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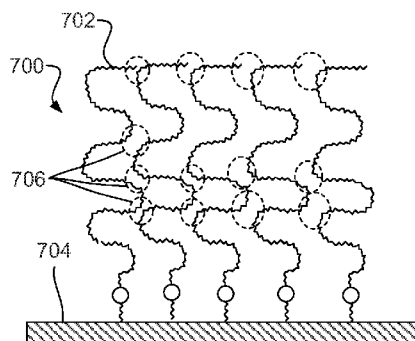


Fig. 7

(57) Abstract: An implantable device has an elongated body with an electrically insulating material having a hydrophobic surface. A polymer brush has hydrophilic polymer chains tethered to at least a portion of the hydrophobic surface, such that the polymer brush reduces sliding friction between the elongated body and biological tissue of an internal wall of a fluid filled cavity.



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Low Friction Implantable Device

BACKGROUND

[0001] In human hearing, hair cells in the cochlea respond to sound waves and produce corresponding auditory nerve impulses. These nerve impulses are then conducted to the brain and perceived as sound.

[0002] Hearing loss, which may be due to many different causes, is generally of two types: conductive and sensorineural. Conductive hearing loss typically occurs where the normal mechanical pathways for sound to reach the hair cells in the cochlea are impeded, for example, from damage to the ossicles. Conductive hearing loss may often be helped by using conventional hearing aids that amplify sounds so that acoustic information can reach the cochlea and the hair cells. Some types of conductive hearing loss are also amenable to alleviation by surgical procedures.

[0003] Many people who are profoundly deaf, however, have sensorineural hearing loss. This type of hearing loss can arise from the absence or the destruction of the hair cells in the cochlea which then no longer convert acoustic signals into auditory nerve impulses. Individuals with complete sensorineural hearing loss are unable to derive any benefit from conventional hearing aid systems no matter how loud the acoustic stimulus is. This is because the mechanism for converting sound energy into auditory nerve impulses has been damaged. Thus, in the absence of properly functioning hair cells, auditory nerve impulses cannot be generated directly from sounds.

[0004] To overcome sensorineural deafness, cochlear implant systems or cochlear prostheses have been developed that can bypass the hair cells located in the vicinity of the radially outer wall of the cochlea by presenting electrical stimulation directly to the auditory nerve fibers. This leads to the perception of sound in the brain and provides at least partial restoration of hearing function. Thus, most of these cochlear prosthesis systems treat sensorineural deficit by stimulating the ganglion cells in the cochlea directly using an implanted lead that has an electrode array. Thus, a cochlear

5 prosthesis operates by directly stimulating the auditory nerve cells and bypassing the defective cochlear hair cells that normally convert acoustic energy into electrical activity to the connected auditory nerve cells.

BRIEF DESCRIPTION OF THE DRAWINGS

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[0005] The accompanying drawings illustrate various examples of the principles described herein and are a part of the specification. The illustrated examples are merely examples and do not limit the scope of the claims.

15 [0006] Fig. 1 is a diagram of an example of a cochlear implant system according to principles described herein.

[0007] Fig. 2 is a diagram of an example of an implantable device during an insertion procedure according to principles described herein.

20 [0008] Fig. 3 is a diagram of an example of a graph depicting a low friction characteristic of an implantable device according to principles described herein.

[0009] Fig. 4 is a diagram of an example of a cochlea showing an internal structure of the cochlea and insertion location of an implantable device according to principles described herein.

25 [0010] Fig. 5 is a diagram of an example of electrodes on an implantable device generating electric fields adjacent to polymer brushes formed on the implantable device according to principles described herein.

[0011] Fig. 6 is a diagram of an example of a polymer brush according to principles described herein.

30 [0012] Fig. 7 is a diagram of an example of a polymer brush according to principles described herein.

DETAILED DESCRIPTION

35 [0013] The insertion of the electrode into the cochlea is typically performed by creating an opening in the cochlea and then inserting the electrode array through the opening into the cochlea. This insertion can

5 generate forces that are large enough to damage the sensitive tissues within the cochlea. This damage can further reduce any residual hearing capability of the patient.

[0014] If hearing can be preserved, many advantages are afforded to the patient such as 1) the possibility of combined acoustic-electric hearing for
10 better hearing in noise and music enjoyment and 2) the preservation of sensory structures that produce homeostatic levels of various growth factors that help to support spiral ganglion cell survival. Further by keeping the sensory structures intact, the patient is not foreclosed from pursuing alternative medical treatments to use the sensory structures, as such treatments are developed.

15 [0015] In order to increase the probability of hearing preservation in the face of electrode insertion trauma the forces exerted during insertion should, at a minimum, not exceed the mechanical limits of the tissue residing within the inner ear. It should be pointed out that the structures responsible for hearing transduction can be irreversibly damaged by forces that are below the threshold
20 of tactile detection during manual insertion.

[0016] The principles described herein include a low friction, implantable device that lowers the insertion force when implanting the device. Such an implantable device is well suited for cochlear implants because insertion of a cochlear implant can generate high friction forces due to the
25 curvature of the cochlear wall. The friction between the implant's surface and the cochlear wall significantly increases as the implant is inserted deeper into the cochlea, in part because more of the implant's surface progressively comes into contact with the inward curvature as the implant slides deeper into the cochlea.

30 [0017] Such an implantable device has an elongated body with an electrically insulating material having a hydrophobic surface. A polymer brush has hydrophilic polymer chains tethered to at least a portion of the hydrophobic surface, such that the polymer brush reduces sliding friction between the elongated body and biological tissue of an internal wall of a fluid filled cavity.

35 [0018] In the following description, for purposes of explanation, numerous specific details are set forth in order to provide a thorough

5 understanding of the present systems and methods. It will be apparent, however, to one skilled in the art that the present apparatus, systems, and methods may be practiced without these specific details. Reference in the specification to "an example" or similar language means that a particular feature, structure, or characteristic described is included in at least that one
10 example, but not necessarily in other examples.

[0019] Fig. 1 is a diagram of an example of a cochlear implant system (100) surgically placed within the patient's auditory system. Ordinarily, sound enters the outer ear (110) and is directed into the auditory canal (120) where the sound wave vibrates the tympanic membrane (130). The motion of the
15 tympanic membrane is amplified and transmitted through the ossicular chain (140) which consists of three bones in the middle ear. The third of the ossicles, or stirrup, (145) contacts the outer surface of the cochlea (150) and causes movement of the fluid within the cochlea (150). Cochlear hair cells respond to the fluid-borne vibration in the cochlea (150) and trigger neural electrical signals
20 that are conducted from the cochlea (150) to the auditory cortex by the auditory nerve (160).

[0020] As indicated above, the cochlear implant (100) is a surgically implanted electronic device that provides a sense of sound to a person who is profoundly deaf or severely hard of hearing. As also noted above, in many
25 cases, deafness is caused by the absence or destruction of the hair cells in the cochlea, i.e., sensorineural hearing loss. In the absence of properly functioning hair cells, there is no way auditory nerve impulses can be directly generated from ambient sound. Thus, conventional hearing aids, which amplify external sound waves, provide no benefit to persons suffering from complete
30 sensorineural hearing loss.

[0021] Unlike hearing aids, the cochlear implant (100) does not amplify sound, but works by directly stimulating any functioning auditory nerve cells inside the cochlea (150) with electrical impulses. Consequently, providing a cochlear prosthesis typically involves the implantation of an electrode array
35 into the cochlea. The cochlear prosthesis operates by direct electrical

5 stimulation of the auditory nerve cells, bypassing the defective cochlear hair cells that normally traduce acoustic energy into electrical energy.

[0022] External components of the cochlear implant include a microphone (170), speech processor (175), and transmitter (180). The microphone (170) picks up sound from the environment and converts it into
10 electrical impulses. The speech processor (175) selectively filters and manipulates the electrical impulses and sends the processed electrical signals through a cable to the transmitter (180). The transmitter (180) receives the processed electrical signals from the processor (175) and transmits them to the receiver (185) by electromagnetic, radio frequencies, optical communication,
15 and/or other wireless communication technology.

[0023] The internal components of the cochlear implant may include an antenna (187) and an internal processor (185). The antenna (187) and internal processor (185) are secured beneath the user's skin, typically above and behind the outer ear (110). The internal processor (185) includes electronic
20 circuitry housed in a hermetically sealed enclosure. This electronic circuitry is connected to via a hermetically sealed feedthrough to the antenna (187). The antenna (187) receives power and signals from the transmitter (180) via electromagnetic induction, radio frequency signals, optical communication, and/or other wireless communication. The internal processor (185) processes
25 the received signals and sends modified signals through the hermetic feedthrough to elongated body (190) and electrode array (195). The electrode array (195) is wound through the cochlea (150) and provides direct electrical stimulation to the auditory nerve inside the cochlea (150) which bypasses the normal mechanics of hearing and results in synaptic excitation of the auditory
30 nerve. A polymer brush (197) is attached to an outer surface of the electrode array (195) that reduces the insertion forces when inserting the electrode array (195) into the cochlea. The polymer brush will be described in more detail below.

[0024] The implant works by using the tonotopic organization of the
35 basilar membrane of the inner ear. The tonotopic organization, also referred to as "frequency-to-place" mapping, is the way the ear differentiates between

5 sounds of different frequencies. In a normal ear, sound vibrations in the air are converted into resonant vibrations of the liquid within the cochlea. High-frequency sounds do not pass very far through the liquid and the structures of the cochlea that contain the liquid. Low-frequency sounds pass farther down the cochlear channels. Consequently, the nerve cells at the basal end of the
10 cochlear spiral sense higher frequencies, while progressively lower frequencies are sensed at different portions of the cochlear spiral moving towards the apex. The movement of hair cells located all along the basilar membrane stimulates the surrounding nerve cells which conduct electrical impulses to the brain. The brain is able to interpret the nerve activity to determine which area of the basilar
15 membrane is resonating and, therefore, what sound frequencies are being heard.

[0025] For individuals with sensorineural hearing loss, hair cells are often fewer in number and/or damaged. The cochlear implant bypasses the hair cells and stimulates the cochlear nerves directly using electrical impulses. The
20 cochlear implant stimulates different portions of the cochlea (150) according to the sound detected by the microphone (170), just as a normal functioning ear would experience stimulation at different portions of the cochlea depending on the frequency of sound vibrating the liquid within the cochlea (150). This allows the brain to interpret the frequency of the sound as if the hair cells of the basilar
25 membrane were functioning properly.

[0026] Fig. 2 is a diagram of an example of an implantable device (200) during an insertion procedure. Here, the implantable device (200) is a cochlear implant that has an electrode array (202) supported by an elongated body (204). Each electrode (206) is secured within an electrically insulating
30 material (208) of the elongated body (204). The electrically insulating material prevents shorting between the electrodes (206). When inserted into scala tympani (209), the electrodes (206) are positioned in close proximity to the nerve cells that are to be stimulated, which are located towards the center of the cochlea (210). This allows for optimal selectivity and maximal stimulation
35 efficiency.

5 **[0027]** As the electrode array is pushed into scala tympani (209), the distal end (212) of the array makes contact with the cochlea's internal wall (214). The internal wall (214) has an inward curvature (216), and the distal end (212) is shaped to guide the elongated body (204) along the curvature (216) such that the elongated body bends inward as the elongated body (204) moves
10 deeper into the cochlea (210).

[0028] During the insertion procedure, the insertion forces are resisted by friction at multiple locations on the elongated body (204). A hole is cut into a part of the cochlea (210) through which the elongated body (204) is inserted. Friction may occur at the interface of the hole and the surface of the elongated
15 body. Also, the length (218) and surface area of the elongated body (204) in contact with the internal wall (214) increase generate friction. This friction increases as more of the length (218) makes contact with the internal wall (214) and progressively increases during the insertion procedure. Increased friction may cause the electrode array to bend, which may cause the electrode array to
20 buckle and prevent it from reaching the desired insertion depth. In some case, the friction can cause the electrode array to bend enough that it is misguided into a lumen other than the scala tympani.

[0029] Bioresponses are triggered merely by introducing a foreign material into a biological environment of, for example, the human body. Such
25 bioresponses are the body's mechanism for protecting itself from harmful foreign materials and are called foreign body reactions (FBR). When inserting an implant however, a spectrum of responses to the injury caused by insertion, inflammation, and wound healing in addition to the foreign body reactions and fibrous encapsulation is expected. Formation of a fibrous capsule around the
30 electrode of an active device (e.g. cochlear implant) acts as a barrier between the electrodes of the implant and the target nerves and can lead to a degradation of the performance of the device by increasing the electrodes interface impedance. Since bioresponse to materials is a surface dominated phenomenon by modifying the surface of a cochlear implant, both the
35 mechanics of insertion and the invivo tissue response can be optimized to improve the surgical outcome.

5 **[0030]** The electrically insulating material (208) of the elongated body (204) is usually made of silicone which is a hydrophobic material. The hydrophobic surface of the silicone repels water, a constituent of the fluid within the cochlea's cavities (209). Domination of the hydrophobic surface interactions hinder formation of a stable lubricating film, and increase friction between the internal wall (214) and the elongated body. The present inventor has found that by coating the elongated body (204) with a hydrophilic polymer brush (220) that attracts the fluid within the cavity (209) to create a lubricating layer between the internal wall (214) and the elongated body (204). Not only does the hydrophilic polymer brush minimize the friction between the internal wall (214) and the elongated body (204), but it also reduces friction's downstream effects, such as damage to the sensory structures within the cochlea (210) and internal wall (214) and consequently minimizes the insertion trauma. Also, the polymer brush (220) is thin and flexible enough so that intended mechanical requirements of the elongated body (204) remains intact. This allows the elongated body (204) to comply and bend as being inserted into the desired cavity of the cochlea. Further, the tissue response to the implant is mitigated by presence of this hydrophilic layer, resulting into no or minimal encapsulation (204). In absence of a thick fibrous tissue, the charge injected by the electrodes (206) is efficiently transferred to the target nerve cells.

25 **[0031]** The polymer brush (220) is made of polymer chains that are covalently tethered to the surface of the hydrophobic, electrically insulating material (208) of the elongated body (204). The properties of the polymer brush (220) that aid with wet lubricity, the ability to reduce friction when wet, are affected by the polymer chain density, polymer chain length, molecular weight of the polymer chain, the hydrophilicity of individual monomers in the polymer chains, monomer type, other properties, or combinations thereof.

[0032] In the example of Fig. 2, the polymer brush (220) is attached to the entire surface area of the portion of the elongated body (204) that is intended to be implanted within the cochlea (210). In other examples, the polymer brush (220) is attached to only the distal end (212) and the sliding side (222) of the elongated body (204), where frictional forces play a role during an

5 insertion procedure. In some examples, the polymer brush's properties are uniform throughout the surface. However, in alternative examples, the hydrophilicity, polymer chain length, or other properties of the polymer chain (220) are different depending on the area of the elongated body (204) to which the polymer brush (220) is attached. A suitable elongated body (204)
10 compatible with the principles described herein is disclosed in U.S. Patent Application Serial No. 12/781,137, which is herein incorporated by reference for all that it contains.

[0033] The elongated body (204) has a stiffness profile that allows the elongated body (204) to bend inward when sliding due to an inward curve of an
15 internal wall (214) of the human cochlea (210). The polymer brush (220) is flexible enough not to substantially interfere with this well-designed stiffness profile. For example, the stiffness profile of the elongated body (204) before applying the polymer brush (220) may be similar to the designed stiffness profile after the polymer brush (220) is attached. Thus, the application of the polymer
20 brush (220) has a minimal if any affect on the elongated body's stiffness profile.

[0034] Fig. 3 is a diagram of an example of a graph (300) depicting a low friction characteristic of an implantable device according to principles described herein. In this example, the y-axis (302) of the chart schematically represents total force exerted to the cochlea during insertion of electrode array
25 through a cochleostomy into the cochlear cavity. The x-axis (304) schematically represents the vertical travel, which translates into the depth of insertion of an electrode array inside the cochlea. Line (306) schematically represents the forces generated with an elongated body having a hydrophobic surface. As is generally depicted, the insertion force needed to push the elongated body
30 deeper into the cochlea increases as the insertion distance increases. Line (308) schematically represents the forces generated with an elongated body having a hydrophilic polymer brush attached to the hydrophobic surface of the elongated body. The friction and resulting insertion force increased in direct proportion to the contact area between the electrode array and the model
35 surface area used in the test until the insertion forces exceeded the electrode arrays' column strength and caused the electrode array to buckle. Initial testing

5 has demonstrated that a surface modified electrode array can be inserted deeper into the cochlea with much less force exerted to the structure.

[0035] As is generally depicted, an exponential relationship exists between the insertion force and the insertion distance. This relationship reflects the increasing contact area between the elongated body and the internal wall of the cochlea, and the increased friction due to the increased contact area.

[0036] Fig. 4 is a diagram of an example of a cochlea (400) showing an internal structure of the cochlea (400) and insertion location (402) of an implantable device (404) according to principles described herein. The internal walls of the cochlea (400) are made of bone, with a thin, delicate lining of epithelial tissue. The primary structure of the cochlea (400) is a hollow tube that is helically coiled, similar to a snail shell. The coiled tube is divided into three fluid-filled spaces (scalae), the scala vestibuli (406), the scala tympani (408), and the scala media (410). Reissner's membrane, or the vestibular membrane, (412) separates the scala media (410) from the scala vestibuli (406), and the basilar membrane (414) separates the scala tympani (408) from the scala media (410). The scala vestibuli (406) joins the scala tympani (408) at the apex of the cochlea.

[0037] The cochlea (400) is filled with a watery liquid, which moves in response to the vibrations coming from the middle ear via the stirrup (145, Fig. 1). As the fluid moves, thousands of "hair cells" (416) in a normal, functioning cochlea are set in motion and convert that motion to electrical signals that are communicated via neurotransmitters to many thousands of nerve cells (418). These primary auditory neurons (418) transform the signals into electrical impulses known as action potentials, which travel along the auditory nerve cells (418) to structures in the brainstem for further processing. The terminal end of the elongated body (190, Fig. 1) is inserted into the scala tympani with the electrode array (195, Fig. 1) being positioned in close proximity to the nerve cells (418).

[0038] The hydrophilic polymer brush (420) forms a barrier that prevents proteins and other constituents of the cochlea's fluid from reaching the hydrophobic surface of the elongated body (404), and consequently, eliminating

5 the nonspecific adsorption of proteins which is a precursor to FBR and fibrous tissue formation.

[0039] Fig. 5 is a diagram of an example of electrodes (500) on an implantable device (502) generating electric fields (504) adjacent to a polymer brush (506) formed on the implantable device (502) according to principles
10 described herein. The electrodes (500) are in communication with a power source that is part of the cochlear implant system (100, Fig. 1), such as a power source connected to the system's transmitter (180, Fig. 1). Here, the polymer brush is formed on the sliding side (508) of the implantable device (502) and on an activation side (510) of the implantable device (502).

15 [0040] The implanted intracochlear electrodes demonstrate a very special electrode-electrolyte system. Bioresponses to the implant and surgical trauma lead to many changes in the electrolyte surrounding the implant, including the chemical constituents and the amount of electrolyte adjacent to the electrodes. The fluid in the cochlea contains mobile ions which transfer the
20 charge from the electrodes to the target neuron. Therefore absence of the electrolyte surrounding the electrodes will increase the series resistance of the system and adversely affect the performance of the device. Hydrophobicity of the polymer brush modified surfaces (506) on the activation side (510) of the implantable device (502) promotes cochlea fluid retention around the electrodes
25 (500) and enables optimal charge delivery which in turn

[0041] Fig. 6 is a diagram of an example of a polymer brush (600) according to principles described herein. In this example, each of the polymer chains (602) of the polymer brush (600) is covalently bonded to the hydrophobic surface (604). A monomer (608) attached to an attachment end (610) of the
30 polymer chains (602) forms the covalent bond between the polymer chain (602) and the hydrophobic surface (604). In some examples, an ionic or another kind of bond is formed between the hydrophobic surface and the polymer chain, and a covalent bond is formed later in a subsequent step. For example, exposure to ultraviolet wavelengths may convert the ionic bond to a covalent bond.

35 [0042] The polymer chains (602) may be made of any polymer that provides a hydrophilic effect to at least the free ends (612) of the polymer chain

5 to improve lubricity of the implantable device. A non-exhaustive list of such hydrophilic polymers includes polyethylene glycol, dextran, polyacrylamide, polymethacrylic acid, hydrophilic proteins, mucins, hyaluronic acid, other hydrophilic polymers, or combinations thereof.

[0043] The polymer brush (600) may be formed using a "grafting to" approach, a "grafting from" approach, a hybrid approach, other approaches, or combinations thereof. A "grafting to" approach includes attaching pre-synthesized polymers to the hydrophobic surface through anionic polymerization, cationic polymerization, chemisorptions, other attachment methods, or combinations thereof. In one example, the hydrophobic surface
10 may be dipped into a solution of an end functionalized, pre-synthesized polymers, where the polymers in the solution bind directly to the hydrophobic surface. The "grafting from" approach includes growing the polymer chains layer by layer off of the hydrophobic surface.

[0044] The polymer brush (600) has a polymer chain density such that
20 each of the polymer chains is attached on one end and the rest of the chain is forced to stretch away from the hydrophobic surface. Such a density prevents the polymer chains (602) from folding over, binding to itself, or lying over the hydrophobic surface (604). In some examples, the hydrophilic monomers bind to water molecules in the cochlear fluid to form hydrate monomers. The free
25 ends (612) of the polymer chains may bind with water molecules to form a super-hydrated layer (614) at the outer most portion of the polymer brush (600). Such a super-hydrated layer (614) may be a lubricating layer that reduces friction between the polymer brush (600) and the tissues in the fluid filled cavities of the cochlea.

[0045] In some examples, all of the polymer chains (602) in the polymer brush (600) are hydrophilic. In other examples, at least 95 percent of the composition of the polymer brush (600) has polymer chains (602) that are hydrophilic. In yet other examples, at least 75 percent of the composition of the polymer brush (600) has polymer chains (602) that are hydrophilic.

[0046] In examples where a hydrophilic polymer brush is attached to a
35 hydrophobic surface made of silicone, the implantable device has the

5 advantage of modifying just the surface properties without altering silicone's bulk properties. Silicone has several desirable characteristics, such as its electrical insulation, bioinertness, mechanical compliance, and ease of fabrication. With the principles described herein, these desirable properties of silicone are retained in the implantable device, while silicone's undesirable
10 surface properties are optimized.

[0047] The polymer brush (600) may have polymer chains (602) with a length in the nanometer scale. However, any polymer chain length is within the principles described herein.

[0048] Fig. 7 is a diagram of an example of a polymer brush (700)
15 according to principles described herein. In this example, the polymer chains (702) are tethered to the hydrophobic surface (704) and are cross linked to one another. Crosslinking may occur where two binding sites (706) are close to one another and create a bond. The binding sites (706) may be made of monomers configured to attract targeted monomers of the adjacent polymer
20 chains. Such bonds may include ionic bonds, hydrogen bonds, disulfide bonds, Van der Waals forces, other non-covalent bonds, or combinations thereof.

[0049] A crosslinked polymer brush according to the principles described herein may form a barrier that prevents proteins and other fluid constituents in the biological environment of the fluid filled cavity from reaching
25 the hydrophobic surface. Such a barrier prevents proteins or other fluid constituents from binding to the hydrophobic material as part of a bioresponse to foreign materials. As a consequence, inflammation, and encapsulation are reduced or even eliminated.

[0050] While the examples above have been described with reference
30 to a specific cochlear implant system, any cochlear implant system that is compatible with the principles described herein may be used. Further, while the elongated body has been described above with reference to specific shapes and insertion positions, any elongated body shape and insertion position that are compatible with the principles described herein may be used. Also, while
35 the above examples have been described with reference to specific electrode materials, electrode positions on the elongated body, electrode geometries, and

5 electrode proximity to the polymer brush, any electrodes with various materials, positions, geometries, and proximities to the polymer brush may be used in accordance with the principles described herein.

[0051] Further, while the examples above have been described with reference to specific polymer brush thickness, polymer types, and formation
10 methods, and polymer brush with various thicknesses, polymer types, and formation methods may be used in accordance with the principles described herein. Also, while the above examples have been described with reference to specific ways that the polymer brush forms a barrier against adsorption of proteins and other constituents in the biological environment of the cochlea, any
15 polymer brush chemistry that forms the barrier may be used in accordance with the principles described herein.

[0052] The preceding description has been presented only to illustrate and describe examples of the principles described. This description is not intended to be exhaustive or to limit these principles to any precise form
20 disclosed. Many modifications and variations are possible in light of the above teaching.

5 CLAIMS

WHAT IS CLAIMED IS:

1. An active, implantable device, comprising:
10 an elongated body comprising:
an electrically insulating material having a
hydrophobic surface; and
a polymer brush comprising hydrophilic polymer
chains tethered to at least a portion of the hydrophobic
15 surface, wherein the polymer brush reduces sliding friction
between the elongated body and biological tissue of an
internal wall of a fluid filled cavity.
2. The device of claim 1, wherein the hydrophilic polymer chains are
20 covalently bonded to the at least a portion of the hydrophobic
surface.
3. The device of claim 1, wherein the elongated body is sized to slide
within a human cochlea.
25
4. The device of claim 3, wherein fluid filled cavity is a human
cochlea and the elongated body bends inward when sliding due to
an inward curve of the internal wall of the human cochlea.
- 30 5. The device of claim 1, wherein the elongated body further
comprises at least one electrode in communication with a power
source and the at least one electrode is positioned to generate an
electric field adjacent the hydrophobic surface.
- 35 6. The device of claim 5, wherein the electric field is focused away
from the polymer brush.

- 5 7. The device of claim 1, wherein the polymer brush forms a barrier that prevents proteins and other constituents of a biological environment of the fluid filled cavity from reaching the hydrophobic surface.
- 10 8. The device of claim 7, wherein the hydrophilic polymer chains are crosslinked to one another to form the barrier.
9. The device of claim 1, wherein the hydrophobic surface is made of silicone.
- 15 10. The device of claim 1, wherein the hydrophilic polymer chains make up at least 75 percent of a composition of the polymer brush.
- 20 11. The device of claim 1, wherein the hydrophilic polymer chains make up at least 95 percent of a composition of the polymer brush.
- 25 12. The device of claim 1, wherein the polymer brush has a characteristic of forming hydrated monomers when the polymer brush is immersed in an aqueous environment.
- 30 13. The device of claim 1, wherein the polymer brush does not substantially interfere with a designed stiffness profile of the elongated body.
14. A cochlear implant, comprising:
 an elongated body comprising:
 an electrically insulating material having a
35 hydrophobic surface located adjacent electrodes; and
 a polymer brush comprising hydrophilic polymer

5 chains tethered to at least a portion of the hydrophobic surface, wherein the polymer brush reduces sliding friction between the elongated body and biological tissue of an internal wall of a fluid filled cavity and the electrodes are positioned to form an electric field adjacent to the polymer
10 brush.

15 15. The implant of claim 14, wherein the hydrophilic polymer chains make up at least 95 percent of a composition of the polymer brush.

16. The implant of claim 14, wherein the polymer brush has a characteristic of forming hydrated monomers when the polymer brush is immersed in an aqueous environment.

20 17. The implant of claim 14, wherein the polymer brush forms a barrier that prevents proteins and other constituents of an biological environment of the fluid filled cavity from reaching the hydrophobic surface.

25 18. The implant of claim 17, wherein the hydrophilic polymer chains are crosslinked to one another to form the barrier.

30 19. The implant of claim 14, wherein the elongated body has a compliancy that allows the elongated body to bend inward when sliding due to an inward curve of an internal wall of the human cochlea.

35 20. The implant of claim 19, wherein the polymer brush does not substantially interfere with a designed stiffness profile of the elongated body.

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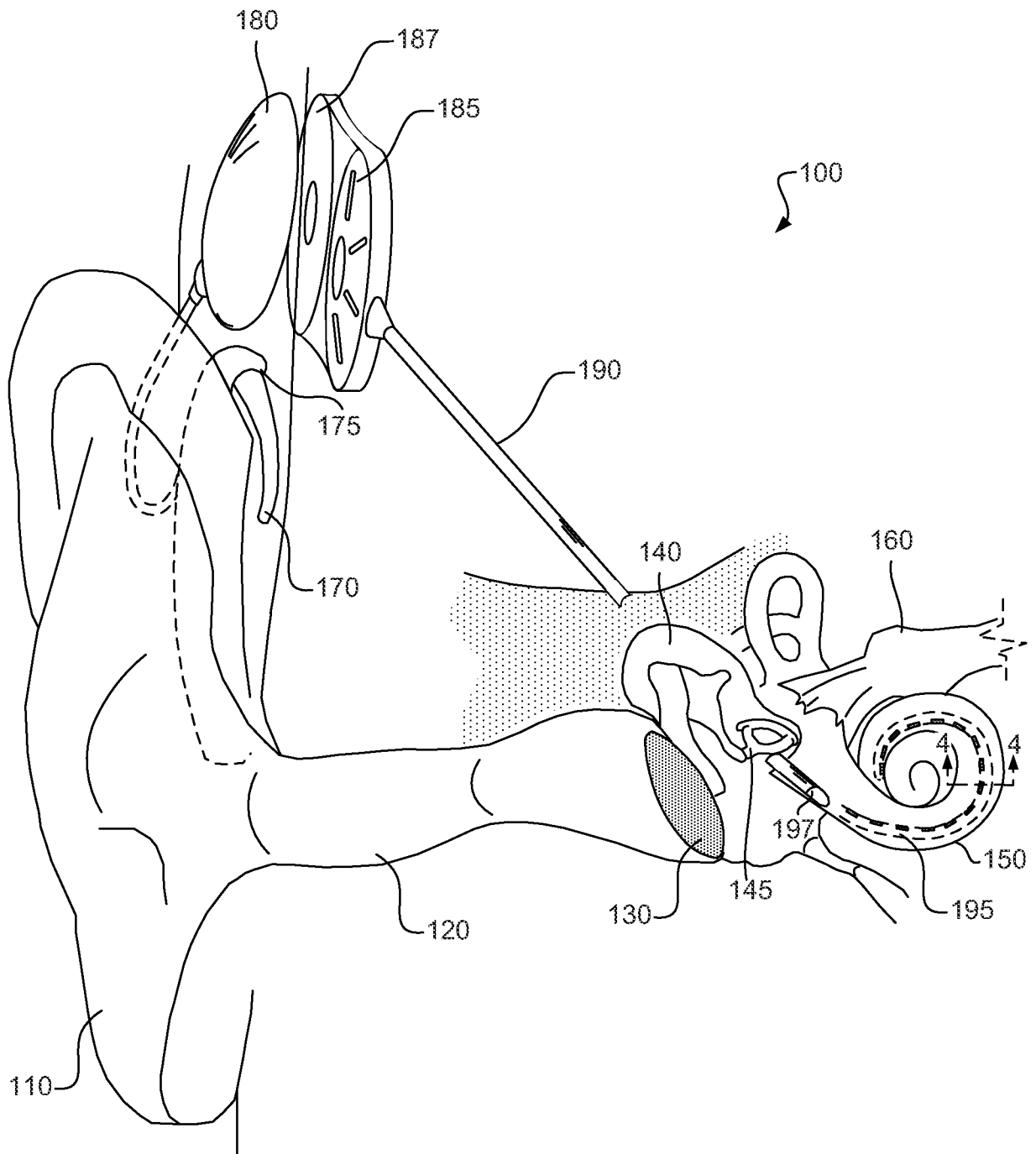
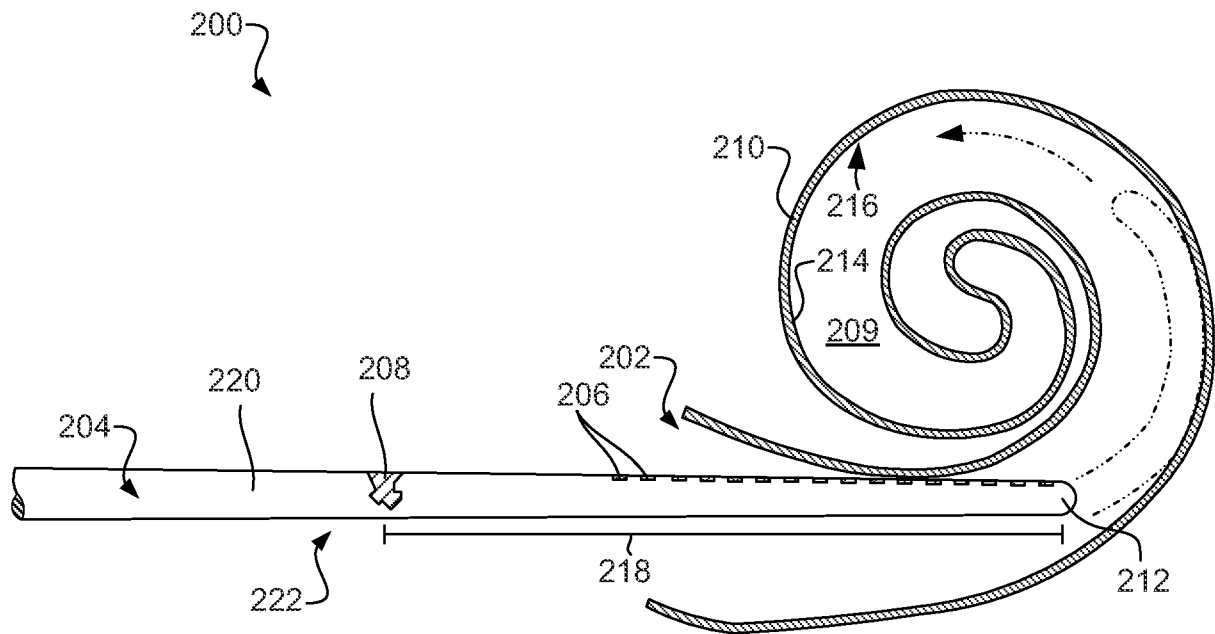
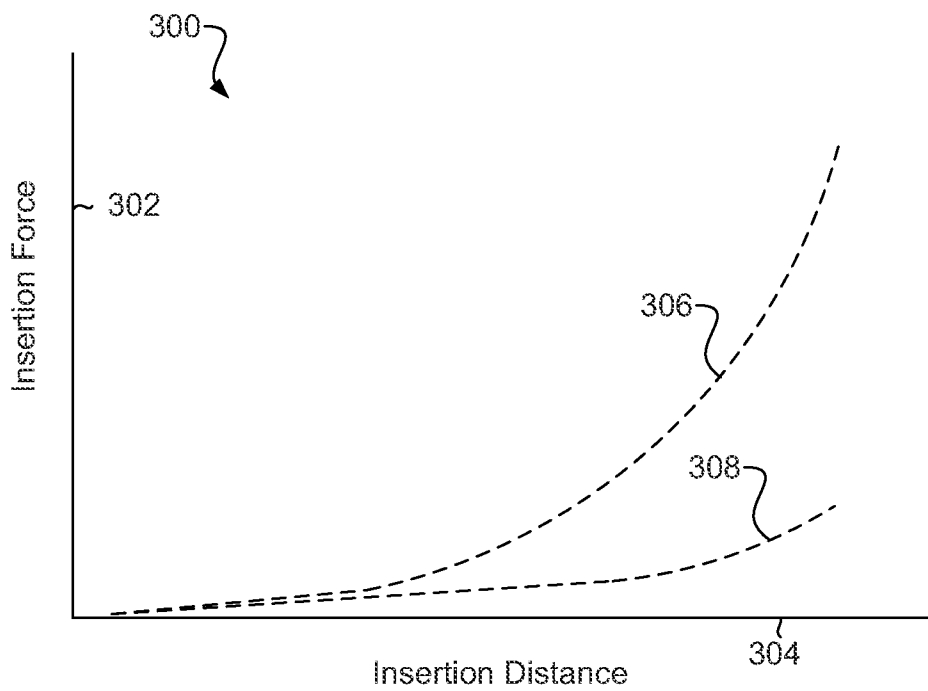


Fig. 1

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**Fig. 2****Fig. 3**

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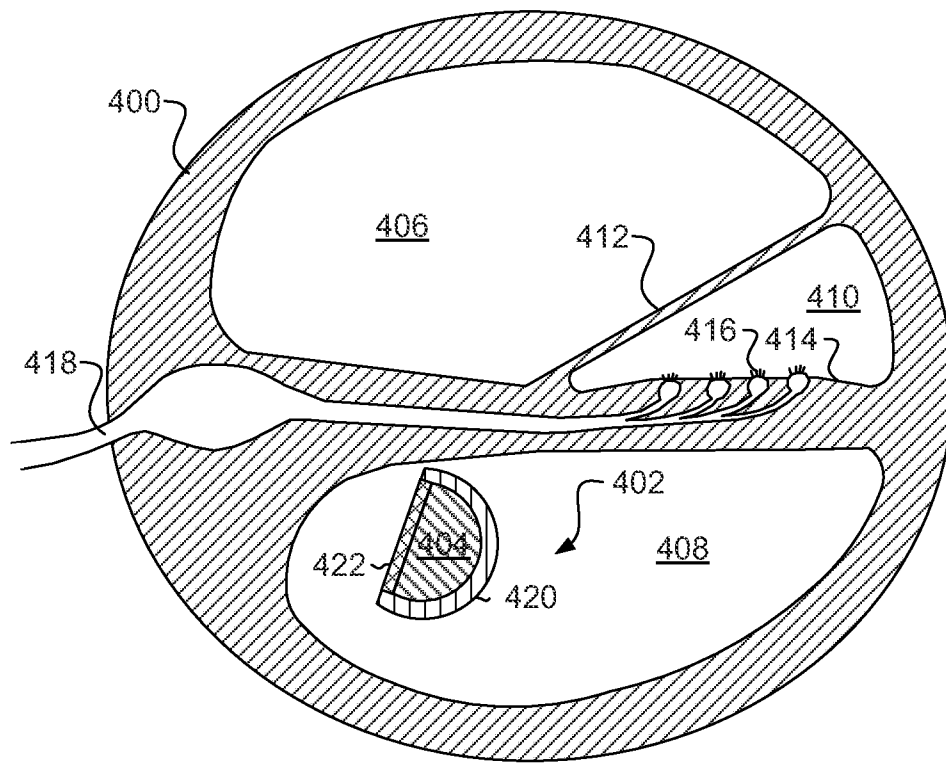


Fig. 4

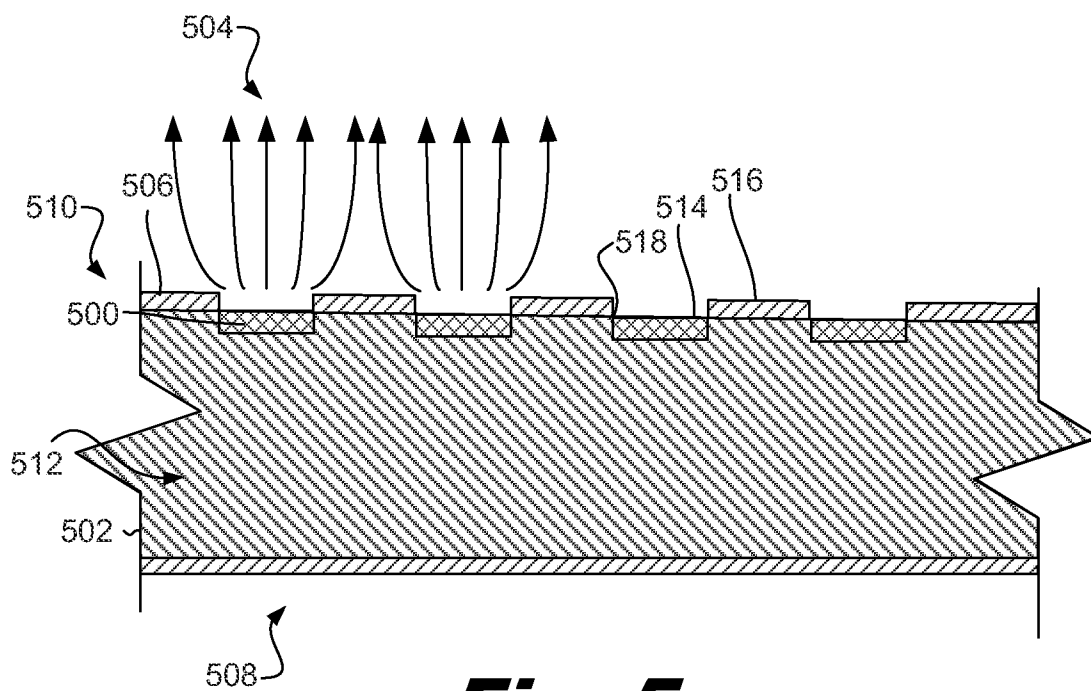
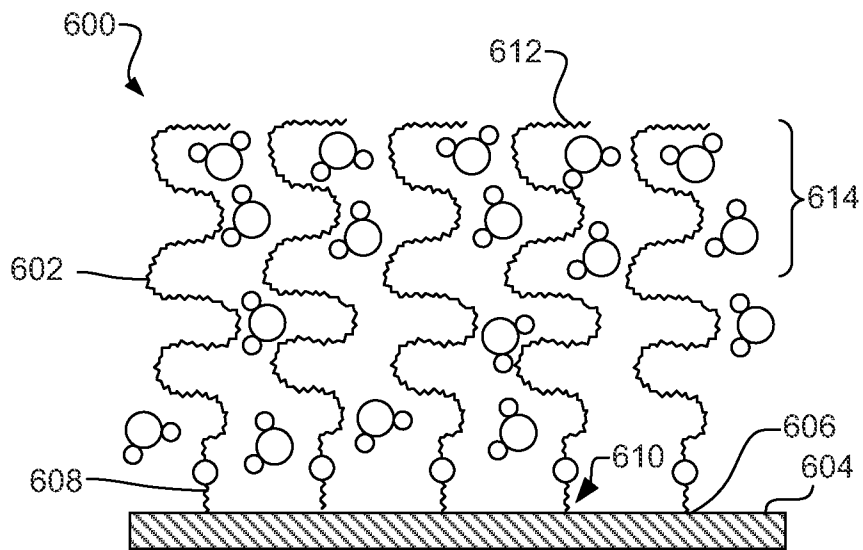
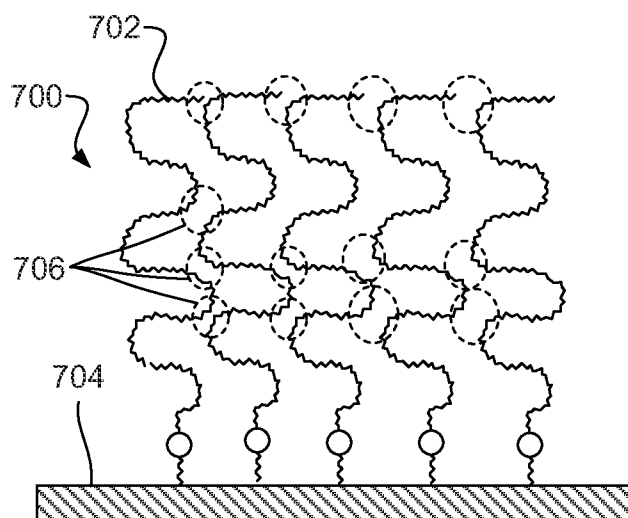


Fig. 5

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**Fig. 6****Fig. 7**

INTERNATIONAL SEARCH REPORT

International application No
PCT/US2012/071806

A. CLASSIFICATION OF SUBJECT MATTER

INV. A61N1/05 A61L29/14
ADD. A61N1/36 A61F2/00 A61F11/04

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

A61N A61L A61F

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

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

EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

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X	Vendra, V. K., Wu, L. and Krishnan, S.: "Polymer thin films for biomedical applications" In: "Nanotechnologies for the life sciences", 15 January 2011 (2011-01-15), Wiley, Wiley Online Library, XP002713107, pages 1-54, DOI: 10.1002/9783527610419.ntls0179, the whole document	1,7,8, 10-12, 14-18
Y A	----- US 2009/292237 A1 (OVERSTREET EDWARD H [US] ET AL) 26 November 2009 (2009-11-26)	1,14 2-6,9, 13,19,20
X	paragraphs [0029], [0051] - [0073], [0093]; figures 5-8 ----- -/-	1-11, 13-15, 17-20 1,14 12,16



Further documents are listed in the continuation of Box C.



See patent family annex.

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Date of the actual completion of the international search

18 September 2013

Date of mailing of the international search report

05/11/2013

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Authorized officer

Smit, Jos

INTERNATIONAL SEARCH REPORT

International application No

PCT/US2012/071806

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT		
Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
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