US011635064B1

(12) **United States Patent**
Kartalov et al.

(10) **Patent No.: US 11,635,064 B1**
(45) **Date of Patent: Apr. 25, 2023**

(54) **MICROFLUIDIC-BASED ARTIFICIAL MUSCLES AND METHOD OF FORMATION**

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(*) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 153 days.

(21) Appl. No.: **17/338,393**

(22) Filed: **Jun. 3, 2021**

Related U.S. Application Data

(63) Continuation of application No. 16/442,092, filed on Jun. 14, 2019, now Pat. No. 11,060,511.
(Continued)

(51) **Int. Cl.**
A61F 2/08 (2006.01)
H02N 1/00 (2006.01)
(Continued)

(52) **U.S. Cl.**
CPC ***F03G 7/06*** (2013.01); ***A61F 2/08*** (2013.01); ***A61F 2/70*** (2013.01); ***A61H 1/02*** (2013.01);
(Continued)

(58) **Field of Classification Search**
CPC F15B 15/103; F15B 15/10; F15B 21/065; F15B 21/06; F15B 2215/305; H02N 1/004; H02N 1/00; H02N 1/002; H02N 1/006; H02N 1/008; H02N 1/06; H02N 1/08; A61F 2/08; A61F 2/70; A61F 2/68; A61F 2/02;
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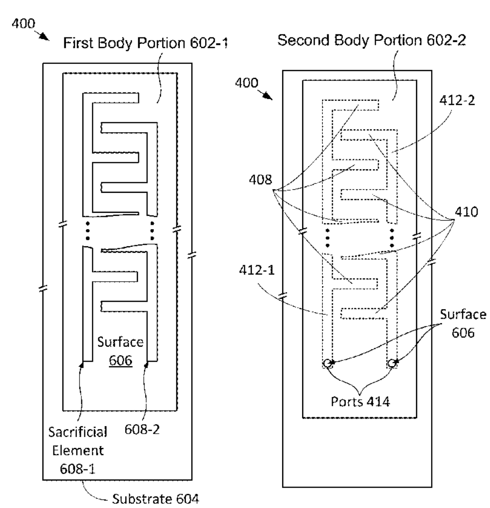
(57) **ABSTRACT**

Artificial muscles comprising a body of dielectric elastomer, wherein the body contains a pair of microfluidic networks are presented. Each microfluidic network includes a plurality of channels fluidically coupled via a manifold. The channels of the microfluidic networks are interdigitated and filled with conductive fluid such that each set of adjacent channels functions as the electrodes of an electroactive polymer (EAP) actuator. By using the manifolds as compliant wiring to energize the electrodes, artificial muscles in accordance with the present disclosure mitigate some or all of the reliability problems associated with prior-art artificial muscles.

13 Claims, 8 Drawing Sheets



Related U.S. Application Data

(60) Provisional application No. 62/745,599, filed on Oct. 15, 2018, provisional application No. 62/684,856, filed on Jun. 14, 2018.

(51) **Int. Cl.**

B33Y 80/00	(2015.01)
A61H 1/02	(2006.01)
F15B 21/06	(2006.01)
F03G 7/06	(2006.01)
A61F 2/70	(2006.01)
A61H 3/00	(2006.01)
B25J 9/10	(2006.01)
F15B 15/10	(2006.01)
F15B 13/044	(2006.01)
A61F 2/48	(2006.01)

(52) **U.S. Cl.**

CPC ***A61H 3/00*** (2013.01); ***B25J 9/10*** (2013.01); ***B25J 9/1075*** (2013.01); ***B33Y 80/00*** (2014.12); ***F15B 13/044*** (2013.01); ***F15B 15/103*** (2013.01); ***F15B 21/065*** (2013.01); ***H02N 1/004*** (2013.01); *A61F 2/482* (2021.08); *A61F 2002/0894* (2013.01); *A61H 2201/1207* (2013.01); *F15B 2215/305* (2013.01)

(58) **Field of Classification Search**

CPC A61F 2/48; A61F 2/482; A61F 2002/0894; A61H 1/02; A61H 2201/1207; A61H 2201/12; B25J 9/10; B25J 9/1075; B33Y 80/00

See application file for complete search history.

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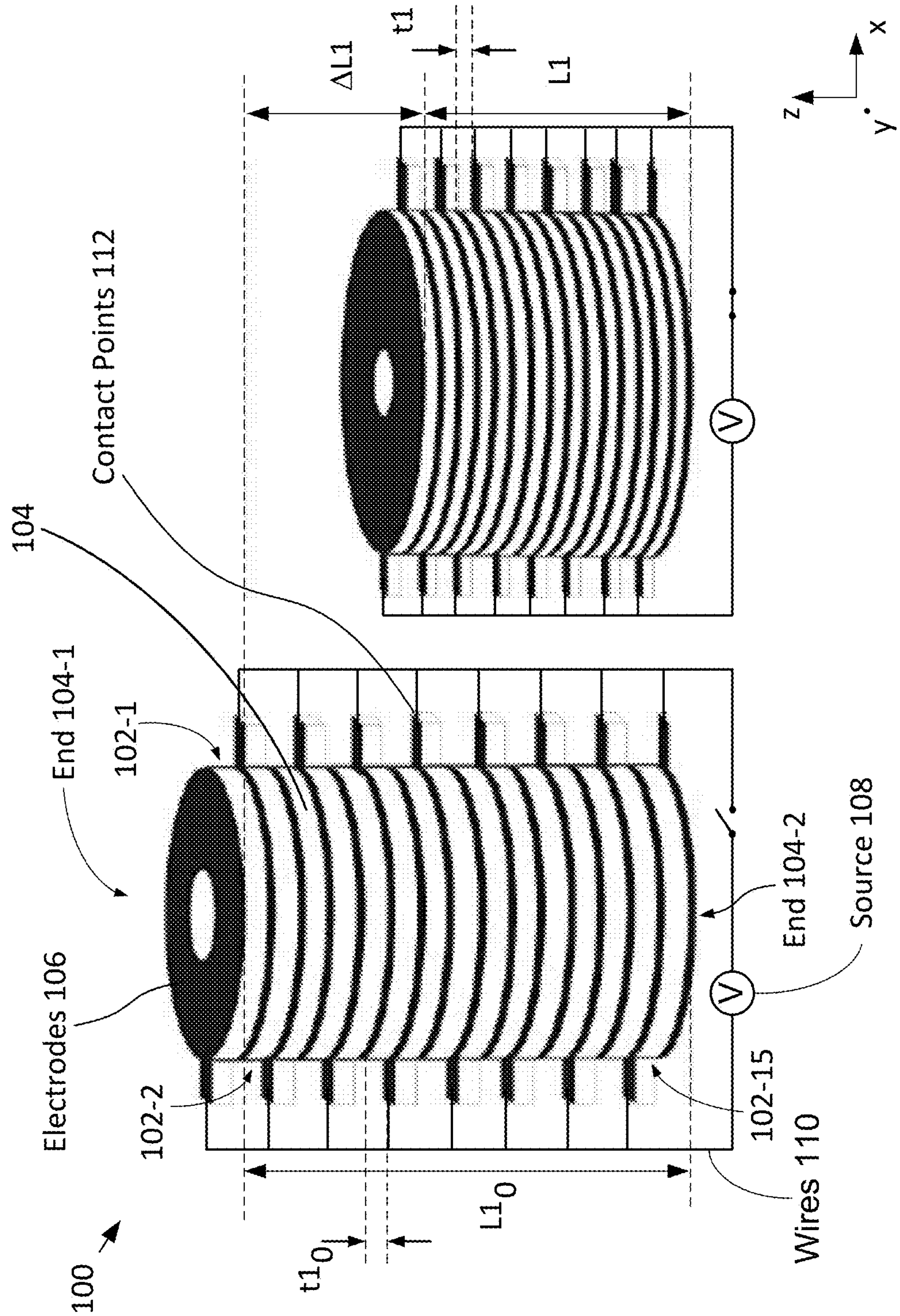


FIG. 1 (Prior Art)

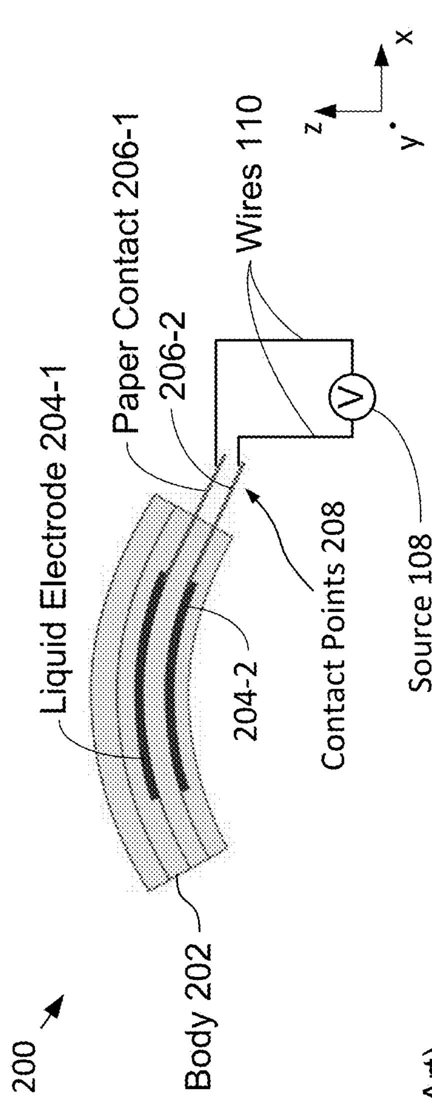


FIG. 2 (Prior Art)

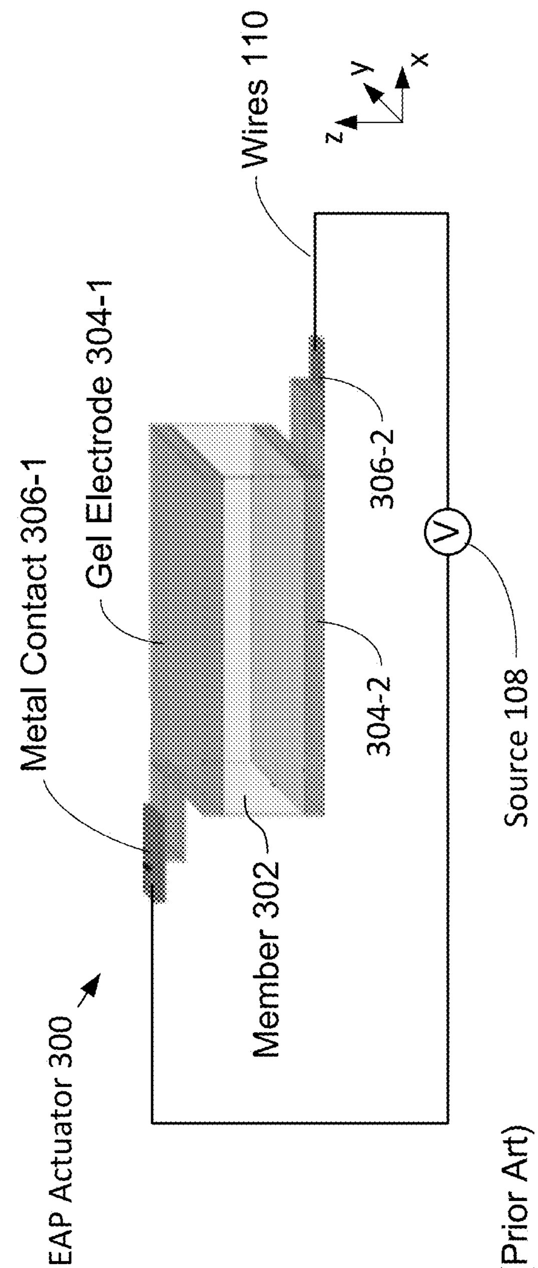


FIG. 3 (Prior Art)

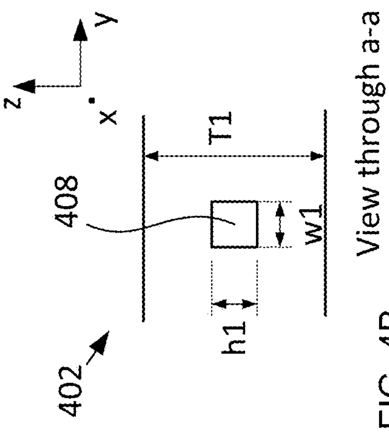


FIG. 4B

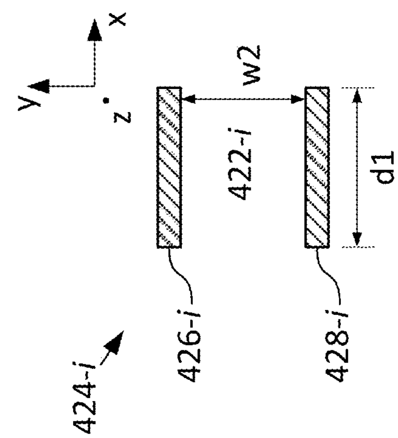


FIG. 4C

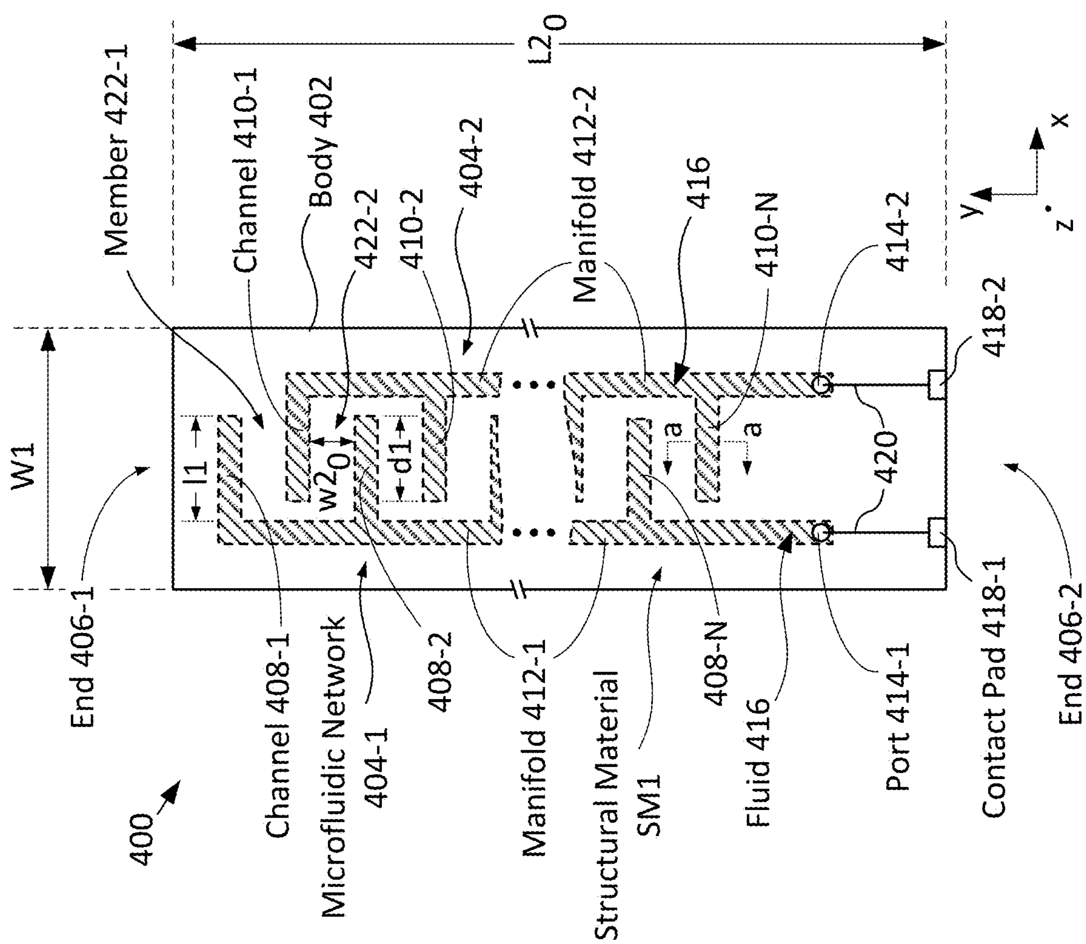
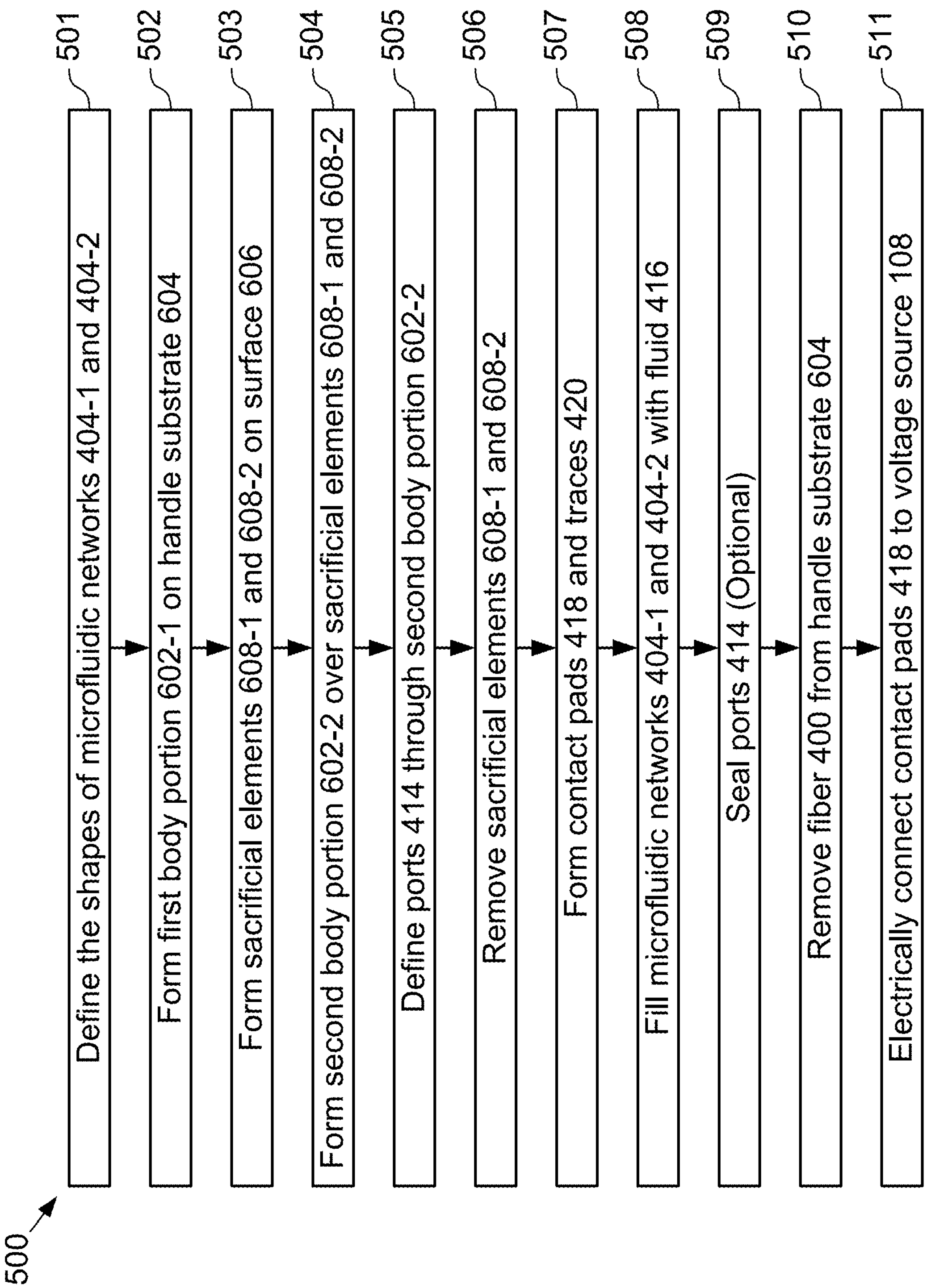


FIG. 4A

FIG. 5


500
501
Define the shapes of microfluidic networks 404-1 and 404-2
502
Form first body portion 602-1 on handle substrate 604
503
Form sacrificial elements 608-1 and 608-2 on surface 606
504
Form second body portion 602-2 over sacrificial elements 608-1 and 608-2
505
Define ports 414 through second body portion 602-2
506
Remove sacrificial elements 608-1 and 608-2
507
Form contact pads 418 and traces 420
508
Fill microfluidic networks 404-1 and 404-2 with fluid 416
509
Seal ports 414 (Optional)
510
Remove fiber 400 from handle substrate 604
511
Electrically connect contact pads 418 to voltage source 108

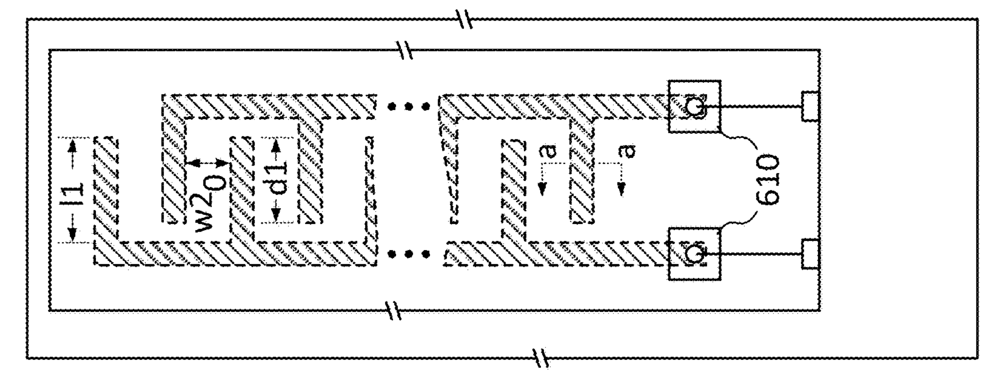


FIG. 6C

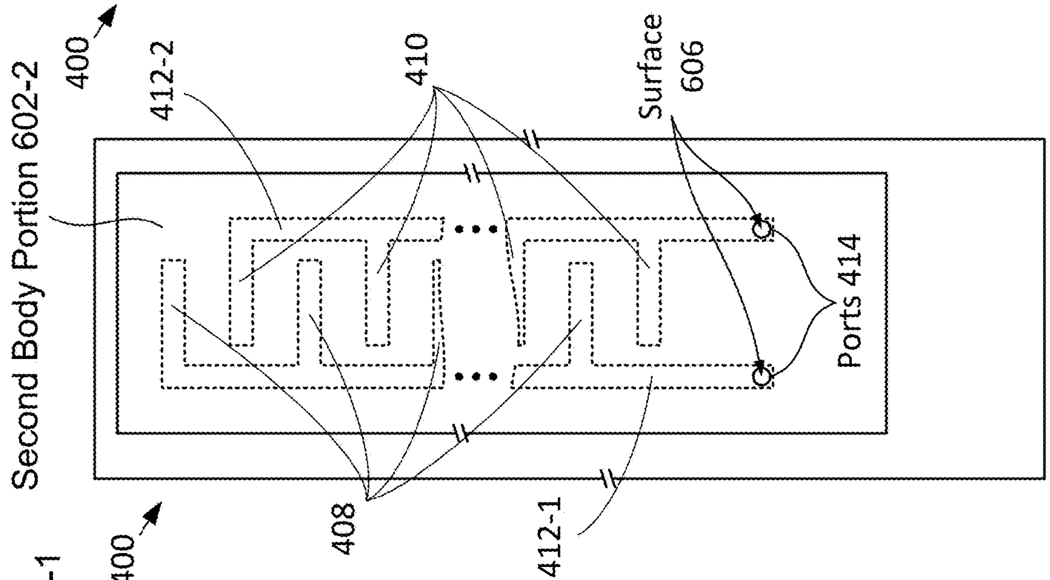


FIG. 6B

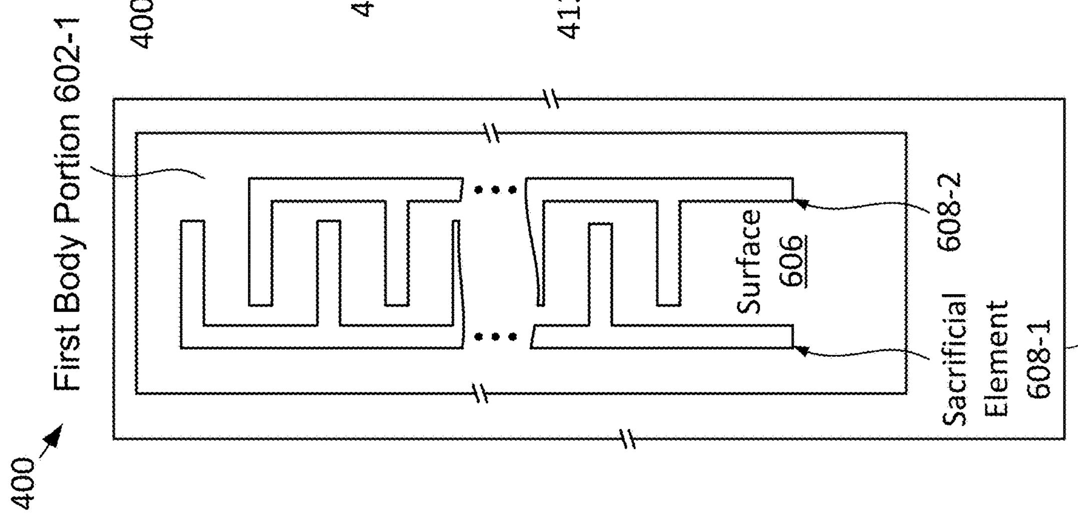


FIG. 6A

FIG. 7

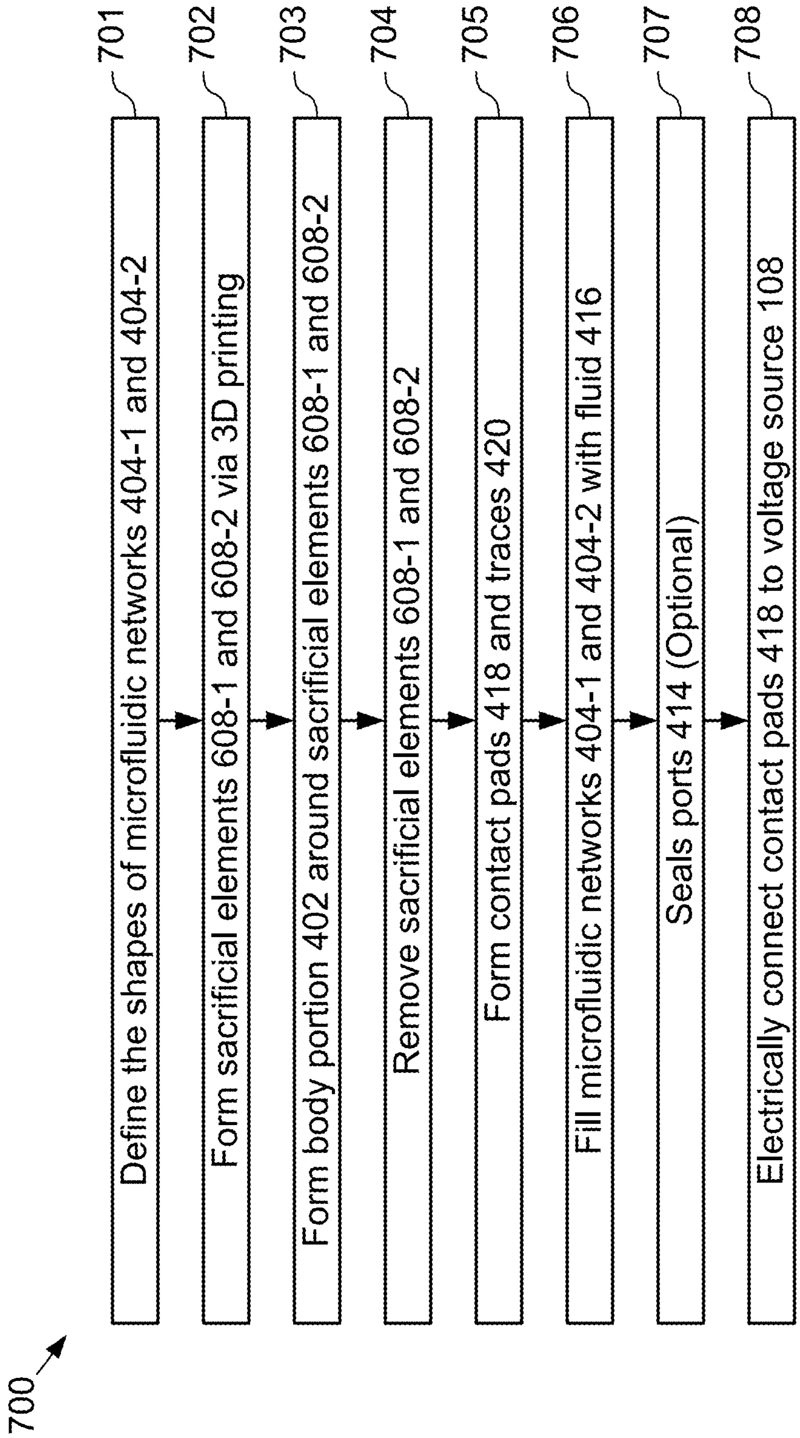

700
701
Define the shapes of microfluidic networks 404-1 and 404-2
702
Form sacrificial elements 608-1 and 608-2 via 3D printing
703
Form body portion 402 around sacrificial elements 608-1 and 608-2
704
Remove sacrificial elements 608-1 and 608-2
705
Form contact pads 418 and traces 420
706
Fill microfluidic networks 404-1 and 404-2 with fluid 416
707
Seals ports 414 (Optional)
708
Electrically connect contact pads 418 to voltage source 108

FIG. 8

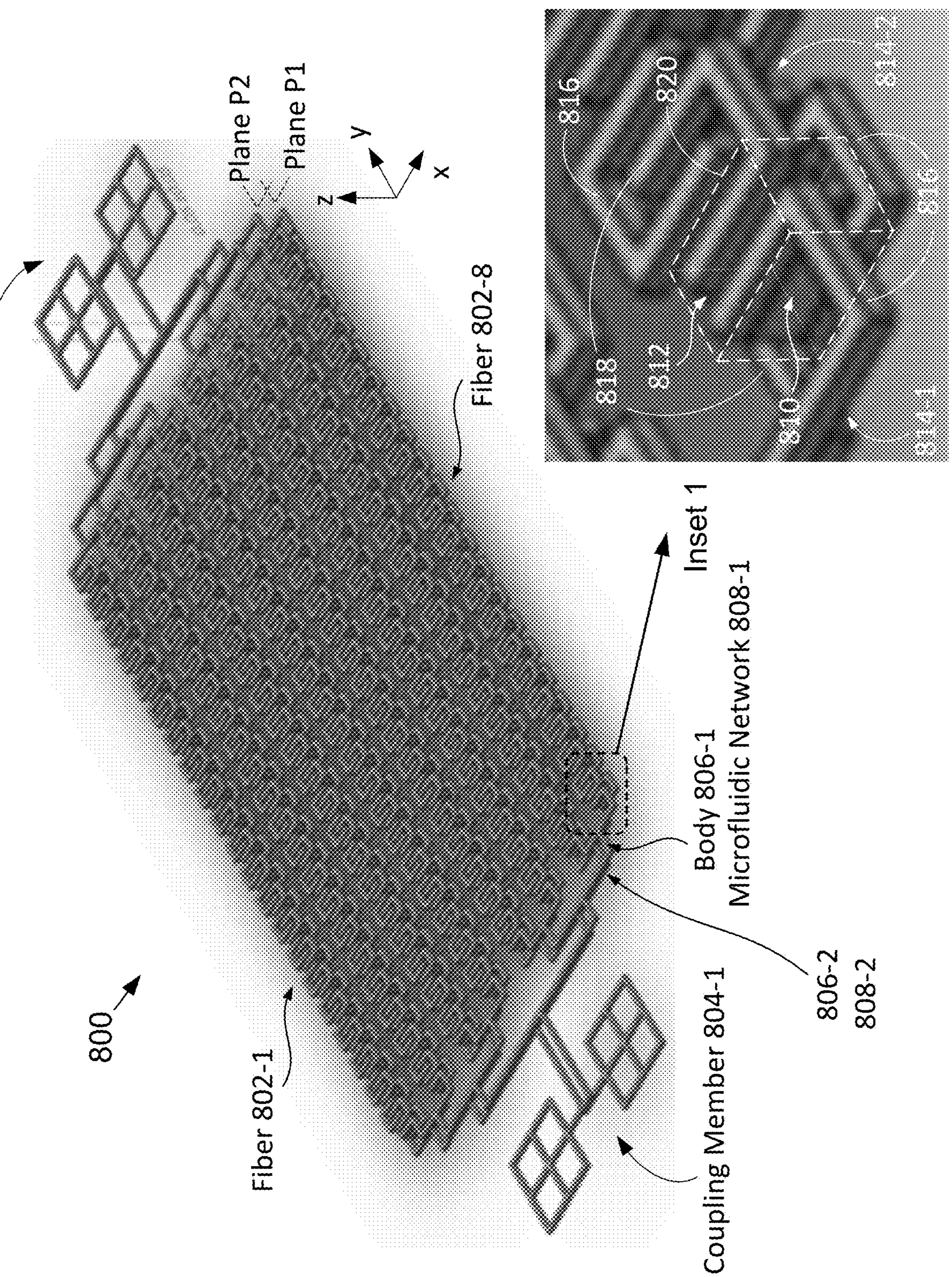
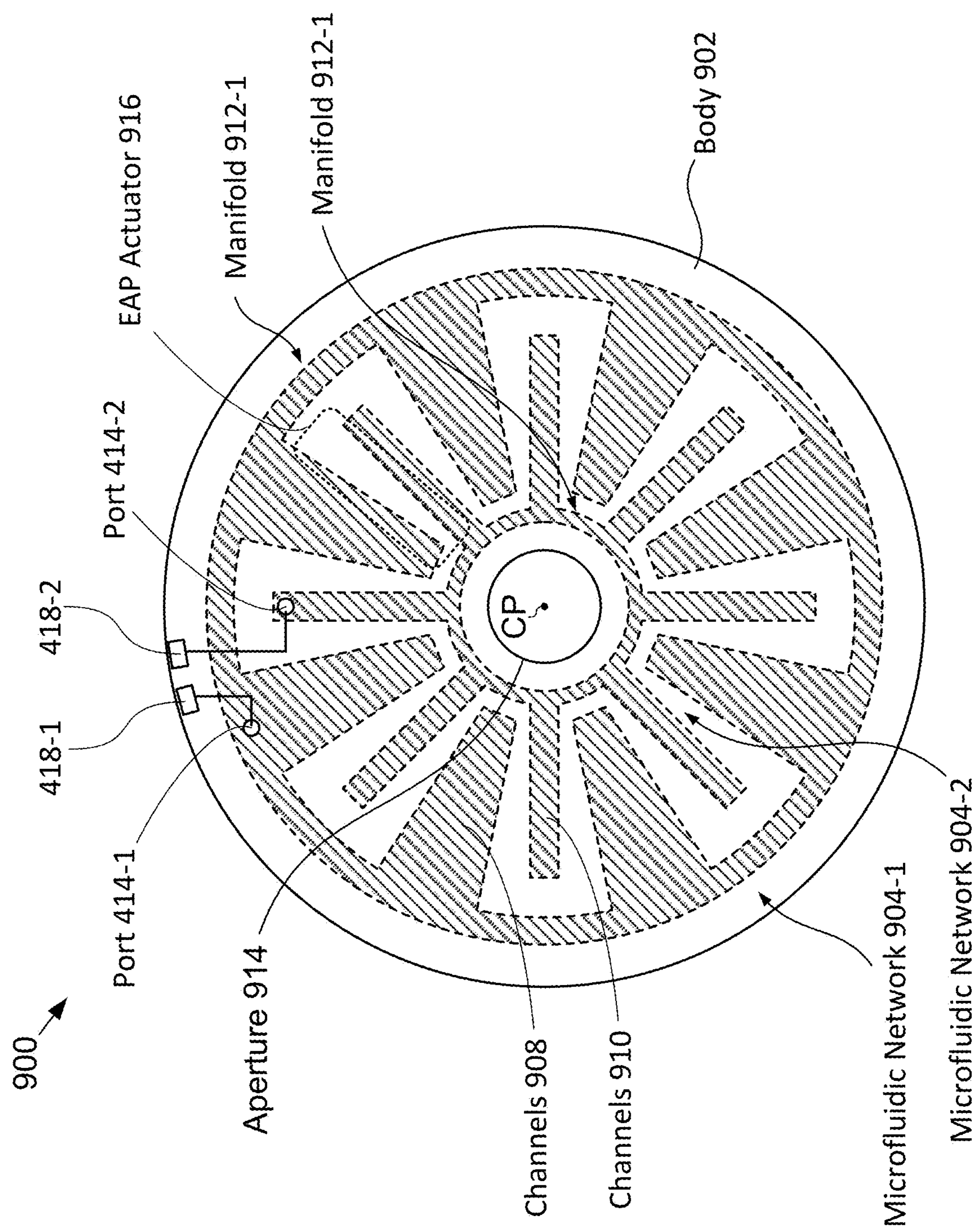

800
Coupling Member 804-2
Plane P2
Plane P1
y
x
z
Fiber 802-8
Fiber 802-1
Inset 1
Body 806-1
Microfluidic Network 808-1
806-2
808-2
Coupling Member 804-1
816
820
814-2
818
812
810
814-1
816

FIG. 9


900
Port 414-2
418-2
418-1
Port 414-1
Aperture 914
EAP Actuator 916
Manifold 912-1
Manifold 912-1
Body 902
CP
Channels 908
Channels 910
Microfluidic Network 904-1
Microfluidic Network 904-2

MICROFLUIDIC-BASED ARTIFICIAL MUSCLES AND METHOD OF FORMATION

CROSS REFERENCE TO RELATED APPLICATIONS

This application is a continuation of co-pending U.S. patent application Ser. No. 16/442,092, filed Jun. 14, 2019, entitled "Microfluidic-Based Artificial Muscles and Method of Formation", which claims the benefit of U.S. Provisional Application Ser. No. 62/684,856, filed Jun. 14, 2018, entitled "Biomimetic Force Generation By Soft Microfluidic Capacitor Stack" and U.S. Provisional Application Ser. No. 62/745,599, filed Oct. 15, 2018, entitled "Biomimetic Force Generation By Soft Microfluidic Capacitor Stack", each of which is incorporated herein by reference. If there are any contradictions or inconsistencies in language between this application and one or more of the cases that have been incorporated by reference that might affect the interpretation of the claims in this case, the claims in this case should be interpreted to be consistent with the language in this case.

TECHNICAL FIELD

The present disclosure relates to electrostatic actuators in general, and, more particularly, to electrostatic actuators suitable for use in exoskeletons, prosthetics, and vehicle propulsion.

BACKGROUND

Practical artificial muscles are sought after for use in a wide range of applications, including exoskeletons for augmenting human strength and capability, wearable electronic devices, prosthetic devices, walking robots, and acoustically quiet underwater propulsion systems.

Artificial muscles have been the subject of investigation for several years, with the strongest emphasis being on conventional technologies such as electromagnetics, pneumatics, hydraulics, thermal actuators, shape-memory alloys, and electrically active polymers (EAP). Unfortunately, as practiced in the prior art, these technologies have many drawbacks that make them impractical for use in many artificial-muscle applications.

Electromagnetic devices, such as linear and rotary electric motors, require high magnetic fields to generate significant force. The generation of high magnetic fields, however, requires large numbers of windings and/or high electric currents. As a result, electromagnetic devices tend to be large, bulky, and power hungry.

A pneumatic or hydraulic device can generate high force, but requires bulky and complex arrangements of tubing that are not suitable for many applications. Furthermore, the force-generation capability of such devices does not scale well into the small size domain required in most artificial muscles.

Thermal actuators can also generate significant force, but are slow and consumer a great deal of power. Furthermore, it can be difficult to remove heat effectively, once it has been generated to induce actuation. Since actuation speed and frequency are limited by heat-transfer rates, such actuators are not fast enough for artificial muscles in typical applications.

Shape-memory alloys (SMA) are metal alloys that can be easily deformed when below a "transition temperature" but return to the shape in which they were formed when heated above this temperature. While SMA artificial muscles have been demonstrated, the material is heavy, actuation tends to be slow, they consume considerable power, and the removal of heat to reverse actuation can be a challenge.

Electroactive polymers (EAP) actuators are elements that change shape under the influence of an applied electric field and are typically considered to most closely emulate biological muscles. Unfortunately, conventional approaches to fabricating EAP actuators are cumbersome, complicated, and not reproducible. Furthermore, many conventional EAP actuators have very poor efficiencies and are not durable. In addition, EAP actuators have historically proven difficult to manufacture.

The need for a simple, easily fabricated actuation technology suitable for use in artificial muscles remains, as yet, unmet in the prior art.

SUMMARY

The present disclosure is directed to artificial muscles based on arrays of electrostatic actuators that include resilient dielectric material disposed between liquid, semi-liquid, or gel electrodes. Artificial muscles in accordance with the present disclosure are particularly well suited for use in acoustically quiet propulsion systems for underwater vehicles, armored exoskeletons, and prosthetics.

Like artificial muscles known in the prior art, embodiments in accordance with the present disclosure employ electrostatic actuators that comprise a resilient dielectric material that is sandwiched between a pair of compliant electrodes. Unfortunately, prior-art electrostatic actuators, such as EAP actuators, are difficult to fabricate in high volume, complex, and unreliable due, in part, to the poor quality of the flexible electrodes used in them. Furthermore, these flexible electrodes require complicated interconnection schemes to actuate arrays of actuators.

In sharp contrast to the prior art, embodiments in accordance with the present disclosure employ a body of dielectric elastomer that contains a pair of microfluidic networks, each comprising a plurality of channels and a manifold that are filled with conductive fluid (e.g., a liquid, semi-liquid, gel, fluid suspension, etc.). The body and the channels collectively define one or more arrays of electrostatic actuators having a dielectric-elastomer member sandwiched between a set of opposing electrodes that are defined by the conductive-fluid-filled channels. The channels of each microfluidic network are electrically and fluidically coupled via its respective manifold. The use of such manifolds enables fluidically and electrically interconnection of large numbers of channel electrodes with high reliability.

An illustrative embodiment is an artificial muscle fiber comprising a serially connected array of micro-capacitors each having liquid positive and negative electrodes separated by an intervening dielectric material. The positive electrodes of the capacitors are electrically and fluidically coupled via a first manifold. The negative electrodes of the capacitors are electrically and fluidically coupled via a second manifold. In response to a voltage differential applied to the first and second manifolds, the electrodes of each capacitor generate an electrostatic force that induces a stress (Maxwell stress) on its intervening dielectric. The induced stress gives rise to a compressive force in each actuator and/or a compression of each capacitor, which manifests as a reduction in the total length of the muscle fiber and/or a "pull-in" force (i.e., tension) between the two ends of the muscle fiber.

In the illustrative embodiment, the artificial muscle fiber is formed via 3D printing of bulk structural material that

comprises a resilient dielectric. Microfluidic channels that define each manifold and its associated channels are formed simultaneously. In some embodiments, a network of sacrificial material is first formed in the shapes of the manifolds and electrodes. The structural material is then 3D printed around the sacrificial material. Afterward, the sacrificial material is removed, thereby generating the microfluidic channels that define the manifolds and their associated channels. These channels are then filled with a liquid conductor via their manifolds using a conventional process, such as dead-end priming, etc.

In some embodiments, a sphincter-like artificial fiber is formed by arranging first and second arrays of electrodes radially about a center point of a disk of resilient dielectric such that each of the second array of electrodes is between a pair of electrodes of the first array. The electrodes of the first array are electrically and fluidically coupled via a first manifold and the electrodes of the second array are electrically and fluidically coupled via a second manifold. When a voltage differential is applied to the first and second manifolds, the first and second arrays of electrodes are attracted to one another giving rise to radial compression of the dielectric that manifests in the bulging of the disk.

In some embodiments, one- or two-dimensional arrangements of linear muscle fibers are defined to realize a muscle-fiber bundle capable of generating significant force.

An embodiment in accordance with the present disclosure is an artificial muscle comprising: a first body comprising a first material that is a dielectric elastomer; a first plurality of channels; a second plurality of channels, wherein the channels of the first plurality thereof and the channels of the second plurality thereof are interdigitated and spaced apart within the first body; a first manifold located within the first body, wherein the first manifold is fluidically coupled with each of the first plurality of channels; and a second manifold located within the first body, wherein the second manifold is fluidically coupled with each of the second plurality of channels; wherein each of the first manifold, the first plurality of channels, the second manifold, and the second plurality of channels is filled with a second material that is electrically conductive, and wherein the second material is selected from the group consisting of a liquid and a gel; and wherein the first and second plurality of channels are configured such that the application of a voltage differential between the second material in the first manifold and the second material in the second manifold generates a first attractive force between each set of adjacent first and second channel.

Another embodiment in accordance with the present disclosure is an artificial muscle comprising: a first microfluidic network comprising a first manifold and a first plurality of channels, the first microfluidic network being filled with a first material that is electrically conductive; a second microfluidic network comprising a second manifold and a second plurality of channels, the second microfluidic network being filled with a second material that is electrically conductive; a plurality of members comprising a third material that is a dielectric elastomer; wherein the first microfluidic network, the second microfluidic network, and the plurality of members are arranged such that (1) the channels of the first plurality thereof and the channels of the second plurality thereof are interdigitated and (2) each member of the plurality thereof is between a set of adjacent first and second channels; and wherein the first and second plurality of channels are configured such that the application of a voltage differential between the first material in the first microfluidic network and the second material in the second microfluidic network gives rise to a compressive force on each member of the plurality thereof.

Yet another embodiment in accordance with the present disclosure is a method for forming an artificial muscle, the method comprising: forming a body that comprises a first material that is a dielectric elastomer, wherein the body includes (1) a first microfluidic network that includes a first plurality of channels and a first manifold that is fluidically coupled with each channel of the first plurality thereof and (2) a second microfluidic network that includes a second plurality of channels and a second manifold that is fluidically coupled with each channel of the second plurality thereof, wherein the first microfluidic network and second microfluidic network are arranged such that the channels of the first plurality thereof and the channels of the second plurality thereof are interdigitated and spaced apart; filling the first microfluidic network with a second material that is electrically conductive, wherein the second material is selected from the group consisting of a liquid and a gel; and filling the second microfluidic network with the second material.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1** depicts a schematic drawing of side views of a prior-art linear actuator proposed for use in an artificial muscle in its unactuated and actuated states.

FIG. **2** depicts a schematic drawing of a cross-sectional view of another prior-art actuator proposed for use in an artificial muscle.

FIG. **3** depicts a schematic drawing of a cross-sectional view of yet another prior-art actuator proposed for use in an artificial muscle.

FIG. **4**A depicts a schematic drawing of a top view of an illustrative embodiment of an artificial muscle fiber in accordance with the present disclosure.

FIG. **4**B depicts a schematic drawing of a sectional view of a representative channel **408** in accordance with the illustrative embodiment.

FIG. **4**C depicts a schematic drawing of a sectional view of actuator **424**-*i* in its nascent state.

FIG. **5** depicts operations of a first exemplary method suitable for use in fabricating an artificial muscle in accordance with the illustrative embodiment.

FIGS. **6**A-C depict schematic drawings of top views of fiber **400** at different stages of its fabrication.

FIG. **7** depicts operations of a second exemplary method suitable for use in fabricating an artificial muscle in accordance with the illustrative embodiment.

FIG. **8** depicts a schematic drawing of a perspective view of an artificial muscle fiber bundle in accordance with the present disclosure.

FIG. **9** depicts a schematic drawing of a top view of another artificial muscle in accordance with the present disclosure.

DETAILED DESCRIPTION

FIG. **1** depicts a schematic drawing of side views of a prior-art linear actuator proposed for use in an artificial muscle in its unactuated and actuated states. Actuator **100** includes electroactive polymer (EAP) actuators **102-1** through **102-15**, which are stacked in series to define a linear structure having ends **104-1** and **104-2**. Actuator **100** is analogous to actuators disclosed by Chuc, et al., in "Fabrication and Control of Rectilinear Artificial Muscle Actuator," *IEEE/ASME Trans. on Mechatronics*," Vol. 16, pp. 167-176 (2011).

Each of EAP actuators **102-1** through **102-15** (referred to, collectively, as EAP actuators **102**) includes layer **104** having an electrode **106** disposed on each of its top and bottom surfaces. Adjacent EAP actuators share a common electrode **106** at the interface between them. Each EAP actuator **102** has a nascent thickness, $t1_0$, in its unactuated state.

Layer **104** is typically a dielectric elastomer, such as acrylic elastomers, polydimethylsiloxane (PDMS), and the like.

To accommodate deformation of EAP actuator **102**, each of electrodes **106** preferably comprises a compliant electrically conductive material or layer, such as carbon grease, saline solutions, liquid metals, microcracked metals, serpentine-shaped metal layers, carbon nanotubes (CNT), gel electrolytes, ionic hydrogels formed with salt water, graphene, ionic conductors (e.g., electrolytes, etc.), or surface-implanted layers of metallic nanoclusters.

Electrodes **106** are electrically connected to voltage source **108** by conventional wires **110**, each of which is electrically connected to an electrode at a contact point **112**.

In its unactuated state, actuator **100** has a total nascent length, $L1_0$, between ends **104-1** and **104-2**, which is equal to N times the nascent thickness, $t1_0$, of each EAP actuator **102**.

When voltage, V, is applied between the electrodes **106** through wires **110-1** and **110-2**, an electrostatic force develops between the electrodes of each EAP actuator giving rise to a compression of its elastomer layer and a reduction of its thickness. As a result, each EAP actuator is compressed to a reduced thickness, t**1**, and the total length of actuator **100** is reduced to actuated length, L**1**. Ends **104-1** and **104-2** are drawn together by displacement, $\Delta L1$, which is equal to N times the compression induced in each EAP actuator **102**.

It should be noted that, as actuator **100** shrinks in length, contact points **112** move closer together. As a result, shear stress can develop at these contact points, which can cause delamination of the electrode material, breakage of a wire, or another failure that results in an electrical disconnection of one or more wires from their respective contact points.

FIG. **2** depicts a schematic drawing of a cross-sectional view of another prior-art actuator proposed for use in an artificial muscle. Actuator **200** is a cantilever actuator whose curvature depends on the voltage applied across its electrodes. Actuator **200** comprises body **202**, liquid electrodes **204-1** and **204-2**, and paper contacts **206-1** and **206-2**. Actuator **200** is analogous to actuators disclosed by Finkenauer, et al., in "Compliant liquid metal electrodes for dielectric elastomer actuators," *Proc. of SPIE*," Vol. 9056, pp. 905631-1 to 905631-7 (**2014**).

Body **202** is a cantilever structure comprising a dielectric elastomer, such as PDMS. Body **202** encloses liquid electrodes **204-1** and **204-2**, each of which comprises a liquid metal alloy—specifically, eutectic Gallium-Indium.

Paper contacts **206-1** and **206-2** are provided to enable electrical contact to the liquid electrodes. Each paper contact is a partial sheet of conductive paper that is inserted into the structure by hand before the material of body **202** is fully cured.

Wires **110** are electrically connected to paper contacts **206-1** and **206-2** at contact points **208** to enable voltage source **108** to provide a voltage differential across the paper contacts.

As will be apparent to one skilled in the art, actuator **200** has several significant drawbacks that make it unattractive for use in practical artificial muscles. First, its assembly requires steps that must be performed by hand; therefore, it is not well suited to mass manufacture. Second, the use of paper contacts gives rise to serious reliability issues. Third, as discussed above with respect to actuator **100**, motion of actuator **200** during actuation can cause shear stress at contact points **208** that can lead to device failure.

FIG. **3** depicts a schematic drawing of a cross-sectional view of yet another prior-art actuator proposed for use in an artificial muscle. Actuator **300** is an EAP actuator whose thickness, t**2**, depends on the voltage applied across its electrodes. Actuator **300** comprises member **302**, gel electrodes **304-1** and **304-2**, and contacts **306-1** and **306-2**. Actuator **300** is analogous to actuators disclosed by Keplinger, et al., in Stretchable Transparent Ionic Conductors," *Science, Vol.* 341, pp. 984-987 (2013).

Member **302** is a slab of dielectric elastomer, such as PDMS. The top and bottom surfaces of member **302** are coated with an electrolytic elastomer (e.g., polyacrylamide hydrogel containing sodium chloride).

Metal contacts **306-1** and **306-2** are copper sheets that are placed on their respective gel electrodes. Wires **110** are joined with the metal contacts at contact points **112**, as discussed above, to electrically connect each of the metal contacts to voltage source **108**.

Unfortunately, actuator **300** has all of the same drawbacks as those discussed above with respect to actuators **100** and **200**. Furthermore, the electrolytic nature of gel electrodes **304-1** and **304-2** add an additional layer of complexity to the operation of actuator **300**. Specifically, the interface between the metal contacts and the electrolyte material can form a double layer under certain bias conditions, which results in the double layer functioning as an additional capacitor.

It is an aspect of the present disclosure that at least some of the problems of prior art artificial muscles comprising pluralities of electrostatic actuators, such as EAP actuators, can be mitigated by employing microfluidic networks in the body of an artificial muscle, where the microfluidic networks are filled with an electrically conductive fluid. Furthermore, each microfluidic network includes a port at which a robust electrical connection to its conductive fluid can be made. For the purposes of this Specification, including the appended claims, the term "fluid" is defined as a liquid, semi-liquid, fluid suspension, gel, or equivalent.

FIG. **4**A depicts a schematic drawing of a top view of an illustrative embodiment of an artificial muscle fiber in accordance with the present disclosure. Fiber **400** includes body **402**, within which microfluidic networks **404-1** and **404-2** reside.

Body **402** is a substantially straight strip of structural material SM**1**. In the depicted example, structural material SM**1** is a dielectric-elastomer—specifically, PDMS. Body **402** has width W**1** and thickness T**1** and extends from first end **406-1** to second end **406-2** to define nascent length $L2_0$. In the depicted example, body comprises PDMS, L2=20 cm, W1=10 mm, and T1=100 microns; however, in some embodiments, body **402** comprises a different suitable dielectric elastomer and/or has a different width and/or thickness. Furthermore, although fiber **400** is depicted as being straight, it should be noted that fiber **400** can have any practical shape (e.g., curved in one or more dimensions, helical, etc.) without departing from the scope of the present disclosure.

Fluidic network **404-1** includes channels **408-1** through **408**-N (where N is any practical number), manifold **412-1**, and port **414-1**.

Fluidic network **404-2** includes channels **410-1** through **410**-N (referred to, collectively, as channels **410**), manifold **412-2**, and port **414-2**.

FIG. **4**B depicts a schematic drawing of a sectional view of a representative channel **408** within body **402** in accordance with the illustrative embodiment. The sectional view shown in FIG. **4**B is taken through line a-a shown in FIG. **4**A and the depicted channel is representative of each of channels **408-1** through **408**-N (referred to, collectively, as channels **408**) and **410-1** through **410**-N (referred to, collectively, as channels **410**).

Each of channels **408** and **410** is a microchannel that resides completely within body **402**. channels **408** and **410** are substantially identical and have length l**1**, width w**1**, and height h**1**. In the depicted example, channels **408** and **410** are approximately 6 mm long (i.e., l**1**=6 mm) and have a substantially square cross-section and w**1** and h**1** are each approximately 30 microns. channels **408** and **410** are interdigitated to define overlap distance, d**1**. In the depicted example, d**1** is equal to approximately 5.3 mm.

Returning now to FIG. **4**A, each of manifolds **412-1** and **412-2** (referred to, collectively, as manifolds **412**) is a microfluidic channel that resides within body **402**. In the depicted example, each of manifolds **412** has the same cross-sectional dimensions as channels **408** and **410**; however, one or more of the manifolds and channels can have different dimensions without departing from the scope of the present disclosure.

Ports **414-1** and **414-2** (referred to, collectively, as ports **414**) are openings through body **402** that enable microfluidic networks **404-1** and **404-2**, respectively, to be filled with conductive fluid. In some embodiments, each of ports **414** is located at a position in body **402** that does not undergo significant motion during the actuation of fiber **400** to mitigate the potential for electrical failure.

Manifold **412-1** extends from channel **408-1** to port **414-1**, thereby fluidically coupling port **414-1** with each of channels **408**.

In similar fashion, manifold **412-1** extends from channel **410-1** to port **414-2**, thereby fluidically coupling port **414-2** with each of channels **410**.

Each of microfluidic networks **404-1** and **404-2** is filled with fluid **416**, which is a conventional electrically conductive fluid. In the depicted example, fluid **416** is saline; however, myriad conductive liquids and/or gels can be used in one or both microfluidic networks without departing from the scope of the present disclosure. Conductive fluids suitable for use in accordance with the present disclosure include, without limitation, carbon grease, saline solutions, liquid metals, liquid electrolytes, gel electrolytes, ionic hydrogels formed with salt water, ionic conductors, and the like.

Typically, microfluidic networks **404-1** and **404-2** are filled with the same conductive fluid; however, in some embodiments, the microfluidic networks are filled with different conductive fluids.

Contact pads **418-1** and **418-2** (referred to, collectively, as contact pads **418**) are conventional metal contacts disposed on the outer surface of body **402**. Each of contact pads **418** is electrically connected to its respective port **414** via a conventional electrical trace **420**. When microfluidic networks **404-1** and **404-2** are filled with fluid **416**, therefore, contact pads **418** are in electrical contact with the conductive fluid, thereby enabling robust electrical connectivity between the microfluidic networks and voltage supply **108** (not shown).

Members **422-1** through **422**-N (referred to, collectively, as members **422**) are portions of body **402** having nascent width $w2_0$. Each of members **422** resides between a different set of adjacent interdigitated channels **408** and **410**. For example, member **422-1** lies between channels **408-1** and **410-1**, member **422-2** lies between channels **408-2** and **410-2**, and so on. In the depicted example, nascent width $w2_0$ is equal to 10 microns; however, any suitable nascent width can be used for members **422** without departing from the scope of the present disclosure.

As will be appreciated by one skilled in the art, when channels **408**-*i* and **410**-*i*, where i=1 through N, are filled with fluid **416**, they function as electrodes that oppose each other across their respective member **422**-*i*, thereby collectively defining parallel-plate capacitor-based electrostatic actuator **424**-*i*.

FIG. **4**C depicts a schematic drawing of a sectional view of electrostatic actuator **424**-*i* in its nascent state. The sectional view shown in FIG. **4**C is taken through the middle of the thickness of the electrostatic actuator. Electrostatic actuator **424**-*i* includes electrodes **426**-*i* and **428**-*i*, which are defined by fluid-filled channels **408**-*i* and **410**-*i*, respectively.

When each of actuated by applying a voltage between ports **414-1** and **414-2**, the compressive force applied to member **422**-*i* is given by:

$$F = \frac{A\varepsilon_0\varepsilon_r^2 V^2}{2 \cdot w2^2}, \tag{1}$$

where V is the applied voltage, A is the area of the opposing faces of electrodes **426**-*i* and **428**-*i* (i.e., h**1**×d**1**), w**2** is the separation between these faces, and ε_r is the relative dielectric constant of structural material SM**1**.

Expressed in terms of force per unit cross-sectional area of the electrodes, equation (1) becomes:

$$F = \frac{\varepsilon_0\varepsilon_r^2 V^2}{2 \cdot w2^2}. \tag{2}$$

For the dimensions of fiber **400** provided above, and using a value of 2.5 for the relative dielectric constant of PDMS, for an applied voltage of 3 kV, the generated force per unit area for electrostatic actuator **424**-*i* is approximately 249 N/cm^2 (i.e., 362 $lbs./in^2$).

Electrostatic actuators **424-1** through **424**-N are stacked in series such that the compressive force generated at each of the actuators gives rise to "pull-in" force (i.e., a tension) between ends **406-1** and **406-2**. Since electrostatic actuators **424-1** through **424**-N are connected in series, however, they function as springs attached in series; thus, the compressive forces they generate are not cumulative. As a result, the tension generated between ends **406-1** and **406-2** is substantially equal to the force generated by a single electrostatic actuator. In some embodiments, the tension generated between ends **406-1** and **406-2** results in a reduction in the length of fiber **400**.

It should be noted, however, that the contractions that occur at each electrostatic actuator are cumulative and can give rise to a macroscopic length change that is equal to N times the contraction of a single electrostatic actuator (assuming all actuators contract the same amount). In the depicted example, nascent length L**2***o* is approximately 20 cm, which enables approximately 2 cm of total contraction of fiber **400**—approximately the amount a human bicep can contract.

As discussed below in more detail, an artificial muscle capable of generating more force can be formed by using a plurality of fibers **400** that are mechanically coupled in parallel.

While fiber **400** is, in some ways, analogous to actuator **100** described above, the use of microfluidic techniques in embodiments in accordance with the present disclosure provides significant advantages to that make practical artificial muscle fibers feasible. These advantages include:

i. conductive-fluid-filled channels are highly reliable compliant electrodes that function as compliant electrodes for electrostatic actuators; or

ii. conductive-fluid-filled manifolds that can flex and bend without the danger of wires snapping and disconnecting enables them to function as flexible and robust electrical wiring for electrically connecting the capacitor electrodes to a voltage source; or

iii. the formation of the microfluidic networks within the body of a muscle fiber eliminates the need for separate dielectric material and hard electrodes to be built into the muscle fiber; or

iv. formation of the microfluidic networks within the body of a muscle fiber enables high-volume manufacture because it can be easily automated by using, for example, three-dimensional (3D) printing, photolithographic-based patterned molding, MEMS fabrication techniques, silk-screening, etc.; or

v. any combination of i, ii, iii, and iv.

It should be further noted that muscles and/or muscle fibers in accordance with the present disclosure can be made such that they are monolithic using, for example, 3D-printed bulk dielectric and channels for soft wiring. Such monolithic structures are significantly sturdier and more robust than prior-art devices that comprise heterogeneous stacks of dissimilar materials (e.g., metals, polymers, paper electrodes, etc.). As a result, muscles and/or muscle fibers in accordance with the present disclosure can deliver and tolerate higher stresses before failure and can provide greater force safely and reliably.

FIG. **5** depicts operations of a first exemplary method suitable for use in fabricating an artificial muscle in accordance with the illustrative embodiment. Method **500** is described herein with continuing reference to FIGS. **4**A-C, as well as reference to FIGS. **6**A-C.

FIGS. **6**A-C depict schematic drawings of top views of fiber **400** at different stages of its fabrication.

Method **500** begins with operation **501**, wherein the shapes of microfluidic networks **404-1** and **404-2** are defined.

At operation **502**, first body portion **602-1** is formed on handle substrate **604**. Handle substrate **604** is a conventional substrate suitable for use in planar-processing fabrication.

First body portion **602-1** is a slab of structural material SM**1** that is formed in conventional fashion. First body portion **602-1** includes top surface **606** and has a thickness of approximately 35 microns and lateral dimensions based on the desired size of fiber **400**.

At operation **503**, sacrificial elements **608-1** and **608-2** are formed on surface **606** of first body portion **602-1**. In the depicted example, sacrificial elements **608-1** and **608-2** are photoresist patterns defined in the shapes of microfluidic networks **404-1** and **404-2**, respectively, using conventional photolithography.

FIG. **6**A shows nascent fiber **400** after the definition of sacrificial elements **608-1** and **608-2**.

At operation **504**, second body portion **602-2** is formed of structural material SM**1** such that it encapsulates sacrificial elements **608-1** and **608-2**. Second body portion **602-2** is formed such that it is a substantially conformal layer over the sacrificial elements and has a substantially planar top surface. The thickness of second body portion **602-2** is approximately 65 microns (outside of the regions of the sacrificial elements). The first and second body portions collectively define body **402** having a total thickness, therefore, of approximately 100 microns.

At operation **505**, ports **414** are defined such that they extend through second body portion **602-2** and expose portions of sacrificial elements **608-1** and **608-2**.

At operation **506**, sacrificial elements **608-1** and **608-2** are removed in conventional fashion, thus defining empty manifolds **412-1** and **412-2** and channels **408** and **410**.

FIG. **6**B shows nascent fiber **400** after the removal of sacrificial elements **608-1** and **608-2**.

At operation **507**, contact pads **418-1** and **418-2** and conductive traces **420** are formed in conventional fashion. In some embodiments, traces **420** are defined such that they extend into ports **414** and onto surface **606** in each of manifolds **412-1** and **412-2**.

At operation **508**, microfluidic networks **404-1** and **404-2** are filled with fluid **416** using a conventional technique, such as dead-end priming under static pressure through ports **414**, and the like.

At optional operation **509**, seals **610** are bonded over ports **414** to mitigate leakage of fluid **416** out of the microfluidic networks. In the depicted example, seals **610** are regions of structural material SM**1** that are formed separately and placed over the ports. In some embodiments, seals **610** are formed in a different conventional manner.

FIG. **6**C shows nascent fiber **400** after seals **610** have been bonded over ports **414**.

At operation **510**, fiber **400** is removed from handle substrate **604**.

At operation **511**, wires **110** are attached to contact pads **418** to connect fiber **400** to voltage source **108**.

It should be noted that method **500** represents merely one fabrication method suitable for use to form artificial muscles and/or muscle fibers in accordance with the present disclosure.

Another exemplary fabrication method suitable for use to fabricate muscles and/or muscle fibers disclosed herein is 3D printing via stereolithography (SLA), which requires no sacrificial material. In a representative SLA process, nascent body **402** is first formed from a liquid-monomer starting material. The monomer material in the structural regions of body **402** are the selectively polymerized using a directional laser beam to convert the monomer into structural material SM**1**, during which, the structure of the microfluidic networks is left as monomer material. After this polymerization step, the monomer material is removed by solvent, etc. to realize a device of only structural material SM**1** that includes empty microfluidic networks **404-1** and **404-2**. Microfluidic networks **404-1** and **404-2** are then filled with conductive fluid **416**.

Yet another exemplary fabrication method suitable for use to fabricate muscles and/or muscle fibers disclosed herein is replication molding, in which the final shape of body **402** is separated into multiple layers, each of which can be formed via casting structural material SM**1** into a mold designed for that layer. Each layer is then separated from its mold and the layers are assembled to realize the completed structure of body **402**.

In some embodiments, body **402** and the conductive-fluid-filled microfluidic networks are formed by bulk-printing operations. Such approaches offer the advantage that no sacrificial material must be formed and subsequently removed; therefore, these methods are potentially faster and more efficient. Furthermore, their use offers the potential for simpler designs.

FIG. **7** depicts operations of a second exemplary method suitable for use in fabricating an artificial muscle in accordance with the illustrative embodiment. Like method **500**, method **700** begins with operation **701**, wherein the shapes of microfluidic networks **404-1** and **404-2** are defined.

At operation **702**, sacrificial elements **608-1** and **608-2** are formed in the shapes of microfluidic networks **404-1** and **404-2**, respectively, via 3D printing. Sacrificial elements **608-1** and **608-2** are formed such that they comprise a material that can be removed (etched, melted, etc.) without significant damage to the structural material of body **402**. It should be noted that, when using 3D printing to define the sacrificial elements, no handle substrate is necessary (although one can be used without departing from the scope of the present disclosure).

At operation **703**, body **402** is formed around sacrificial elements **608-1** and **608-2** via 3D printing of structural material SM**1**. As noted above, in the depicted example, structural material SM**1** is PDMS.

During operation **703**, body **402** is formed such that ports **414** are defined during the printing process and a portion of each of sacrificial elements **608-1** and **608-2** is exposed.

In some embodiments, the structure of microfluidic networks **404-1** and **404-2** is written directly into body **402** as it is formed via 3D printing. In such embodiments, operation **702** is not included in method **700**.

Furthermore, in some embodiments, operation **702** forms microfluidic networks **404-1** and **404-2**, which are written directly into body **402** using an electrically conductive material (e.g., a polymer, etc.) that is flexible enough to function as compliant electrodes in electrostatic actuators **424**. In such embodiments, no sacrificial material is needed and the entire muscle fiber can be completed in a single 3D printing process. This can be accomplished using, for example, stereolithography (SLA) 3D printing, as discussed above. At operation **704**, sacrificial elements **608-1** and **608-2** are removed through ports **414** in conventional fashion.

At operation **705**, contact pads **418-1** and **418-2** and traces **420** are formed.

At operation **706**, microfluidic networks **404-1** and **404-2** are filled with fluid **416**, as described above.

At optional operation **707**, ports **414** are sealed to mitigate leakage of fluid **416** out of the microfluidic networks.

At operation **708**, method **700** is completed with the attachment of wires **110** to contact pads **418** to connect fiber **400** to voltage source **108**.

Although the methods presented herein are described with reference to fiber **400**, it should be noted that any method in accordance with the present disclosure can be used to fabricate any artificial muscle or artificial muscle fiber in accordance with the present disclosure.

As noted briefly above, increasing the number of electrostatic actuators **424** connected in series increases the amount of contraction a muscle fiber can realize but does not increase the amount of force it can generate. In the depicted example, each of electrostatic actuators **424** can generate approximately 249 N/cm^2. The capacitive plate area of each electrostatic actuators **424** is only about 1.8×10^{-3} cm^2, however, so each actuator can generate approximately 0.44 N of force. In order to generate an appreciable amount of force, a plurality of fibers **400** can be connected in parallel so that the forces they generate combine, as discussed below and with respect to FIG. **8**.

FIG. **8** depicts a schematic drawing of a perspective view of an artificial muscle fiber bundle in accordance with the present disclosure. Bundle **800** includes fibers **802-1** through **802-8** and coupling members **804-1** and **804-2**.

Fibers **802-1** through **802-8** (referred to, collectively, as fibers **802**), are analogous to fiber **400** described above; however, the bodies of each of fibers **802** are configured such that they follow the paths of their respective microfluidic networks. Fibers **802** are linearly arrayed along the x-direction, as indicated in FIG. **8**.

Each fiber **802** consists of two entwining but non-crossing bodies **806-1** and **806-2** that encase microfluidic networks **808-1** and **808-2**, respectively. In similar fashion to the microfluidic networks of fiber **400**, microfluidic network **808-1** includes a plurality of channels **810**, which are fluidically and electrically connected via manifold **814-1**. Microfluidic network **808-2** includes a plurality of channels **812**, which are fluidically and electrically connected via manifold **814-2**.

As shown in Inset **1**, bodies **806-1** and **806-2** and microfluidic networks **808-1** and **808-2** serpentine within each of planes P**1** and P**2** and periodically transition between the planes via an orthogonal member. As a result, manifold **814-1** has multiple manifold segments **816** in each plane and manifold **814-2** has multiple manifold segments **818** in each plane. Channels **810**, located within the orthogonal members of body **806-1**, run between planes P**1** and P**2** to fluidically couple manifold segments **816**. In similar fashion, channels **812**, located within the orthogonal members of body **806-2**, run between planes P**1** and P**2** to fluidically couple manifold segments **818**. When filled with conductive fluid, channels **810** and **812** form the opposing electrodes of electrostatic actuators **820**.

As will be apparent to one skilled in the art, after reading this Specification, fabrication of the three-dimensional structure of fibers **802** via sacrificial molding requires a slight modification to method **500** described above.

For example, after the formation of first body portion **602-1**, a first layer of sacrificial material is deposited on the top surface of the first body portion and defined in the shape of the manifold segments **816** and **818** that are located in plane P**1**. An intermediate body portion is then formed over the first body portion and the first layer of sacrificial material, where the intermediate body portion has a top surface that defines plane P**2**. Vias are then formed through the intermediate body portion to expose portions of the manifold segments in plane P**1** and the vias are filled with sacrificial material to define the shapes of channels **810** and **812**. A second layer of sacrificial material is then defined the top surfaces of the intermediate body portion and sacrificial-material-filled vias to form the shapes of the manifold segments **816** and **818** that are located in plane P**2**. A top body portion is then formed over the intermediate body portion and second layer of sacrificial material to complete body **802**. Once body **802** is complete, method **500** can continue as described above with the removal of the sacrificial material, formation of ports and interconnect traces, the filling of the microfluidic networks with conductive fluid, and so on.

It should be noted that bundle **800** can be fabricated using any suitable fabrication process, including those discussed above (e.g., direct 3D printing with or without sacrificial material, replication molding, silk-screening, etc.).

Fluidic networks **808-1** of fibers **802** are fluidically and electrically connected at common ports (not shown) formed at each of coupling members **804-1** and **804-2**. In similar fashion, microfluidic networks **808-2** of fibers **802** are also fluidically and electrically connected at different common ports formed at each of coupling members **804-1** and **804-2**. As a result, bundle **800** has only one pathway for filling all of microfluidic networks **808-1** and one separate pathway for filling all of microfluidic networks **808-2**. In addition, this enables the entire bundle to be electrically connected to voltage source **108** (not shown) via only two connection points. Still further, bundle **800** is configured in a binary fluidic architecture such that fluidic bias is mitigated and the electric and fluidic resistance is substantially equal among fibers **800**.

It should be noted that the parallel fiber arrangement of bundle **800** is analogous to a parallel arrangement of M identical springs, which will produce M times the force generated by each spring for the same shared elongation. As a result, bundle **800** generates 8 times the force produced by each fiber **802**.

In accordance with the present disclosure, therefore, an increase in the absolute elongation of bundle **800** can be realized by increasing the number of electrostatic actuators **816** along the length of each of its muscle fibers **802**, while the force generated by bundle **800** can be increased by adding additional muscle fibers **802** and connecting all of the muscle fibers in parallel. Furthermore, additional bundles **800** can be arrayed along both x-axis and z-axis to gain even greater force.

Although bundle **800** includes fibers **802** that are arrayed in only the x-dimension, in some embodiments, fibers **802** are also arrayed in the z-direction to realize a two-dimensional bundle of muscle fibers and greater force-generation capability.

FIG. **9** depicts a schematic drawing of a top view of another artificial muscle in accordance with the present disclosure. Muscle **900** is an artificial sphincter that includes body **902**, within which microfluidic networks **904-1** and **904-2** reside.

Body **902** is a substantially circular plate of dielectric-elastomer-based structural material. Body **902** is analogous to body **402** described above.

Fluidic network **904-1** includes channels **908-1** through **908-8**, (referred to, collectively, as channels **908**), manifold **912-1**, and port **414-1**.

Fluidic network **904-2** includes channels **910-1** through **910-8** (referred to, collectively, as channels **910**), manifold **912-2**, and port **414-2**.

Channels **908** are uniformly spaced and arranged in a radial pattern about center point CP. Channels **908** are fluidically coupled with port **414-1** via manifold **912-1**.

In similar fashion channels **910** are also uniformly spaced and arranged in a radial pattern about center point CP. Channels **910** are fluidically coupled with port **414-2** via manifold **912-2**.

Fluidic networks **904-1** and **904-2** are analogous to microfluidic networks **404-1** and **404-2** described above, and filled with conductive fluid **416**. As a result, each set of adjacent channels **908** and **910** function as the electrodes of an electrostatic actuator **916**, which is analogous to electrostatic actuators **424** discussed above.

In the depicted example, muscle **900** also includes optional aperture **914**, which runs through body **902** and is centered on center point CP. In the nascent state of muscle **900**, aperture **914** has a nascent circumference.

In analogous fashion to electrostatic actuator **424**, when a voltage is applied between contact pads **418-1** and **418-2**, an attractive force is generated between each adjacent set of channels **408** and **410**.

In contrast to the linear force (i.e., tension) collectively generated by electrostatic actuators **424**, however, the compressive force generated by each electrostatic actuator **916** is directed circumferentially such that they collectively generate a constrictive force about center point CP (much like a rubber band around a rolled-up newspaper provides a constrictive force along the circumference of its loop, which squeezes the paper inward while also attempting to shrink the rubber band radially).

In some embodiments, this constrictive force gives rise to a reduction in the circumference of aperture **914**, which can be advantageously used in numerous applications. For example, if aperture **914** were to surround a portion of a fluidic channel, the constrictive force could be used to create a pressure wave in the fluid within the channel. A plurality of muscles **900** operated in such a manner could be used as a peristaltic pump. In some cases, aperture **914** could be closed nearly completely, mimicking the behavior of a biological sphincter muscle. It should be noted that sphincter-like operation can also be achieved using by wrapping a linear muscle fiber (e.g., fiber **400**) in a helix about an element (e.g., a tube, etc.) and actuating it.

In some embodiments, muscle **900** does not include aperture **914** and/or muscle **900** is non-planar, which enables a variety of functions, such as peristaltic propulsion, volumetric compression, and the like.

It should be noted that the microscopic scale of the individual muscle fibers lands itself very well to building super-arrays of muscle fiber bundles that can substantially mimic the anatomic structure of human and animal muscles, locomotion, and/or propulsion systems. For example, ideally, a biomimetic prosthetic limb or biomimetic-powered exoskeletal limb that has a range of motion, degrees of freedom, and intuitive use that approximates the original limb it replaces is highly desirable. Such a prosthetic device is possible within the scope of the present disclosure via arrangements of muscle fibers in orientation, arrangements, and suspension points to the artificial endoskeleton or exoskeleton that are substantially anatomically correct. In other words, muscle fibers and muscles in accordance with the present disclosure lend themselves to virtually "verbatim" biomimetics, with significant advantages over prior-art artificial muscles with respect to fidelity and range of motions, as well as wearability and ease of use. Moreover, fabrication of artificial muscles in accordance with the present disclosure using 3D printing enables low-cost direct printing of monolithic 3D artificial muscles with the exact biomimetic structure necessary, thereby avoiding complex design, assembly, and optimization procedures.

It should be further noted that the ability to substantially mimic a biological muscle system, such as those of fish, etc., enables the application of muscles and/or muscle fibers in accordance with the present disclosure to many important applications, such as biomimetic propulsion of unmanned underwater vehicles (UUV) and unmanned surface vehicles (USV). For example, vehicles that propel themselves by mimicking the motion of the fins of pelagic fish (e.g. sharks, tuna, etc.) have inherent advantages in preventing cavitation of rotary propellers and, therefore, would have acoustic signatures more closely matching biologicals than man-made vehicles. As a result, such propulsion could be provided in an energetically efficient way.

In some applications, muscles and/or muscle fibers in accordance with the present disclosure can be fabricated to match the anatomy of large walking animals, such as ostriches or large cats, thereby offering more optimized land locomotion with different objectives. For example, ostrich-like limbs on ground vehicles offer the potential for maximizing cross-country access and large ground clearance (e.g., for anti-mine and anti-IED (improvised explosive devices) stratagem, etc.). Furthermore, in some applications, large-cat-like limbs offer the potential for high speed and jumping capability in a ground vehicle, which would be highly useful in families of scout vehicles, robots, armored personnel carriers with multiple pairs of limbs, and the like. It can also enable improved protection against enemy fire or explosive devices, since a multi-limbed vehicle would not be totally disabled by the loss of some of its limbs—in contrast to four-wheel ground vehicles that are disabled by the loss of a single wheel, or a tracked vehicle that can be disabled by the loss of one track.

It is to be understood that the disclosure teaches only examples of embodiment in accordance with the present disclosures and that many variations of these embodiments can easily be devised by those skilled in the art after reading this disclosure and that the scope of the present invention is to be determined by the following claims.

What is claimed is:

1. A method for forming an artificial muscle, the method comprising:
- forming a body that comprises a first material that is a dielectric elastomer, wherein the body includes (1) a first microfluidic network that includes a first plurality of channels and a first manifold that is fluidically coupled with each channel of the first plurality thereof and (2) a second microfluidic network that includes a second plurality of channels and a second manifold that is fluidically coupled with each channel of the second plurality thereof, wherein the first microfluidic network and second microfluidic network are arranged such that the channels of the first plurality thereof and the channels of the second plurality thereof are interdigitated and spaced apart;
- filling the first microfluidic network with a second material that is electrically conductive, wherein the second material is selected from the group consisting of a liquid and a gel; and
- filling the second microfluidic network with the second material.

2. The method of claim **1** wherein the body is formed by operations comprising:
- defining a first shape that includes the first microfluidic network;
- defining a second shape that includes the second microfluidic network;
- forming a first sacrificial element having the first shape;
- forming a second sacrificial element having the second shape;
- forming a nascent body that comprises the first material, wherein the nascent body encloses a portion of each of the first and second sacrificial elements; and
- removing the first and second sacrificial elements.

3. The method of claim **2** wherein the nascent body is formed by operations comprising:
- forming a first portion of the nascent body on a substrate, the first portion comprising the first material; and
- forming a second portion of the nascent body over the portion of each of the first and second sacrificial elements, the second portion comprising the first material.

4. The method of claim **1** wherein the body is formed by three-dimensional (3D) printing of the first material.

5. The method of claim **4** wherein the body is formed by operations comprising:
- defining a first shape that includes the first microfluidic network;
- defining a second shape that includes the second microfluidic network;
- forming a first sacrificial element having the first shape, wherein the first sacrificial element is formed by 3D printing of a third material;
- forming a second sacrificial element having the second shape, wherein the second sacrificial element is formed by 3D printing of the third material;
- forming the body such that it encases a portion of each of the first and second sacrificial elements; and
- removing the third material.

6. A method for forming an artificial muscle, the method comprising:
- forming a first sacrificial element having a first shape that is based on a first microfluidic network, the first sacrificial element comprising a first material;
- forming a second sacrificial element having a second shape that is based on a second microfluidic network, the second sacrificial element comprising the first material;
- forming a body such that it encases a portion of each of the first and second sacrificial elements, the body comprising a second material that is resilient;
- removing the first material to define the first and second microfluidic networks within the body; and
- filling each of the first and second microfluidic networks with a third material that is electrically conductive, wherein the third material is selected from the group consisting of a liquid and a gel.

7. The method of claim **6** further electrically each of the first and second microfluidic networks with a voltage source.

8. The method of claim **6** wherein the first and second sacrificial elements are formed by 3D printing of the first material.

9. The method of claim **6** wherein the body is formed by 3D printing of the second material.

10. The method of claim **6** wherein second material is a dielectric elastomer.

11. The method of claim **6** wherein the first microfluidic network includes a first plurality of channels and a first manifold that is fluidically coupled with each channel of the first plurality thereof, and wherein the second microfluidic network includes a second plurality of channels and a second manifold that is fluidically coupled with each channel of the second plurality thereof, and further wherein the first microfluidic network and second microfluidic network are arranged within the body such that the channels of the first plurality thereof and the channels of the second plurality thereof are interdigitated and spaced apart.

12. The method of claim **6** wherein the body is formed by operations comprising:
- forming a first portion of the body on a substrate, the first portion comprising the second material;
- disposing the first and second sacrificial elements on the first portion; and
- forming a second portion of the body over at least some of the first and second sacrificial elements.

13. The method of claim **12** wherein the first material is a photoresist.

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