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(54) **Microfabricated implantable wireless pressure sensor for use in biomedical applications and pressure measurement and sensor implantation methods**

Mikrofabrizierter implantierbarer Funkdrucksensor zur Verwendung in Biomedizinischen Anwendungen sowie Druckmessungs- und Sensorenimplantationsverfahren

Capteur de pression sans fil implantable microfabriqué destiné à être utilisé dans des applications biomédicales, et procédés de mesure de pression et d'implantation de capteur

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Description

FIELD OF THE INVENTION

[0001] The field of the invention relates to pressure sensors and, more particularly, to microfabricated implantable pressure sensors for use in biomedical applications including monitoring of intraocular pressure.

BACKGROUND

[0002] Pressure sensor devices have been used to study various physiological conditions in biomedical applications. One known application is to monitor intraocular pressure, for example, in connection with treatment of glaucoma. Glaucoma is a well-known ocular disease that affects millions of people. Persons afflicted with this disease require treatment for life. The disease causes visual field loss and if left untreated, may result in permanent loss of vision, and is a primary cause of blindness in the United States and elsewhere.

[0003] The exact cause of glaucoma is not known, but it is characterized by pathological changes in the optic disc and nerve fiber of the retina. Studies suggest that development of the disease may be attributable to various factors including elevated intraocular pressure. Normal intraocular pressure typically ranges from about 10 to about 21 mm Hg, e.g., about 15 mm Hg. Intraocular pressures of eyes of patients having glaucoma often exceed 21 mm, although glaucoma may exist when intraocular pressures are at lower levels. Elevated intraocular pressures are believed to be responsible for slowly damaging the optic nerve which, in turn, can cause blind spots in the field of vision. Total blindness may occur if the entire optic nerve is damaged.

[0004] One known manner of measuring intraocular pressure is to use an external pressure measurement device that acquires intraocular pressure readings from outside of the eye. One known pressure measurement device is known as a tonometer, which measures an external deformation of an eye and relates that measurement to intraocular pressure. Such external measurement devices, however, may not have the desired level of accuracy since they operate in an external environment rather than within the eye itself. Further, such devices do not provide for continuous monitoring of intraocular pressure since a tonometer must be utilized each time intraocular pressure is to be determined and, therefore, provides discontinuous intraocular pressure monitoring.

[0005] It is also known to implant a sensor into an eye for purposes of measuring an electrical parameter related to intraocular pressure, and to use telemetry to obtain an electrical parameter measurement and relate the electrical parameter measurement to intraocular pressure. In one known system, an external instrument generates a signal to remotely energize an in vivo intraocular pressure sensor. The response generated by the in vivo sensor is measured and correlated to intraocular pressure.

[0006] For example, referring to Figure 1, a known intraocular telemetry system 10 includes an external system 20 and an internal or implanted intraocular sensing circuit 30. The external system 20 includes an excitation circuit 21 and a measurement device 22. The sensing circuit 30 typically includes a resistor (R_{sensor}) 33 and an inductor (L_{sensor}) 34 and a capacitor (C_{sensor}) 35. The capacitor 35 may be configured to vary with the intraocular pressure applied to the capacitor 35.

[0007] The excitation circuit 21 typically includes an inductor (L) 24. During use, the excitation circuit 20 generates energy, which is delivered to the sensing circuit 30 by inductive coupling between the inductors 24, 34, thereby energizing the sensing circuit 30. The resulting response (e.g., resonant frequency or impedance) of the sensing circuit 30 is measured by the measurement device 22 and correlated to intraocular pressure.

[0008] The implanted sensing circuit 30 is essentially an RLC resonance circuit. The resonant frequency and the Quality (Q) factor of the circuit 30 are determined by resistance, capacitance and inductance parameters as provided by Resonant Frequency $(f) = 1/(2\pi\sqrt{(LC)})$; and Q Factor $= 1/R\ (\sqrt{(L/C)})$. A change of capacitance causes a shift in resonant frequency of the implanted sensor circuit 30, which can be wirelessly measured by the external measurement device 22. Examples of such intraocular implants and telemetry systems are described in U.S. Patent No. 6,579,235 to Abita et al., "Passive Silicon Transensor Intended for Biomedical, Remote Pressure Monitoring," by Backlund et al., "A system for wireless intra-ocular pressure measurements using a silicon micromachined sensor," by Rosengren et al., and "A system for passive implantable pressure sensors"; by Rosengren et al.

[0009] One known capacitor for use in intraocular pressure sensors is manufactured using MEMS technologies and includes a membrane, a flat bottom portion and a chamber. The capacitor is part of a pressure sensor that is implantable to monitor pressures through a remote telemetry connection. Another known capacitor device used in pressure sensors is referred to as a comb-drive capacitor unit. One known capacitor unit is described in "Design and Simulation of a MEMS-Based Comb-Drive Pressure Sensor for Pediatric Post-Operative Monitoring Applications," by Duck-Bong Seo et al. Seo et al. describe an implantable MEMS-based pressure sensor to monitor pressures through a remote telemetry connection in the context of monitoring pressures of the right side of the heart following surgery. Seo et al. show a flat membrane and a comb drive and explain that a change of overlapping area changes the capacitance of the device, and that no bending or other deformation of the membrane was found for the comb-drive sensor.

[0010] While known sensor devices and telemetry systems may provide some improvements over known external pressure measurement devices, they can be improved. For example, certain known sensor devices present performance, biocompatibility, packaging and/or size challenges. Certain known devices also lack sensitivities and detection ranges suitable for various biomedical applications. Further, certain known devices utilize wafer bonding techniques, which typically require additional fabrication time and result in larger or thicker devices. Additionally, bonding often results in reduced yield rate, e.g. due precise component alignment requirements. Thus, devices that are fabricated using wafer bonding are not desirable. Certain known devices also may not be adaptable to commercial fabrication on a large scale. Additionally, the inductor element of the implanted sensor circuit can be improved to provide a more effective sensor circuit and more accurate intraocular pressure determinations. Known devices may also require larger incisions or blades for implantation of sensor devices due to their large size. Such incisions are not desirable. Further, certain known implants require sutures to remain implanted in the eye, which are also not desirable.

[0011] US 2003/0139677 A1 discloses a microfabricated implantable pressure sensor in accordance with the preamble of claim 1.

[0012] Duck-Bong Seo ET AL: "DESIGN AND SIMULATION OF A MEMS-BASED COMB-DRIVE PRESSURE SENSOR FOR PEDIATIC POST-OPERATIVE MONITORING APPLICATIONS", , 1 January 2003 (2003-01-01), 2003 Summer Bioengineering Conference, Sonesta Beach Resort in Key Biscayne, Florida Retrieved from the Internet: URL:http://www.tulane.edu/~sbc2003/pdfdocs/1239.PDF [retrieved on 2013-02-18] discloses a comb-drive variable capacitor for a microfabricated implantable pressure sensor.

[0013] Therefore, it would be desirable to have implantable sensor devices that can be fabricated using known micromachining and MEMS technologies. It would also be desirable to have implantable sensor devices that are sufficiently small or miniature in size so that they may be delivered through a needle rather than through a large incision using a blade. It would also be desirable to have sensor devices that may be implanted without the need for sutures and in various locations of an eye. Further, it would also be desirable to have biocompatible and implantable microfabricated sensor devices with improved capacitor and inductor components for enhanced sensitivity, dynamic range and accuracy. It would also be desirable to continuously and passively monitor intraocular pressure by telemetry using such sensor devices. Such capabilities would enhance biomedical applications and pressure-dependent physical conditions and diseases including monitoring of intraocular pressure.

SUMMARY

[0014] The invention is directed to a microfabricated implantable pressure sensor according to claim 1. Further developments of the invention are disclosed in the dependent claims 2 to 12.

[0015] The microfabricated implantable pressure sensor includes a variable capacitor and an inductor. The variable capacitor and the inductor are electrically connected to each other. The variable capacitor includes a substrate, a flexible member and a plurality of capacitor elements. The substrate defines a plurality of channels, and edges of the flexible member are on the substrate. A middle portion of the flexible member is raised above the substrate, thereby defining a chamber. Capacitor elements extend indirectly from the flexible member. Fluid pressure changes on the middle portion cause the middle portion to move, thereby causing the capacitor elements to move within respective channels and causing capacitance to vary with changes in an overlapping area of the capacitor elements and the substrate. An electrical circuit including the variable capacitor and the inductor can generate a detectable resonant frequency shift in response to a change of fluid pressure on an outer surface of the flexible member.

[0016] In one or more embodiments, capacitor elements extend indirectly from a flexible member by an indirect connection, e.g., by an indirect connector including an intermediate member and a cross bar or member. The capacitor elements are carried by the cross bar or member, which is connected to or extends from an intermediate member, which extends between the flexible member and the cross bar or member. Thus, capacitor elements that move within channels do not extend directly from the flexible member. In one or more embodiments, the middle portion of the flexible member may be deformed in a non-linear manner, e.g., to assume a bowl-like shape, while the intermediate member / cross bar configuration permits the capacitor elements to remain movable within respective channels in a direction that is perpendicular to a plane defined by a top surface of the substrate. The intermediate member and at least one capacitor element may lie within a common vertical plane, and at least one capacitor element may lie within a vertical plane that is offset from and parallel to a vertical plane defined by the intermediate member.

[0017] In one or more embodiments, a middle portion of a flexible member may be flexible and resilient (e.g., made of Parylene) so that movement or deformation of the flexible member alters the overlapping area of capacitor elements and the substrate, thereby changing capacitance. Channels in the substrate and capacitor elements may form mating comb structures.

[0018] In one or more embodiments, variable capacitor and the inductor components are configured to detect fluid pressure changes with a sensitivity of about 1 mmHg within a fluid pressure range of about 1-50 mmHg.

[0019] In one or more embodiments, the inductor may be stationary and have a fixed inductance and be formed by a stack of insulated inductor elements that encircle a variable capacitor. Inductor components may extend through the entire substrate or extend partially through or be deposited on the substrate. The inductor may also be in the form of a ring, which can be collapsed or compressed configuration for delivery through a needle, e.g., a 20-25 gauge needle, and expanded when delivered at the treatment site. Embodiments also provide for sutureless implantation.

BRIEF DESCRIPTION OF THE DRAWINGS

[0020] Referring now to the drawings in which like reference numbers represent corresponding parts throughout and in which:

Figure 1 illustrates a known telemetry system for monitoring intraocular pressure;

Figure 2 is a perspective cross-sectional view of a variable capacitor of a microfabricated implantable pressure sensor constructed in accordance with one embodiment;

Figure 3 is a cross-sectional view of the variable capacitor shown in Figure 2 in which a flexible member is in an initial or relaxed state when external fluid pressure is less than an internal chamber pressure;

Figure 4 is a cross-sectional view of a variable capacitor shown in Figure 2 in which a flexible member is in a compressed or deformed state when external fluid pressure is greater than the internal chamber pressure;

Figure 5 further illustrates a channel formed within a substrate and a capacitor element extending from a flexible member and being moveable within the channel to alter the overlapping area and capacitance;

Figure 6 is a graph showing a relationship between the overlapping area and change in capacitance achieved with the capacitor configuration shown in Figures 2-5;

Figure 7 is a partial cross-sectional view of a substrate and a capacitive element at a first depth within a channel, corresponding to overlapping area A1 in Figure 6;

Figure 8 is a partial cross-sectional view of a substrate and a capacitive element at a second depth deeper within a channel, corresponding to overlapping area A2 in Figure 6;

Figure 9 is a partial cross-sectional view of a substrate and a capacitive element at a third depth deeper within a channel, corresponding to overlapping area A3 in Figure 6;

Figure 10 is a graph showing a relationship between changes of capacitance and pressure, and measurement sensitivity achieved with the capacitor configuration shown in Figures 2-5;

Figure 11 is a perspective view of a lump inductor of an implantable pressure sensor having integrated metal lines according to one embodiment;

Figure 12 is a perspective cross-sectional view of a lump inductor constructed having stacked metallic layers separated by insulative material according to another embodiment;

Figure 13 is a perspective cross-sectional view of a lump inductor constructed having metallic elements formed through or embedded within a substrate according to another embodiment;

Figure 14 is a perspective view of a lump inductor having metallic elements formed through or embedded within a substrate and a foldable or rollable inductor sheet or ring according to a further embodiment;

Figure 15 further illustrates a structure of the foldable or reliable inductor sheet or ring of the inductor shown in Figure 14 in accordance with one embodiment;

Figure 16A is a flow diagram illustrating one embodiment of a method of fabricating an implantable pressure sensor having a variable capacitor and a lump inductor in which inductor elements are formed by metal lines extending through a substrate;

Figure 16B illustrates an alternative sensor configuration having a variable capacitor and a variable inductor that can be fabricated using process steps shown in Figure 16A;

Figure 17 is a table summarizing expected electrical parameters of microfabricated pressure sensors constructed according to embodiments and having a variable capacitor shown in Figures 2-4 and different lump inductor configurations shown in Figures 12-15.

DETAILED DESCRIPTION OF ILLUSTRATED EMBODIMENTS

[0021] Certain embodiments are directed to a variable capacitor that is a component of a microfabricated implantable pressure sensor for use in various biomedical applications. A variable capacitor constructed according to embodiments includes a substrate having trenches or channels defined therein and a flexible member. A portion of the flexible member is raised above the substrate. Capacitor elements extend indirectly from the flexible member and movable together and simultaneously within the channels, thereby varying capacitance as a result of changing the overlapping area of the substrate and capacitor elements. Certain other embodiments are directed to a microfabricated implantable pressure sensor and configurations of a variable capacitor and an inductor. An inductor may have a fixed or variable inductance. The inductor may be fixed or stationary, or be movable, e.g., with a component of the variable capacitor. Certain other embodiments are directed to a microfabricated sensor having a variable capacitor and a variable inductor that are carried by or embedded within a common flexible member. Certain embodiments are directed to methods of fabricating implantable pressure sensors using surface micromachining and MEMS technologies.

[0022] Embodiments advantageously provide implantable pressure sensors that may be fabricated using known micromachining and MEMS technologies and are of a miniature size so that they may be delivered through a needle and implanted in various locations without the need for sutures. Embodiments also advantageously provide biocompatible pressure sensors having variable capacitors, lump/variable inductors with enhanced accuracy, sensitivity and range for use in various biomedical applications including passive monitoring of intraocular pressure using telemetry and other biomedical applications involving, e.g., aneurysms and the brain.

[0023] Figures 2-10 illustrate embodiments of a variable capacitor of a microfabricated implantable pressure sensor for use in biomedical applications. The variable capacitor includes a flexible member, a portion of which is raised above a substrate and capacitor elements or plates that are moveable within channels or trenches formed within the substrate to vary capacitance. Figures 11-17 illustrate different lump or fixed inductor configurations that may be used with a variable capacitor and electrical characteristics thereof and related methods of fabrication. Figures 18-21 illustrate embodiments of a microfabricated implantable pressure sensor for use in biomedical applications and having a flexible member, a portion of which is raised above a substrate that does not include channels or trenches, variable capacitance and fixed inductance. Figures 23-25 illustrate embodiments of a microfabricated implantable pressure sensor for use in biomedical applications and having a flexible member that carries elements of a variable capacitor and also elements of a variable inductor.

[0024] Referring to Figures 2-4, a variable capacitor 200 constructed in accordance with one embodiment and configured for use in a microfabricated implantable pressure sensor includes a substrate 210, a flexible member 220 disposed on the substrate 210, and a capacitor component 230 that includes a plurality of capacitor elements 232 extending indirectly from the flexible member 220. The capacitor elements 232 are movable within trenches, grooves or channels (generally channels 216) defined through the substrate 230, e.g., partially through the substrate 230 as shown in Figures 2-4. In the illustrated embodiment, all of the capacitor elements 232 are the same length, but other capacitor element 232 configurations may be utilized.

[0025] Movement of the capacitor elements 232 to different depths within the channels 216 alters the overlapping area of the capacitor elements 232 and the substrate 220. Changing the overlapping area alters capacitance and the resonant frequency response of a sensor circuit that includes the variable capacitor 200. For example, Figure 3 illustrates a plurality of capacitor elements or plates 232 positioned at a first depth resulting in an overlapping area A1, whereas the overlapping area increases by ΔA to A2 when the capacitor elements 232 are moved deeper down into the channels 216.

[0026] The substrate 210 may be composed of silicon and may be in the form of a wafer having a thickness of about 500 microns. Although this specification refers to silicon, the substrate 210 may be composed of other materials including a conductive polymer or another suitable micromachinable substrate material having sufficiently high conductivity. The substrate 210 has a top surface 212 and a bottom surface 214. One or more channels 216 are formed through the top surface 212 of the substrate, thereby forming corresponding projections, walls or fingers 218. In the illustrated embodiment, the channels 216 and projections 218 form a comb structure.

[0027] In the illustrated embodiment, the substrate 210 defines a plurality of channels 216, e.g., five channels 216, and four corresponding projections 218. It should be understood, however, that the substrate 210 may define other numbers of channels 216, e.g., about 3 to 10 channels 216. The number of channels may depend on the capacitor 200 configuration, e.g., the width of the substrate 210 and/or the number of capacitor elements 232. Further, although the illustrated embodiment shows channels 216 and projections 218 that are the same width, the channels 216 and projections 218 may have different widths to provide different variations of capacitance and to accommodate different numbers of channels 216 and different capacitor element 232 configurations.

[0028] For example, in embodiments including a 500 micron substrate 210, each channel 216 may have a depth of about 200 microns, a width of about 20 microns, a spacing (projection 218 width) of about 20 microns. The capacitor elements 232 may be movable by about 50 microns within the channels 216, resulting in an overlapping area of the capacitor elements 232 and substrate 210 that may range from about 10^6 to about 10^7 square microns. It should be understood that other dimensions and configurations may be utilized as necessary.

[0029] The flexible member 220 includes an outer or top; surface 221 and an inner surface 222. First and second edges or bottom surfaces 223, 224 are disposed on, connected to, formed on, or sealed to the top surface 212 of the substrate 210. During fabrication of the variable capacitor 200, another material or coating, such as a layer of silicon dioxide (not shown in Figures 2-4), may be applied on the top surface 212 of the substrate 210. Thus, the edges 223, 224 of the flexible member 220 may be in direct contact with a silicon dioxide layer rather than the top surface 212. For ease of explanation and illustration, Figures 2-4 show edges 223, 224 being disposed on, connected to or formed on the top surface 212 of the substrate 210, whether such contact is direct or indirect as a result of an intermediate silicon dioxide layer.

[0030] The flexible member 220 also includes a middle portion 225 that extends between the first and second edges 223, 224. The middle portion 225 is raised above the top surface 212 of the substrate 210, thereby defining an inner space or chamber 226 between the top surface 212 and the inner surface 222 of the flexible member. The capacitor 200 is eventually sealed so that the inner space or chamber 226 is also sealed and has a fixed internal or chamber pressure (Pc).

[0031] In the illustrated embodiment, the middle portion 225 includes first and second arcuate or "shoulder" sections 227, 228. In the illustrated embodiment, each shoulder section 227, 228 extends inwardly and upwardly from respective first and second edges 223, 224 to a middle section 229 that extends between the shoulder sections 227, 228. In the illustrated embodiment, the middle section 229 is flat and parallel to the top surface 212 of the substrate 210, whereas the shoulder sections 227, 228 extend upwardly in some manner (e.g., as a result of having an arcuate shape) so that the middle section 229 is raised above the substrate 210. It should be understood that the middle portion 225 may have other shapes and that the shoulder sections 227, 228 may be arcuate or shoulder shapes or other shapes as necessary in order to raise the middle section 229 above the substrate 210.

[0032] The flexible member 220 is made of a material that allows the middle portion 225, e.g., the middle section 229 and/or one or more shoulder sections 227, 228 depending on the capacitor 200 configuration and fluid pressure application, resulting in deformation, deflection or bending of the middle portion 225 under fluid pressure (P_f) if the fluid pressure is greater than the internal chamber pressure (P_c) (as shown in Figure 4). The flexible member 220 may be resilient to return from a deformed shape (as shown in Figure 4) to an initial or relaxed shape (as shown in Figure 3) when the external fluid pressure is less than the internal chamber pressure.

[0033] For this purpose, the flexible member 220 may be composed of a material having a suitable Young's modulus of about 1 GPa to about 10 GPa, e.g., about 4 GPa. One example of a suitable material for the flexible member 120 is Parylene, e.g., Parylene C, D, N, F, HT, A and AM. For ease of explanation, reference it made to the flexible member 220 being made of a polymer or Parylene, but it should be understood that the flexible member 120 may be composed of other suitable materials that provide desired flexibility and/or resiliency attributes. Selection of flexible member 120 materials may also depend on, for example, ease of micromachining, CMOS/MEMS process compatibility and biocompatibility (e.g., USP Class VI implantable grade).

[0034] In one embodiment, the flexible member 220 may be made of Parylene, have a width of about 500 microns, and the shoulder sections 227, 228 may be configured so that the middle section 229 is raised above the top surface212 of the substrate 210 by about 10 microns. The middle portion 225 may be moved or deflected by about 10 microns towards the substrate 210. It should be understood that these dimensions are provided as one example of how a variable capacitor 200 having a raised flexible member 220 may be implemented, and other configurations may be utilized for different applications.

[0035] Referring to Figures 2-4, and with further reference to Figure 5, capacitor elements 232 may be in the form of fingers or plates that extend indirectly from the flexible member 220 and are arranged in a comb structure. In the illustrated embodiment, capacitor elements 232 extend directly from, or are carried by, one or more cross bars or members 234. An intermediate member 236 extends between the flexible member 220 and the cross bars or members 234. Figures 2-4 illustrate an embodiment that includes a single intermediate member 236 that connects the middle section 229 of the flexible member 220 and the crossbar member 234. Portions of the chamber 226 are defined by the inner surface 222 of the middle section 229, cross bar members 234, the intermediate member 236 extending between the flexible member and a cross bar member 234.

[0036] According to one embodiment, the number of intermediate members 236 is less than the number of capacitor elements 232. In the illustrated embodiment, a single intermediate member 236 joins the middle section 229 and a cross bar 234 that carries a plurality of capacitor elements 232. This configuration advantageously provides a flexible member 220 having sufficient flexibility and advantageously provides linear or vertical movement or substantially linear or vertical movement of capacitor elements 232 within channels 216 even when the flexible member 220 is deformed.

[0037] More specifically deformation of the flexible member 220 by fluid pressure results in downward movement of the flexible member and downward movement of the intermediate member 236 extending from the flexible member. This results in downward movement of the capacitor elements 232 carried by the cross bar 234, which extends from the intermediate member 236. The structural configuration of embodiments advantageously prevents outward bowing of capacitor elements 232 that may result if the capacitor elements 232 extended directly from the flexible member 220 (i.e., without any intermediate member 236, as in known comb structure devices), thereby causing capacitor elements 232 to scrape against inner surfaces of the channels 216, or causing the capacitor elements 232 to not be positioned within the channels 216 depending on the configuration of the capacitor. Thus, embodiments advantageously utilize an intermediate member 236 / cross bar 234 configuration so that capacitor elements 232 extend indirectly from the flexible member, thereby preventing the capacitor elements 232 from being pushed out at an angle when the flexible member 220 is deformed, e.g., in a bowl-like shape, by fluid pressure.

[0038] It should be understood that other structural configurations may be utilized while achieving these advantages. For example, rather than having a single intermediate member 236, other numbers of intermediate members 236 may be utilized so long as the number of intermediate members 236 provides sufficient flexibility and maintains the vertical orientation of the capacitor elements 232 when the flexible member 220 is deformed.

[0039] In the illustrated embodiment, at least one capacitor element 232 is in-line with, or within the same vertical plane defined by, the intermediate member 236, and at least one other capacitor element 232 is within a vertical plane that is offset from the vertical plane defined by the intermediate member 236. In the illustrated example, the middle capacitor element 232 lies within the same vertical plane defined by the intermediate member 236, and the other capacitor elements lie within different vertical planes and are parallel to the plane defined by the intermediate member 236 and the middle capacitor element 232. In other embodiments, the capacitor elements 232 may be arranged so that no capacitor element 232 is in-line with or within the same vertical plane defined by the intermediate member 236, but all capacitor elements 232 are parallel to the plane defined by the intermediate member 236. The particular configuration utilized may depend on, e.g., the number of intermediate members 236, the number of capacitor elements 232 and the arrangement of these components.

[0040] Referring to Figure 5, capacitor elements 232 are configured and have a suitable shape and size so that they may move with the flexible member 220 within channels 216, e.g., within channels 216 of a corresponding substrate 120 comb structure. According to one embodiment, a capacitor element 232 is a conductive material 510, such as a metal, and may be optionally coated with an insulation material 512. In another embodiment, the capacitor element 232 may include a metal coating that is applied over a conductive, non-metallic material. A channel 216 may also include an insulative coating 520 and a conductive or metal coating 522 that is applied within the channel 216 using metallization.

[0041] During use, the flexible member 220 having capacitor elements 232 extending there from is used as a variable capacitor electrode, and the substrate 110 is used as a ground electrode. If the internal chamber 226 pressure is greater than the external fluid pressure, then the flexible member 220 will not be deformed or bent and will retain its original or initial shape. If the fluid pressure exceeds the chamber 226 pressure, then the middle portion 229, e.g., the middle section 225 of the flexible member 220, will be deformed or deflected by the fluid pressure. The flexible member 220 may be sufficiently thin (e.g., about 10 microns) so that the amount of deflection of the middle portion 229 is proportional to the difference between the external fluid pressure and the internal chamber pressure, $((\delta)(\alpha)(\Delta P))$. At the same time, the position of the capacitor elements 232 extending from the flexible member 220 is changed, i.e., the capacitor elements 232 move with the moving flexible member 220.

[0042] As a result, the effective overlapping area between interdigitated electrodes is changed which, in turn, alters the capacitance across the electrodes. More specifically, the capacitance increases as the capacitor elements 232 are moved deeper within respective channels 216 and the overlapping area of the substrate 220 and the capacitor elements 232 increases, and capacitance decreases as the capacitor elements 232 are moved to a shallower depth within the channel 216 and the overlapping area of the substrate 220 and the capacitor elements 232 decreases.

[0043] For example, referring to Figure 6, capacitor elements 232 may assume an initial, relaxed position, generally illustrated as (0,0). The initial position may be the capacitor elements 232 being positioned partially within respective channels 216. Alternatively, a capacitor element 232 may be positioned outside above the channels 216, e.g., above the top surface 112 in the illustrated example. The initial relaxed position may depend on the variable capacitor 200 configuration, e.g., how far the flexible member 220 may be deflected or deformed and the length of the capacitor elements 232.

[0044] Figure 7 illustrates one example in which the initial, relaxed position is a position in which distal portions of capacitor elements 232 are positioned partially inside respective channels 216. When the chamber 226 pressure is greater than the external fluid pressure, the flexible member 220 is in its initial, relaxed state, and the capacitor elements 232 are positioned at a first depth within the channels 216. This arrangement results in an initial overlapping area (A1) of the distal portions of the capacitor elements 232 and the substrate 210, and a corresponding capacitance C1.

[0045] Referring to Figures 6 and 8, as fluid pressure on the outer surface 221 of the flexible member 220 increases, the fluid pressure will exceed the internal chamber 226 pressure, causing the flexible member 220 to bend or deflect towards the substrate 210. This causes the capacitor elements 232 to be moved from the initial depth to a second, deeper depth within the channels 216. This movement of the flexible member 220 results in the overlapping area of the capacitor elements 232 and the substrate 210 to increase from A1 to A2 and results in a corresponding increase in capacitance from C1 to C2.

[0046] Similarly, as shown in Figures 6 and 9, as fluid pressure increases further, the flexible member 220 will bend or deflect towards the substrate 210 to a greater degree, thereby moving the capacitor elements 232 to a third, depth within the channels 216. This movement results in the overlapping area of the capacitor elements 232 and the substrate 210 to increase from A2 to A3 and a corresponding increase in capacitance from C2 to C3.

[0047] The capacitance behavior of this structure can be expressed as $\Delta C = (\varepsilon A / d) (\alpha)(\delta)(\alpha)(\Delta P)$, where ΔC = change of capacitance for a deflection of the flexible member 220 and corresponding movement of capacitive elements 232 within channels 216; ε = permittivity of the channel 216 space; A = overlapping area of capacitor elements 232 and substrate 210; d = distance between a conductive portions 510 of a capacitor element 232 and a conductive layer 520 of the channel 216 of the substrate 210; α is the proportional symbol and ΔP = the change in fluid pressure on the flexible member 220.

[0048] Figure 10 illustrates how a change in capacitance may be correlated to a change in fluid pressure on the flexible member 220. In the illustrated example, a 0.4 pF change of capacitance corresponds to a pressure change of 1 mm Hg. Thus, embodiments are capable of pressure measurements with 1 mm Hg sensitivity.

[0049] The total capacitance may be expressed as $C(total) = C0 + \Delta C(\Delta P)$ where C(total) = total capacitance; C0 = a fixed capacitance (when $\Delta P = 0$); ΔC = change of capacitance as a function of pressure difference ΔP on the flexible member 220 and ΔP = pressure difference on the flexible member 220. The total capacitance should be sufficiently high to allow a variable capacitor 200 to be used in telemetry systems (e.g., in the system generally illustrated in Figure 1). Total capacitance may be increased by increasing the area of the capacitor elements 232 (larger electrode overlapping area), providing a larger number of capacitor elements 232, structuring the flexible member 220 so that it may be deflected to greater depths within channels 216 to increase overlapping areas, and decreasing the distance between interdigitated electrodes.

[0050] Additional considerations for effective telemetry include having a pressure sensor with sufficiently high inductance and sufficiently high coupled capacitance in order to allow the resulting resonant frequency of the sensor circuit to lie within a reasonable detection range. For example, the resonant frequency of an implantable sensor circuit should lie between 10-500 MHz for telemetry involving biomedical applications. For this purpose, in addition to having a variable capacitor 200 and sufficient capacitance as discussed above with reference to Figures 1-10, microfabricated implantable pressure sensors should also have inductor elements that allow the sensor to be implantable and provide electrical characteristics (e.g. resonant frequency) suitable for use in biomedical applications and telemetry. Figures 11-15 illustrate different embodiments of pressure sensors having lump inductors or inductors having a fixed inductance.

[0051] Figure 11 illustrates a lump inductor 1110 constructed in accordance with one embodiment for use in a microfabricated implantable pressure sensor 1100 includes a variable capacitor 200 (not illustrated in Figure 11 for clarity). Further, Figure 12 is a perspective cross-sectional view illustrating metallic layers 1211 along two sides of the variable capacitor 200 in order to illustrate how the variable capacitor 200 and the inductor 1210 may be integrated within the sensor 1200, but it should be understood that the stacked metallic layers 1211 are arranged around the variable capacitor 200.

[0052] The inductor 1110 is formed by metal lines 1112 that are integrated within the top surface 212 of the substrate 210 and surround the variable capacitor 200. In the illustrated embodiment, a single wire 1112 is wound in a spiral pattern around the variable capacitor 200. One example implementation of the inductor 1110 shown in Figure 11 may include a metallic line or element 1112 having a thickness of about 2 microns, a width of about 20 microns, and being wound to form about five overlapping sections. Overlapping metal lines 1112 may be spaced apart by about 10 microns.

[0053] Referring to Figure 12, in another embodiment, a microfabricated pressure sensor 1200 includes an inductor 1210 that is formed as a stack of metallic layers 1211 that are fabricated using surface-micromachining methods. In this embodiment, the inductor 1210 is arranged so that alternating insulative layers 1212 and metallic layers 1211 are stacked together. This inductor configuration may be particularly suited for configurations that required increased lump inductance and lump capacitance. The insulative layer 1212 may be a polymer such as Parylene or the same material that is used to form the flexible member 220. All of the metallic layers 1211 may be embedded within an insulative material 1212, or a top metallic layer 1211 may be exposed (as shown in Figure 12). In one embodiment, the inductor 1210 may include a stack of about two to four metallic layers 1211. The thickness of a metallic layer 1211 may be about 2 microns, a width of a metallic layer 1211 may be about 20 microns and the thickness of the insulative layer 1212 between metallic layers 1211 may be about 2 microns.

[0054] Referring to Figure 13, it may be desirable to increase inductance while reducing resistance in order to increase the quality (Q) factor for higher sensing capabilities in terms of both sensitivity and sensing distance. For this purpose, a microfabricated pressure sensor 1300 may include a variable capacitor 200 (as shown in Figures 1-10) and a high aspect ratio inductor 1310. Figure 13 is a perspective cross-sectional view illustrating the inductor 1210 elements along two sides of the variable capacitor 200 in order to illustrate how the variable capacitor 200 and the inductor 1310 may be integrated within the sensor 1300, but it should be understood that the metal lines 1311 are arranged around the variable capacitor 200.

[0055] The inductor may include thick metal lines 1311 that fill channels 216 that are formed completely through the portions of the substrate 210. In other embodiments, the metal lines 1311 may fill channels 216 formed partially through the substrate 210 depending on the desired inductance and resistance. The high aspect ratio inductor 1310 configuration shown in Figure 13 is well suited to maximize the capacitance and inductance of the sensor 1300 while reducing resistance as a result of the dimensions of the thick metal lines 1311 based on the expression R = p L/A, where p = resistivity of the metal material, L = length of the metal line, and A = area of the metal line. For example, the thickness of the substrate 210 may be about 500 microns, metal lines 1311 may extend through the substrate 210 to have a depth that is also about 500 microns, the width of the metal lines 1311 may be about 20 microns and the metal lines 1311 may extend along the length of the substrate 210, e.g., about 3 millimeters.

[0056] Referring to Figures 14 and 15, in another embodiment, a microfabricated pressure sensor 1400 may include a variable capacitor 200 and a lump inductor 1410 in the form of an inductor sheet. For purposes of illustration, not limitation, the sensor 1400 is shown as having an inductor sheet 1410 that is coupled to metal lines 1311 of the high aspect ratio inductor 1310 shown in Figure 13. In other embodiments, the inductor sheet 1410 may be used as the sole inductor element, or in combination with other types of inductors, e.g., as shown in Figures 11 and 12. Thus, Figures 14 and 15 are provided as one example of how embodiments may be implemented.

[0057] In the illustrated embodiment, the inductor sheet 1410 has a circular shape (when in an expanded or relaxed shape) and includes alternating metallic layers 1411 and insulative layers 1412. The metallic layers 1411 may be platinum, titanium and gold, or another suitable biocompatible metal or conductive materials. The insulative layers 1412 may be a polymer such as Parylene.

[0058] The inductor sheet 1410 is preferably configured for implantation through a clinical gauge needle (e.g., having a 20-25 gauge size). For this purpose, the inductor sheet 1410 may be configured to assume a stressed or compressed shape when being delivered through a needle and an expanded or relaxed shape after the sensor 1400 is deployed from the needle and implanted. For example, the inductor sheet 1410 may be rolled or folded while positioned within the needle and may expand to assume a circular shape (as shown in Figures 14 and 15) when the pressure sensor 1400 is deployed from the needle.

[0059] Figure 16A illustrates an embodiment of a method 1600 of fabricating a micromachined pressure sensor having a variable capacitor (e.g., as shown in Figures 1-10) and a lump inductor (e.g., the inductor 1310 as shown in Figure 13). It should be understood that method steps shown in Figure 16A can be utilized and/or adapted to fabricate pressure sensors having other variable capacitors and other lump inductors (e.g., as shown in Figures 11 , 12, 14 and 15). For ease of explanation, reference is made to a method for fabricating the pressure sensor having a variable capacitor and lump inductor shown in Figure 13.

[0060] At stage 1605, a substrate 210, such as a silicon wafer, is provided. The substrate 210 may have a thickness of about 500 microns. The substrate 210 is etched, e.g., deep reactive-ion etching (DRIE). In the illustrated embodiment, DRIE may be used to etch partially through a central portion of the substrate 210 to form channels (for the eventual variable capacitor 200) and to form other channels 216 completely through the substrate 210 (for the eventual inductor 1310). The width of the channels 216 in the central portion of the substrate 210 may be about 20 microns, and the depth of the channels 216 in the central portion of the substrate 210 may be about 200 microns. The width of the other channels 216 formed through the substrate 210 may also be about 20 microns. A tissue anchor (not shown in Figure 16A) may be created on the backside 214 of the substrate 210. One example of a suitable tissue anchor is described in U.S. Publication No. 2006/0247664, entitled "Micromachined Tissue Anchors for Securing Implants Without Sutures by E. Meng et al..

[0061] At stage 1610, a first insulative layer 520 (e.g., as shown in Figure 5) is deposited over the top surface 212 of the substrate 210. The insulative layer 520 may be Parylene and may have a thickness of about 2 microns. As shown in Figure 16A, the first Parylene layer 520 is applied and patterned to coat surfaces that were exposed as a result of the etching at stage 1605, i.e., the inner surfaces of the open channels 216 formed partially and completely through the substrate 210.

[0062] At stage 1615, metal electroplating is performed on the open channels 216 that were formed through the substrate 210 so that these channels 216 are filled with metal 1311 (as further illustrated in Figure 13). These metal- filled channels or lines 1311 will eventually form the high aspect ratio inductor 1310 that is integrated within the substrate 210.

[0063] At stage 1620, surface metallization is performed on channels 216 that were formed partially through the substrate 210, thereby forming a layer of metal 522 over the first Parylene layer 520 (as further illustrated in Figure 5).

[0064] At stage 1625, a first sacrificial coating of photoresist 1626 is applied (e.g., by spin coating) over a portion of the substrate 210. The thickness of the first photoresist coating 1626 may be about 10 microns. One suitable photoresist 1626 that may be utilized with embodiments is a layer of AZ4620 type photoresist (supplied by Clariant Corp., Charlotte, NC). The photoresist 1626 may be hard-baked at about 120<0>C for smoothing of edges and degassing purposes. In the illustrated embodiment, the first photoresist coating 1626 is applied over the metal-filled channels 1311 positioned between other open channels 216 formed partially through the substrate 210.

[0065] At stage 1630, Parylene is applied and patterned a second time to fill with Parylene open channels 216 that were previously coated with metal, and to coat the photoresist 1626 with Parylene. The second Parylene layer may have a thickness of about 2 microns and will eventually form capacitor elements 232 and the cross bar 234 (as further illustrated in Figures 2 and 13).

[0066] At stage 1635, a second sacrificial photoresist coating 1636 is applied and patterned over the second Parylene coating that forms capacitor elements 232 and cross bar 234 elements, over portions of the substrate 210 and over channels 216 filled with metal 1311. The thickness of the second photoresist coating 1636 may be about 15 microns.

[0067] At stage 1640, metal connections are formed on electrodes (not shown for clarity) for purposes of connecting the metal-filled channels 216 (inductor wires 1311) and capacitor elements or interdigitated electrodes.

[0068] At stage 1645, Parylene is applied and patterned a third time. The third Parylene coating may have a thickness of about 5 microns and forms the flexible member 220 and an intermediate member 236 that extends between the flexible member 220 and the cross bar 234 elements formed at stage 1635. In the illustrated embodiment, the third Parylene layer covers the second photoresist coating 1636, portions of the substrate 210 and metal filled channels 216. The third Parylene coating is applied over sections that will eventually form the variable capacitor 220 and other sections that will eventually form the lump inductor 1310.

[0069] At stage 1650, the backside 212 of the substrate 210 is etched, e.g., using DRIE, and at stage 1655, the first and second photoresist layers 1626, 1636 that were applied at stages 1625 and 1630 are stripped away, thereby releasing the device components.

[0070] More specifically, metal 1311 that fills the channels 216 formed through the entire substrate 210 form the high aspect ratio fixed inductor 1310 (as further illustrated in Figure 13), the top electrode plates or capacitor elements 232 are joined by cross bar elements 234 and are connected to the intermediate member 236, which extends between the cross bar elements 234 and the flexible member 220 (as further illustrated in Figures 2-4 and 13), and the bottom electrode plates form projections or fingers 218 and corresponding channels 216 in which capacitor elements 232 move to vary capacitance.

[0071] It should be understood that method fabrication steps can be modified or adapted for fabrication of other structures of embodiments. The inductor may be a fixed inductor (e.g., as shown in Figure 13), or method embodiments can be applied to fabricate a structure having a variable capacitor (as discussed above) and a variable inductor, e.g., as shown in Figures 16B. The variable inductor shown in Figure 16B may be formed by stage 1660 during which further etching 1660 of the silicon substrate 210 is performed to release the metal 1311 components and form a variable inductor. Thus, embodiments can be adapted for fabrication of various variable capacitor / lump inductor and variable capacitor / variable inductor configurations, and it should be understood that Figures 16A-B are provided to show examples of how embodiments may be implemented.

[0072] Figure 17 is a table summarizing expected physical, electrical and microelectromechanical attributes of microfabricated pressure sensors having a variable capacitor as shown in Figures 1-10 and different lump inductors having fixed inductance as shown in Figures 12-15. Data in Figure 17 was derived using finite element analysis and accepted electrical model calculations.

[0073] Embodiments advantageously provide microfabricated pressure sensors having sufficiently high capacitance, inductance, resonant frequency (f_r), f_r shift (Δf) and sensitivity ($\Delta f/fr$), and sufficiently low resistance. For example, Figure 17 shows that the pressure sensor 1200 shown in Figure 12 has high inductance (about 40nh), the pressure sensor 1300 shown in Figure 13 low resistance (about 0.03 ohm) and a high Q factor (~600), and the pressure sensor 1400 shown in Figure 14 has high inductance (about 145nh) and high capacitance (about 127 pF). The pressure sensor 1300 including the high aspect ratio inductor 1310 has the lowest resistance (~0.03 ohm). Ratios of ($\Delta F/f_r$) for all three pressure sensors 1200, 1300, 1400 were determined to exceed 10^{-3}, indicating that sensor embodiments would be suitable for detection by an external measurement device of a telemetry system.

[0074] Figure 17 also shows that microfabricated pressure sensors constructed according to embodiments should have sufficient sensitivity to be able to measure 1mm Hg pressure changes, which correspond to a capacitance change of about 0.4pF, while providing for a detection range of about 1- 50mmHg. Figure 17 also shows that microfabricated pressure sensors that include a variable capacitor and inductors according to embodiments are advantageously sufficiently small in size so that they may be implanted through a clinical gauge needle and be implanted in various parts of an eye. For example, pressure sensors 1200 having the variable capacitor 200 (Figures 2-4) and the inductor 1210 (Figure 12) or the inductor 1310 (Figure 13) have dimensions of about 0.5mm x 0.5mm 3.0 mm, and pressure sensors 1400 having the variable capacitor 200 (Figures 2-4) and the inductor 1410 including a rollable sheet has dimensions of about 0.5mm x 0.5mm x 4.0mm (when in a stressed or compressed configuration). Other minimally invasive incisions may also be utilized if desired, e.g. incisions in the cornea that are smaller than about 3mm to allow self-healing of the cornea. Further, tissue anchors may be utilized to implant sensor embodiments without the need for sutures, e.g., as described in U.S. Publication No. 2006/0247664.

[0075] Although particular embodiments have been shown and described, it should be understood that the above discussion is not intended to limit the scope of these embodiments. Various changes and modifications may be made without departing from the scope of the disclosure. For example, pressure sensors may include a variable capacitor and a lump inductor, or a variable capacitor and a variable inductor. Further, the dimensions and configurations of variable capacitor, lump inductor and variable inductor components are provided as examples of how embodiments may be implemented, and other dimensions and configurations may be utilized to suit pressure sensing specifications and applications.

Further, fabrication process parameters and steps may vary with fabrication of different capacitor and inductor configurations. Although embodiments are described with reference to a polymer, e.g., Parylene, flexible member and capacitor elements may be other materials, e.g., a biocompatible metal, and may be the same or different materials. Embodiments may also be utilized with variable capacitors having capacitor elements that are movable within channels formed in a substrate and with variable capacitors that are implemented without substrate channels.

[0076] Although reference is made to ocular implantation of a sensor without sutures by delivering the sensor through a needle, it should be understood that other minimally invasive implantation procedures and devices may be utilized as needed. For example, sensor devices may be implanted through corneal or scleral incisions of a suitable size. Sensor devices may also be implanted using tissue anchors or hooks. It should also be understood that embodiments may be utilized in various biomedical applications. Although reference is made to a microfabricated pressure sensor for passive monitoring of intraocular pressure using telemetry, embodiments may also be used or adapted for use in other applications including, but not limited to, monitoring pressure of other bodily fluids and physiological parameters such as monitoring pressure of blood within an aneurysm, monitoring pressure of cerebrospinal fluid and monitoring pressure in other biomedical applications.

Claims

1. A microfabricated implantable pressure sensor, comprising:

   a variable capacitor (200) including a substrate (210), a flexible member (220) having first and second edges (223, 224) disposed on a substrate (210) and a middle portion (225) extending between the first and second edges (223, 224), a chamber (226) being defined between the substrate (210) and the middle portion (225); and
   an inductor (1110, 1210, 1310, 1410) electrically connected to the variable capacitor (200),
   an electrical circuit including the variable capacitor and the inductor being configured to generate a detectable resonant frequency shift in response to a change of fluid pressure on an outer surface of the flexible member;
   characterized in that
   the substrate (210) defines a plurality of channels (216);
   the middle portion (225) is raised about the substrate (210); and
   the sensor further comprises a plurality of capacitor elements (232) extending indirectly from the flexible member (220), the plurality of capacitor elements (232) being movable within respective channels (216) with changes of fluid pressure on an outer surface of the flexible member (220), capacitance varying with changes in an overlapping area of the plurality of capacitor elements (232) and the substrate (210).

2. The pressure sensor of claim 1, the variable capacitor (200) and the inductor (1110, 1210, 1310, 1410) being configured for detection of fluid pressure changes with a sensitivity of about 1 mmHg within a fluid pressure range of about 1-50 mmHg.

3. The pressure sensor of claim 1, the inductor (1110, 1210, 1310, 1410) having a fixed inductance.

4. The pressure sensor of claim 3, the inductor (1210) including a stack of inductor elements (1211) positioned around the variable capacitor (200), inductor elements (1211) in the stack being separated from each other by a polymer material (1212).

5. The pressure sensor of claim 3, the inductor (1310) comprising metallic elements (1311) extending through the entire substrate (210).

6. The pressure sensor of claim 3, the inductor (1410) comprising a ring of inductor elements (1411) positioned around the variable capacitor (200).

7. The pressure sensor of claim 3, the variable capacitor (200) and the inductor (1410) being configured for intraocular implantation through a needle.

8. The pressure sensor of claim 7, the inductor (1410) being capable of assuming a compressed shape when positioned inside the needle and an expanded shape that is different than the compressed shape when deployed from the needle.

9. The pressure sensor of any one of claims 1 to 8, the flexibility of the middle portion (225) varying across a width of the middle portion, wherein non-linear deformation of the middle portion results in movement of the plurality of capacitor elements (232) within respective channels (216) in a direction that is perpendicular to a plane defined by a top surface of the substrate (210).

10. The pressure sensor of any one of claims 1 to 9, the sensor being sufficiently small in size so that it can be implanted through a clinical gauge needle and be implanted in various parts of an eye.

11. The pressure sensor of any one of claims 1 and 2, the inductor having a variable inductance.

12. The pressure sensor of any one of claims 1 to 11, further comprising a cross bar (234) carrying the plurality of capacitor elements (232); and an intermediate member (236) extending between the flexible member (220) and the cross bar (234).

Patentansprüche

1. Mikrofabrizierter implantierbarer Drucksensor, umfassend.

einen Drehkondensator (200), der ein Substrat (210), ein flexibles Element (220) mit auf einem Substrat (210) angeordneten ersten und zweiten Rändern (223, 224) und einem sich zwischen dem ersten und dem zweiten Rand (223, 224) erstreckenden Mittelteil (225) beinhaltet, wobei zwischen dem Substrat (210) und dem Mittelteil (225) eine Kammer (226) definiert ist; und

eine Induktorspule (1110, 1210, 1310, 1410), die elektrisch an den Drehkondensator (200) angeschlossen ist,

einen Stromkreis, der den Drehkondensator und die Induktorspule beinhaltet, der dazu gestaltet ist, eine erfassbare Resonanzfrequenzverschiebung in Reaktion auf eine Änderung von Fluiddruck an einer Außenfläche des flexiblen Elements zu erzeugen;

dadurch gekennzeichnet, dass

das Substrat (210) eine Vielzahl von Kanälen (216) definiert;

der Mittelteil (225) über dem Substrat (210) angehoben ist; und

der Sensor weiter eine Vielzahl von Kondensatorelementen (232) umfasst, die sich indirekt von dem flexiblen Element (220) erstrecken, wobei die Vielzahl von Kondensatorelementen (232) in jeweiligen Kanälen (216) bei Änderungen des Fluiddrucks an einer Außenfläche des flexiblen Elements (220) bewegbar sind, wobei die Kapazität bei Änderungen in einem überlappenden Gebiet der Vielzahl von Kondensatorelementen (232) und des Substrats (210) variiert.

2. Drucksensor nach Anspruch 1, wobei der Drehkondensator (200) und die Induktorspule (1110, 1210, 1310, 1410) zur Erfassung von Fluiddruckänderungen mit einer Empfindlichkeit von etwa 1 mmHg in einem Fluiddruckbereich von ungefähr 1-50 mmHg gestaltet ist.

3. Drucksensor nach Anspruch 1, wobei die Induktorspule (1110, 1210, 1310, 1410) eine festgelegte Induktivität aufweist.

4. Drucksensor nach Anspruch 3, wobei die Induktorspule (1210) einen Stapel von Induktorelementen (1211) beinhaltet, die um den Drehkondensator (200) herum positioniert sind, wobei Induktorelemente (1211) in dem Stapel durch ein Polymermaterial (1212) voneinander getrennt sind.

5. Drucksensor nach Anspruch 3, wobei die Induktorspule (1310) Metallelemente (1311) umfasst, die sich durch das gesamte Substrat (210) erstrecken.

6. Drucksensor nach Anspruch 3, wobei die Induktorspule (1410) einen Ring von Induktorelementen (1411) umfasst, die um den Drehkondensator (200) herum positioniert sind.

7. Drucksensor nach Anspruch 3, wobei der Drehkondensator (200) und die Induktorspule (1410) zur intraokularen Implantation durch eine Nadel gestaltet sind.

8. Drucksensor nach Anspruch 7, wobei die Induktorspule (1410) fähig ist, eine komprimierte Form anzunehmen, wenn sie innerhalb der Nadel positioniert ist, und eine expandierte Form, die von der komprimierten Form verschieden ist, wenn sie aus der Nadel heraus im Einsatz ist.

9. Drucksensor nach einem der Ansprüche 1 bis 8, wobei die Flexibilität des Mittelteils (225) über eine Breite des Mittelteils variiert, wobei eine nichtlineare Verformung des Mittelteils in einer Bewegung der Vielzahl von Kondensatorelementen (232) in jeweiligen Kanälen (216) in einer Richtung resultiert, die senkrecht zu einer von einer Oberseite des Substrats (210) definierten Ebene verläuft.

10. Drucksensor nach einem der Ansprüche 1 bis 9, wobei der Sensor klein genug ist, um durch eine Nadel von klinischer Dicke implantiert werden zu können und in verschiedenen Teilen eines Auges implantiert werden zu können.

11. Drucksensor nach einem der Ansprüche 1 und 2, wobei die Induktorspule eine variable Induktivität aufweist.

12. Drucksensor nach einem der Ansprüche 1 bis 11, weiter einen Querstab (234) umfassend, der die Vielzahl von Kondensatorelementen (232) trägt; und ein Zwischenelement (236), das sich zwischen dem flexiblen Element (220) und dem Querstab (234) erstreckt.

Revendications

1. Capteur de pression implantable micro-fabriqué, comprenant :

un condensateur variable (200) englobant un substrat (210), un membre flexible (220) possédant des premier et deuxième bords (223, 224) disposés sur un substrat (210) et une portion médiane (225) s'étendant entre les premier et deuxième bords (223, 224), une chambre (226) étant définie entre le substrat (210) et la portion médiane (225) ; et

un inducteur (1110, 1210, 1310, 1410) relié par voie électrique au condensateur variable (200) ;

un circuit électrique englobant le condensateur variable et l'inducteur étant configuré pour générer un déplacement détectable de fréquence de résonance en réponse à un changement de pression de fluide sur la surface externe du membre flexible ;

caractérisé en ce que

le substrat (210) définit plusieurs canaux (216) ;

la portion médiane (225) est surélevée par rapport au substrat (210) ; et

le capteur comprend en outre plusieurs éléments (232) faisant office de condensateurs s'étendant indirectement à partir du membre flexible (220), lesdits plusieurs éléments (232) faisant office de condensateurs étant mobiles au sein de canaux respectifs (216) en conformité avec des changements de pression de fluide sur la surface externe du membre flexible (220), la capacité variant en conformité avec des changements dans une zone de chevauchement desdits plusieurs éléments (232) faisant office de condensateurs et du substrat (210).

2. Capteur de pression selon la revendication 1, dans lequel le condensateur variable (200) et l'inducteur (1110, 1210, 1310, 1410) sont configurés pour détecter des changements de pression de fluide avec une sensibilité d'environ 1 mmHg dans une plage de pressions de fluide d'environ 1 à 50 mmHg.

3. Capteur de pression selon la revendication 1, dans lequel l'inducteur (1110, 1210, 1310, 1410) possède une inductance fixe.

4. Capteur de pression selon la revendication 3, dans lequel l'inducteur (1210) possède une pile d'éléments (1211) faisant office d'inducteurs disposés autour du condensateur variable (200), les éléments (1211) faisant office d'inducteurs dans la pile étant séparés les uns des autres par une matière polymère (1212).

5. Capteur de pression selon la revendication 3, dans lequel l'inducteur (1310) comprend des éléments métalliques (1311) s'étendant à travers l'ensemble du substrat (210).

6. Capteur de pression selon la revendication 3, dans lequel l'inducteur (1410) comprend un anneau d'éléments (1411) faisant office d'inducteurs disposés autour du condensateur variable (200).

7. Capteur de pression selon la revendication 3, dans lequel le condensateur variable (200) et l'inducteur (1410) sont configurés pour une implantation intraoculaire via une aiguille.

8. Capteur de pression selon la revendication 7, dans lequel l'inducteur (1410) est capable de prendre une configuration comprimée lorsqu'il est disposé à l'intérieur de l'aiguille et une configuration élargie qui est différente de la configuration comprimée après son déploiement à partir de l'aiguille.

9. Capteur de pression selon l'une quelconque des revendications 1 à 8, dans lequel la flexibilité de la portion médiane (225) varie sur la largeur de la portion médiane, une déformation non linéaire de la portion médiane donnant lieu au mouvement desdits plusieurs éléments (232) faisant office de condensateurs au sein de canaux respectifs (216) dans une direction qui est perpendiculaire à un plan défini par la surface supérieure du substrat (210).

10. Capteur de pression selon l'une quelconque des revendications 1 à 9, dans lequel le capteur possède une dimension suffisamment petite pour pouvoir être implanté via une aiguille de calibre clinique et pour pouvoir être implanté dans différentes parties d'un oeil.

11. Capteur de pression selon l'une quelconque des revendications 1 et 2, dans lequel l'inducteur possède une inductance variable.

12. Capteur de pression selon l'une quelconque des revendications 1 à 11, comprenant en outre une barre transversale (234) qui supporte lesdits plusieurs éléments (232) faisant office de condensateurs et un membre intermédiaire (236) qui s'étend entre le membre flexible (220) et la barre transversale (234).

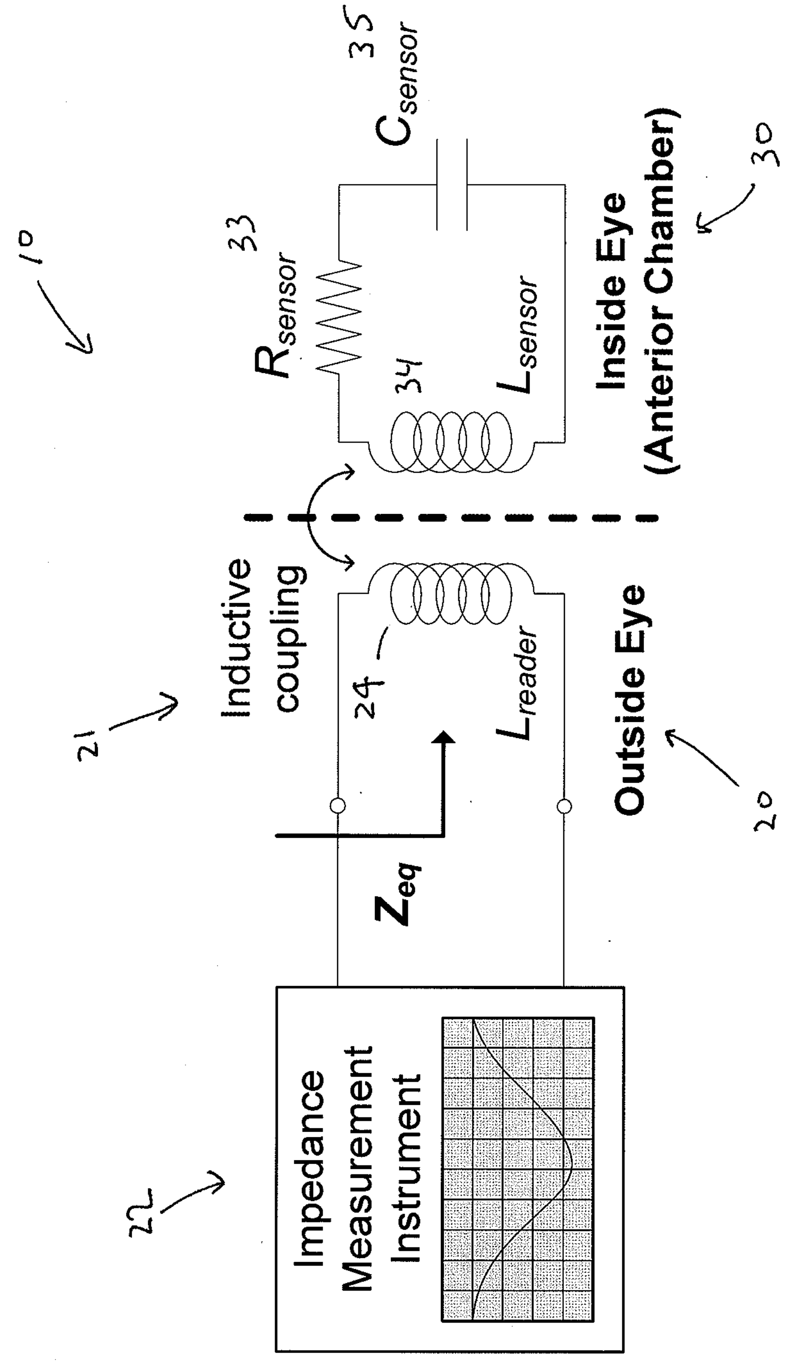


Figure 1
(Prior Art)

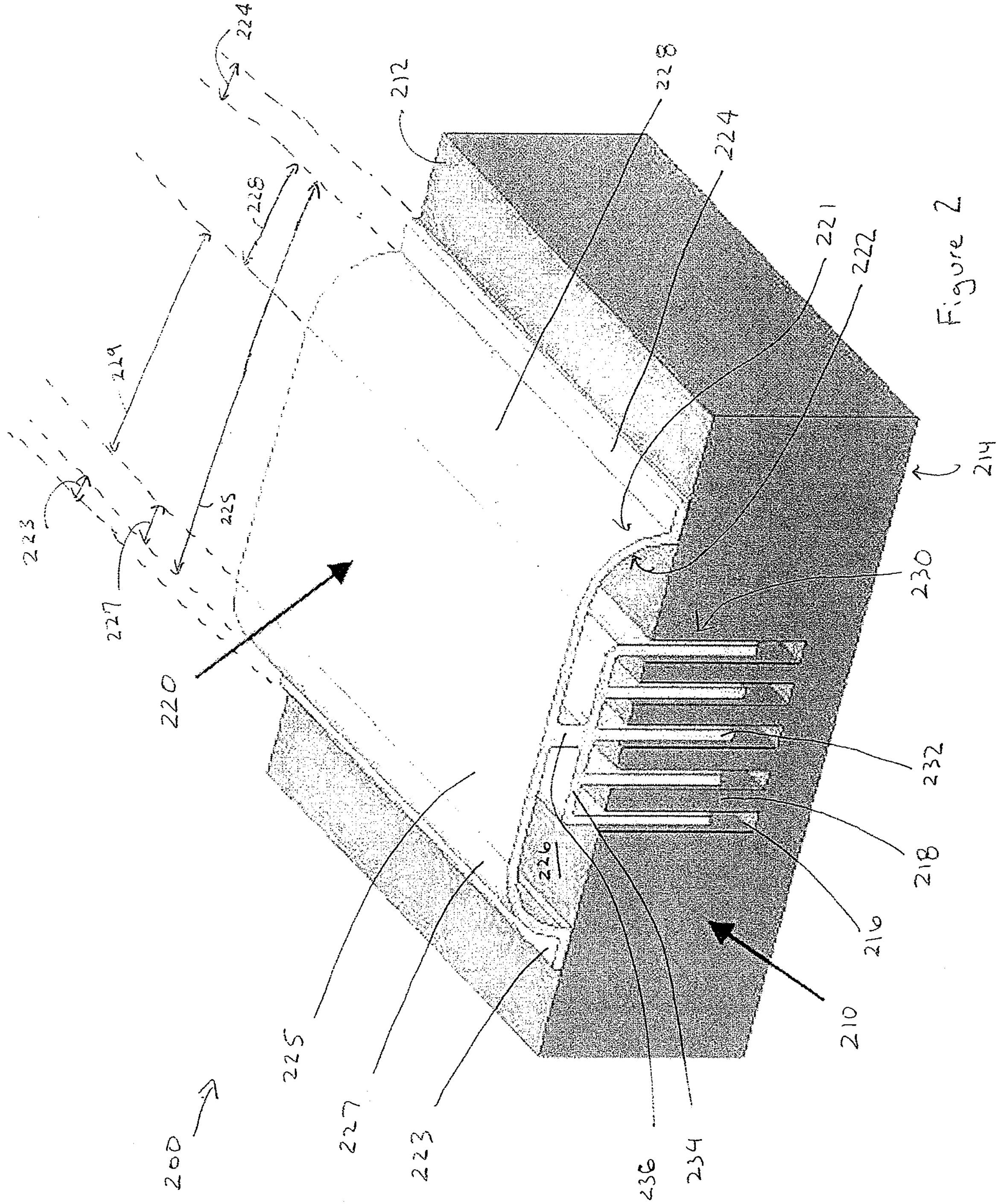

200
210
212
214
216
218
220
221
222
223
224
225
226
227
228
229
230
232
234
236

Figure 2

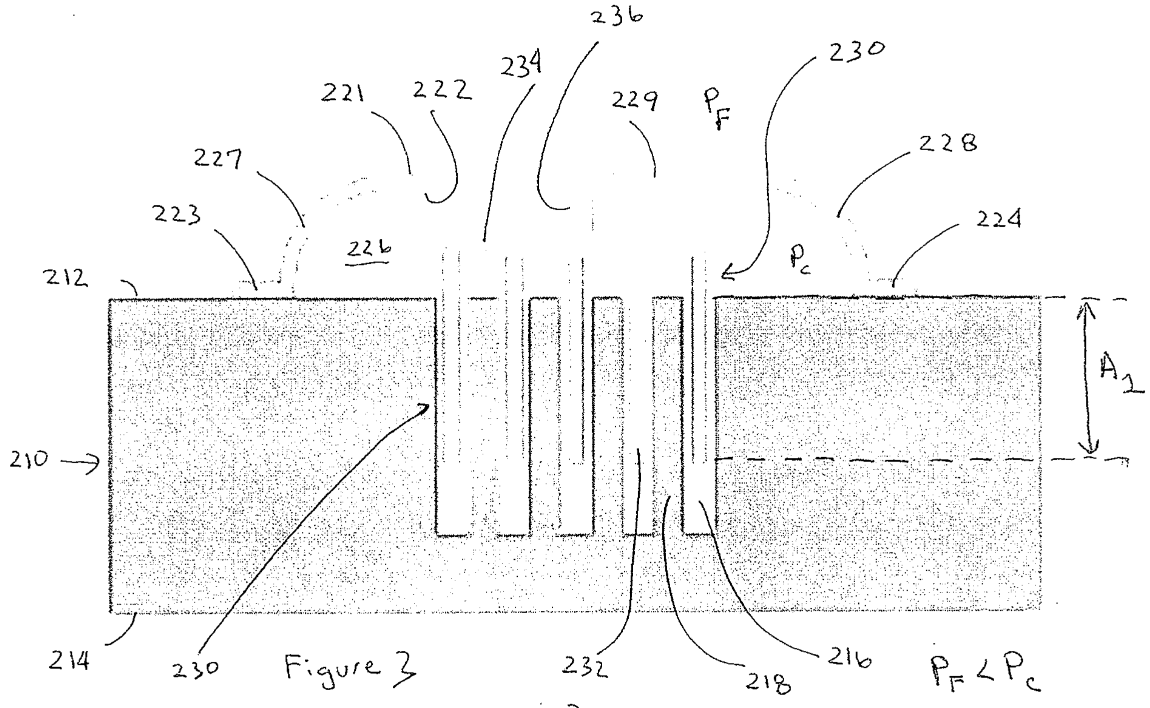


Figure 3

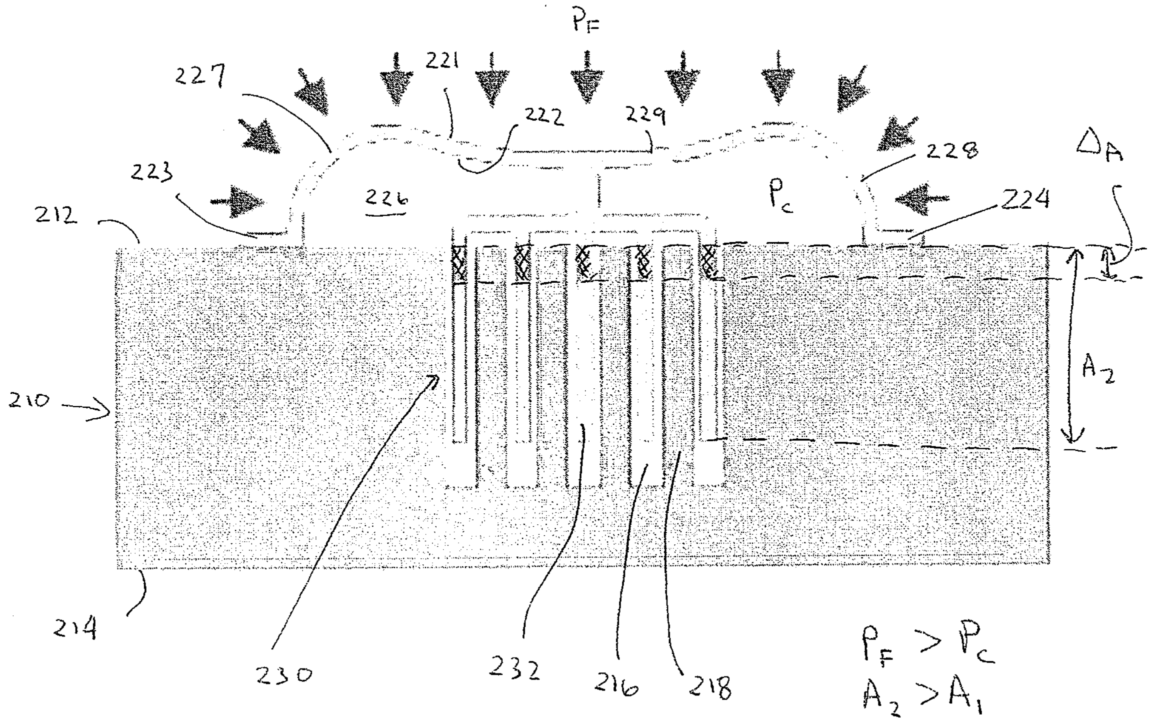


Figure 4

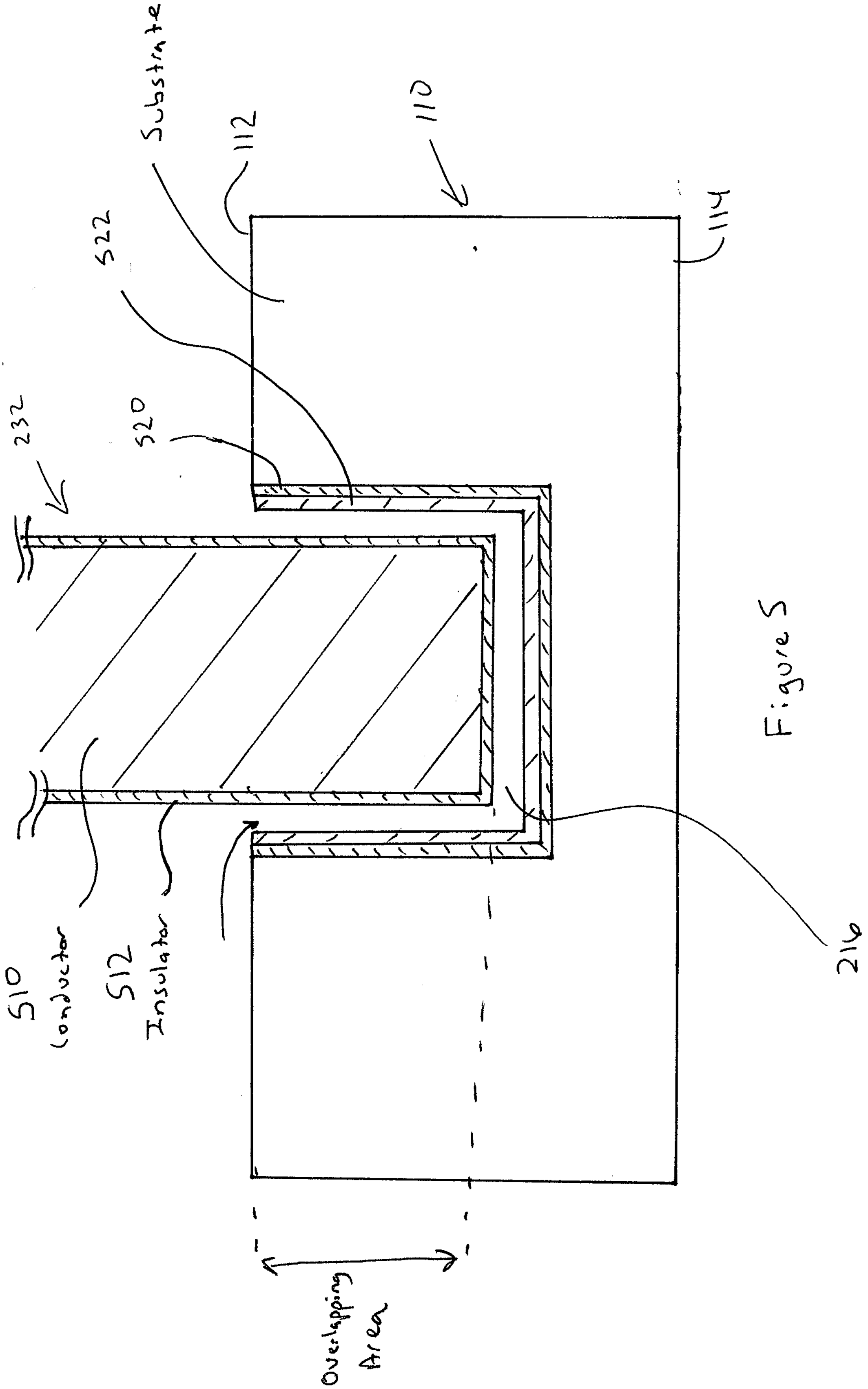
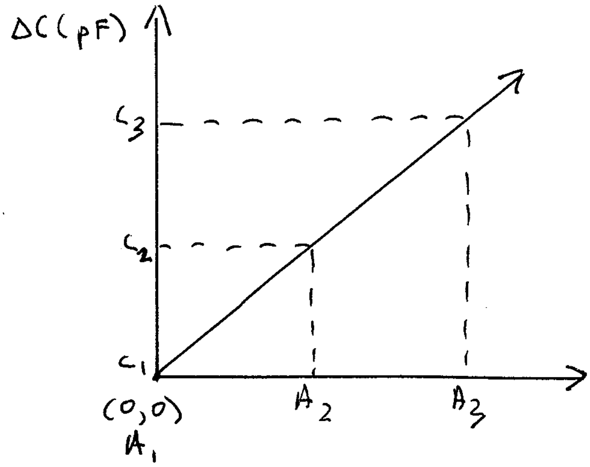
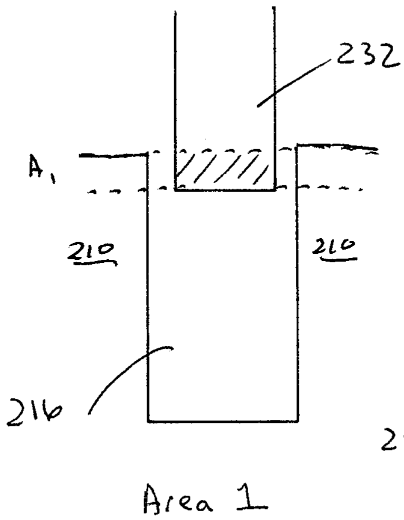

Substrate
112
110
114
522
520
232
510
Conductor
512
Insulator
216
Overlapping
Area

Figure 5

Figure 6





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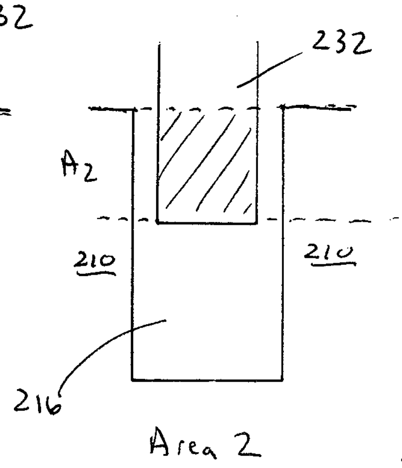
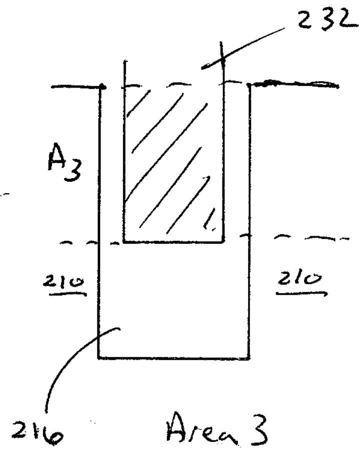


Figure 8



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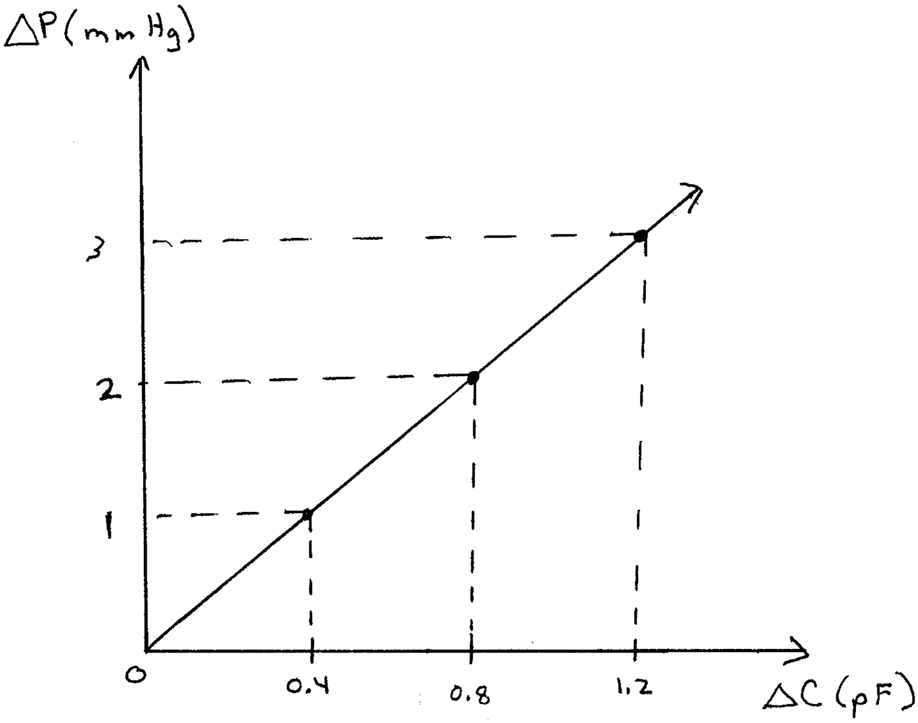

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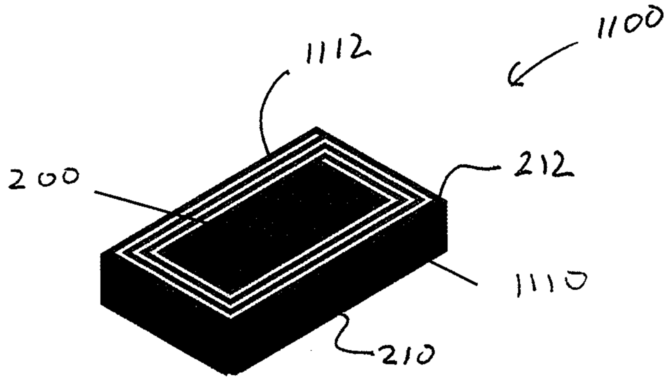


Figure 11

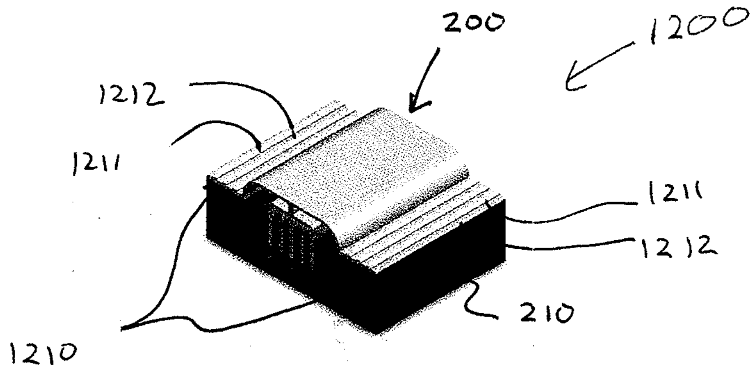


Figure 12

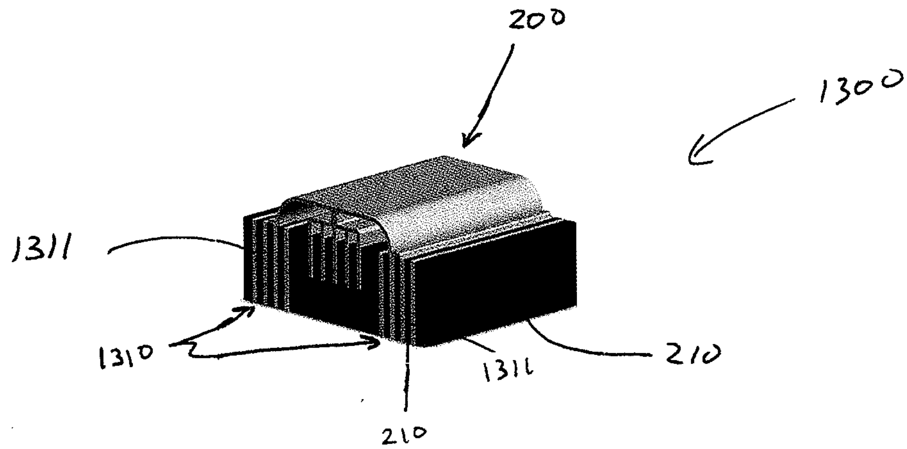


Figure 13

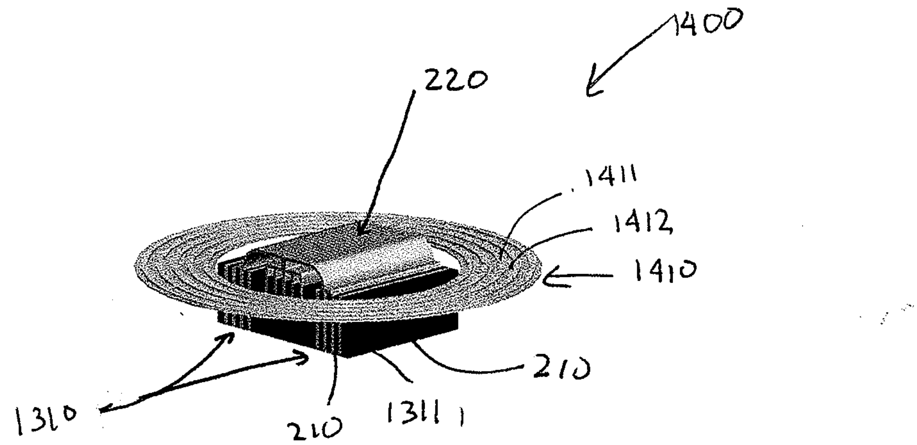


Figure 14

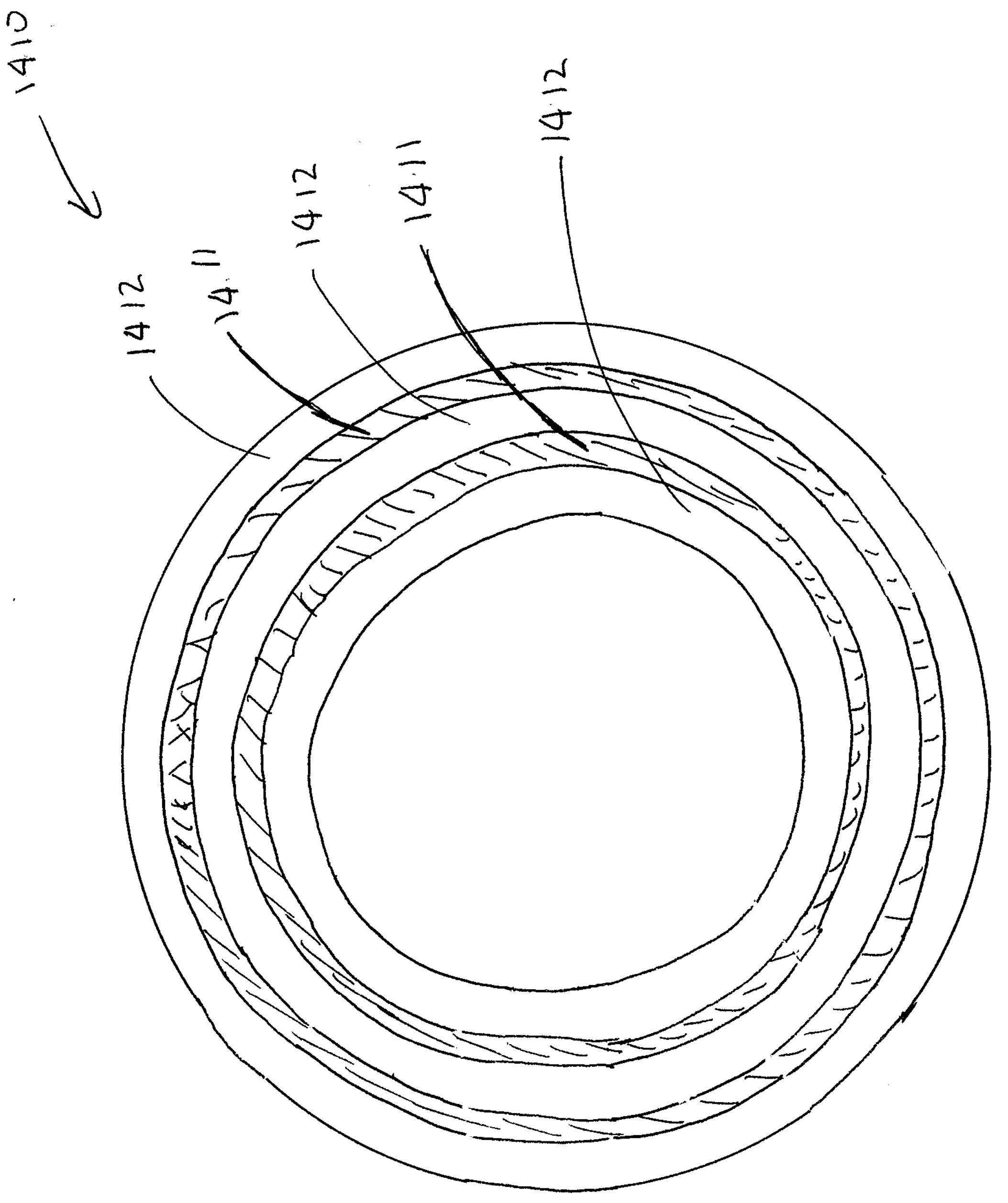


Figure 15

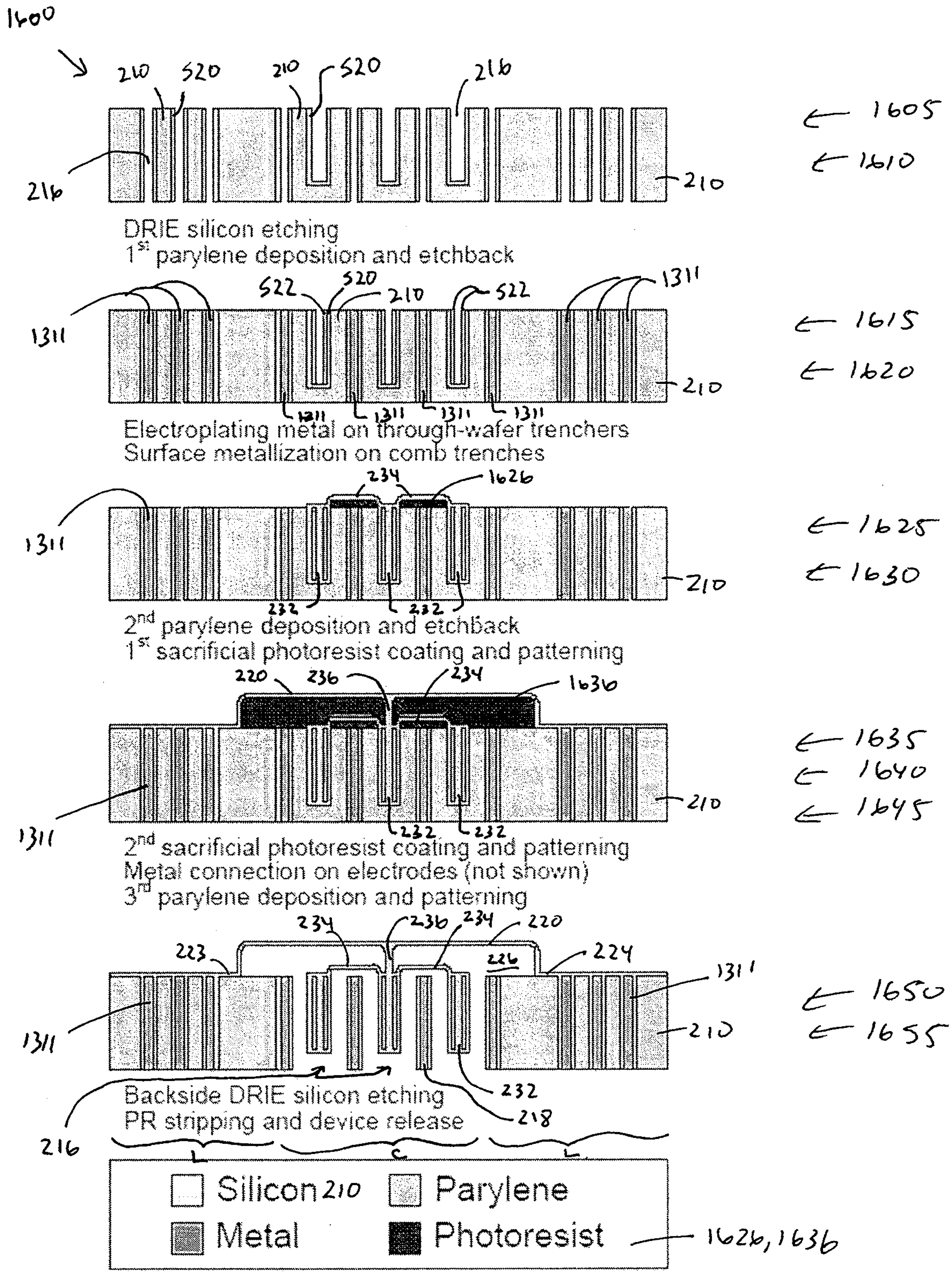

1600
210
520
210
520
216
216
210
1605
1610
DRIE silicon etching
1st parylene deposition and etchback
1311
522
520
210
522
1311
1311
210
1615
1620
1311
1311
1311
1311
Electroplating metal on through-wafer trenchers
Surface metallization on comb trenches
234
1626
1311
210
1625
1630
232
232
2nd parylene deposition and etchback
1st sacrificial photoresist coating and patterning
220
236
234
1636
1311
210
1635
1640
1645
232
232
2nd sacrificial photoresist coating and patterning
Metal connection on electrodes (not shown)
3rd parylene deposition and patterning
234
236
234
220
223
226
224
1311
1311
210
1650
1655
216
232
218
Backside DRIE silicon etching
PR stripping and device release
L
C
L
Silicon 210
Parylene
Metal
Photoresist
1626, 1636

Figure 16A

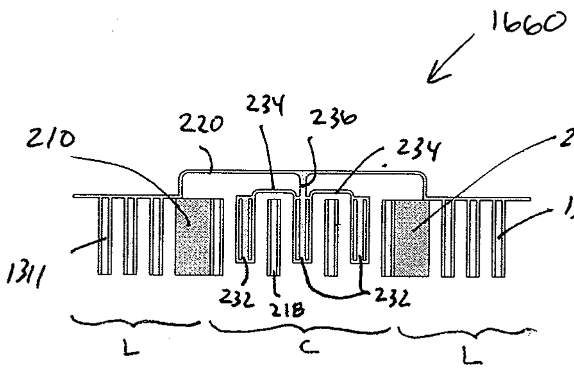


Figure 16 B

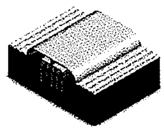
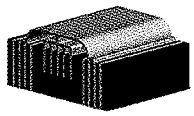
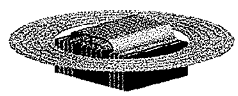
Pressure sensitivity	**1 mmHg**		
Detection range	**1–50 mmHg**		
Design	A (1200)	B (1300)	C (1400)
Schematic			
Variable capacitor scheme	Comb plates (vertical movement)	Comb plates (vertical movement)	Comb plates (vertical movement)
Lump inductor scheme	Surface-micromachined (thin lines)	Embedded (HAR thick lines)	Embedded + rollable
Overall implant size	0.5 x 0.5 x 3 mm^3	0.5 x 0.5 x 3 mm^3	0.5 x 0.5 x 4 mm^3 (after rolled)
Capacitance	~ 25 pF	~ 31 pF	~ 127 pF
Capacitance change (ΔP = 1 mmHg)	~ 0.4 pF	~ 0.4 pF	~ 0.4 pF
Inductance	~ 40 nH	~ 10 nH	~ 145 nH
Resistance	~ 2.8 Ω	~ 0.03 Ω	~ 3.8 Ω
Resonance frequency f_R	**~ 159 MHz**	**~ 286 MHz**	**~ 37 MHz**
f_R shift Δf (ΔP = 1 mmHg)	**~ 1.6 MHz**	**~ 1.8 MHz**	**~ 59 kHz**
$\Delta f / f_R$	**~ 1 x 10^{-2}**	**~ 6.4 x 10^{-3}**	**~ 1.6 x 10^{-3}**
Q factor	**~ 14**	**~ 600**	**~ 9**

Figure 17

REFERENCES CITED IN THE DESCRIPTION

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- US 20060247664 A **[0060] [0074]**

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- **ROSENGREN.** *A system for wireless intra-ocular pressure measurements using a silicon micromachined sensor* **[0008]**
- **ROSENGREN.** *A system for passive implantable pressure sensors* **[0008]**
- **DUCK-BONG SEO.** *Design and Simulation of a MEMS-Based Comb-Drive Pressure Sensor for Pediatric Post-Operative Monitoring Applications* **[0009]**
- **DUCK-BONG SEO et al.** *DESIGN AND SIMULATION OF A MEMS-BASED COMB-DRIVE PRESSURE SENSOR FOR PEDIATRIC POST-OPERATIVE MONITORING APPLICATIONS,* 01 January 2003 **[0012]**

patsnap

专利名称(译)	微加工植入式无线压力传感器，用于生物医学应用和压力测量和传感器植入方法		
公开(公告)号	[EP2786701B1](EP2786701B1)	公开(公告)日	2015-12-23
申请号	EP2014171832	申请日	2007-08-29
[标]申请(专利权)人(译)	加州理工学院		
申请(专利权)人(译)	加州理工大学		
当前申请(专利权)人(译)	加州理工大学		
[标]发明人	TAI YU CHONG CHEN PO JUI RODGER DAMIEN C HUMAYUN MARK S		
发明人	TAI, YU-CHONG CHEN, PO-JUI RODGER, DAMIEN C. HUMAYUN, MARK S.		
IPC分类号	A61B3/16 A61B5/03 A61B5/00 G01L9/00 A61B5/0215		
CPC分类号	A61B3/16 A61B5/0031 A61B5/0215 A61B5/03 G01L9/0072		
优先权	60/841113 2006-08-29 US		
其他公开文献	EP2786701A3 EP2786701A2		
外部链接	Espacenet		

摘要(译)

可变电容器，包括可变电容器和电感器的微制造可植入压力传感器，以及相关的压力测量和植入方法。电感器可以具有固定或可变的电感。可变电容器和压力传感器包括柔性构件，该柔性构件设置在基板上并限定腔室。电容器元件间接地从柔性构件延伸。施加到柔性构件的外表面的足够的流体压力导致柔性构件移动或变形，从而导致电容和/或电感改变。可以检测共振频率或阻抗的所得变化以确定压力，例如眼内压。