# Simulations of 3D-Printable biomimetic artificial muscles based on microfluidic microcapacitors for exoskeletal actuation and stealthy underwater propulsion 

Michelangelo A. Coltelli ${ }^{\text {a }}$, Jeffrey Catterlin ${ }^{\text {a }}$, Axel Scherer ${ }^{\text {b }}$, Emil P. Kartalov ${ }^{\text {a,* }}$<br>${ }^{a}$ Physics Department, Naval Postgraduate School, 833 Dyer Road, Monterey, CA, 93943, United States<br>${ }^{\mathrm{b}}$ Electrical Engineering Dept, California Institute of Technology, 1200 E California Blvd, Pasadena, CA, 91125, United States

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#### Abstract

Practical artificial muscles are highly desirable in a wide range of applications, including strength augmentation in military exoskeletons, medical prosthetics for amputees, locomotion boosters for geriatric and handicapped patients, walker robots, and acoustically quiet underwater propulsion systems. So, artificial muscles have been a subject of active research through a variety of approaches, e.g. electromagnetics, pneumatics, hydraulics, thermal expansion/contraction, piezoelectrics, shape memory alloys, and electrically active polymers. Herein we propose a new approach based on a combination of microfluidics, 3D printing/additive manufacturing (AM), and electrostatic actuation. Back-of-the-envelope calculations promise 33 MPa generated stress under feasible conditions. Respective integral architectures are described. Individual devices and $2 \times 2$ arrays are analyzed through COMSOL simulations. Simulations predict $10-20 \%$ strain, which is ample for most applications. Parameter sweeps in the simulations offer quantitative insights into optimal values for maximizing the output force density. The simulations demonstrate that alternative wiring schemes produce muscle or counter-muscle behavior of the same arrays, offering novel capabilities. The proposed technology promises a major impact on a range of important applications, e.g. exoskeletons, prosthetics, walker vehicles, and stealthy undersea propulsion. Published by Elsevier B.V.


## 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 really 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 output less force when scaled down for use in compact systems. Furthermore, complex fluid motions are difficult to achieve by pneumatics because pressure is typically either on or

[^0]off, producing jerky choppy motion that may be acceptable in an industrial robot but impractical in exoskeletons, prosthetics, etc.

Due to these limitations, a wide range of applications requiring actuation, such as exoskeletal locomotion, walking robots, biomimetic underwater propulsion, prosthetics, medical servoassists, and small-scale biomimetic robots, look to different actuation systems as a potential solution, including artificial muscles [2,3]. Artificial muscles can be organized in several large groups: piezoelectrics, pneumatic artificial muscles (PAM), thermal actuators, and electroactive polymers (EAP).

Piezoelectric actuators [4-6] offer large forces in small devices at low voltages but the range of motion is very small. Devices are often stacked to mitigate that drawback. For example, such stacks are successfully used in the beam control circuitry of atomic force microscopes. However, the cost of individual devices and manufacturing difficulties severely limit the size of practical stacks, with the resulting overall elongation still being too small for typical artificial muscle applications.

PAMs, e.g. McKibben muscles [7], cloth muscles [8], and RIPAs [9], employ a flexible bladder structure enmeshed in braided, crisscrossed, or wound fibers. As the bladder is filled with air, it deforms
and displaces the fibers, outputting force. While possessing advantages in compactness and force output compared to piston system, PAMs use the same basic principles as hydraulics/pneumatics and thus suffer from the same basic limitations in unfavorable scaling and control issues.

Thermal actuation has also been proposed, e.g. with anisotropic materials that curl up with a temperature change, producing torsional artificial muscles $[10,11]$ and SMA (shape-memory alloys) muscles $[12,13]$. Thermal expansion and contraction can generate high forces, but heat transfer severely limits the thermal actuators' response speed and cycling frequency. As a result, such actuators are not practical for most propulsion applications.

EAPs [14-18] change shape under the influence of an applied electric field. They are considered closest to the biological muscles among all the above-mentioned approaches. They avoid the use of magnetic fields and thus avoid the concomitant limitations. However, EAP actuators are typically complicated heterogeneous materials that are difficult to fabricate and suffer from low reproducibility, very low efficiency, and low durability. As a result, they have proven very difficult to manufacture to the standards and at the scale and price-point required by practical applications.

Dielectric Elastomer Actuators (DEA) [19] are a particular subclass of EAP, 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 should increase both force output and motion distance. However, manufacturing such arrays from traditional materials (e.g. metal electrodes and polymer dielectrics) by traditional manufacturing means (e.g. photolithography) to the necessary scale is impractical for reasons similar to the difficulties experienced with the piezoelectric and EAP approaches.

Based on the above analysis, the need for practical artificial muscles remains unmet. An alternative approach is offered herein, as a potential solution to this problem. It combines electrostatic actuation with microfluidics, liquid electrodes, and additive manufacturing (AM). Calculations indicate that the proposed devices could generate up to 33 MPa stress under the current extreme limits of manufacture and materials. COMSOL simulations of both individual devices and $2 \times 2$ arrays offer strong evidence for the feasibility of the proposed techniques. Parameter sweeps of the simulations offer insights into the behavior of the proposed devices as well as suggest optimal values maximizing device performance. These results allow for efficient design to maximize the generated output force density.

The $2 \times 2$ array simulations indicate different behaviors of the same devices based on different wiring schemes. These alternatives suggest muscle-like and counter-muscle-like actuations, leading to novel capabilities and applications.

Finally, the described features combined with the architectural flexibility of AM promise novel biomimetic actuators with the motion range, complexity, and dexterity potentially approaching those of biological muscles. These would be welcomed in a wide range of applications, e.g. high-fidelity prosthetics, ergonomic agile exoskeletons, all-terrain walkers, intuitive robotic controls, and stealthy undersea propulsion systems.

## 2. Materials and methods

### 2.1. Software

We used COMSOL to generate the simulation models and parameter sweeps presented herein. Specific settings follow.


Fig. 1. The basic device of the COMSOL simulation. Two rectangular cavities (blue and red) are arranged in parallel and filled with conducting material inside the bulk dielectric (orange). Critical parameters are indicated: separation $D$ between the electrodes; tendon width e ; dielectric thickness c above and below the electrodes.

### 2.2. Structures

The basic structures of the model included the overall bulk material polymer/dielectric (orange) and two electrode chambers (red and blue) situated in parallel within the dielectric, as drawn in Fig. 1. The three structures were joined together using the union function.

### 2.3. Material inputs

Silicone (polydimethylsiloxane (PDMS)) was assigned to the bulk dielectric. Liquid water was assigned to the electrode chambers.

### 2.4. Node use

Three physics nodes were used: electrostatics, solid mechanics, and moving mesh. The solid mechanics node was applied to the bulk dielectric. The moving mesh node was applied to the electrode chambers. The electrostatics node was applied to all domains. Within the solid mechanics node, a fixed boundary node was applied to one side of the bulk dielectric that is parallel to the electrode plates, while the opposite side was allowed to deform. A boundary load was applied to all surfaces of the model with the outputs for each component force being equated to their respective Maxwell Upward Surface Tension Equations. The boundary node is key to the simulation working properly, because it connects the physics nodes and allows for capacitive force reaction in the simulation.

### 2.5. Voltage setting

Within the electrostatic node, a potential of 3000 V was applied to all surfaces of one of the electrodes and a ground potential was applied to all surfaces of the other electrode.

### 2.6. Derived values

The total output force of the device was calculated as the surface integral of the Von Mises Stress along the unconstrained outer sur-


Fig. 2. Structure of proposed artificial muscles. (left) A single fiber consists of a stack of micro-capacitors defined by microfluidic channels inside a polymer. The polymer serves as the dielectric (orange), while the electrodes are channels filled with conducting fluid or gel (blue and red). Applying voltage to the two electrodes would generate force contracting the fiber. (right) Geometry controls the direction of the generated force. Shown here is one geometry for a sphincter-like muscle, e.g. for peristaltic propulsion of UUVs.
face of the device, opposite to the boundary-constrained surface. The total deformed surface area was calculated as the surface integral of the area of the unconstrained outer surface of the device, opposite the boundary-constrained surface. The average output force density was calculated by dividing the total output force by the total deformed surface area.

## 3. Results and discussion

### 3.1. Basic ideas

Electrostatic actuation is based on the fact that the plates of a charged capacitor exert an attractive force on each other. That force is small for typical macroscale capacitors. However, the force is inversely proportional to the separation between the plate. Thus, miniaturizing the capacitors while arraying them in 3D structures should produce significant forces. Here is a back-of-the-envelope estimate for these forces.

The unit device can be thought of as a parallel-plate capacitor with ignored edge effects. From basic electrostatics, the electric field E generated in vacuum by one of the plates is
$E=\frac{\sigma}{2 \varepsilon_{0}}=\frac{Q}{2 \varepsilon_{0} A}$
wherein $\sigma$ is the surface charge density, Q is the charge on the plate, and A is the area of the plate. This electric field would act on the charges in the other plate and produce a total force $F$ on them given by:
$F=Q E=\frac{Q^{2}}{2 \varepsilon_{0} A}$
In the presence of a linear dielectric between the plates, this result gets adjusted by the respective dielectric constant $\varepsilon_{r}$, so:
$F=Q E=\frac{Q^{2}}{2 \varepsilon_{0} \varepsilon_{\mathrm{r}} A}$
On the other hand, the capacitance of the parallel-plate capacitor filled with a linear dielectric is given by
$C=\frac{Q}{V}=\frac{\varepsilon_{0} \varepsilon_{r} A}{D}$
wherein V is the applied voltage and D is the distance between the plates. Then
$F=\frac{Q^{2}}{2 \varepsilon_{0} \varepsilon_{r} A}=\frac{1}{2 \varepsilon_{0} \varepsilon_{r} A}\left(\frac{\varepsilon_{0} \varepsilon_{r} A V}{D}\right)^{2}=\frac{\varepsilon_{0} \varepsilon_{r} A V^{2}}{2 D^{2}}$
Since this estimate is for the lattice unit device in an array, the more relevant parameter here is the force density $f$, defined as the generated force per unit cross-sectional area, i.e.
$f=\frac{F}{A}=\frac{\varepsilon_{0} \varepsilon_{r} A V^{2}}{2 D^{2} A}=\frac{\varepsilon_{0} \varepsilon_{r} V^{2}}{2 D^{2}}$
Clearly, minimizing $D$ would maximize $f$. Then manufacturability points to AM. Hence, the limit on D would stem from the best resolution of AM in acceptable materials. AM includes many fabrication methods, of which stereolithography (SLA) can yield both high resolution and functional product. The current resolution of top-of-the-line SLA printers is $10 \mu \mathrm{~m}$.

Next, the allowed voltages are limited by the breach voltage of typical materials. For example, the breach voltage of some silicones is $\sim 600 \mathrm{~V} / \mu \mathrm{m}$, it should be possible to set V to 5 kV for D $=10$ microns and still avoid breaching the dielectric. Finally, the dielectric constant for silicone is typically around 3 . So

$$
\begin{aligned}
& f_{\max } \cong \frac{\left(8.85 \times 10^{-12} \frac{\mathrm{~F}}{\mathrm{~m}}\right) \times 3 \times(5000 \mathrm{~V})^{2}}{2 \times\left(10 \times 10^{-6} \mathrm{~m}\right)^{2}} \cong 3.3 \\
& \quad \times 10^{7} \frac{\mathrm{~N}}{\mathrm{~m}^{2}} \cong 33 \mathrm{MPa} \cong 4,815 \mathrm{psi}
\end{aligned}
$$

This estimate is a high upper bound, based on frontier values in SLA and dielectric breach fields. Even if more conservative values result in 10x reduction in force density, the system would still generate hundreds of pounds of force per square inch. This should be more than sufficient for most artificial muscle applications.

The electrostatic actuation idea itself is not new. However, as far as we know, we are the first to propose combining it with microfluidics and with AM. The basic idea is to AM-fabricate microfluidic devices in such a way that the channels would form wiring when filled with conducting fluid or gel, while the bulk of the material would serve both as the dielectric and as the mechanical medium of force transfer. This approach offers manufacturability, scalability, monolithic fabrication, parallelization, easy addressability, sufficient motion distance and force output, and biomimetic architectural flexibility. As a result, the proposed approach may be the only truly practical way of building artificial muscles.

### 3.2. Basic architectures

A simple architecture exemplifying the proposed idea is shown in Fig. 2A. The device is essentially an artificial muscle fiber. The dielectric material (orange) is built to include microchannels connected in two systems in a comb-like arrangement. The microchannels are subsequently filled with conducting fluid or gel. Then voltage is applied between the two systems. The shown geometry produces an array of microcapacitors among the alternating prongs of the interdigitated combs. Each microcapacitor generates an attractive force between its plates that contracts the dielectric pad between the plates of the respective microcapacitor. That contraction is transferred to the surrounding material. As a result, the overall structure contracts along its longitudinal axis.

Each microcapacitor acts as a contracting spring, while the array acts as a chain of springs connected in series. The device would shrink longitudinally and apply a force to the outside world equal to the force generated in each microcapacitor. Each capacitor would shrink by a small distance but the overall device would shrink by the cumulative distance and by the same percentage. So, longitudinal
arraying increases the range of motion. The overall structure acts as a muscle fiber.

Next, the fiber is arrayed along the other two dimensions, i.e. laterally. The arrayed fibers then act as springs connected in parallel and output force equal to the sum of the forces contributed by the individual fibers. So, lateral arraying gains force. Thus, the fibers would act as a muscle fiber bundle, in direct analogy to human anatomy. This view also shows why the force density is the correct parameter to calculate, as producing a desired force output is simply a matter of multiplying the force density by enough crosssectional area.

The microscopic scale of the individual fiber allows an unprecedented flexibility in designing truly biomimetic architectures of force generation. For example, to produce a peristaltic pumping structure, it would be sufficient to organize the fibers concentrically, e.g. as shown in Fig. 2B.

### 3.3. Simulations - general approach and conventions

We ran simulations in COMSOL to study the projected behavior of the basic versions of such actuators with the objective of producing useful predictions for the optimal values of their geometric parameters before the prototyping of physical devices takes place. To do so, we generated several studies, performed parametric sweeps, and plotted the output force density as a function of the varied parameter.

To limit the volume of the parametric space, we adopted certain conventions. The non-flexed lateral size of the electrode plates was kept at $400 \times 400 \mu \mathrm{~m}$. The non-flexed plate thickness was kept at $100 \mu \mathrm{~m}$ for each plate. The non-flexed separation between paired plates within the same microcapacitor was kept at $100 \mu \mathrm{~m}$. While AM resolution approaches $10 \mu \mathrm{~m}$, we picked $100 \mu \mathrm{~m}$ to gain easier prototyping, at the cost of lower force output.

The voltage applied between the electrodes was kept at 3000 V. This ensures that even if the electrically induced deformation would significantly shrink the distance between the plates of the microcapacitor, the resulting field would still be well below the typical dielectric breach value of several hundred volts per micron. We also applied a fixed boundary condition to the lower outer surface of the device, parallel to the electrode plates. This significantly simplified the calculations for COMSOL while still producing valid results.

### 3.4. Simulations of a single microcapacitor

The simplest device to study is a single microcapacitor embedded in surrounding bulk dielectric material, as illustrated in Fig. 1. Fig. 3A shows the COMSOL simulation results when the structure flexed under 3000 V applied between the plates. The imposed boundary conditions fixed the bottom surface of the device to remain flat. A heat map was used to indicate the stress as calculated by COMSOL.

The simulation demonstrates an interesting beneficial feature of the system. As the material flexes under the applied voltage, the distance $D$ between the plates roughly halves, in essence improving the achieved effective fabrication resolution by a factor of two. Due to the inverse square dependence of the plate force on D , the electrostatic force between the plates would roughly quadruple. This means that the limit on the achievable output force density would be roughly quadruple as well, compared to the limit set by the SLA resolution of non-flexed structures. Thus, the system has a reactive feature that dramatically improves the performance of the overall device as contraction increases. While this effect is expected from first principles, the presented COMSOL simulations confirm it and estimate it quantitatively.

The actuation process can be viewed into two parts: force generation and force transfer. The force is proportional to the total area of the microcapacitor plates. Hence, maximizing the plate share of the device cross-section area would maximize the generated force. On the other hand, force transfer is more efficient when there is more material on the lateral sides of the plates, likely with diminishing returns beyond a certain thickness. The biological analogy is muscles and tendons, respectively. Both need to be present for effectual actuation.

As both generation and transfer compete for the available crosssection area, while they work together to output macroscopic force, there must be a golden point of optimal tradeoff that maximizes the outputted force density. Determining that point ought to be highly beneficial to device design, as it would bring the prototype parameters close to the optimal values and thus minimize the scope of the subsequent experimental optimizations.

Hence, a parameter sweep was conducted over the width of the bulk material region surrounding the plates along both horizontal axes in Fig. 3A. This parameter "e" was swept from $100 \mu \mathrm{~m}$ to 400 $\mu \mathrm{m}$, in steps of $10 \mu \mathrm{~m}$. This kept the muscle area constant while the tendon area was allowed to increase in steps. Simultaneously, the thickness of the bulk polymer between the outer top surface of the device and the top plate was the kept at $\mathrm{c}=50 \mu \mathrm{~m}$, the same as the thickness between the bottom surface of the device and the bottom plate. The value for c was selected as half the conventional plate separation, to allow for efficient subsequent arraying of the basic device. For every value of e in the swept set, the simulation calculated the output force by integrating the stress over the top surface of the device, and then calculated the force density by dividing that force by the total area of the top surface. The results are plotted in Fig. 3B.

As expected, increasing tendon width increased the output force, while the force generation area was kept constant. This confirms that force transfer improves with e. Also, there is a saturation point, perhaps around $\mathrm{e}=310 \mu \mathrm{~m}$. As expected from the tradeoff between generation and transfer, the sweep showed that the outputted force density indeed reaches a peak, specifically at around e $=220 \mu \mathrm{~m}$. Both identified values of e (transfer saturation and peak force density) are important to know in future design.

The simulation indicates a peak force density $\mathrm{f}=1.44 \mathrm{kPa}$. The original back-of-the-envelope calculations were done based on 10 $\mu \mathrm{m}$ plate separation and applied voltage of 5 kV , to estimate an upper bound. On the other hand, the simulation was based on a more conservative choice of input parameters ( $100 \mu \mathrm{~m}$ and 3 kV ), to facilitate prototyping and experimental testing. So, the calculation must be redone:
$f_{\text {sim }} \cong \frac{\left(8.85 \times 10^{-12} \frac{\mathrm{~F}}{\mathrm{~m}}\right) \times 3 \times(3000 \mathrm{~V})^{2}}{2 \times\left(100 \times 10^{-6} \mathrm{~m}\right)^{2}} \cong 11948 \frac{\mathrm{~N}}{\mathrm{~m}^{2}} \cong 12 \mathrm{kPa}$
The above assumes $100 \%$ muscle. At $\mathrm{e}=220 \mu \mathrm{~m}$, only about a fifth of the area is muscle. Hence, we should expect only a fifth of the above value, or $\sim 2.4 \mathrm{kPa}$. That is less than a factor of 2 away from the COMSOL simulation result. Considering the roughness of the back-of-the-envelope estimate, this is a reasonable agreement.

In the above simulation, the thickness " c " of the bulk material above the top plate and below the bottom plate was set at $50 \mu$ m . That value was picked as it is optimal for efficient longitudinal arraying with $100 \mu \mathrm{~m}$ plate separation. However, it still remained to be determined what that setting meant for force output. Hence, c was swept next from $50 \mu \mathrm{~m}$ to $400 \mu \mathrm{~m}$, while keeping $\mathrm{e}=220$ $\mu \mathrm{m}$ (the apparent optimal value).

The results of the c sweep are shown in Fig. 3C. The force density starts at $\sim 1.44 \mathrm{kPa}$ at $\mathrm{c}=50 \mu \mathrm{~m}$, then rapidly declines, then asymptotically settles at $\sim 400 \mathrm{~Pa}$ for large c . This makes sense as increasing c makes the material thicker above and below the micro-


Fig. 3. Simulations of Single Microcapacitor. (A) COMSOL simulation result for a single microcapacitor flexing under 3000 V applied between the plates. Imposed boundary conditions fix the bottom surface flat. The stress is indicated by a heat map. Plate separation is roughly halved in the flexed state, resulting in roughly quadrupling the force between the plates compared to the non-flexed state. (B) Sweeping side thickness e in COMSOL shows that the force (red) saturates, while the force density (blue) peaks. (C) Sweeping the thickness c of the material above and below the microcapacitor shows maximal force density is achieved for $\mathrm{c}=50 \mu \mathrm{~m}$, which is also the optimal value for arraying.
capacitor. So, the same force applied to a thicker slab will result in less deformation. Less deformation means the flexed plates narrow the plate separation less, which means less electrostatic force is generated at the same voltage. Fig. $3 \mathbf{A}$ shows that at $\mathrm{c}=50 \mu \mathrm{~m}$, $D$ is roughly halved from its non-flexed value, so (as the electrostatic force is inverse quadratic in D ) the force should be roughly quadrupled with respect to its value for non-flexed D. Indeed, the maximal and asymptotic values of the force density in the c sweep are roughly 4 x apart, i.e. consistent with small deformations at large
c. The conclusion is that $\mathrm{c}=50 \mu \mathrm{~m}$ is the optimal value for force density maximization in this geometry.

### 3.5. Simulations with $2 \times 2$ array of microcapacitors

The next logical step was to array 4 microcapacitors in a $2 \times 2$ formation, in the plane parallel to the plates. In such a structure, there are two obvious choices for polarity: Planar) all top plates have the same polarity and all the bottom plates have the oppo-
site polarity, resulting in a planar alternation pattern; Checkered) alternate polarity in a checkered formation, e.g. bottom distal left, bottom proximal right, top proximal left, and top distal right plate have all the same polarity, while the other four all hold opposite polarity.

Fig. 4 shows the results of the simulation for Planar (A) and Checkered (B) configuration of polarity, with the distance $h$ between the microcapacitors set to $60 \mu \mathrm{~m}$ along both horizontal axes. Based on the previous simulations, the side width and vertical slab width were set to their apparent optimal values of $\mathrm{e}=220 \mu$ m and $\mathrm{c}=50 \mu \mathrm{~m}$.

In the Planar configuration (Fig. 4A), there is noticeably less bulging of the plates in the inward area (i.e. between capacitors) than in the corresponding area of the same capacitors on the outward side of the array. Simultaneously, that outward side shows similar behavior to the one in Fig. 3A. This suggests an unexpected benefit to the arraying in the horizontal plane with Planar wiring - the neighboring capacitors seem to decrease each other's deformation around the edges. That would likely help flatten the non-linear response of the device to applied voltage and ultimately offer easier control. Finally, the array behaves as desired in the vertical direction, showing significant contraction. This suggests that when arrayed laterally, the planar formation would produce correct behavior of the overall device at the macroscale.

With Checkered wiring (Fig. 4B), there is a large bulge in the center due to the attraction between neighboring plates. This bulge is opposite to the desired behavior as it would produce expansion instead of contraction in the longitudinal direction of the muscle fiber. However, while undesirable for contractive actuation, this phenomenon may potentially produce expansive actuation instead, i.e. a form of counter-muscle.

The $2 \times 2$ array simulations were repeated to parameter sweep h from $50 \mu \mathrm{~m}$ to $400 \mu \mathrm{~m}$ in steps of $10 \mu \mathrm{~m}$. As mentioned above, $h$ is the horizontal distance between neighboring microcapacitors in the array. The interchangeability of the two horizontal axes suggested that the two distances would be kept equal to the same h , while sweeping $h$ itself.

The results of the sweep are shown in Fig. 5. As expected, the force densities of both configurations decrease with increasing $h$, since a decreasing percentage of the total area is devoted to force generation. The decrease in force density is far steeper in the checkered configuration since in its case the deformations along the horizontal axes are far more pronounced. In both configurations, force density is maximal for the minimal $\mathrm{h}=50 \mu \mathrm{~m}$.

If two microcapacitors maximize output force density for $\mathrm{h}=50$ $\mu \mathrm{m}$, it stands to reason that that ought to be the chosen spacing to use in further arraying. The maximal output force density increases from 1.44 kPa for the single microcapacitor, to 1.79 kPa for the 2 $\times 2$ array. That suggests further increase in output force density as more microcapacitors are arrayed laterally.

### 3.6. Analysis of range of motion

The adult human bicep shrinks by roughly 10-15 \% on full contraction. Such strain cannot be achieved by relatively hard materials such as silicon, metals, or piezo-electrics; only softer materials can do it. In a sense, we can move around as we do only because our muscles are mostly water. To produce life-like agility and range of motion, artificial muscles must offer comparable elongation and strain.

The single micro-capacitor simulation (Fig. 3A) produced $\sim 50 \mu$ m contraction over a $400 \mu \mathrm{~m}$ starting thickness, or $\sim 12.5 \%$ strain. The simulation of a $2 \times 2$ array with planar wiring (Fig. 4A) produced $\sim 40 \mu \mathrm{~m}$ contraction over a $400 \mu \mathrm{~m}$ starting thickness, or $\sim 10 \%$ strain. The simulation of a $2 \times 2$ array with checkered wiring (Fig. 4B) produced $\sim 80 \mu \mathrm{~m}$ expansion over a $400 \mu \mathrm{~m}$ starting
thickness, or $\sim 20 \%$ strain. Hence, the strain results from the proposed devices are comparable to the human bicep strain. This is evidence of the applicability of the proposed technology.

### 3.7. Future work

The size of the simulated array must be increase in both the longitudinal and lateral axes. The parameter e would then become less important as an edge effect of decreasing overall impact. Parameter $h$ is likely optimized at half the plate thickness. Parameter $h$ will remain important as the array grows in size. Simulations of larger arrays will need parameter sweeps in $h$.

Symmetry arguments suggest that the boundary conditions on the bottom of the device will remain valid and useful regardless of the size of the array. That boundary surface can be regarded as the plane of mirror symmetry passing through the middle of an actual macroscopic device. The middle should not bend at all, since symmetry dictates it should experience equal and opposite forces along the longitudinal axis at every point of the surface.

As the simulated array grows in size, the calculated maximal force density in each h sweep ought to converge to a set number. That number should be a good estimate of the force density output of a macroscopic prototype of the same geometry and dimensions. Once that force density is determined, the cross-sectional area can be calculated as necessary to achieve a desired force output.

Finally, improvements of fabrication capability would necessitate changes in the picked conventional parameters and iteration of the parameter sweeps to recalculate the optimal values before prototyping.

### 3.8. Potential applications

### 3.8.1. Biomimetic reactive propulsion

The overall structure in Fig. 2B would shrink concentrically like a bladder, similar to the way squids and octopi expel water from their body cavity to achieve reactive propulsion. This may offer a biomimetic method of propulsion for unmanned underwater vehicles (UUV). Such propulsion would produce no cavitation and would look and sound like a biological, thereby offering stealth.

### 3.8.2. Biomimetic fin propulsion

Pelagic fish propel themselves to high velocity by the slow motion of a large back fin that displaces large amounts of water. As the individual points on the fin do not travel fast enough, they do not cavitate like traditional high-speed propellers. If artificial muscles are constructed appropriately and installed in UUV's following the anatomy of pelagic fish, the UUV's would move and sound like a biological of similar size. The result would be stealthy propulsion for the UUV's.

### 3.8.3. Acoustic translucence

Replacing electromagnetic motors with polymer artificial muscles would decrease the effective density of the propulsion module from around $9 \mathrm{~g} / \mathrm{cm}^{3}$ to around $1 \mathrm{~g} / \mathrm{cm}^{3}$. If this is combined with switching to polymer or plastic hull, the same should be true for the overall craft. Since the intensity of reflected sonic waves depends on the density mismatch between water and the material of the craft, the switch should allow for sonar beams to pass through the craft, for the most part. As a result, much less energy is reflected back towards the source. The result is an acoustically translucent craft that appears to sonar to be a biological of similar size.

### 3.8.4. Rotational actuation

In principle, the muscle fibers can also be arranged in macroscopic helical structures akin to the human forearm muscles that


Fig. 4. Simulations of a $2 \times 2$ Microcapacitor Array. COMSOL simulations for a $2 \times 2$ array of microcapacitors flexing under 3000 V applied voltage, with horizontal separations set to $\mathrm{h}=60 \mu \mathrm{~m}$. Imposed boundary conditions fix the bottom surface flat. The stress is indicated by a heat map. Side thickness is set to $\mathrm{e}=220 \mu \mathrm{~m}$, while horizontal slab thickness is set to $\mathrm{c}=50 \mu \mathrm{~m}$. (A) Result for planar configuration (planes of alternating polarity) shows correct behavior for contractive actuation. (B) Result for checkered pattern of polarity shows a vertical bulge that is likely to interfere with vertical arraying and macroscale function of contractive actuation. However, the bulge may be usable for expansive actuation, i.e a form of counter-muscle.
allow for axial rotation of the wrist with respect to the elbow. Contraction of a helical fiber bundle will produce a torque that can be used for rotational actuation.

### 3.8.5. Exoskeletal locomotion

Modern personal armor offers effective protection at the price of high weight. A modern infantryman with combat load approaching 90 lbs tends to sacrifice protection to retain mobility. The solution can be an armored self-propelled exoskeleton ("powered armor"). Artificial muscles offer a pathway to that goal. Their architectural
flexibility means they can be built to follow human anatomy explicitly, thereby offering a similar range of motion to the exoskeleton and highly intuitive efficient controls to the wearer. Essentially, the artificial muscles outside could mimic the motion of the human muscles within. That would simplify training, increase precision, decrease fatigue, and allow for complex motions and combat techniques that would be inaccessible by other types of actuators. Once developed, powered armor would revolutionize ground combat particularly in dense terrain, urban environments, ship boarding, and breach actions. The same locomotion technology would also


Fig. 5. Force Densities of Polarity Configurations of a $2 \times 2$ Microcapacitor Array. COMSOL parameter sweeps of the horizontal separation h between microcapacitors in a 2 $\times 2$ array. The force densities of checkered configuration (blue) and planar configuration (red) of polarities both decrease with h , as expected. Maximal values are achieved for $\mathrm{h}=50 \mu \mathrm{~m}$.
be useful in servo-assists to the elderly and incapacitated, and as a fatigue-reducer to lightly equipped troops and physical workers.

### 3.8.6. Ground walker vehicles

Mines and improvised explosive devices (IEDs) pose a persistent threat to ground vehicles. Typical solutions involve heavily armored V-shaped bottom and increased ground clearance to reduce the damage effects. However, that approach produces heavy expensive vehicles. Artificial muscles may offer an alternative solution. A ground walker vehicle (e.g. with 8,10 , or 12 long legs) could ensure very high ground clearance reducing the need for heavy armor on the bottom and lightening the vehicle. Furthermore, such a walker retains effectiveness even after the loss of some of the mechanical limbs. Moreover, such legs would provide variable ground clearance offering unmatched cross-country maneuverability and superior ability to take advantage of cover. Thus, artificial muscles could be the pathway to the next generation of military ground all-terrain vehicles, e.g. recce vehicles and armored personnel carriers.

### 3.8.7. Alternating actuators

Biological muscles can only contract. So, reverse motion is achieved by another contracting muscle acting from the opposite side of a joint, e.g. as in the human bicep-tricep system. In contrast, the described devices have the ability to produce muscle and counter-muscle motion, e.g. by using the two polarity configurations described above. This is a potentially very useful feature, as it would be an efficient way to build alternating-motion actuators while circumventing the typical need for a joint.

## 4. Conclusions

Artificial muscles based on microfluidics, arrayed microcapacitors, electrostatic forces, and AM, have been described and analyzed. Calculations indicate 33 MPa achievable stress. COMSOL simulations and parameter sweeps of a single device and a $2 \times 2$ array indicate $10-20 \%$ strain and optimal parameter values to maximize output force density. Alternative wiring schemes of the same array show muscle-like and counter-muscle-like behaviors, offering novel capabilities for building actuators. The proposed technology promises a major impact on a range of applications, e.g. exoskeletal locomotion, prosthetics, servo-assists, walker allterrain vehicles, and stealthy biomimetic underwater propulsion.

## Data Availability Statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

## 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

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## Biography

Dr Kartalov was born and grew up in Bulgaria. In 1994, he moved to the US to matriculate in the California Institute of Technology as an undergraduate. Dr Kartalov earned a B.S. in Physics, M.S. in Applied Physics, and Ph.D. in Applied Physics, all from California Institute of Technology. In 2004, Dr Kartalov moved to the University of Southern California as a postdoctoral scholar. In 2006, he won a K99/R00 career award from NIH and became faculty at USC in 2008. In 2016, Dr Kartalov moved to the Naval Postgraduate School to take his current position as Associate Professor of Physics. Dr Kartalov has 26 issued US patents and 25 peer-reviewed publications. He lives in Pacific Grove and has two sons aged 3 and 9.


[^0]:    * Corresponding author.

    E-mail address: epkartal@nps.edu (E.P. Kartalov).

