Sensors & Actuators: A. Physical 340 (2022) 113543

Contents lists available at ScienceDirect

Sensors and Actuators: A. Physical

journal homepage: www.journals.elsevier.com/sensors-and-actuators-a-physical

Scalable microfluidic double-helix weave architecture for wiring of microcapacitor arrays in 3D-printable biomimetic artificial muscles

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ARTICLE INFO

Keywords:
Actuator
Muscle
Biomimetic
Electrostatic
Architecture
Microfluidic

ABSTRACT

Practical artificial muscles are highly desirable in a wide range of applications: acoustically quiet underwater propulsion, exoskeletons, walker robots, prosthetics, and medical augments. 3D-printable microfluidic electrostatic biomimetic artificial muscles in particular hold high promise for low-cost, energy-efficient, high-strength-to-weight-ratio, manufacturable actuator solutions. Their basic design and operational principles have been established. However, there remains a major problem to solve as to how to wire them fluidically and electrically in a scalable, efficient, and practicable fashion. This short communication offers an innovative solution to this very problem. Herein, each muscle fiber is a double helix of microfluidic channels connecting longitudinal arrays of microcapacitor plates of alternating polarity. The fibers are arrayed in the two lateral dimensions to produce muscle fiber bundles that are connected by binary-tree architectures that taper off to only two inputs and two outputs for the entire muscle. This solution ensures full scalability, efficient fluidic loading, simple electrical interface, and resilience to single-point failures. Hence, the offered solution is a major step towards the practical implementation of 3D-printable artificial muscles and their applications.

1. Introduction

Traditional robotic actuation is done via electric motors or pneumatics/hydraulics. Electromagnetic step motors [1] offer precision, use a convenient form of power, and have some capability for miniaturization, making them the usual choice for small robots and prosthetics. However, these motors are in fact electromagnetic (EM) motors, which require a strong magnetic field generated either by strong permanent magnets or solenoids running large currents. Conventional EM motors often choose the latter path and require significant power to operate, while generating excess heat.

Pneumatic systems provide more force in large systems, e.g. construction vehicles, industrial assembly lines, the US Army's Mule walking robot, etc., but they require compressors, can spring leaks, and the output force scales disadvantageously as devices shrink in size. Furthermore, complex fluid motions are difficult to achieve by pneumatics because pressure is typically either on or off, producing choppy motion that might be acceptable in an industrial robot but impractical in exoskeletons, prosthetics, etc.

Due to these limitations, a wide range of applications, such as exoskeletal locomotion, walking robots, biomimetic underwater propulsion, prosthetics, medical servo-assists, and small-scale biomimetic robots, look to alternative actuators, including artificial muscles [2,3]. Perhaps the most promising artificial muscles are Dielectric Elastomer Actuators [4] (DEA), wherein the actuation is a result of the deformation of a polymer (elastomer) slab under the electrostatic force between the charges built on the slab's surfaces, under applied voltage. That force is small macroscopically, but it scales as the inverse square of the separation between the plates. Hence, shrinking the devices to the microscale would gain a disproportionate increase in force. Arraying such devices in 3D would increase both force output and motion distance. However, manufacturing such arrays to the necessary scale by traditional means would be prohibitively expensive and thus impractical.

We recently reported on an innovative alternative solution [5] – artificial muscles based on a combination of microfluidics, electrostatic actuation, liquid electrodes, and 3D-printing. Briefly, 3D printing would be used to build the artificial muscle, wherein microfluidic channels would connect to columns of parallel flat microcavities. When the microchannels are filled with conducting fluid, they would become flexible electrical wiring, while the flat microcavities would become the plates of arrays of microcapacitors connected in parallel, as the plates

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https://doi.org/10.1016/j.sna.2022.113543
Received 24 October 2021; Received in revised form 4 February 2022; Accepted 1 April 2022
Available online 5 April 2022
0924-4247/Published by Elsevier B.V.

would be wired in alternating electric polarity. Applying voltage to the system would make each microcapacitor shrink by a small amount. So, each column of microcapacitors would act as a muscle fiber, wherein the shrinking at the microscale would add up to significant contractive elongation on the macroscale. Fibers would be arranged laterally in parallel to form artificial muscle bundles. The overall result would be a longitudinal contractive force density that would scale as the square of applied voltage and the inverse square of the distance between the microcapacitor plates [5]. COMSOL simulations [5] have predicted up to 33 MPa force density at the current extreme limits of manufacture and materials.

While these muscles boast high promise and utility, a major problem remains to be solved, i.e. how to connect and organize the microcapacitor arrays fluidically and electrically in a scalable fashion that would also allow reliable loading of the liquid/gel electrodes. In this short communication, we offer an innovative solution for the needed large-scale microfluidic architecture. The presented solution is thus a major step to the practical implementation of this type of artificial muscles.

2. Results and discussion

The basic problem of wiring the microcapacitor arrays stems from microfluidic, mechanical, electrical, and scaling restrictions, with the added challenge that all such must be satisfied simultaneously for the architecture to be functional and practicable.

From mechanical perspective, the microcapacitors must be arranged in columns, so that their individual microscopic contractions add up to a macroscopic elongation along the longitudinal direction of the device, to produce the required length of motion during actuation. Furthermore, the individual columns of stacked microcapacitors must be arranged in parallel to the longitudinal axis and arrayed laterally in the remaining two orthogonal dimensions. This ensures that the columns contract in parallel and the generated forces add constructively to output a cumulative force to the outside world.

As a third requirement, the bulk polymer between adjacent columns must be as monolithic and mechanically strong as possible, since it would serve as a tendon equivalent. In biological muscles, the individual fibers are grouped in bundles while their sheathing is made of connective tissue that becomes the tendons by which the muscle attaches to the bones. As the biological muscle fibers contract, they pull on the tendons, which transfer the generated force to the outside world. Similarly, the microcapacitor stacks convey their contraction to the surrounding polymer material, which acts as tendon and transfers a portion of the generated force to the outside world [5]. The tendon would be strongest if there are as few disruptions as possible in the lateral directions. So, any channel serving as wiring inside the structure should be ideally precluded from running perpendicular to the tendons as it would weaken them structurally. The periodicity in the design must reflect that as well.

From electrical perspective, each microcapacitor is a set of two plates of opposite polarity, so each plate must be connected to an outside electrode of the respective polarity. Because the microcapacitors are stacked in columns, this means the polarity must alternate along each column, from plate to adjacent plate. The simplest design to achieve that is two combs kept at opposite polarity and facing each other with their prongs interdigitated.

This would work electrically but produces a fluidic problem: The prongs are dead-end channels which would be difficult to fill with conducting fluid. If the matrix of the polymer is permeable to air (e.g. silicone), this can be done by dead-end priming [6]. While hard 3D-toner resin is less permeable to air than silicone, artificial muscles would have to be printed in soft resin [5], which ought to be more permeable to air than the hard resin. Thus, dead-end priming could still work. A related solution would be a new toner resin chemically designed to be sufficiently air-permeable to allow dead-end priming.

An independent solution would be to design the fluidic matrix based on through-channels only. That would avoid dead ends, while still allowing dead-end priming by closing off the exit while applying pressure at the entrance of the matrix.

To ensure proper evacuation and avoid tendon weakening, the wiring must be done in the longitudinal direction with longitudinal periodicity and no dead ends. This means each plate must be accessed by its connecting input and output channels ideally at opposite corners, e.g. top left and bottom right (e.g. blue in Fig. 1A). Moreover, this must be done twice, since there are two polarities and two subsets of plates in the same column. It stands to reason that if one polarity uses top left and bottom right, then the other must use bottom left and top right (e.g. red in Fig. 1A). This logic derives the unit device of the artificial muscle fiber (Fig. 1A). The unit contains two plates of each polarity as connections must alternate from plate to plate.

The unit device is then arrayed longitudinally to produce the structure of a single muscle fiber (Fig. 1B). Curiously, the resulting weave is a double helix akin to dsDNA. The two polarities never cross but connect to alternating plates along the column. There are only two inputs and two outputs regardless of the length of the fiber and the number of microcapacitor units within it. Also, each of the helices is a single channel, greatly facilitating reliable filling with conducting fluid.

Mechanical considerations dictate a limit to the aspect ratio between plate width and plate thickness. Excessive aspect ratio can lead to plate collapse and improper filling. That would degrade electrical performance since it would decrease the electrode area and capacitance, leading to diminished charge and force at the same voltage. Furthermore, force transfer to the tendon ought to be more efficient with smaller plates. As a result, the optimal design must be a bundle of a large number of thin parallel fibers arrayed laterally. Also, such fibers must be wired together in a scalable fashion. The solution is again microfluidic and based on a binary tree [6,7]. Fig. 1C shows a pair of fibers wired in a binary fashion. The symmetry of the design ensures theoretically equal fluidic resistance along both pathways for each polarity.

In a printed device, fabrication artifacts would cause asymmetries that would imbalance the fluidic resistances of different pathways. In addition, if the fluid does not wet the channel surface well, surface tension would offer high resistance and right-angle corners would be difficult to evacuate. Transport would be difficult until a pathway is entirely filled with fluid from entry to exit. Then any further fluid pushed into the device would flow through that pathway, shunting alternative higher-resistance pathways blocked by air bubbles. This is a typical scenario in microfluidics of aqueous fluids in hydrophobic channels [6].

Thankfully, our application is not limited to aqueous fluids, so the simple solution is to use hydrophobic fluids. Then surface tension becomes an ally and would help fill the channels quickly and efficiently. Any imbalance in resistance from small asymmetry in the geometry should not matter. Finally, right-angle corners can be avoided by using rounded turns and by replacing T-splits with Y-splits.

If any bubbles do form, they should be few and isolated. They could be removed by dead-end priming as described above. Another method is to cool the device to increase the solubility of air into the fluid and so help remove bubbles by dissolving them. Finally, if any bubble remains, it is likely to be limited to a particular microcapacitor, which would have negligible effect on the overall device.

Simultaneously, hydrophobic fluids can also be made electrically conductive by adding e.g. silver microparticles or graphene nanoparticles, and might also be cured in place if necessary. For example, Electron Microscopy Sciences (Hatfield, PA 19440) offers a curable silver acrylic with volume conductivity of 5×10^5 S/m (compared to sea water at $\sim$5 S/m).

The structure in Fig. 1C is ultimately a 2D array. This needs to be expanded in a scalable fashion to 3D. One of the ways to do so is to expand the array by binary tree in 2D first, to a number N of parallel fibers lying in the same horizontal plane, where N is equal to a power of

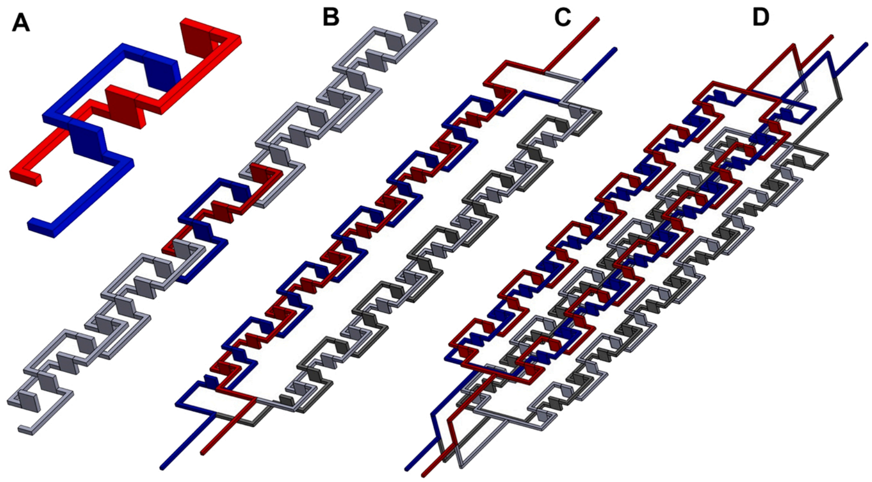

Fig. 1.. Schematic of Scalable Weave Architecture for 3D-printable artificial muscles. (A) Basic unit of the weave in the two electrical polarities (red and blue). (B) A single artificial muscle fiber generated through longitudinal arraying of the basic unit. (C) A two-dimensional binary fiber bundle with two common inputs and outputs. (D) A three-dimensional fiber bundle of 2×2 fibers and two common inputs and outputs. This bundle can be used as the basic 3D unit together with binary-tree wiring to allow for arbitrary size of the artificial muscle array while retaining only two global inputs and outputs. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2. Then, the plane can be arrayed vertically and connected by an analogous but vertical binary tree, to produce a NxN fiber bundle. Another way to go to 3D arrays is shown in Fig. 1D. The original horizontal pair of fibers is arrayed vertically and connected by binary tree, producing a 2×2 fiber bundle. That bundle itself can then be arrayed first vertically then horizontally etc., in a binary fashion.

Because the output force density would scale as the inverse square of the plate separation within each microcapacitor, it would pay to make that separation as small as possible while still avoiding dielectric breakdown. Hence, in practice the plates would be arranged more densely than depicted in Fig. 1 wherein they are spaced out to improve visualization. Simultaneously, the connecting channels in the double helix would be shorter than depicted in Fig. 1, thereby offering less fluidic resistance than the figure would suggest. Finally, the coupling channels in the binary tree architecture would grow wider as they rise in hierarchy, to minimize the rise of overall fluidic resistance of the structure and help with efficient loading.

3. Conclusions

Artificial muscles based on microfluidics, arrayed micro-capacitors, electrostatic forces, and 3D printing offer a great promise for a wide range of applications. Making those a reality requires a complex wiring scheme that must simultaneously satisfy a list of mechanical, microfluidic, electrical, and scaling requirements. Herein we have presented such a solution – an innovative practicable scalable double-helix weave architecture that satisfies all requirements. Hence, the presented work is a major development towards practical implementation of artificial muscles.

CRediT authorship contribution statement

M. Coltelli: Software, Drafting, Visualization. **E. Kartalov:** Conceptualization, Visualization, Supervision, Project administration, Funding acquisition.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

This work was supported by the ONR CRUSER Program at NPS and ONR grant N0001421WX01682.

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