Force Control of Textile-Based Soft Wearable Robots for Mechanotherapy

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Abstract—Soft robotic devices have been utilized in a number of biomedical applications involving human interaction. An emerging opportunity for soft robotic wearable devices is in mechanotherapeutic applications for the recovery and regeneration of soft tissues. Previous studies have implied that judicious force application during mechanotherapy plays an important role in the functional outcome of tissue regeneration. In this paper, we propose soft robotic devices with closed-loop force control to precisely manipulate muscular tissue. The developed devices incorporate fully soft sensors and actuators using textile-based materials and fabrication methods. The closed-loop force control system is demonstrated in bench studies to regulate massage-magnitude forces at frequencies akin to those expected in manual mechanotherapy practices. Testing of the device on human limbs demonstrates the precision and accuracy of the closed-loop force control methodology across different body shapes and types. When commanded to regulate sinusoidal force profiles (with amplitudes of 30N, 45N and 60N), the soft robotic force control device could regulate peak compressive loads to within 0.7N of the desired force. Conversely, open-loop pressure-based control resulted in up to +/-6.6N force tracking variability between participants. A soft robotic system with independently actuatable modules was also fabricated to demonstrate force-controlled actuation patterns to mimic manual massage techniques.

I. INTRODUCTION

Soft robotic devices fabricated from low modulus materials have gained significant interest in recent years [1, 2]. The inherent compliance of smart soft-bodied devices makes them particularly attractive for biomedical applications that involve human interaction [3]. Soft robotic approaches have been leveraged for pneumatically-powered assistive gloves [4, 5], soft wearable exosuits for rehabilitation [6, 7] and soft orthotic sleeves [8]. Soft robotic implants have also demonstrated the ability to recover heart function by direct massage of the cardiac tissue using contractile soft actuation [9-11].

Another exciting possibility for wearable soft robotic devices is in the manipulation of soft tissue for muscle recovery and regeneration. Therapeutic mechanical manipulation of soft tissue is an ancient form of alternative medicine that has been speculated to reduce pain, augment blood flow, reduce tissue stiffness and increase limb range-ofmotion [12]. The biological mechanism(s) underlying the effect of massage-like therapy are still not well understood but recent studies have demonstrated that massage attenuates inflammatory signaling [13]. Striking functional outcomes of soft tissue manipulation therapies have recently been demonstrated. Studies have shown that cyclical mechanical manipulation can accelerate the recovery of musculoskeletal tissue and minimize muscle fiber damage after injury from eccentric exercise [14]. Recent work in an animal model has also demonstrated that severely injured musculoskeletal tissue can be rapidly regenerated using cyclical tissue compression without the assistance of biological therapies [15]. These studies underscore the potentially significant impact of mechanotherapeutic approaches for soft tissue regeneration. However, mechanically-induced tissue regeneration remains poorly understood and clinical data showing functional outcomes is limited.

Many devices for manipulating muscle tissue have been developed for a plurality of clinical applications. Electrical stimulation is a common method of inducing native muscle contraction and has been demonstrated to aid muscle recovery [16]. Mechanical means of tissue manipulation such as intermittent pneumatic compression devices are also commonplace; these inflatable cuffs are placed around limbs for the treatment of many conditions from edema recovery to deep vein thrombosis prevention [17]. Compression devices can incorporate sequential actuation of individual pneumatic chambers to create peristaltic-like squeezing motion of soft tissue. Examples of this include the NormaTec PULSE (NormaTec, MA, USA) for sports recovery and the Flexitouch system (Tactile Medical, MN, USA). These products are designed to augment mechanical actions within the body including the transport of lymph fluid or lactic acid. However, these systems have limited control modalities and no capacity to monitor the forces being exerted to the body during operation. Mechanotherapy studies have shown that the loading conditions of soft tissue play a critical role on the functional outcomes of mechanotherapy [14].

Robotic force control systems for mechanotherapy have previously been explored, for example researchers have recently developed robotic implants to accelerate tissue

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growth [18]. Rigid-body industrial robotic manipulators have also been utilized for mechanotherapeutic soft tissue manipulation [19-21]. Despite having the ability to precisely manipulate tissue and apply controlled forces, industrial robots are expensive and not intrinsically safe for human interaction, which makes them impractical for large-scale human subject testing. Conversely, soft robotic devices that can be worn during operation have significant potential to assist the therapeutic recovery of muscle. Emerging new soft sensor designs [1, 22-26] are increasingly being integrated into artificial muscles to enable enhanced control [27]. Nonetheless, the focus of these sensor-integrated actuators has largely been for proprioception rather than for exerting controlled loads to the external environment, with limited focus in wearable applications.

In this paper we propose the use of textile/silicone-based soft force sensors to enable closed-loop force control of wearable soft actuators. We demonstrate the ability of the developed system to apply controlled and repeatable forces in bench testing and to the gastrocnemius muscle in a user study. The experimental results illustrate how the closed-loop control system can deliver much improved consistency in force application between different subjects compared to a pressurecontrolled system. We also demonstrate the ability to generate force-controlled pressure gradients using sequential actuation of multiple actuators.

II. SOFT ROBOTIC SYSTEM DESIGN

A. Device Overview

We present textile-based inflatable devices that can be wrapped around a limb so as to apply controlled compressive loading to a user's muscle during actuation (see Fig 1).



Fig 1. Concept sketch showing a section view of the device on a limb and control schematic for the textile-based soft robotic wearable device (3 sensor-actuator module system shown).

The actuator is fabricated from two textile layers to form a sleeve which houses an inflatable bladder fabricated from thermoplastic elastomer (TPE). The outer textile layer of the actuator sleeve is inextensible and the inner layer is elasticated so as to direct force towards the muscle when the bladder is inflated. The device is bound around a limb and secured using Velcro. A soft force sensor is positioned at the interface of the actuator and muscle to provide feedback for closed-loop force control. An electronically-controlled air pressure regulator is used to pressurize the bladder according to the desired load. We first demonstrate an initial single actuator-sensor module and finally, an integrated three actuator-sensing module that can facilitate sequential actuation to emulate massage-like therapy. The actuator-sensor modules are comprised of two discrete layers for sensing and actuation (Fig 2A). Each actuator module has an inflation area of 45mm×100mm so as to fully cover the sensor profile, which has an area of 991mm² (sensor profile and dimensions are shown in Fig 2B).

The sensing layer incorporates the soft force sensor(s) which are sewn in to an elastomeric compression wrap material (Fabrifoam, Fabrifoam Products, PA, USA). Actuator sleeves are sewn together from an inextensible Kevlar-reinforced sail cloth (Dimension-Polyant, Kempen, Germany). An elasticated warp-knitted, raschel polyamide-elastane textile (24710, Darlington, USA) is sewn to the inner layer of the actuator layer (see Figs 1 and 2). The knit structure is free to distend upon inflation of the inner bladder and provides recoil to assist deflation of the bladder. The internal TPE bladder (Stretchlon 200, Fibreglast, USA) is fabricated using a heat press. A standard 1/8" polyurethane airline is bonded to the TPE using Loctite Vinyl Fabrics Adhesive (Henkel Corp, CT, USA) to form an airtight seal.



Fig 2. (A) Image showing a single actuator-sensor module with detail of the soft fabric-silicone based sensor. (B) Sensor profile and dimensions. (C) Photograph of single module device being used on human muscle and (D) Photograph of 3 actuator-sensor module device.

B. Soft Textile-Silicone Force Sensor

We adopt silicone-textile sensors that utilize a capacitivebased transduction mechanism (Fig 2A) [26, 28]. The sensors are fabricated from two layers of conductive knit electrode fabric (M-130, Shieldex) that are separated by an elastomeric dielectric layer (Ecoflex 0030, SmoothOn, USA) to form a deformable capacitor. This sensor design has previously been utilized in strain sensing applications [26] although the same design can be utilized for sensing normally applied load. This soft sensing approach has several advantages for the proposed application. Firstly, the sensors are conformable which makes them ideal for integration into wearable devices. This sensor implementation does not incorporate potentially hazardous or toxic materials utilized in other soft sensor designs, and prior sensor characterization studies have demonstrated good linearity. A thin layer of dielectric elastomer (thickness 200µm) is applied to a conductive fabric (thickness 150µm) layer using a film applicator. After curing, a second elastomeric layer is applied and a second conductive fabric layer is placed on to the uncured elastomer and left to cure. A laser (VLS 6.60, Universal Laser Systems, USA) is used to cut the final sensor elements. A pair of micro coaxial cables with 300µm diameter (50MCX-37, Molex Temp-Flex, USA) are mechanically-adhered to the conductive knit layers with a thermoplastic film (3914 Sewfree Tape, Bemis Associates Inc., USA). These micro cables are connected to a measurement circuit which contains a linear capacitor-tovoltage converter (CVC) (CAV444, Analog Microelectronics, Mainz, Germany). Our system is designed to receive capacitance signals from 0-200pF. A low-stretch polyestercotton material (Typhoon Wide DWR - 00552v, Milliken, USA) was bonded to both sides of the sensor using a thermoplastic film (3914 Sewfree Tape, Bemis Associates Inc., USA) in order to minimize transverse sensor strains during compressive loading.

C. Control System

A real-time controller (cRIO 9030, National Instruments, MA, USA) is used to facilitate closed-loop force control. The analog voltage signals generated from each CVC are acquired by the control system; these force sensor signals are digitized and passed to a proportional-integral (PI) control algorithm which is implemented on a field programmable gate array (FPGA). Feedforward control can also be implemented to improve force tracking; the force set point is multiplied by a pre-determined gain and added to the PI control component. The feedforward gain can be tuned empirically based on the actuator pressure required to generate a given set point force once the device has been placed on an individual or test rig. The controller output is converted back to analog form and the signal is received by electronically-controlled pressure regulators (ITV1011-21N2BL4). The pressure regulators have integrated controllers to enable accurate pressure control of the inflatable bladders. A host PC is used to issue commands to the control system and facilitates adjustments to the controller gains.

III. EXPERIMENTS

A. Sensor Calibration and Force Control Testing

A calibration of the force sensor was performed in order to obtain the sensor sensitivity. A compression test machine (Instron 5567, Instron, MA, USA) was used to applied compressive loads to the sensor. In order to gradually and uniformly load the sensor during the calibration procedure, an 18mm thick plate fabricated from silicone elastomer (Dragonskin FX Pro, Smooth On USA) was used to support the sensor. A loading fixture of the same profile as the sensor was affixed to a 1kN load cell and aligned with the sensor so that all the compressive load experience by the load cell was transmitted to the sensor. The sensor output voltages were directly acquired from the CVC at 10N increments over a range of 0-60N. A least squares linear regression model was computed to obtain the sensor sensitivity which was 4.5mV/N (R²=0.9991) based on three loading curves (Fig 3).



Fig 3. Calibration of the soft force sensor demonstrating voltage-force relationship for three loading cycles (grey curves). The linear regression model is overlaid (black and white dashed line).

A test rig was developed to assess the controllability of the soft robotic system (Fig 4A). For the initial control parameter characterization, a single actuator-sensor module was assessed. The device was wrapped around a cylinder of 85mm diameter that incorporated a soft 5mm foam interface. In order to determine appropriate proportional and integral gains (K_p and K_i), the soft robotic system was subject to step inputs of 45N for a range of gains. Data was acquired using a data logging system (PowerLab, ADInstruments, New Zealand) at 1kHz to capture the force response and set point input signal.



Fig 4. (A) Test rigs to validate control performance (control scheme also shown) and (B) test rig with integrated pressure mapping system for quantification of contact pressure with sequential actuation.

The control system was tested with only proportional control ($K_i=0$) and no feedforward component in order to obtain a K_p value. The system was stable with a K_p gain of up to 0.5, higher gains resulted in small oscillations. A range of integral gain values were then tested to obtain the fastest response with minimal overshoot. Fig 5A shows the system response to a step input for with the proportional gain set at 0.5 and a range of integral gains from 0.0002-0.001. An integral gain of 0.001 yield a response time of 600ms (to 90% of the set-point) with a 4.4% overshoot. Lowering the integral gain to 0.0008 yielded a 720ms response time with negligible overshoot.

To demonstrate improved tracking with feedforward control, the pressure regulator was adjusted until the force sensor reached 45N in order to obtain the open loop pressure-force relationship for the test rig. Fig 5B shows the control system responding to a 45N sinusoidal input (0.1Hz) with PI control alone and with the feedforward component added (with gain set to 0.12PSI/N). The incorporation of feedforward control reduced the peak tracking error from 1.0N to 0.2N.



Fig 5. (A) Control system response to step input of 45N showing the incorporation of integral gain to eliminate steady state error where Kp=0.5 and Ki=0.0002-0.001. (B) Force tracking of a sinusoidal input (0-45N amplitude at 0.1Hz) with PI control (Kp=0.5, Ki=0.0008) and; with PI + feedforward control (feedforward gain of 0.12PSI/N).

To evaluate the bandwidth of the system, the system was actuated with 10 sinusoidal input cycles of frequencies between 0.05-3Hz for a peak-to-peak force input of 0-30N, 0-45N and 0-60N. These peak-to-peak force ranges were chosen as representative force levels that might be exerted during manual massage. Previous research has indicated massage pressures to be in the range of 14-34kPa when applied through the thumb of a massage therapist [29].

In this series of studies, the proportional gain was set to 0.5 and integral gain to 0.0008, the feedforward gain was set to 0.12PSI/N. A bode plot (Fig 6) illustrates the system response. The device bandwidth was determined to be 0.53Hz, 0.69Hz and 1.01Hz for 0-60N, 0-45N and 0-30N amplitude sinusoidal inputs, respectively. This bandwidth limitation is largely a result of the actuator's inability to deflate quickly enough, which results in residual loading before the next actuation cycle. These results demonstrate that the developed system can generate mechanical forces at time-scales in the same order of those expected in manual massage.



Fig 6. Bode plot for bandwidth characterization of soft robotic control scheme at peak-to-peak setpoint amplitudes of 30N, 45N and 60N.

B. User Study

We performed a study to 1) demonstrate that the proposed device can apply controlled loads to a wearer's limb and 2) to compare force-control with existing control methods of therapeutic pneumatic compression. Specifically, we hypothesized that closed-loop force-controlled actuation would apply more consistent and accurate loads to a human limb compared to open-loop, pressure-controlled actuation. Pressure-control is the standard approach for existing pneumatic compression wearables, but the intrinsic variability in muscle shape and compliancy between users will lead to variability in the compression forces applied. We recruited 7 healthy participants for this study, all participants gave informed consent.

This study considers the ability of the open-loop and closed-loop regimes to accurately regulate three desired sinusoidal force profiles of amplitude 0-30N, 0-45N and 0-60N at 0.2Hz. An initial pre-test was performed on a single participant to obtain a consistent set of parameters for testing on the other 6 participants. A single sensor-actuator module was secured around the gastrocnemius muscle of the reference participant. To determine the open-loop pressure parameters, the device was actuated at 0.2Hz sinusoidal input with an initial amplitude of 1PSI. The commanded pressure amplitude was then incrementally increased until the desired peak forces applied to the muscle were measured by the soft sensor to within <1N of the peak force. The pressures required to generate peak loads of 30N, 45N and 60N on this single individual were 4.2PSI, 8.0PSI and 12.5PSI. For the closedloop force-controlled case, we commanded the control system to track 0-30N, 0-45N and 0-60N sinusoidal force profiles at 0.2Hz. Separate control parameters were set for each force profile to obtain force control to within <1N of the peak force. K_p was set to 0.5 for all cases. K_i was set to 0.0008, 0.001 and 0.0015 for the 30N, 45N and 60N cases respectively. Feedforward control was not implemented on the 30N case to prevent overshooting, and the feedforward gain was set to 0.12PSI/N for both the 45N and 60N cases.

The device was placed around the gastrocnemius muscle of the remaining 6 participants and actuated to compare openloop and closed-loop force control. The pressure and force control parameters determined from testing on the reference participant were used on all other participants. Testing between open and closed-loop regimes was randomized. For all participants, the actuator was wrapped around the calf muscle with just sufficient pre-load to prevent movement during actuation. The soft force sensor was zeroed before each actuation period. Participants were requested to remain still and seated, with their muscles relaxed for the duration of the study. Data was acquired using a data logging system (PowerLab, AD Instruments, New Zealand) at 1kHz to capture the force response and set point input signal. We recorded the peak force measured by the soft force sensor for 10 actuation cycles at both conditions (pressure control vs force control). The mean force errors for all participants and standard deviation are plotted in Fig 7.



Fig 7. Bar plot showing the mean force errors for pressure-controlled approach vs the force-controlled approach in applying mechanical forces to the gastrocnemius muscle for desired peak forces of 30N, 45N and 60N (n=6 participants). Error bars denote +/- SD. A positive error refers to less force than the desired peak force.

To make a fair assessment of the intrinsic variability of forces exerted between participants using the two control regimes (irrespective of the absolute force tracking accuracy), we considered the forces exerted on each individual participant relative to the mean force for each group. The mean force was subtracted from each individual force measurement to obtain a force deviation from the mean of each group. These force deviations were converted to an absolute value, and a mean force deviation was computed for each individual participant at all conditions. To assess statistical significance between the pressure and force control groups, we use the Mann-Whitney U test, with p<0.05 being considered significant. A box and whisker plot showing the distribution of force deviation from the group mean is shown in Fig 8.



Fig 8. Box and whisker plot showing the absolute force devitation relative to the group mean, boxes represent interquartile range and median, whiskers represent maximum and minimum values. Datasets are based on n=6 participants.

The results of this study confirm that consistent loads can be applied to human limbs using force-controlled soft actuation. With force control, the desired peak forces could be regulated to within 0.7N of error for the majority of cases. For the pressure controlled case, the peak forces applied to each participant were lower than the desired peak forces (4.9-5.0N, on average), although in some cases greater force was applied than desired. There was significantly greater variability in the forces applied between participants using the pressure control regime. The standard deviations in force error with pressure control were approximately an order-of-magnitude greater than those observed with the force control approach (4.8-6.6N) compared to (0.1-0.7N).

The consistency of forces applied to each individual participant was within a standard deviation of 100mN for both pressure and force control regimes. This is due to participants remaining stationary during the study, with minimal environmental disturbances. It is likely that a tuned open-loop pressure controlled approach may be sufficient for conducting mechanotherapy studies if participants remain static, although this may impose practical limitations for testing over long periods of time. Irrespective of the control regime used, this study underscores why force sensing at the actuator-tissue interface is important for determining the correct magnitude of force being applied to soft tissue.

Limitations to this study include evaluation of the device at a single frequency. It is not known what the optimum loading frequency will be for optimal mechanotherapy, 0.2Hz is faster than devices like the NormaTec PULSE (NormaTec, MA, USA) which inflate over a period of several seconds. Mechanotherapy studies with small animals have typically utilized higher frequencies (1-2Hz) [14, 15], the bandwidth of the current system may not be sufficient to accurately regulate force at these rates. The forces reported in this study are also those as measured by the soft force sensor which only facilitates force regulation in a localized area of the muscle. Furthermore, we do not currently account for any time-



Fig 9. Sequential actuation of individual actuators allows phased compression. Plots represent the unrolled pressure mat with the actuators located approximately at the center. Actuators were commanded to 10N load at 0.2Hz for 0°, 45° and 90° phase shifts. Time is in seconds, the pressure map is denoted in kPa.

dependent relaxation artefacts in the soft sensor or soft tissue.

C. Force-controlled Pressure Gradients with Sequential Actuation

To demonstrate the potential of a soft robotic device to perform force-controlled actuation patterns that mimic manual massage, we evaluated the pressure profile generated by the three module implementation of the device. A pressure-mapping system (Tekscan 5250, Tekscan Inc., Boston, MA) was used to quantify pressure distribution. A two point calibration was performed on the mat before being wrapped around a cylinder of 85mm diameter (experimental rig shown in Fig 4B). The device was lightly wrapped around the cylinder to minimize preload and both the mat and soft sensors were zeroed. The soft robotic system was actuated at a 10N sinusoidal peak-to-peak set-point at the following phase offsets: 0° , 45° and 90° , the actuation frequency was 0.2Hz. Fig 9 shows a sequential pressure map during actuation at these different phases. The reaction forces supported by the wrap can also be observed during actuation. Peak pressures of up to 10kPa were observed in each phase case. These contact pressures are of the same order as those exerted in manual massage [29].

IV. CONCLUSIONS

This paper demonstrates soft robotic force control in a textile-based device for proposed applications in mechanotherapy. Despite being a popular therapy, data outlining the optimal force parameters (magnitude, frequency, spatial distribution and force direction) for effective mechanotherapy is sparse. Soft robotic wearable devices will facilitate, safe and controlled clinical testing to evaluate mechanical-based, biologic-free soft tissue regeneration and recovery. With multiple sensor-actuator modules, we demonstrate the ability to perform sequential actuation patterns to mimic manual massage which may augment lymph and blood flow within the body.

Future actuation schemes may enable more versatile soft tissue manipulation, such as the ability to apply shear loads or lengthwise stripping motions [21] commonly practiced in massage. Soft sensing arrays that can monitor distributed forces over larger areas will provide greater fidelity monitoring of the forces being applied to the soft tissue. Incorporation of a regulated vacuum source in to the control system will also assist with actuator recoil to improve the bandwidth of these devices. Whilst the primary focus of this work has considered soft robotic force control for wearable mechanotherapy devices, the ability to apply controlled loads with inflatable devices may have applicability to soft orthotics and to assist the synchronization of soft robotic ventricular assist devices [9].

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