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(54) **IMPLANTABLE PRESSURE MONITOR**

continuation of application No. 12/729,589, filed on Mar. 23, 2010, now Pat. No. 7,931,598, which is a continuation of application No. 11/452,920, filed on Jun. 15, 2006, now Pat. No. 7,686,768.

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CPC *A61B 5/0215* (2013.01); *A61B 5/0031* (2013.01); *A61B 5/6882* (2013.01)

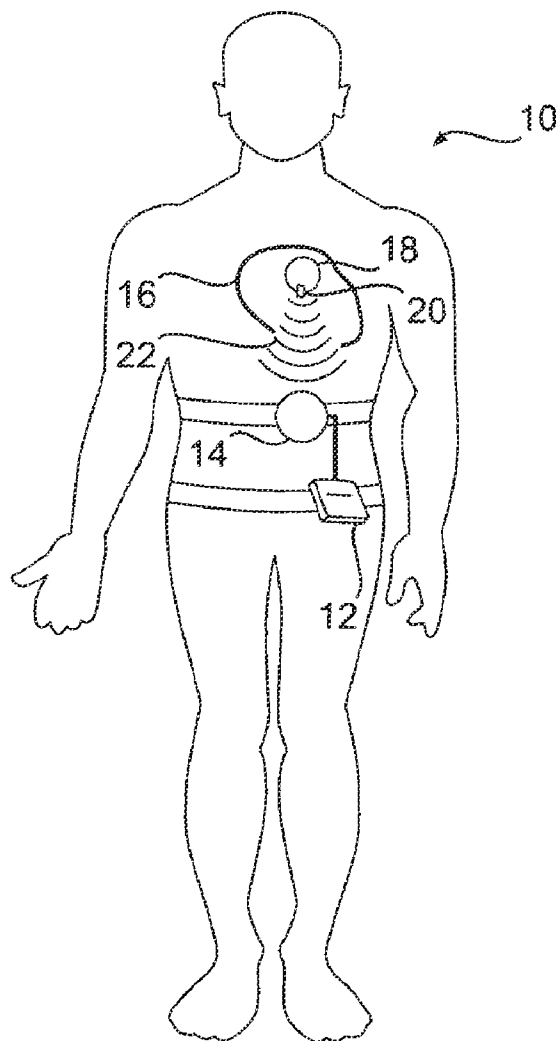
(22) Filed: **Feb. 19, 2013**

USPC **600/486**

Related U.S. Application Data

(63) Continuation of application No. 13/093,790, filed on Apr. 25, 2011, now Pat. No. 8,376,953, which is a

(57) **ABSTRACT**
An implantable pressure monitor.



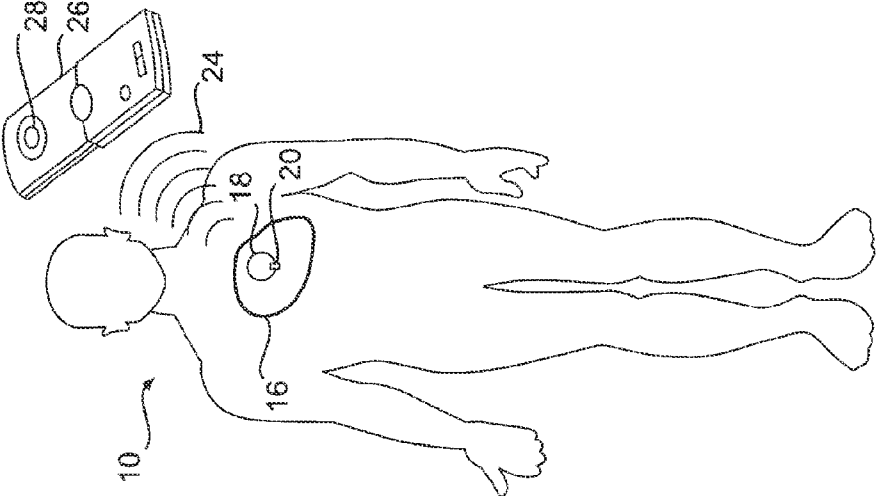


FIG. 2

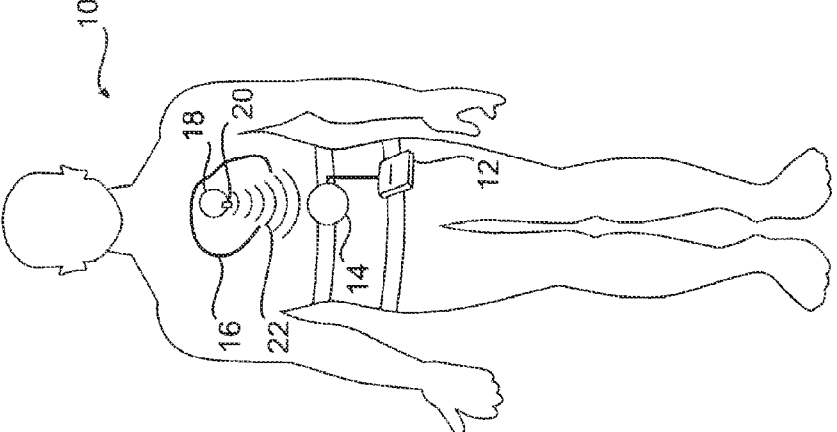


FIG. 1

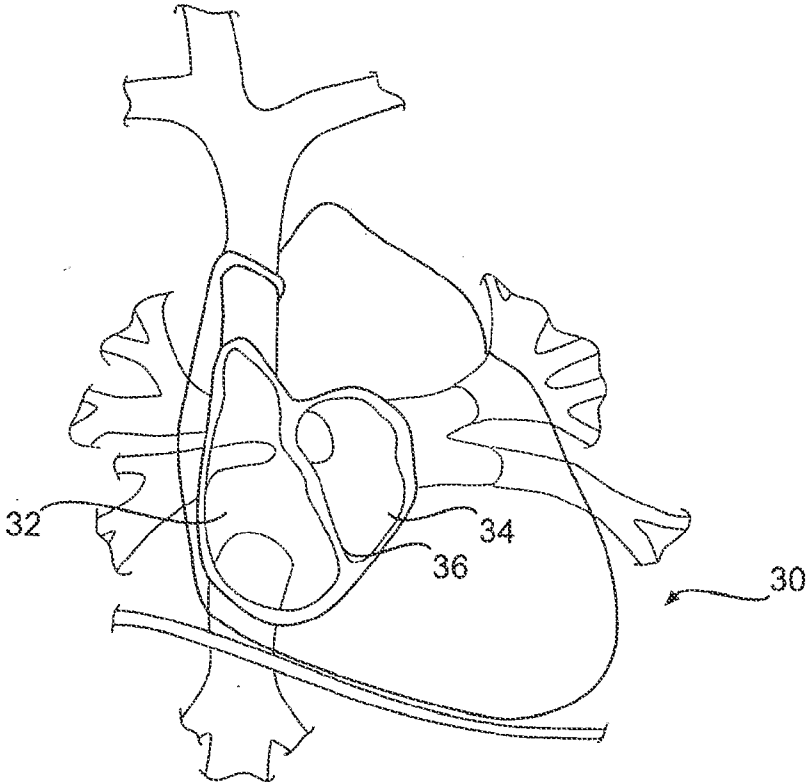


FIG. 3

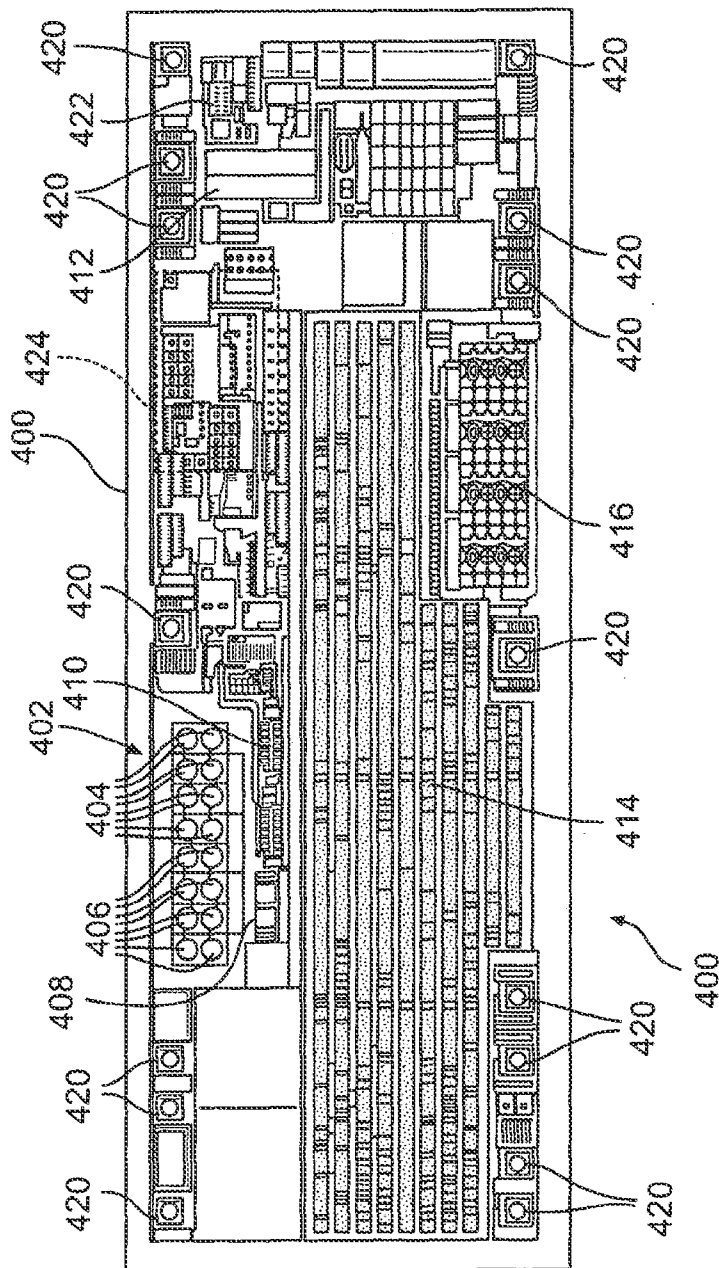


FIG. 4

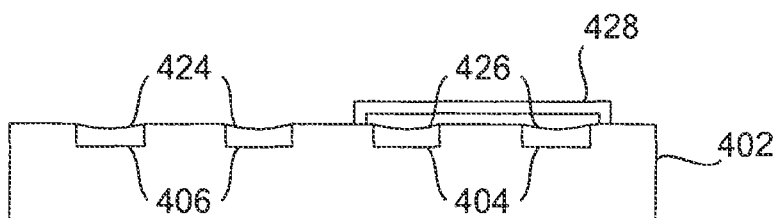


FIG. 5

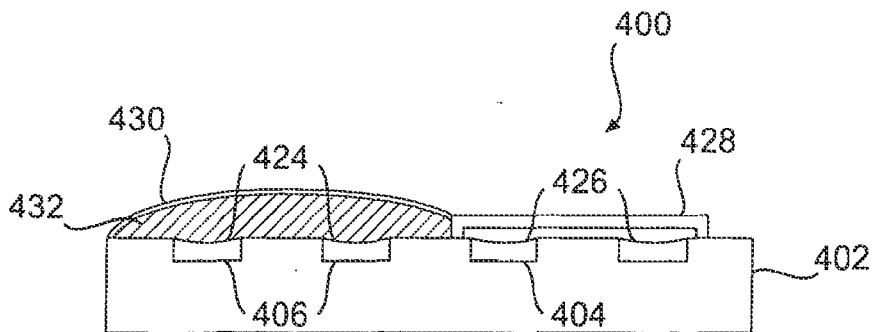


FIG. 6

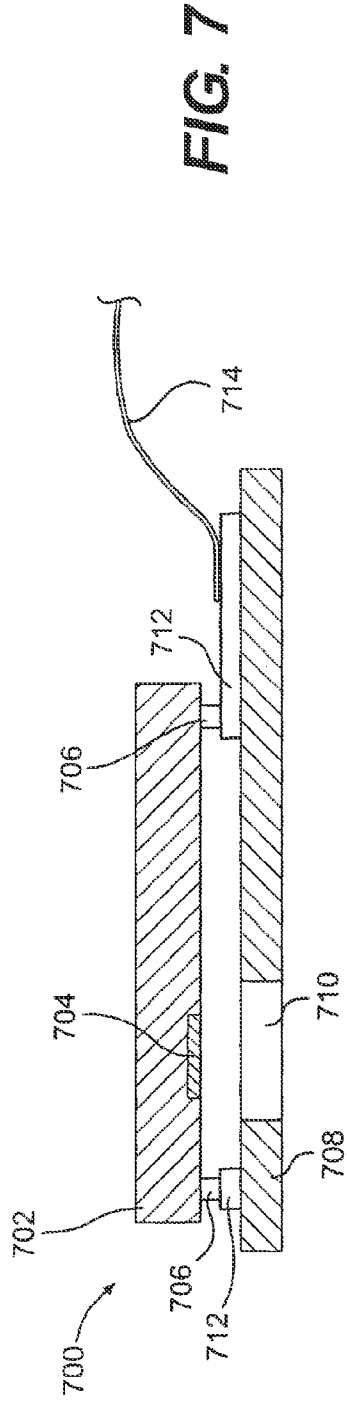


FIG. 7

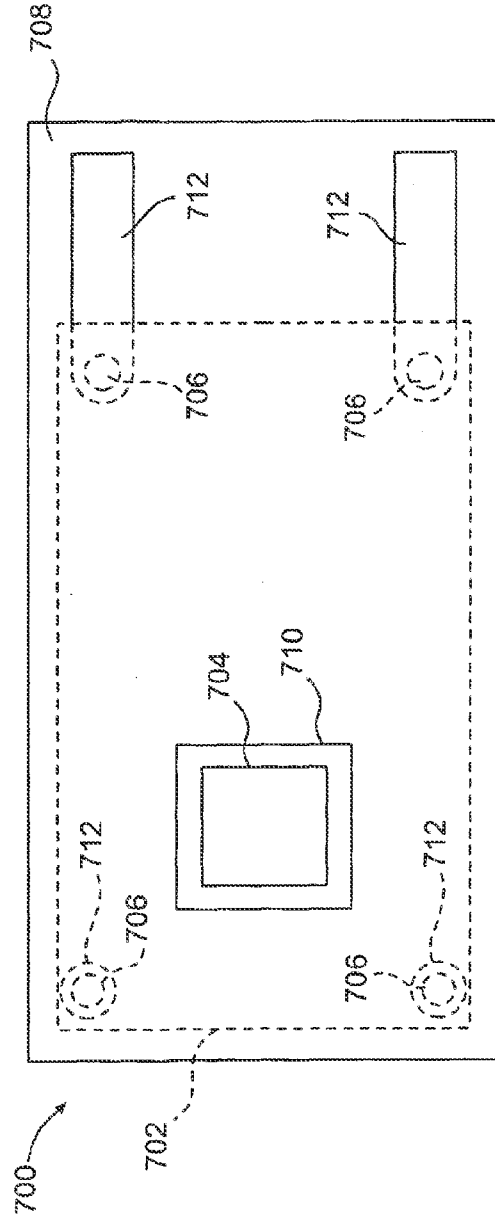


FIG. 8

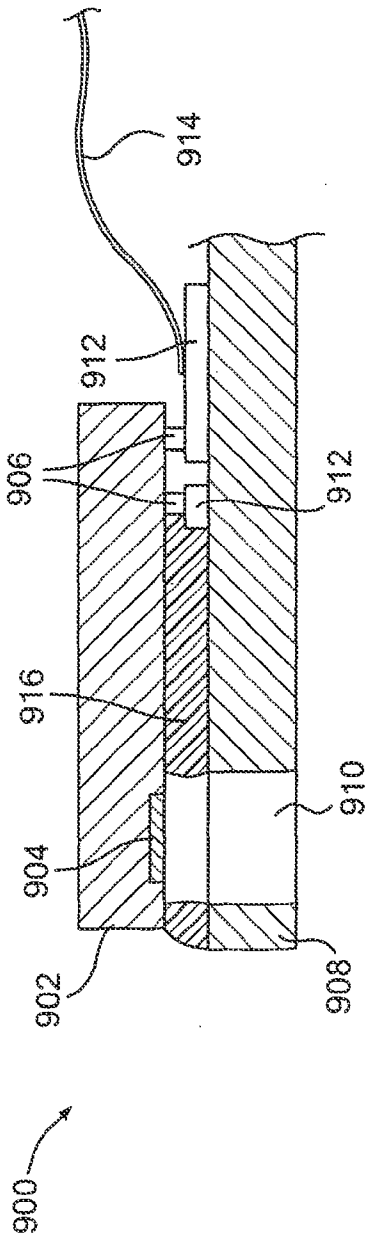


FIG. 9

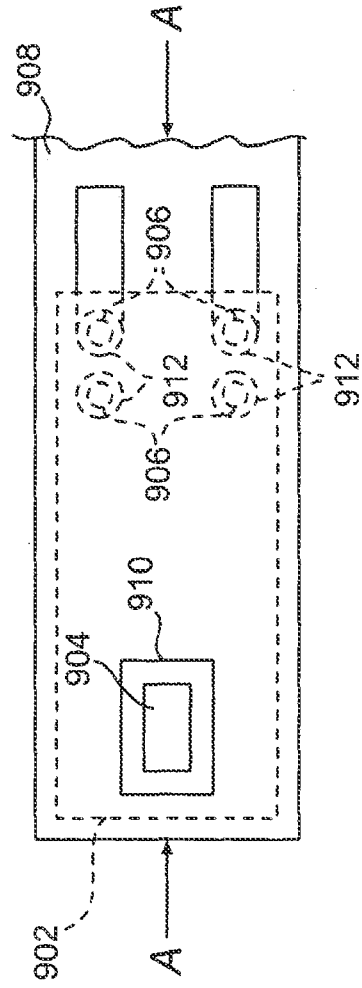


FIG. 10

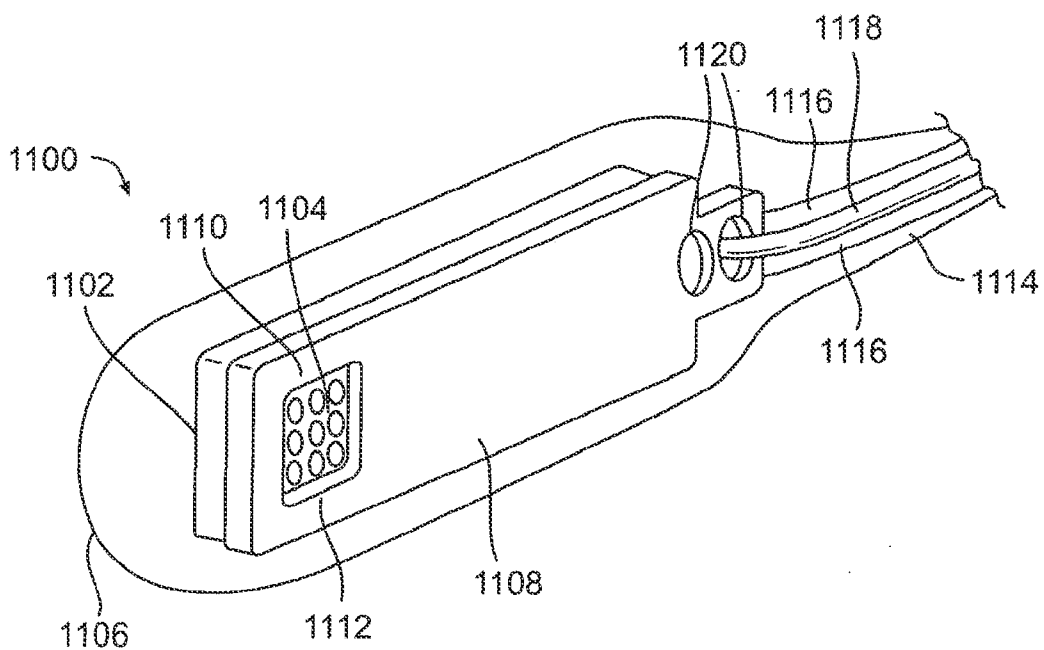


FIG. 11

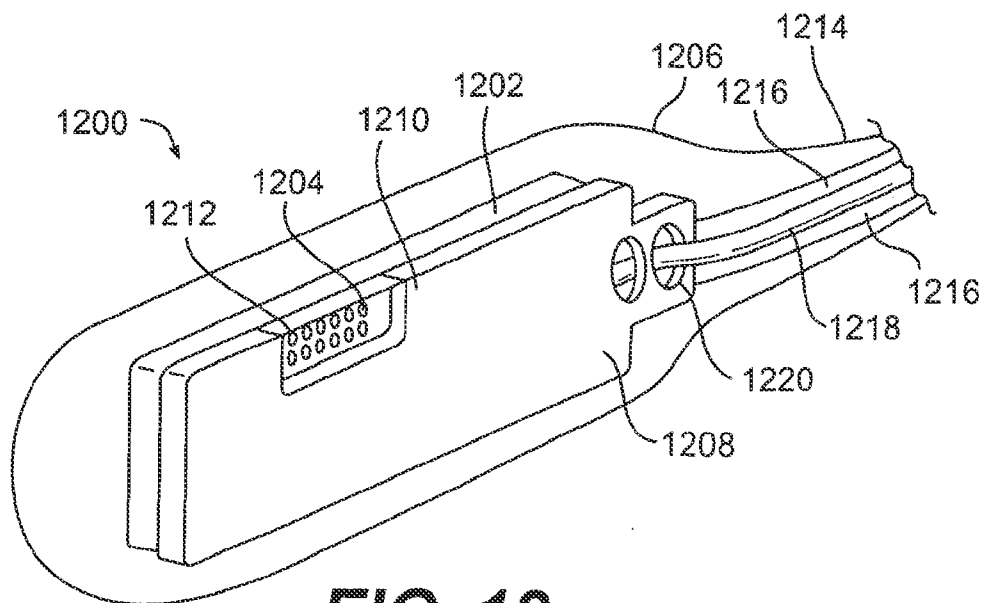


FIG. 12

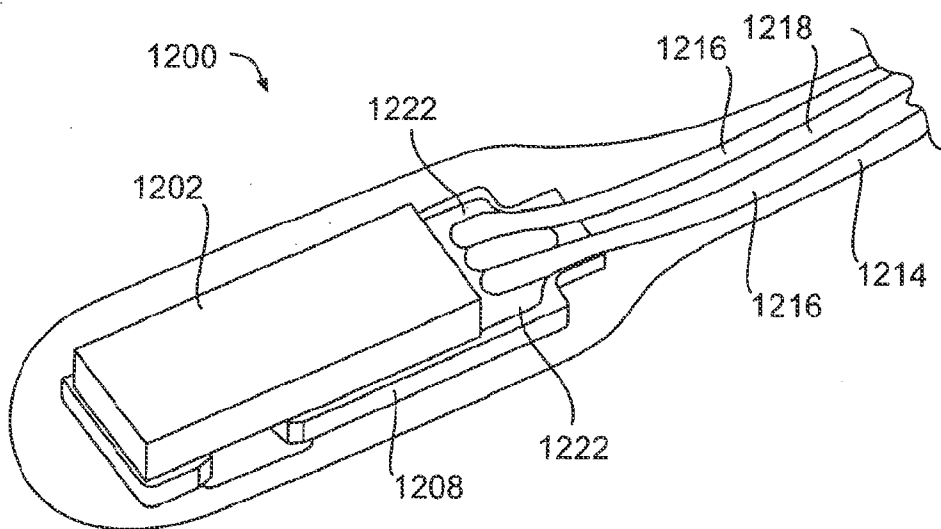


FIG. 13

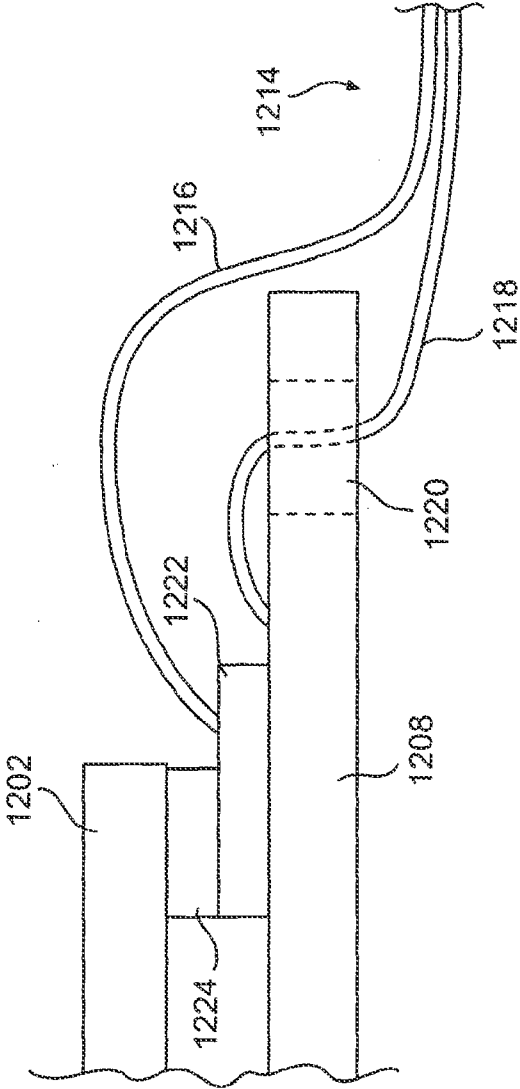


FIG. 14

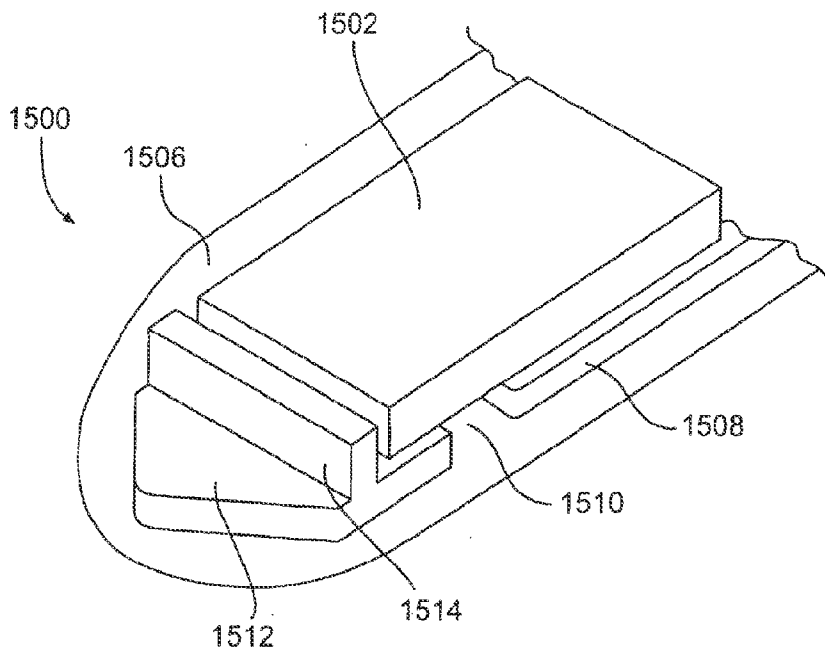


FIG. 15

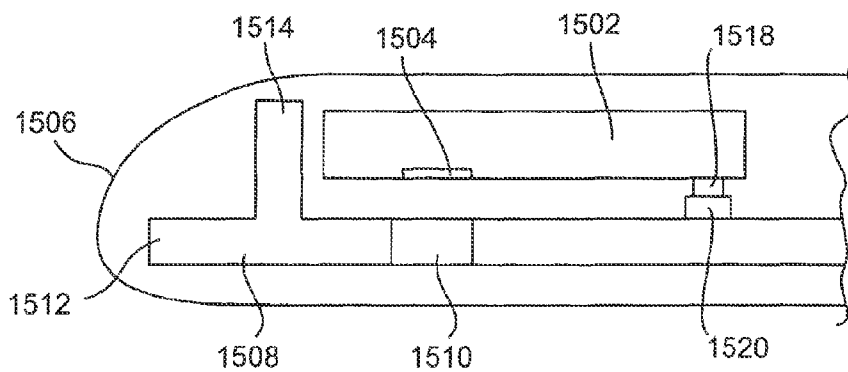


FIG. 16

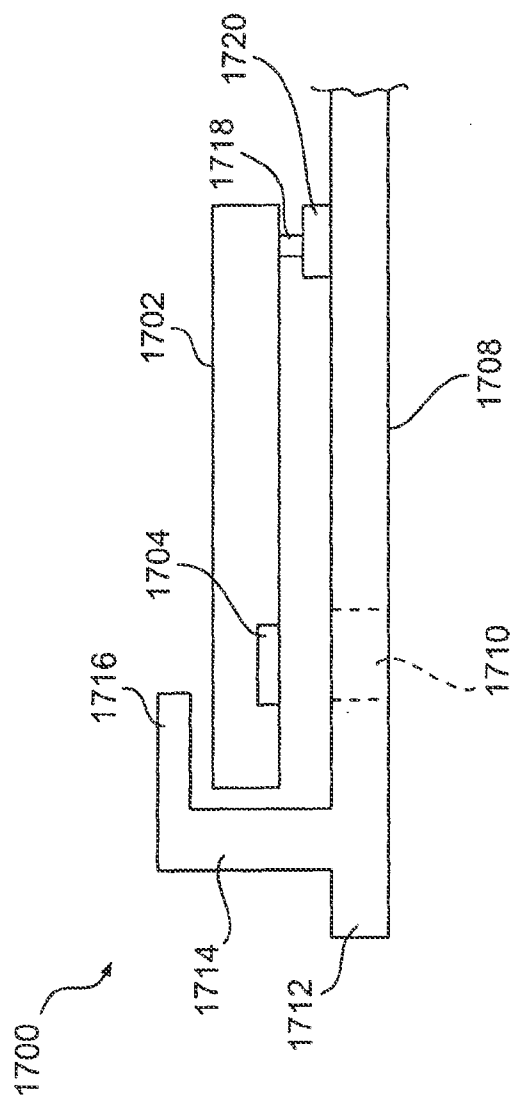


FIG. 17

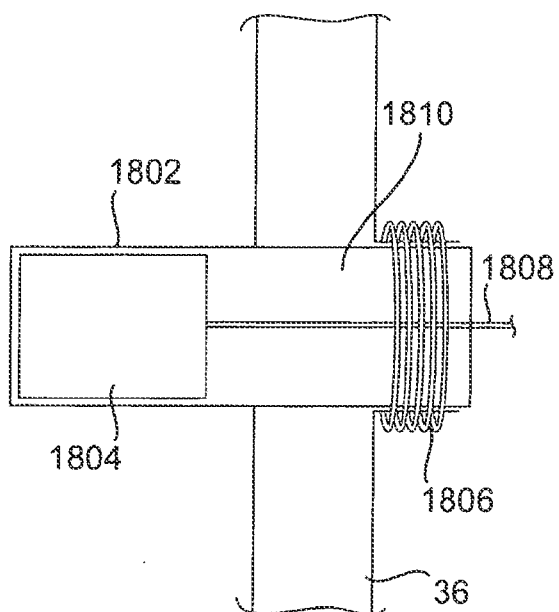


FIG. 18

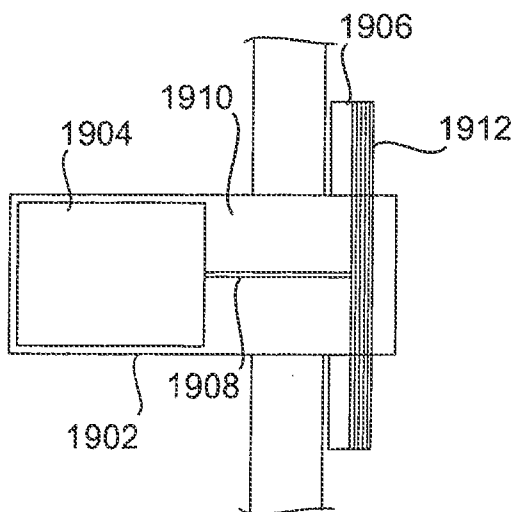


FIG. 19

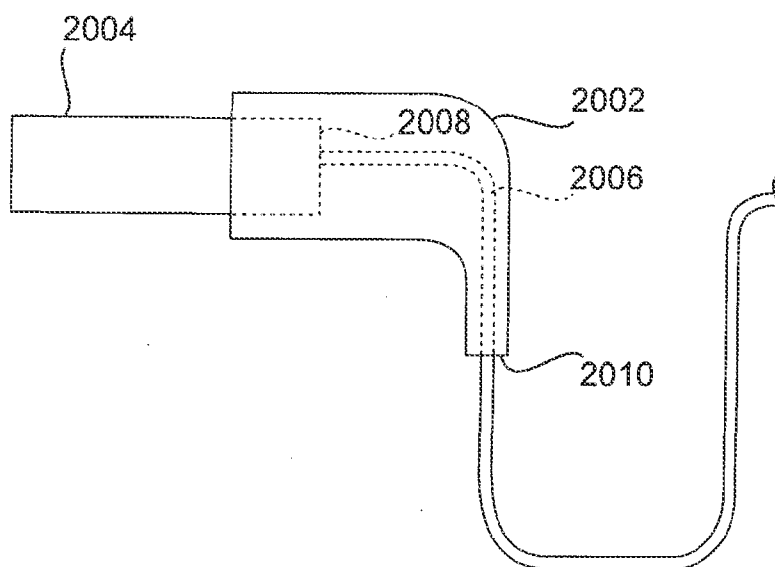


FIG. 20

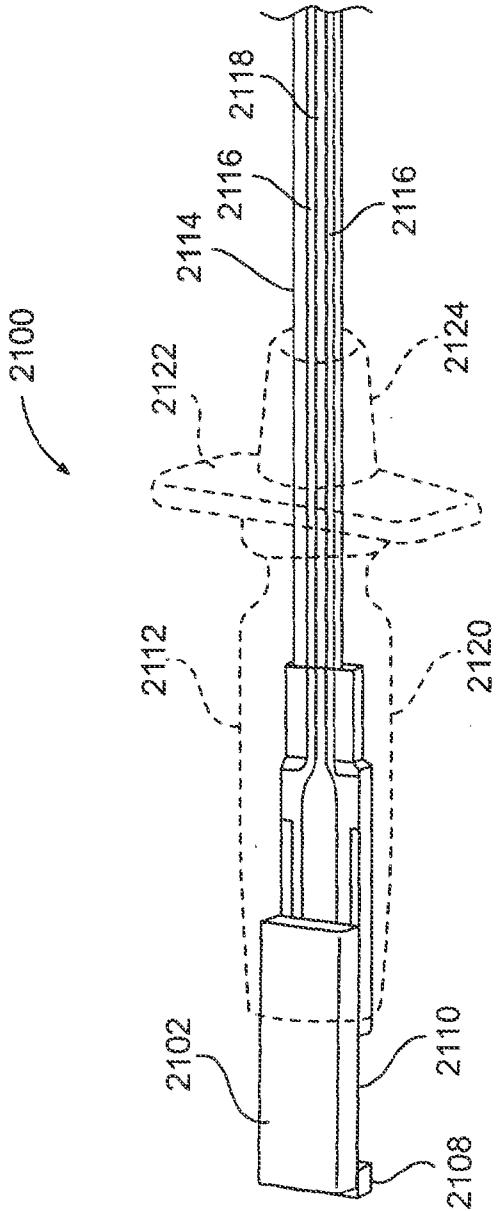


FIG. 21

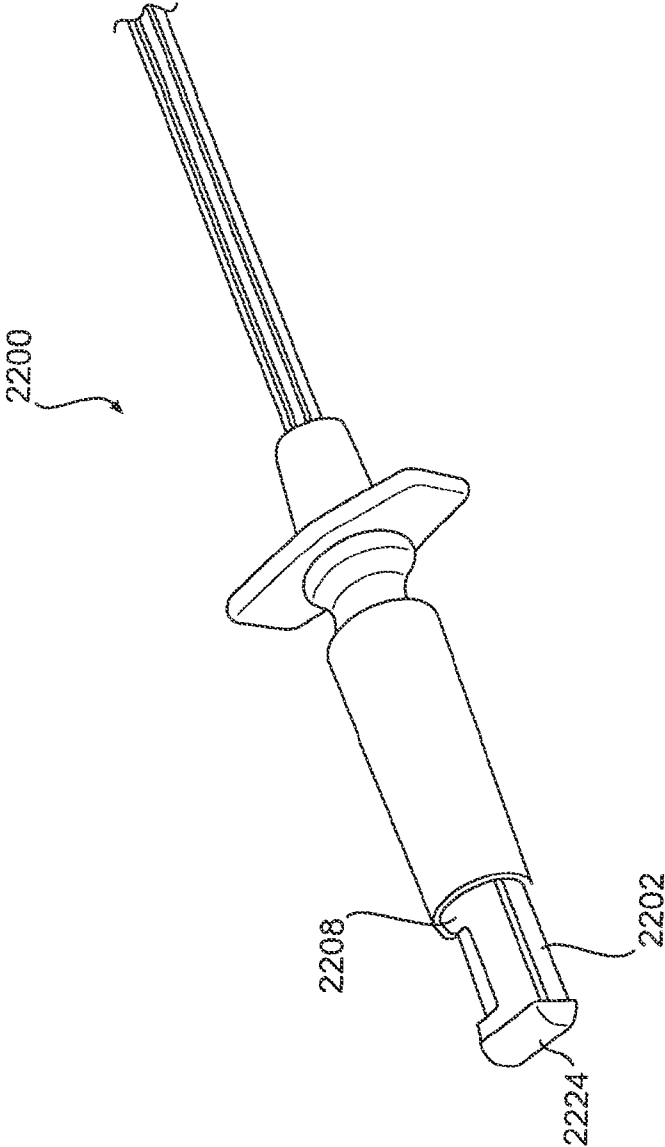


FIG. 22

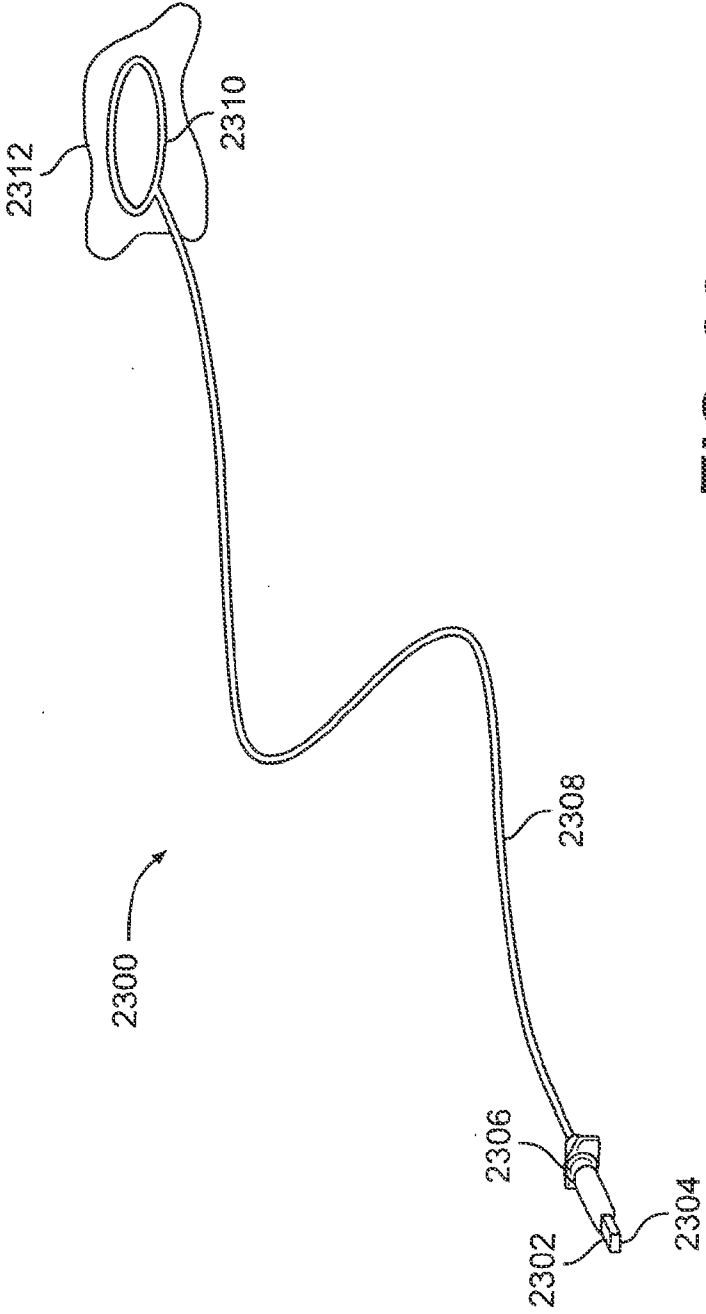


FIG. 23

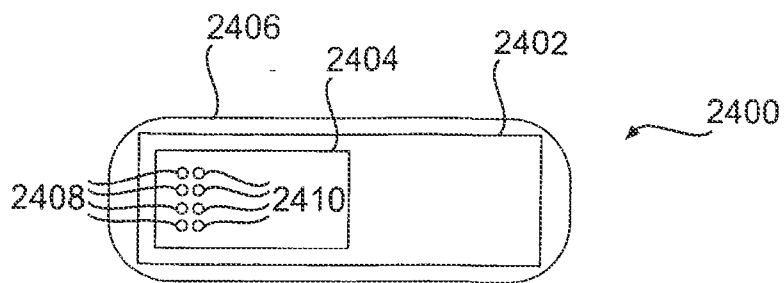


FIG. 24

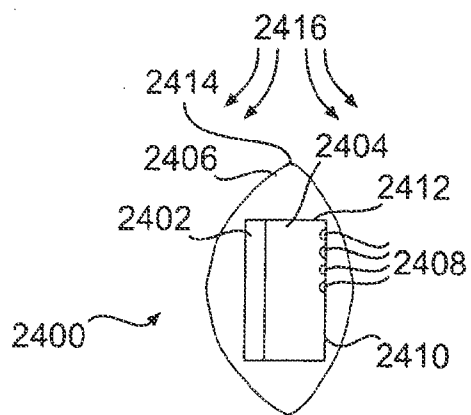


FIG. 25

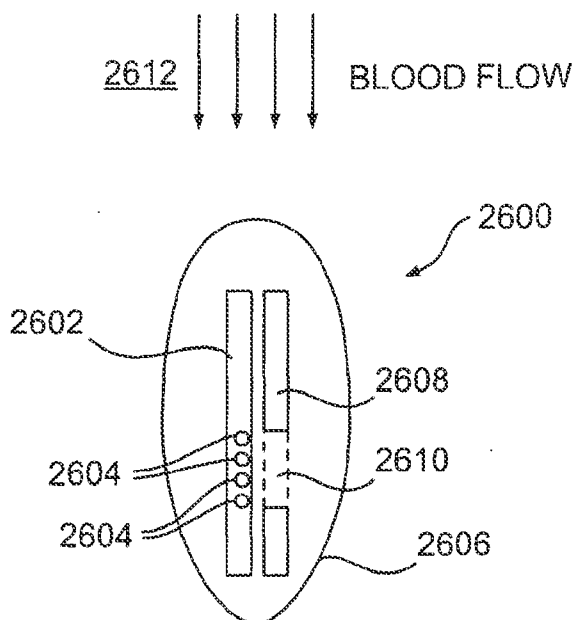


FIG. 26

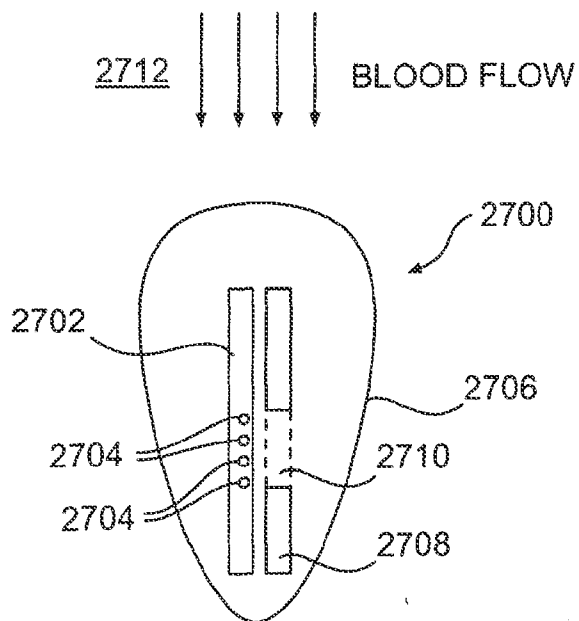


FIG. 27

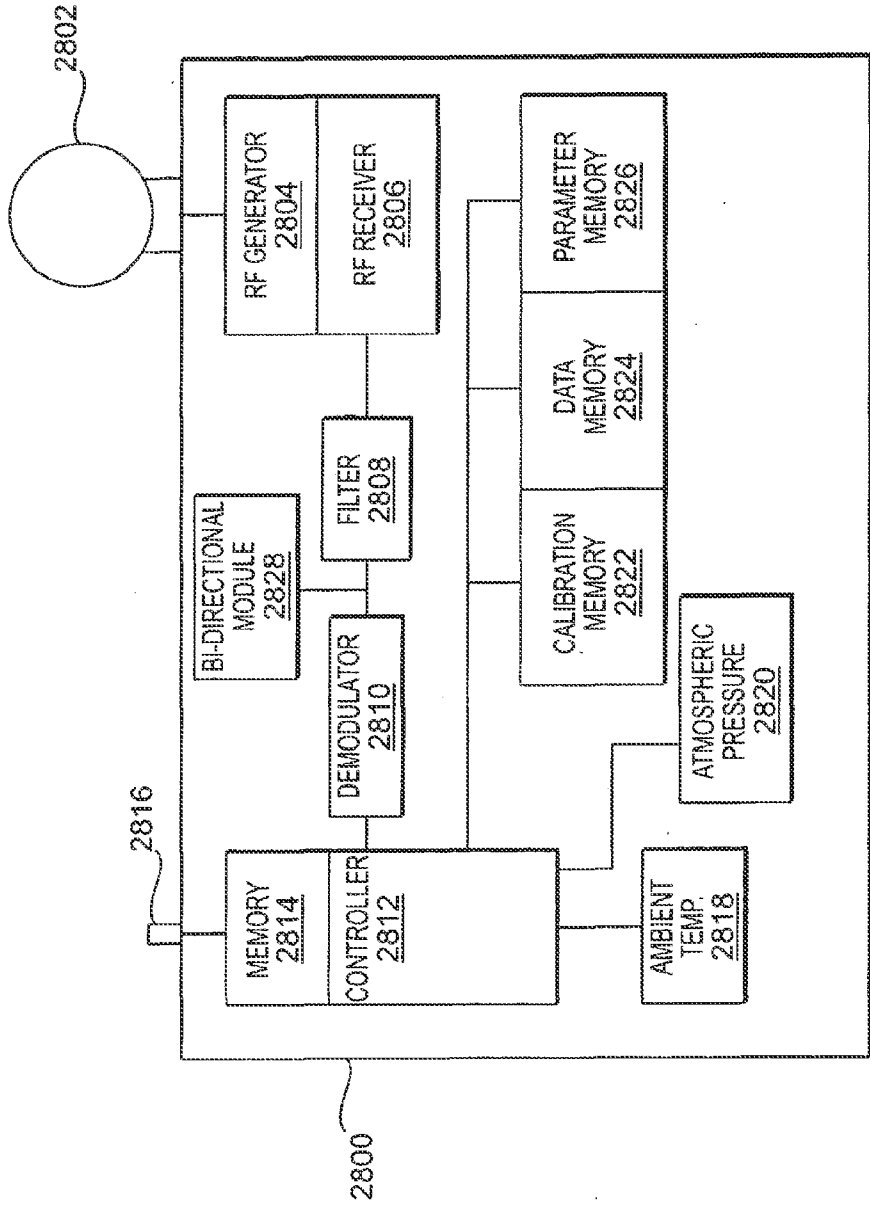


FIG. 28

IMPLANTABLE PRESSURE MONITOR

CROSS-REFERENCE TO RELATED APPLICATIONS

[0001] This application is a continuation of U.S. application Ser. No. 13/093,790, filed Apr. 24, 2011, which U.S. application Ser. No. 12/729,589, filed Mar. 23, 2010, which is a continuation of U.S. application Ser. No. 11/452,920, filed Jun. 15, 2006, which claims the benefit of U.S. provisional application Ser. No. 60/738,980 filed Nov. 23, 2005, and of U.S. provisional application Ser. No. 60/773,344, filed Feb. 15, 2006, the disclosures of which are hereby incorporated herein by reference.

FIELD

[0002] The invention is directed generally to a method and apparatus and for sensing a characteristic of a patient, such as blood pressure and/or temperature, and more particularly to methods and devices particularly adapted for telemetric measurement of blood pressure via a device implanted within the cardiovascular system during surgery and particularly within the heart.

BACKGROUND

[0003] The National Institute of Health (NIH) has concluded that heart failure constitutes "a new epidemic" in the USA. Heart failure, a chronic, progressive and incurable disease, affects over 20 million people worldwide. In the US alone, some 5 million people have been diagnosed with heart failure. Heart failure is estimated to cost the US economy today more than \$40 billion annually.

[0004] Intracardiac pressure management is an important aspect of heart failure treatment. For example, a rise of the intracardiac pressure, such as in the left atrium is an important early indication of disease progression and the first opportunity for therapeutic intervention. Current blood pressure-measuring methods only can be applied in the coronary care unit (CCU) or the intensive care unit (ICU) and provide no more than an occasional snapshot of intracardiac pressure when the patient is already in a very critical situation. The limitations on current intracardiac pressure measurement methods are a serious impediment to early and optimal treatment. Current treatment methods require hospitalization and may be extremely costly (on average, over \$16,000 per patient admittance). The ability to monitor patients and intervene outside of the hospital setting would greatly reduce the number of hospitalizations and extend the lives of those affected by the diagnosis.

[0005] Various sensors and devices have been used or proposed for the measurement and analysis of the blood pressure and/or temperature of a patient with mixed success. The currently contemplated sensors have certain disadvantages. For example, the telemetric sensor described in U.S. Pat. No. 6,855,115 can be implanted in the heart by a catheter and is not designed for surgical implantation. Moreover, the sensor, which is rolled up during the implantation procedure, must be made of a flexible material of a specific configuration so that any change of the blood pressure inside the heart effectuates a change in the distance of the sensor height, i.e., the distance between the two capacitor plates used in the sensor. This flexible sensor is folded for delivery via a catheter and then unfolded at the place of implantation. However, a disadvantage of such a configuration is its required flexibility as con-

stant and precise acquisition of measurement data may not be possible when the sensor is placed on or close to the cardiac muscle, and therefore is exposed to the cardiac motions, which may influence correct pressure readings. In addition, the flexible material of a sensor made in accordance with U.S. Pat. No. 6,855,115 may deform due to exposure to constantly streaming liquids, especially a turbulent blood stream likely encountered inside the heart. As a consequence, the capacitance of the capacitor may be changed and measurement values may deteriorate and/or deviate from the true value. Another disadvantage of this type of sensor is due to its use of a pressure-dependent LC-oscillator. The resonant frequency of this oscillator can be analyzed telemetrically. In principle, this kind of device can be applied to measure the pressure that affects the measurement capacitor. Thus, any damage to the material can affect the pressure measurements obtained. Further, as the sensor is influenced by the surrounding media of the sensor, a corruption of measurement values may occur. In addition, there is no circuitry in this type of sensor to digitize the pressure measurement values acquired. Using analog signals may result in external interference during the acquisition and transmission of data, which causes inaccuracies in readings.

[0006] Another exemplary implantable device, described in U.S. Pat. No. 6,409,674, uses a catheter filled with a pressure transmitting fluid or gel-like material. The catheter transmits pressure to a pressure transducer within a housing. The sensed pressure is then telemetrically transmitted to an external reader. However, such a device requires a housing for the electronic signal processing circuitry, which results in a larger and heavier sensor structure that can cause strain on the heart when implanted into a heart wall. Moreover, the catheter and housing configuration creates a more complicated, mechanical structure that may be at increased risk for mechanical failure, and therefore is not suitable for long term implantation.

[0007] Another device, described in U.S. Pat. No. 6,970,742, has a pressure sensor placed within the heart. A signal from the pressure sensor is transmitted to a housing outside the heart which contains the electronic processing circuits. The signal is processed by the electronic processing circuits, such as converting the signals from analog to digital, and then telemetrically transmitted to an external reader. However, housing the electronic processing circuitry requires additional components and a relatively larger implanted device. Moreover, because digitization of the signal does not occur until outside of the heart, there is a risk of interference in the wire connecting the sensor and the electronic processing circuitry, as analog interference may result from external sources.

[0008] Small pressure sensor chips including the electronic processing circuits have been used in other applications. For example, integrated chips having pressure sensors have been used for pressure measurement in optical and cranial applications. These sensors are compact and have fewer mechanical components. Examples of such pressure sensor chips are described in EP 1 312 302 A2 and German patent application DE 10 2004 005 220.7, in which the inventors of the present invention were involved. However, these integrated chips are used in a relatively stable environment, with little movement in the fluids of the eye or brain. Nor are these pressure sensors subject to the cyclical, dynamic movements found in the heart. Such movement may harm connections, such as connections between wires and the pressure sensing chip. Thus,

the use of such pressure sensor chips is not suited for the environment of the heart, where there is cyclical and dynamic movement, and where there is continuous and turbulent fluid movement around the pressure sensor.

[0009] Conventional techniques to provide stability and support to such known pressure sensing chips to enable their use as a cardiovascular pressure sensors would not likely succeed. Directly attaching a wire to a pressure sensing chip may have a negative impact on the functionality of the chip. For example, when soldering is used for the connection, the heat may damage the chip. One known method of avoiding that problem is to adhere a substrate to the back of the pressure sensing chip, solder the wire to a bond tack on that substrate, and then connect the wire to the chip. However, such substrates have different coefficients of thermal expansion than the chip. Thus, as the temperature changes, the substrate expands and contracts at a different rate than the pressure sensor chip, thereby causing stress and strain on the pressure sensing chip and increasing the risk of damage and/or inoperability.

[0010] Other known pressure sensors require a cable connection between the pressure sensor inside the heart and the external body monitoring device. However, such a cable clearly requires an entry into the body. An entry may be inconvenient and require the implantation of both the device and the entry, as well as increase the risk of infection for the patient.

[0011] Thus, there is a need for intra-cardiac pressure sensors that are more reliable and accurate, and which cause less irritation when implanted in the heart and are more compatible with the dynamic conditions encountered in a moving heart. Also, a need exists for such a sensor to be used at other locations within the cardiovascular system with little or no modifications.

SUMMARY

[0012] The invention meets the above needs and avoids the disadvantages and drawbacks of the prior art by providing a substantially rigid, chip-based telemetric sensor and system in which an extremely small and lightweight chip, including at least one pressure sensor and all necessary electrical circuitry, may be implanted into the heart or other portion of the cardiovascular system during surgery, to monitor blood pressure and/or temperature.

[0013] In this manner, pressure signals may be digitized at or near the sensing location in the heart or other location in the cardiovascular system and data may be telemetrically directed to the place of data acquisition to reduce or eliminate data transmission interference from external sources.

[0014] In particular, the chip may be a substantially rigid structure that provides improved durability, long term stability, and long term accuracy, and resistance to damage or a change in membrane characteristics from the blood flow due to turbulences and the like within the bloodstream. For example, the chip may be an application specific integrated chip (ASIC) containing all the necessary sensing elements and digital signal processing electronics. The ASIC preferably is very small and lightweight to avoid undue stress on the heart and is orientated within the body in a position to minimize turbulent flow and reactionary forces. The ASIC may be used with an antenna in the form of a coil created with very small dimensions. This minimal configuration of ASIC and coil may reduce and/or eliminate mechanical tensions effecting the connection between ASIC and a coil.

[0015] The ASIC and the coil may be electrically and physically connected by a flexible coil. The ASIC, cable and coil may be encapsulated within a seamless biocompatible and flexible sheathing, such as silicone or similar material, to form an integrated sensor unit. The seamless sheathing may maintain the integrity of the sensor by reducing or eliminating the exposure of the sensor to body fluids, such as blood. It may also be shaped and/or orientated to reduce turbulent flow.

[0016] A liquid or gel may be placed between the pressure sensing elements, such as capacitive membrane sensors of the sensor and the sheathing, to reduce or eliminate the effects of endothelialization on the surface of the sensor. The sheathing material itself may act as a pressure transmitting material. The liquid or gel allows for integrating the pressure across the entire area of pressure sensing portion of the sensor to minimize the effects of localized plaque or endothelialization. Of course, heparin and other preventative coatings known in the art also may be used to prevent or reduce endothelialization.

[0017] To protect the ASIC and particularly the membrane sensor elements from damage due to handling, e.g., as a consequence of contact with a surgical instrument during implantation and/or during use, the sensor design may have a unique geometry. For example, the ASIC may be connected to a substantially rigid substrate in a spaced apart relationship from the ASIC such that the substrate is opposite the pressure sensing elements of the sensor chip, with an aperture in the substrate providing access to the pressure elements to expose them to fluid pressure to be sensed. A silicone or other similar flexible material may be disposed between the ASIC and the substrate. Moreover, a pressure transmitting material may be placed within the gap between the ASIC and the rigid substrate so that pressure from the blood can be transmitted to the pressure elements via the material.

[0018] The ASIC may incorporate a robust system to compensate for drift due to the age and use of the sensor. For example, the ASIC may include inactive pressure sensing elements that determine the change in the measurement in the sensor due to age and usage, and may account for this change when active pressure sensing elements determine the pressure.

[0019] The ASIC may be supported in a holder particularly adapted for anchoring the ASIC in a wall of the heart or other location in the cardiovascular system. The holder may include a stop to position the ASIC and limit its movement.

[0020] The ASIC is powered by induction from a wireless signal from an external reader, thereby avoiding the need for an internal power source. Use of a transponder power supply at the external reader allows for a substantially rigid sensor device with a longer life. The external reader provides power to the substantially rigid sensor and receives pressure and temperature information from the substantially rigid sensor. The external reader stores and displays measurement and parameter data, calculates certain values. The external reader stores and displays measurement and parameter data, and may transmit the data to a computer or other device for further processing. The external reader may have a separate antenna coil to facilitate prolonged periods on a patient's body. The external reader may store one or more calibration curves for different sensors.

[0021] The telemetric pressure and/or temperature sensor of the invention may be used for continuous or on demand sensing. A specific identification number may be transmitted with each single measurement or measurement cycle. In this way, a continuous measurement value and sensor identifica-

tion, and therefore the measurement value and the identity of the patient, is provided. The identification number may allow a single external reader to receive data from multiple sensors and systems and to assign them to the correct calibration curve for that sensor system and the patient.

[0022] The invention may be implemented in a number of ways. According to one aspect of the invention an intracardiac pressure measuring system for measuring blood pressure inside the heart of a patient includes an antenna and an integrated chip. The integrated chip may include a first substantially rigid substrate, at least one pressure sensor disposed within the substrate to generate signals indicative of a sensed pressure, and electronic signal processing components to process the signals generated by the at least one pressure sensor. The electronic signal processing components may be operatively connected to the antenna and the integrated chip may be powered by a signal received at the antenna. An implantable holder may support the integrated chip and include an anchor structure to mount the integrated chip within a wall of the heart during surgery such that the at least one pressure sensor is exposed to blood flow in the heart. The system may also include a remote receiver, wherein the integrated chip is operative to send digital signals indicative of the pressure sensed in the heart telemetrically via the antenna to the remote receiver.

[0023] The at least one pressure sensor may be capacitive-based pressure sensitive membranes housed within the substrate. The at least one pressure sensor may generate an analog signal in response to a sensed pressure and the electronic signal processing components may include at least one analog to digital (A/D) converter to digitize within the heart the analog signals from the at least one pressure sensor. The system may further include a flexible wire connecting the antenna and the integrated circuit, with the antenna being configured to be implanted within the patient underneath the skin to facilitate telemetric data transmission to the receiver. The integrated chip may weigh less than about one gram, have a surface area on one side of less than or equal to about 10 mm² and have a thickness of less than about 1 mm.

[0024] The antenna, the chip, wire and holder may be encapsulated in a seamless, one-piece biocompatible sheathing. A pressure transferring medium may be interposed between the biocompatible sheathing and the at least one pressure sensor. The biocompatible sheathing may act as the pressure transferring medium, and may be shaped to minimize turbulence in blood flow within the heart. The integrated chip may further include a unique digital identification, which is sent telemetrically to the receiver. The receiver may obtain calibration information associated with the integrated chip based on the unique digital identification. The receiver may include a stored parameter and produce an alert based on the signals indicative of the pressure sensed in the heart and of the stored parameter.

[0025] The system may further include a second substantially rigid substrate located opposite the at least one pressure sensor in the first substrate and in a spaced apart configuration to protect the chip from mechanical damage. The second substrate may include an aperture permitting blood flow within the heart to act on the at least one pressure one pressure sensor. A pressure transferring medium may be interposed between the at least one pressure sensor and the second substantially rigid substrate to transfer blood pressure to the at least one pressure sensor. At least one bond pad may be disposed between the first and second substantially rigid sub-

strates and electrically connected to the integrated chip. At least one bond tack may be provided on the second substantially rigid substrate and be connected to the at least one bond pad such that the antenna is operatively connected to the integrated chip via the at least one bond pad and the at least one bond tack to provide a strain relief connection. An antenna connector may connect the antenna to the second substantially rigid substrate, wherein the antenna connector includes a signal portion electrically connecting the antenna to the integrated chip and a support portion connected to the second substantially rigid substrate. The antenna connector may be attached to the second substantially rigid substrate such that there is slack in the signal portion when the support portion is taut. The signal portion may be connected to the integrated chip via the at least one bond tack and the support portion may be connected to the second substantially rigid substrate via an opening in the second substantially rigid substrate. The at least one bond tack may include at least two bond tacks disposed on opposite sides of the aperture. The second substantially rigid substrate may include a protective barrier connected thereto, and a biocompatible sheathing may encapsulate at least the integrated chip and the second substantially rigid substrate. The protective barrier prevents the first substantially rigid substrate from puncturing the biocompatible sheathing. A flexible support material may be provided between the integrated chip and the second substantially rigid substrate.

[0026] The holder and the first substantially rigid substrate may be integrally formed into a single piece. The antenna may be supported by said holder. The holder may include a stop that limits the movement of the integrated chip into the heart chamber. The integrated chip and the holder may be removable from the heart after implantation. The at least one pressure sensor may include a plurality of pressure sensors including at least one active sensor responsive to changes in pressure within the heart and at least one passive sensor that is isolated from the changes in pressure within the heart, and the electronic signal processing components may provide a signal based at least in part on a signal from the at least one active pressure sensor and a signal from the at least one passive pressure sensor. The structure of the active pressure sensor may be substantially the same as a structure of the passive pressure sensor. The plurality of pressure sensors may include capacitive pressure sensors each having a flexible movable membrane. The passive pressure sensor signal may be responsive to a change in position of the membrane of the passive pressure sensor, which is due to a drift effect comprising a sag of said membrane. The change of position of the membrane of the active pressure sensor may be due to a change in pressure within the heart and a drift effect comprising a sag of the membrane. The pressure signals may be the result of offsetting the signal from the at least one active pressure sensor with the signal from the at least one passive pressure sensor.

[0027] According to another aspect of the invention a method of sensing blood pressure within the cardiovascular system of a subject includes the steps of (a) implanting within the subject an integrated chip including a substantially rigid substrate and at least one capacitive-based pressure sensor disposed within said substrate in a position to sense blood pressure within the cardiovascular system; (b) powering on the integrated chip telemetrically by activating a power source located outside the subject; (c) obtaining one or more analog signals from the at least one pressure sensor indicative

of the pressure at the position in the cardiovascular system; and (d) converting the analog signals to digital signals at or directly adjacent to the position in the cardiovascular system where the sensing occurs.

[0028] The implanting step may include implanting an ASIC having a capacitive-based pressure sensor in the heart. The method may further include the step of limiting the ASIC from entering the heart chamber with a stop device. The integrated chip may include a unique digital identification, and the method may include the step of telemetrically communicating the unique digital identification to an external reader. The method may further include the step of obtaining calibration information associated with the integrated chip at the external reader based on the unique digital identification. The integrated chip may be supported in a holder and the implanting step may include the steps of delivering the holder to the position in the cardiovascular system, and mounting the holder at the position such that the at least one pressure sensor is exposed to the pressure in the cardiovascular system to be sensed.

[0029] The at least one capacitive-based pressure sensor may include a plurality of capacitive-based pressure sensors including an active pressure sensor and a passive pressure sensor located within the subject in a position to directly sense blood-pressure within a position in the cardiovascular system, and the step of obtaining one or more analog signals may further include obtaining one or more analog signals from the active pressure sensor indicative of the pressure at the position in the cardiovascular system, obtaining one or more analog signals from the passive pressure sensor indicative of the pressure at the position in the cardiovascular system, and generating one or more combined analog signals based on the one or more analog signals from the active pressure sensor and the one or more analog signals from the passive pressure sensor indicative of the pressure at the position in the cardiovascular system. The step of converting the analog signals may further include converting the combined analog signals to digital signals. The step of generating the one or more combined analog signals may include offsetting the signal from said active pressure sensor with the signal from said passive pressure sensor.

[0030] In yet another aspect of the invention an integrated chip for intra-cardiac blood pressure measurement inside the heart of a patient includes a first substantially rigid substrate, at least one pressure sensor disposed within the substrate to generate signals indicative of a sensed pressure, and electronic signal processing components to process the signals generated by the at least one pressure sensor. The electronic signal processing components may be operatively connected to an antenna, and the integrated chip may be powered by a signal received at the antenna. The integrated chip is operative to send digital signals indicative of the pressure sensed in the heart telemetrically via an antenna to a remote receiver.

[0031] The at least one pressure sensor may generate analog signals and the electronic signal processing components may include at least one analog to digital (A/D) converter to digitize within the heart the analog signals from the at least one pressure sensor. The integrated chip may weigh less than about one gram, have a surface area on one side of less than or equal to about 10 mm² and have a thickness of less than about 1 mm. The integrated chip may further include a second substantially rigid substrate located opposite the at least one pressure sensor in the first substrate and in a spaced apart configuration. The second substrate may include an aperture

permitting blood pressure within the heart to act on the at least one pressure one pressure sensor. The integrated chip may also include a pressure transferring medium interposed between the at least one pressure sensor and the second substantially rigid substrate to transfer blood pressure to the at least one pressure sensor.

[0032] Additional features, advantages, and embodiments of the invention may be set forth or apparent from consideration of the following detailed description, drawings, and claims. Moreover, it is to be understood that both the foregoing summary of the invention and the following detailed description are exemplary and intended to provide further explanation without limiting the scope of the invention as claimed.

BRIEF DESCRIPTION OF THE DRAWINGS

[0033] The accompanying drawings, which are included to provide a further understanding of the invention, are incorporated in and constitute a part of this specification, illustrate embodiments of the invention and together with the detailed description serve to explain the principles of the invention. No attempt is made to show structural details of the invention in more detail than may be necessary for a fundamental understanding of the invention and the various ways in which it may be practiced. In the drawings:

[0034] FIG. 1 schematically illustrates an embodiment of an implantable telemetric measuring device and reader constructed according to principles of the invention providing for continuous or regular intra-cardiac pressure monitoring;

[0035] FIG. 2 schematically illustrates another embodiment of an implantable telemetric measuring device and reader constructed according to principles of the invention providing for on-demand intra-cardiac pressure monitoring;

[0036] FIG. 3 illustrates a cross-sectional view of the heart area of a patient where the implanted device of the invention may be employed, including the left and right atrium and the crossing of veins at the posterior septum;

[0037] FIG. 4 illustrates a greatly enlarged, plan view of a substantially rigid ASIC constructed according to principles of the invention for sensing intra-cardiac pressure and temperature including active and passive, capacitive membrane sensing elements and on-chip electronics for digital signal processing and telemetric power supply.

[0038] FIG. 5 is an enlarged, cross-sectional view of the ASIC of the invention showing some of the active pressure sensors and passive pressure sensors;

[0039] FIG. 6 is an enlarged, cross-sectional view of the ASIC of the invention showing a pressure transmitting gel or fluid between a sheathing and the active pressure sensors;

[0040] FIGS. 7 and 8 schematically illustrate a cross-sectional and top plan view, respectively, of one embodiment of an implantable sensor device of the invention including a substantially rigid ASIC connected at two ends to a substantially rigid substrate having a cut out;

[0041] FIGS. 9 and 10 schematically illustrate a cross-sectional and top plan view, respectively, of another embodiment of an implantable sensor device of the invention including a substantially rigid ASIC connected at one end to a substrate having a cut out;

[0042] FIG. 11 is a perspective illustration of the implantable device of FIGS. 7 and 8 or 9 and 10, showing an electrical wire and filament core connection between the ASIC and the antenna;

[0043] FIG. 12 illustrates a perspective view of the implantable sensor device of the invention with a cut out located at an edge of the substrate,

[0044] FIG. 13 is a perspective view showing the electrical wire and filament core connection of the FIG. 12 embodiment;

[0045] FIG. 14 is a cross-sectional view that schematically illustrates the electrical wire and filament core connection to a substrate of the invention;

[0046] FIGS. 15 and 16 schematically illustrate yet another embodiment of an implantable sensor device of the invention having a cut out located at an edge of the substantially rigid substrate and a protective barrier wall located at one end;

[0047] FIG. 17 schematically illustrates a cross section view of a further embodiment of the protective barrier wall of the invention;

[0048] FIG. 18 schematically illustrates an integrally formed ASIC and holder of the invention implanted in a wall of the heart during surgery;

[0049] FIG. 19 schematically illustrates another integral ASIC and holder of the invention having a "T"-shaped anchor implanted in a wall of the heart during surgery;

[0050] FIG. 20 schematically illustrates an ASIC and separately formed holder of the invention in the form of an elbow connector holding the ASIC and guiding a slack portion of the connecting wire;

[0051] FIG. 21 is a three-dimensional representation of a holder of the invention that mounts the ASIC in the wall of the heart and includes suture wings that limit movement of the ASIC and serve as suture mounts;

[0052] FIG. 22 is a three-dimensional representation of another embodiment of a holder of the invention where the holder includes an end cap;

[0053] FIG. 23 is a three-dimensional representation of a holder of the invention illustrating how it connects the ASIC via a flexible cable to an antenna;

[0054] FIG. 24 schematically illustrates an implantable sensor device of the invention encased in a biocompatible sheathing;

[0055] FIG. 25 is a side view of the device illustrated in FIG. 24 illustrating how the shape of the sensor may be configured as a football shape to minimize turbulence and reactionary fluid forces in the heart;

[0056] FIG. 26 is a side view of a dual substrate sensor device of the invention showing how the shape of the sheathing may be configured to minimize turbulence and reactionary fluid forces in the heart;

[0057] FIG. 27 is a side view of another embodiment of the invention illustrating how the shape of the sheathing may be configured to minimize turbulence and reactionary fluid forces in the heart; and

[0058] FIG. 28 is a block diagram of the major electronic components of an external reader constructed according to the principles of the invention for telemetrically receiving data from an implanted sensor device.

DETAILED DESCRIPTION

[0059] The embodiments of the invention and the various features and advantageous details thereof are explained more fully with reference to the non-limiting embodiments and examples that are described and/or illustrated in the accompanying drawings and detailed in the following description. It should be noted that the features illustrated in the drawings are not necessarily drawn to scale, and features of one

embodiment may be employed with other embodiments as the skilled artisan would recognize, even if not explicitly stated herein. Descriptions of well-known components and processing techniques may be omitted so as to not unnecessarily obscure the embodiments of the invention. The examples used herein are intended merely to facilitate an understanding of ways in which the invention may be practiced and to further enable those of skill in the art to practice the embodiments of the invention. Accordingly, the examples and embodiments herein should not be construed as limiting the scope of the invention, which is defined solely by the appended claims and applicable law. Moreover, it is noted that like reference numerals reference similar parts throughout the several views of the drawings.

[0060] FIG. 1 schematically illustrates an embodiment of an implantable telemetric measuring device and reader constructed according to principles of the invention providing for continuous or regular intra-cardiac pressure monitoring. A coil or antenna 14 connected with an external reader 12 generates a radio frequency (RF) field in a manner known in the art. The coil 14 and external reader 12 may be fixed to individual belts that wrap around the patient and connect to each other via a standard cable. The RF field induces a current in a coil 18 connected to a substantially rigid sensor device 20, such as described herein, implanted within the heart 16 of the patient 10, such as the septum or the wall of the left atrium, to sense pressure in the left atrium. The sensor device 20 may consist of an application specific integrated circuit (ASIC) such as described herein, having power conditioning circuitry that detects when adequate power is being delivered and switches on sensing, analog-to-digital, and data processing circuits. The data processing circuitry sends the sensor data to the ASIC transmitter, which uses the coil 18 as an antenna. The coil 18 telemetrically transmits, via signal 22, the data to the antenna 14 of the external reader 12. The external reader 12 may provide secure reception and storage of pressure and temperature values, compare the pressure reading of the implanted device 20 to ambient pressure via an internal sensor in the reader, and deliver the intracardiac data to other devices, such as computers, personal digital assistants (PDAs), cell phones, etc., via standard protocols.

[0061] In this embodiment, the external reader 12 may obtain data from the sensor device 20 at continuous or regular intervals. By way of example, the external reader 12 may continuously generate an RF signal to activate the sensor device 20 to obtain pressure and/or temperature readings (in order to describe even the waveform of the blood pressure, if desired by the doctor, the sensor device should take up to 100 or more measurements per second). Alternatively, the external reader 12 may generate an RF signal at regular intervals (e.g., every half hour, once every four hours, once a day) to activate the implanted rigid sensor device 20 to obtain pressure and/or temperature readings.

[0062] FIG. 2 schematically illustrates another embodiment of an implantable telemetric measuring device and reader, which may operate similarly to the FIG. 1 embodiment but provides for on demand intra-cardiac pressure monitoring according to principles of the invention. In this embodiment, a coil 28 in a hand-held reader 26 generates an RF field that induces a current in the coil 18 of the substantially rigid sensor device 20 implanted within the heart 16 of the patient 10, as in the FIG. 1 embodiment. As described above, sensor device 20 may include an ASIC that operates similarly to the FIG. 1 embodiment. Thus, power condition-

ing circuitry in the sensor device **20** detects when adequate power is being delivered, and turns on the sensing, analog-to-digital, and data processing circuits. The data processing circuitry sends the sensor data to the ASIC transmitter, which uses the coil **18** as an antenna. The coil **18** transmits, via signal **24**, the data to the antenna **28** of the hand-held reader **26**. The hand-held reader **26** may be extendable to expose the antenna **28** and provides reception and storage of pressure and temperature values, and compares the implant's pressure reading to ambient pressure via an internal sensor in the readout device. The hand-held device **26** may deliver the intracardiac data to other devices, such as computers, PDAs, cell phones, etc., via standard protocols.

[0063] In this embodiment, the reader unit **26** may obtain data from the implanted sensor device **20** on demand. By way of example, a user may activate and cause the reader unit **20** to generate an RF signal by extending the top portion containing the antenna from the bottom portion of the reader unit housing to activate the implanted rigid sensor device **20** to obtain pressure and/or temperature readings.

[0064] FIG. 3 illustrates a cross-sectional view of the heart area of a patient where the implanted device may be employed, including the left and right atrium and the crossing of veins at the posterior septum. The heart **30** has a right atrium **32** and a left atrium **34**, which are divided by the septum **36**. As described in more detail herein, it may be advantageous to locate and/or anchor the implantable sensor device **20** at the septum **36** separating the right atrium **32** and the left atrium **34**, such that a portion of the sensor **20** extends into the chamber to be sensed, e.g., the left atrium **34**. The implantable device may work as a short-term implant as well as a long-term implant, and may be implanted at the "Waterston's groove" near the access of the pulmonary vein or other locations chosen by a doctor. The implantable sensor device **20** also may be designed to facilitate ready removal of the device if medically necessary. An embodiment of such a removable device is illustrated in FIG. 23.

[0065] FIG. 4 illustrates a substantially rigid sensor ASIC constructed according to principles of the invention for sensing intra-cardiac pressure and temperature in any of the embodiments of the invention. The ASIC **400** contains pressure sensing elements **402**, such as eight passive sensors **404** and eight active sensors **406**, temperature sensor **408**, an analog-to-digital (A/D) converter **410**, data transmission circuitry **412**, power conditioning circuitry including components such as smoothing and resonance capacitors (not shown), a digital state control **414** including a code redundancy check for secure data transmission and memory **416**, such as Electrically Erasable Read-Only Memory (EEPROM) cells, for unit identification, which are components known in the art. An example of the suitable ASIC structure is described in U.S. Pat. Nos. 5,321,989 and 5,431,057, the contents of which are expressly incorporated by reference in their entirety.

[0066] According to the principles of the invention, the ASIC **400** should be an extremely small and lightweight chip to avoid placing undue stress on the heart and/or producing turbulent flow in the heart chamber(s). For example, an ASIC particularly adapted for use in the embodiments described herein as being implanted during open chest surgery should weigh less than fractions of a gram, have a surface area of less than or equal to about 10 mm² per side, and a thickness of about ¼ mm to about 1 mm. In one advantageous embodiment, the ASIC may be about 2 mm wide by 5 mm long by

about 250 to 800 microns thick. Other dimensions may also be used depending upon the particular application or location in the cardiovascular where the sensing will occur and depending upon the delivery method. In general, the dimensions of the ASIC **400** may range from about 3 mm to about 8 mm long, about 0.6 mm to about 2.5 mm wide, and about 0.2 mm to about 1.3 mm high. Other dimensions, such as an ASIC that is substantially square, may also be used.

[0067] In the embodiment of the invention shown in FIG. 4, the ASIC **400** includes sixteen capacitive pressure sensors cells **402**, eight of which are active pressure sensors **406** and provide pressure data, and eight of which are passive pressure sensors **404** and act as an internal reference. The pressure sensor cells **402** may include minute, flexible membranes that are housed within the substantially rigid ASIC structure as shown schematically in FIG. 5. Specifically, the active pressure sensors **406** have flexible membranes **424** and passive pressure sensors **404** have flexible membranes **426**. The membranes **424** of the active pressure sensors **406** are distortable based on the level of cardiac blood pressure. The distortion may be mainly in a direction generally perpendicular to the planar top surface of the ASIC **400**. The distortion may be determined based on capacitive measurements or by use of distension measuring tapes. By way of one specific exemplary embodiment, the sixteen capacitive pressure sensing elements **402** of the ASIC **400** may each be about 96 microns in diameter.

[0068] As the pressure changes in the heart, the capacitance measured in the pressure sensors **402** changes. The pressure sensors **402** generate signals based on the change in capacitance, and thus indicative of the pressure in the heart. As will be described below, the signals preferably are processed by components located in or on the ASIC **400** and transmitted to an external reader.

[0069] Thus, the blood pressure measuring process may be a capacitive pressure measurement process via measuring membranes **424**, **426** that are integrated into the ASIC **400**, such as the planar top surface of the chip as shown in FIG. 5. The ASIC **400** may have a substantially inflexible, substrate made of silicon that cannot be folded or rolled up. The thin, but mechanically inflexible substrate creates a mechanically stable device providing a substantially rigid structure to house the measuring membranes **424**, **426** as shown in FIG. 5. Changes in the geometry of the ASIC **400**, such as twisting due to blood turbulences, may be avoided due to this substantially rigid, chip-based configuration, even when the ASIC **400** is exposed to turbulent, blood flow. Thus, the implanted ASIC **400** provides a durable device capable of withstanding the internal environment of the heart and other locations in the cardiovascular without producing dangerous stresses within the heart.

[0070] Numerous small membranes **424**, **426** having relatively small dimensions (e.g., a diameter of less than 0.2 mm) may be used as capacitive pressure sensors. Such small dimensions may result in membranes **424** that are less vulnerable to mechanical forces, such as the force of blood flow within the heart, and therefore more reliable.

[0071] The ASIC **400** contains mechanical and electrical elements that are subject to wear and need drift compensation to obtain measurements of suitable quality and reliability for their intended cardiovascular uses. Drift in a sensor may occur as time passes and physical properties of the structure change. Over time and usage, changes in electronics in a chip may effect the measurements. Further, when a pressure sen-

sor uses a membrane, the membrane may sag in the middle as it ages. The capacitance at the pressure sensor membrane varies based on the change in position of the pressure sensor membrane. These changes, unrelated to the change in blood pressure, may alter the true value of the measurements being sensed. Drift compensation is particularly important in an intra-cardiac long-term pressure sensor.

[0072] The drift compensation scheme employed in ASIC 400 should reduce or eliminate the effects of the change in the physical properties of the ASIC 400. According to the principles of the invention, the drift of the pressure values obtained from the sensor structure 400 may be minimized to a value of about 5.0 mmHg/year to about 2.5 mmHg/year or even smaller than 1 mmHg/year.

[0073] In accordance with drift compensation principles of the invention, a plurality of active sensors 406 and a plurality of passive sensors 404 are provided, such as eight of each. According to an embodiment of the invention, the structure of the active sensors 406 and the structure of the passive sensors 404 are identical. However, as illustrated in FIG. 5, the membranes 424 of the active sensors 406 are open to the sensing environment (e.g., a heart chamber) for sensing pressure, while the membranes 426 of the passive sensors 404 are isolated from the environment, e.g., by placing a glass layer 428 or other suitable material over the surface of the membrane 426 so that pressure in heart will not affect the passive sensors 404. Both the active sensors 406 and passive sensors 404 are affected substantially the same by age, usage and sagging and any other effects of the environment. Using the passive sensors 404, the ASIC 400 may determine how much of the change in position of the pressure sensor membrane 426 is effected by the age and sagging. The change in capacitance based on the change in position of the passive pressure sensor membrane 426 is determined. This amount is then used to offset the change capacitance measured in the active pressure sensor membrane 424. This system allows the change in capacitance due the pressure within the heart to be more accurately determined. Compensating for drift may allow a doctor or patient to better determine short term (e.g., days, weeks) trends in pressure within the patient, such as the heart.

[0074] The implantable sensor device, which may include the ASIC 400, a connector and an antenna, may be completely encapsulated within a seamless biocompatible sheathing (not shown in FIGS. 5-6). The material areas around the measuring membranes 424 maintain their flexibility after encapsulation to allow transmission of the pressure to the measuring membranes 424. The biocompatible sheathing will be described in greater detail below.

[0075] FIG. 6 is a cross-sectional view of the ASIC 402 of the invention with a gel or fluid between a sheathing and the active pressure sensors 406. As described above, a glass substrate 428 or other suitable material isolates the passive pressure sensors 404. A liquid or gelatinous pressure transmitting medium 432 is used between sheathing 430 and the active pressure sensors 406. As will be described below, this liquid or gelatinous medium 432 may improve the measurement or reception of blood pressure values within the chamber to be sensed, e.g., within the left atrium. Even though fibrous tissue or plaque may grow in the area of the implant over time (e.g., months or years after the implantation), encapsulating the pressure sensors within a separate gel-filled membrane may allow reliable measurement values to still be obtained.

[0076] For example, endothelialization may result in endothelia being deposited on the surface of the sensor structure. If endothelia and/or plaque are deposited on the surface of one of the active pressure sensors, or on the biocompatible sheathing at the surface of one of the active pressure sensors, pressure measurement readings may be adversely affected. One way to reduce such an effect is to coat the sheathing and/or sensors with a drug, e.g., heparin, to reduce or eliminate endothelia. However, such treatments may not always be effective.

[0077] Thus, as illustrated in FIG. 6, the surface of the active pressure sensors 406 are coated with a gel or fluid 432 and encapsulated in the membrane 430. In this manner, endothelial growth or plaque on the membrane 430 directly over the surface of one of the active pressure sensors 406 will have a reduced or negligible effect on the pressure sensor measurement, as the pressure is transmitted via the endothelial growth and the membrane 430 through the gel/fluid 432 to the active pressure sensors 406. Further, plaque growth and/or endothelialization on the entire surface would still allow pressure sensing measurements to be obtained, as the pressure exerted on the endothelia is transmitted via the gel/fluid 432 to the active pressure sensors 406. In particular, the gel/fluid filled membrane 430 may function to integrate the change in pressure over a larger area than the individual active pressure sensors 406 themselves. This minimizes the effects of endothelialization and/or plaque adherence to the sheathing 430. Although sheathing 430 is shown as only covering the gel/fluid 432, it is understood that the sheathing 430 or other sheathings could cover part or all of sensor device 400, as described below.

[0078] As described above, the ASIC 400 includes an A/D converter 410. As is known in the art, the pressure sensors 402 provide analog signals indicative of the pressure in the heart. The A/D converter 410 converts the signals from the pressure sensors 402 to digital signals.

[0079] Thus, the transmission and digitizing of measurement values into appropriate signals in the invention is preferably carried out within or very closely adjacent to the heart chamber or chambers to be sensed, such as the left and/or right atrium and/or the left or right ventricle, and most preferably are processed inside the ASIC 400. Using a fully digital system may result in greater accuracy of the readout. In an analog system, where the amplitude of the signal is proportional to the pressure reading provided by the sensors, the value of pressure recorded by an external reader depends upon the distance between body and reader. As a result, either the distance from the body to reader must be very tightly controlled, or the accuracy of the system will suffer. According to the invention, the distance from body to reader has little or no effect on the pressure value measurement received due to the use of a digital signal and to processing the signals at or very near the sensor. This may make the system more robust and accurate than analog systems.

[0080] In addition, the fully digitized data can be handled for more easily by data transmission systems, making the external readers compatible with computer, Internet and telemedicine interfaces. For example, highly accurate pressure sensors and a 9-bit analog-to-digital converter may impart high resolution to the sensing systems, where an accuracy of about +/-2 mm Hg or less may be achieved.

[0081] Further, digitization at the ASIC 400, as opposed to analog signal transmission via an antenna before digitization, may avoid interference issues from other, unrelated RF

sources. In prior devices, analog signals are sent from the sensor to the antenna structure via a wire. By processing and converting the analog signals to digital signals prior to transmission over the wire to the antenna, the system may avoid analog interference that may be induced in the wire by external RF signals and noise, such as radio broadcasts, electronics, and the like.

[0082] The ASIC 400 measures pressure at the pressure sensing elements 402 and transfers the absolute pressure signals to an external reader. A pressure value is calculated from the difference of absolute pressure value, measured with the ASIC 400, and the atmospheric pressure surrounding the patient as is well-known in the art. This atmospheric pressure may be measured within the external reader, which is normally in the surrounding environment of the patient.

[0083] The operation of the ASIC 400 is based on the interaction between a connected antenna, such as shown in FIG. 23, and an external reader according to well-known principles of transponder technology. Therefore, no internal power source is required. The ASIC 400 and the external reader may be tuned so that continuous measurements, e.g. up to 120 single measurements per second, may be processed and transmitted. As described above in FIG. 1, the total system may be programmed so that measurements are taken and stored in given intervals or at defined time periods. Retrieval, monitoring, and recording of data may be possible at any time.

[0084] According to an embodiment of the invention, the ASIC 400 preferably consists of a single integrated chip. All relevant functions and components for the measuring process, digitizing, identification number transmission, power supply, and telemetric data transmission are integrated into the single integrated chip. As described above, the ASIC 400 may contain a specific identification number, as well as a chip specific calibration file and further circuit and storage components. Alternatively, the circuit components may also be placed on two or more chips, e.g. if sensing in separate locations is desired.

[0085] The ASIC 400 may be formed from a single complementary metal oxide semiconductor ("CMOS") chip to produce a smaller implantable device than with other methods, and help minimize power use and maximize measurement accuracy reliability. Since the consumption of power produces heat, minimization of power may be desirable in implantation applications. In a one-chip solution, the ASIC 400 may be highly resistant to mechanical or electrical interference from the outside, as there is no interaction between multiple chips.

[0086] The power consumption of the chip may be low, so that if an increase of temperature occurs in the course of inductive/transponder related power insertion, difficulty in measuring or data transmission may be reduced or avoided. The optimized circuit design may result in a very low power consumption, such as only about 210 microwatts at about 3 volts DC. The sampling rate may be about 20 to about 120 Hz. The high integration factor of the logic circuit combined with the high speed of data transmission may allow the use of a very secure data transmission protocol, thereby addressing concerns of the regulatory authorities.

[0087] An integrated temperature sensor 408 may be provided in the ASIC 400 to allow for temperature sensing as shown in FIG. 4. The temperature sensor 408 may use the circuit in the ASIC 400 and base the temperature measurement on current characteristics within the circuit, thereby determining the temperature in the heart based on the tem-

perature based current characteristics within the ASIC 400. Each ASIC 400 may be individually calibrated to determine its current characteristics (magnitude, frequency, etc.) at a given temperature (e.g., body temperature). As the temperature changes, the current characteristics within the ASIC 400 change. Using the information on the current characteristics and the specific calibration determination for the ASIC 400, the temperature at a particular time can be determined based on current characteristics at that time. The raw pressure data must be corrected for temperature and other external and/or internal influences, and calibration information, such as a calibration curve of the embedded chip, may be established for each ASIC 400 or system that implements an ASIC 400. Each ASIC may have a unique identification number to facilitate calibration and use of data as discussed below.

[0088] The ASIC 400 includes a data memory 416, such as the EEPROM cells, in which the unique identification number may be stored. This identification number is transmitted telemetrically together with the measurement values. The identification number may be used to determine the appropriate calibration information for an ASIC 400. Also, a single external reader may then be used to interrogate multiple implanted ASICs, as described below.

[0089] The unique identification number may be transmitted along with the sensor data to the external reader to allow the external reader to use the correct calibration information to calculate pressure and/or temperature. An external reader (as described in greater detail below), may have a memory to store calibration information for a number of ASICs 400 or systems that implement ASICs 400. The appropriate calibration information is associated with the appropriate ASIC 400 or system via the identification number. With the identification number, or other identification indicia, the external reader accesses the calibration information associated with the particular ASIC 400 or system that includes the particular ASIC 400. The data received by the external reader is processed using the appropriate calibration information to achieve more accurate results.

[0090] Each ASIC 400 and/or system also may be zeroed prior to implantation. When inside the patient, the system compares the measured pressure to the pressure in a vacuum. Outside the patient, the external reader compares the ambient pressure to the pressure in a vacuum. Pressure inside the heart is calculated by comparing the difference between the pressure measured inside the heart and the pressure measured outside the patient. Zeroing the ASIC 400 or the system may involve using the ASIC 400 system to measure the pressure outside the patient and comparing this measurement to the pressure obtained by another external device. The difference between these two readings may be stored with the calibration information associated with the ASIC 400 or system and used to adjust future pressure measurements by the ASIC 400 or system once it has been implanted to account for the difference.

[0091] Using one or more transponder coils, an external reader may be used for the power supply of the ASIC. This unit also may be used for telemetric data acquisition. The range for telemetric power supply and data transmission may be from about 3 cm to about 35 cm or other ranges as can be readily determined by a skilled artisan. This range also may depend on the distance between the external reader and the implanted antenna and the size of the antennas.

[0092] Measurement data are processed and preferably are digitized on the ASIC 400 for transmission from the sensor

structure to the interior transponder coil. The transmission of the measurement data from the ASIC 400 to the interior transponder coil may be realized via one or more electric conductors, preferably designed as flexible thin wires, embedded in silicone on other nonconducting material. Measurement data are transmitted telemetrically from the interior transponder coil to the external reader. The external reader capacities may be designed for an exterior supply of all power resources which are required for the continuous operation of ASIC 400, including measurements and data transmission.

[0093] The ASIC 400 also includes a bi-directional power circuitry 424 for working with the reader to evaluate the strength of the signals sent between the reader and the ASIC 400. The components in the bi-directional power circuitry 424 interact with a reader to ensure that appropriate signal strength and data transmission is achieved. The interaction between the bi-directional power evaluation module 424 and the reader is described in greater detail below with respect to FIG. 28.

[0094] FIGS. 7 and 8 schematically illustrate an embodiment of an implantable sensor device 700 of the invention including a substantially rigid sensor chip connected at two ends to a substantially rigid substrate 708 having a cut out. A sensor chip 702, such as ASIC 400, includes pressure sensing membranes 704 and four spaced chip bond pads 706. A substantially rigid substrate 708 having an aperture 710 and bond tracks 712 connected to bond pads 706 are also provided. The substrate 708 is configured in a spaced apart relationship to the sensor chip 702. More particularly, the aperture 710 of the substrate 708 is located substantially opposite of the capacitive pressure membranes 704 of the sensor chip 702 so pressure from the blood surrounding the device may be transmitted readily to the pressure membranes 704. A pressure transferring material (not shown) may be located at the aperture 710 to ensure that pressure from the blood is transferred to the pressure membranes 704.

[0095] The sensor chip 702 and the substrate 708 may be configured in a fixed relationship, so that the distance, or offset, between the sensor chip 702 and the substrate 708 does not change. The chip bond pads 706 may be connected to the substrate bond pads 712 to fix the distance between the sensor chip 702 and the substrate 708. As shown in the embodiment of FIGS. 7 and 8, the sensor chip 702 and the substrate 708 both have four bond pads. However, it is understood that other amounts of bond pads may also be used.

[0096] At least one of the substrate bond pads 712 may be elongated in the form of a track to facilitate connection to an electrical wire 714 that connects to an antenna (not shown). Electrical wire 714 is connected to the substrate bond pad 712 by any conventional method, such as by using heat and pressure. Connecting the electrical wire 714 to a substrate bond pad 712, as opposed to being directly connected to chip 702, may reduce or eliminate damage to or malfunction by the sensor chip due to the connection process. The electrical wire 714 is electrically connected to the sensor chip 702 via the electrical connection between the substrate bond pad 712 and the chip bond pad 706.

[0097] FIGS. 9 and 10 schematically illustrate another embodiment of an implantable sensor device including a substantially rigid sensor chip connected at one end to a substantially rigid substrate having a cut out. The device 900 of FIGS. 9 and 10 has similar components and operation to the device 700 illustrated in FIGS. 7 and 8. However, device 900 has chip bond pads 906 located in generally close prox-

imity to each other at one end of the sensor chip 902. In addition, the substrate bond pads 912 are generally located in close proximity to each other on the substrate 908. When the chip bond pads 906 and the substrate bond pads 912 are connected, the sensor chip 902 and the substrate 908 are fixed at one end, with the other free end being supported in a cantilevered manner. This arrangement of chip bond pads 906 and substrate bond pads 912 may reduce stress on the sensor chip 902, as changes in the size of the substantially rigid substrate 908, such as due to thermal expansion, may have less of an effect on the sensor chip 902 due to the location of the chip bond pads 906 on the sensor chip 902.

[0098] The device 900 may further include a flexible filler material 916 located between the sensor chip 902 and the substrate 908. As shown, the filler 916 may be located throughout the area between the sensor chip 902 and the substrate 908 except at the aperture 910 that is opposite the capacitive pressure membranes 904. Filler 916 may be any flexible material that can provide support to reduce or eliminate movement in the offset direction between the sensor chip 902 and the substrate 908. The filler 916 may be the same material used to surround the implanted device 900, such as a biocompatible material like silicone or other similar material.

[0099] FIG. 11 is a perspective illustration of an implantable sensor device such as the FIGS. 7 and 8 or FIGS. 9 and 10 embodiments showing the electrical wire and core filament connection to the ASIC and antenna. The device 1100 includes a substantially rigid sensor chip 1102 having pressure membranes 1104, and a substantially rigid substrate 1108 with an aperture 1110 exposing the pressure membranes 1104. A pressure transmitting material 1112, such as a liquid or gelatinous material, is located within the aperture 1110 to transmit pressure from the blood to the pressure membranes 1104. The entire device 1100 is enclosed by a biocompatible sheathing 1106, such as silicone. In addition, the sheathing 1106 can also be used as the pressure transmitting material 1112 within the aperture 1110.

[0100] Substrate 1108 may further include connector holes 1120 for facilitating attachment of an antenna connector 1114 to the substrate 1108 and the sensor chip 1102. The connector 1114 includes electrical wires 1116 and a filament core 1118, such as nylon. Electrical wires 1116, which may be formed of gold cable, or other appropriate material, provide an electrical connection between the sensor chip 1102 and an antenna (not shown). Electrical power from the antenna may be conducted via the electrical wires 1116 to the sensor chip 1102 for powering the sensor chip 1102 to obtain measurements. Signals, such as pressure measurements and identification indicia, may be transmitted over the electrical wires 1116 from the sensor chip 1102 to the antenna for transmission to a reader. The filament core 1118 provides strength to the connector 1114 to reduce or eliminate strain on the connection between the substrate bond pad (not shown) and the electrical wires 1116. The filament core may be made of nylon or other similar, synthetic flexible material that does not conduct electricity and has a low coefficient of thermal expansion. This connection will now be described in greater detail below with reference to the examples of FIGS. 12-14.

[0101] FIGS. 12 and 13 illustrate an implantable sensor device 1200 with a cut out located at an edge of the substantially rigid substrate, including a cable and core filament connection, while FIG. 14 schematically illustrates the cable and core filament connection to the substrate. The implantable device 1200 includes a substantially rigid sensor chip

1202 having pressure membranes **1204**. In this embodiment, the capacitive pressure membranes **1204** are located near the edge of one side of the sensor chip **1202**. The device **1200** further includes a substantially rigid substrate **1208** having connector holes **1220** and a cut out **1210** opposite of the pressure membranes **1204**. A pressure transmitting material **1212** is located within the cut out **1210** to transmit pressure from the blood to the pressure membranes **1204**. The device **1200** is surrounded by a biocompatible sheathing **1206**, such as silicone. According to a preferred embodiment of the invention, the pressure transmitting material **1212** may be the same as the sheathing material **1206**.

[0102] The device **1200** further includes a connector **1214** which includes electrical wires **1216** and a filament core **1218**. The electrical wires **1216**, which may be formed of gold, or any other suitable similar material, connect to substrate bond pads **1222**, and the substrate bond pads **1222** are connected to chip bond pads **1224**. This results in an electrical connection between the electrical wires **1216** and the sensor chip **1202**. The filament core **1218** may be attached directly to the substrate **1208**, such as by an adhesive. As shown in FIG. 14, the filament core **1218** is threaded through the connector hole **1220** for attachment to the substrate **1208** such that the electrical wires **1216** have extra slack when the filament core **1218** is pulled straight. This configuration may reduce or eliminate the strain on the connection between the electrical wires **1216** and the substrate bond pad **1222** when there is movement of either the connector **1214** or the substrate **1208**. Other methods for strain relief on electrical wires **1216** may also be used.

[0103] FIGS. 15 and 16 schematically illustrate an implantable sensor device with a cut out located at an edge of the substrate and a protective barrier wall located at one end of the substrate. The device **1500** has a substantially rigid sensor chip **1502** having capacitive pressure membranes **1504** (shown in FIG. 16). The device **1500** further includes a substantially rigid substrate **1508** having a cut out **1510** located substantially opposite the pressure membranes **1504**. A chip bond pad **1518** on the sensor chip **1502** is connected to a substrate bond pad **1520** of the substrate **1508** in a conventional manner. The device **1500** is encapsulated in a biocompatible sheathing **1506**.

[0104] The substrate **1508** includes a barrier wall **1514** that may be substantially perpendicular to the plane of the substrate **1508**. The height of the barrier wall **1514** may be such that the top of the barrier wall **1514** is at or above the top of the sensor chip **1502** when it is attached to the substrate **1508**. The barrier wall **1514** may provide additional protection to the chip sensor **1502**, such as preventing the sharp ends of the chip **1502** from wearing or puncturing the sheathing **1506**. In addition, a front portion **1512** of the substrate **1508** shaped like an arrow is located beyond the barrier wall **1514** and is tapered to reduce or eliminate the effects of blood turbulence on the chip sensor **1502**, as well as aid in the implantation of the device **1500** within the heart. This may occur when the tapered portion **1512**, and thus the device **1500**, is inserted into the heart. The edges of the barrier walls may be slightly rounded (not shown in the drawings) to avoid any wearing or puncturing of the sheathing. Although not shown, it is understood that a pressure transmitting material and/or a filler material may be used with the device **1500**.

[0105] FIG. 17 schematically illustrates a cross sectional view of a further embodiment of a barrier wall formed as an end cap at one end of the sensor device. The device **1700** has

a sensor chip **1702** having capacitive pressure membranes **1704**. The device **1700** further includes a substrate **1708** having a cut out **1710** located substantially opposite the pressure membranes **1704**. A chip bond pad **1718** on the sensor chip **1702** is connected to a substrate bond pad **1720** of the substrate **1708** in a conventional manner. The device **1700** may also be encapsulated in a biocompatible sheathing (not shown in FIG. 17).

[0106] The substrate **1708** may include a barrier wall **1714** that is substantially perpendicular to the plane of the substrate **1708**, as in the prior embodiment. The height of the barrier wall **1714** may be such that the top of the barrier wall **1714** is at or above the top of the sensor chip **1702** when it is attached to the substrate **1708**. In addition, the barrier wall includes a top cover **1716** extending inwardly from the top of the barrier wall **1714** substantially parallel to the substrate **1708** over the sensor chip **1702** to provide protection to the top of the sensor chip **1702**. Although shown in FIG. 17 as extending over only a small area of the sensor chip **1702**, it is understood that the barrier top cover **1716** could extend further, including along the entire length of the sensor chip or beyond. The barrier wall **1714** and the barrier top cover **1716** may provide additional protection to the chip sensor **1702** and the sheathing as discussed above. More specifically, the barrier top cover **1716** may inhibit damage to the sheathing **1706** that could occur by rubbing of sharp edges of the chip sensor **1702** against the sheathing material **1706**. In addition, a front portion **1712** of the substrate **1708** is located beyond the barrier wall **1714** and may be tapered to reduce turbulence, as well as aid in the implantation of the device **1700** within the heart, also as discussed above.

[0107] FIG. 18 schematically illustrates an integrally formed ASIC and holder of the invention in a greatly enlarged scale implanted in a wall of the heart after surgery. The implantable sensor device of this embodiment may include a substantially rigid substrate **1802** including an ASIC **1804**, which is located at one end of the substrate **1802**. The substrate may be elongated such that a holder portion **1810** is integrally formed therewith and anchored within a wall of the heart such that the ASIC is at least partially exposed within the chamber of the heart to be sensed. At a minimum, the active pressure sensors must be exposed. The other end of the substrate **1802** is the holder portion that is affixed within a heart wall **36**, such that part of the holder portion **1810** of the substrate **1802** is located on the other side of the heart wall **36**. A wire **1808** may also be affixed to the substrate **1802** and connect the ASIC **1804** to an antenna (not shown). The holder portion **1810** of the substrate **1802** may be configured to form part of an anchor structure. For example, the outer portion may include a bend, elbow, or other configuration to facilitate anchoring with or without separate affixing means known in the art, such as sutures or the tobacco pouch suture **1806** schematically shown in FIG. 18 surrounding the exposed end of the holder portion **1810**.

[0108] FIG. 19 schematically illustrates on a greatly enlarged scale another integral ASIC and holder of the invention having a "T" shape anchor implanted in a wall of the heart. In this embodiment, the substantially rigid substrate **1902** includes an ASIC **1908** and an integral holder **1910** having an anchor **1906** such that the substrate **1902** and the anchor portion **1906** are configured in a generally "T" shape, with the free end extending through the heart wall. An antenna **1912** may be attached directly to the anchor **1906** and/or substrate **1902**. The antenna **1912** may be in the form of a coil

wrapped around the T-shaped end of the holder **1910** and connected to the ASIC **1904** by a wire **1908** located on the substrate **1902**. This may reduce the distance of the wire **1908** required or essentially eliminate it.

[0109] FIG. 20 schematically illustrates an ASIC and a holder of the invention in the form of an elbow connector for holding the ASIC and guiding a slack portion of the connecting wire. As illustrated in FIG. 20, an elbow or bend, separately connected to or integrally formed with the substrate **2002**, may be used to connect and/or guide the wire **2006** to an ASIC **2004** held at the other end of the elbow connector. The wire **2006** may be connected in such a way as to minimize the movement of the wire **2006** at the connection point **2008**. As shown, the ASIC **2004** is attached to the elbow **2002**. The wire **2006** is attached to both the ASIC **2004** (at **2008**) and the elbow **2002** (at **2010**). The connection of the wire to the elbow **2002** at **2010** may absorb the stress of wire movement, thereby ensuring the fail-safe connection between the wire **2006** and the substantially rigid sensor chip **2004** through the numerous cycles anticipated during the life of a patient. While various embodiments of the invention show a holder with an anchor, separate anchor structure may not be necessary, as the heart wall naturally closes around the sensor device.

[0110] The wire **2006** connects the ASIC **2004** to the antenna (not shown) and may be made of flexible material that allows a signal to be transmitted between the sensor structure and the antenna structure. The wire may be made of gold, platinum, iridium, stainless steel, spring steel, or similar material. The wire is totally embedded (surrounded) by a flexible mantle.

[0111] As illustrated in FIG. 20, it may be desirable for wire connection to have some predetermined amount of slack in the wire to account for the movement of the heart without placing undue stress on the components. Thus, the slack may take the form of a bend or loop in the wire **2006** at or near the connection **2010** of the wire **2006** to the bend **2002**, as shown. This slack will avoid excessive bending and/or stress on the wire **2006**. Slack may also reduce or eliminate additional strain on the heart, as there is little or no increased effort necessary for normal heart pumping due to the sensor implant. Other configurations to provide slack, reduce bending or stress in the wire **1006** and reduce strain on the heart may also be used.

[0112] FIG. 21 is a three-dimensional representation of a holder of the invention that mounts the ASIC in the wall of the heart and includes suture wings that limit movement of the ASIC, e.g., during implantation by a doctor, and serve as suture mounts. The implantable sensor device **2100** includes a ASIC **2102** having capacitive pressure membranes (not shown) formed on a substantially rigid substrate **2108** having a cut out **2110** substantially opposite the pressure membranes as shown and described above. A connector **2114** having electrical wires **2116** and a filament core **2118** is also provided. The device **2100** also includes a holder **2112** (shown in dashed lines) for supporting the ASIC **2102** and the substrate **2108** and mounting it to the wall of a heart. A receiving portion **2120** of the holder surrounds and fixedly receives a portion of the substrate **2108** such that cut out **2110** is exposed. When the device **2100** is inserted into the heart wall, transversely extending flange **2122** forms a pair of suture wings that act as stops to ensure that the device **2100** is not inserted too far. In addition, the suture wings **2122** may be used in conjunction with sutures for anchoring the device **2100** to the heart. While two wings are shown in FIG. 21, the

flange **2122** could be formed with three or more wings if desired. The holder **2112** may also include a connector attachment portion **2124** to provide a conduit for connecting the wires **2116** and core **2118** to the sensor chip **2102** via the substrate **2108**.

[0113] FIG. 22 is a three-dimensional representation of another embodiment of a holder of the invention which is similar to the FIG. 21 embodiment but includes an end cap **2224** to provide further protection to the sensor chip **2202** and substrate **2208**, such as when the device **2200** is inserted through a heart wall.

[0114] FIG. 23 is a three-dimensional representation of a holder of the invention illustrating how it connects the ASIC via a flexible cable to an antenna. The implantable device **2300** generally includes a holder **2306** receiving an ASIC **2302** formed on a substantially rigid substrate **2304**. A flexible connector **2308** electrically connects the ASIC **2302** and an antenna **2310**. The entire device **2300**, including the holder **2306**, connector **2308**, and antenna **2310**, is encapsulated in a biocompatible sheathing, such as a seamless sheathing of silicone.

[0115] The antenna **2310** serves both as power transducer and antenna, and may be fabricated of any type of conductive metal. The antenna **2310** may be made of pure gold, or any other suitable material, to provide both biocompatibility and the necessary degree of electrical conductivity. According to a preferred embodiment of the invention, the antenna **2310** and the connector **2308** may be made of the same material, and the connector **2308** may be part of the antenna **2310**, e.g., the connector **2308** and the antenna **2310** are integrally formed. The antenna **2310** may be extremely thin (for example, about 25 microns) and light weight. All components of the implantable device **2300**, including the antenna **2310**, the connector **2308**, and the holder **2306**, may be very small and light weight to avoid strain and irritation of the heart when implanted. Thus, by of example, the holder **2306** may be made of a light weight plastic, and coils in antenna **2304** and wires in connector may be made of a relatively thin and lightweight wire material, such as thin gold or other suitable materials.

[0116] The number and size of the coils of the antenna **2310** may be dimensioned in such a way that an appropriate telemetric range between the internal and external coils is achieved. For the embodiment shown in FIG. 23 in which the antenna is fixed subcutaneously, the minimum range for the transmission of measurement data from the internal coil to the extra corporal emitter/receiver unit is about 2 cm to about 25 cm. If the antenna is fixed at or near the heart, the range may be closer to the high end, i.e. about 25 cm. However, this required range may change based on the position of the antenna.

[0117] As described above, all energy which is required for the acquisition of measurement data may be provided telemetrically by the antenna **2310**, where the coil may be designed as a passive coil. Examples of suitable coils are illustrated in German patents DE 199 45 879 A1 or DE 101 56 469 A1. By way of example, the antenna **2310** may be formed from a cable that is wound into a plurality of coils. The cable may also connect the antenna **2310** to the sensor chip.

[0118] FIG. 24 schematically illustrates an implantable sensor device of the invention encased in a biocompatible sheathing. A sensor device **2400** includes an ASIC **2404** positioned on a substantially rigid substrate **2402** and encapsulated in a biocompatible sheathing **2406**. The sensor system,

i.e., the ASIC **2404**, substrate **2402**, cable (not shown) and antenna (not shown) may be encapsulated in a biocompatible sheathing such as, silicone, polyurethane or other suitable material. The encapsulation of the system preferably is seamless, i.e., has no break or seam. This reduces or eliminates the risk of contamination or damage to the sensor system structure by fluids within the body. By way of example, the thickness of the encapsulation may be in the range of about 0.01 mm to about 0.8 mm. A seamless sheathing may be obtained by seamless molding or by dipping the entire sensor device **2400** (sensor ASIC, cable and antenna) into the biocompatible material.

[0119] The implanted device may be positioned in heart to minimize turbulence of the blood flow within the heart chamber and reactionary forces. As illustrated in FIG. 25, the sensor device **2400** may be orientated such that its shortest side, such as side **2412**, may be positioned to be in the most upstream position in the blood flow path **2416**. This presents the minimum area in the blood flow and reduces and/or minimizes currents and reactionary forces caused by the implanted device **2400**. This positioning may be done regardless of location of the pressure sensors **2408** on the chip. As illustrated, the longer sides **2410** of the sensor device **2400** containing the top surfaces of the capacitive pressure membranes **2408** may be parallel to the blood flow **2416**.

[0120] As illustrated in FIG. 25, the shape of the sheathing surface **2414** also may be curved or shaped, e.g., similar to the football shape shown in FIG. 25, to further reduce the turbulence caused by blood flow **2416** around the sensor device. The biocompatible sheathing **2408** may be applied to the implanted device **2400** to form the curved surface **2414**. Such curves or other shapes may be designed to minimize hydrodynamic forces.

[0121] The encapsulation in a fully biocompatible material, such as silicone may result in very little change in the sensitivity of the pressure sensor. Further, a small offset due to the influence of the encapsulation material may be compensated for during calibration. This may allow, for example, measurements of about ± 2 mm Hg or less.

[0122] FIG. 26 is a side view of a dual substrate sensor embodiment of the invention illustrating how the shape of the sheathing may be configured to minimize turbulence and reactionary fluid forces in the heart. The implantable device **2600** includes a substantially rigid ASIC **2602** having capacitive pressure membranes **2604** and a substrate **2608** with an aperture **2610** substantially opposite the pressure membranes **2604**. As illustrated, the device **2600** is encapsulated in a sheathing **2606**. The sheathing **2606** may be made of a biocompatible material, such as silicone, that is flexible so that the pressure from the blood may be transmitted to the pressure membranes **2604**. A pressure transmitting material (not shown) may be placed in the aperture **2610** to aid in transmitting the pressure. The sheathing **2606** may be shaped, such as an oval or football configuration, to reduce or eliminate hydrodynamic forces from the blood flow **2612**. Again, the smallest sides of the device are orientated into the upstream portion of **2112** of the blood flow.

[0123] FIG. 27 is a side view of another embodiment of the invention illustrating how the shape of the sheathing may be configured to minimize turbulence and reactionary fluid forces in the heart. The device **2700** has the same components as the FIG. 26 embodiment and is like-numbered. However, in this embodiment, the sheathing **2706** may be shaped, such

as in a rounded triangle configuration, to reduce or eliminate hydrodynamic forces from the blood flow **2712**.

[0124] FIG. 28 is a block diagram of the major electronic components of an external reader **2800** of the invention for telemetrically receiving data from an implanted sensor device. An antenna **2802** may be integrated with the external reader **2800**, as shown schematically in FIGS. 1-2. Also, as illustrated in FIG. 1, the external reader **12** may be attached to the patient, or alternatively, as illustrated in FIG. 2, the external reader **26** may be incorporated inside a handheld device. In other cases, the antenna **2802** may be attached to the patient's body or connected to the external reader **2800** by cable connections. Another method attaches the external reader to the patient's bed or seat. Numerous such arrangements may be employed to adapt to the particular application, as the skilled artisan will recognize.

[0125] The antenna **2802** is used for receiving data, in the form of digital signals, from the implanted sensor device. The digital signals are received at the RF receiver **2806** via an RF generator **2804**. The RF generator **2804** generates an RF signal to be transmitted via the antenna **2802** to power the implanted device.

[0126] The digital data signals received by RF receiver **2806** are processed by filter **2808** and demodulator **2810** before being received and then processed as appropriate by controller **2812**. A bidirectional power module **2828**, described in detail below, is connected between the filter **2806** and demodulator **2810**. Separate memory devices, such as calibration memory **2822**, data memory **2824**, and parameter memory **2826**, may be provided and communicate with controller **2812**. The calibration memory **2822** stores calibration information associated with a particular ASIC sensor system, and the calibration memory **2822** may store calibration information for a number of different ASIC sensor systems. Calibration information may be obtained from an external source, such as a computer, through communications port **2816**. The appropriate calibration information, based on a unique identification number of the ASIC sensor system being interrogated, is obtained from the calibration memory **2822**. Thus, a medical professional, such as a doctor or nurse, can use one reader to obtain pressure readings from multiple patients.

[0127] The data memory **2824** stores data related to the pressure and/or temperature received from the ASIC of the implanted sensor device. The data may be stored in the data memory **2824**, and then transferred, via a data memory module **2814**, to another device through data communications port **2816**, such as a computer. Using information obtained from an atmospheric pressure module **2820**, the controller **2812** uses the data received from antenna **2802** and stored in data memory **2824** to determine the pressure within the heart, as is known in the art. Using the information from the ambient temperature module **2818**, the controller **2812** also uses the data to determine the temperature within the heart, as is known in the art.

[0128] The pressure and temperature calculations, which are performed in controller **2812**, as well as the data from the implanted sensor device, may be stored in the data memory module **2814**. These calculations and data may then be communicated to another device, such as a computer through communications port **2816**.

[0129] The pressure sensor readings, and the parameter alerts described below, may be displayed by the reader on a display (not shown), such as an LCD display or the like. The measured pressure values and parameter alerts also may be

displayed on a monitor of the external reader (not shown) and recorded in an appropriate storage device. The system may be equipped for purposes of telemedicine, so that data is transmitted from the external reader to a medical department or healthcare provider via wire connection, telephone, internet or any other suitable telecommunication source is possible, via known wired or wireless protocols.

[0130] The parameter memory **2826** stores parameter thresholds. The data received from the implanted sensor device is compared by the controller **2812** to the parameter thresholds. If the data fails to meet a particular threshold, or exceeds a particular threshold, an alarm may occur to alert a user. The threshold parameters may be set by a doctor or other health care professional. By way of example, the threshold range for pressure may set to 25 to 30 mm Hg, depending on the patient. If the reader receives a measurement of 25 mm Hg, which is above the threshold range, it alerts the user that the measurement exceeds the threshold. Further, an alert may occur based on the raising of pressure per time. Other parameters may also be used.

[0131] The parameter thresholds may be provided to the external reader **2800**, such as by a user manually entering a parameter threshold. Alternatively, the parameter thresholds may be provided to the external reader **2800** from another device, such as computer through communications port **2816**. The parameter thresholds may be provided via a direct connection, such as by a wire, or by a wireless connection, such as by a LAN, a WAN or the like.

[0132] The calibration memory **2822**, the data memory **2824** and the parameter memory **2826** may be separate memory storage devices within the external reader, or each may be a portion of a single memory storage device. The reader may use a signal at 13.56 MHz, or other known frequencies. One example of a suitable reader is disclosed in published patent application No. PCT/EP2004/012670.

[0133] Bi-directional power evaluation module **2828** assists in evaluating the strength of signals received from the implanted device to ensure that a minimum signal strength is received. The signal received from the implanted devices via antenna **2802** is evaluated by the bi-directional power evaluation module **2828**. The evaluation may be implemented via various methodologies. According to an embodiment of the invention, the reader may increase the power of the signal sent via the antenna **2802** to the implanted device over small increasing increments. At each increment, the bi-directional power evaluation module **2828** evaluates the signal received back from the implanted device to determine the quality and strength of the signal. This process is repeated until a successful signal is received from the implanted device. The reader then uses the minimum power necessary to achieve an acceptable signal and begins performing the reading of data, such as pressure and temperature measurements, from the device. This may be performed by taking a predetermined number of readings (e.g., five readings) in a row. All the readings may be taken after the minimum power level has been determined. Alternatively, the bi-directional power evaluation module **2828** may determine the minimum power level after each of the predetermined readings.

[0134] By way of another embodiment of the invention, the reader can increase the power level supplied by the antenna **2802** by larger increments, such as by quarter Fourier steps (FS steps), until a valid signal is received from the implanted device. Once a valid signal is obtained, the power is decreased by one step, such as $\frac{1}{4}$ FS, then increased in smaller steps,

such as $\frac{1}{8}$ FS steps) until a valid signal is received. This process is repeated using progressively smaller steps ($\frac{1}{16}$ FS, $\frac{1}{32}$ FS) until a minimum power level is determined. The reader then uses the resulting minimum power level to compute the required power setting and obtains a predetermined number of readings.

[0135] Another methodology involves assessing the demodulation quality level (DQL) of the signal in addition to the signal state analysis described above. The DQL of a signal changes as the coil geometry and/or distance from the reader changes. It does not use an incremental algorithm to assess the required starting power, but the last power setting of the last measurement. The reader sets the power to a previously used level. If a reading is possible, the reader increases or decreases the power for the next reading according to DQL. If no reading is possible, the reader increases the power in increments until valid signal is received from the implanted device. After a predetermined number (e.g., five) of successful readings, the reader obtains the measurement readings. During these measurement readings, the reader continues to increase, decrease, or hold the power level according to DQL and obtaining valid signals.

[0136] Use of power conditioning may result in various beneficial characteristics and features for the voltage controller/stabilizer supply voltage (V_{DDA}) used in the ASIC. When using power conditioning, there is generally a high common mode rejection ratio (CMRR) for V_{DDA} , as well as good radio frequency (RF) suppression for V_{DDA} . In addition, a fast power on reset (POR) signal is used if V_{DDA} falls below tolerance, which would happen if supply power is not sufficient. Because there is no measurement or signal transmission if POR not "1," determining a proper power supply using the power conditioning may prevent this drawback.

[0137] The implantable sensor device of the invention also may be incorporated or attached to other devices implanted within the body. Examples of such devices may include a pacemaker, defibrillator or a drug dispenser.

[0138] While the invention has been described in terms of exemplary embodiments, those skilled in the art will recognize that the invention can be practiced with modifications in the spirit and scope of the appended claims. For example, while the embodiments described above have been directed to implantation of the telemetric sensing device of the invention within the heart, one or more such devices may be implanted within other positions in the cardiovascular system of a patient, such as the aorta, pulmonary artery, or other great vessel. These examples given above are merely illustrative and are not meant to be an exhaustive list of all possible designs, embodiments, applications or modifications of the invention.

Claimed are:

1. An implantable pressure monitor comprising:
 - a substantially rigid chip including:
 - a proximal end and a distal end;
 - pressure sensors exposed on a first surface of the chip in a sensor region of the distal end;
 - signal processing circuitry receiving pressure-indicative signals from the sensors and producing pressure-indicative output signals; and
 - a chip electrical connector in the proximal end communicating the output signals;
 - a substantially rigid substrate that:
 - is spaced apart from the chip;
 - faces the first surface of the chip;

- is connected to the chip electrical connector in the chip's proximal end by a substrate electrical connector; defines an aperture positioned over the sensor region of the chip's distal end, thereby exposing the pressure sensors; and covers the distal end of the chip except for the sensor region;
 - a flexible filler material located throughout space between the chip and the substrate except beneath the aperture, thereby leaving the pressure sensors exposed, such that (a) the flexible filler material connects the chip to the substrate, and (b) the distal end of the chip is connected to the substrate by only the flexible filler material;
 - a wire that:
 - extends from the substrate;
 - is electrically connected to the substrate electrical connector;
 - communicates the output signals; and
 - is not connected to the chip;
 - a biocompatible sheath that encapsulates the chip, substrate, filler, and wire, and is sufficiently flexible to transmit pressure exerted on the sheath exterior through the sheath.
2. The implantable pressure monitor of claim 1, wherein the sheath has a curved shape to reduce or eliminate hydrodynamic forces.
 3. The implantable pressure monitor of claim 1, wherein the substantially rigid substrate extends proximally from the chip to a proximal end which comprises an anchor.
 4. The implantable pressure monitor of claim 3, wherein the anchor comprises a transversely extending flange that forms suture wings.
 5. The implantable pressure monitor of claim 1, further comprising a holder that fixedly receives the chip and the substantially rigid substrate and extends proximally to a proximal end which comprises an anchor.
 6. The implantable pressure monitor of claim 5, wherein the holder further comprises a distal end cap protecting the chip and the substantially rigid substrate.
 7. The implantable pressure monitor of claim 1, wherein the flexible filler material comprises silicone.
 8. The implantable pressure monitor of claim 1, wherein the flexible filler material holds the chip and the substrate together in a fixed relationship.

9. The implantable pressure monitor of claim 1, wherein the substantially rigid substrate extends distally from the chip to a distal end which comprises a barrier wall protecting a distal end of the chip.
10. The implantable pressure monitor of claim 9, wherein the barrier wall forms an end cap at the distal end of the chip.
11. The implantable pressure monitor of claim 10, wherein the barrier wall extends in a direction substantially perpendicular to a plane of the substrate and to a height such that a top of the barrier wall is at or above a top of the chip.
12. The implantable pressure monitor of claim 9, wherein the substantially rigid substrate further extends distally from the barrier wall to a tapered front portion.
13. The implantable pressure monitor of claim 1, wherein the sheath comprises a one-piece, seamless silicone covering.
14. The implantable pressure monitor of claim 13, wherein the flexible filler material comprises silicone.
15. The implantable pressure monitor of claim 1, wherein the substrate is sufficiently rigid such that it cannot be folded or rolled up.
16. The implantable pressure monitor of claim 1, wherein the substrate is sufficiently rigid to protect the pressure sensors from damage as a consequence of contact with a surgical instrument during implantation and from mechanical damage during use.
17. The implantable pressure monitor of claim 1, wherein the substrate is sufficiently rigid to avoid twisting of the chip due to turbulent blood flow.
18. The implantable pressure monitor of claim 1, wherein the substrate is rigid.
19. The implantable pressure monitor of claim 1, wherein the substrate is mechanically inflexible.
20. A method of monitoring pressure with the implantable pressure monitor of claim 1 positioned in a space in which pressure is to be monitored such that pressure sensors of the monitor are exposed to the pressure, the method comprising:
 - conveying power to the monitor, thereby causing the monitor to operate;
 - receiving from the monitor the pressure-indicative output signals; and
 - storing the pressure-indicative output signals.

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