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(54) **DRUG-DELIVERY PUMP WITH DYNAMIC, ADAPTIVE CONTROL**

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CPC ***A61M 5/16831*** (2013.01); ***A61F 9/0017*** (2013.01); ***A61M 5/14276*** (2013.01); ***A61M 5/16804*** (2013.01); ***A61M 5/172*** (2013.01); ***G06F 19/3468*** (2013.01); *A61M 5/148* (2013.01); *A61M 5/14244* (2013.01); *A61M 5/16854* (2013.01); *A61M 31/002* (2013.01); *A61M 2005/14204* (2013.01); *A61M 2205/3331* (2013.01); *A61M 2205/3368* (2013.01); *A61M 2210/0612* (2013.01); *A61M 2210/0693* (2013.01)

(58) **Field of Classification Search**

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See application file for complete search history.

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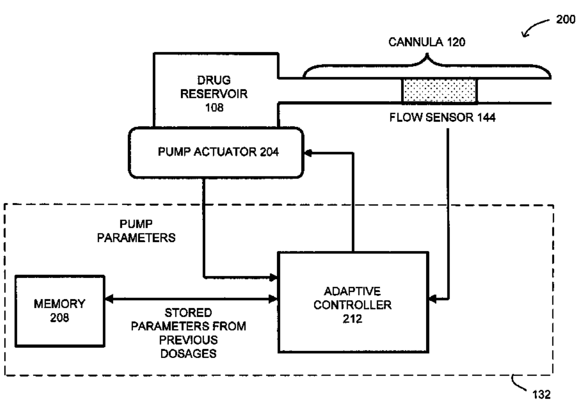
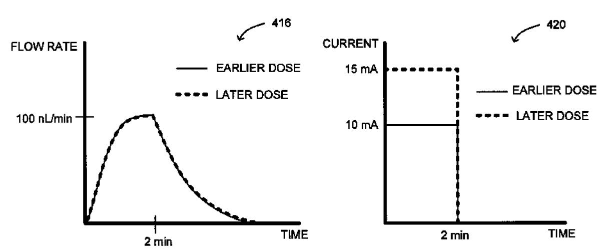
(57) **ABSTRACT**

In various embodiments, actuation of a drug-delivery pump is controlled based on a change in a condition of the pump.

9 Claims, 8 Drawing Sheets





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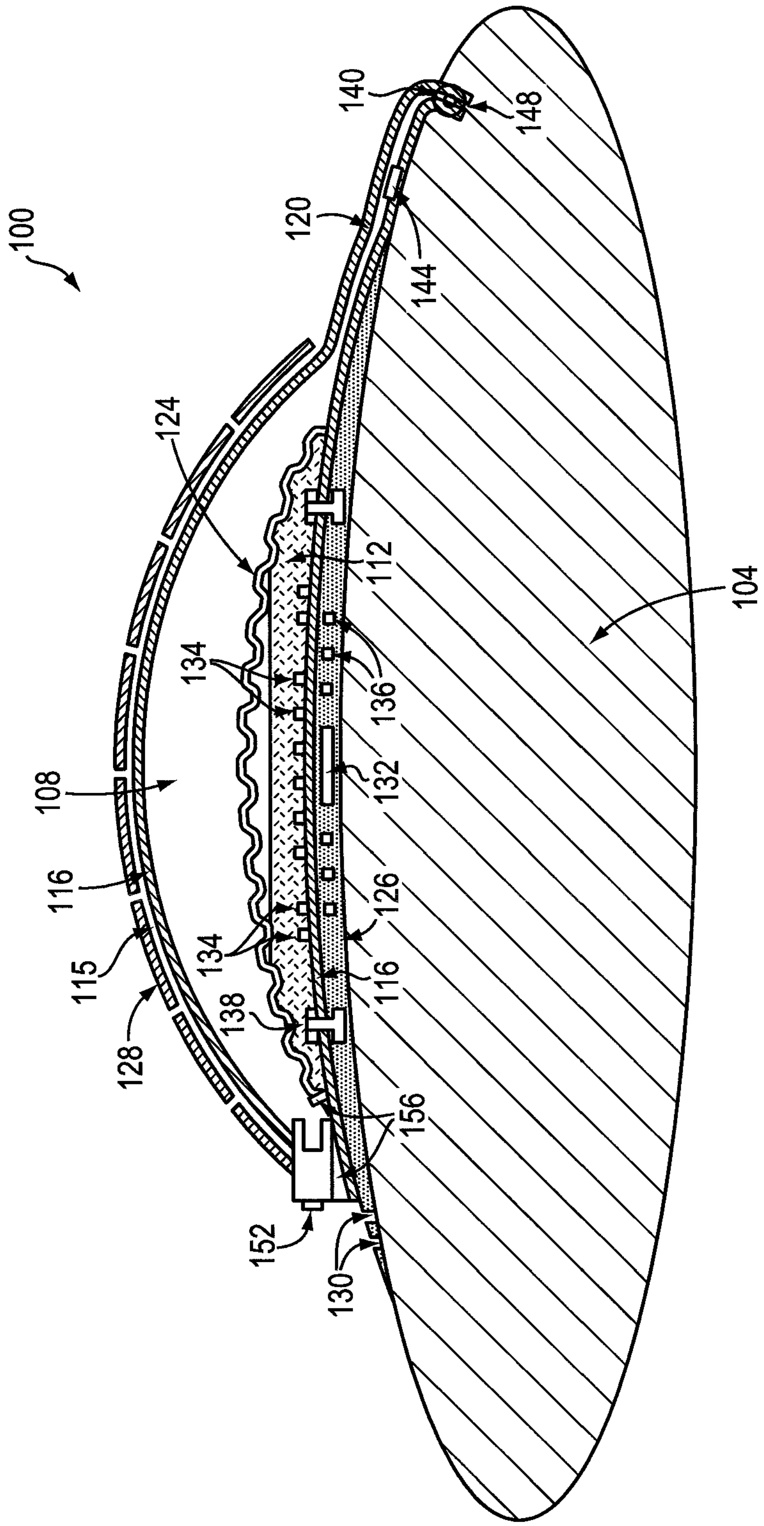
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130

FIG. 1

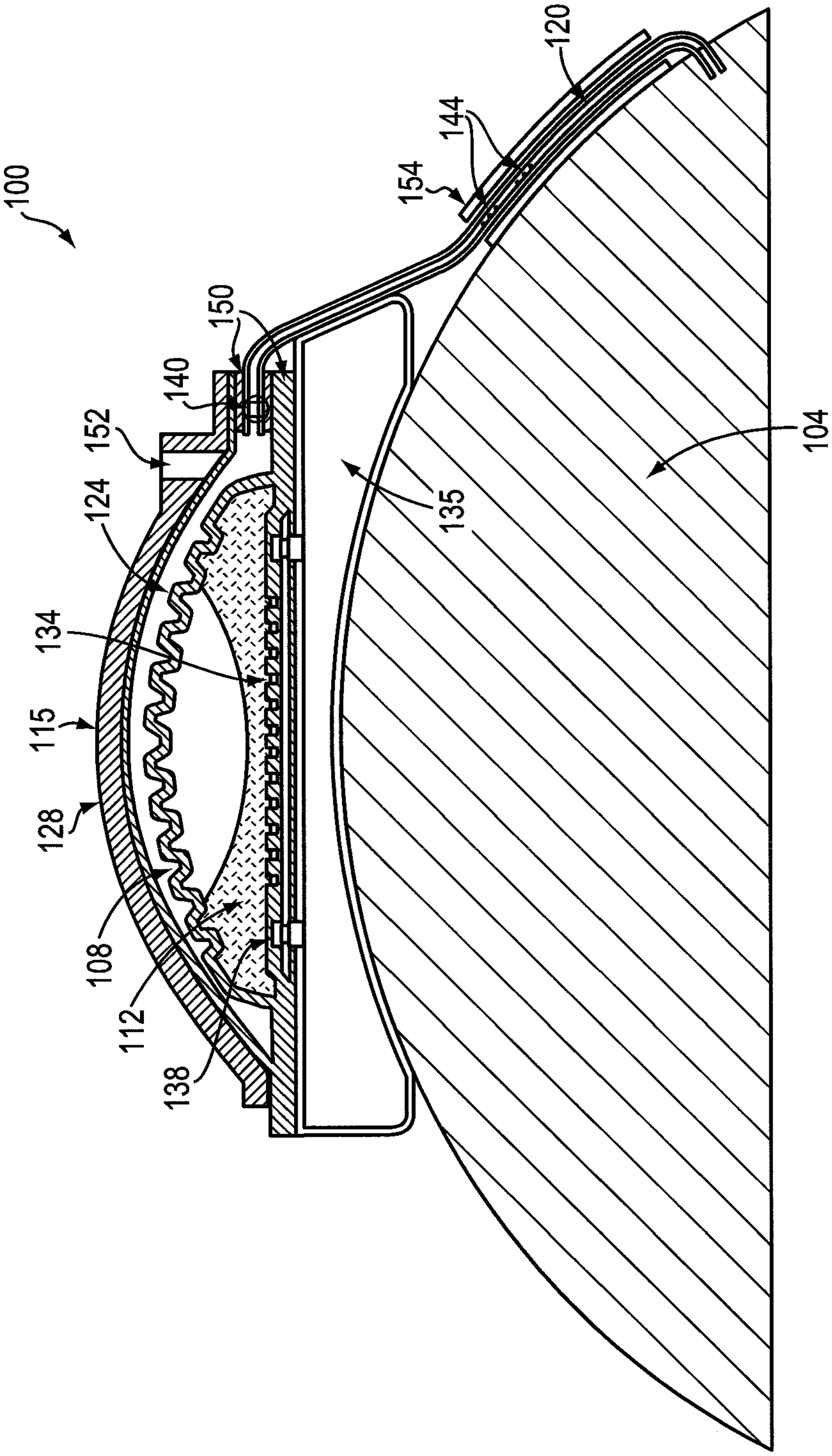
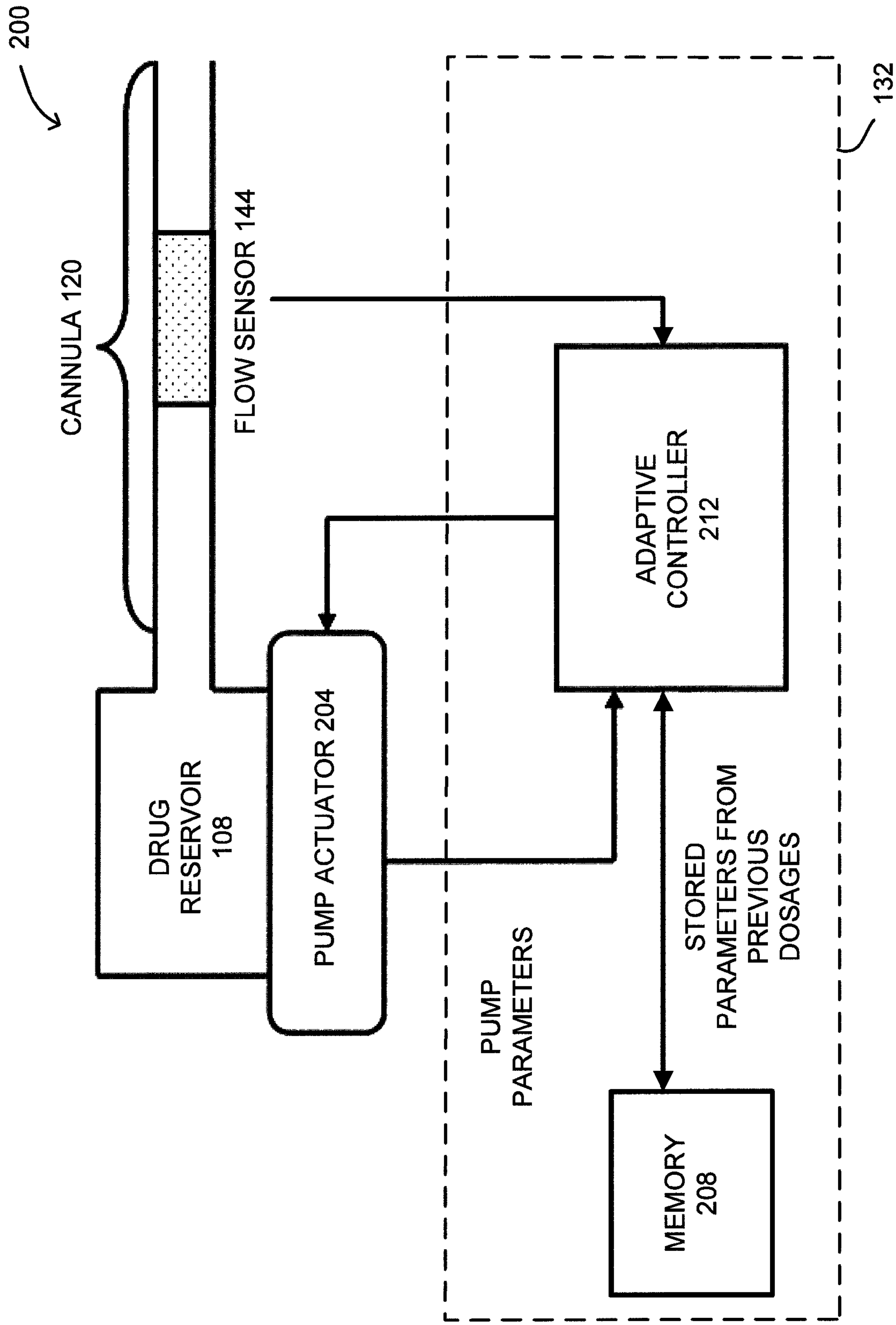


FIG. 2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FLOW SENSOR 144
132
DRUG RESERVOIR 108
PUMP ACTUATOR 204
ADAPTIVE CONTROLLER 212
PUMP PARAMETERS
STORED PARAMETERS FROM PREVIOUS DOSAGES
MEMORY 208

FIG. 3

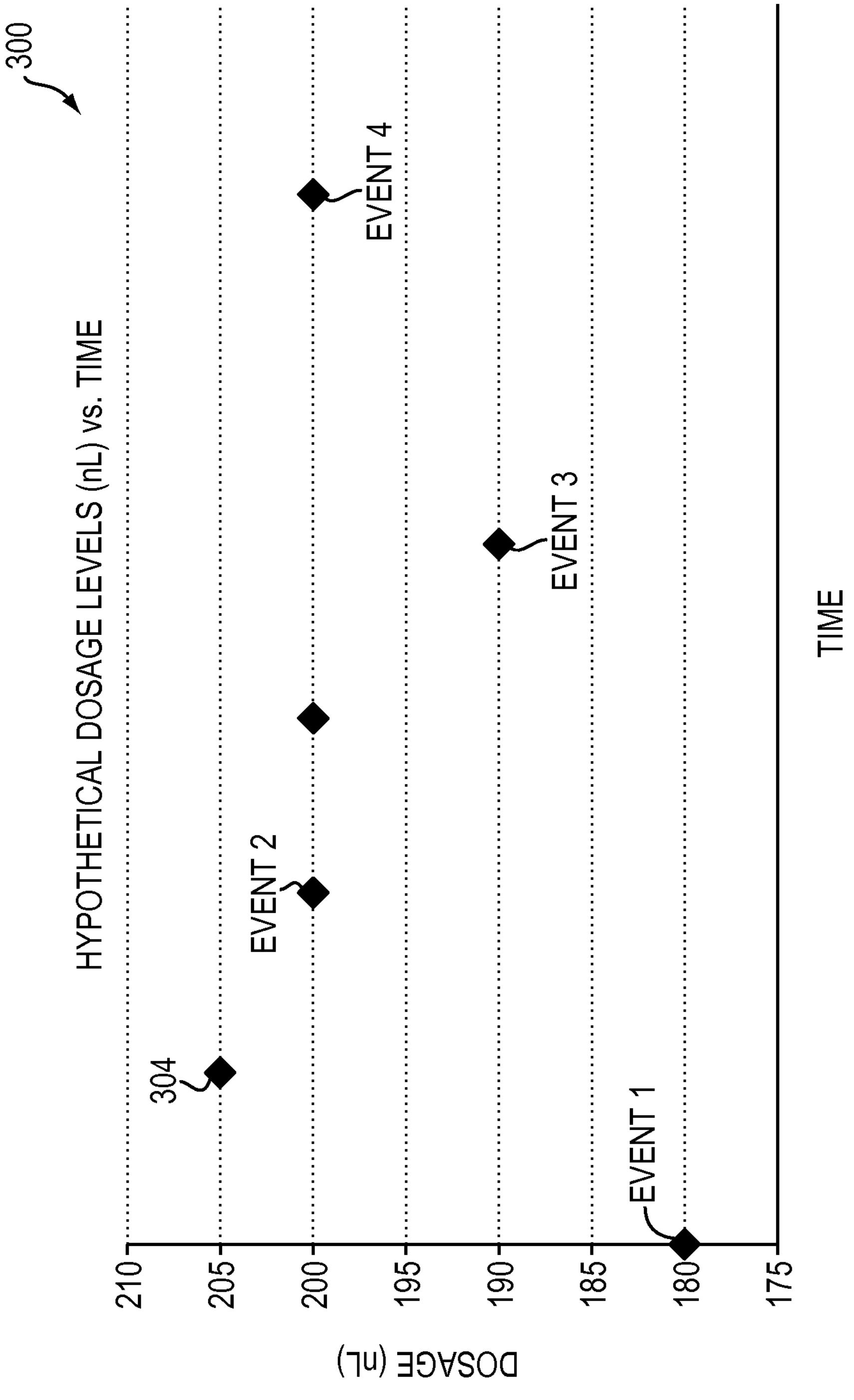
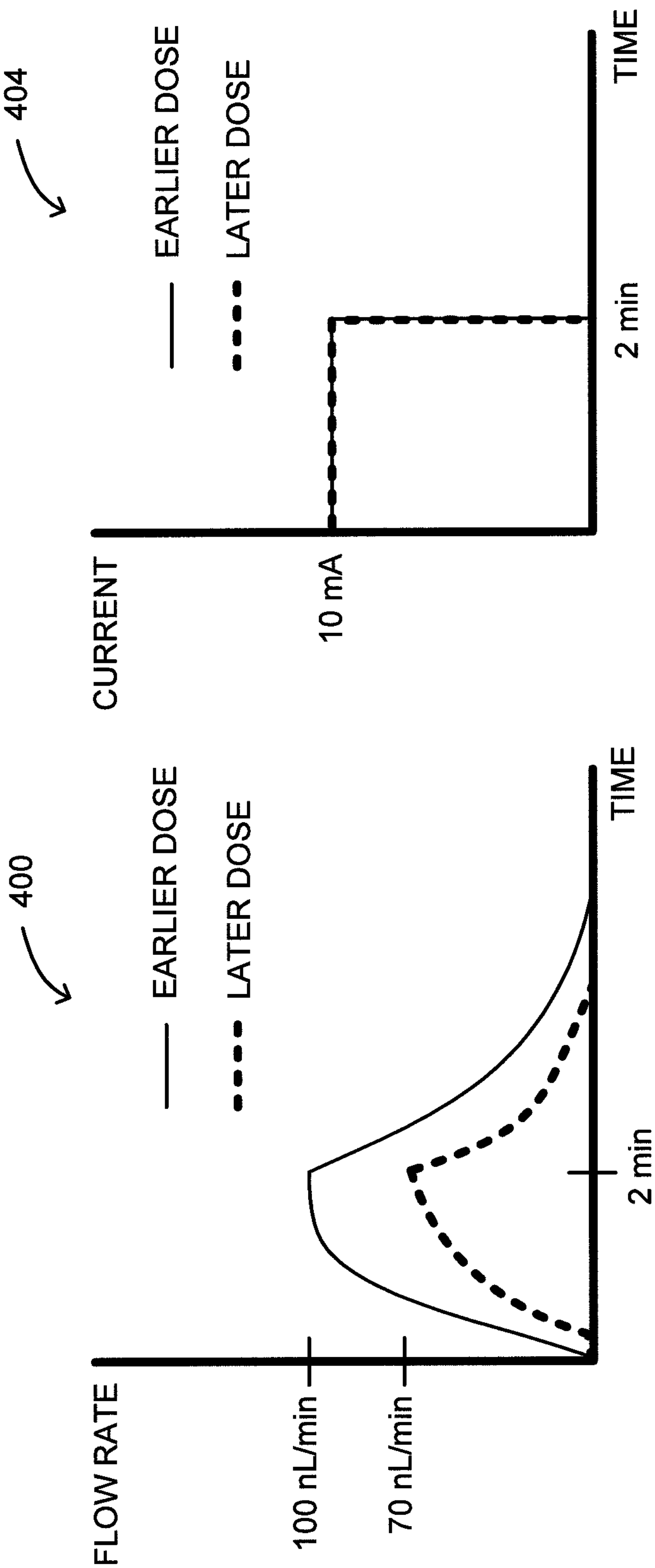

300
HYPOTHETICAL DOSAGE LEVELS (nL) vs. TIME
DOSAGE (nL)
210
205
200
195
190
185
180
175
TIME
304
EVENT 1
EVENT 2
EVENT 3
EVENT 4

FIG. 4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CURRENT
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TIME
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70 nL/min
2 min
TIME

FIG. 5A

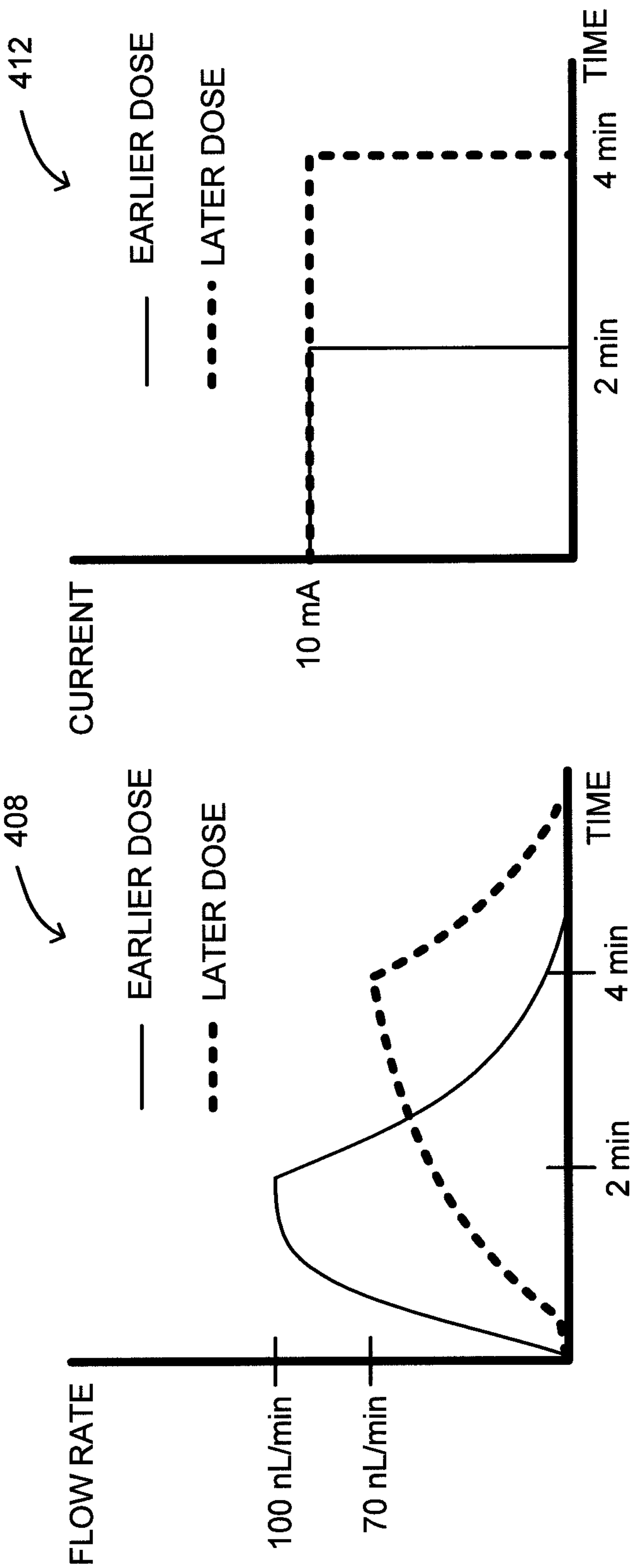
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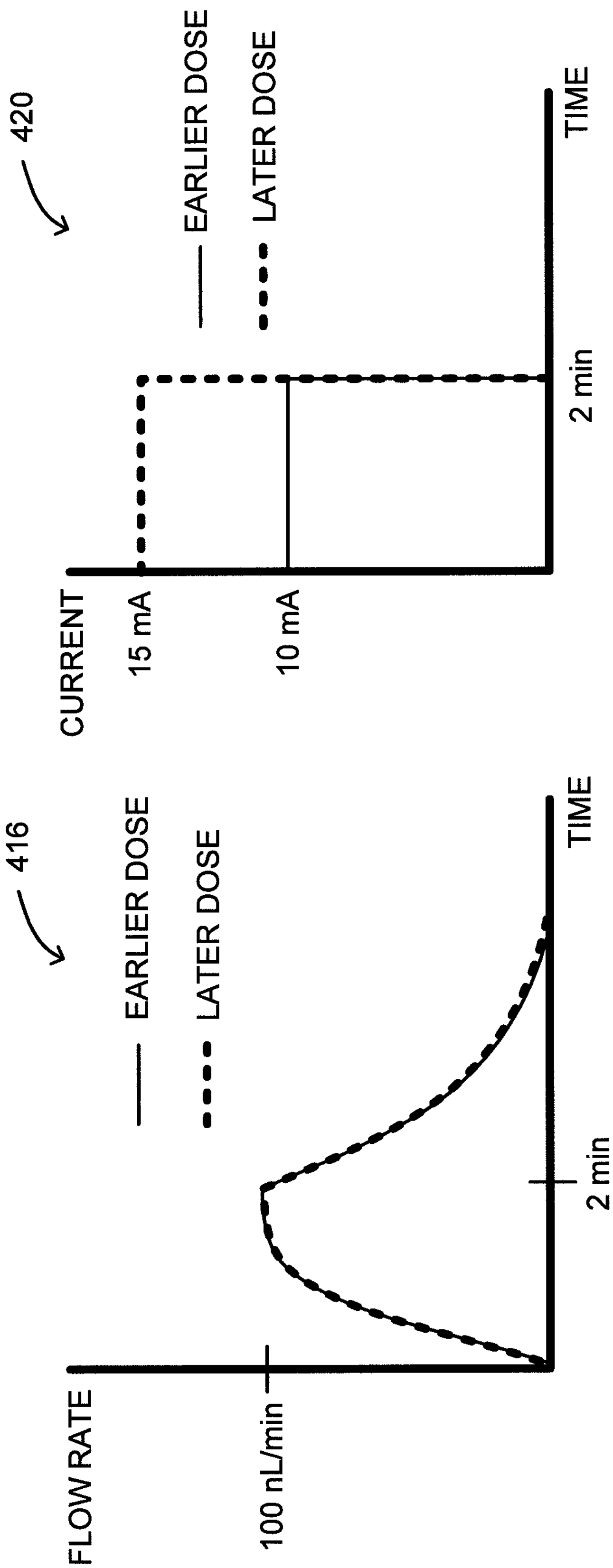
FIG. 5B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TIME
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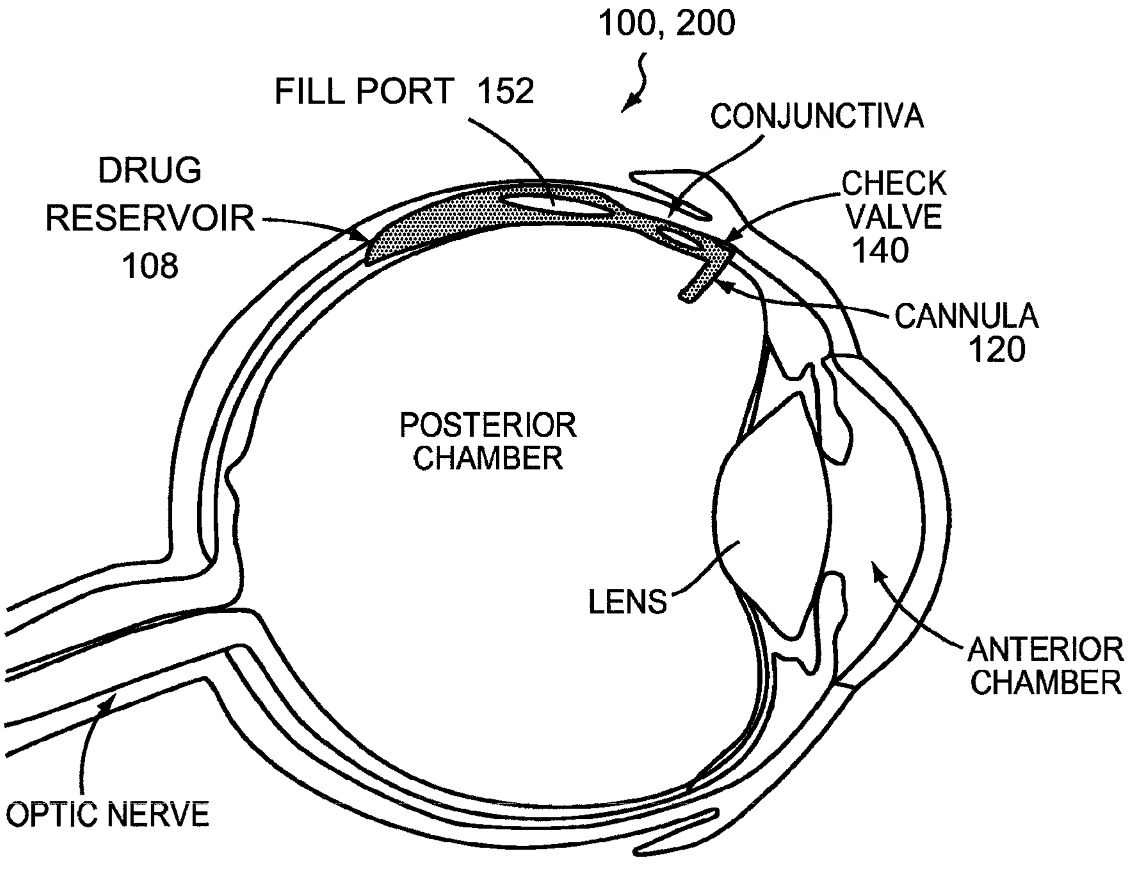
2 min
TIME

FIG. 5C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FILL PORT 152
CONJUNCTIVA
DRUG RESERVOIR 108
CHECK VALVE 140
CANNULA 120
POSTERIOR CHAMBER
LENS
ANTERIOR CHAMBER
OPTIC NERVE

FIG. 6

DRUG-DELIVERY PUMP WITH DYNAMIC, ADAPTIVE CONTROL

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is a divisional application of, and incorporate herein by reference, U.S. patent application Ser. No. 12/858,808, filed on Aug. 18, 2010 continuation-in-part of, claims priority to and the benefit of, and incorporates herein by reference in its entirety U.S. patent application Ser. No. 12/463,265, which was filed on May 8, 2009, and which claimed priority to and the benefit of U.S. Provisional Patent Application Nos. 61/051,422, filed on May 8, 2008; 61/197,751, filed on Oct. 30, 2008; 61/197,769, filed on Oct. 30, 2008; 61/198,090, filed on Nov. 3, 2008; and 61/198,131, filed on Nov. 3, 2008. This application also claims priority to and the benefit of, and incorporates herein by reference in its entirety, U.S. Provisional Patent Application No. 61/234,742, which was filed on Aug. 18, 2009.

TECHNICAL FIELD

In various embodiments, the invention relates to drug-delivery pumps. In particular, embodiments of the invention relate to drug-delivery pumps whose actuation may be dynamically and adaptively controlled.

BACKGROUND

Medical treatment often requires the administration of a therapeutic agent (e.g., medicament, drugs, etc.) to a particular part of a patient's body. As patients live longer and are diagnosed with chronic and/or debilitating ailments, the likely result will be an increased need to place even more protein therapeutics, small-molecule drugs, and other medications into targeted areas throughout the patient's body. Some maladies, however, are difficult to treat with currently available therapies and/or require administration of drugs to anatomical regions to which access is difficult to achieve.

A patient's eye is a prime example of a difficult-to-reach anatomical region, and many vision-threatening diseases, including retinitis pigmentosa, age-related macular degeneration (AMD), diabetic retinopathy, and glaucoma, are difficult to treat with many of the currently available therapies. For example, oral medications can have systemic side effects; topical applications may sting and engender poor patient compliance; injections generally require a medical visit, can be painful, and risk infection; and sustained-release implants must typically be removed after their supply is exhausted (and generally offer limited ability to change the dose in response to the clinical picture).

Another example is cancer, such as breast cancer or meningiomas, where large doses of highly toxic chemotherapies, such as rapamycin, bevacizumab (e.g., AVASTIN), or irinotecan (CPT-11), are typically administered to the patient intravenously, which may result in numerous undesired side effects outside the targeted area. Yet another example is drug delivery to the knee, where drugs often have difficulty penetrating the avascular cartilage tissue for diseases such as osteoarthritis.

Implantable drug-delivery devices (e.g., drug-delivery pumps), which may have a refillable drug reservoir, a cannula for delivering the drug, a check valve, etc., generally allow for controlled delivery of pharmaceutical solutions to a specified target. As drug within the drug reservoir depletes, the physician can refill the reservoir with, for example, a syringe, while leaving the device implanted within the patient's body. This approach can minimize the surgical incision needed for implantation and typically avoids future or repeated invasive surgery or procedures.

Implantable drug-delivery pumps, particularly in ocular applications, often utilize a passive mechanism for drug delivery (e.g., pumping the drug out when a finger is pressed on the drug reservoir). One limitation of these conventional, passively-driven drug-delivery pumps is their inability to dynamically respond to changes inside the pump (e.g., failures, blockages, etc.) or to changes in the drug-delivery target area (e.g., increased pressure, bending of the pump's cannula, inflammation causing pressure around the cannula, etc.). The ability to respond to such changes can improve not only the therapeutic value of a pump, but also safety.

Active drug-delivery pumps, particularly feedback-driven ones, represent a substantial improvement over passively-driven pumps. Typically, these feedback-driven pumps are electrically-driven mechanical pumps. They generally employ controller units that receive inputs from sensors that monitor the target treatment area and, in response, direct the release of a pharmaceutical or therapeutic agent to achieve a desired result. The amount of drug released in each dosage period is thus largely determined by the current conditions of the target area and is intended to be variable depending on what the conditions of the target area warrant.

Pharmaceutical treatment regimens may, however, require that a drug be administered in fixed amounts at regular time intervals regardless of the changing conditions in the drug-delivery target area. Since the dosage levels produced by existing closed-loop feedback-driven systems can be highly dependent on the parameters of the treatment area and thus prone to fluctuations, they are inadequate for delivering fixed drug dosages at periodic intervals. For example, changes in the conditions of the target area, such as blockages or other biochemical or physiological events, may lead to variable levels of drug being delivered to the target area. Accordingly, there is a need for a feedback-driven pump that maintains the target dosage level despite such changes.

Furthermore, while feedback based on the conditions of the target area is important in numerous therapeutic applications, errors in drug administration can also arise from changing conditions within the pump itself. Conventional pumps generally do not account for such changes, which can also lead to variable amounts of drug being released. Accordingly, there is also a need for a drug-delivery pump that dynamically responds to changing conditions within the pump itself in order to, for example, consistently release a fixed dosage of drug at periodic time intervals.

SUMMARY OF THE INVENTION

In various embodiments, the present invention features an external or implantable drug-delivery pump that includes a dynamic, adaptive control system. The control system may operate the pump so as to release substantially fixed amounts of pharmaceutical or therapeutic agents to a target treatment area at regular intervals. In certain embodiments, the control system continuously monitors (either directly or indirectly) conditions internal to the pump that have an effect on the degree and duration of pump actuation and, consequently, the amount of drug that is released. As used herein, the term "substantially" means±10% (e.g., by weight or by volume), and in some embodiments, ±5%.

In one embodiment, the drug-delivery pump is an electrochemically-actuated pump, such as an electrolysis-driven pump. Electrochemically-actuated pumps, as compared to electrically-driven mechanical pumps, offer several advantages for drug-delivery systems. For example, they generally have few moving parts, which enables them to be small and portable, and which makes them less prone to mechanical breakdown than electrically-driven mechanical pumps. In particular, electrochemically-actuated pumps are suitable for environments that require small pump sizes, such as the ocular environment. As further described herein, an electrolysis-driven pump generally employs electrodes to generate an electrochemically active gas that variably pressurizes a drug contained in a separate chamber in order to dispense the drug in a controlled fashion. The amount of drug dispensed depends on the gas pressure variably generated by the pump actuator, which in turn depends on the current that passes through the electrodes. Because of the inherent variability in these electrolysis-driven pumps (e.g., the volume of gas and/or the amount of electrolyte can change between every pump cycle), the adaptive control design described herein can confer substantial advantages, as further explained below.

In general, in one aspect, embodiments of the invention feature a drug-delivery pump that includes a drug reservoir, a cannula for conducting liquid from the reservoir to a target site, a pump actuator for forcing the liquid from the reservoir through the cannula, and circuitry for controlling the actuator based on a change in a condition of the pump.

In general, in another aspect, embodiments of the invention feature a method of delivering a drug to a patient using such a drug-delivery pump. The method involves establishing fluid communication between the drug reservoir and the patient (i.e., the target site) and controlling the pump actuator based on a change in a condition of the pump so as to deliver a dosage of liquid from the drug reservoir into the patient.

In various embodiments, the control circuitry maintains delivery of a substantially fixed dosage of the liquid at periodic time intervals to the target site. Moreover, the circuitry may include memory for storing the conditions of the pump at the time of previous delivery events (e.g., at the time of each delivery interval). In one embodiment, the drug-delivery pump includes a flow sensor for measuring a flow rate of the liquid through the cannula and into the patient, and the circuitry controls the pump actuator based, at least in part, on an analysis of the flow rate. The circuitry may also control the actuator based on the stored conditions of the pump from the previous doses and/or on real-time data from the actuator. In another embodiment, the control circuitry maintains delivery of a substantially fixed dosage of the liquid over time through continuous infusion to the target site.

As mentioned, the drug-delivery pump may be an electrolysis-driven pump. More particularly, the pump actuator may include an electrolyte chamber, an expandable diaphragm that separates the electrolyte chamber from the drug reservoir and provides a fluid barrier therebetween, and electrolysis electrodes that cause evolution of a gas in the electrolyte chamber. The evolution of the gas expands the diaphragm so that the liquid is forced from the drug reservoir into the cannula. In various embodiments, the diaphragm expansion is adjusted by varying the actuation current supplied to the electrodes. In other embodiments, the diaphragm expansion is adjusted by varying an actuation duration of the electrodes. As described herein, the electrolysis electrodes may be driven with either a constant current or a time-varying current waveform.

In general, in yet another aspect, embodiments of the invention feature a drug-delivery pump that includes a drug reservoir, an electrolyte chamber, electrolysis electrodes, an expandable diaphragm that separates the electrolyte chamber from the drug reservoir and provides a fluid barrier therebetween, a cannula for conducting liquid from the drug reservoir to a target site, and circuitry for adjusting expansion of the diaphragm based on conditions of the target site (e.g., changes in one or more biochemical parameters of the target site, in electrical activity at the target site, and/or in pressure at the target site). The pump may include a sensor for detecting such conditions. For their part, the electrolysis electrodes may be activated to cause evolution of a gas in the electrolyte chamber, which expands the diaphragm so that the liquid is forced from the drug reservoir into the cannula.

In general, in still another aspect, embodiments of the invention feature a drug-delivery pump that includes a drug reservoir, a cannula for conducting liquid from the reservoir to a target site, a pump actuator for forcing the liquid from the reservoir through the cannula, and circuitry for controlling the actuator. In particular, the circuitry controls the actuator i) to initially deliver a substantially fixed dosage of the liquid at periodic time intervals to the target site, and ii) to compensate for a change in a condition of the pump so as to maintain or resume the delivery of the substantially fixed dosage of the liquid at the periodic time intervals to the target site.

In general, in a further aspect, embodiments of the invention feature a method of delivering a drug to a patient from a drug-delivery pump that includes a drug reservoir and a pump actuator for forcing liquid from the reservoir into the patient. The method involves establishing fluid communication between the drug reservoir and the patient, and controlling the pump actuator. In particular, the actuator is controlled i) to initially deliver a substantially fixed dosage of the liquid at periodic time intervals from the drug reservoir into the patient, and ii) to compensate for a change in a condition of the pump so as to maintain or resume the delivery of the substantially fixed dosage of the liquid at the periodic time intervals into the patient.

These and other objects, along with advantages and features of the embodiments of the present invention herein disclosed, will become more apparent through reference to the following description, the accompanying drawings, and the claims. Furthermore, it is to be understood that the features of the various embodiments described herein are not mutually exclusive and can exist in various combinations and permutations, even if not made explicit herein.

BRIEF DESCRIPTION OF THE DRAWINGS

In the drawings, like reference characters generally refer to the same parts throughout the different views. Also, the drawings are not necessarily to scale, emphasis instead generally being placed upon illustrating the principles of the invention. In the following description, various embodiments of the present invention are described with reference to the following drawings, in which:

FIG. **1** schematically illustrates, in cross-section, an implantable drug-delivery pump in accordance with one embodiment of the invention;

FIG. **2** schematically illustrates, in cross-section, an implantable drug-delivery pump in accordance with another embodiment of the invention;

FIG. **3** is a block diagram of a drug-delivery pump in accordance with one embodiment of the invention;

FIG. **4** is a graph representing an example of how each of the drug-delivery pumps depicted in FIGS. **1**-**3** may adapt to changing conditions within the pump to deliver a target dosage level;

FIG. **5**A illustrates exemplary flow and actuation profiles of a pump that operates without feedback control;

FIG. **5**B illustrates exemplary flow and actuation profiles of a pump whose actuator is actuated for a longer period of time as the pump's efficiency decreases;

FIG. **5**C illustrates exemplary flow and actuation profiles of a pump whose actuation current is increased as the pump's efficiency decreases; and

FIG. **6** is a sectional view of a patient's eye illustrating implantation therein of a drug-delivery pump in accordance with one embodiment of the invention.

DESCRIPTION

In general, embodiments of the present invention pertain to external or implantable drug-delivery pumps (whether they be reusable and refillable pumps, disposable pumps, etc.) whose actuation may be dynamically and adaptively controlled. For example, embodiments of the drug-delivery pumps may be implantable within a patient's body, such as within the patient's eye or brain. In certain embodiments, the implantable drug-delivery pumps combine small size and a refillable drug reservoir. The small size minimizes discomfort from the drug-delivery pump to the patient, while the refillable reservoir allows the pump to be refilled in situ, rather than having to be replaced. As such, a fluid, such as a solution of a drug, can be supplied to the patient over extended periods of time.

A. Exemplary Drug-Delivery Pump

Embodiments of the invention may be employed in connection with various types of drug-delivery pumps, whether they be external pumps or pumps implantable within a patient's body. FIGS. **1** and **2** schematically illustrate two variations of an exemplary implantable drug-delivery pump **100** (namely, an exemplary electrolytic or electrolysis-driven pump **100**) implanted within a patient's eye **104**. The pump **100** may, however, instead be implanted in other portions of a patient's body. For example, it may be implanted in the sub-arachnoid space of the brain to provide chemotherapy or to provide another type of treatment for the brain (e.g., by dosing the brain's parenchyma directly); near a tumor in any portion of the patient's body to provide chemotherapy; in a pancreas that does not respond well to glucose to provide agents (e.g., proteins, viral vectors, etc.) that will trigger insulin release; external to a patient but with a cannula placed under the skin or inside the abdominal cavity to deliver insulin; in the knee to provide drugs that will treat osteoarthritis or other cartilage diseases; near the spine to provide pain medications or anti-inflammatories; or elsewhere.

As illustrated in FIGS. **1** and **2**, embodiments of the pump **100** may include two main components: a pair of chambers **108**, **112** surrounded, at least in part, by a wall **115**, and a cannula **120**. As illustrated in FIG. **1**, the wall **115** that surrounds the chambers **108**, **112** may include or consist of a stand-alone parylene film **116** and, thereover, a separate protection shell **128** made of a relatively rigid biocompatible material (e.g., medical-grade polypropylene). Alternatively, as illustrated in FIG. **2**, the wall **115** may correspond only to the protective shell **128**, which may be coated with parylene.

The top chamber **108** defines a drug reservoir that, when being used to treat a patient, may contain the drug to be administered in liquid form. For its part, the bottom chamber **112** may contain a liquid that, when subjected to electrolysis, evolves a gaseous product. For example, that liquid may be water, which may be electrolytically separated by an applied voltage into hydrogen gas and oxygen gas. Alternatively, as other examples, the electrolyte liquid may be a saline solution (i.e., NaCl in H_2O) or a solution that contains either magnesium sulfate or sodium sulfate. In one embodiment, the two chambers **108**, **112** are separated by a corrugated diaphragm **124**. In other words, the diaphragm **124** provides a fluid barrier between the two chambers **108**, **112**. Like the stand-alone film **116**, the diaphragm **124** may be constructed from, for example, parylene.

As illustrated in FIG. **1**, the stand-alone film **116** may act as an outer barrier for the drug reservoir **108** and the protective shell **128** may provide a hard surface against which the film **116** exerts pressure. In such a case, the shell **128** may be perforated to allow for eye, brain, or other bodily fluid movement. Alternatively, as illustrated in FIG. **2**, the protective shell **128** may itself act as the outer barrier for the drug reservoir **108** and be unperforated. In both embodiments depicted in FIGS. **1** and **2**, the protective shell **128** may prevent outside pressure from being exerted on the drug reservoir **108**. As illustrated in FIG. **1**, a bottom portion **126** (i.e., a floor **126**) of the protective shell **128** may include suture holes **130**. Similarly, although not shown in either FIG. **1** or FIG. **2**, the cannula **120** may also include suture holes along its sides. The suture holes **130** may be employed in suturing (i.e., anchoring) the pump **100** in place in the patient's body.

As also illustrated in FIG. **1**, to provide power to the pump **100** and to enable data transmission therewith, a battery and control circuitry **132** may be embedded (e.g., hermetically sealed) under the chambers **108**, **112** (i.e., between a bottom portion of the stand-alone parylene film **116** of the drug reservoir **108** and the floor **126** of the protective shell **128**), and an induction coil **136** may be integrated in the protective shell **128** (e.g., by injection molding). FIG. **2** more clearly illustrates a hermetic case **135** for housing the battery and conventional control circuitry **132**, but, for simplicity, does not depict the components housed therein. The hermetic case **135** may be made from biocompatible metals (e.g., titanium) or metal alloys. The bottom of the hermetic case **135** may be flat, or it may be concave to help the implantable pump **100** fit on the patient's eye **104**.

In one embodiment, the induction coil **136** permits wireless (e.g., radio-frequency) communication with an external device (e.g., a handset). The handset may be used to send wireless signals to the control circuitry **132** in order to program, reprogram, operate, calibrate, or otherwise configure the pump **100**. In one embodiment, the control circuitry **132** communicates electrically with electrolysis electrodes **134** in the electrolyte chamber **112** by means of metal interconnects (vias) **138** spanning a bottom portion of the electrolyte reservoir **112**. The electrolysis electrodes **134** may be made from, for example, platinum, gold, and/or other metal(s). As further described below, the control circuitry **132** controls the pumping action of the pump **100**, including the below-described closed-loop control process.

In one embodiment, as illustrated in FIG. **1**, the cannula **120** connects the drug reservoir **108** to a check valve **140** inserted at the site of administration. The check valve **140** may be a one-way check valve that prevents the backflow of any fluid into the drug reservoir **108**. Alternatively, or in addition, as illustrated in FIG. **2**, the check valve **140** may be integral with and located at a proximal end of the cannula **120** (i.e., at the end closest to the drug reservoir **108**). More generally, however, the check valve **140** may be located anywhere along the cannula **120**. In addition, one or more flow sensors **144** for monitoring the flow of the drug, and thereby enabling the measurement of the drug volume delivered and/or the flow rate of the drug through the cannula **120**, may be associated with one or more of a proximal, middle, or distal portion of the cannula **120**. Optionally, as illustrated in FIG. **1**, one or more target site sensor(s) **148** may also be integrated at a distal end of the cannula **120** (i.e., at the end furthest from the drug reservoir **108**) in order to measure one or more parameters at the site of administration (e.g., the intravitreal chamber, shoulder capsule, knee capsule, cerebral ventricals, spinal canal, etc.). For example, the target site sensor(s) **148** may be employed to sense one or more of a change in a biological or biochemical parameter at the target site (e.g., a change in a specific analyte concentration, the presence or absence of a specific biochemical marker, etc.), a change in electrical activity at the target site (which may, for example, be brought on by a physiological change), and a change in pressure at the target site. In one embodiment, the target site sensor(s) **148** provide feedback (i.e., real-time measurements) to the control circuitry **132** so that the flow of drug may be metered by a closed-loop control process. For example, increased pressure in the drug target region may warrant a decrease in the flow of drug from the pump **100**.

As illustrated in FIG. **1**, the cannula **120** may be an extension of the stand-alone parylene film **116**. Alternatively, as illustrated in FIG. **2**, the cannula **120** may be a separate component (e.g., a parylene component) that is coupled to the protective shell **128**. For example, a proximal end of the cannula **120** may be inserted through a fluid connection port formed in the protective shell **128** and bonded thereto by way of, e.g., a biocompatible epoxy glue **150**. A silicone sheath **154** may be placed around a portion of the cannula **120** (see FIG. **2**), but this is optional (see FIG. **1**).

In one embodiment, as illustrated in FIG. **1**, a fill port **152** is assembled with the drug reservoir **108** and sealed by a sealant (e.g., a biocompatible epoxy) **156** to the stand-alone film **116** and protective shell **128**. In yet another embodiment, as illustrated in FIG. **2**, a hole may be formed through the protective shell **128** and the fill port **152** featured therein. In still another embodiment, the fill port **152** may be formed elsewhere on the pump **100** and be connected to the drug reservoir **108** through tubing. For example, the fill port **152** may be molded from biocompatible materials, coupled to a matching notch on the hermetic case **135**, and connected to the drug reservoir **108** through the tubing. In one embodiment, the tubing is inserted through a fluid connection port formed in a wall surrounding the drug reservoir **108** and bonded thereto by way of a biocompatible epoxy glue. In either case, the fill port **152** is in fluid communication with the drug reservoir **108** and permits an operator of the pump **100** (e.g., a physician) to refill the drug reservoir **108** in situ (e.g., while the pump **100** is implanted within the patient's eye **104**). In general, the drug reservoir **108** can be refilled by inserting a refill needle into and through the fill port **152**.

In various embodiments, the main parts of the pump **100** (i.e., the pair of chambers **108**, **112** and the cannula **120**) are amenable to monolithic microfabrication and integration using multiple parylene layer processes. The fill port **152**, the protective shell **128**, and other components may be assembled with the pump **100** after the microfabrication steps.

In operation, when current is supplied to the electrolysis electrodes **134**, the electrolyte evolves gas, expanding the corrugated diaphragm **124** (i.e., moving the diaphragm **124** upwards in FIGS. **1** and **2**) and forcing liquid (e.g., drug) out of the drug reservoir **108**, into and through the cannula **120**, and out the distal end thereof to the targeted site of administration. The corrugations or other folds in the expandable diaphragm **124** permit a large degree of expansion, without sacrificing volume within the drug reservoir **108** when the diaphragm **124** is relaxed. When the current is stopped, the electrolyte gas condenses back into its liquid state, and the diaphragm **124** recovers its space-efficient corrugations.

B. Adaptive Control Based Upon Internal Pump Conditions

In general, the response of the electrolysis-driven pump **100** to a given input current supplied to the electrolysis electrodes **134** depends on how much liquid is remaining in the drug reservoir **108**. For example, if the drug reservoir **108** is nearly empty, more current is needed to bring the drug reservoir **108** to its "full" configuration before pressure can begin to build up and pumping can commence. On the other hand, if the drug reservoir **108** is completely full, very little current is needed before delivery of the drug begins. Similarly, the response of the electrolysis-driven pump **100** to a given input current also depends on the gas/liquid ratio in the electrolysis chamber **112**. In particular, the response of the pump **100** will be very different when the drug reservoir **108** is full with drug (e.g., when the electrolysis chamber **112** operates with a low gas/liquid ratio) than when the drug reservoir **108** is nearly empty (e.g., when the electrolysis chamber **112** operates with a high gas/liquid ratio). In addition, other factors can cause the response of the electrolysis-driven pump **100** to change over time including, for example, degradation of the electrolysis electrodes **134**, changes in the concentration of the electrolyte in the electrolysis chamber **112**, changes in the flow characteristics of the check valve **140**, and restrictions that form at the output of the cannula **120** due to tissue growth or some other mechanism.

Because of these factors, the electrolysis pump **100** is inherently variable. Accordingly, adaptive control in accordance herewith can confer substantial advantages upon the pump **100**. For example, as further explained below, by analyzing previous doses to ascertain how the pump **100** responded to given input currents, the optimal settings (e.g., the settings which give the most accurate and shortest dose) for the current dose can be derived. This can be particularly beneficial when the dose volume is small compared to the volume of the drug reservoir **108**. In such a situation, the state parameters of the pump **100** (e.g., the drug volume remaining in the drug reservoir **108**, the liquid/gas ratio in the electrolysis chamber **112**, the condition of the electrodes **134**, the characteristics of the check valve **140**, etc.) are nearly identical from one dose to the immediately following dose, and, as such, the previous doses are an excellent predictor for the current dose.

FIG. **3** is a block diagram of a drug-delivery pump **200** that depicts the control circuitry **132** in greater detail. The drug-delivery pump **200** may be any type of external or internal pump having an actuator **204** that forces the liquid from the drug reservoir **108** into and through the cannula **120**. For example, the drug-delivery pump **200** may be an electrolysis-driven pump and, with reference to FIGS. **1** and **2** described above, the pump actuator **204** may include the electrolyte chamber **112**, the expandable diaphragm **124**, and the electrolysis electrodes **134**. For its part, the control circuitry **132** includes computer memory **208** for storing one or more conditions of the pump **200**, and an adaptive controller **212** for controlling the pump actuator **204** based on a change in a condition of the pump **200**. Optionally, the control circuitry **132** may also include one or more module(s) to convert raw data received from the flow sensor **144** into a meaningful value (e.g., into a flow rate in nL/min) and/or to convert similarly raw data received from the pump actuator **204** into a meaningful value. Alternatively, the functions performed by such module(s) may instead be performed by the adaptive controller **212**.

The computer memory **208** may be implemented as any type of volatile or non-volatile (e.g., Flash) memory, while the adaptive controller **212** and/or the module(s) described above may each be implemented as any software program, hardware device, or combination thereof that is capable of providing the functionality described herein. For example, the adaptive controller **212** and/or the module(s) described above may each be an application-specific integrated circuit (ASIC) or a field-programmable gate array (FPGA). Alternatively, the adaptive controller **212** may be implemented using a general-purpose microprocessor (e.g., any of the PENTIUM microprocessors supplied by Intel Corp.) that is programmed using any suitable programming language or languages (e.g., C++, C#, Java, Visual Basic, LISP, BASIC, PERL, etc.). Suitable control programming is straightforwardly implemented by those of skill in the art without undue experimentation.

In one particular embodiment, as further described below, the control circuitry **132** is programmed to deliver a fixed dosage of the drug from the drug reservoir **108** to the target site at periodic time intervals, and is configured to store the conditions of the pump **200** at each of those time intervals in the computer memory **208**. Some exemplary and non-limiting conditions internal to the pump **200** that may be stored at each dosing interval (or at other periodic intervals) include the current through, voltage across, or resistance of the electrolysis electrodes **134**; the total electrical charge used to drive the electrolysis electrodes **134**; the maximum flow rate of the drug through the cannula **120**; any variations in flow patterns of the drug through the cannula **120**; the actuation time required for the pump **200** to achieve a particular flow rate of the drug through the cannula **120**; the time required for the flow of drug to ramp down from a particular flow rate to a flow rate of zero; the time delay between the initial actuation of the pump **200** and the initial flow of drug through the cannula **120**; the efficiency of the pump actuator **204** (which, in the case of an electrolysis-driven pump **200**, may be defined as the ratio between the amount of charge pumped through the actuator **204** and the amount of gas generated thereby); the internal pressure of the drug reservoir **108**; the acceleration experienced by the pump **200**; flow sensor parameters particular to the flow sensor **144** architecture (e.g., where the flow sensor **144** is a resistive temperature detector, the resistance of the sensor and heater elements may be stored); and the physical dimensions of the pump actuator **204**, the drug reservoir **108**, and/or the cannula **120**, which may change due to blockages, scarring, or other biochemical/physiological events.

In one embodiment, these parameters are measured either directly or indirectly by using physical sensors, such as, for example, the flow sensor(s) **144**, pressure sensors in the drug reservoir **108** or cannula **120**, accelerometers, gyroscopes, altimeters, sensors in proximity to the electrolysis electrodes **134** (to measure, for example, their resistance, the current passing therethrough, and/or the voltage thereat or thereacross), or any other sensor dispersed throughout the pump **200**. In other embodiments, these parameters are determined by using known relationships. For example, the flow rate of the drug through the cannula **120** may be determined by using a pressure sensor in the cannula **120** and by utilizing the well-known linear relationship between pressure and flow rate. In still other embodiments, many of these parameters may ascertained by analyzing the electrical waveforms used to drive the pump actuator **204**, and/or by analyzing the flow profiles sensed by the flow sensor(s) **144**.

In all cases, as further described below, the adaptive controller **212** of the control circuitry **132** can receive and process this parameter data and compensate for any change in a condition of the pump **200** in order to adjust its operation to maintain a target dosage level. This “self-compensation” may be achieved by storing, as mentioned above, parameter data from the pump **200** state at the time of the previous dosages and by considering real-time parameter values to determine the optimal actuation current for the electrolysis electrodes **134** and/or their actuation duration at the next dosing event. For example, as illustrated in FIG. **3**, the adaptive controller **212** may receive, analyze, and process the stored parameters from previous doses, real-time data from the pump actuator **204**, and real-time data from the flow sensor(s) **144** (e.g., flow rate data) to ascertain and direct appropriate output signals to the pump actuator **204** (i.e., in order to drive the pump **200** in the appropriate manner). For initial dosing, or in cases where the above-described data may be unavailable (e.g., due to a reset action in the pump **200**), the adaptive controller **212** may employ a set of pre-defined reference parameter values. These reference values may be specific to the characteristics of the particular pump **200** employed, for example specific to the types of electrolysis electrodes **134** employed, the type of electrolytic solution used, and/or the physical dimensions of the pump actuator **204**, drug reservoir **108**, and cannula **120**.

In one mode of operating an electrolysis-driven pump **200**, the electrolysis electrodes **134** are driven using a constant current for a variable amount of time. In this mode, the constant current results in a monotonic rise in the flow rate of the drug through the cannula **120** until the current is shut off, at which point the residual pressure in the pump **200** gives rise to a slow decay in the flow rate until the flow rate reaches zero. In one functional example for this mode of operation, the following three parameters are stored in the computer memory **208** at each dosing interval: the current supplied to the electrolysis electrodes **134** in order to drive the pump **200** (I); the maximum flow rate of the drug through the cannula **120** (F_{max}); and the volume of liquid (i.e., drug) that is delivered by the pump **200**, due to residual pressure, after the pump actuator **204** is deactivated ($V_{shutoff}$). This stored information is then used, in future doses, to improve the dosing speed and accuracy. For example, the current used to drive future doses may be adjusted based on previous dose data (e.g., increased if the maximum flow rate is too low, and decreased if the maximum flow rate is too high) in order to keep the duration of each dose, and the volume of the drug delivered on each dose, relatively consistent. In one embodiment, this is done in a linear fashion as follows:

$$I_{current}=F_{optimal}/F_{max,previous}\times I_{previous}$$

where $I_{current}$ is the current to be supplied to the electrolysis electrodes **134** during the current dose, $F_{optimal}$ is the desired maximum flow rate of the drug through the cannula **120**, $F_{max,previous}$ was the maximum flow rate of the drug through the cannula **120** during the previous dose, and $I_{previous}$ was the current supplied to the electrolysis electrodes **134** during the previous dose.

As another example, the shut-off time of the pump actuator **204** may instead, or in addition, be adjusted (e.g., shut off later if the volume of the liquid delivered after the pump actuator **204** is deactivated is lower than expected, and shut off earlier if the volume of the liquid delivered after the pump actuator **204** is deactivated is higher than expected) in order to keep the volume of the drug delivered relatively consistent. Once again, this may be done using a linear approximation, where the pump actuator **204** is deactivated as soon as the following condition is met:

$$V_{accumulated}+F/F_{max,previous}\times V_{shutoff,previous}=V_{target}$$

where $V_{acuumulated}$ is the total volume of the drug delivered so far in the current dose, F is the real-time flow rate of the drug through the cannula **120**, $F_{max,previous}$ was the maximum flow rate of the drug through the cannula **120** from the previous dose, $V_{shutoff,previous}$ was the volume of the drug delivered after the pump actuator **204** was shut off in the previous dose, and V_{target} is the target volume of the drug to be delivered. In this manner, the adaptive controller **212** constantly adjusts the way in which the pump **200** is actuated, and accounts for systematic, non-random changes in the pump **200** characteristics.

Determining and controlling both the amount of current needed to initiate the flow of drug through the cannula **120** and then to reach a particular flow rate, as well as the amount of liquid delivered from the drug reservoir **108** after the current is no longer applied to the electrolysis electrodes **134**, is of particular benefit when the pump **200** is an electrolysis-driven pump. In particular, the first parameter is important because the amount of current needed to initiate the flow of drug through the cannula **120** and to reach a particular flow rate depends on how much liquid is left in the drug reservoir **108**. Using too low a current would be power-inefficient, since all systems would be running even though there would be no or very low flow of drug through the cannula **120**. On the other hand, using too high a current could cause the flow rate of the drug to overshoot to unsafe levels. The second parameter is also of importance since the volume of drug delivered after the pump **200** is turned off is dependent primarily on the gas/liquid ratio in the electrolysis chamber **112**. For doses later in the life-cycle of the pump **200** (e.g., where the pump **200** runs with a high gas/liquid ratio in the electrolysis chamber **112**), there is much more gas that needs to be dissipated before the pump **200** can fully stop. The opposite is true for earlier doses.

As will be understood by one of ordinary skill in the art, in addition to the two examples given above, the adaptive controller **212** may recognize and analyze numerous other changes in conditions internal to the pump **200** when controlling the pump actuator **204** and, ultimately, the dispensing of the drug from the drug reservoir **108**. For example, there may be situations where is it desirable for the pump **200** to reach an optimal flow rate ($F_{optimal}$) for each dose in a specified period of time ($t_{optimal}$) and to then maintain that flow rate for the remainder of the dose. One way to achieve this is to begin each dose by using a constant current ($I_{starting}$) to drive the electrolysis electrodes **134** of the pump **200** until the optimal flow rate ($F_{optimal}$) is reached, at which point feedback from the flow sensor **144** and an algorithm (e.g., a proportional-integral-derivative ("PID") algorithm or another algorithm) may be used to adjust the current supplied to the electrolysis electrodes **134** to maintain that optimal flow rate ($F_{optimal}$) for the remainder of the dose. In other words, the pump **200** may be driven using a time-varying current waveform. In one embodiment, in order to achieve the optimal flow rate ($F_{optimal}$) in the specified period of time ($t_{optimal}$), the starting current ($I_{starting}$) is adjusted from dose to dose. In a manner similar to before, this can be done, for example, using a linear approximation (although, as will be understood by one of ordinary skill in the art, non-linear approximations may also be employed for any of the parameters derived herein). More specifically, the starting current for the current dose ($I_{starting,current}$) can be calculated using the starting current from the previous dose ($I_{starting,previous}$) and the time it took for the flow rate to reach the optimal flow rate ($F_{optimal}$) in the previous dose ($t_{previous}$), as follows:

$$I_{starting,current}=t_{previous}/t_{optimal}\times I_{starting,previous}$$

Referring now to FIG. **4**, an exemplary graph **300** illustrating the effects of the above-described adaptive control on the drug dosage level is depicted. In this example, the target dosage level to be delivered during each release event is 200 nanoliters (nL). Event **1** corresponds to an initial dosing of 180 nL based on calculations using the reference parameter values. The adaptive controller **212** then calculates appropriate adjustments to the pump **200** parameters (e.g., as described above, the amount of current supplied to the electrolysis electrodes **134** and/or the actuation time thereof may be increased in order to increase the volume of drug delivered to the target site) until a target delivery of 200 nL is achieved at Event **2**. As illustrated, there may be a point **304** in time between Event **1** and Event **2** during which the adaptive controller **212** overcompensates and the pump **200** delivers more than the target dosage (e.g., 205 nL). In this case, the adaptive controller **212** refines its adjustments to the pump **200** parameters (e.g., as described above, the amount of current supplied to the electrolysis electrodes **134** and/or the actuation time thereof may be decreased in order to decrease the volume of drug delivered to the target site) until the target delivery of 200 nL is in fact achieved at Event **2**.

Continuing with the example depicted in the graph **300** of FIG. **4**, the dosage at Event **3** then drops to 190 nL due to a change in one or more of the pump **200** parameters. Exemplary conditions within the pump **200** itself that may change and lead to such a decrease in the dosage of the drug delivered (i.e., to a decrease in the efficiency of the pump **200**) can include the degradation (e.g., erosion or corrosion) of the electrolysis electrodes **134**, a decrease in the concentration of the electrolytes in the solution present in the electrolysis chamber **112**, and/or general mechanical or chemical wear. In response, the adaptive controller **212** then compensates as described above so that the pump **200** releases the correct amount of drug at Event **4**. The pump **200** thus dynamically reacts to changing conditions of the pump **200**.

FIG. **5**A depicts exemplary flow profiles **400** and actuation profiles **404** for a pump that operates without the feedback control provided by the control circuitry **132** (e.g., for a pump employing an open-loop control system). As shown, the amount of drug delivered at later times decreases even though the actuation current remains the same (the actuation profiles **404** for the earlier and later doses overlap in FIG. **5**A), due to decreasing pump efficiency.

FIG. **5**B depicts exemplary flow profiles **408** and actuation profiles **412** for a pump **200** that operates with the feedback control provided by the control circuitry **132**. In particular, FIG. **5**B shows how increasing the pumping time for a later dose can compensate for reduced pump **200** efficiency. More specifically, for the later dose, the pump **200** actuates for a longer period of time at the same current in order to successfully deliver the target dosage amount.

FIG. **5**C also depicts exemplary flow profiles **416** and actuation profiles **420** for a pump **200** that operates with the feedback control provided by the control circuitry **132**. In particular, FIG. **5**C shows how the dosing time for the earlier and later doses can be kept constant while still compensating for decreased pump **200** efficiency by increasing the actuation current of the later dose. The flow profiles **416** for the earlier and later doses overlap, illustrating that the same amount of drug is delivered during both dosages.

C. Adaptive Control Based Upon Conditions of the Target Site

In other embodiments, with reference again to FIGS. **1-3**, the adaptive controller **212** can also receive information from the target site sensor(s) **148** that monitor the drug-delivery treatment area, and thereafter change the target dosage for certain time periods. More particularly, if changes in the treatment area (e.g., worsening or improvement of symptoms, changes in biological or biochemical parameters, changes in electrical activity, changes in pressure, etc.) require a higher or lower dosing level or a change in the frequency of dosages, the adaptive controller **212** can control the pump actuator **204** so as to adjust the dosage and maintain it at a new level until another change is required. In other words, the adaptive controller **212** may actuate the pump **200** to achieve a desired result, such as the regulation of a specific physiological state or biochemical parameter. As before, the parameters sensed by the target sensor(s) **148** (e.g., pressure, temperature, etc.) may be stored in the computer memory **208** for later use (e.g., for comparison in determining the appropriate dosage of drug to be delivered).

As an example, assume that the pump **200** delivers an initial target dosage of 200 nL every 30 minutes. After a period of time, either due to a change in the treatment area or dosing regimen, the dosage may need to be decreased to 150 nL. The adaptive controller **212** may then operate the pump actuator **204** so as to deliver 150 nL of the drug every 30 minutes until instructed otherwise, either by another change in the treatment area or by a user of the pump **200**.

Advantageously, this flexibility facilitates the use of the pump **200** with a wide range of treatment regimens that may require the staggering of different dosages or dosage frequencies over prolonged periods of time.

Optionally, the adaptive controller **212** may be programmed to respond to both a change in a condition of the pump **200** itself and, at the same time, to a change in condition of the target treatment area. In other words, the adaptive controller **212** may receive data from both sensors or other devices internal to the pump **200** and from the target site sensor(s) **148**, analyze both sets of data, and control the pump actuator **204** to account for both sets of data. Alternatively, in another embodiment, if the deterministic parameters are to be those of the pump **200** itself rather than those of the treatment area, the adaptive controller **212** may be programmed to refrain from initiating actions based on, for example, blockages that may form within the target area due to physiological changes or scarring.

D. Exemplary Uses of the Dynamic, Adaptively Controlled Drug-Delivery Pumps

FIG. **6** schematically illustrates a drug-delivery pump **100**, **200** implanted in the eye of a patient in accordance with one embodiment of the invention. As illustrated, the pump **100**, **200** is placed upon the conjunctiva of the eye, and a distal end of the cannula **120** is inserted therethrough in to the posterior chamber of the eye. As such, the distal end of the cannula **120** (and, hence, the drug reservoir **108**) is in fluid communication with the patient. The drug-delivery pump **100**, **200** then administers a therapeutic liquid to the posterior chamber of the eye through the cannula **120** and the check valve **140**, which, as previously mentioned, may be employed to prevent the backflow of the liquid. In particular, the pump actuator **204** may be controlled through use of the adaptive controller **212** and the other control circuitry **132** in any of the manners described hereinabove (e.g., based on a change in a condition of the pump itself and/or based on conditions of the target site) so as to deliver one or more dosages of the liquid from the drug reservoir **108**, through the cannula **120**, and into the patient's eye.

In other embodiments, the pump **100**, **200** is used to administer the liquid to the anterior chamber of the eye, which is separated from the posterior chamber by the lens. More generally, however, the pump **100**, **200** may, as previously mentioned, be employed to administer liquid to any portion of the patient's body.

As an additional example, the pump **100**, **200** may be a body-adhered electrolysis-driven pump for the infusion of medication into a patient's subcutaneous tissue. For example, the pump **100**, **200** may continuously deliver insulin to the patient's body over three to seven days. A patient may need, however, to recalculate his or her insulin delivery (e.g., increase or decrease basal rates over time), as well as program the pump **100**, **200** to give an intermittent bolus spike of insulin after a meal. Accordingly, the pump **100**, **200** in this example can adapt the electrolysis to increase or decrease the flow of insulin to accurately deliver the correct fluidic volumes over time. Furthermore, infusion of a drug over an extended period of time, such as three days, may subject the pump **100**, **200** to new environmental conditions. For example, a patient may drive from low to high altitudes or fly in a pressurized plane. The pump **100**, **200** can use both environmental signals (e.g., altimeter, pressure change, flow rate change, etc.) to adjust the flow of the drug and to ensure the accurate delivery of the drug.

As yet another example, the pump **100**, **200** may use input from an accelerometer or gyroscope in order to sense a patient's position. For example, the pump **100**, **200** may sense that the patient was horizontal during the hours of 10 pm to 6 am for the previous 7 days (because, for example, the patient was sleeping). In this case, the pump **100**, **200** may then recognize the patient's sleep time (i.e., from sensing the patient to be in a horizontal position) or REM sleep cycle and then use that information to infuse a different volume of drug (or drug at specific times) to accommodate optimal conditions. For example, the flow rate of the pump **100**, **200** may be adjusted to an amount pre-prescribed by a physician for infusion during sleep (e.g., it is often best to inject some glaucoma medications to a patient's eye during REM sleep cycle in order to better distribute the medication throughout the eye, while some medications such as Anti-VEGF drugs for the retina act over a period of a month and should be injected calmly into the vitreous; in addition, a lower basal rate of insulin or less pain medication may be injected during sleep). In contrast to understanding when a patient is sleeping, the pump **100**, **200** may also recognize when the patient is exercising or when the patient is not supine, and adjust its infusion of drug accordingly (e.g., such as to that which is pre-programmed by the physician for infusion during certain activities).

Advantageously, the control circuitry **132** described herein can be employed for pumps that are not uniform in their characteristics, either due to user-selected preferences or variations arising during the manufacturing process. The types of electrodes and electrolytic solution used, for example, determine the performance of electrolysis-driven pumps. The control circuitry **132** is, however, robust and versatile enough to accommodate pumps that operate across a wide range of parameter values. As another example, manufacturing process variations in the resistance of the flow sensor elements can be mitigated by the adaptive nature of the control circuitry **132**. More specifically, mismatched resistances in the flow sensor elements resulting from the process variations will result in an offset for which the control circuitry **132** can compensate.

Optionally, the control circuitry **132** may also serve to enhance safety and efficacy of the pump **100**, **200** by monitoring certain key pump parameters. For example, acceptable ranges may be defined for each parameter or for some overall combination of parameters corresponding to a specific pump state, during which the pump **100**, **200** continues to operate normally. Should an individual parameter or some combination of parameters not fall within these pre-defined ranges, an action may then be triggered within the pump **100**, **200**, such as shutting off or alerting the user that a response is required. For example, the pump **100**, **200** may alter a patient by illumination, sound, vibration, or shock. In one embodiment, the alert is programmed to occur when the patient is moving to maximize the likelihood that the patient will receive the alert and also to conserve battery power by avoiding alerts while the patient is sleeping.

In one particular example, the control circuitry **132** can respond to and predict the failure of a flow sensor **144**. Where, for example, the flow sensor **144** includes a group of heaters and resistive temperature detectors, one of its elements may begin to fail after an indeterminate number of doses due to thermal stresses experienced during its use. The control circuitry **132** can monitor the resistance of the heater elements periodically (e.g., from dose to dose) and detect changes in resistance that may indicate the start of failure or outright failure (such as an open-circuit). Other pump components including sensors and actuators that employ resistive or capacitive elements can likewise be monitored by the control circuitry **132** to ensure proper functional operation.

Having described certain embodiments of the invention, it will be apparent to those of ordinary skill in the art that other embodiments incorporating the concepts disclosed herein may be used without departing from the spirit and scope of the invention. For example, although the adaptive controller **212** and the other control circuitry **132** has primarily been described for use in connection with an electrolysis-driven pump, this is for illustrative purposes only. Those of ordinary skill in the art will readily appreciate and understand that the adaptive controller **212** and the other control circuitry **132** may also be usefully employed in other types of drug-delivery pumps, such as those that rely on, for example, electroosmosis, mechanical actuation, or pressure-driven mechanisms. Accordingly, the described embodiments are to be considered in all respects as only illustrative and not restrictive.

What is claimed is:

1. A method of delivering a drug to a patient from drug-delivery pump comprising a drug reservoir, a cannula, and a pump actuator for forcing a liquid drug from the drug reservoir into the patient via the cannula at period time intervals, the method comprising:

establishing fluid communication between the drug reservoir and the patient;

measuring at least one quantitative electrical or flow pump operating parameter;

storing (i) a fixed dosage of the liquid drug to be delivered by the pump actuator through the cannula during each of a plurality of dosing intervals and (ii) a value of the at least one quantitative pump operating parameter measured during a previous dosing interval, the pump actuator being operative during the previous dosing interval to deliver the fixed dosage; and

controlling the pump actuator by computing actuator settings based at least in part on the stored fixed dosage and a change in a condition of the pump actuator specified by a quantitative difference between the stored value of the pump operating parameter and a current value of the pump operating parameter, and adjusting the pump actuator in accordance with the computed actuator settings to thereby compensate for the change in actuation time required for the drug-delivery pump to achieve a target flow rate of the liquid drug through the cannula or a time required for the flow of the liquid drug to decrease from the target flow rate to a flow rate of zero;

wherein the drug-delivery pump is an electrolysis-driven pump and the pump actuator comprises electrolysis electrodes driven by a current, and wherein the controlling the pump actuator comprises varying actuation current supplied to the electrolysis electrodes.

2. The method of claim **1**, wherein controlling the pump actuator comprises maintaining delivery of a substantially fixed dosage of the liquid at the periodic time intervals to the patient.

3. The method of claim **2** further comprising storing conditions of the pump actuator at each time interval.

4. The method of claim **1** further comprising measuring a flow rate of the liquid drug into the patient.

5. The method of claim **4**, wherein controlling the pump actuator comprises analyzing at least one of the flow rate, stored conditions of the pump actuator from previous dosing interval, or real-time data from the pump actuator.

6. The method of claim **1**, wherein controlling the drug-delivery pump comprises varying an actuation duration of the electrolysis electrodes.

7. The method of claim **1**, wherein controlling the drug-delivery pump comprises driving the electrolysis electrodes with a constant current.

8. The method of claim **1**, wherein controlling the drug-delivery pump comprises driving the electrolysis electrodes with a time-varying current waveform.

9. The method of claim **1**, wherein controlling the pump actuator comprises maintaining delivery of a substantially fixed dosage of the liquid drug over time through continuous infusion to the patient.

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