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(54) **EXTRACORPOREAL GAS EXCHANGE DEVICES, SYSTEMS AND METHODS**

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A61M 1/16 (2006.01)
A61M 1/10 (2006.01)

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(Continued)

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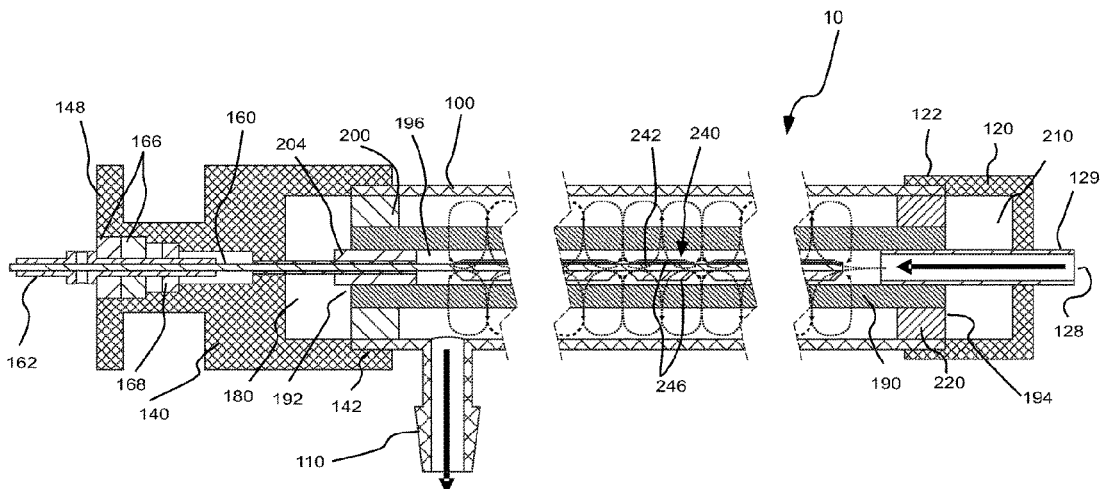
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(57) **ABSTRACT**

An extracorporeal gas exchange device includes a housing, a rigid shaft rotatable within the housing, a plurality of agitation mechanisms positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the plurality of hollow gas permeable fibers. The plurality of hollow gas permeable fibers are positioned radially outward from the plurality of agitation mechanisms. The rotational speed of the rigid shaft is adjustable independent of the flow rate of fluid through the housing.

18 Claims, 14 Drawing Sheets



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A61M 1/265
See application file for complete search history.

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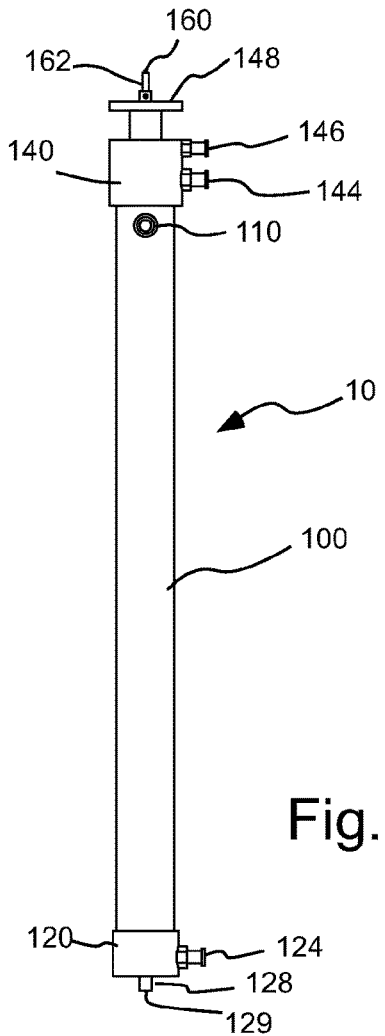


Fig. 1A

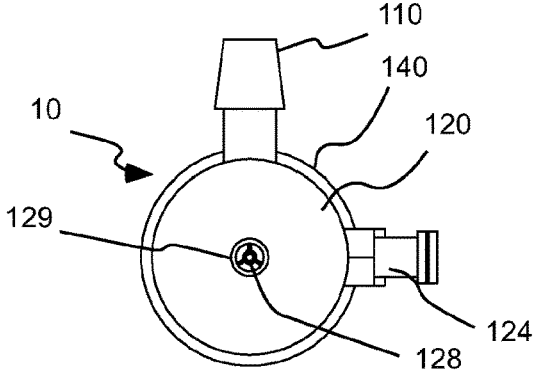


Fig. 1B

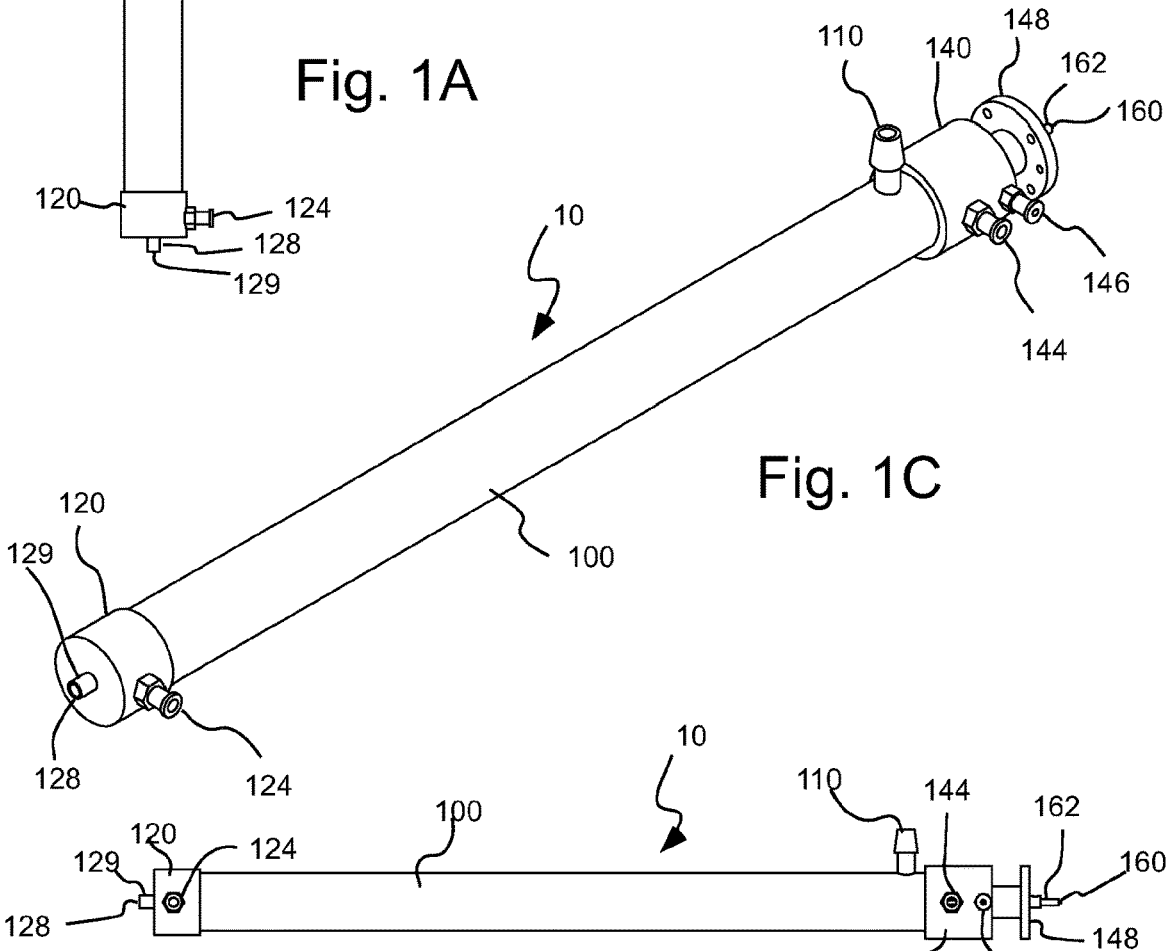


Fig. 1C

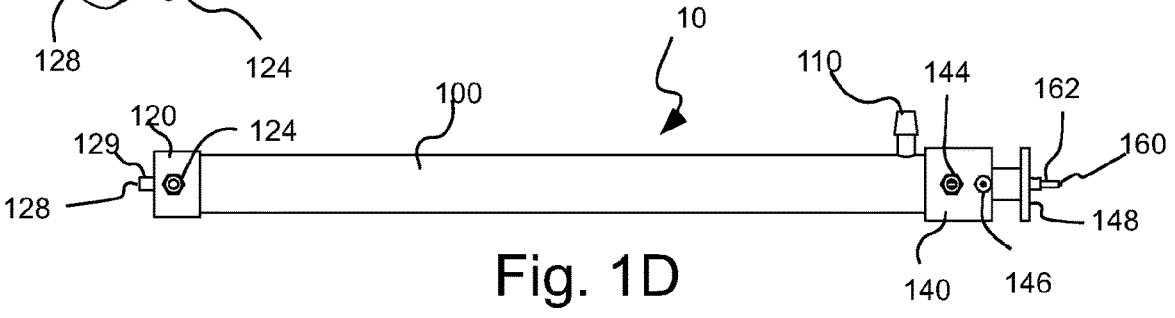
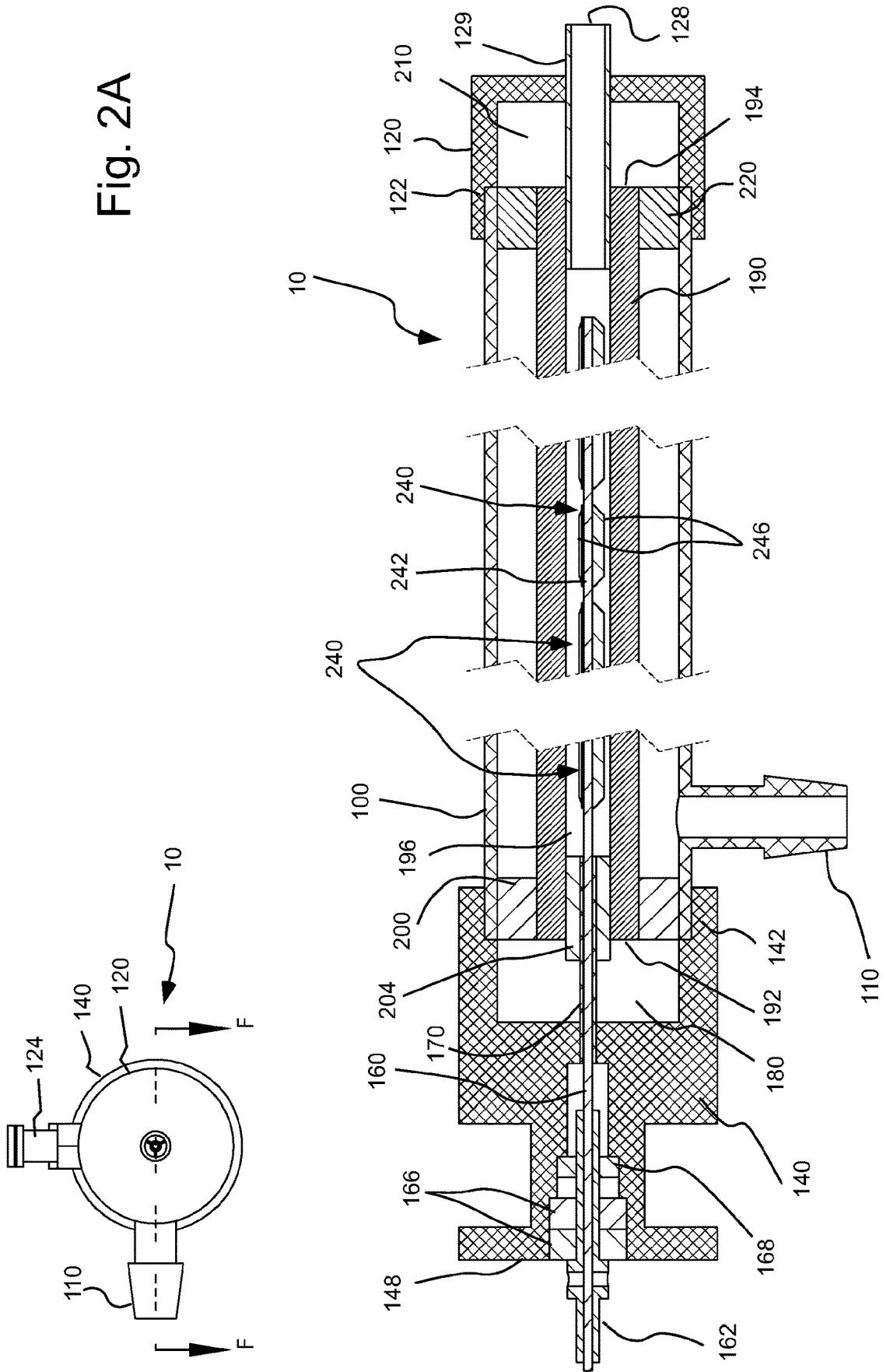


Fig. 1D

Fig. 2A



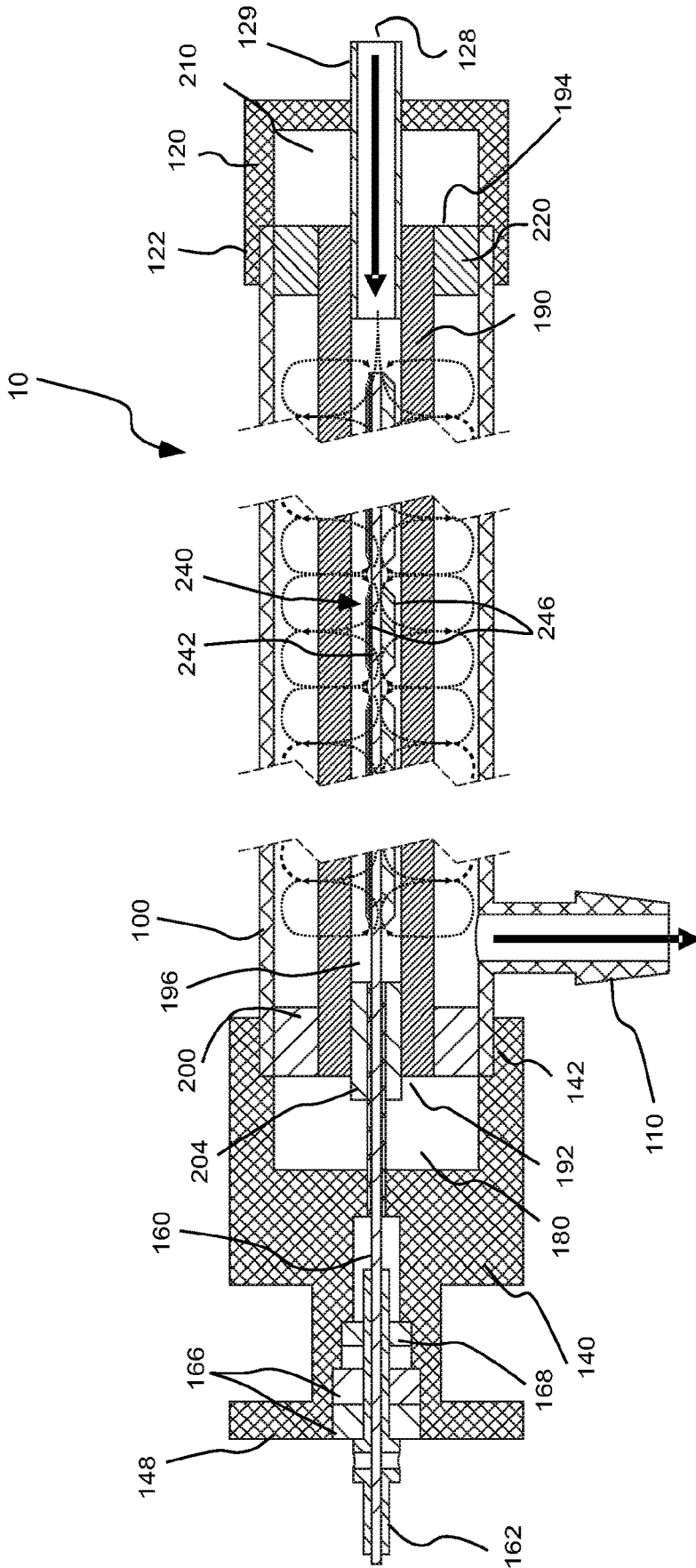


Fig. 2B

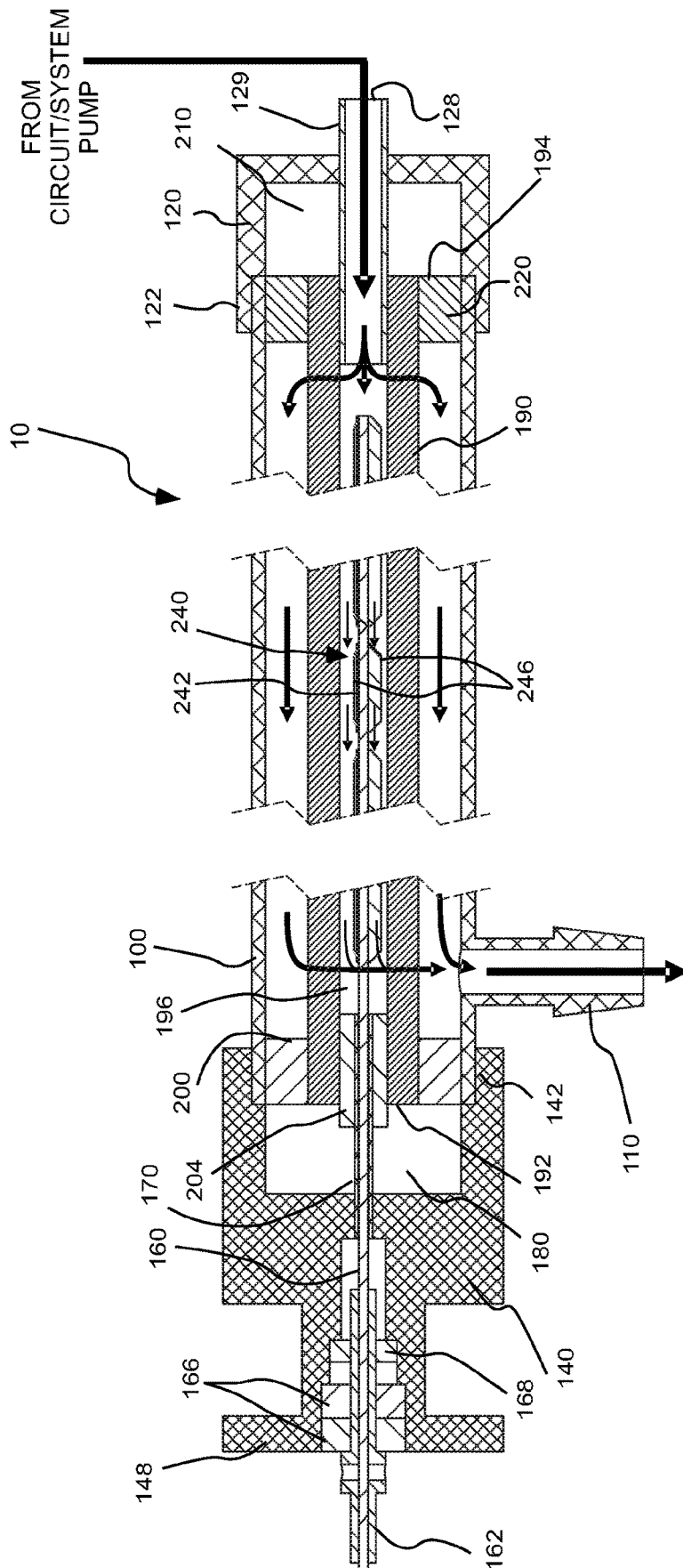


Fig. 2C

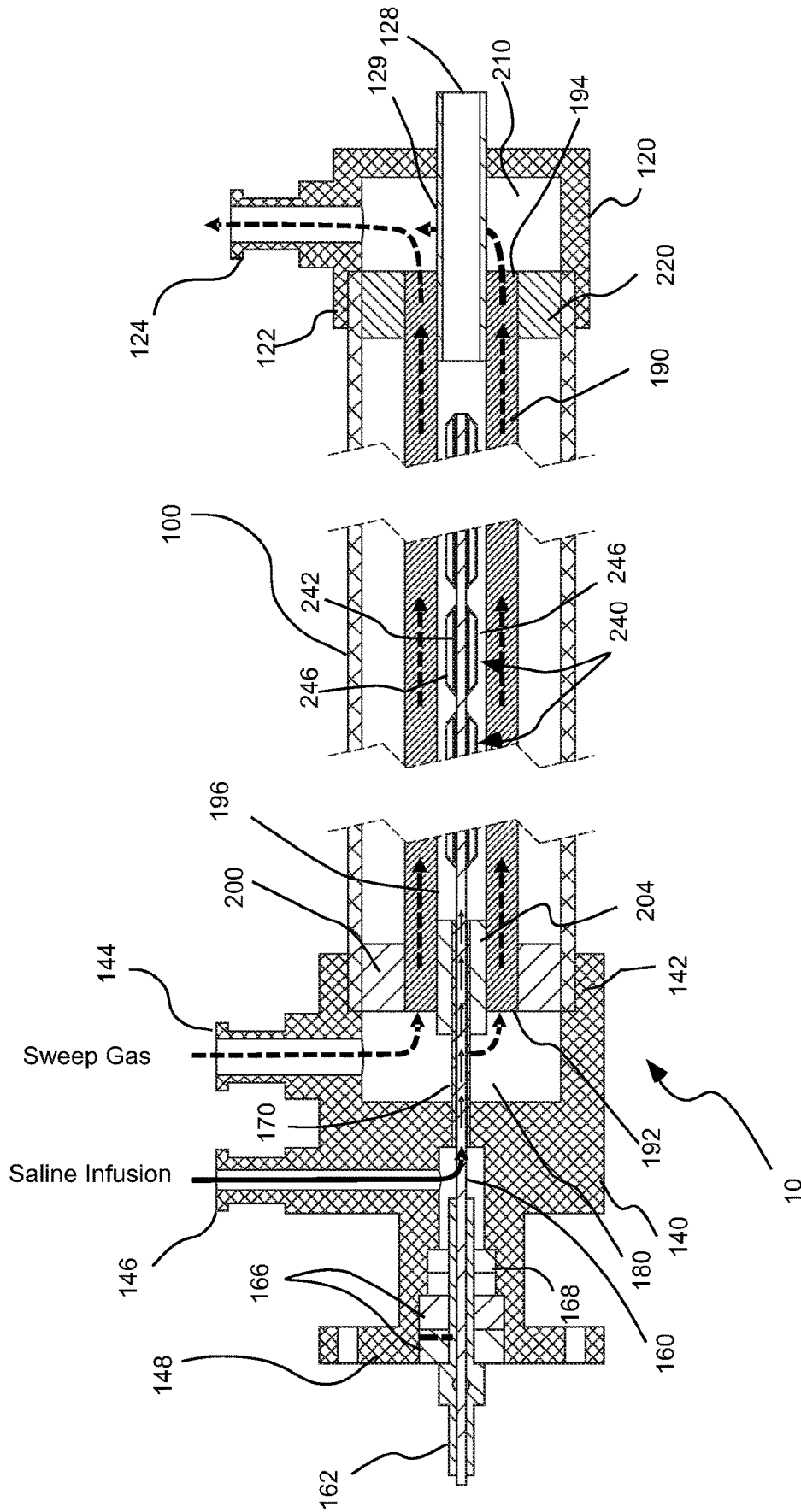


Fig. 2E

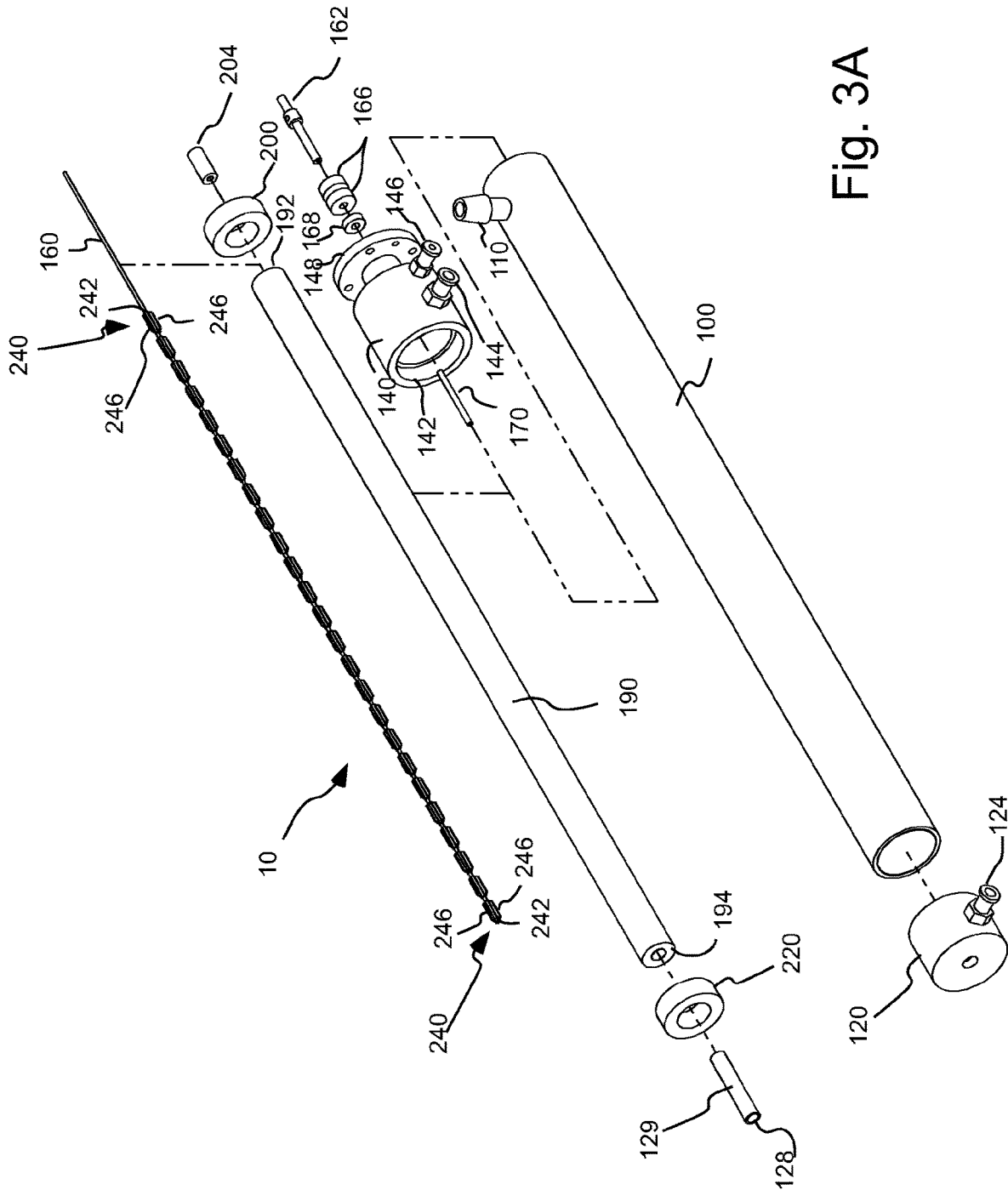


Fig. 3A

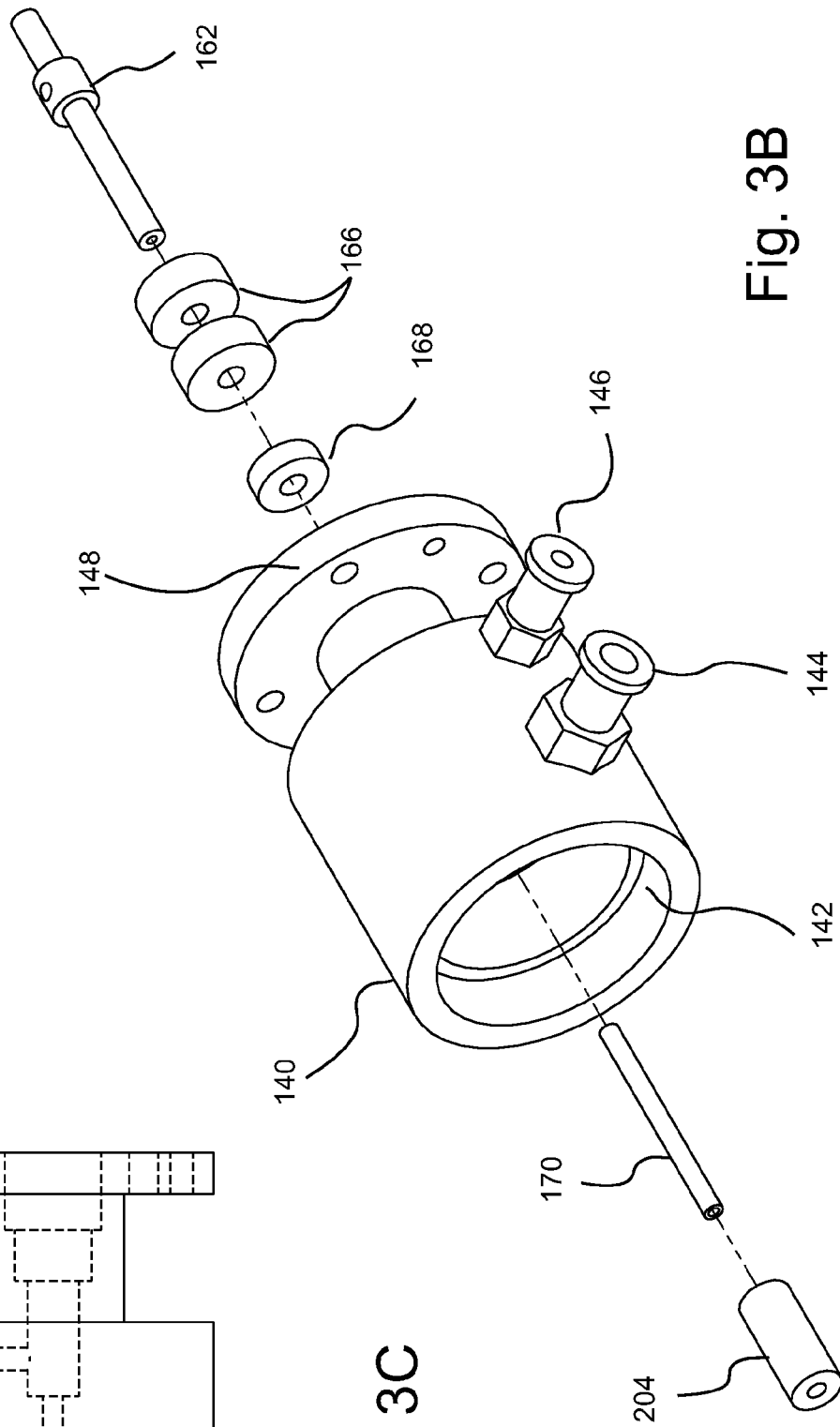
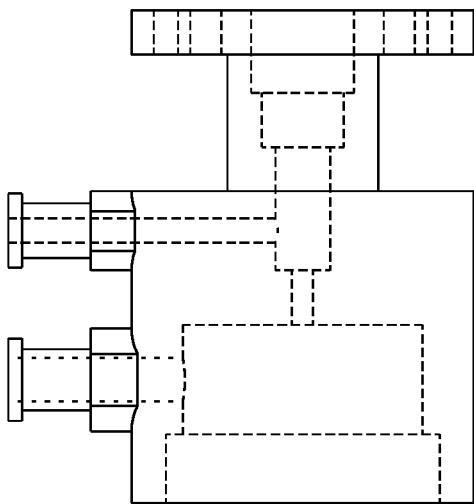


Fig. 3C

Fig. 3B

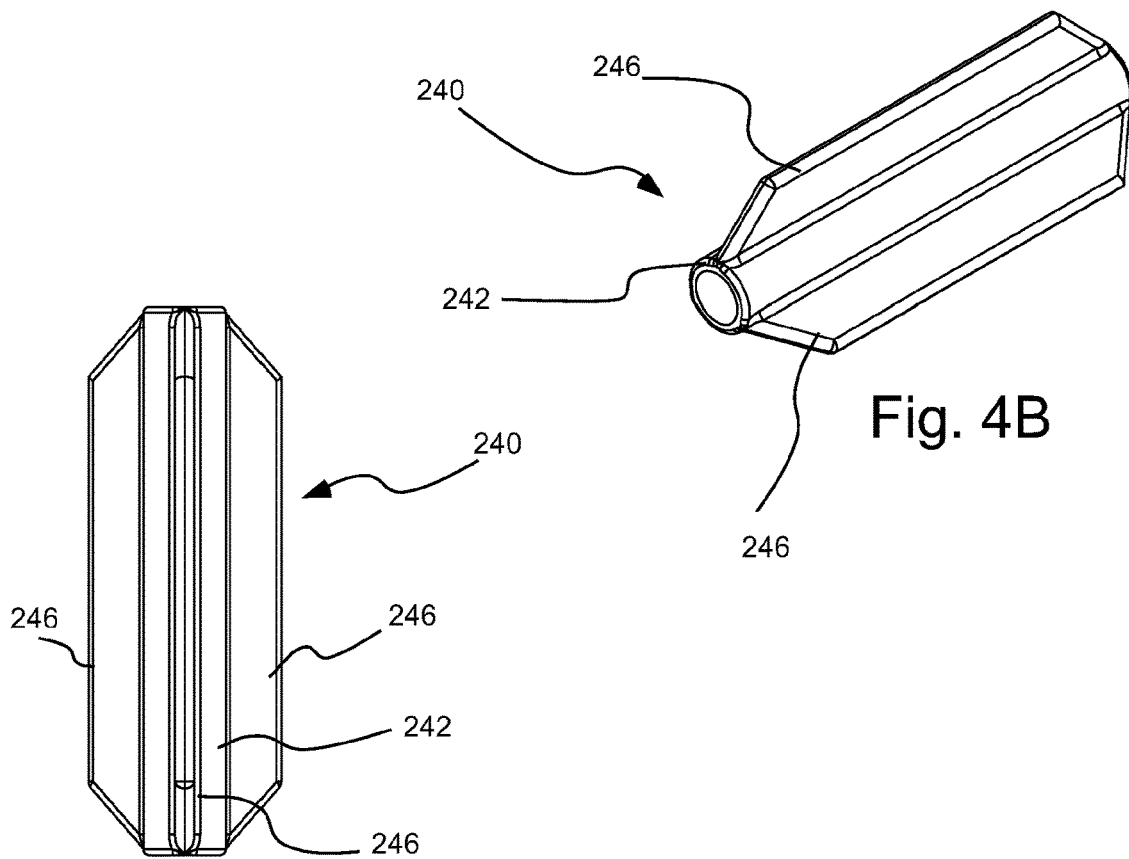


Fig. 4A

Fig. 4B

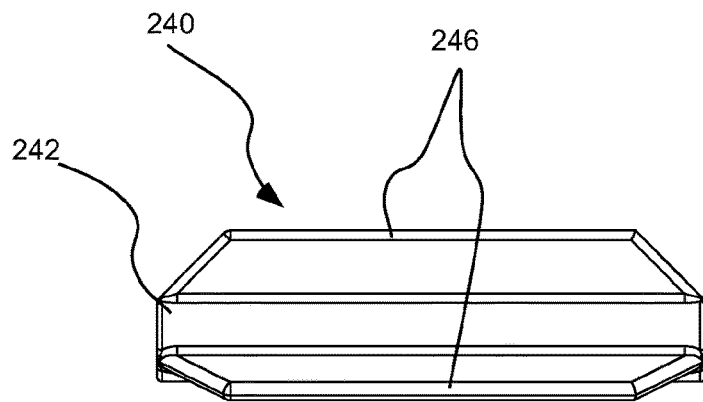


Fig. 4C

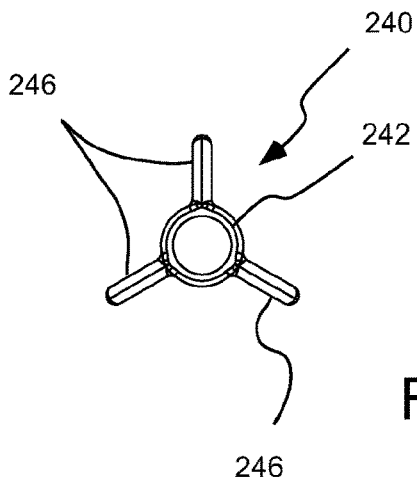


Fig. 4D

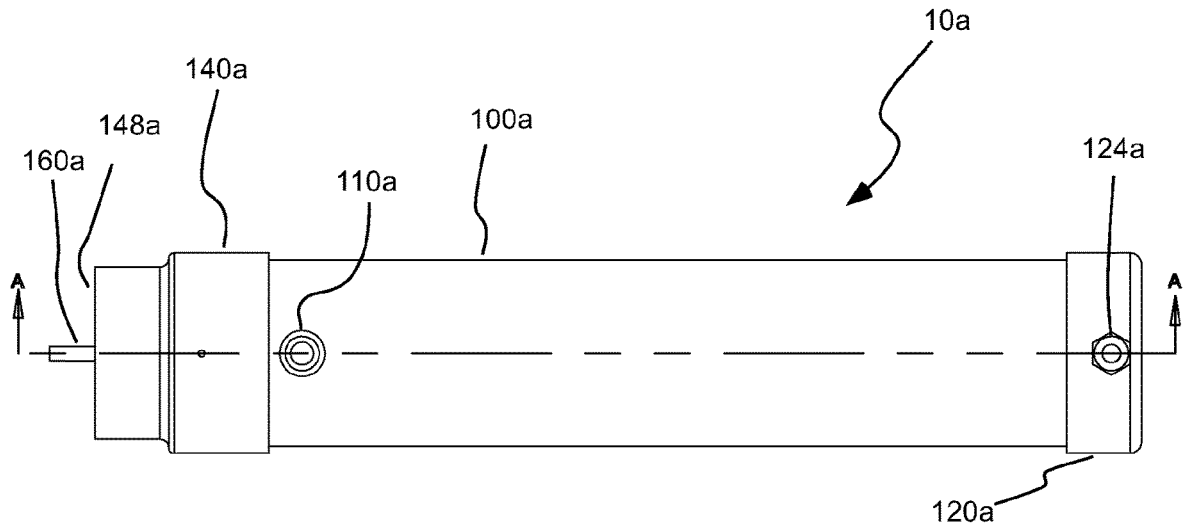


Fig. 5A

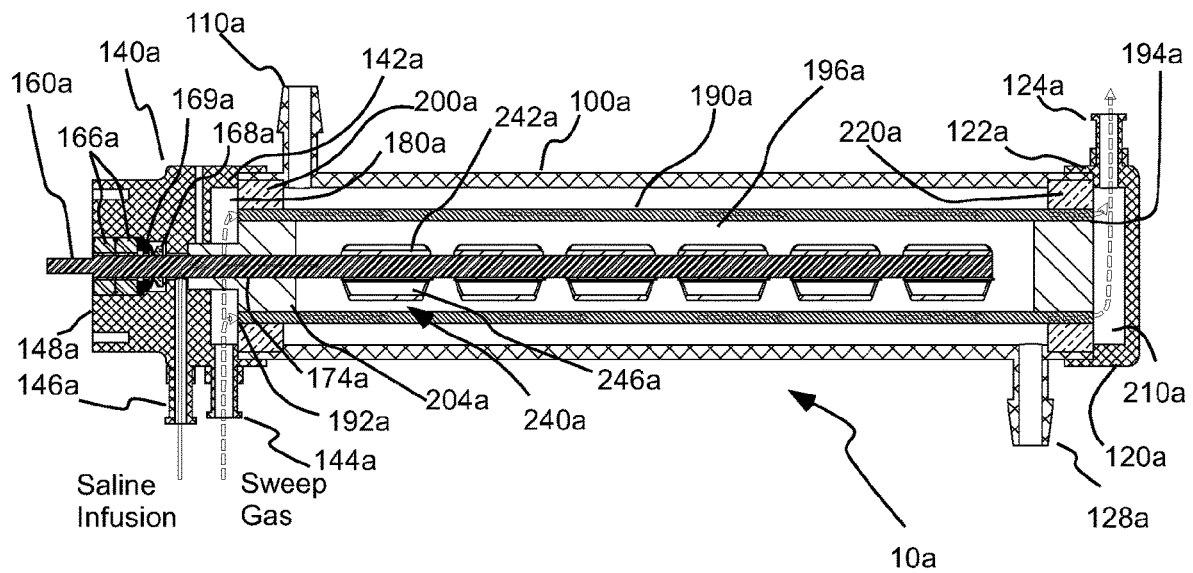


Fig. 5B

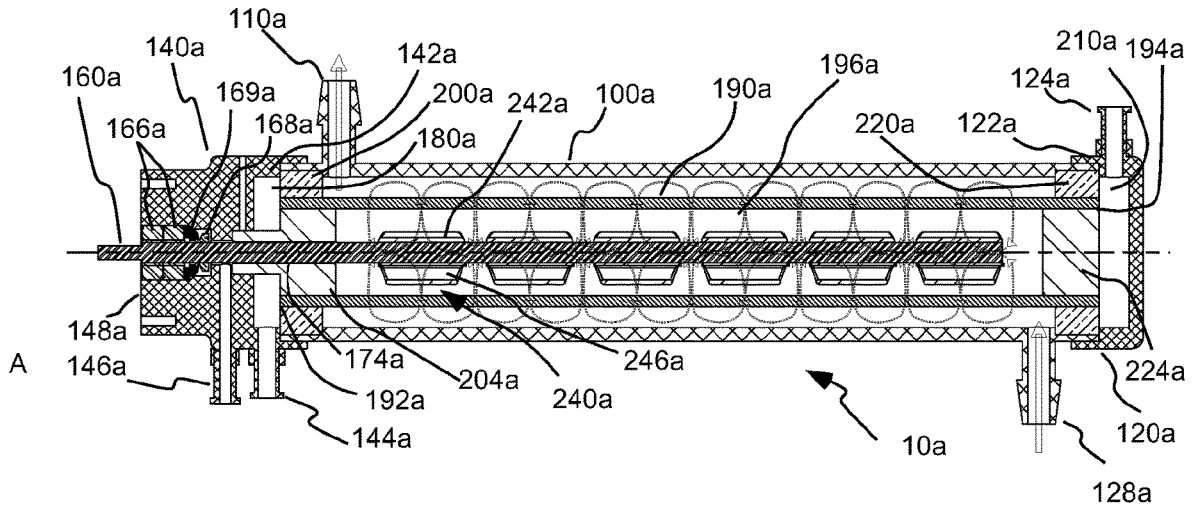


Fig. 5C

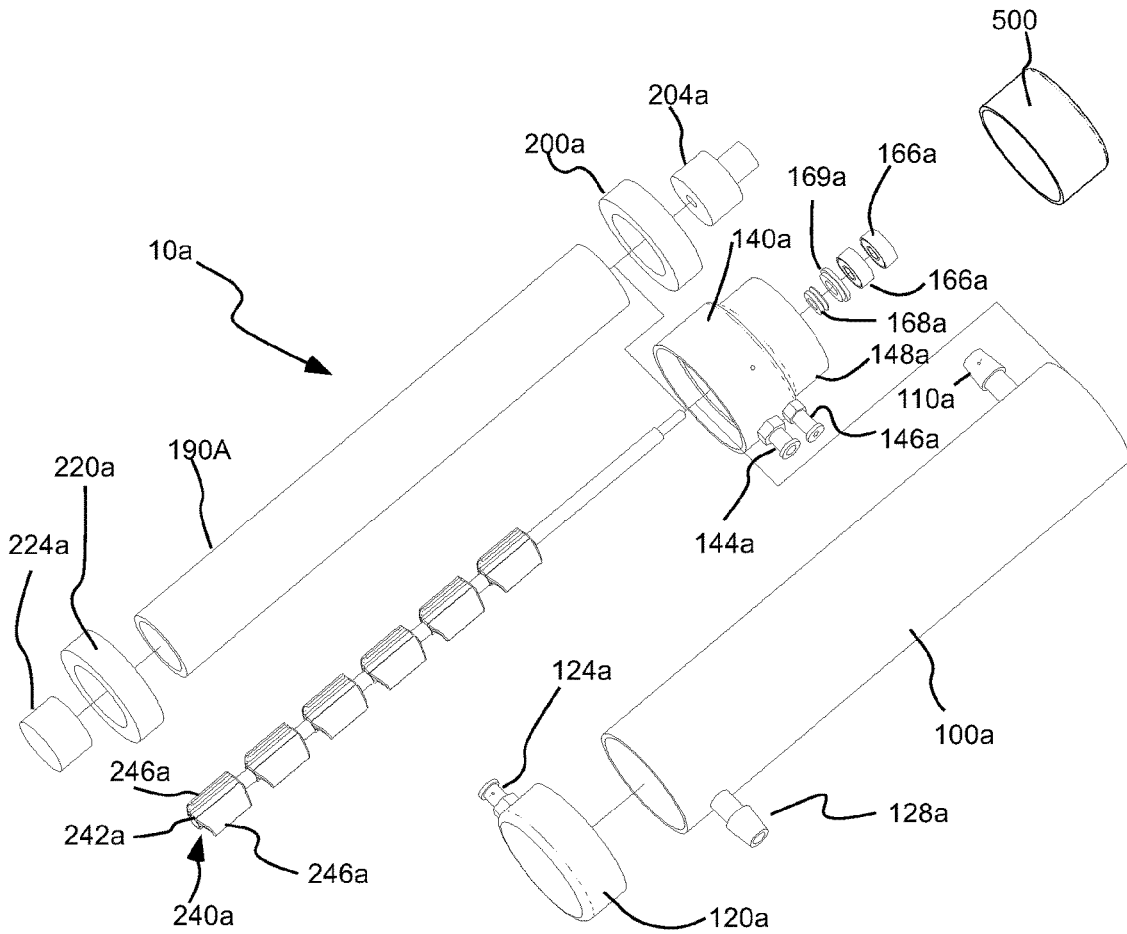


Fig. 5D

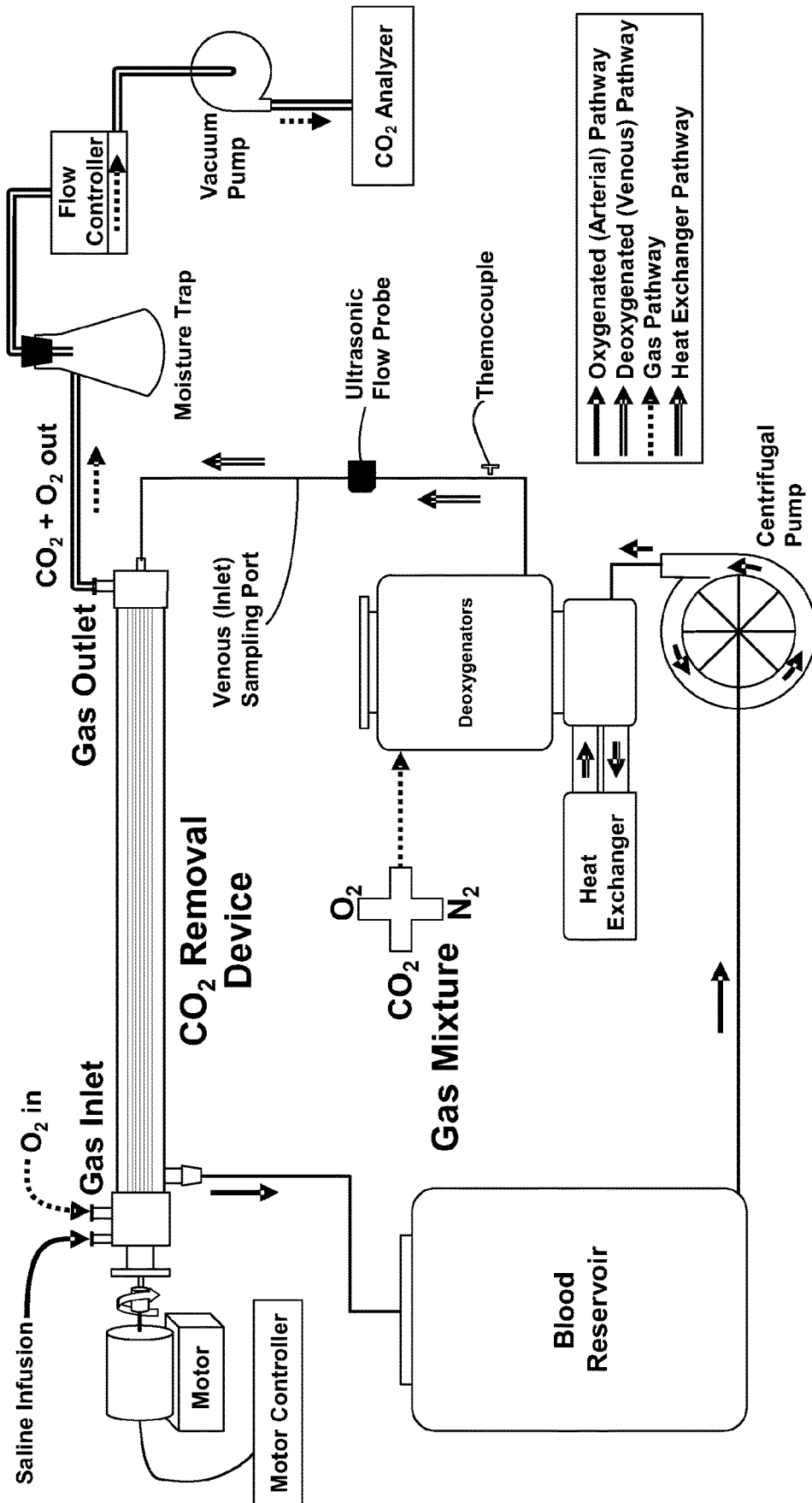


Fig. 6A

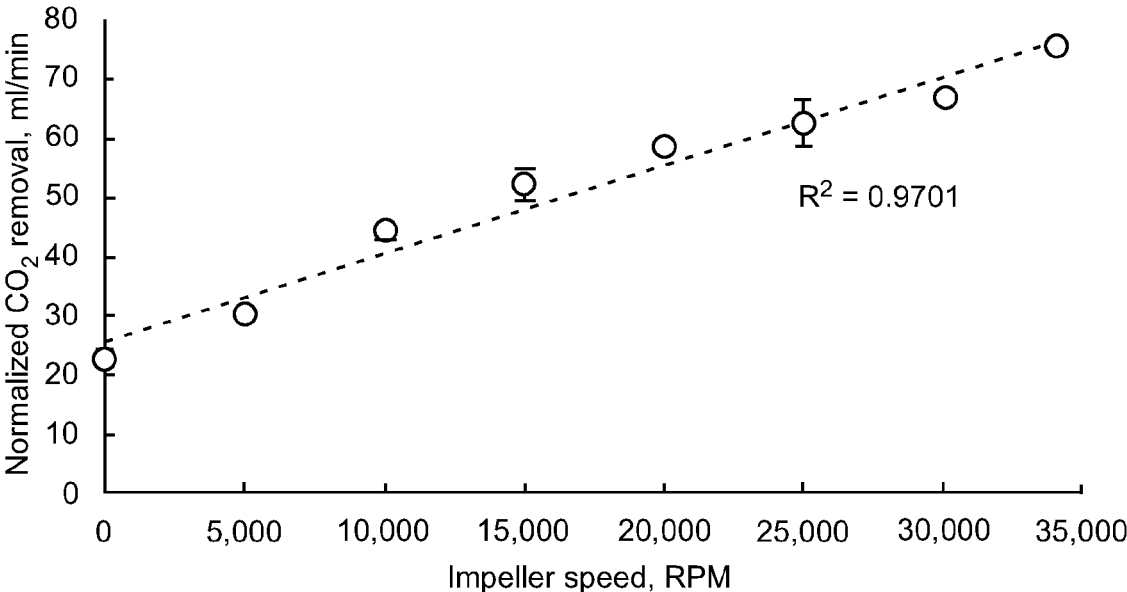


Fig. 6B

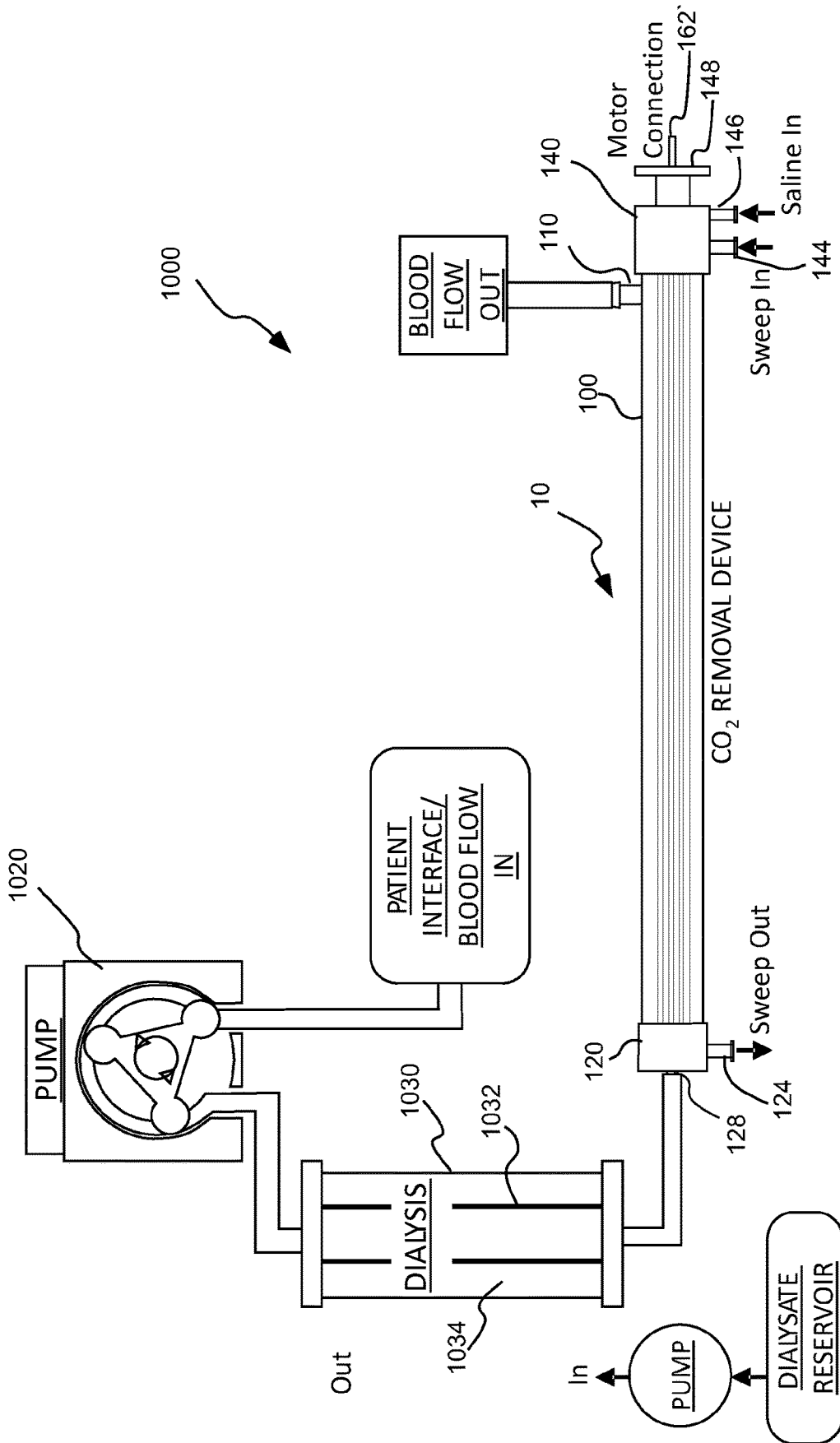


Fig. 7

EXTRACORPOREAL GAS EXCHANGE DEVICES, SYSTEMS AND METHODS

CROSS-REFERENCE TO RELATED APPLICATIONS

This application claims benefit of U.S. Provisional Patent Application Ser. No. 62/105,467, filed Jan. 20, 2015, the disclosure of which is incorporated herein by reference.

GOVERNMENTAL INTEREST

This invention was made with government support under grants numbers HL070051 and HL117637 awarded by the National Institutes of Health. The government has certain rights in this invention.

BACKGROUND

The following information is provided to assist the reader in understanding technologies disclosed below and the environment in which such technologies may typically be used. The terms used herein are not intended to be limited to any particular narrow interpretation unless clearly stated otherwise in this document. References set forth herein may facilitate understanding of the technologies or the background thereof. The disclosure of all references cited herein are incorporated by reference.

Lung disease is one of the major healthcare problems present in the United States today. Respiratory failure is a syndrome in which the respiratory system fails in one or both of its gas exchange functions: oxygenation and carbon dioxide elimination. In practice, respiratory failure may be classified as either hypoxemic or hypercapnic. Hypoxemic respiratory failure, which is sometime referred to as type I respiratory failure, is characterized by an arterial oxygen tension (Pa O₂) lower than 60 mm Hg with a normal or low arterial carbon dioxide (CO₂) tension (Pa CO₂). Hypoxemic respiratory failure is the most common form of respiratory failure. It may be associated with nearly all acute diseases of the lung. Hypercapnic respiratory failure, which is sometimes referred to as type II respiratory failure, is characterized by a PaCO₂ higher than 50 mm Hg. Hypoxemia is also common in patients with hypercapnic respiratory failure who are breathing room air. Hypercapnic respiratory failure is, for example, commonly associated with severe airway disorders such as asthma, chronic obstructive pulmonary disease or COPD, and acute respiratory distress syndrome or ARDS.

Extracorporeal support systems are under development to effect carbon dioxide removal. However, clinical adaptation of developing extracorporeal carbon dioxide removal systems (ECCO₂R) for management of hypercapnic respiratory failure has been hindered by the high blood flow rates necessary to provide adequate support. The high blood flow rates require a larger cannula inserted into the patient and increases the invasiveness of the procedure as a result.

SUMMARY

In one aspect, an extracorporeal gas exchange device includes a housing, a rigid shaft rotatable within the housing, at least one agitation mechanism positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the hollow fibers. The plurality of hollow fibers are positioned radially outward from the

agitation mechanism. The rotational speed of the rigid shaft may, for example, be adjustable independent of the flow rate of fluid through the housing.

The extracorporeal gas exchange device may, for example, include a sweep gas inlet in fluid connection with an inlet end of the plurality of hollow gas fibers, a sweep gas outlet in fluid connection with an outlet end of the plurality of hollow gas fibers, a fluid inlet, and a fluid outlet. The fluid inlet and the fluid outlet are isolated from fluid connection with the sweep gas inlet and the sweep gas outlet.

In a number of embodiments, the extracorporeal gas exchange includes a plurality of agitation mechanisms (for example, 3 to 30 or 3 to 15 agitation mechanism) positioned on the rigid shaft. The plurality of agitation mechanisms may for example, include a plurality of impellers in spaced positions on the rigid shaft. The plurality of impellers (or other agitation mechanisms) may, for example, be generally evenly spaced on the rigid shaft. Each of the plurality of impellers may, for example, include a plurality of vanes extending along the length thereof and extending radially outward therefrom. In a number of embodiment, each of the vanes are curved. The agitation mechanism(s) may, for example, be adjacent the plurality of hollow gas permeable fibers without an intervening component.

The fluid inlet may, for example, be adapted to be placed in fluid connection with a patient and the fluid may be blood (or a blood-derived fluid; blood and blood-derived fluids are referred to herein collectively as blood). A sweep gas placed in fluid connection with the sweep gas inlet may, for example, be adapted to remove carbon dioxide from the blood. The sweep gas may, for example, include oxygen.

In a number of embodiments hereof, extracorporeal gas exchange devices hereof are adapted to have a flow rate of blood through the housing in the range of approximately 200 to approximately 500 or approximately 200 to approximately 400 mL/min.

In another aspect an extracorporeal gas exchange device includes a housing, a rigid shaft rotatable within the housing, at least one agitation mechanism positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the hollow fibers, wherein the plurality of hollow fibers are positioned radially outward from the agitation mechanism. The agitation mechanism may, for example, be positioned adjacent the plurality of hollow gas permeable fibers without an intervening component. The rotational speed of the rigid shaft may, for example, be adjustable independent of the flow rate of fluid through the housing.

In a further aspect, an extracorporeal gas exchange device includes a housing, a shaft rotatable within the housing, at least one agitation mechanism positioned on the shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the hollow fibers. The plurality of hollow fibers are positioned radially outward from the agitation mechanism. In a number of embodiment, the plurality of hollow fibers are oriented generally parallel to the orientation of the rigid shaft (that is, within 5, 2 degrees or 1 degree of parallel). The rotational speed of the shaft is adjustable independent of the flow rate of fluid through the housing.

In another aspect, a system includes at least one pump device adapted to effect fluid flow, at least one dialysis system in fluid connection with the at least one pump device; and at least one extracorporeal gas exchange device hereof in fluid connection with the at least one pump device and the at least one dialysis system.

The present devices, systems, methods and compositions, along with the attributes and attendant advantages thereof, will best be appreciated and understood in view of the following detailed description taken in conjunction with the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1A illustrates a side view of an embodiment of an extracorporeal CO₂ removal device hereof.

FIG. 1B illustrates a front view of the device of FIG. 1A.

FIG. 1C illustrates a perspective view of the device of FIG. 1A.

FIG. 1D illustrates a top view of the device of FIG. 1A.

FIG. 2A illustrates a cross-sectional view of the device of FIG. 1A along plane F-F.

FIG. 2B illustrates a cross-sectional view of the device of FIG. 1A showing idealized flow lines of blood flow through the device including active mixing induced by impellers.

FIG. 2C illustrates a cross-sectional view of the device of FIG. 1A showing idealized flow lines of bulk flow blood of blood through the device without active mixing induced by impellers.

FIG. 2D illustrates a cross-sectional view of the device of FIG. 1A along plane A-A.

FIG. 2E illustrates a cross-sectional view of the device of FIG. 1A showing bulk flow of sweep gas through the device.

FIG. 3A illustrates a perspective, exploded or disassembled view of the device of FIG. 1A.

FIG. 3B illustrates an enlarged perspective, exploded or disassembled view a portion of the device of FIG. 1A including a second housing end member thereof.

FIG. 3C illustrates a hidden line view of the second housing end member of the device of FIG. 1A.

FIG. 4A illustrates a top view of an embodiment of an impeller or rotor for use in the device of FIG. 1A.

FIG. 4B illustrates a perspective view of the impeller of FIG. 4A.

FIG. 4C illustrates a side view of the impeller of FIG. 4A.

FIG. 4D illustrates a front view of the impeller of FIG. 4A.

FIG. 5A illustrates a side view of another embodiment of an extracorporeal CO₂ removal device hereof.

FIG. 5B illustrates a cross-sectional view of the device of FIG. 5A along plane A-A as defined in FIG. 5A, showing idealized flow lines of gas through the device.

FIG. 5C illustrates a cross-sectional view of the device of FIG. 5A showing idealized flow lines of blood flow through the device and from active mixing induced by impellers.

FIG. 5D illustrates a perspective, exploded or disassembled view of the device of FIG. 5A.

FIG. 6A illustrates an in vitro blood flow circuit used for gas exchange testing of a number of devices hereof.

FIG. 6B illustrates a graph of measured CO₂ removal rate as a function of impeller rotation rate in studies of a CO₂ removal devices hereof using the in vitro blood flow circuit of FIG. 6A at a flow rate of 250 mL/min.

FIG. 7 illustrated a schematic representation CO₂ removal device hereof in fluid connection with a renal dialysis device.

DETAILED DESCRIPTION

It will be readily understood that the components of the embodiments, as generally described and illustrated in the figures herein, may be arranged and designed in a wide variety of different configurations in addition to the

described representative embodiments. Thus, the following more detailed description of the representative embodiments, as illustrated in the figures, is not intended to limit the scope of the embodiments, as claimed, but is merely illustrative of representative embodiments.

Reference throughout this specification to “one embodiment” or “an embodiment” (or the like) means that a particular feature, structure, or characteristic described in connection with the embodiment is included in at least one embodiment. Thus, the appearance of the phrases “in one embodiment” or “in an embodiment” or the like in various places throughout this specification are not necessarily all referring to the same embodiment.

Furthermore, described features, structures, or characteristics may be combined in any suitable manner in one or more embodiments. In the following description, numerous specific details are provided to give a thorough understanding of embodiments. One skilled in the relevant art will recognize, however, that the various embodiments can be practiced without one or more of the specific details, or with other methods, components, materials, et cetera. In other instances, well known structures, materials, or operations are not shown or described in detail to avoid obfuscation.

As used herein and in the appended claims, the singular forms “a,” “an,” and “the” include plural references unless the context clearly dictates otherwise. Thus, for example, reference to “an impeller” includes a plurality of such impellers and equivalents thereof known to those skilled in the art, and so forth, and reference to “the impeller” is a reference to one or more such impellers and equivalents thereof known to those skilled in the art, and so forth. Recitation of ranges of values herein are merely intended to serve as a shorthand method of referring individually to each separate value falling within the range. Unless otherwise indicated herein, and each separate value, as well as intermediate ranges, are incorporated into the specification as if individually recited herein. All methods described herein can be performed in any suitable order unless otherwise indicated herein or otherwise clearly contraindicated by the text. Unless the context clearly dictates otherwise, use of the terms “approximately,” “about” and similar terms refer to a value within 10% of the stated value.

In a number of embodiments of devices, cartridges, or systems hereof, an array of rotating impellers or rotors facilitates highly efficient gas exchange in, for example, an artificial lung device and allows operation at significantly lower blood flow rates than existing technology. The blood flow rates used in systems hereof may, for example, approximate or match those used in hemodialysis (for example, approximately 250 ml/min). The ability of devices, systems and methods hereof to achieve clinically relevant levels of CO₂ removal at flow rates typical used in renal hemodialysis is unprecedented. The relatively low flow rates through devices hereof provide, for example, for simplified cannulation strategies (for example, the use of smaller and/or less invasive cannula) and potential integration of devices and systems hereof into existing renal dialysis circuits.

In a number of embodiments hereof, blood pumped from the body to an extracorporeal device, system, or cartridge hereof is recirculated in and out of a fiber membrane bundle (a plurality of hollow fiber membranes) as a result of, for example, rotor- or impeller-generated circulation/pumping before exiting the device and returning to the body. Intimate interactions between rotating impellers of the devices hereof and the surrounding hollow fiber membranes (HFMs) (see, for example, FIG. 2B) generates an active mixing effect. As used herein, the term active mixing refers generally to

generating flow past the surface of hollow fiber membranes, preferably perpendicular thereto, at a higher velocity than would exist if the velocity arose only from the bulk blood flow through the device. Because the size of the diffusional boundary layer scales roughly as the inverse square root of velocity, the diffusional boundary layer is reduced as a result. Active mixing provides a method of enhancing gas transfer capabilities in, for example, hollow fiber membrane (HFM) based artificial lung devices hereof. High fluid velocities at the gas exchange surfaces generated by impeller pumping, which diminishes the thickness of the diffusive boundary layer at the fiber surface. Convective delivery and removal of blood gases from the membrane surface is improved, enhancing gas transfer efficiency. In a number of embodiments hereof, impellers which are used to create active mixing pump fluid only in the radial direction, generally perpendicular to the orientation of the HFM and to bulk fluid flow through the device, and therefore have no or substantially no effect on circuit blood flow rate. One or more pumping or pressurizing mechanisms or devices, separately controllable from the impellers, is/are used to effect bulk fluid flow through the device. As a result, the level of active mixing (and associated gas exchange) is controllable independently of circuit flow rate (or control of the pressurizing mechanism(s) which causes flow through the circuit or system).

As described above, rotating impellers within a HFM bundle generate an active mixing effect that enhances gas exchange capabilities, allowing operation of the device at lower blood flow rates than comparably sized devices. The impellers hereof pump (that is, impart motion to the fluid flow) in the radial direction (generally perpendicular to bulk flow) without a significant effect on bulk flow, and therefore do not contribute to or inhibit blood flow through the device, allowing impeller speed (i.e. level of active mixing and associated improvement in gas exchange) to be controlled independently of device blood flow. This feature improves the versatility of the devices and system hereof, allowing such devices or systems to be operated in a stand-alone circuit or spliced/integrated into an existing blood circuit (for example, a renal hemodialysis flow circuit).

In a number of embodiments, extracorporeal gas exchange (for example, CO₂ removal) devices hereof create active mixing through rotation of impellers on a rigid driveshaft. The driveshaft may, for example, be suspended concentrically within a hollow fiber membrane bundle. As described above, the impellers may, for example, generate flow substantially only or only in the radial direction, thereby contributing to substantially no or no change in total flow rate through the device (that is, bulk flow rate from the inlet to the outlet of the device). In addition to the ability of operation at blood flow rates similar to renal hemodialysis, the devices hereof may be spliced directly into existing dialysis circuitry with minimal disturbance in that system's normal operation, or be used with dialysis pump circuits.

The rigid impeller driveshaft of the devices hereof assists in preventing impeller blade contact with any surrounding device components. The rigid driveshaft substantially reduces or eliminates the potential for blood cell trauma resulting from impeller vane/blade contact with static surfaces (thereby, significantly improving hemocompatibility). Compared to flexible shaft technology, a rigid shaft provides for significantly improved impeller durability because the impeller surface wear is greatly reduced. Therefore, in a number of embodiments of devices hereof, protective coils (or other support member) surrounding the impeller may be omitted. In that regard, the agitation mechanism used in the

devices hereof (for example, impeller vanes) may be directly adjacent to the hollow fiber bundle without intervening components. A coil or support may be used adjacent the inner wall of the hollow fiber bundle to assist in maintaining a well-formed annular configuration of the hollow fiber bundle. However, such a coil or support need only be sufficiently robust to support the annular fiber bundle and need not operate to protect the fiber bundle from contact with the agitation mechanisms supported on the rotating rigid shaft.

An embodiment of a device **10** hereof and components thereof are illustrated in FIGS. **1A** through **4D**. As, for example, illustrated in FIG. **1A** through **3A**, device **10** includes a housing having an extending housing section or body **100** which may, for example, be formed from a polymeric material or materials such as an acrylic polymer. In a number of embodiments, a blood outlet port **110** is formed in housing body **100**. In the illustrated embodiment, the housing further includes a first end member **120** connected to a first end of housing body **100**. First housing end member **120** includes a sweep gas outlet port **124** and a blood inlet port **128**. Sweep gas outlet port **124** may, for example, include a luer lock connector or luer connector as known in the medical arts. In the illustrated embodiment, a second housing end member **140** is connected to a second end of housing body **100**. In the illustrated embodiment, the housing further includes a second housing end member **140**, which includes a sweep gas inlet port **144** and a saline port **146**. Each of sweep gas inlet port **144** and saline port **146** may, for example, include a luer lock connector or luer connector as known in the medical arts.

In a number of embodiments, second housing end member **140** further includes a seating or coupling **148** to interface with a motor to provide rotation to a drive shaft **160** (as illustrated, for example, in FIG. **2D**). Drive shaft **160** may, for example, be formed to be rigid or inflexible (under the conditions of operation of device **10**) from, for example, stainless steel. Drive shaft **160** is sufficiently rigid such that, during operation at full speed, the drive shaft **160** does not deflect from the axis of rotation thereof (as a result of centrifugal forces associated with the weight of drives shaft **160** plus all impellers **240** fixed thereto) to a degree to come in contact with any surrounding components of device **10**, specifically a surrounding fiber membrane bundle **190**. In a number of embodiments, the clearance between impeller(s) **240** and the fiber membranes of hollow fiber membrane bundle **190** is in the range of approximately 0.5 to 3 mm or 0.5 to 2 mm. In the illustrated embodiment, a driveshaft cap **162** surrounding a first end of driveshaft **160** was used to form an operative connection with the motor. Bearings **166**, such as ball bearings, may, for example, be seated within coupling **148**. In the illustrated embodiment, a seal **168** is also provided in operative connection with drive shaft **160**. Saline may be introduced via saline inlet or infusion port **146** to provide continuous flushing of driveshaft **160** and the blood side face of seal **168**. A sheath **170** may, for example, be provided around driveshaft **160** to seal the saline infusion pathway from a sweep gas inlet manifold chamber **180**. In a number of embodiments, sheath **170** was formed from a lubricious polymer such as polytetrafluoroethylene or PTFE.

Inlet manifold chamber **180** is in fluid connection with sweep gas inlet **144** and an inlet end **192** of an annular shaped hollow fiber bundle **190**. In a number of studied embodiments of system **10**, a polypropylene hollow fiber bundle **190** such as Celgard® x30-240 available from Membrana GmbH of Wuppertal, Germany was used. In the illustrated embodiment, inlet manifold chamber **180** was

formed in second housing end member **140** which was connected to and formed a sealing engagement with housing body **100** via a seating **142** formed therein. Inlet manifold chamber **180** was sealed from the blood flow path of system **10** via a polyurethane fiber potting **200** extending between an inner wall of housing body **100** and an outer surface of hollow fiber bundle **190** and a fiber potting insert **204** extending between an inner surface of hollow fiber bundle **190** and sheath **170**. In a number of embodiments, fiber potting insert **204** was machined from stainless steel.

An outlet manifold chamber **210** is in fluid connection with sweep gas outlet **124** and an outlet end **194** of annular-shaped hollow fiber bundle **190**. In the illustrated embodiment, outlet manifold chamber **210** was formed in first housing end member **120**, which was connected to and formed a sealing engagement with housing body **100** via a seating **122** formed therein. Outlet manifold chamber **210** was sealed from the blood flow path of system **10** via a polyurethane fiber potting **220** extending between an inner wall of housing body **100** and an outer surface of hollow fiber bundle **190** and a blood inlet tube **129** which was sealed against an inner surface of hollow fiber bundle **190**. In a number of embodiments, blood inlet tube **129** was formed from stainless steel.

A plurality of impellers **240** (23 in one studied embodiment of system **10**) were positioned along the length of rigid drive shaft **160** within interior volume **196** formed by the annular hollow fiber bundle **190**. As, for example, illustrated in FIGS. 4A through 4D, impellers **240** included an extending sleeve **242** through which drive shaft **160** passes and a plurality of vanes **246** extending radially outward from sleeve **242**. In a number of studied embodiments, impellers **240** included three radially extending vanes. In a number of embodiments, the vanes extended longitudinally along the length of sleeve **242** and extended perpendicular to the axis of sleeve **242**. In a number of embodiments, impellers **240** were generally evenly spaced along the length of rigid drive shaft **160**.

FIGS. 2A through 2E illustrate cross-sectional views of device **10** via which the flow paths of sweep gas and blood through device **10** are discussed. As, for example, illustrated in FIGS. 2B and 2C, blood enters through a blood inlet port **128** into interior volume **196** formed by annular hollow fiber bundle **190**. In other embodiments, blood may, for example enter through the outer gap between annular hollow fiber bundle **190** and housing body **100**. Rotation of drive shaft **160**, and thereby impellers **240**, causes the majority of blood flow to enter a recirculating path through hollow fiber bundle **190** (as represented by dashed arrows in FIG. 2B), starting at the center of each impeller **240**. After blood passes through hollow fiber bundle **190** toward the interior wall of housing body **100**, the blood flow re-traverses hollow fiber bundle **190** in the opposite direction (that is, radially inward) toward the gap between impellers **240**.

The bulk blood flow (illustrated in FIG. 2C) pushes flow gradually toward the blood outlet port **110** at a rate equal to the circuit blood flow rate of the system into which device **10** is incorporated. Once again, blood enters through blood inlet port **128**. Most blood recirculates through hollow fiber bundle **190** as illustrated in FIG. 2B, but (as a result of the circuit flow rate), blood moves in a general direction from blood inlet port **128** to blood outlet port **110**. Blood can move from inlet to outlet in the impeller region at center, in the fiber bundle (interstitial space between fibers), or the outermost region between the bundle and the housing wall.

As, for example, illustrated in FIG. 2E, sweep gas (represented by dashed arrows) enters sweep gas inlet port **144**

and travels to inlet manifold chamber **180** before entering inlet ends **192** of the lumens of hollow fiber bundle lumens **190**. As described above, the fibers of hollow fiber bundle **190** are potted concentrically around fiber potting insert **204**, sealing the gas pathway from driveshaft **160**, which is sheathed inside driveshaft sheath or tubing **170**. Sweep gas exits the fiber lumens via outlet ends **194** thereof into outlet manifold chamber **210**. Sweep gas exits device **10** through sweep gas outlet port **124**.

In the illustrated embodiment, a flushing fluid such as saline (represented by solid arrows in FIG. 2E) is infused through flushing fluid/saline inlet port **146** and travels radially inward toward driveshaft **160**. Seal **168** on driveshaft **160** ensures that infused flushing fluid/saline exits through the driveshaft sheath/tubing **170** toward the blood flow path of device **10**, rather than leaking through bearings **166**.

Another embodiment of a device **10a** hereof and components thereof are illustrated in FIGS. 5A through 5D. Similar to device **10**, device **10a** includes a housing having an extending housing section or body **100a** which may, for example, be formed from a polymeric material. A blood outlet port **110a** is formed in housing body **100a**. A first end member **120a** is connected to a first end of housing body **100a**. First end member **120a** includes a sweep gas outlet port **124a**. Unlike device **10**, in device **10a**, a blood inlet port **128a** is formed in housing body **100a** rather than first end member **120a**. A second housing end member **140a** is connected to a second end of housing body **100a**. Second housing end member **140a** includes a sweep gas inlet port **144a** and a saline port **146a**.

Similar to device **10**, second housing end member **140a** includes a seating or coupling **148a** to interface with a drive system which may include a motor **500** (illustrated schematically in FIG. 5D) to provide rotation to a drive shaft **160a**, which may, for example, be formed to be rigid or inflexible (under the conditions of operation of device **10a**) from, for example, stainless steel. As described above, drive shaft **160a** is sufficiently rigid such that, during operation at full speed, drive shaft **160a** does not deflect from the axis of rotation thereof (as a result of centrifugal forces associated with the weight of drive shaft **160a** plus all impellers **240a** fixed thereto) to a degree to come in contact with any surrounding components of device **10a** (and, specifically, surrounding fiber membrane bundle **190a**). As further described above, the clearance between impeller(s) **240a** and the fiber membranes of hollow fiber membrane bundle **190a** may, for example, be in the range of approximately 0.5 to 3 mm or 0.5 to 2 mm. In the illustrated embodiment of FIGS. 5A through 5D, driveshaft **160a** operatively connects directly with the motor without an intervening drive shaft cap. Bearings **166a**, such as ball bearings, may, for example, be seated within coupling **148a**. A seal **168a** and a seal backing **169a** (seated between bearings **166a** and seal **168a**) are provided in operative connection with drive shaft **160a**. Seal backing **169a** assist in providing a seal between coupling **148a** (for, for example, from an acrylic polymer) and shaft **160a**. Saline may be introduced via saline inlet or infusion port **146a** to provide continuous flushing of drive shaft **160a** and the blood side face of seal **168a**. In the embodiment of device **10a**, a sheath similar to sheath **170** of device **10** was not used. Instead, a potting insert component **204a** was machined using the material of sheath **170** (PTFE) or a similar lubricious polymeric or other lubricious sealing material. This embodiment reduces the number of parts in the assembly and improves precision in manufacturing. Sealing contact between driveshaft **160s** and potting insert component **204a** seals the saline infusion pathway from a

sweep gas inlet manifold chamber **180a**. In alternative embodiments, the rigid shafts and the operatively connected agitation mechanisms may be driven by a drive systems other than motor. For example, rigid shafts and agitation mechanisms hereof may be magnetically suspended and driven, thereby obviating the need for any bearings and seals

As described above, inlet manifold chamber **180a** is in fluid connection with sweep gas inlet **144a** and an inlet end **192a** of an annular shaped hollow fiber bundle **190a**. As also described above, a polypropylene hollow fiber bundle **190a** such as Celgard® x30-240 available from Membrana GmbH of Wuppertal, Germany may be used. Inlet manifold chamber **180a** was formed in second housing end member **140a** which was connected to and formed a sealing engagement with housing body **100a** via a seating **142a** formed therein. Inlet manifold chamber **180a** was sealed from the blood flow path of system **10a** via a polyurethane fiber potting **200a** extending between an inner wall of housing body **100a** and an outer surface of hollow fiber bundle **190a** and fiber potting insert **204a** (machined from PTFE as described above) extending between an inner surface of hollow fiber bundle **190a** and drive shaft **160a**.

An outlet manifold chamber **210a** was provided in fluid connection with sweep gas outlet **124a** and an outlet end **194a** of annular-shaped hollow fiber bundle **190a**. Outlet manifold chamber **210a** was formed in first housing end member **120a**, which was connected to and formed a sealing engagement with housing body **100a** via a seating **122a** formed therein. Outlet manifold chamber **210a** was sealed from the blood flow path of system **10a** via a polyurethane fiber potting **220a** extending between an inner wall of housing body **100a** and an outer surface of hollow fiber bundle **190a**. A solid plug **224a** (formed, for example from stainless steel) formed a seal with the inner surface of annular fiber bundle **190a** and sealed the blood compartment from gas outlet manifold chamber **210a**.

As described above in connection with device **10**, device **10a** includes a plurality of impellers **240a** (6 in the illustrated embodiment) positioned along the length of rigid drive shaft **160a** within interior volume **196a** formed by the annular hollow fiber bundle **190a**. In a number of embodiments, devices hereof include from 3 to 30 or 3 to 15 impellers. Without limitation to any mechanism, it is desirable to maximize the amount of flow at an angle to (for example, perpendicular to) the direction of bulk flow of blood (that is, in the direction of axis A as illustrated in FIG. 5C) to maximize active mixing. This result may be accomplished by providing a plurality of impellers at spaced positions along shaft **160a**.

Impellers **240a** include an extending sleeve **242a** through which drive shaft **160a** passes and a plurality of vanes **246a** extending radially outward from sleeve **242**. In the embodiment of device **10a**, vanes **246a** extended longitudinally along sleeve **242a** and were curved as illustrated in FIG. 5D. The curve in vanes **246a** acted on the fluid/blood during rotation of vanes **246a** around shaft **160a** to reorient the fluid flow (that is, pump the fluid/blood) with velocity vectors directed radially outward. In a number of embodiments, impellers **240a** were generally evenly spaced along the length of rigid drive shaft **160a**.

FIGS. 2A through 2E illustrate cross-sectional views of device **10** via which the flow paths of sweep gas and blood through device **10** are discussed. As, for example, illustrated in FIGS. 2B and 2C, blood enters through an blood inlet port **128** into interior volume **196** formed by annular hollow fiber bundle **190**. In other embodiments, blood may, for example enter through the outer gap between annular hollow fiber

bundle **190** and housing body **100**. Rotation of drive shaft **160**, and thereby impellers **240**, causes the majority of blood flow to enter a recirculating path through hollow fiber bundle **190** (as represented by dashed arrows in FIG. 2B), starting at the center of each impeller **240**. After blood passes through hollow fiber bundle **190** toward the interior wall of housing body **100**, the blood flow re-traverses hollow fiber bundle **190** in the opposite direction (that is, radially inward) toward the gap between impellers **240**.

Similar to device **10a**, the bulk blood flow (illustrated in FIG. 5C) pushes flow gradually toward the blood outlet port **110** at a rate equal to the circuit blood flow rate of the system into which device **10a** is incorporated. Blood enters through blood inlet port **128a** and recirculates through hollow fiber bundle **190a**. As a result of the circuit flow rate, blood moves in a general direction from blood inlet port **128a** to blood outlet port **110a**. Sweep gas (represented by dashed arrows in FIG. 5B) enters sweep gas inlet port **144a** and travels to inlet manifold chamber **180a** to inlet ends **192a** of the lumens of hollow fiber bundle lumens **190a**. Sweep gas exits the fiber lumens via outlet ends **194a** into outlet manifold chamber **210a** and exits device **10a** through sweep gas outlet port **124a**.

In a design configuration such as device **10a** wherein the blood flow is introduced within the housing into a space of volume between fiber bundle **190a** and an inner wall of housing body **100a**, gas exchange was found to approximately equivalent to gas exchange in a similarly device such as device **10**, wherein blood is introduced within a volume defined by the annular fiber bundle **190**, at higher rotation speeds, but slightly lower at very low rotation speeds. Relocating blood inlet port **128a** to housing body **100a** in device **10a** facilitate manufacture as compared to blood inlet port **128** of device **10**. At higher rotation speeds typically used in devices hereof, the agitation generated by impellers **240a** is significant enough that the location of blood inlet port **128a** minimally affects resulting performance as compared to blood inlet port **128** of device **10**.

In a number of studies of devices and systems hereof, gas exchange performance was evaluated in an in vitro flow loop, system or circuit illustrated in FIG. 6A. In a number of studies, the inlet blood gas tensions were maintained at $p\text{CO}_2=45\pm 5$ mmHg and $s\text{O}_2=65\pm 5\%$ using commercially available oxygenators (Deoxygenators) and an $\text{O}_2/\text{CO}_2/\text{N}_2$ gas mixture. Circulating blood was maintained constant at 250 ml/min with a commercially available blood pump and tubing clamp, as verified by an on-line ultrasonic flow probe. Pure O_2 sweep gas was pulled through the fibers of the hollow fiber bundle (counter-current to the blood flow) at a flow rate of 6.2 L/min. The percent CO_2 in the exiting sweep gas was measured and used to calculate a CO_2 removal rate. Evidence of active mixing was demonstrated by experimental comparison of the achieved CO_2 removal rate versus impeller rotation speed as set forth in FIG. 5B. In a number of studied embodiments, the impeller rotation speed was in the range of 500-35,000 RPM, approximately 10,000-35,000 RPM, or approximately 30,000-34,000 RPM. In a number of other embodiments, the impeller rotation speed may, for example, be in the range of approximately 500-10,000 RPM, approximately, 2000-8000 RPM, or approximately 5500-6500 RPM. Even at relatively high impeller rotational speeds, the impellers contribute little (for example, less than 5%, less than 2% or less than 1%) or zero to the bulk flow rate through the devices and systems hereof, providing for control of active mixing that is substantially independent or independent of control of bulk flow. At a blood flow rate of 250 ml/min, measured CO_2 removal increased 229% at the

highest rotation speed. Inlet pCO₂ was verified to equal 45±5 mmHg prior to each measurement in the studies of FIG. 6B. Removal rates were normalized to an inlet pCO₂ of 45 mm Hg.

In the studies hereof, device **10** measured approximately 35-40 cm in length (wherein, the blood-primed portion was approximately 30 cm in length. Device **10a** was approximately 20 to 25 cm in length and the blood-primed portion was approximately 17 cm in length. Device **10** had a diameter in the range of 2.5 to 4 cm (wherein the blood primed portion had a diameter of approximately 2.22 cm), while device **10a** had a diameter in the range of 4.45 cm (wherein the blood primed portion had a diameter of approximately 4.13 cm). The priming volume of the studied device **10** was approximately 100 mL, while the priming volume of device **10a** was approximately 150 mL. The number of fiber membranes (0.03 cm diameter, polypropylene fibers) of studied device **10a** was 750, and 23 impellers were fixed on drive shaft **160**. The number of fiber membranes (0.03 cm diameter, polypropylene fibers) of studied device **10a** was 1331, and 6 impellers were fixed on drive shaft **160a**. Alternative device dimensions with various aspect ratios (length versus fiber bundle diameter) may yield similar gas exchange performance. In a number of embodiments, devices hereof may, for example, have blood primed regions in the range of 10-30 cm in length, have diameters in the range of 2.22-5 cm, and/or have priming volumes in the range of 75-150 mL. In a number of embodiments hereof, devices hereof may, for example, include 750-4000 hollow fiber membranes, and/or include 1-30, 3-30 or 3-15 impellers **240** on drive shaft **160**.

FIG. 7 illustrates a representative example of a system **1000** for reducing the concentration of CO₂ in blood plasma and effecting renal dialysis. Oxygenation of blood plasma can be effected using a sweep gas including (or consisting of) oxygen. In the case of CO₂ removal, the sweep gas supply includes a lower concentration of CO₂ than present in the blood plasma, giving rise to a concentration gradient/driving force across the membranes of the hollow fibers. Typically, the sweep gas supply will include no or substantially no CO₂ therein to maximize the driving force. System **1000** includes patient interfaces for removal of blood from the patient and return of filtered blood to the patient and system **10** for removal of CO₂/oxygenation as described above. One or more blood pumps **1020** as known in the art may be used for effecting flow of blood through system **1000** as required. A dialyzer or dialysis filter system **1030** may, for example, include a plurality of, semipermeable fibers **1032**. In dialysis filter system **1030**, the blood may, for example, flow through semi-permeable fibers **1032** and the dialysate flows in a volume **1034** around fibers **1032**. After passing through dialysis filter system **1030**, the filtered blood is returned to the patient. As described above, blood flow rates currently used in dialysis systems can be used in system **1000** (typically, approximately 250 mL/min). For example, the blood flow rate may be in the range of approximately 200 to approximately 500 mL/min, in the range of approximately 200 to 400 mL/min, or in the range of approximately 225 to 275 mL/min. The order of circuit components in the systems hereof can be varied. In system **1000**, of FIG. 6, for example, CO₂ removal device **10** receives flow after both pump **1020** and dialysis filter system **1030**. Device **10** could, for example, also be placed between pump **1020** and dialysis filter system **1030**. Typically, device **10** is placed after a pump in the systems hereof to provide a positive fluid pressure (to reduce or eliminate the risk of pulling gas out of the fibers into the blood pathway).

In a manner similar to that described in connection with oxygenators or lung assist systems in U.S. Pat. Nos. 7,763, 097 and 8,043,411, the disclosure of which are incorporated herein by reference, carbonic anhydrase or CA may be used on or in the vicinity of the fibers of hollow fiber membrane **190** of device **10** to drive or increase the removal of bicarbonate from blood. CA reversibly catalyzes hydration of CO₂ into carbonic acid, which then rapidly dissociates into bicarbonate ion. Immobilized CA may, for example, be used to facilitate diffusion toward a membrane including the immobilized enzyme. CA immobilized on or in the vicinity of the surface of the fibers of hollow fiber membrane **190** enables "facilitated diffusion" of CO₂ as bicarbonate towards the fibers of hollow fiber membrane **190** and enhances the removal rate of CO₂. Indeed, creating velocity streams across the orientation of the fiber membranes may increase the effectiveness of CA. In that regard, in passive test devices (without active mixing) the enhancement of gas exchange by carbonic anhydrase and blood has been found to be limited by the diffusional boundary layer and not by the amount or kinetics of the carbonic anhydrase.

The foregoing description and accompanying drawings set forth a number of representative embodiments at the present time. Various modifications, additions and alternative designs will, of course, become apparent to those skilled in the art in light of the foregoing teachings without departing from the scope hereof, which is indicated by the following claims rather than by the foregoing description. All changes and variations that fall within the meaning and range of equivalency of the claims are to be embraced within their scope.

What is claimed is:

1. An extracorporeal gas exchange device, comprising:
 - a housing, a rigid shaft rotatable within the housing, a plurality of agitation mechanisms positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the plurality of gas permeable hollow fibers, the plurality of hollow gas permeable fibers being positioned radially outward from the plurality of agitation mechanisms, the rotational speed of the rigid shaft being adjustable independent of the flow rate of fluid through the housing.
2. The extracorporeal gas exchange device of claim 1 wherein the housing comprises a sweep gas inlet in fluid connection with an inlet end of the plurality of hollow gas permeable fibers, a sweep gas outlet in fluid connection with an outlet end of the plurality of hollow gas permeable fibers, a fluid inlet, and a fluid outlet, the fluid inlet and the fluid outlet being isolated from fluid connection with the sweep gas inlet and the sweep gas outlet.
3. The extracorporeal gas exchange device of claim 2 comprising between 3 and 30 agitation mechanisms positioned on the rigid shaft.
4. The extracorporeal gas exchange device of claim 2 comprising between 3 and 15 agitation mechanisms positioned on the rigid shaft.
5. The extracorporeal gas exchange device of claim 3 wherein the plurality of agitation mechanisms comprise a plurality of impellers in spaced positions on the rigid shaft.
6. The extracorporeal gas exchange device of claim 5 wherein the plurality of impellers are generally evenly spaced on the rigid shaft.
7. The extracorporeal gas exchange device of claim 5 wherein each of the plurality of impellers comprise curved vanes.

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8. The extracorporeal gas exchange device of claim 6 wherein the fluid inlet is adapted to be placed in fluid connection with a patient and the fluid is blood.

9. The extracorporeal gas exchange device of claim 8 wherein a sweep gas placed in fluid connection with the sweep gas inlet is adapted to remove carbon dioxide from the blood.

10. The extracorporeal gas exchange device of claim 9 wherein the sweep gas comprises oxygen.

11. The extracorporeal gas exchange device of claim 10 wherein the flow rate of blood through the housing is in the range of 200 to 500 mL/min.

12. The extracorporeal gas exchange device of claim 10 wherein the flow rate of blood through the housing is in the range of 200 to 400 mL/min.

13. The extracorporeal gas exchange device of claim 1 wherein the plurality of agitation mechanisms are adjacent the plurality of hollow gas permeable fibers without an intervening component.

14. A system, comprising:
 at least one pump device adapted to effect fluid flow;
 at least one dialysis system in fluid connection with the at least one pump device; and
 at least one extracorporeal gas exchanged device in fluid connection with the at least one pump device and the at least one dialysis system, the extracorporeal gas exchange device comprising a housing, a rigid shaft rotatable within the housing, a plurality of agitation mechanisms positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the hollow fibers, the plurality of hollow gas permeable fibers being positioned radially

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outward from the plurality of agitation mechanisms, the rotational speed of the rigid shaft being adjustable independent of the rate of fluid flow through the housing.

15. An extracorporeal gas exchange device, comprising: a housing, a rigid shaft rotatable within the housing, a plurality of agitation mechanisms positioned on the rigid shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the plurality of hollow gas permeable fibers, the plurality of hollow gas permeable fibers being positioned radially outward from the plurality of agitation mechanisms.

16. The extracorporeal gas exchange device of claim 15 wherein the plurality of agitation mechanisms are adjacent the plurality of hollow gas permeable fibers without an intervening component.

17. The extracorporeal gas exchange device of claim 15 wherein the rotational speed of the rigid shaft is adjustable independent of the flow rate of fluid through the housing.

18. An extracorporeal gas exchange device, comprising: a housing, a shaft rotatable within the housing, a plurality of agitation mechanisms positioned on the shaft, and a plurality of hollow gas permeable fibers adapted to permit diffusion of gas between fluid flowing within the housing and an interior of the plurality of hollow gas permeable fibers, the plurality of hollow gas permeable fibers being positioned radially outward from the plurality of agitation mechanisms, the rotational speed of the shaft being adjustable independent of the flow rate of fluid through the housing.

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