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Characterization of Pneumatic Touch Sensors for a Prosthetic Hand

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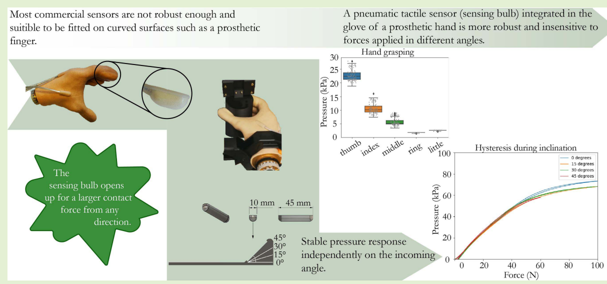
***Abstract*—This paper presents the results from the characterization of pneumatic touch sensors (sensing bulbs) designed to be integrated into myoelectric prostheses and body-powered prostheses. The sensing bulbs, made of silicone, were characterized individually (single sensing bulb) and as a set of five sensors integrated into a silicone glove. We looked into the sensing bulb response when applying pressure at different angles, and also studied characteristics such as repeatability, hysteresis, and frequency response. The results showed that the sensing bulbs have the advantage of responding consistently to pressure coming from different angles. Additionally, the output (pneumatic pressure) is dependent on the size of interacting object applied to the sensing bulb. This means that the sensing bulb will give higher sensation when picking up sharper objects than blunt objects. Furthermore, the sensing bulb has good repeatability, linearity with an error of 2.95±0.40%, and maximum hysteresis error of 2.39±0.17% on the sensing bulb. This well exceeds the required sensitivity range of a touch sensor. In summary, the sensing bulb shows potential for use in prosthetic hands.**



***Index Terms*— Pneumatic, touch sensor, non-invasive, sensory feedback, sensing glove.**

I. Introduction

THE tactile receptors in the human skin are exceptional sensors, the presence of which helps us to interact and explore our surroundings in activities such as manipulation and exploration [1]. There are several types of cutaneous receptors in the skin that detect vibration, force, shear, temperature, and pain. The receptors that respond to mechanical stimuli are called mechanoreceptors, which include Meissner's corpuscles, Pacinian corpuscles, Merkel cells, and Ruffini corpuscles [2]. The mentioned mechanoreceptors detect heavy pressure, vibration, light touch respectively skin stretching [3]. There is high density of mechanoreceptors in the human hand which makes it sensitive to delicate touch where the glabrous skin in the volar part of the hand is more sensitive than the hairy skin on the dorsal part, and the central whorl of the finger pulp being the most sensitive part as it contains the highest density of receptors [4]. In order to perceive different kinds of tactile sensations, all four mechanoreceptor types contribute to the flow of sensory information to the brain where the percepts are formed. Losing a hand entails the loss of thousands of mechanoreceptors but the receptors still remain in the residual limb. However, wearing a myoelectric hand prosthesis, which is a widely used choice among the commercially available prostheses [5], hinders the usage of the remaining receptors in the residual limb and takes away the ability to feel [6]. Thus, leaving the user with only visual input, the sound from the prosthesis, and sensations at the residual limb [7]. For upper limb prosthesis users, providing sensory feedback is highly desired. It has also been shown to improve the motor control of the prosthesis [8] and to reduce the need for visual input. Additionally, it helps the user to adapt to a new prosthesis and to learn how to use it more effectively [9].

The tactile sensing in an artificial hand should be capable of detecting temperature, texture, shape, and force [4]. However, the top priority is to provide feedback that enables grasping (e.g., of an object), touch, and proprioception [7]. Different kinds of sensors have been explored and developed to record sensory input, but few of them have been integrated into commercial prosthetic hands because this often leads to increased cost or to added difficulty during implementation [10].

In prosthetic hands, exteroceptive sensors are used to measure data during interaction with objects and environment. Depending on which sensing techniques are used, detection of normal or tangential forces, vibration, point contact, and temperature can be performed. To be fitted into the prostheses the sensors should meet criteria such as low hysteresis, robustness, and broad dynamic range [11]. The most common

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sensors used in prosthetic hands are force sensitive resistors (FSRs) [10], [12], [13], piezoelectric sensors [14], [15], and capacitive sensors [16]–[18]. Capacitive sensors have good frequency response, spatial resolution, and a wide dynamic range. Such a sensor can detect a normal force by a change in distance between the plates of the sensor or a tangential force by a change in the effective area (overlap of the plates) of the sensor. Nevertheless, it cannot distinguish between these two types of forces [2]. However, they are non-linear, have low accuracy, are more susceptible to environmental noise, and can be influenced by stray capacitance [19]. Resistive touch sensors are often simple, cheap, and have good sensitivity, although, they also have poor repeatability, high hysteresis, and poor frequency response. Furthermore, looking at FSRs, which have a limited active area, the actuator should look a certain way for the FSR to have good repeatability. Typically, the FSR sensors need a mechanical setup that positions the sensing force orthogonal to the FSR surface. The contact area to the FSR sensor should be a flat area that is slightly smaller than the FSR sensing area to provide evenly distributed pressure on the FSR [20]. These characteristics of the FSR make it less recommendable for use in prosthetics hands. This is because the FSR will be placed on a curved surface and the output signal from the FSR will be inconsistent when the force is applied at different angles. A piezoelectric tactile sensor is a suitable candidate to detect slip because it is able to measure vibration due to its very high-frequency response. However, such sensors have limitations. They are only able to measure dynamic force and tend to have poor spatial resolution [1], [2], [21]–[24]. Amongst the sensors mentioned above, FSRs and capacitive sensors are most commonly used to detect contact force. Frequently mentioned reasons are because they are cheap, simple, and lightweight.

Because most commercial sensors only detect a load applied in the center of a sensor, there is now a trend toward making sensors that are flexible and can be attached to a curved surface, such as a fingertip, and also to cover a larger area. Some studies have developed sensing skins [25]–[27], whereas others have developed sensors that use liquids [28]–[30] or air bladders [31], [32]. The latter sensors, integrated into a glove, are constructed to be as anthropomorphic as possible to imitate a real hand. Whether it is on a robotic hand that will interact with humans or on a prosthetic hand, such sensors make the prosthetic more comfortable during interaction and bring to the robotic hand the ability to handle fragile objects. A proposed soft tactile sensor uses a magnet which is immersed in a soft body structured finger, which also consists of a Hall-effect sensor to measure the intensity of the magnetic field generated by the magnet. This sensor detects normal forces [33]. However, such magnetic sensors are susceptible to other interfering magnetic sources and noise [34]. Because of interference of other magnetic objects, integration in robotics is of limited use [35]. The BioTac®tactile sensor (SynTouch, LLC) senses force, vibration and temperature [36] and has been crafted according to the human fingertip to be used in biomimetic systems. In prosthetic hands it has been used to provide closed-loop control of grasping force within the prosthetic hand.

Most commercial prostheses have a silicone glove and in the study by Antfolk *et al.* [31], silicone encapsulated bulbs were developed (hereafter called sensing bulbs) in the shape of fingertips, and were then integrated as a part of the glove. The sensing bulbs cover an area roughly equivalent to the proximal and intermediate phalanx of each finger. Plastic tubes connect the sensing bulbs to other silicone pads that act as a tactile display, which in turn, creates a non-invasive closed pneumatic sensory feedback system. The bulbs of this design spread to provide larger contact force in any direction with an anthropomorphic appearance. We suggest that this kind of pneumatic sensing bulb is robust and insensitive to direction. The design and the integration of the sensing bulbs into the glove is described in the aforementioned study. Consequently, in this study, the tactile properties of the sensing bulbs are characterized so that they can be used in conjunction with processing electronics to provide a finely adjusted transfer function and measure of the force and feedback provided to the user.

The paper is structured as follows. In Section II we briefly summarize some important features of the sensing bulbs that were mentioned in the previous study. Then, in Section III we describe the different ways used to characterize a single sensing bulb and also when the sensing bulbs are integrated into the glove. The results are presented in Section IV, with some further discussion in Section V where the characteristics of the sensing bulbs are compared, theoretically, to other state-of-the-art tactile sensors. A summary of the mentioned sensors can be seen in Table I. Finally, in Section VI, we draw our conclusions and make suggestions for future development.

II. Background: Pneumatic Touch Sensors

The design of this system was similar to that of an earlier study [31] with some minor changes: the sensing bulbs are of different size and are placed in different positions, only covering the distal phalanx of the rubber glove. It is mentioned in the aforementioned study that the sensing bulbs will be placed individually according to the type of prosthesis and also to how the prosthesis is handled by the user.

The sensing bulbs are made of high-temperature vulcanized (HTV) 20 shore silicone with the Young's modulus of 0.843 MPa. The thickness of the sensing bulb is 0.5 mm and the semi-rigid bottom-support is 1.75 mm with a 65 shore silicone. The semi-rigid bottom is to withstand the created pressure in the sensing bulb, resulting in bulging only on the sensing part of the sensing bulb. This eventuates a more accurately reading of the applied pressure. Some conclusions about the design was made in Antfolk et al. study during the development phase [31], where the thickness of the wall has to be compromised between durability and sensibility, therefore, the thickness should not be too thin nor too thick. The durability and the sensitivity were desirable in the design to make the sensing system durable to fit in a prosthetic glove, and meanwhile, provide with enough sensitivity to feed back the sensations. The force is sensed through the sensing bulbs and is mediated using air in a plastic tube that is connected to an actuator, providing the amputee with sensory feedback

TABLE I
THEORETICAL SUMMARY OF THE PROPERTIES OF THE SENSING BULB AND THE COMPARED STATE-OF-THE-ART SENSORS

	Advantages	Disadvantages
Sensing bulb	✓flexible ✓low hysteresis ✓good precision ✓large sensitive area ✓consistent response to applied pressure at different angles	✗low spatial resolution[a] ✗output inverted dependent on size of object[b]
Capacitive sensor	✓spatial resolution ✓frequency response ✓wide dynamic range	✗very low precision[c] ✗non-consistent response to applied pressure at different angles
Resistive touch sensors	✓cheap ✓good sensitivity ✓large sensitive area ✓simple to use/connect	✗low precision ✗high hysteresis ✗output dependent on area size of object[d] ✗non-consistent response to applied pressure at different angles
Piezoelectric sensor	✓high frequency response	✗detects only dynamic force ✗output dependent on area size of object[d] ✗non-consistent response to applied pressure at different angles

[a] With the current design, spatial resolution is bad as the sensor is fairly large.
[b] Smaller area larger response, which actually could be an advantage, see discussion.
[c] Low precision due to; highly non-linear, susceptible to environmental noise. Often used as on/off sensor.
[d] Larger area larger response.

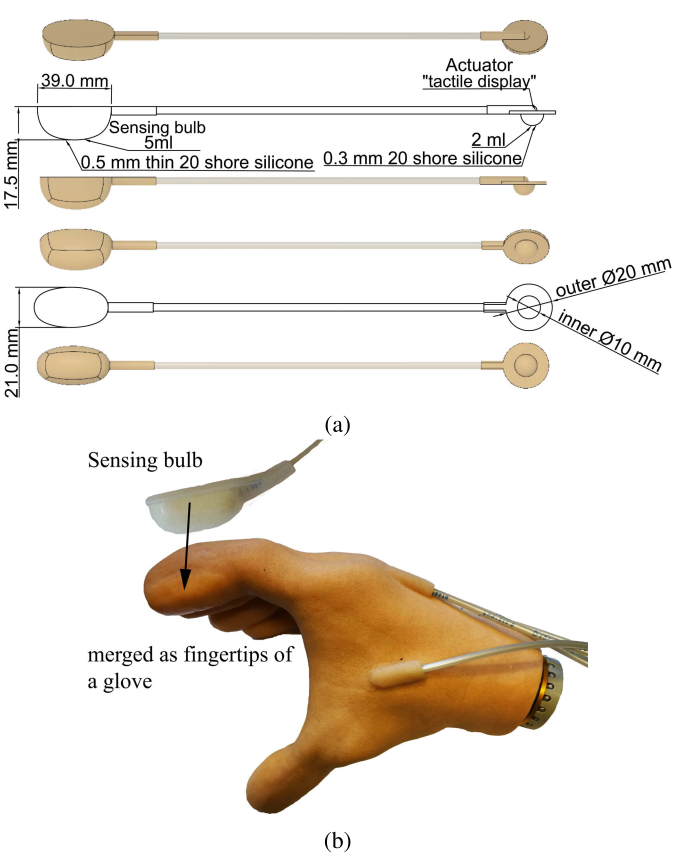

Fig. 1. (a) Pneumatic Sensing System: The sensing system is seen in different angles with dimensions. The actuator is seen in its active state, i.e. applied pressure on the sensing bulb makes the actuator bulge against the skin. When there is no pressure on the sensing bulb, the actuator is flat. (b) Sensing bulb: Top image, a single silicone bulb (not integrated into glove), Bottom image, glove with merged sensing bulbs in the fingertips for the MyoHand VariPlus.

on the residual limb when the silicone bulge against the skin (Fig. 1a). The sensing bulb is shaped to imitate a real fingertip (distal phalanx) (Fig. 1b) and having polyurethane foam inside to make the sensor stiffer and to facilitate a quick return to its initial form (e.g., after an object is released).

In this study the sensing bulbs were characterized by different experimental setups for an individual sensing bulb (top image in Fig. 1b) and for a prosthetic glove with five integrated sensing bulbs.

III. CHARACTERIZATION METHODS

The set-up of the experiments contained i) a VEXTA PK254-E2.04A stepper motor (Oriental Motor Co.,LTD., USA) having a step size of 1.8°/step ii), a force gauge (MARK-10, USA) with accuracy of $\pm 0.1\%$ full scale and a resolution of 0.02 N. This was mounted on iii) an miniature linear positioning system (Parker 402002, Parker Hannifin Corporation, Pennsylvania) that drove the force gauge on the z-axis to press on the sensing bulb to make it deflate and inflate (Fig. 2a). The stepper motor was driven using a NI MID-7604 stepper motor driver (National Instruments, USA). A setting of 10 microsteps per step was using on the motor driver. The linear positioning system uses a 1 mm lead screw, leading to a displacement of 500 nm/step of the system. The speed of the positioning system during the experiments were set at 5 μm/s. The speed is lower than the normal grasping speed, but could eliminate the rate dependencies.

Different sizes of objects with rectangular surfaces were attached to the shaft of the force gauge to act upon the sensing bulb. An integrated pressure sensor (MPXV5100G, NXP Semiconductors, The Netherlands) with accuracy of $\pm 2.5\%$ V_{FSS} and a sensing range of 0-100 kPa, was used to measure the pressure from the sensing bulb. The data was analyzed to evaluate the characteristics of the sensing bulb, such as, hysteresis, frequency response, and repeatability. Experiments, explained in section Sections III-A to III-C, were done on a single sensing bulb. While, for the experiment, in section III-D, tests were made on five different sensing bulbs that were integrated into a silicone glove [31] covering a MyoHand VariPlus Speed (Otto Bock, Germany).

LabVIEW, a visual programming language and environment, was used to control the measurement setup. Multiple tasks were done in LabVIEW: control of the stepper motor

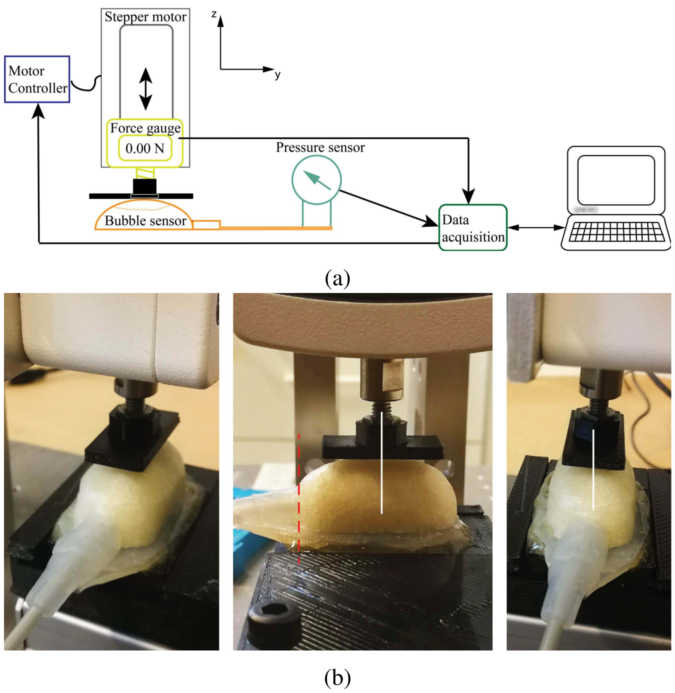

Fig. 2. (a) Experimental setup for Force-pressure characterization of the sensing bulb: Motor controller moves the gantry with the attached force gauge on the z-axis to press the sensing bulb and to measure the force applied to the sensor. The pressure sensor measures the pressure induced from the sensing bulb in the silicone tube. Further analysis was done after data acquisition. (b) Photograph of the positioning of the sensing bulb and the indenter. The center of the indenter was inlined with the center of the sensing bulb as the white line. However, the indenter should not exceed the dashed red line which is the attachment to the plastic tube.

in the z-axis, gathering of the data from the force gauge and from the pressure sensor using a NI USB-6341 DAQ device (National Instruments, USA). The DAQ device was also used to control the prosthetic hand in the second part of the measurement, which included a fully sensorized glove. The update rate of the experimental setup was set to 10 Hz.

Post-processing and visualization of data were conducted using Python with packages such as Pandas (https://pandas.pydata.org/). From the measured output voltage of the MPXV5100G sensor [37], the pressure was calculated according to equations supplied in the datasheet.

A. Setup: Compression With Indenters of Different Sizes

This experiment was done on a single sensing bulb, to evaluate how the sensing bulb behaved when it was compressed using objects of different sizes. The objects were rectangular plates, which acted as indenters in the experiments. The indenters were 3D-printed, using PLA filament (Young's modulus, 2.960 MPa), according to the sizes displayed in Table II. The size of indenter no. (v) was the same as that of the base of the sensing bulb, but when the sensor was compressed, the silicone of the sensing bulb widens, exceeding the size of the indenter. Therefore, an indenter twice the size (indenter no. (vi)) was chosen to cover the volume exceeded. The stepper motor was set to move in increments of one step during loading and unloading. The indenters were changed after each measurement, but the sensor remained

TABLE II
DIFFERENT INDENTERS

No.	Indenter surface area regardless of the area of the base of the sensing bulb	Base Area (mm^2)
(i)	12.5%	100
(ii)	25%	200
(iii)	50%	400
(iv)	75%	600
(v)	100%*	800
(vi)	200%	1600

* the same size as of the base of the sensing bulb.

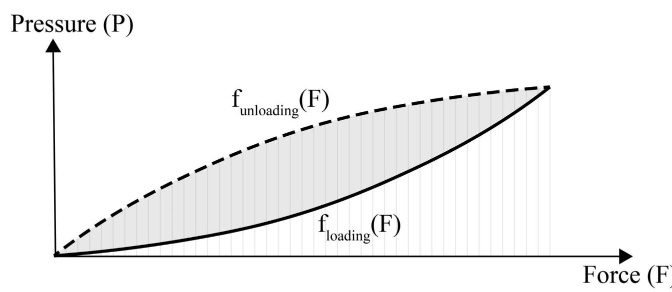

Fig. 3. Integral hysteresis definition: An approximated area is calculated between the two functions $f_{loading}$ and $f_{unloading}$ using a step size of 0.5 N. To get the error in percentage the approximated area is divided by the area for the function f_{loaded}.

at the same fixed position. The stepper motor slowly lowered the force gauge to apply pressure on the sensing bulb until the force reached 100 N (loading), and thereafter raised the force gauge until reaching 0 N (unloading). The slow pace of changes in pressure ensured isothermic conditions within the air compartment and the silicone membrane. It was chosen to apply pressure where the force gauge read up to 100 N, even though no more than 20 N is required for sensitivity range [4]. This was done to examine the characteristics of the sensing bulb under an extreme condition.

For the hysteresis evaluation, both loading and unloading phases were included. The hysteresis was only calculated for the three biggest indenters, (iv)-(vi), because of their linearity in the force range 0-100 N. Hysteresis was calculated for each indenter. The hysteresis was calculated according to the equation below:

$$hyst\% = \frac{\int_{x=0}^{100} f_{unloading}(x)dx - \int_{x=0}^{100} f_{loading}(x)dx}{\int_{x=0}^{100} f_{loading}(x)dx} * 100$$

The area between the function of loading and unloading state was calculated with the integral definition, which was divided by the loading area to get the hysteresis error in percentage (Fig. 3).

In the further analysis of the data, only the linear part was considered. Therefore, different end-forces were chosen for indenters of different sizes.

B. Setup: Compression From Different Angles

This measurement setup evaluated the angular dependency of the output of the sensing bulb when pressure was applied at different angles. This experiment was done on a single sensing bulb. Furthermore, four different mounting plates were 3D-printed, and then used to press against a tilted sensing

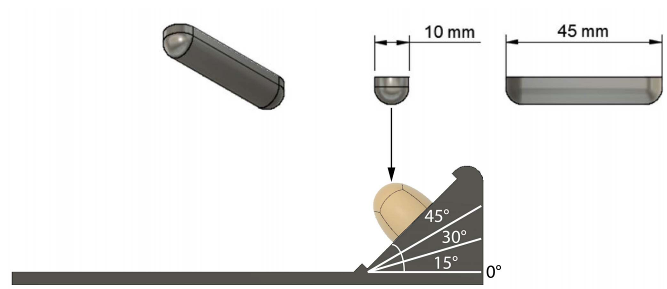

Fig. 4. Cylindrical indenter used in compression from different angles: The rounded bottom of the indenter was applied to the sensing bulb. Top image, on the left, seen from the bottom; In the middle, seen from the short-side; On the right, seen from the long-side. Bottom image shows the attachment used to position the sensing bulb in four different tilted positions: 0, 15, 30, and 45°.

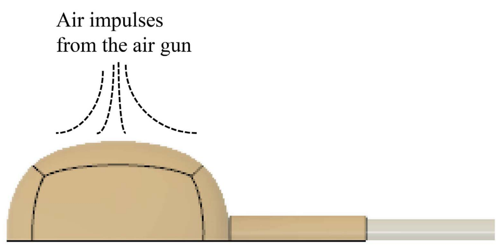

Fig. 5. A soft air gun was used to apply six successive air pulses to a single sensing bulb.

bulb at different angles (0°, 15°, 30°, and 45°). The indenters were pressed against the sensing bulb until the force gauge reached 100 N, however, at larger angles the pressure saturated before 100 N. Therefore, with the largest angle the loading stopped when the force gauge reached 60 N. A cylindrical indenter was 3D-printed to avoid uneven deformation of the sensor at the edges of the indenter (Fig. 4). The cylindrical indenter had a diameter of 10 mm and a potential contact area of 706 mm. The cylindrical indenter was aligned to the sensing bulb. Between the measurements with different pressure angles, the sensor position was adjusted to remain centered (although tilted) with respect to the centerline of the mounting indenter.

C. Setup: Impulse Response

The evaluation of the cutoff frequency was done by looking at the impulse response. A soft air gun was used to apply impulses to a single sensing bulbs (Fig. 5). The soft air gun used a 12 g CO_2 capsule. When the trigger was pulled, a pulse of air hit the sensing bulb. After each firing the CO_2 decreased and thereby, the amplitude of the impulses decreased. The soft air gun was mounted just above the sensing bulb, and six impulses were generated manually by pulling the trigger. A Fast Fourier transform was applied to the extracted sensor responses sampled at 100 kHz.

D. Setup: Glove With Sensing Bulbs

To assess the behavior of the sensing bulb in a realistic scenario, a measurement was conducted on five sensing bulbs that had been integrated into the glove as fingertips. The myo-electric prosthetic, with the sensing glove, was mounted and

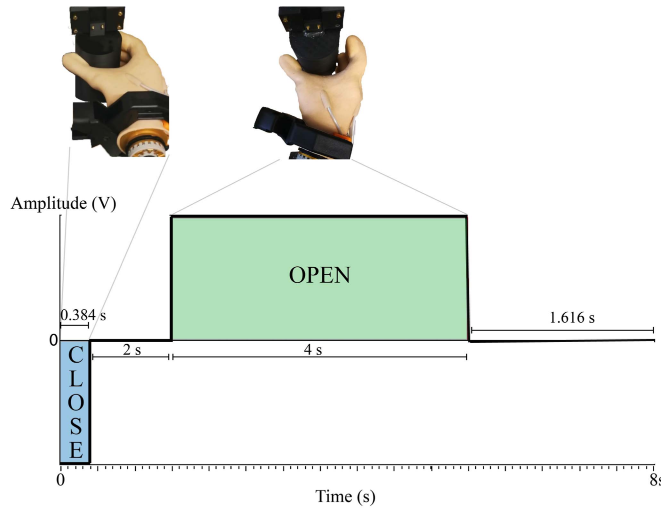

Fig. 6. (a) Control signal for the prosthetic hand. Amplitude: 0.5 (150 mm/s); Closing duration: 384 ms; Opening duration: 4 s; Time frame: 8 s. Duration between closing and opening was 2 s. Top figures illustrate the sensing bulb hand during the the closing and opening state.

remained in a fixed position during all the measurements. The MyoHand VariPlus speed was controlled by modulating a square wave signal in LabVIEW. The modulated signal was set by the amplitude and duty cycle, which defined the closing and opening speed of the hand, respectively, and how much the hand was opened or closed. The amplitude was set to 0.5, which is 50% of the hand's speed (300 mm/s), in other words, during the experiment the hand was moving at the speed of 150 mm/s. The sensor output was measured, and collected in LabVIEW, for all five sensing bulbs while the hand was repeatedly grasping and releasing an attached 3D printed cylindrical object (ϕ 60 mm) resembling the shape of a 0.5 L bottle. To investigate the sensor response repeatability, the hand grasped the object 100 times. The grasp was maintained for 2 s before release. The closing duration lasted for 384 ms and the opening duration lasted for 4 s to ensure that the hand opened completely. The hand remained open the last 1.616 s before repeating the grasping sequence. The change from close to open state happened within a time frame of 8 s. The experimental setup can be seen in Fig. 6.

IV. Results

A. Compression With Indenters of Different Sizes: Hysteresis

Driving the stepper motor until the force gauge reached 100 N showed a monotonically increasing non-linear curve with the indenter no. (i)-(iii). This naturally indicates that the indenter compressed the sensing bulb against the rigid bottom of the sensor earlier with smaller indenters, than with the larger indenters. As can be seen in Fig. 7, the curve becomes saturated with the smaller indenters (no. (i)-(iii)) because the indenter reached the sensor base at around 20, 30, and 60 N respectively, while for the larger indenters the curves are linear until 100 N is reached. Looking at the three biggest indenters with linear curves (indenter no. (iv)-(vi)), the maximum hysteresis occurs at 60.2±4.59 N (marked as the gray area in Fig. 7) with an error of 2.39±0.17% at the full-scale range

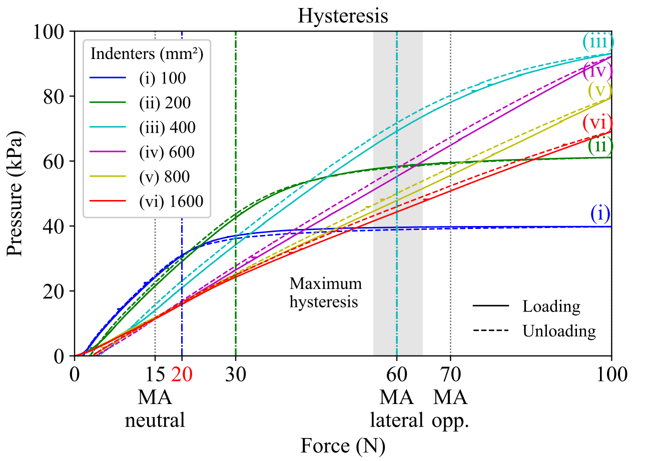

Fig. 7. Hysteresis of the sensing bulb: The solid lines represent pressure added (loading) to the sensing bulb and the dashed lines represent when pressure relieves (unloading). The vertical lines represent where the linearity occurs for each indenter and also the MyoHand VariPlus Speed, which has proportional gripping force of 0-100 N [39]. The force marked in red, indicates the maximum detected force required in the prosthetic hands. The gray area shows the occurrence of the maximum hysteresis for the (iv)-(vi).

of the measured force (0-100 N). The error is less than the maximum error ($< 5\%$) suggested for sensors used in an artificial hand [38]. The reason for the hysteresis might be induced due to the characteristics of the silicone glove, such as the visco-elasticity of the silicone. Moreover, the figure shows that the sensor output is higher during unloading than during loading. This indicates that the sensing bulb was more compressed during loading. The maximal pneumatic pressure difference between loading and unloading was 3.69 Pa. The figure also compares the pressure applied on the sensing bulb with the maximum gripping force generated by the Otto Bock prosthetic hand, MyoHand VariPlus Speed [39].

B. Compression With Indenters of Different Sizes: Linear Regression

The response of the pneumatic pressure, when the sensing bulb was compressed by indenters of six different sizes, is shown in Fig. 8. It can be noted that the sensory output was approximately linear when forces below 20, 30, 60 and 100 N were applied with indenters (i), (ii), (iii) and (iv, v, vi), respectively. pressure was applied on the sensing bulb. For the three smallest indenters, (i)-(iii), the sensing bulb output was approximately linear until the force gauge reached 20, 30, and 60 N, respectively, when applying pressure on the sensing bulb Fig. 8a. For the remaining three indenters, (iv)-(vi), the sensor readings were approximately linear within the full range (0-100 N) (Fig. 8b). However, with a non-linearity for the biggest indenter (vi). Using linear regression, the coefficient of determination, R^2, was 0.9957, 0.9971, 0.9969, 0.9995, 0.9996, and 0.9960 with respect to increasing size of the indenters. The root-mean-square-error (RMSE) was 0.6025, 0.6806, 1.1101, 0.5762, 0.4592, and 1.1794 kPa. This implies that over 99% of the change in pneumatic pressure, for all the indenters, was related to the pressure applied. While $< 1\%$ depended on other variables, such as variable mechanical

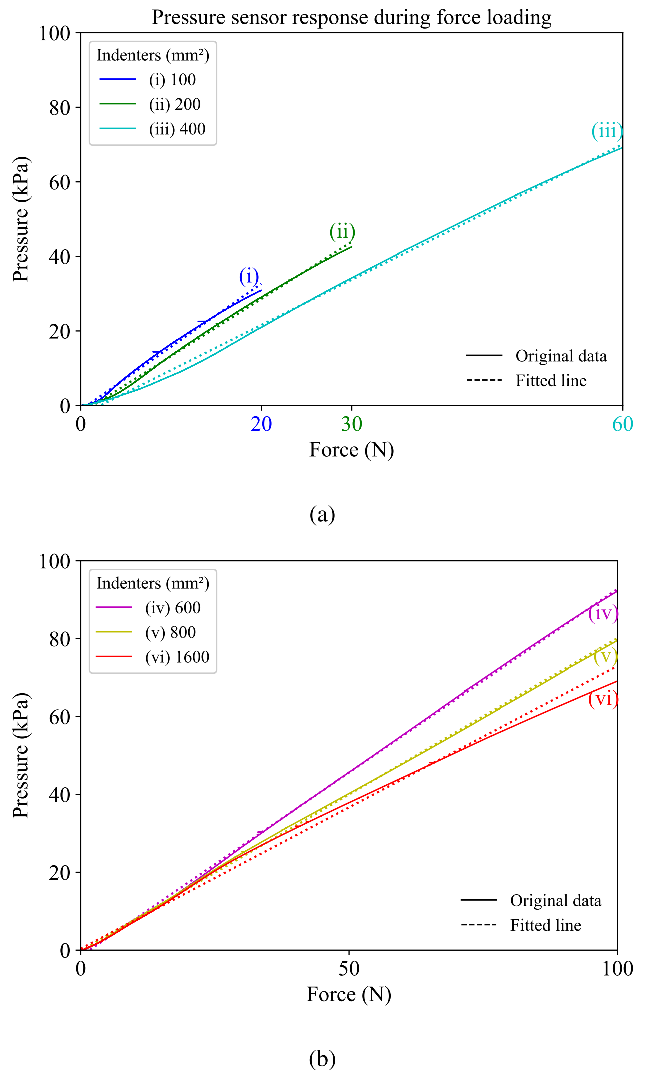

Fig. 8. Force-pressure curve: Pneumatic pressure measured when the sensing bulb is compressed by indenters of different sizes: relatively small (a) indenter no. (i)-(iii) or larger (b) indenter no. (iv)-(vi). The curves show linearity at different load forces, where the end-force for each indenter was chosen as the point where the indenters reach the sensing bulb base. The dotted line shows the fitted line obtained from a linear regression.

properties of the sensing bulb because of its soft nature. It can be speculated that if the wall of the sensing bulb would have been thicker, the sensing bulb could have managed greater force. However, this would occur at the expense of the minimum force sensible. The deviation could also depend on the pressure sensor utilized: MPXV5100G [37], which had an accuracy of $\pm 2.5\%$ V_{FSS}. It can also be concluded that the calculated slope constants decrease when the sensor is deflated, when applying pressure with larger indenters.

C. Compression From Different Angles

As shown in Fig. 9, the characteristics of the sensing bulb remains the same when the pressure is applied at different angles. There is a difference in the pressure level depending on when indenter reaches the sensor base, resulting in pressure saturation for the sensing bulb. The reason for this is that, when the angle is larger, there is less area to counteract the pressure

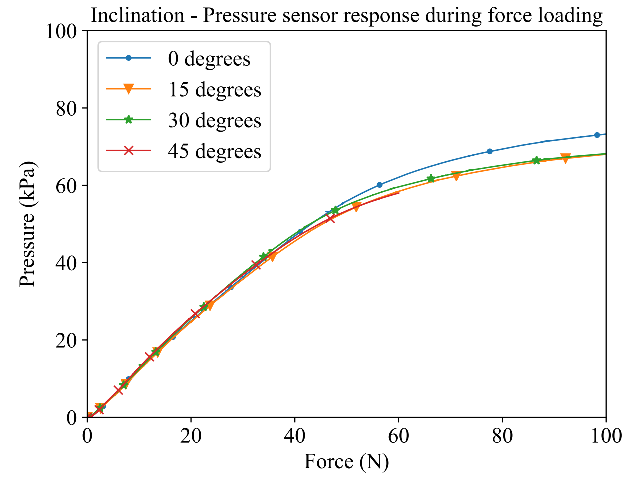

Fig. 9. Loading and unloading on the sensing bulb at different angles.

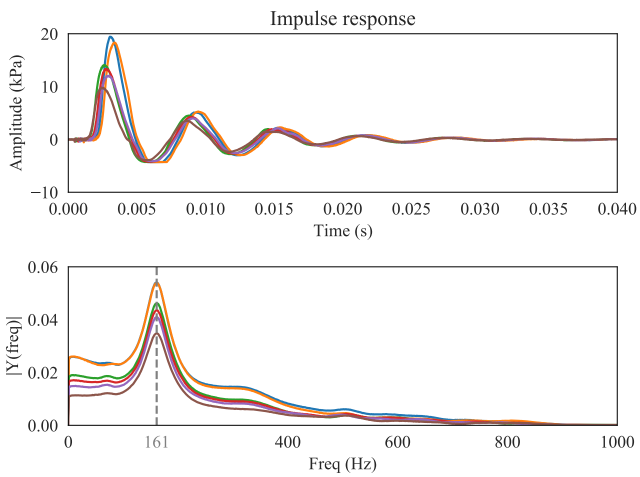

Fig. 10. Six manually generated impulse responses: top image, Response from the gun shot, bottom image. FFT analysis showing there is a resonant frequency at 161 Hz.

from the indenter. The consequence is that the indenter is not able to press with a great force. The contact behavior between the cylindrical indenter and the sensing bulb is different depending on the angles and during the loading on the sensing bulb. The contact area increases during loading. The two materials, silicone (sensing bulb) and plastic (indenter) create friction that stretches the sensing bulb and initiates greater tension on one side, whereas on the other side the mass gathers under the indenter when the angle increases. With no angle on the sensing bulb the mass encloses the cylindrical indenter.

D. Impulse Response

The response of the sensing bulb when generating an air impulse response is shown in Fig. 10. The amplitude of the impulse amplitude was between 10-20 kPa. The sensing bulb, with its resonant frequency at 161 Hz, covers the detectable vibrotactile frequency range of a Meissner corpuscle (3-40 Hz), a Merkel disk (0.4-3 Hz), and part of a Pacinian corpuscle (40-500 Hz) and Ruffini endings (100-500 Hz) [3], [19]. The sensing bulb had a rise time of 0.57 ms, which is within the range specified in the guidelines for a tactile sensing system in which the response time was limited to 1 ms [21].

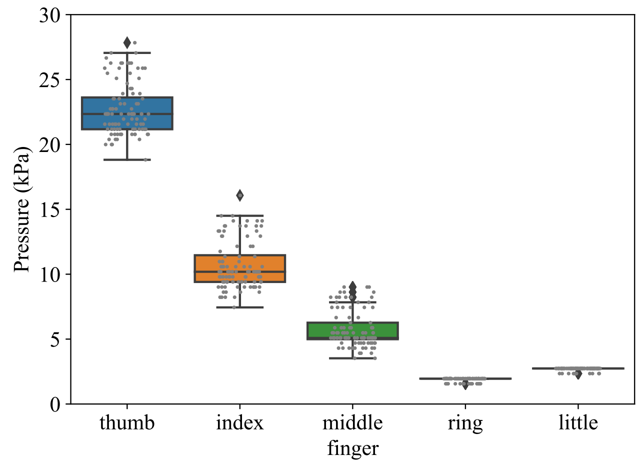

Fig. 11. Repeated measurement when the sensing bulb grasps an object 100 times, looking at the maximum pressure in each grasp.

TABLE III
PARAMETERS FOR THE SENSING BULB

Parameter	Value
Main advantage	Indifferent to angles
Hysteresis	2.95±0.40%
Coefficient of determination	>0.99
Sensitivity	0.82 kPa/N
Resonant frequency	161 Hz
Rise time	0.57 ms
Range	0-9.020 kPa (for the thumb)

The values in the table are based on the results when using the indenters (iv)-(vi), where the response did not saturate.

E. Glove With Sensing Bulbs

The data from the repeated measurements can be seen in Fig. 11. Because the MyoHand VariPlus Speed is a single-degree-of-freedom tripod prehensor, it forms a palmar grasp with thumb, index, and middle finger. This explains the results for the ring and little finger in the figure, where not enough pressure was put on the sensing bulbs to give a noticeable response. The prosthetic hand was placed to give a maximum response at the thumb and index finger and less at the middle finger. It could be noted that the median force measured by the thumb sensor was higher than on the other sensors. This is expected because the force is distributed among the opposing fingers. The repeatability of the sensing bulb hand shows a maximum discrepancy of 9.020 kPa for the thumb and a minimum discrepancy of 0.392 kPa for the ring and little fingers.

V. DISCUSSION

In this paper we characterized a pneumatic sensing bulb designed as an add-on for myoelectric and body powered prosthetic hands. The results can be seen in Table III. This was done by measuring the pneumatic pressure from the sensing bulb when applying pressure with various sizes of indenters and with different incoming angles. Additionally, the sensing bulbs were evaluated in a use-like scenario, while integrated within glove and fitted on a commercial powered prosthetic hand.

A. Comparative Analysis

The evaluation of the sensing bulb not only showed low hysteresis and good repeatability, but also showed stable pneumatic pressure response when pressure was applied at different angles. This is an advantage in prostheses if incoming pressures are not perpendicular to the hinge joint, because parts of the force vectors will be absorbed by the joint. Therefore, when using commercial sensors, such as FSR and capacitive sensors, the response will be inconsistent. However, use of a sensing bulb, which is not dependent on the incoming angle, will give consistent readings. Moreover, to get correct readings from the FSR, the pressure has to be applied evenly over the surface area. During grasping, the objects can have different shapes or cover only some part of the FSR active area. This makes the sensing bulb a better candidate for measuring grasping force. It was mentioned that FSR, instead of being a sensor for absolute force measurements, would be a better fit for detecting motion and position [23]. Furthermore, during manipulation of objects, the maximum contact force is encountered at the distal phalanx [40]. This requires the sensors to be flexible so that they can be placed on the fingertips of the prosthetic hands and still provide stable gripping of objects. Applying FSR on prosthetic fingers, which are soft to resemble the human body, gives an inaccurate sensor response. As for the sensing bulb, it already has the shape of a fingertip and it gives stable values wherever the pressure is applied. Despite the well-known characteristics of FSRs (simple and cheap), this kind of sensing technology is rarely used in commercial devices because of its poor performance [41]. A more suitable option is the capacitive force sensor, SingleTact, which has better accuracy than do other thin-film sensors [42]. Comparing a load cell and FSR, a calibrated SingleTact was recommended for use in prosthetic hands because of its adaptability to the shape of an artificial fingertip and its higher accuracy [16]. On the contrary, it has been shown that the error increases with indenter curvature [16]. The sensing bulb, itself, is a sensor with the shape of a fingertip which eliminates error caused by integrating a sensor on a curved surface.

B. Hysteresis

The sensing bulb showed only small hysteresis during the experiment. Because the pressure was applied slowly, see section A, small hysteresis was measured (Fig. 7). In this study, the hysteresis may be explained by deformation of the silicone or by a very small air leakage, because the sensing bulb did not regain its shape during unloading compared to the corresponding position during loading. Another explanation for the hysteresis could be that the viscoelastic HTV silicone might have induced properties such as creep and stress relaxation. It is desirable to make sure that the connection between the sensing bulb and the plastic tube is sealed properly to make it airtight. Even though it is often required that the tactile sensors, used in prosthesis, should have low hysteresis [11], human skin itself has high hysteresis [4].

C. Linearity

There is a higher sensitivity when smaller indenters are applied on the sensing bulb compared with larger indenters, which indicates that the sensing bulb hand has greater sensitivity when grabbing smaller objects. This is related to the physical relationship between force, pressure, and contact area for the sensor. However, what is interesting is that the sensor bulb response, stronger for smaller object, correlates to the natural perceived sensation using your hand. A smaller object of equal weight to a bigger will give a stronger perceived sensory [43]. As a consequence, having the sensing bulb fitted on a MyoHand VariPlus Speed, which has a grasp force of 100 N, the sensing bulb will give a non-linear response when grabbing small objects, although, such high force might not be necessary when handling small objects. Furthermore, compared to FSR [44] with its nonlinearity, where it has a higher sensitivity at low forces and wider variations [38], the sensing bulb is an arguably better candidate.

A non-linear characteristic could also be seen when applying an indenter larger than the sensing bulb itself. A possible explanation could be same as for the smaller indenters where pressure saturation is obtained early. The MyoHand Variplus performs a tripod prehension. The grasp will not adapt to the object since it has one degree-of-freedom. For this kind of prosthesis, the average contact force is detected at the thumb and the distal phalanges of the index and middle finger. Compared to human hand and prostheses with adaptive grasp, the contact area is smaller resulting in higher force holding an object. However, considering the force being distributed on a smaller contact area the maximum force was shown to only reach 24.9 N at the fingertips [40]. For the sensing bulb the force prediction discrepancy of nearly 10 N would not be applicable for such low forces as the sensing bulb has a linear characteristic up to 30, 60 and 100 N for the indenters (ii), (iii) and (iv, v, vi), respectively. With an exception of indenter (i) for which the sensor is linear up to $\tilde{2}0$ N which is a little shy of the maximum force of 24.9 N.

The sensing bulb is linear within the range of 0-100 Hz, and has a resonant frequency of 161 Hz. The required frequency response for force and position sensors should be the basis for such requirements (>100 Hz) [38]. The rationale is that it can detect deep pressure and high frequency vibration (e.g., can detect smooth surface objects). Looking at the skin's time response, which is ∼15 ms [45], the sensing bulb is relatively faster at 6 ms.

D. Intensity Resolution

When applying pressure on the sensing bulb, before it became flat, the force gauge showed a minimum value of 20 N for the smallest indenter (i). No more than 20 N is necessary to apply to the sensing bulb because the required sensitivity range of a tactile sensor to mimic a hand is 0.01-10 N [4]. However, doing a characterization in the full range (0-100 N) showed us the limits of the sensing bulb.

E. Spatial Resolution

When making an artificial sensor that aims to mimic a human fingertip, the spatial resolution should be 1-2 mm (>2 mm interpoint distance, from which the pattern discrimination should be more sensitive) [4]. However, because the sensing bulb is one unitary sensor it will respond to wherever the pressure is applied, thus it will not sense the position of the pressure, which provides neither interpoint discrimination nor pattern recognition. Moreover, the sensation will be fed back to the residual limb and compared to the two-point discrimination in the fingertips, which at the level of forearm varies between 30-45 mm [46].

VI. Conclusion

This study evaluates a pneumatic sensing bulb that could be used in body-powered and myoelectric prostheses. We validated that the sensing bulb has an advantage of sensing incoming pressure from different directions better than with current commercial sensors, such as most of the resistive sensors and piezoelectric sensors. The results showed an angle dependency during high forces, however, during use it has been shown that the contact force on the fingertips is maximum 24.9 N [40]. Furthermore, the sensing bulb is stretchable, which is highly desired when designing a sensor glove to be used on prostheses. Due to the material and construction of the sensing bulb, it will not add any significant weight when being implemented in the prosthesis. Furthermore, since the sensing bulb does not contain any electrical elements it is not susceptible to electrical interference while interacting with the surroundings. The sensing bulb has the potential for further development, such as integrating other sensing elements, adapting the glove to be fitted on a prosthesis with a higher degree-of-freedom hand and technological improvements to reduce the hysteresis (air leakage) further and also, to eliminate the non-linearity when interacting with bigger objects. As mentioned, one drawback is that, if the sensing bulb is not sealed properly there could be leakage and if it breaks, the whole glove has to be replaced.

Future work would be to integrate the sensing bulb into an active sensory feedback system with actuators to provide the user with sensory feedback with different modalities. Compared to the previous study [31], where the system was passive, the active system opens up the possibility to adapt the feedback to the user. It would also be possible to filter sensor data and transform the sensor data before feeding back the information to the user, thus avoiding redundant information. This would especially be informative if the number of sensors increases. The sensing bulb provides with solely one type of sensory feedback, pressure. Inasmuch as the hand contains different kinds of tactile sensing modalities, the tactile system should also contain different sensors (hybrid tactile sensors). Regarding the angle dependency effect during large compressions, the effect could be minimized by having the sensor bulb at a higher internal pressure. This would make the sensing bulb deform less. In conclusion, a comparison with different design configuration could be taken into consideration for future development. Moreover, focusing on adding more features to the sensing bulb glove, such as texture sensing. Other avenues that will be pursued is to subdivide the sensor bulb into smaller parts. Potentially, by having a 2x2 sensing bulb in the prosthetic finger new features such as force direction will be extracted.

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