Soft Pneumatic Actuator for Rendering Anal Sphincter Tone

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Abstract—Sphincter tone examination, as part of digital rectal examination (DRE), can provide essential information to support the early detection of colorectal cancer. Mastering DRE skills for junior doctors is always challenging due to the lack of real training cases. We developed a soft pneumatic active actuator, made of a compound of silicone rubber materials, to mimic human sphincter muscles and simulate various anal sphincter tones for the purpose of training. Different pumping actuation (syringe and bellows) and driving mechanisms (linear, stepper, and servo motor) were implemented and compared for their effect on the rendered tones. A further comparison was made with a previous prototype based on a cable-driven mechanism. Both quantitative and qualitative assessments were conducted to evaluate the performance of each mechanism. A differential pressure sensor was used to measure applied pressure on a catheter balloon placed inside the sphincter, comparing the readings with anorectal manometry data obtained from real patients. Qualitative feedback was gathered through a user study with 10 colorectal expert practitioners. Four questions were asked targeting reaction/response time, pressure level, pressure quality, and similarity to a real case. The results show the capacity and limitation of each mechanism, with the one based on a servo motor and a bellows being the most favourably rated.

Index Terms—Sphincter tone assessment, sphincter actuator, soft pneumatic actuator (SPA), soft haptics, pneumatic mechanism, cable-driven mechanism.

I. INTRODUCTION

HAPTICS is the science of touch-base interaction between a human and a computer application. Combined with anatomical and diagnostic knowledge, it has the potential to revolutionize medical education. Typically, students devote several years for mastering fine motor skills [1]. Haptic simulators have been introduced to address a growing need for effective training and evaluation of clinical skills [1]. In a wide range of medical disciplines and professions, such simulators can be applied [2]. This technology can complement traditional training approaches, especially where hands-on training is not applicable or is unethical. Conventional haptic technologies rely on the user manipulating an external device or being constrained by an inflexible joint [3]. However, this is incompatible with the soft and flexible nature of human bodies. To match the mechanical behaviour of the human body and in order to provide a seamless experience for haptic interactions, it would be appropriate, in this particular application/framework, for haptic devices to be soft and flexible. Soft feedback can be applied to the user in many ways, including through soft pneumatic actuators.

The main difference between rigid and soft actuators is that conventional rigid haptic interfaces can be used in many applications, they are general purpose tools and the rendering can be associated to many suitable applications [4]. However, in terms of soft haptic interfaces, they are usually built and customized for a specific purpose/application. Another difference is that, while rigid haptic interfaces can be very expensive, soft interfaces may be more affordable.

Soft haptics is gaining popularity among researchers due to the technological advances in the field of soft actuators. There are a variety of soft actuator system approaches that operate with different principles, such as pneumatic, dielectric, and magnetorheological fluid [5]. Among them, soft pneumatic actuators (SPA) consisting of a pneumatic chamber, a frame constraint structure and a body made of flexible materials are of particular interest because they are lightweight, affordable and easily customized to a given application [6], [7]. They enable the achievement of safer and flexible interaction in a wide range of applications such as mobile exploratory robots, assistive wearable devices, and rehabilitative technologies [7], [8]. These actuators can achieve combinations of contraction, extension, bending and rotary motion and can be rapidly fabricated in a multi-step moulding process [7]. McKibben muscle (or the pneumatic artificial muscle) was introduced in [9] and used as part of a soft glove for haptic feedback in [10]. Other soft wearable devices include a wrist for haptic feedback [3], a tactile finger [11], and a multi-finger haptic palpation actuator [12]. The medical application of soft pneumatic actuators has been studied for rehabilitation [13], haptic-enabled medical simulators [14], and human assist technology [15]. Most of these applications use a compressor or syringe based mechanism to control the pneumatic actuators. To date, actuators for rendering anal sphincter tone have not been investigated or developed.

Many conventional pumping techniques can drive air to inflate or deflate chamber-based pneumatic systems. Proportional pressure regulators require air supply and the proximity of a wall-mounted air inlet or a bulky and noisy compressor. Small size peristaltic pumps have been widely used, but the flow rate is too low to generate a fast response as required for our application. In many applications where syringe pumps are used to inflate air chambers, stepper motors are often preferred to servo motors as the driving mechanism [20], [21]. On the one hand, stepper motors have precise increments in movement, enabling excellent speed control and providing a constant holding torque without the need for the motor to be powered. On the other hand, they have a few limitations which can cause implementation and operational issues: they lose a significant amount of their torque as they approach their maximum driver speed, sometimes resulting in skipped steps and a loss in position. Besides, they generate significant noise during operation. Servo motors can generate a higher peak torque than stepper motors as they are able to operate at higher speeds. Their main drawback is that the feedback mechanism in the servo will constantly try to correct any drift from the desired position. This constant adjustment may result in twitching while trying to hold a steady position. Replacing syringe based systems with bellows can simplify the mechanism as it reduces any mechanical backlash of the timing belt, crankshaft or lead screw mechanism.

Comparing and choosing among all the driving mechanisms is hence of crucial importance as it may affect the haptic interaction of the pneumatic sphincters on the examining finger.
The simulation results (of the numerical sphincter model that surrounded a finger, as per the Fig. 2). In the simulation, air pressure was applied in the chambers to bond well to Dragon Skin series. The latter is a relatively stiff material that has the advantage of two independent air chambers that can be inflated separately to deliver higher pressure as well as an improved bonding between the soft material for the inner chambers and the hard material for the outer casing. The inner ring which embeds the air chambers is made of silicone with two hollow air chambers each with a pipeline (left). Air pressure is independently applied to the chambers via a tube. When inflated, the chambers expand and push the inner wall of the actuator against the anal canal of the sphincter model (right).

In this paper, we continue and extend our previous work [16] which aims at developing a low-cost, low-power and high torque soft pneumatic actuator for rendering several anal sphincter tones (pressures).

Firstly, we present the design and simulation of an improved sphincter model for delivering higher pressure as well as an improved mould design for easier fabrication. We then study and compare the performance of several pumping systems (syringe, pipette or bellows) and driving mechanisms (linear, stepper, or servo motor). Generated pressure is measured and compared against reference profiles obtained from real patients anorectal manometry. Finally, a group of clinicians tests and evaluates the system. Their feedback further shows promising correlation between the model and real patients sphincter tones when using a servo with a bellows.

II. SETUP

In this section, we first review the design and fabrication of the soft pneumatic actuator and then present the various pumping and driving mechanisms implemented.

A. Sphincter Fabrication

The human external anal sphincter is formed of two symmetrical muscles wrapped around the anus. They are thick, red voluntary muscles whose function is to squeeze the anus and the anal canal [17]. With this in mind, we designed a pneumatic sphincter that consists of two independent air chambers that can be inflated separately to better mimic human sphincter muscles (Fig. 1).

The sphincter model is based on a multi-layer design using two different silicone rubbers, a soft material for the inner chambers and a hard material for the outer casing. The inner ring which embeds the air chambers is made of Dragon Skin 20 (Smooth-On Inc., Easton PA.) by mixing equal weight of the two-part silicone rubber part A and part B. For the outer ring, Mold Star 16 Fast (1A:1B mix ratio) was used to fix boundary conditions by locally rigidifying the structure. The latter is a relatively stiff material that has the advantage to bond well to Dragon Skin series.

The choice of the elastomer compounds and the geometry used for the realisation of the sphincter model was determined by finite element analysis (FEA) design with the SolidWorks simulation package (Fig. 2). In the simulation, air pressure was applied in the chambers of the numerical sphincter model that surrounded a finger, as per the rectal examination. The simulation results (i.e. the maximum pressure exerted on the finger) were compared with real data of anorectal manometry obtained clinically on patients. Several iterations of the computation were needed by modifying the structure of the soft actuator, its size and the geometry of the air chambers, in order to get an optimal design. The simulation model was validated when a good correlation between simulation and experimental results was achieved. As opposed to the initial design presented in [16], where an anal canal made of soft material was used to apply the force on the finger, the proposed design was improved (geometry of the chambers and their relative position to the inner wall) to ensure that the pressure of the air chambers was directly applied on the finger. The material used for this new design was found by expert colorectal surgeons to create a better haptic perception to render human anal muscles.

Confining air chambers fabrication required a multi step process [7], [18], [19], with a specific order as illustrated in Fig. 3. To this end, two sets of moulds were designed, one for the inner chambers and the other for the outer casting. The inner mould was designed to include two kernels held with minimal support and with each kernel having one pipeline as inlet. The support was used to remove the chambers, after pouring the silicone material. The outer mould is intended to further limit the expansion in the outward radial and axial directions. In addition, it also firmly seals the air pipes to the chambers. The moulds were designed in Fusion 360 (Autodesk, San Rafael, CA, USA) and printed in PLA using a PRUSA i3 MK3S 3D printer (Prusa Research, Prague Cz). To avoid inhomogeneous inflation of the air chambers caused by the possible presence of air bubbles being trapped inside the cured silicone, the mixtures were subjected to a vacuum during the pot life of silicone once poured into the mould and cured at room temperature. Fig. 4 illustrates the fabrication process of the pneumatic sphincter used in this study.

B. Pumping mechanisms

Pneumatic actuators require a mechanism to pump air in and out of the chambers, resulting in inflation and deflation. As our final simulator should be used independently of direct access to an air inlet, and as the noise should be minimal, solution using compressors were not investigated. Our starting point was a disposable syringe as a basic and affordable option. Here, the air flow is controlled by the back-and-forth movement of the plunger. Travel distance is important considering a motorized system to push or pull the plunger. The larger the syringe, the shorter the travel and the faster the response. After preliminary tests, syringes of 100 ml capacities were selected with regard to the response time.

Plastic transfer pipettes are possible candidates for an alternative to syringe pump based systems (Fig. 5). They are made of an air bulb connected to a narrow stem.
1) Assembling the moulds for creation of the inner part embedding the air chambers.
2) Pouring Dragon Skin 20 silicone mixture and curing.
3) Extracting each chamber kernel from the small aperture and then sealing the apertures with a thin layer of Dragon Skin 20 silicone.
4) Pouring Mold Star silicone mixture in a bigger mould and curing.
5) Final sphincter model.

While the syringe pump and pipettes are rudimentary, the motorized systems required to actuate them could be potentially complex and expensive. A third possible option considered is a special type of pipette known as bellows pipette (Fig. 5). In this case, the air bulb is made of a bellows and the air flow is controlled by the contraction and expansion of the bellows. Any basic motorized system, such as a servo motor, could be easily utilized to press the bellows. For a smaller form factor, we used a bellows dispenser with a short stem (Fig. 5).

C. Driving mechanisms

Different mechanisms to either move the plunger or press the bellows were tested for the purpose of identifying which mechanism better fulfills the speed and pressure requirements. These requirements are directly related to the pressure profiles of the sphincter tones we intend to render using our sphincter actuator. Four conditions (tones) have been considered: healthy, decaying, weak, and cough reflex with their pressure profiles given in Fig. 6 [16]. These pressure profiles were obtained through anorectal manometry of various patients and generalized into a reference profile for each condition [16]. Among these profiles, cough reflex sets up our boundary conditions as it requires the fastest and highest peak pressure response.

The different mechanisms (pump + driver) implemented and tested are illustrated in Fig. 7. In addition, an example of actuated sphincters is given in Fig. 8. For the syringe-based pumping, linear movement is required to move the plunger. Widely used options include linear motor and stepper motor with lead-screw or timing belt transmission. The former is a natural choice for linear movement and the latter has been extensively used for precise linear movements. Most 3D printers use stepper motors to achieve a fine movement with a resolution of 0.1 mm. However, in common practice if the design allows, rotary movement is preferred over linear movement as it significantly reduces the complexity of hardware components and electronics. For the bellows-based pumping, rotational movement is adequate. Thus, a crankshaft mechanism actuated by a servo motor can be effective in pressing a confined bellows.

1) Syringe pump - linear motor: An Actuonix P16-50-22-12-P linear motor (Actuonix Motion Devices Inc, Victoria, BC Canada) with stroke length of 50 mm, maximum speed of 46 mm/s, and maximum force of 50 N when operated under 12 VDC was used. The motor comes with its dedicated controller (Linear Actuator Controller Board, ACTUONIX, Canada) with options to command the motor through an analogue or PWM input. The motor shaft was coupled to a 100 ml syringe plunger with the base, coupler, and holder 3D printed (Fig. 7(a)). Since our sphincter has two air chambers, two sets of linear motors were required. Each chamber was connected to

Fig. 3: Fabrication steps of the pneumatic sphincter. 1) Assembling the moulds for creation of the inner part embedding the air chambers. 2) Pouring Dragon Skin 20 silicone mixture and curing. 3) Extracting each chamber kernel from the small aperture and then sealing the apertures with a thin layer of Dragon Skin 20 silicone. 4) Pouring Mold Star silicone mixture in a bigger mould and curing. 5) Final sphincter model.

Fig. 4: The fabrication of the pneumatic sphincter. (a) 3D printed inner mould, (b) pouring silicone, (c) removing kernels, (d) 3D printed outer mould, (e) pouring silicone, and (f) final product.

Fig. 5: Various pumping options. (a) 20 ml syringe, (b) 50 ml syringe, (c) 100 ml syringe, (d) 5 ml pipette, (e) 5 ml pipette with shorter stem, (f) 5 ml bellows pipette, and (g) 30 ml bellows dispenser. Our final choices were (c) and (g).

Fig. 6: Pressure profiles from left to right: Healthy Squeeze (HS), Decaying Squeeze (DS), Cough Reflex (CR), and Weak Squeeze (WS). The x-axis refers to time in seconds whereas the y-axis refers to the degree of relax/squeeze in mmHg.
2) Syringe pump - stepper motor: A NEMA 17 stepper motor (Stepperonline, China) with a holding torque of 59 Ncm was used, along with a 10 mm lead-screw to convert motor rotation to linear movement. Each motor was driven by a stepper driver (TMC2208, Watterott, Germany) for reliable and silent performance, and connected to a separate voltage source (24 VDC) in order to guarantee smooth and stable movement. Same as before, each chamber was connected to a separate motor. The same Arduino Uno as before was used to send commands (steps and directions) to the stepper drivers. The command profiles were generated by estimating the number of steps required to move forwards or backwards at any given time stamp.

3) Bellows pump - servo motor: A high torque servo motor HS-5685MH HV Ultra Torque (Hitec RCD USA, Inc., Poway, CA, USA) was chosen to press/release a 30 ml bellows pipette. From each servo, operating under 7.4 VDC, a maximum torque of 126 Ncm and a speed of 0.17 S/60° are achievable. We identified the maximum and minimum servo position (angle) to press and release the bellows. These values were used to linearly transform pressure profiles to command profiles. Similarly, each chamber was connected to a separate motor and the same Arduino Uno was used to control both motors. The command profiles were sampled at a frequency of 10 Hz and imported as arrays to the Arduino script.

4) Cable-driven mechanism: The design of this cable-driven mechanism was introduced previously in [16] (Fig. 7(d)). It includes two 900-00005 servo motors (Parallax, Inc., Rocklin, CA, USA) with a maximum torque of 27 Ncm at 6 VDC. A silicon sphincter was made to hold the cables in position, as well as to accommodate the examining finger.

III. EVALUATION

To evaluate the performance of the developed actuators, both quantitative and qualitative assessments were conducted by measuring sphincter pressure using a pressure sensor and comparing the readings with reference profiles, and by a user study with expert colorectal practitioners who were asked four subjective questions. The experimental set-up, including both pneumatic and cable-driven actuators with their corresponding pumping/driving mechanisms/electronics, is shown in Fig. 9.

A. Pressure measurement

The pressure generated by each sphincter mechanism was measured using an MPX5050GP pressure sensor (NXP, Eindhoven, Netherlands) in range 0-50 kPa with a sensitivity of 90 mV/kPa, attached to an anorectal manometry catheter (CAT003, Laborie Medical Technologies, Canada) (Fig. 9). The catheter balloon was pre-inflated by a syringe and inserted inside the sphincter actuator. By pumping air into the chambers, the balloon was squeezed and its pressure was recorded by the sensor given above. An Arduino Mega 2560 was used to read the pressure values at a rate of 20 Hz. For each mechanism, each of four pressure profiles was rendered three times and corresponding pressure values were recorded and averaged.

B. Human user study

A human user study was conducted to assess the quality of the rendered tones. Ten clinical participants (nurse practitioners in continence and colorectal surgeons) with prior experience assessing sphincter tone were recruited. The participants were asked to wear medical gloves and insert their index finger inside the sphincter. Each test, commanding the motor to move and observing/experiencing the inflation level, the travel distance for the maximum required pressure was marked.
of four pressure profiles was then rendered twice by each mechanism. This was followed by the participant answering all four questions for the particular mechanism. The user study was approved by the Imperial College Ethics Committee (EERP1819-054). Informed Consent was obtained from each participant prior to their participation and standard procedures were implemented regarding anonymity and confidentiality.

The experimental set-up for the user study is shown in Fig. 10. All hardware components and electronics were covered and hidden from participants to reduce distraction. In addition, both sphincter actuators were enclosed in white boxes. Participants were asked to sit on a chair in front of the experimental set-up with their finger inserted inside a sphincter. The order of four set-ups (linear-syringe, stepper-syringe, servo-bellows, and cable-driven) was randomized for each participant. For a chosen set-up, the sequence of four pressure profiles (HS, DS, CR, and WS) was kept same and repeated twice. After finishing a set-up, participants were given time and instructed to answer a questionnaire with four questions (Table I) to be rated on a scale of 1 to 5. The questions were for each different mechanism in general and not for each profile. As explained earlier, the speed and the pressure level were the two most important requirements the. Hence, Q1 and Q2 directly targeted these features. Q3, the pressure quality (localized vs. distributed), was mainly included to highlight the differences between cable-driven and pneumatic. The last question, Q4, is for overall assessment with respect to real cases.

IV. RESULTS AND DISCUSSION

The results of both quantitative (pressure measurement) and qualitative assessments (user study) are presented and discussed in this section.

A. Quantitative results

The pressure readings for each mechanism are shown in Fig. 11. Several observations can be made comparing readings (solid blue line) and reference profiles (dashed red line). All pressure levels, except for weak squeeze, are below the corresponding reference pressures. This means none of the mechanisms could achieve the intended maximum pressure. However, the differences across mechanisms were not the same. In terms of pressure level, servo-bellows was able to achieve the highest pressure, followed by linear-syringe, cable-driven and finally stepper-syringe. In terms of profile shape, cable-driven had the closest shape, followed by stepper-syringe, servo-bellows and lastly linear-syringe. In the case of WS, all mechanisms could perfectly follow and generate the same reference profile.

With the linear motor, the output pressure is jerky and fluctuating around a constant pressure. This if more obvious in HS profile. A potential reason could be that, when the air chambers are fully inflated to reach the maximum pressure level, the resistive load is higher than what the linear motor can handle. A similar behaviour is observed with the finger inside sphincter. In a no-load condition, the syringes not connected to the pipes, the motor could stably maintain the high pressure position. With the bellows pump and servo motor, the shape of HS and DS are slightly different than the references. In HS, the overshoot is small and in DS, the decay starts with some delay. It seems that the pressure readings are somehow saturated. Re-checking the values, we realized that they are all within the maximum range. Hence, this saturated behaviour did not come from the sensor side. We speculate that it comes from the servo motor itself. Most likely, the maximum resistive load from the air chambers is higher than what they can continuously and reliably handle.

Among the CR profiles, the one resulting from the stepper motor is the smallest. Not only it could not reach the peak value, but also it is wider. This is because of the stepper motor speed; it cannot rotate fast enough. In these motors, the speed and torque are inversely proportional; the higher the speed, the lower the torque and vice versa.

An additional comparison between the mechanisms is given in Table II. As can be seen, servo-based mechanisms are electrically

![Fig. 9: Experimental set-up components. (a) electronics including micro-controllers and motor drivers, (b) pressure sensor connected to catheter balloon, (c) servo-bellows mechanism, (d) linear-syringe mechanism, (e) stepper-syringe mechanism, (f) pneumatic sphincter with catheter balloon, and (g) cable-driven sphincter actuator. As it can be seen, two sets of motors have been used for each mechanism.](Image)

![Fig. 10: Experimental Study set-up: pneumatic sphincter actuator (left) and cable-driven sphincter actuator (right). The mechanisms were covered from participants to reduce distraction.](Image)

<table>
<thead>
<tr>
<th>TABLE I: Four subjective questions of the user study.</th>
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<tbody>
<tr>
<td>Q1. Response/reaction time?</td>
</tr>
<tr>
<td>1: Very slow (not realistic)</td>
</tr>
<tr>
<td>Q2. Pressure level on finger?</td>
</tr>
<tr>
<td>1: Very low (not realistic)</td>
</tr>
<tr>
<td>Q3. Quality of pressure?</td>
</tr>
<tr>
<td>1: Very localized (not realistic)</td>
</tr>
<tr>
<td>Q4. Overall similarity to real case?</td>
</tr>
<tr>
<td>1: Very low (not realistic)</td>
</tr>
</tbody>
</table>
TABLE II: Additional comparison between the mechanisms.

<table>
<thead>
<tr>
<th>Feature</th>
<th>Linear</th>
<th>Stepper</th>
<th>Servo</th>
</tr>
</thead>
<tbody>
<tr>
<td>Driver</td>
<td>Dedicated</td>
<td>Common</td>
<td>None</td>
</tr>
<tr>
<td>Power</td>
<td>12 VDC</td>
<td>24 CDV</td>
<td>6-7.4 VDC</td>
</tr>
<tr>
<td>Torque</td>
<td>High</td>
<td>Medium</td>
<td>High</td>
</tr>
<tr>
<td>Speed</td>
<td>Medium</td>
<td>Medium</td>
<td>High</td>
</tr>
<tr>
<td>Form factor</td>
<td>Large</td>
<td>Large</td>
<td>Small</td>
</tr>
<tr>
<td>Weight</td>
<td>Medium</td>
<td>Heavy</td>
<td>Light</td>
</tr>
<tr>
<td>Noise level</td>
<td>High</td>
<td>Low</td>
<td>Low</td>
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<tr>
<td>Price</td>
<td>High</td>
<td>Medium</td>
<td>Medium</td>
</tr>
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B. Qualitative results

Fig. 12 summarises answers from all ten participants. Servo-bellows achieved the highest accumulated score (15.60), whereas stepper-syringe had the lowest (12.60). Mechanisms with servo motor, servo-bellows and cable-driven, achieved the highest score for response time (4.10). The highest score for pressure level was registered for servo-bellows (3.60), followed by linear-syringe (3.50) and cable-driven (3.50). Regarding pressure quality (Q3; localized vs. distributed), on average, pneumatic sphincter scored higher than cable-driven with 3.50 vs. 3.20. For the overall similarity (Q4), servo-bellows rated the highest with score of 3.70, followed by linear-syringe (3.20), then by stepper-syringe and cable-driven, both with a score of 2.90. Such results seem to suggest that the perceived feeling of servo-bellows mechanism is the most favourable of all. On the other hand and as expected, the cable-driven mechanism was regarded as the least favourable due to its localized and less-realistic haptic feedback.

These qualitative results are in agreement with the quantitative results that highlighted the speed limitation of the stepper motor. As a result, the stepper-syringe is regarded as the one with the lowest response time. The jerky behaviour of the linear motor is reflected in the pressure quality (Q3) by being the second lowest. From Fig. 11, it can be seen that the pressure level of the stepper-syringe mechanism, particularly for CR, is not as high as others. This has been reflected in Q2 by being the lowest amongst all.

We have also conducted a statistical analysis to investigate the statistical significance of the results. 1-way ANOVA was applied to each question separately and the p-values and box-plots are given in Fig. 13. As it can be seen, for questions Q1, Q2, and Q4, the p-values are larger than 0.05 meaning that there was no statistical significant difference between the mechanisms. For Q3, there is a marginal trend towards significance. It appears that one of driving and/or pumping mechanisms (most likely SB) has the tendency to make a difference in the answers to Q3. While the mean values for LS, SS, and CD are equal to 3.0 (neutral: neither localized nor distributed), it is 4.5 for SB (realistically distributed). However, we expected a low mean value, close to 1.0, for CD as the cable tends to feel more localized, and a large mean value, close to 5.0, for the rest, as pneumatic chambers tend to feel more distributed. A possible reason for this results could be that the participants misinterpreted the meaning of Q3 (localized vs distributed) and mixed it with Q4 (overall similarity).
Roger Kneebone, Dr. Carmen Sofia Chacon, Dr. Anna Przedlacka, realistic simulation environment”. We also would like to thank Dr.
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Detection Project [Award Reference: C63674/A26161] - “Improving 
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As future work, we are planning to test a miniature electrical pump 
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We have developed a soft pneumatic actuator to render anal 
the intention was to create more realistic haptic feedback for the purpose of training junior doctors. Various pumping 
and driving mechanisms have been developed and tested. In addition, 
comparisons were made with a previously developed cable-driven 
and qualitative assessments (user study with clinical participants) 
were conducted. The pressure readings revealed that almost all the 
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We have developed a soft pneumatic actuator to render anal 
and driving mechanisms have been developed and tested. In addition, 
comparisons were made with a previously developed cable-driven 
mechanisms could follow and produce the reference pressure profiles. 
Even the cough reflex, which required a fast response and a high 
torque, was successfully regenerated. Through a user study with 
clinicians, it was shown that the servo-bellows mechanism (SB) was 
the most favourably rated in terms of re-creating sphincter tones that 
are more similar to the real ones. Furthermore, an ANOVA analysis, 
plied over each question, revealed that the difference between 
mechanisms was not statistically significant, except for Q3, which 
was marginal. SB is also electrically less demanding, has a smaller 
form factor, lower noise and is reasonably priced. 

As future work, we are planning to test a miniature electrical pump 
for its response time and pressure level, as well as conduct further 
experiments comparing male and female pressure profiles with a 
larger pool of clinicians.

V. CONCLUSIONS AND FUTURE WORK

ACKNOWLEDGMENT

Fig. 13: Summary of statistical analysis. P-values are reported applying 
1-way ANOVA to each question.

Q1 (p-value = .41)

Q2 (p-value = .48)

Q3 (p-value = .05)

Q4 (p-value = .15)

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REFERENCES

HapticsTouch feedback technology widening the horizon of medicine. 

Haptics in medicine and clinical skill acquisition [special section intro.]. 

waist device for kinesthetic haptic feedback. Frontiers in Robotics and 
AI, 5, 83.

Multidisciplinary Digital Publishing Institute.

A soft pneumatic actuator as a haptic wearable device for upper limb amputees: Toward a soft robotic liner. IEEE Robotics and 

“Design of soft multi-material pneumatic actuators based on principal 

J. (2016). Modeling, design, and development of soft pneumatic actuators 
with finite element method. Advanced engineering materials, 18(6), 978-988.

pneumatic actuator fascicles for high force and reliability. Soft robotics, 
4(1), 23-32.

McKibben pneumatic artificial muscles. IEEE Transactions on robotics 
and automation, 12(1), 90-102.

(2017). Soft robotic glove for kinesthetic haptic feedback in virtual reality 
environments. Electronic Imaging, 19-24.

tactile rendering for wearable haptics. In IEEE World Haptics Conference 
(WHC) (pp. 327-332).

medical simulators: A review. IEEE Access, 6, 3184-3200.

Multidisciplinary Digital Publishing Institute.

[14] Li, M., Luo, S., Nanayakkara, T., Seneviratne, L. D., Dasgupta, P., 

assist technology. In Symposium on fluid power.

[16] Marechal, L., Granados, A., Ethapami, L., Qiu, S., Kontovounisios, C., 
based on pneumatic and cable-driven mechanisms. In IEEE World Haptics 
Conference (WHC) (pp. 376-381).

http://ddc.musc.edu/public/diseases/colon-rectum/ 
anal-stenosis.html

[18] Ilievski, F., Mazzeo, A. D., Shepherd, R. F., Chen, X., & Whitesides, G. 
Edition, 50(8), 1890-1895.

[19] Li, Wanying, Fabrication of Soft Robotic Actuators by Using Injection 
Molding Technology, Master’s thesis, Faculty of the Graduate School 
of Cornell University, 2017

[20] Lindenroth, L., Soor, A., Hutchinson, J., Shafi, A., Back, J., Rhode, K., 
Liu, H. (2017, September). Design of a soft, parallel end-effector ap-
plied to robot-guided ultrasound interventions. In IEEE/RSJ International 
Conference on Intelligent Robots and Systems (IROS) (pp. 3716-3721).

[21] Peters, J., Nolan, E., Wiese, M., Miodownik, M., Spurgeon, M., Arezzo, 
in fluid-driven soft robots using low-melting-point material. In IEEE/RSJ 
International Conference on Intelligent Robots and Systems (IROS).