Closed-Loop Temperature Control of Nylon Artificial Muscles

Carter S. Haines and Günter Niemeyer

Abstract— **Coiled actuators made from polymer fibers are fast becoming popular due to their low-cost and ease of fabrication. Unfortunately, reliable real-time temperature measurement has been frustrated by the small fiber diameter of typical actuators. By using coiled polymer fibers wrapped with a metal wire, we demonstrate the ability to concurrently drive a muscle by electrothermal heating, and monitor muscle temperature through the wire resistance. This simple method enables convenient overheat protection for these muscles, as well as the possibility for closed-loop temperature control. Using this platform, we demonstrate a nested controller using temperature and position feedback to improve contraction speed, and investigate the cooling rates of various configurations that increase total force output.**

I. Introduction

Recently discovered twisted and coiled polymer fiber actuators can provide over 30% tensile contraction when heated [1]. These "artificial muscle" fiber actuators have shown promise in robotics, where their light weight, low cost, inherent compliance, and slender form factor are highly desirable. Coiled artificial muscles operate via a thermal-expansion-driven effect that pulls adjacent coils together as temperature increases, and pushes them apart during cooling. The resultant actuation stroke and force are relatively linear with temperature, especially when compared against shape-memory alloy wire muscles. This, combined with the low thermal hysteresis of these polymer muscles, makes them attractive for applications where position or force control is required [2, 3]. Unfortunately, failure due to overheating is a major risk when muscle temperature is not known. A simple method for temperature feedback is desired, both to improve controllability and protect the actuator.

Direct temperature measurement has been notoriously difficult, especially for small diameter fibers where the thermal mass of the muscle is low. Traditional thermocouples and thermistors have been used to measure the temperature of large-diameter muscles [4], but this method is typically too slow and inaccurate for small-diameter, fast-actuating muscles due to the non-negligible thermal mass of the sensor. Non-contact methods such as pyrometry and infrared cameras are also able to measure large-diameter muscles [5, 6, 7], but the cost of such measurement equipment far exceeds the cost of most muscles, and these methods typically lack the resolution to accurately determine temperature at the sub-millimeter scale, unless expensive infrared-transmissive optics are used.

Fabricating muscles from commercially available silver-plated multifilament yarn has become popular due to the ease of preparing muscles by simply inserting twist under load [8, 9, 10, 11]. The silver coating allows these coiled yarn muscle to be electrically actuated by Joule heating without requiring additional processing or treatment. Unfortunately, although the resistance of this coating changes during heating, it also changes as the length and geometry of the coiled muscle change during actuation. This dependence on both temperature and strain is complex [12], and can be further affected over time as the silver plating oxidizes and degrades with repeated strain, frustrating attempts to measure muscle temperature based on resistance alone.

Copper-wire-wrapped coiled polymer actuators developed at the University of Texas at Dallas [13] provide an ideal muscle structure for both Joule heating, and resistive temperature measurement. Figure 1 shows an example of one of these coiled fibers made by wrapping a 45-gauge enamel-coated copper wire around a coiled 175-um-diameter nylon 6,6 monofilament sewing thread. The chiralities of wire wrapping and coiling were opposite to prevent the wire from penetrating the coil during actuation (Fig. 1A). By soldering wires onto each end of the muscle, electrical and mechanical connections can be made to allow for easy testing. This wire solves the issues with silver-plated yarns by providing constant resistance with strain, while acting as a resistive

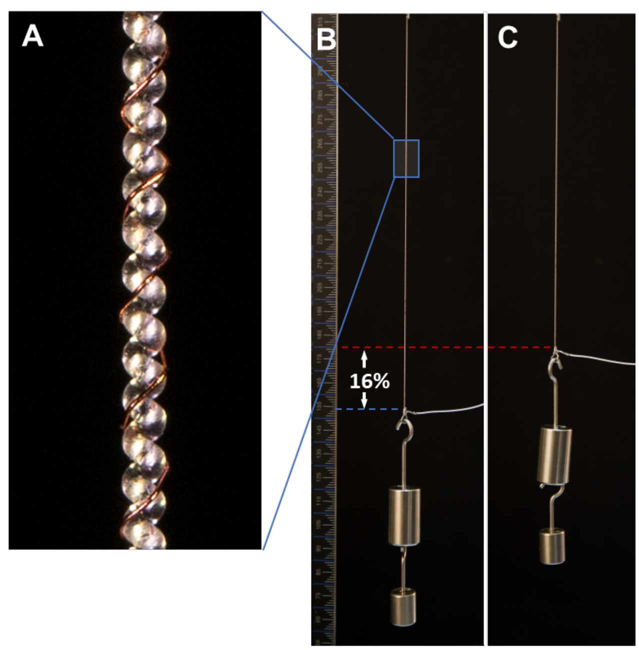

Fig 1. Nylon artificial muscle fiber made by twisting and coiling a 175-µm-diameter sewing monofilament, and wrapping this with a 45-µm-diameter enamel-coated copper wire to enable electrical heating. A) Close-up photograph of the muscle fiber. Photographs of the muscle B) at room temperature and C) when heated to 200°C, while lifting a load of 30 g.

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thermometer for sensing temperature. Fig. 1B shows one such muscle at room temperature (20°C) with a 30 g applied load. When electrothermally heated to 200°C, the muscle contracted by 16% of its initial length (Fig. 1C), and returned back to its original length when cooled.

In this paper, we explore the use of these wire-wrapped, coiled polymer artificial muscles to provide both electrothermal actuation and resistive temperature feedback. Prototype circuitry is developed to implement these features simultaneously, and enable closed-loop control over temperature. Lastly, this real-time temperature measurement is used to measure the thermal performance of various coiled artificial muscles.

II. Temperature Measurement

Electrothermal heating and resistance measurement were performed quasi-simultaneously by quickly switching between a pulse-width modulated power supply input and a resistance measurement circuit. These cycles were driven at 10 kHz, and consisted of a power phase and a resistance-measure phase. During the power phase, a MOSFET was switched on for between 0 and 50 µs to deliver adjustable power to the muscle from the power supply. During the resistance-measure phase, a constant-current source was connected to the actuator for 20 µs and the resulting voltage was recorded by an analog-to-digital converter (ADC) after passing through an op-amp to amplify and limit the signal within the 0 to 3.3V range of the ADC. Fig. 2A shows a

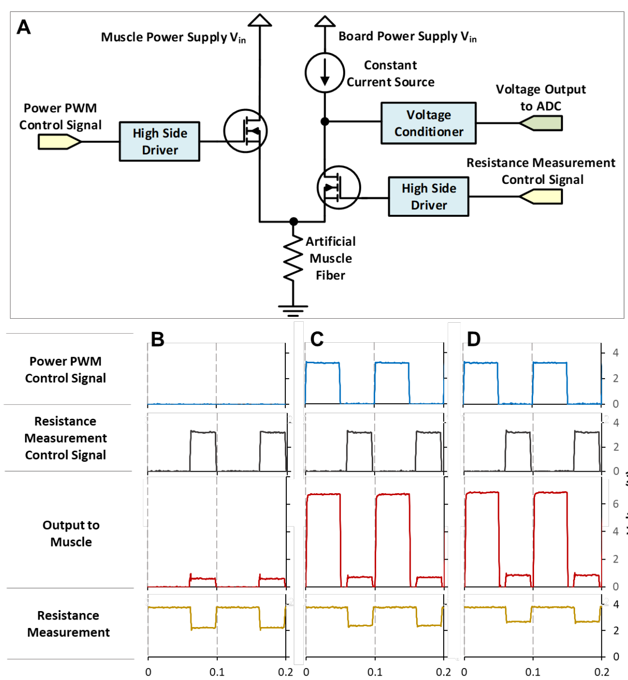

Fig. 2. A) Simplified diagram of the circuit for applying power and measuring the resistance of the muscle fiber. Two digital inputs control the circuit to either apply power from an external supply or to apply a constant current to the muscle to measure resistance. Oscilloscope plots of these control signals, and the resulting outputs are shown in B-D, where B) is with 0% duty power applied to a 20°C muscle, C) is with 100% duty applied to a 20°C muscle, and D) is with 100% duty applied to a 140°C muscle.

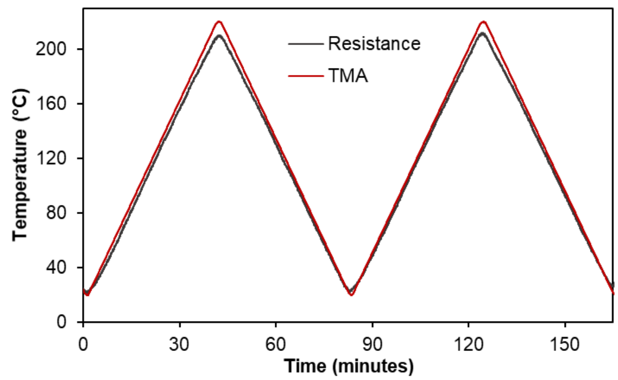

Fig. 3. Temperature measurement of a wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon monofilament, when environmentally heated in the oven of a thermomechanical analyzer (TMA). The red curve is the oven temperature, and the black curve is the temperature calculated by measuring the resistance of the wire.

block diagram of the circuit used. The constant-current source was implemented using an LM317 linear regulator and a variable resistor to adjust the sense current between 10 and 100 mA for accommodating different muscle resistances. Measurement and control was performed by a SAM3X8E microcontroller connected to a computer. The MOSFET control waveforms, and the resulting voltage on the muscle and the resistance measurement circuit are shown in Fig. 2B, C, and D, at 0% power, at 100% power with the muscle at 20°C, and at 100% power with the muscle at 140°C, respectively. Note that the resistance measurement voltage reflects an offset due to Schottky diodes in the circuit, which, while not strictly necessary, were added for protection in case both MOSFET transistors were accidentally turned on simultaneously. This offset voltage was subtracted out prior to calculating temperature.

Muscle temperature was calculated from the resistance of the copper heating wire using a temperature coefficient of resistance (TCR, α) for copper of $\alpha = 0.0039\ K^{-1}$ [14] and the equation

$$T(R) = T_{20^\circ C} + \frac{R/R_{20^\circ C} - 1}{\alpha} \tag{1}$$

To verify the accuracy of these resistance-derived temperature measurements, we heated an actuator in a temperature-controlled oven and compared the resistance-derived temperature measurement against the oven temperature. Fig. 3 shows these results for a polymer muscle made by coiling a 175-µm-diameter nylon sewing monofilament, and wrapping the coil with enameled copper wire. The applied tension was 300 mN.

III. Closed-Loop Control

Closed-loop temperature control was achieved by feeding these real-time temperature measurements into a simple PID controller. Fig. 4 shows the ability of the controller to track a setpoint temperature that linearly ramps from room temperature to 180°C and back over 40 seconds, along with the required input power levels.

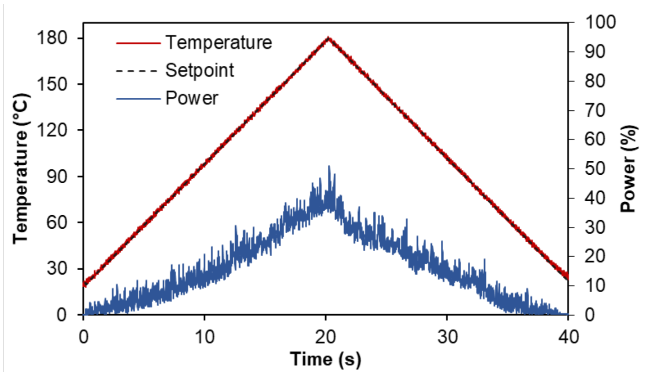

Fig. 4. Closed-loop temperature control of a wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon monofilament. The dashed black line shows the requested triangle temperature ramp from 20°C to 180°C, and the red line shows the measured temperature from the heating wire. The amount of power applied, as a percentage of the maximum 30V power supply output, is shown in blue.

Fig. 5 provides the resultant isometric (fixed length) force generation of the muscle, measured by a load cell (Transducer Techniques LSP-2). These measurements show good agreement with the force measured by a thermomechanical analyzer (TMA) when heated to 150°C and back at a much slower 5°C/min scan rate.

Isometric force generation was also measured in response to a step change in the temperature setpoint. As shown in Fig. 6, when a setpoint of 160°C was requested for 2 seconds, every 10 seconds, the copper wire reached the setpoint nearly instantly at the 60 V power supply input voltage. However, although the temperature increased and settled at its setpoint in just a few milliseconds, the force generated by the muscle rose more gradually, possibly reflecting the need to conduct heat from the heating wire into the polymer muscle fiber.

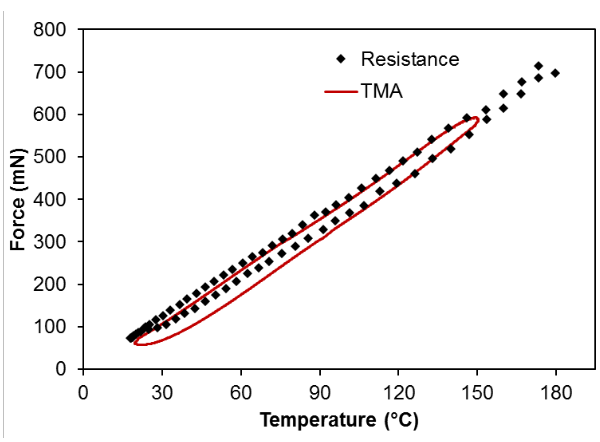

Fig. 5. Force versus temperature of a wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon monofilament. The red line was collected during quasi-static heating in an environmental chamber. The black dots are temperature measurements from the electrical resistance of the copper wire during electrothermal heating.

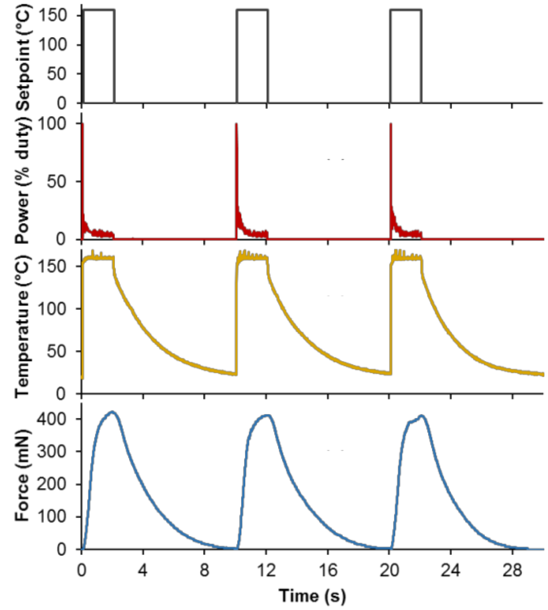

Fig. 6. Step-response of the temperature controller when a setpoint of 160°C was requested for 2 seconds every 10 seconds. The setpoint (black), percent of maximum power (red), resistance-derived temperature (yellow), and force (blue) are shown for a wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon monofilament

As seen in Fig. 6, activation of thermally-actuated muscles can be very fast. The only limitation is how quickly heat can be input to and conducted through the muscle fiber. In the case of electrothermal heating, short high-power pulses can drive rapid actuation in under 1 s [2], with contraction times below 50 ms being possible [15]. However, with closed-loop control employing only position feedback, large gains and fast contraction times can result in the controller requesting large power levels that cause the muscle to reach unsafe temperatures that far exceed its melting point. In the case of nylon 6,6, the muscle completely melts at ~260°C, with irreversible damage occurring at lower temperatures. This issue becomes especially problematic with large inertial loads, or when the requested actuation stroke is close to the maximum stroke of the muscle, leaving little room for overshoot. By adding temperature feedback, the controller can both prevent overheating the muscle and be better damped to reduce overshoot and ringing.

To demonstrate improved position control using temperature feedback, a nested controller was implemented comprising an inner and outer PID loop (Fig. 7A). The outer loop takes a position setpoint and computes a desired temperature. This temperature is limited to a safe maximum value, and then fed into the inner loop which controls power to realize this setpoint temperature. The performance of this nested position and temperature feedback controller was compared against a regular position-only feedback controller with no temperature information. Fig. 7B and 7C show the actuation strain and temperature, respectively, for each controller when a stepwise contraction of 5% was requested from a coiled, wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon coiling thread. The muscle supported a 30 g load. Both controllers were limited to a maximum temperature of 200°C to avoid damaging the muscle, as seen in Fig. 5C. In the case of the position-only controller, this meant that the gain had to be reduced to prevent the controller from exceeding this limit. As a result, the nested controller outperformed the position-only controller, exhibiting significantly faster rise and settling times.

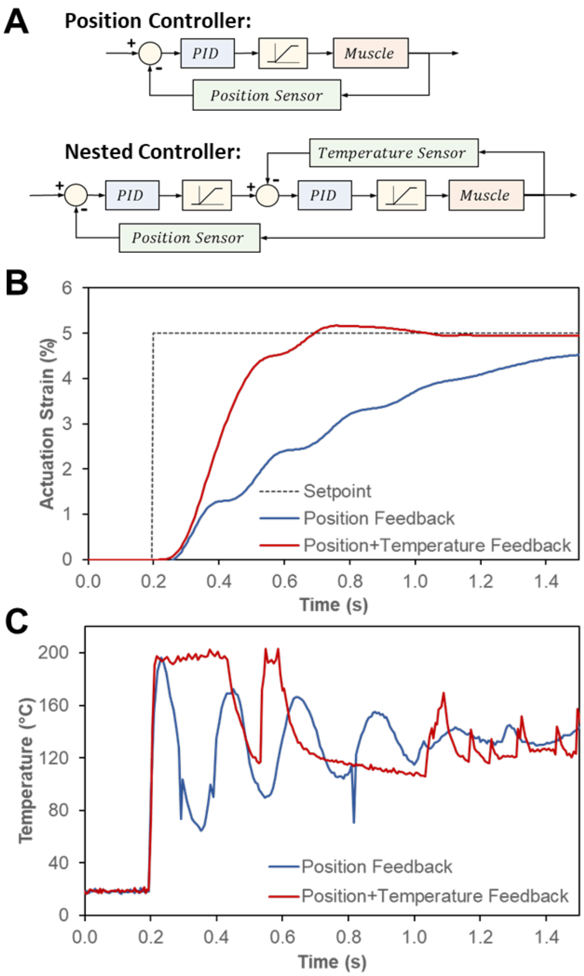

Fig. 7. Position control of a wire-wrapped artificial muscle made by coiling 175-µm-diameter nylon monofilament. A) Diagram of the controllers. Position-only control (top) was implemented using an encoder for feedback, while a nested position and temperature feedback controller (bottom) added a temperature control stage based on measuring wire resistance. The yellow function blocks represent the limited, non-negative range of power or temperature that can be applied. B) Actuation strain and C) temperature are shown comparing the performance of the two controllers when a setpoint of 5% contraction is requested at 0.2 s.

IV. Thermal Performance Of Polymer Muscles

For applications where large actuation forces are required, the force capacity of a single muscle fiber can be increased by increasing the diameter of the fiber it is made from. This force scales well, increasing quadratically with increasing fiber diameter [1, 16]. However, this increase in force comes at the expense of thermal rates during heating and cooling. There are two fundamental processes at work. Within the fiber, heat propagates according to the heat equation and the Fourier number such that the time constants scale with the square of distances, i.e. with the square of the diameter. At the external boundary, the heat transfer is limited by the surface area and hence scales linearly with diameter. In combination, we expect the overall time constants to scale between linearly and quadratically with diameter.

To investigate this trade-off, real-time resistive temperature measurement was performed on three otherwise-similar coiled polymer muscles, made by twisting nylon monofilaments having diameters of 125, 175 and 250 µm (Fig. 8A). The force generated by each muscle was measured using a load cell. Each muscle was heated and held at 160°C for 5 s by closed-loop temperature control, after which time the power was shut off. The cooling rate was measured as the time required for the muscle to relax to $1/e$ of its fully-contracted force. This thermal cooling time constant increases with increasing area for both muscles in stagnant air, and with 70 cm/s of forced-air (Fig. 8B).

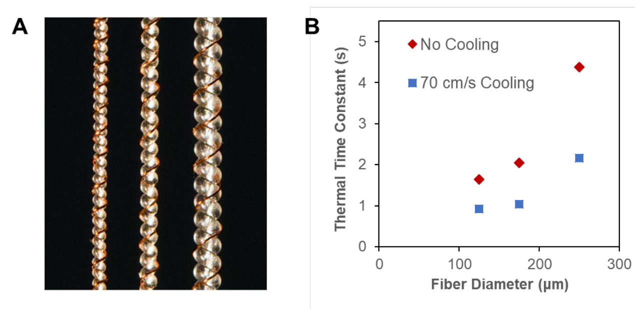

Fig. 8. A) Pictures of copper-wire-wrapped nylon muscles made by coiling 125 µm (left), 175 µm (middle), and 250 µm (right) diameter nylon monofilament sewing thread. B) Comparison of the cooling performance of the muscles in (A) in stagnant air (red) and 70 cm/s forced air (blue), as a function of the cross-sectional area of the precursor monofilament fiber.

Alternatively, the force of a coiled muscle can also be increased by lowering the coil spring index (the ratio of the average coil diameter to the diameter of the constituent filament) [1]. This increases the muscle's stiffness and thus its force, but results in lower actuation stroke. There is also a lower limit on how small the spring index can practically be, since the coil cannot be made smaller than the diameter of the fiber from which it is made.

For a coil delivering a stroke of Δx while working against a force F, the contractile power output, P, is limited by the minimum cycle duration, T, according to:

$$P = \frac{F\Delta x}{T} \tag{2}$$

Since force scales quadratically with fiber diameter, and the thermal time constant also approaches a quadratic relationship with fiber diameter as the fiber diameter becomes large, heating and cooling place an upper bound on the power output of a single coiled muscle fiber of a given length, irrespective of fiber diameter.

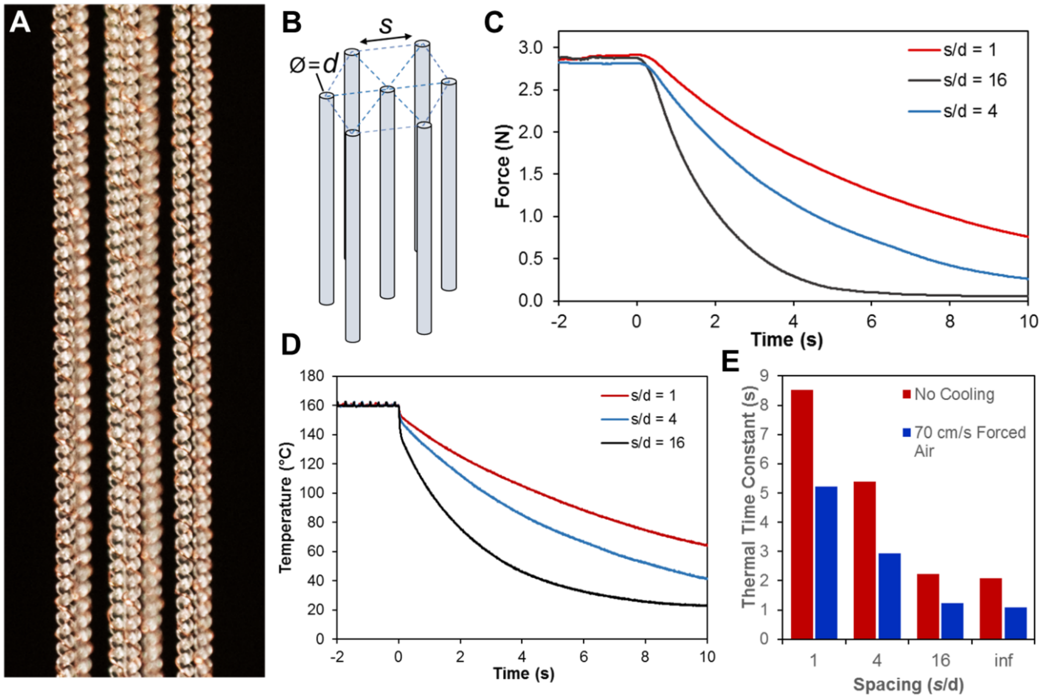

Fig. 9. Performance of bundled artificial muscles. A) Photograph of a bundle of seven wire-wrapped, coiled muscles made by twisting 175-µm-diameter nylon monofilament. Each coiled fiber has an outer diameter, *d*, of 0.37 mm and the spacing between the center of adjacent fibers, *s*, was 1.5 mm, for an *s*/*d* ratio of 4. B) Diagram of the close-packed hexagonal bundle configuration used for testing. Comparison of the C) force and D) temperature of bundles of 7 wires having *s*/*d* spacings of 1, 4, and 16. The muscle was initially held at 160°C by electrical heating, and power was turned off at 0 seconds. E) The time constants for cooling in stagnant and 70 cm/s forced air for the three bundles, and for a single, non-bundled fiber representing a bundle of muscles with infinite separation.

In principle, these issues can be overcome by bundling multiple muscle fibers in parallel to increase force, without commensurately slowing heating and cooling. This is an especially attractive method since the ability to easily and cheaply produce coiled nylon fibers in large quantities enables many small-diameter, fast-cooling fibers to be placed side-by-side to form a muscle bundle. Like muscle bundles found in nature, this also enables techniques such as recruitment of groups of muscle fibers to increase bundle force. When the N muscles within a bundle are spaced sufficiently far apart, these bundles should be able to cool as quickly as each individual fiber, while providing N times the force and power. However, bundling beyond a critical density will negatively impact cooling and the power output benefits.

To explore the effect of bundle spacing, centered hexagonal bundles of seven muscles were constructed with different spacings (Fig. 9). The outer diameter of each muscle fiber, d, was 0.37 mm, and the bundles tested were spaced by a distance between the center of each muscle, s, of 0.37, 1.5, and 6.0 mm, corresponding to s/d ratios of 1, 4, and 16, respectively. These results were also compared to a single muscle fiber, representing the performance of a bundle having infinitely large spacing, $s/d = \infty$. As expected, the 8.5 s cooling rate constant was longest for tight bundles with $s/d = 1$. The rate constant decreased to 5.4 s by increasing spacing to $s/d = 4$. Only at a larger $s/d = 16$ did the 2.2 s time constant approach the faster 2.1 s performance of well-separated $s/d = \infty$ muscles. When actively cooled with 70 cm/s of forced air, the cooling times decreased by between 38% and 48%, but the same general trend was observed with increasing bundle spacing.

We notice that even the worst-case tight bundle ($s/d = 1$) showed only a 4x slowdown in cooling compared to the 7x increase in force. We believe this stems from the relatively large overall surface area making bundles of small fibers superior to large fibers.

V. Conclusions

In this paper we demonstrated a simple, real-time method for powering coiled polymer artificial muscles, and simultaneously measuring their temperatures. Prototype hardware was developed to implement this technique and validate the accuracy of these measurements against muscles heated in an isothermal oven. This method has enabled us to use simple control techniques to provide closed-loop control of muscle temperature. We can thereby avoid overheating as well as improve the performance of position controllers.

Using these temperature measurements, we also investigated the effects of scaling and muscle bundling in an effort to increase force and mechanical power output. Increasing fiber diameter does raise force output, but thermal conduction ultimately limits the possible power output of a single muscle irrespective of fiber diameter. Bundling can improve both force and output power, though again thermal considerations limit the bundle density. Not surprisingly, this speaks to the need for efficient cooling and maximizing total surface area. We hope future work can explore ways to include microscopic surface features and passively improve cooling rates.

We hope our results will enable more capable and robust coiled polymer artificial muscles, eliminating any risk of muscle overheating, and ultimately promote the use of these artificial muscles in robotic and mechatronic applications.

ACKNOWLEDGMENT

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