Identification and Control of a Soft-Robotic Bladder Towards Impedance-Style Haptic Terrain Display

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Abstract—This letter evaluates the capabilities of a soft robotic pneumatic actuator derived from the terrain display haptic device, the “Smart Shoe.” The bladder design of the Smart Shoe is upgraded to include a pressure supply and greater output flow capabilities. Forcing characteristics are identified and incorporated as feedforward terms in a sliding-mode tracking controller to provide impedance control. A bench top setup is created to rigorously test this new type of actuator. The bandwidth and stiffness capability are evaluated relative to forces and displacements encountered during human gait. Four force vs. displacement profiles relevant to haptic terrain display are proposed and evaluated. It was found that the actuator could sustain a stiffness similar to a soft-soled shoe on concrete, as well as other terrain (sand, dirt, etc.). Compressions of the bladder done at 20 mm/s, which is similar to the speed of human gait, showed promising results in tracking a desired force trajectory. The results in this letter show this actuator can display haptic terrain trajectories, providing a basis for future wearable haptic terrain display devices.

Index Terms—Force feedback, haptic interfaces, nonlinear control systems, soft robotics.

I. INTRODUCTION

HAPTIC manipulation allows individuals to interact with virtual objects via robotic interfaces [1]. The field of haptics has been thoroughly studied using handheld devices connected to robotic manipulators to interact with virtual objects [2], [3]. Our previous work introduced the haptic terrain display device, the “Smart Shoe” [4], [5], which uses an array of fabric-composited [6], [7], silicone bladders to simulate features in terrain. During the stance phase of gait, bladders corresponding with terrain obstacles would remain inflated while the remaining bladders deflated. This strategy would render features such as ankle pitch or roll, as well as stepping on small objects like individual stones. An example of the Smart Shoe is shown in Fig. 1(a) where an array of these bladders are attached to a shoe upper to create the haptic device. Similar terrain display devices such as RealWalk [8] and Re-Step [9] could only render gross terrain features such as ankle pitch or roll using linear actuators while the Smart Shoe could render both fine and gross terrain features.

In this work we upgrade the bladder actuator and control to improve dynamics, force output, and ease of assembly in modular designs, Fig. 1(b). The actuator is evaluated rigorously to identify actuator workspace (Fig. 1(c)), bandwidth, maximum stiffness, \( k_{\text{max}} \), output force model, as well as input and output flow models.

The actuator workspace and bandwidth are evaluated relative to two metrics. First, there exists some stiffness, \( k_{\text{req}} \), which is the required actuator stiffness for successful terrain display. If a user of the Smart Shoe is walking over concrete, each actuator should be capable of displaying the forces from the concrete as well as the virtual pair of shoes worn by the user. The equivalent stiffness of the concrete and virtual pair of shoes is the minimum required stiffness, or \( k_{\text{req}} \), which must exist somewhere in the workspace. The second metric is the actuator bandwidth relative to the frequency spectrum of ground reaction forces in gait. Ground reaction forces (GRF) have been studied in [10] and modeled in [11], which found a majority of vertical GRF (vGRF) were contained under 10 Hz, with the heel strike transient (HST) occurring at nearly 75 Hz. The success of the actuator bandwidth should be measured relative to these values.

Soft-robotic systems provide an alternative to rigid actuation methods such as the linear actuators used in [8] and [9]. Wearable soft robots with haptic applications were reviewed in...
[12] where four major actuation methods were identified: 1) cable-driven actuation, 2) fluidic actuators, 3) shape memory alloys, and 4) electroactive polymers. Our application requires an actuator capable of high forces (250 N) and moderate displacement (~20 mm) without being cumbersome to the user. A (1) cable-driven solution would be burdensome to the user by requiring large motors mounted somewhere on the user to output the large, required forces. The slow heating and cooling of (3) shape memory alloys would limit actuator bandwidth below that required for use during gait. Additionally, (4) electroactive polymers are a poor candidate due to very low forces and small displacements. The candidate that best fits the need of our application are (2) fluidic actuators with geometry chosen to best fit the application of gait. Fluidic McKibben actuators were used in the wearable ankle-foot robots in [13], but as is typical, they pulled tendons for joint actuation and mounted actuators on the leg. Fluidic actuators in this work are mounted within haptic devices to directly impede compressive forces during stepping.

This work makes several contributions. To the best of our knowledge, this is the first actively controlled pneumatic, soft-robotic bladder designed for the loading and dynamics of human gait while displaying impedance of terrain and foot-ground contact. While the review in [12] shows that silicone-based pneumatic devices are very common in wearable soft robotics, we find that the geometry of a large cross-sectional bladder capable of high forces (250 N), high strain (~63%), and being packaged in a Smart Shoe makes our design unique. This work is also the first to characterize this new bladder actuator and evaluate it as an impedance-style device meant for haptic terrain display.

The remainder of this letter begins by describing the bladder actuator and testing setup, which is followed by the identification of the force model, PWM valve flow, and closed-loop bandwidth. The sliding-mode tracking controller is then designed and various impedance profiles in the bladder workspace are tested at various compression rates. The result of this testing is presented in the following section. The letter concludes with a discussion of these results.

II. ACTUATOR DESIGN AND CONSTRUCTION

The proposed actuator is a fabric-composited silicone bladder wall molded into a 3D printed manifold. The bladder wall features a cross-sectional area of approximately 60 mm × 55 mm and a height of 40 mm. The 3D printed manifold measures 67.5 mm × 70 mm × 15 mm.

The manifold of our new system contains innovative mechatronic and flow path systems to improve the terrain display capabilities of our previous design. Several features are added in the new manifold: the number of output valves are doubled to improve actuator dynamic response, an input valve connected to an external pressure supply is added to increase actuator force output capacity, and flow paths routed for modularity allowing for single bladder or multi-bladder configurations. A control strategy has also been added to this updated actuator where pulse-width modulation (PWM) allows for intermediate stiffnesses that were unachievable with the previous bladder operation strategy.

The actuator manifold interior is shown in Fig. 2. Three solenoid valves are used to distribute air into and out of the bladder. Two solenoid valves control the output flow (yellow) while the third inputs air into the bladder from the 10 PSI pressure supply (magenta). A pressure tap line allows for external pressure sensing (cyan). Interior and exterior grooves fill with silicone rubber during the molding process to act as sealing gaskets. Multiple bladders can be combined in series (such that pressure supply lines shown in Fig. 2 align) allowing access to the same supply. Combining modules as a Smart Shoe in the configuration shown in Fig. 1 locates all output ports on the perimeter of the shoe.

The construction of the bladder walls follows the procedure outlined in [6] where a two-piece mold is used to shape Mold Max 40 silicone rubber (Smooth-On, Inc., 40 shore A hardness) with poly-cotton fabric compositing into the bladder walls. This results in consistent and durable bladder walls. The use of the same two-piece mold for each set of bladder walls helps maintain consistency in dimensions such as wall thickness. The use of fabric increases the durability of the bladder wall relative to a bladder wall made only from pure rubber. The assembly is completed by molding the walls into the 3D printed manifold.

III. TESTBED SETUP

The testing apparatus used for the system identification and control is shown in Fig. 3. The bladder actuator is controlled from a dSpace system by control input, \( u_{PWM} \), which is a voltage sent to the interior solenoid valves. A pressure supply, \( P_s \), provides pressure of 10 PSI to the bladder while a pressure sensor returns internal pressure, \( P \), to the bladder. The Ankle-Foot Simulator (AFS) [14], [15] robot is controlled by the dSpace system by control input, \( u_{AFS} \). The AFS robot returns measurements of bladder compression, \( z \), and bladder force, \( F \). The dSpace system is interfaced with a PC running ControlDesk and Simulink.

IV. SYSTEM IDENTIFICATION

This work identifies all the models necessary for model-based sliding-mode control of the soft-robotic bladders including bladder force, input and output flow, and closed-loop bandwidth.
Fig. 3. The testing setup for the bladder actuators. ControlDesk is run on a PC and interfaced with a dSpace MicroLabBox. The AFS is controlled by \( u_{AFS} \), while the bladders are controlled by \( u_{PWM} \). An amplifier of custom design circuitry drives the solenoid valves inside the actuator. A pressure supply, \( P_s \), supplies 10 PSI to the bladder. A pressure sensor returns pressure measurement, \( P \), to the dSpace. The AFS robot measures bladder compression, \( z \), and bladder force, \( F \), which is sent back to the dSpace system.

Fig. 4. The free body diagram the describes the bladder forces.

Using experimental data to identify system parameters is a critical step for proper control. Even with consistent manufacturing techniques, some variability will always be present in parameters such as wall thickness or bladder height. System identification techniques should be used to identify parameters each time a bladder is constructed, because they will all be slightly different.

A. Force Modeling

A free-body diagram of the bladder is shown in Fig. 4. A bladder actuator at some compression, \( z \), and internal pressure, \( P \), is subjected to five forces: the stiffness of the air inside the bladder, \( k_a \), applies air pressure force, \( F_a(z, P) \); the stiffness of the fabric-composited silicone walls result in the wall force, \( F_{kw}(z, P) \); various damping terms associated with the walls and air pressure are lumped into \( b_{eq} \), which gives damping force, \( F_b(z, \dot{z}, P) \), a function of compression velocity, \( \dot{z} \); the gravity vector, \( g \), applies a gravity force, \( F_g \), to the bladder; the external force applied to the bladder is written as \( F(t) \).

The system states can be written as

\[
\dot{z} = v
\]
\[
\dot{v} = \frac{1}{m_b} \left[ F(t) + F_g - F_{kw} - F_b - F_a \right]
\]
\[
\dot{P} = \frac{P}{V} \left( Q - \dot{V} \right)
\]

where \( v \) is compression velocity. The pressure dynamics, \( \dot{P} \), include control input of flow, \( Q \), bladder volume, \( V \), and the change in volume, \( \dot{V} \). This equation is derived from the ideal gas law assuming adiabatic conditions.

Previous work [4] used a complex lumped parameter model based on a mass-spring-damper with many parameters determined by a root-mean-square error (RMSE) fit to estimate \( F_{kw} \) and \( F_a \). The bladder walls were simulated in small segments using finite element analysis (FEA) at each timestep, making this model an unlikely candidate for real-time control.

Alternatively, this work proposes a polynomial fit to collected force, pressure, and displacement data, which was found to be more accurate, more applicable to a real-time application, and easier to use in model-based control schemes. The actuator force, \( F(t) \), can be written as

\[
F(t) = f(z, \dot{z}, \ddot{z}, P, t)
\]

where \( z \) is bladder displacement, time derivatives \( \dot{z} \) and \( \ddot{z} \) are velocity and acceleration, respectively, \( P \) is the internal bladder pressure, and \( t \) is time. The updated system states are written as

\[
\dot{z} = v
\]
\[
\dot{v} = \frac{1}{m_b} f(z, \dot{z}, \ddot{z}, P, t)
\]
\[
\dot{P} = \frac{P}{V} \left( Q - \dot{V} \right)
\]

Data was collected using the testbed setup shown in Fig. 3. The AFS robot performed a quasi-static compression of the bladder for 20 mm. This compression test was repeated for bladder pressures from 0 PSI to 10 PSI in 1 PSI increments.

A polynomial model was fit to the collected force data. The polynomial fit (\( R^2 = 0.988 \)) and experimental data are shown in Fig. 5. The polynomial is third order with respect to \( z \), first order with respect to \( P \).
with respect to $P$, and constrained by $\hat{F}(z = 0, P = 0) = 0$. The final model is written as

$$\hat{F} = 21.52z - 2.245z^2 + 3.028zP + 0.07648z^3 - 0.1165z^2P.$$  

(4)

B. Valve Flow Modeling

The manufacturer specifies valve flow, $Q$, as

$$Q = \frac{2Kf_T}{P_{atm}} \sqrt{(P_a - P_b)} P_b,$$

(5)

where $K$ and $f_T$ are supplied constants, $P_a$ and $P_b$ are upstream and downstream pressures, respectively, and $P_{atm}$ is a flow resistance value from [16]. For this system, pressures always operate in the subsonic region. PWM is used to relate $P_{atm}$ to PWM duty cycle, which is a common method used in pneumatic actuators [17], [18], [19], [20], [21], [22] to change the flow resistance in the valve to control flow.

PWM output flow characteristics are then evaluated by applying a pressure to the bladder and measuring flow. The output port of the actuator is connected to a flow meter (Omron D6F-20A6). A 10 PSI pressure is applied via the pressure tap port, bypassing the input valve. The PWM duty cycle of the output valves are increased from 0% to 100% before decreasing in the opposite direction over 20 seconds. This procedure is repeated in 1 PSI increments down to 1 PSI. Multiple PWM frequencies are tested between 10 – 50 Hz.

The collected data is processed by doing an RMSE search for the three coefficients that best fit a power model and minimize RMSE. The best results were found using a PWM pulse frequency between 25–40 Hz. The output flow model that computes the PWM duty cycle ($DC$) as a function of desired flow, $Q_{des}$, atmospheric pressure, $P_{atm}$, the internal pressure, $P$, is written as

$$DC_{out} = 7245 + \frac{2Kf_T \sqrt{(P - P_{atm})P_{atm}}}{Q_{des}} - 0.7814 + 12.6.$$  

(6)

This power model is shown graphically in Fig. 6(a).

PWM input flow characteristics are then evaluated by repeatedly inflating the bladder. The flow meter is placed in line with the 10 PSI pressure supply connected to the input valve. The actuator is set to atmospheric pressure and an input valve PWM is applied where bladder pressure and input flow are measured. This procedure is repeated for PWM frequencies between 10–50 Hz and duty cycles between 10% and 100% in 10% increments. The duty cycle for the desired input flow is characterized using an RMSE search as,

$$DC_{in} = 6000 * \left(\frac{2Kf_T \sqrt{(P_{des} - P)} P_{des}}{Q_{des}}\right)^{-0.7} + 16,$$  

(7)

where $P_{des}$ is the 10 PSI pressure supply. The curve fit to the raw data is shown in Fig. 6(b).

C. Closed-Loop Bandwidth

A frequency sweep is applied to determine system bandwidth. A pressure controller was tuned for the best possible performance to a step input. The desired pressure is slowly varied sinusoidally, and the actual pressure is recorded. The sinusoid frequency is increased slowly from ~0 Hz to ~20 Hz, or until the output magnitude has broken down. This experiment was conducted for an uncompressed bladder with a pressure supply of 10 PSI at three PWM frequencies: 10 Hz, 25 Hz, and 40 Hz. The magnitude of the output relative to the input is recorded and plotted as a Bode plot.

The results of the frequency sweep are shown in Fig. 7. The experimental data was fit using nonlinear least squares to a second-order transfer function

$$\hat{G}(s) = \frac{\omega_n^2}{s^2 + 2\zeta \omega_n s + \omega_n^2},$$  

(8)

where $\omega_n$ is an estimate of the natural frequency, $\zeta$ is an estimate of the damping factor, and $s = j\omega$, which is an input of frequency $\omega$. The bandwidth of the estimated transfer function is computed by finding where the magnitude crosses the -3 dB value. The highest bandwidth was achieved with a PWM pulse frequency of 40 Hz. The estimated second order parameters were computed as $\zeta = 0.86$ and $\omega_n = 58.5$ rad/s with a bandwidth of $\omega_{BW,40} = 7.3$ Hz.
The direct force controller takes actuator compression measurement, $z$, from the AFS and uses a lookup table to find the desired force, $F_{des}$. The controller is used to minimize error $e$, which is the difference between the force measurement, $F$, and $F_{des}$. A low-pass filter and dead zone limit unnecessary valve actuation.

Fig. 8. The direct force controller takes actuator compression measurement, $z$, from the AFS and uses a lookup table to find the desired force, $F_{des}$. The controller is used to minimize error $e$, which is the difference between the force measurement, $F$, and $F_{des}$. A low-pass filter and dead zone limit unnecessary valve actuation.

Fig. 9. The desired force vs. displacement curves: (C1) Stiffness of a soft shoe sole, (C2) stiffness of a soft shoe sole at a 5 mm displacement, (C3) stiffness of dress shoe in sand, (C4) double the stiffness of C1 and C2, but at 9 mm displacement. The bladder haptic workspace, according to the open-loop model, is shown in green.

V. IMPEDANCE CONTROL

This work characterizes the bladders as an impedance-style device, as in some displacement, $x$, is measured, which is then converted to a desired output force, $F_{des}$. The force can be measured directly with a force sensor or indirectly with different sensors and a force model.

The sliding-mode tracking controller is designed to minimize tracking error in the presence of modeling uncertainty, which is a common approach in controlling pneumatic cylinders [23], [24], [25], [26], [27]. The error is defined as

$$ e = F - F_{des} \quad (9) $$

where $F_{des}$ is the desired force trajectory.

A block diagram of the controller is shown in Fig. 8. The AFS robot is used to measure the bladder compression, $z$, and bladder force, $F$. The AFS is used to compress the bladder 20 mm at various compression rates. The compression value, $z$, is used to determine the desired force, $F_{des}$, based on a look up table that defines a desired haptic trajectory. A low-pass filter with a with a frequency of $\tau = 40$ Hz is used to minimize noise in the force measurement. This frequency is sufficiently higher than the closed-loop bandwidth such that it will have minimal effect on characterization. A dead zone is also implemented to only activate the controller should the error be sufficiently large. The dead zone removes unnecessary valve actuation when the error is sufficiently small. The sliding-mode controller takes error, $e$, and computes a desired flow, $Q_{des}$, which is then converted to $DC_{out}$ or $DC_{in}$ using (6) or (7), respectively. The drive circuitry then outputs the necessary PWM power signals to the solenoid valves.

The derivation of the continuous sliding-mode tracking controller closely follows [28]. The remainder of the controller derivation follows the “off-the-shelf” procedure applied to the system states shown in (3). The control input, $Q$, for the SMC is written as

$$ Q = - \left[ \rho (z, \dot{z}, P) + \beta_0 \right] \tanh \left( \frac{e}{\mu} \right) \quad (10) $$

where $\beta_0$ and $\mu$ are gains. The term $\rho (z, \dot{z}, P)$ is solved as

$$ \rho (z, \dot{z}, P) = \frac{a_{max} (z, \dot{z}, P) - \hat{F}_{des}}{b_{min} (z, \dot{z}, P)} \quad (11) $$

where $\hat{F}_{des}$ is the time derivative of the desired force, with terms

$$ a_{max} (z, \dot{z}, P) = \dot{z} \frac{\partial}{\partial z} \hat{f}_u + A_{b, u} \dot{z} \frac{\partial}{\partial P} \hat{f}_u \quad (12) $$

and

$$ b_{min} = \frac{P}{V_{max}} \frac{\partial}{\partial P} \hat{f}_l. \quad (13) $$

Terms $\hat{f}_u$ and $\hat{f}_l$ are the upper and lower force bounds of $\hat{f}$, respectively, $V_{min}$ and $V_{max}$ are the minimum and maximum estimates of bladder volume, respectively, and $A_{b, u}$ is the upper estimate of the bladder top area in contact with the AFS. The specifically chosen upper and lower bounds of the uncertain terms ensure that the controller will always push the tracking error to zero. Additionally, the tanh() function is used in place of the traditional sgn() function due to better performance from the tanh() function, which was found in [29].

VI. EVALUATION

This section evaluates the ability of the bladder actuator to track various trajectories through the force vs. displacement workspace. The four proposed curves, $C_1$, are shown in Fig. 9.

Each curve is motivated by relevant haptic terrain display interactions. The first curve, $C_1$, simulates stepping on a soft shoe sole onto a hard surface such as concrete, which is the lower bound on $k_{max}$, or the actuator stiffness required for successful haptic terrain display. Experimental testing of EVA foam, which is common in Nike running products, was done using the AFS robot. The linear stiffness was found to be approximated as 80 N/mm. The stiffness of soft soled shoes varies in the literature, ranging from 55 N/mm [30] to 257 N/mm [31]. A stiffness of 28.1 N/mm was used in $C_1$. Note that in an actual Smart Shoe design, four of these actuators could be combined in parallel under the heel portion of the shoe, bringing the total stiffness to 112.5 N/mm during heel contact, which is similar in value to the experimentally determined stiffness as well as the values published in the literature.
The second curve, $C_2$, evaluates the same stiffness as $C_1$, but beginning at an offset of 5 mm. This example is motivated by Fig. 10. Simulating a user stepping on a small stone would require immediate force output from the actuator in the location of the stone, but the remaining actuators should output little to no force until the simulated shoe contacts the ground. Curve 3, $C_3$, shows experimentally collected data from a shoe being compressed into sand using the AFS robot. Finally, $C_4$, doubles the stiffness seen in $C_1$ and $C_2$, but at a greater offset of 9 mm. This example is chosen because the maximum stiffness of the actuator might not start at zero displacement, but further in the displacement workspace.

To further evaluate the actuator in the domain of bandwidth, these curves are tested at four different compression rates: 2 mm/s, 4 mm/s, 8 mm/s, and 20 mm/s. The 20 mm/s compression rate compresses the entire bladder in one second, which is like what would be experienced during gait.

The set up in Fig. 3 is used for this evaluation. Sliding-mode controller gains of $(\beta_0, \mu) = (1 \times 10^{-5}, 5)$ were used for both input and output flow cases for $C_1$, $C_2$, and $C_3$, at the 2 mm/s, 4 mm/s, and 8 mm/s compression rates. These were tuned to achieve a critically-damped response to a step input. The gain $\beta_0$ was increased to $\beta_0 = 3 \times 10^{-5}$ for $C_4$ because the doubled stiffness in $C_4$ essentially doubles the dynamics and $\beta_0$ must be increased to account for unmodeled dynamics. Additionally, for all 20 mm/s compressions, $\beta_0$ was increased to $5 \times 10^{-5}$, again, to compensate for unmodeled dynamics in the controller.

**VII. RESULTS**

The results of the evaluation are shown in Fig. 11 where the four different curves, $C_1$, $C_2$, $C_3$, and $C_4$, are plotted at the four compression rates, 2 mm/s, 4 mm/s, 8 mm/s, and 20 mm/s. Each plot includes the bladder workspace in light green with the control results shown in orange. Note that the measured control data is only included if the desired force and displacement reside within the bladder workspace. The mean and standard deviation of the tracking error, $\mu_e$ and $\sigma_e$, respectively, are shown in Table I. Finally, two time response plots are included.

First, Fig. 12 shows the desired flow, $Q_{des}$, computed by the sliding-mode tracking controller during the tracking of $C_3$ at 2 mm/s. The desired and measured forces are included. Second, Fig. 13 shows the same desired flow, desired force, and measured force, but for the tracking of $C_3$ at 20 mm/s.

**VIII. DISCUSSION**

The impedance control results in Fig. 11 and Table I show very small error for $C_1$, $C_2$, and $C_3$ at 2 mm/s, with a mean tracking error of less than 3 N, which is 1.2% of the 250 N workspace of the bladder actuator. The performance at 2 mm/s on $C_4$ has greater error, which was expected because the stiffness is double that of $C_1$ and $C_2$, which excites the unmodeled dynamics in the controller and lowers controller performance. At the rate of 4 mm/s, the tracking error remains under a mean of 3.5 N except for $C_4$ with a mean tracking error of 5.4 N. Tracking error increases at the 8 mm/s compression rate. At this point, the controller performance begins to decline on all four curves as the unmodeled dynamics become more relevant. Tracking $C_3$ gives the best results because at a constant compression rate, the change in the reference signal, $F_{des}$ remains low compared to the other curves with greater stiffness. Finally, the results for the 20 mm/s compression rate are tabulated. This compression rate was chosen to compress the bladder through its 20 mm workspace in 1 second, which is the correct order of magnitude of speeds expected during walking. The same general trend holds where $C_3$ results in the best performance due to $F_{des}$ experiencing the lowest dynamics. $C_4$ performs the worst, where the mean tracking error is up to 35.5 N, or 14.2% of the bladder workspace. The performance in $C_4$ at 20 mm/s can be explained by saturation of the flow into and out of the actuator for such a dynamic trajectory. Future improvements in flow capacity would improve the 20 mm/s results.

The 20 mm/s results also show some hysteresis effects that arise from limited actuator bandwidth. The results in Fig. 7 show the actuator closed-loop bandwidth reaches up to 7.3 Hz. Recall that actuator bandwidth should be compared to a value of 10 Hz, where most vertical ground reaction forces were observed. The lower bandwidth means that hard heel contact will be slightly softer than desired. Bandwidth could be improved by increasing flow capability by either decreasing flow resistance or increasing pressure differential. A higher pressure applied to the input and a vacuum instead of atmospheric pressure on the output could be applied. Applying a vacuum to the output has been shown to have success in [32].

The heel strike transient discussed earlier in this work (reaching up to 75 Hz in [11]) could be addressed in future work. One possible solution would be to superimpose an open-loop transient over the control output to provide a transient response to the user [1]. A subject study would be required to determine if this is necessary and if the resulting open-loop transient is sufficiently close to a heel strike transient during gait.
Fig. 11. The results of the impedance controller on the four workspace trajectories, C1, C2, C3, and C4. Light green shows the haptic workspace, black is the desired trajectory, and orange is the measured force and displacement.

Fig. 12. The desired flow output from the controller as well as desired and measured force during the 2 mm/s compressions on curve C3.

Fig. 13. The desired flow output from the controller as well as desired and measured force during the 20 mm/s compressions on curve C3.

Fig. 7 makes sense intuitively and further illustrates the trade-offs in this work. With increasing PWM frequency, the control bandwidth increases, which will lead to better performance in tracking dynamic trajectories. However, this comes at the cost of potential increases in power consumption and heat while creating distracting audible and haptic sensations.

The results in Figs. 12 and 13 illustrate a few points of interest that occurred while tracking C3. First, even at the 20 mm/s the controller generally tracks well, which is also true according
to Fig. 11 and Table I. Second, in Fig. 13 between 10 and 15 seconds, the desired flow, $Q_{des}$, is approximately $-4 \text{ L/min}$, meaning the actuator is trying to release additional air to decrease force down to zero. However, the mechanical stiffness of the bladder walls results in a force of approximately 20 N, which is a 20 N tracking error. This occurs because the beginning of $C_3$ leaves the actuator workspace (which can be seen in Fig. 8). In fact, this behavior occurred for $C_2$ and $C_1$ as well. This illustrates that future iterations of these actuators could examine designs that minimize wall stiffness forces required to push the lower bound of the workspace closer to zero.

IX. CONCLUSION

This letter demonstrates that active impedance control of a Smart Shoe actuator can mimic terrain encountered during gait. These results push forward the field of haptic terrain display by providing impedance capabilities of rendering fine terrain features. This is accomplished by identification of a force model, input and output flow models, and design of a continuous sliding mode PWM controller for haptic terrain display. Final actuator performance included 7.3 Hz of bandwidth, 20 mm/sec compression rate, a workspace of up to 20 mm range of motion, and 250 N of force output. It was found that the workspace size and bandwidth are sufficient to successfully simulate most terrain features.

Future work should focus on integrating these bladders into a wearable haptic device, which would allow user evaluations. Since this letter used the AFS to measure force and displacement, Smart Shoe embedded systems need to provide sufficient measurement of bladder displacement, air pressure, and forces. The force modelling and controller should be efficient enough for real-time microcontroller implementation. Microcontroller implementation should be verified. Future work could also consider improving bladder bandwidth and workspace.

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