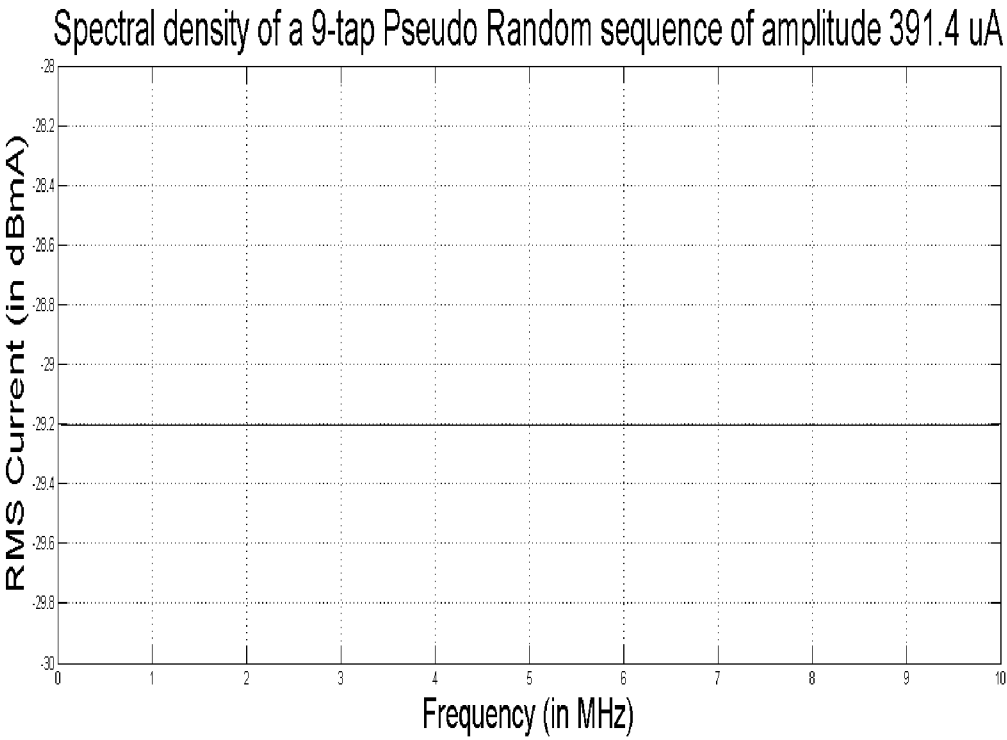
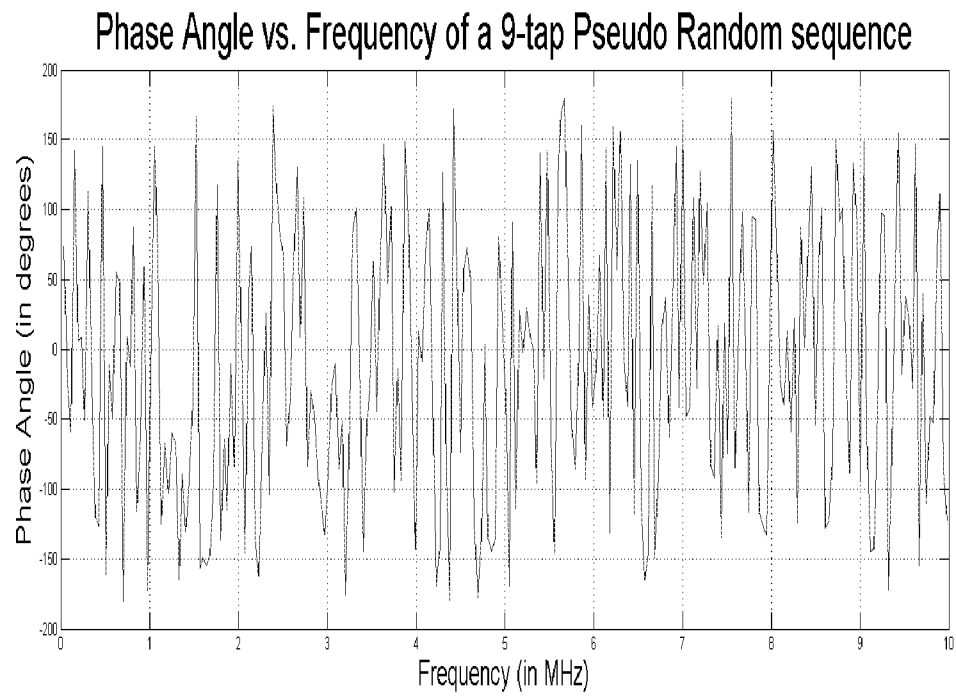


FIG. 26

**FIG. 27**

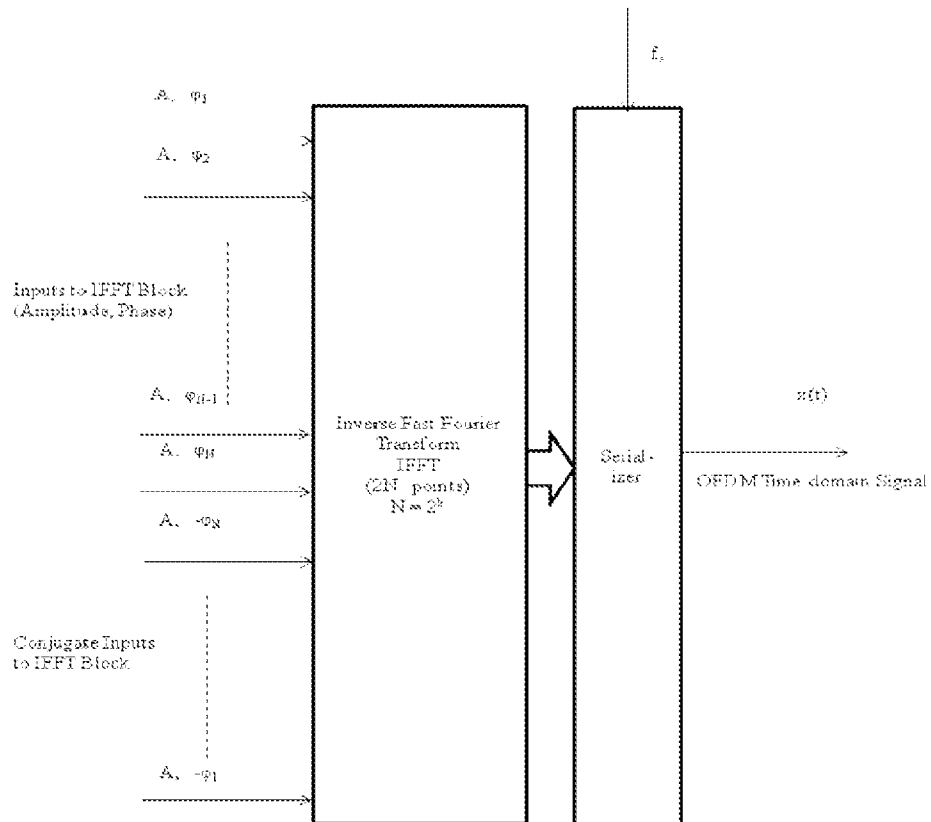
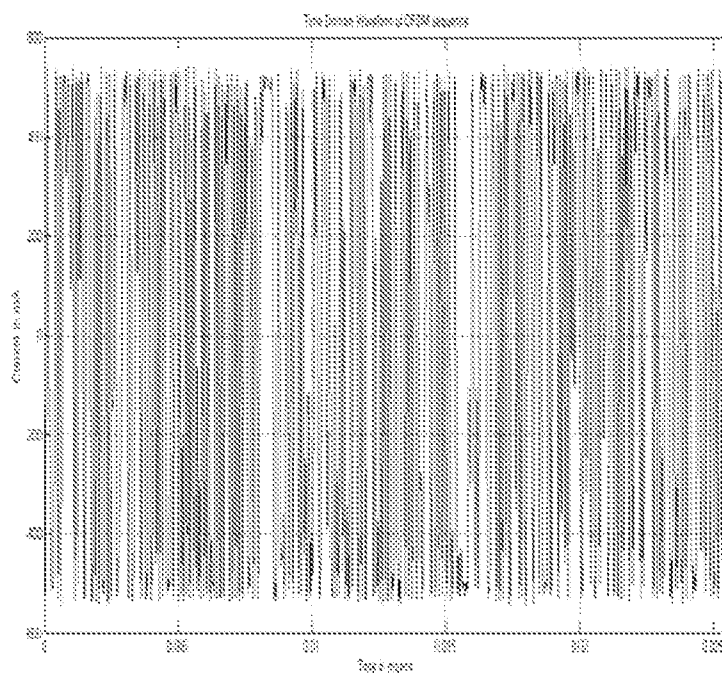
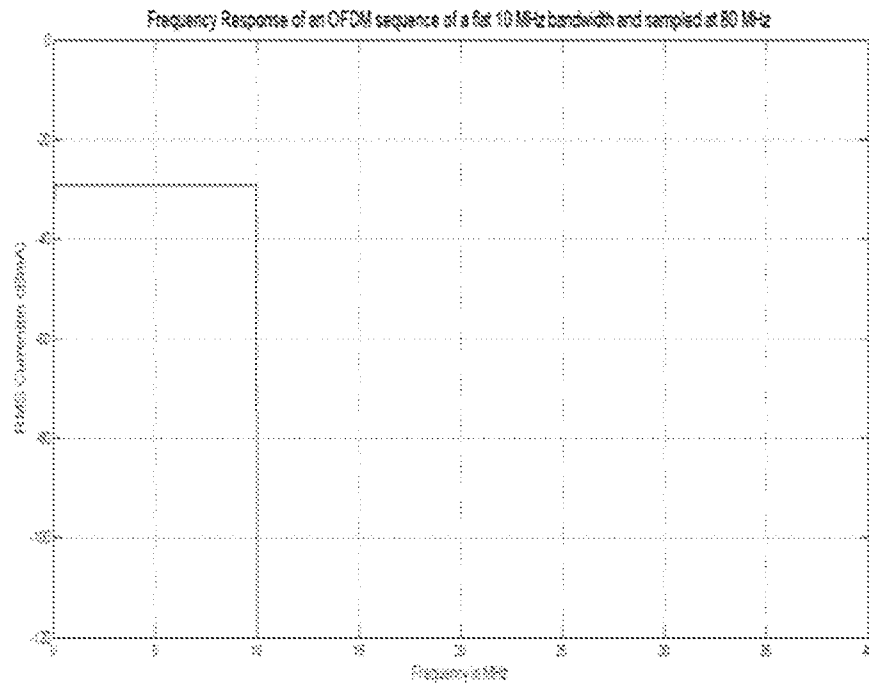


FIG. 28



$f_L = 39.0625$ KHz, $f_H = 10$ MHz, $f_S = 80$ MHz, $I_{RMS} = 390.625$ μ A, PAR = 1.39

FIG. 29



BW = 10 MHz, Sampling Rate = 80 MHz

FIG. 30

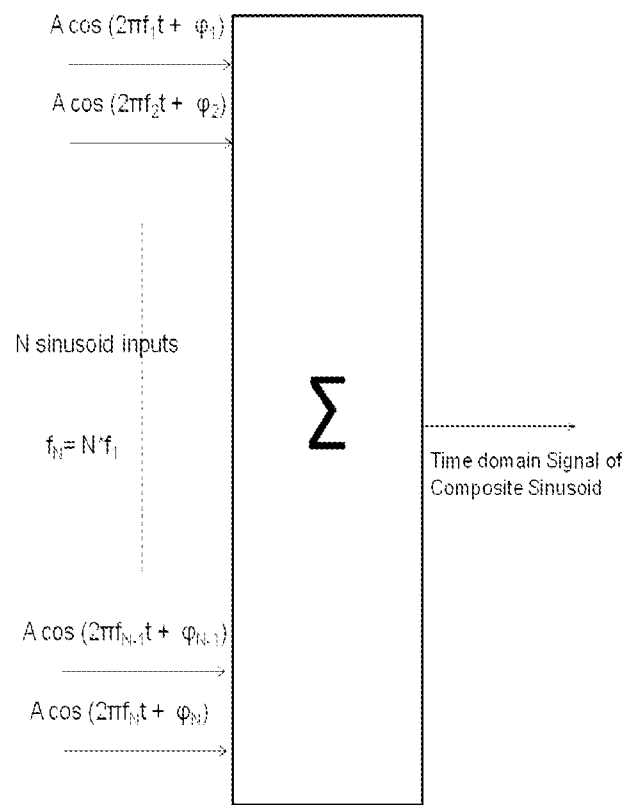


FIG. 31

INTERNATIONAL SEARCH REPORT

International application No.

PCT/IB2011/052553

A. CLASSIFICATION OF SUBJECT MATTER

See extra sheet

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)

IPC: A61B 5/053, A61B 5

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

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

CPRSABS, CNKI, DWPI, SIPOABS: electro⁺, curren⁺, voltage⁺, impedan⁺

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
P,X	CN201564473U(CHONGQING UNIVERSITY), 01 Sept. 2010(01.09.2010), the whole document	1,10
P,X	WO2011013345A1(PANASONIC ELEC WORKS CO LTD et al), 03 Feb. 2011(03.02.2011), the whole document	1
P,X	CN101940469A(ZHONGSHAN UNIVERSITY), 12 Jan. 2011(12.01.2011), the whole document	1-5,9
X	WO2010042855A1(UNIV MINNESOTA et al), 15 Apr. 2010(15.04.2010), specification paragraphs [0005]-[0077], figures 1-10	1-8
X	US2010036455A1(UNIV WASHINGTON), 11 Feb. 2010(11.02.2010), specification paragraphs [0013]-[0078], figures 1-7B	1-3,5-6,8

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

* Special categories of cited documents:	“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
“A” document defining the general state of the art which is not considered to be of particular relevance	“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
“E” earlier application or patent but published on or after the international filing date	“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
“L” document which may throw doubts on priority claim (S) or which is cited to establish the publication date of another citation or other special reason (as specified)	“&” document member of the same patent family
“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	

Date of the actual completion of the international search 09 Sept. 2011(09.09.2011)	Date of mailing of the international search report 20 Oct. 2011 (20.10.2011)
Name and mailing address of the ISA/CN The State Intellectual Property Office, the P.R.China 6 Xitucheng Rd., Jimen Bridge, Haidian District, Beijing, China 100088 Facsimile No. 86-10-62019451	Authorized officer FAN, Wenyang Telephone No. (86-10)62085628

INTERNATIONAL SEARCH REPORT

International application No.

PCT/IB2011/052553

C (Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US2008188730A1(CARDIAC PACEMAKERS INC), 07 Aug. 2008(07.08.2008), specification paragraphs [0003]-[0022], figures 1-4	1-3,5

INTERNATIONAL SEARCH REPORT
Information on patent family members

International application No.
PCT/IB2011/052553

Patent Documents referred in the Report	Publication Date	Patent Family	Publication Date
CN201564473U	01.09.2010	None	None
WO2011013345A1	03.02.2011	JP2011025071A	10.02.2011
		JP4706782B2	22.06.2011
		JP2011024831A	10.02.2011
CN101940469A	12.01.2011	None	None
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		US7356366 B	08.04.2008
		WO2006017446 A	16.02.2006
		EP1781169 A	09.05.2007
		JP2008508078T	21.03.2008
		AT485764T	15.11.2010

INTERNATIONAL SEARCH REPORT

International application No.

PCT/IB2011/052553

Continuation of: **CLASSIFICATION OF SUBJECT MATTER**

A61B 5/053 (2006.01) i

A61B 5/00 (2006.01) i

专利名称(译)	用于测量解剖学参数的系统和方法		
公开(公告)号	EP2579775A1	公开(公告)日	2013-04-17
申请号	EP2011795272	申请日	2011-06-13
[标]申请(专利权)人(译)	戈皮纳坦VENUGOPAL DUTTA GOUTAM PATIL NITIN PATKI作者Abhijit SUBRAMANIYAN拉加万		
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当前申请(专利权)人(译)	戈皮纳坦， VENUGOPAL DUTTA， GOUTAM 帕蒂尔NITIN PATKI， 作者Abhijit SUBRAMANIYAN， 拉加万		
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发明人	GOPINATHAN, VENUGOPAL DUTTA, GOUTAM PATIL, NITIN PATKI, ABHIJIT SUBRAMANIYAN, RAGHAVAN		
IPC分类号	A61B5/053 A61B5/00		
CPC分类号	A61B5/0535 A61B5/0538 A61B5/204 A61B5/6823 A61B5/6831 A61B5/6847 A61B5/6868 A61B2576 /023 A61N1/36521 G16H30/40		
优先权	1637CHE2010 2010-06-13 IN		
外部链接	Espacenet		

摘要(译)

提供了用于测量受试者中感兴趣区域中的解剖学参数的系统和方法。诊断装置包括一对或多对电极，用于传输感兴趣体积中的电信号并用于接收来自感兴趣区域的电压信号。还提供了一种在从表示电压信号的数据导出的频率范围内使用电测量（例如电压或电流）的方法。这些测量值可用于确定受试者的一个或多个解剖学特征的参数。

