Wearable Sensory Substitution for Proprioception via Deep Pressure

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***Abstract*—We propose a sensory substitution device that communicates one-degree-of-freedom proprioceptive feedback via deep pressure stimulation on the arm. The design is motivated by the need for a feedback modality detectable by individuals with a genetic condition known as PIEZO2 loss of function, which is characterized by absence of both proprioception and sense of light touch. We created a wearable and programmable prototype that applies up to 15 N of deep pressure stimulation to the forearm and includes an embedded force sensor. We conducted a study to evaluate the ability of participants without sensory impairment to control the position of a virtual arm to match a target angle communicated by deep pressure stimulation. A participant-specific calibration resulted in an average minimum detectable force of 0.41 N and maximum comfortable force of 6.42 N. We found that, after training, participants were able to significantly reduce targeting error using the deep pressure haptic feedback compared to without it. Targeting error increased only slightly with force, indicating that this sensory substitution method is a promising approach for individuals with PIEZO2 loss of function and other forms of sensory loss.**

***Index Terms*—sensory substitution, wearable devices, proprioception, haptics**

I. Introduction

Proprioception can be considered our "sixth sense." It provides continuous information about body position and movement vital to motor control and coordination, balance, muscle tone, postural reflexes, and skeletal alignment. Many neuromuscular disorders arise from dysfunction of motor efferents, but a deficiency of afferent proprioceptive sensory input is another, often overlooked, cause of impairment that can severely impact motor function, even when strength is preserved. A large host of diseases result in loss of sensation (known as sensory neuropathies), sometimes affecting proprioception and sometimes other touch sensing modalities including vibration, skin deformation, temperature, and pain sensation.

Proprioception in humans is entirely dependent on the non-redundant mechanosensor PIEZO2. Our long-term goal is to address lack of proprioception in individuals with recessive PIEZO2-loss of function, who show complete congenital absence of proprioception leading to motor and functional impairment [1]. Individuals with PIEZO2-LOF also lack vibratory sense and discriminatory touch perception specifically on glabrous skin, although deep pressure, temperature, and some pain sensation is preserved [1]–[4]. No pharmacologic or assistive technology options currently exist for individuals with PIEZO2-LOF. Our goal is to design and test a wearable haptic device that enables proprioceptive feedback using preserved sensory input modalities and evaluate its efficacy to enable intuitive control of limb movement in individuals with PIEZO2-LOF.

Prior work in sensory substitution for proprioception has focused on conveying the state of a prosthetic hand, arm, or leg to a user on an intact body location for an individual with amputation. For the hand, Cheng et al. conveyed the the configuration of a virtual human hand (representing a prosthetic hand) using vibrotactile feedback on a belt worn around the waist [5]. Wheeler et al. applied rotational skin stretch to the forearm in order to provide proprioceptive feedback from a virtual prosthetic arm controlled with myoelectric sensors [6]. On the lower limb, Welker et al. controlled the position of an ankle-foot prosthesis using the wrist, effectively substituting wrist angle for ankle angle [7]. Skin stretch devices were used by Kayhan et al. [8] and Colella et al. [9] to substitute for proprioception in multiple degrees of freedom and at multiple locations on the body. Sensory neuropathy is another condition that is often a direct complication of another comorbidity, such as stroke and diabetes. Tzorakoleftherakis et al. showed that vibrotactile actuators could also be used to convey movement of the arm for patients with loss of proprioception after stroke [10].

Unfortunately, none of the above feedback modalities are appropriate for proprioceptive feedback for individuals with PIEZO2 loss of function due to lack of cutaneous touch sensation. Because deep pressure sensation is intact, we propose a sensory substitution device that communicates proprioception via deep pressure stimulation. We designed a wearable device to provide this stimulation and performed a study to determine whether participants with intact sensation could use the stimulation as a substitute for the elbow angle of a virtual arm. The research questions our work seeks to address are:

- What range of forces applied to the forearm are noticeable and comfortable for participants?
- Can participants learn to map deep pressure applied to the forearm to the angles of a virtual elbow?
- What is the accuracy with which this can be accomplished by deep pressure stimulation, compared to when vision is used?
- Given that sensitivity to changes in force decreases when force magnitude increases (per Weber's Law), will participants' accuracy change with force / elbow angle?

The answers to these questions will inform future designs of wearable haptic devices for sensory substitution of proprioception, especially for individuals with PIEZO2 loss of function, but also for individuals with neuropathy, stroke, amputation, and other diseases that result in sensory loss.

II. System Design and Experimental Setup

A. Deep Pressure Stimulation Device

A wearable device was designed to communicate a user's elbow angle via deep pressure stimuli applied to the forearm. The device includes (1) a deep pressure stimulator and (2) an embedded electronics system. The deep pressure stimulator utilizes a position-controlled micro linear actuator with position feedback (Actuonix L12-30-50-12-I) and a cylindrical tactor (15 mm diameter). The actuator is housed in a rigid enclosure with a flexible plastic interface for contact with skin. Plastic straps are used to fasten the device in place. As the actuator extends its position, the tactor presses directly on the skin and applies a deep pressure stimulus. A low-profile capacitive force sensor (SingleTact, diameter 15 mm and force range 45 N) was embedded into the tactor to measure the applied pressure in real time. The tactor, enclosure, interface, and straps were 3D printed. The device is shown in Figure 1.

The embedded electronics system consists of a microcontroller (ARM Cortex-M4, Teensy 3.2), safety button, and SD card writer. The microcontroller was programmed in C and implements a finite state machine for the device's two distinct modes: calibration and runtime. During calibration mode, the microcontroller adjusts the device parameters to define the maximum and minimum positions to be used in the study. These corresping to the maximum comfortable force and minimum perceivable force selected by each user. In runtime mode, the microcontroller uses open-loop control to read in elbow angle and move the actuator accordingly. A local PID controller on the actuator's board provides the low-level control. Also, the microcontroller records data on force and actuator position feedback via I2C and analog-to-digital conversion, respectively. The device also has 2-way

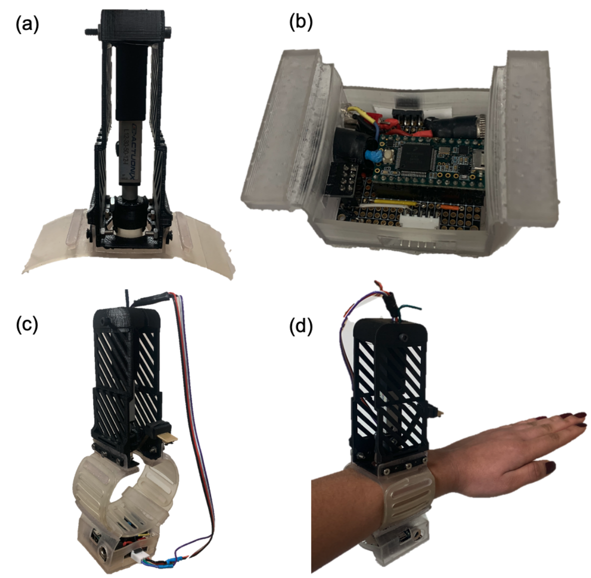

Fig. 1. The wearable device consists of (a) a linear actuator with attached force sensor and (b) embedded electronic system. The parts connect with a (c) flexible 3D-printed arm band and are (d) fastened to the dorsal side of the arm.

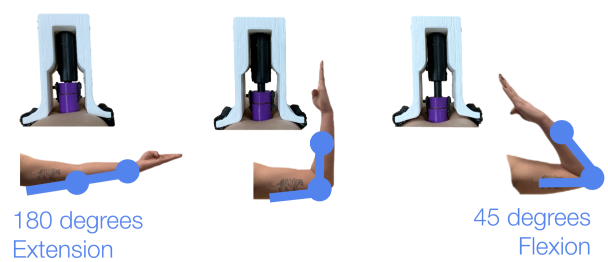

Fig. 2. With the arm fully extended at 180 degrees, the tactor is just in contact with the forearm surface and does not apply significant pressure. As the flexion angle decreases to 45 degrees (i.e., the arm becomes more flexed), the actuator moves the tactor down toward the forearm and increases the amount of deep pressure stimulation.

serial communication to output data and connect with a virtual environment. The electronics are housed in a flexible plastic container that slides onto the plastic straps of the deep pressure stimulator.

The device is fully wearable and programmable, so the deep pressure stimuli can be changed. The default feedback mapping is linear, where the deep pressure stimuli changes with elbow angle (Figure 2).

B. Virtual Environment

To disrupt the intact proprioception of healthy users, we developed a custom software environment for individuals to control a virtual arm. The virtual arm and graphic user interface was created with Python and the canvas drawing library *turtle*. A simple graphic was used to represent the virtual arm and 1-degree-of-freedom elbow movement (Figure II-B). Users utilized the left and right arrows of a keypad to extend and flex their virtual arms from 180 to 45 degrees (as shown in

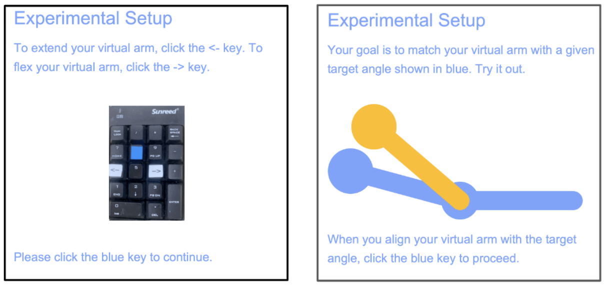

Fig. 3. Instructions shown on-screen for the user are: (Left) Keypad for participants to interact with the graphical user interface to adjust virtual arm with arrows and indicate completion with blue key. (Right) Example of blue target angle and yellow virtual arm graphics displayed in the virtual environment.

Figure 2). For every key press, the virtual arm moved either 1 or 3 degrees; randomness was incorporated in the arm's movement to prevent users from counting key presses and using the keypad as a form of feedback. Despite this, we recognize that users are able to estimate the arm position to some extent using the number of key presses, akin to how people use feedforward, planned motor commands to achieve motion.

The device interfaced with the virtual environment via 2-way serial communication. As users moved their virtual arm, the virtual arm's current position was passed to the haptic device and the device applied the associated deep pressure stimuli. Additionally, the device outputted real-time force and actuator position data to the software environment so the experimenter could view the force and actuator position in real time.

III. Experimental Methods

A. Participants

The criteria for participant recruitment were individuals aged 18 years or older and having no known neurological disorders. Fourteen participants were recruited through community and institution emails (ages 22-60, 7 men, 6 women, 1 declined to share gender, 12 right-handed, 2 left-handed). Individuals with diagnosed neurological or cognitive conditions were excluded from this study. The University Institutional Review Board approved the experimental protocol and all participants gave informed consent.

B. Procedures

To evaluate the efficacy of deep pressure stimuli as proprioceptive feedback, we conducted a user study where we tasked participants with controlling and matching a virtual arm's angle to a given target angle. Target angles ranged from 180 to 45 degrees in increments of 15, resulting in 10 discrete target angles. All participants experienced conditions with and without haptic feedback to complete the task. The order in which they received the conditions was randomized. Participants were randomly assigned to either test conditions with real-time haptic feedback (H) first or no haptic feedback (nH) first. To establish a baseline of optimal proprioceptive performance, participants were evaluated when they experienced visual feedback, where their virtual arm was displayed on the screen. Within these H and nH test conditions, the participants would go through one test condition with no visual feedback (nV) first and then a second test condition with real-time visual feedback (V), in that order.

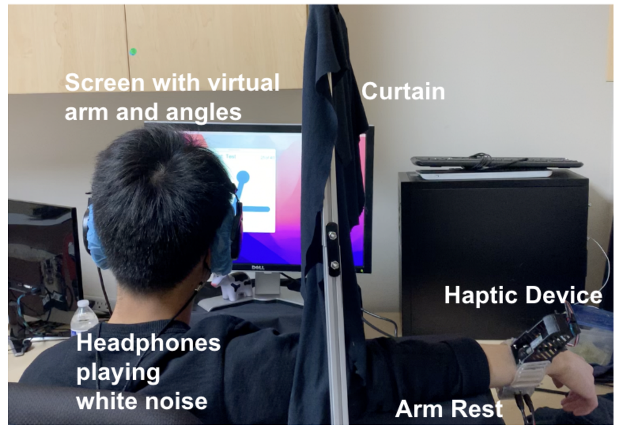

Fig. 4. Participant is situated in front of a display of the virtual environment, which they interact with via a keypad, while the haptic device is worn on the dominant forearm and obscured from the participant's view by a curtain.

Participants visited the lab for an approximately 1-hour long experiment session that consisted of learning and testing tasks. Prior to starting the experiment, all participants completed a pre-experiment survey to collect information on prior experience with human-machine interactive devices and confirm adherence to inclusion criteria.

During the experiment, participants sat in front of a computer screen displaying the virtual environment. The wearable device was fastened to the dorsal side of their forearm to prevent the deep pressure from interfering with circulation. The arm with the device rested in an extended position on an arm rest with a curtain to obscure their view of the device. The participants used their non-dominant hand to control the virtual arm with a keypad.

Participants wore headphones playing white noise throughout the experiment to avoid bias from auditory cues produced by actuator motor sounds. The device was strapped to the participant's forearm with the actuator and enclosure on the dorsal side of the forearm.

1) Calibration: Once the device had been securely fastened to the participant's arm, a calibration sequence was performed to determine a user's minimum detection pressure and maximum comfortable deep pressure. For minimum pressure, the actuator started from a position without contacting the participants' forearm surface and gradually moved toward the forearm surface in pre-defined position increments. The participant verbally notified the experimenter when they first felt pressure, and the corresponding actuator position and force measured were recorded by the software system. For maximum pressure, the actuator also moved in incremental steps while applying pressure on the forearm and the participants verbally noted when they would start feeling uncomfortable. This calibration process was a dialogue between participant and experimenter and was repeated at least three times per participant. The calibrated actuator positions for maximum and minimum pressures were stored and mapped to the minimum and maximum arm flexion angles on the haptic device.

2) Learning: The learning phase was a robust sequence of tasks to ensure the participant had adequate practice with deep pressure feedback. The learning phase of the experiment consisted of four parts:

- *Explore:* Participants moved their virtual arm freely, observed the associated haptic feedback in real time, and learn to associate deep pressure haptic feedback with the virtual arm's angle. This phase lasted for 1 minute.
- *Target:* Participants moved their virtual arm, which was displayed in real time on the screen, to target angles between 45 and 180 degrees and were instructed to pay attention to the haptic feedback once they reached these target angles. Participants had unlimited time for each target angle and were provided a total of 20 target angles in descending and pseudorandom orders.
- *Haptic Feedback:* Participants passively experienced the haptic feedback associated with the different target angles in descending order, for a minimum duration of 10 seconds and for as long as the participant needed. Participants did not have a virtual arm to move in this phase.
- *Practice:* Participants were instructed to match their virtual arm, the position of which was not shown in real time, with the displayed target angle between 45 and 180 degrees by using real-time haptic feedback. For each target angle, once the participant indicated the completion of their angle-matching attempt with a key press, their virtual arm's current position would appear with the target angle displayed for 10 seconds to provide corrective feeback so participants could learn. Practice consisted of 4 batches of 10 angles. The four batches were in descending, pseudorandom, and random order.

3) Testing: Upon completion of the learning phase and an optional break, participants then proceeded to the testing phase.For each of the resulting four test conditions (H nV; H V; nH nV; and nH V), participants were tasked with matching the virtual arm position to the displayed target angle for 10 unique angles in each condition. The 10 unique target angles were spaced 15 degrees apart in the range of 45 to 180 degrees. The order of target angles was randomized for each test condition.

Upon conclusion of the learning and test experiments, participants completed a post-experiment survey that asked participants to rate the level of difficulty in completing the angle-matching tasks for each of the four test cases and describe some of their strategies.

4) Metric: We quantified proprioceptive performance as the absolute value of difference between the target angle and subject's virtual arm angle, called angle error henceforth. Utilizing the absolute value ensures that a participant overshooting and undershooting their virtual arm's position is not neglected in final metric.

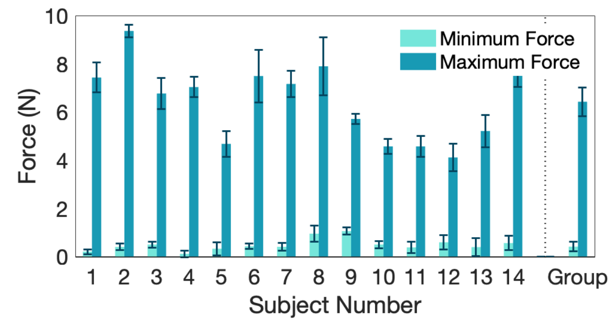

Fig. 5. Average and standard error of forces measured during calibration of minimum and maximum actuator positions for each subject. Across all fourteen subjects, the average minimum detection force was 0.41 N and maximum comfortable force was 6.42 N.

C. Statistical Analyses

All statistical tests were performed with R statistical software in RStudio IDE, and figures were generated with MATLAB (The MathWorks, Inc. Natick, MA, USA).

IV. Results and Discussion

A. Calibration

Figure 5 shows the minimum and maximum calibration forces selected by each participant, as well as the group mean. Across all fourteen subjects, the calibration minimum force mean value was 0.41 N and maximum comfortable force of 6.42 N.

B. Angle Error

The key takeaway of this study is that participants could use deep pressure stimuli to improve their proprioception. Figure IV-B shows that when there is no visual display of the current position of the virtual arm, such that participant relied on the haptic feedback, the haptic feedback condition (H nV) had significantly less error than the no haptic feedback condition (nH nV). The error of the case with no haptic and no visual feedback (nH nV) is not overly large because participants could, to some extent, recall the number of key presses required to move the virtual arm to a target. This is akin to using feedforward (as opposed to feedback) control.

The errors between the user-selected angle and target angle without visual feedback (nV) are much larger than with visual feedback present (V). The errors in the visual feedback conditions are non-zero because of the randomness we injected into the key press control of virtual arm position to avoid over-reliance on counting the number of key presses to achieve a target angle. The visual feedback conditions effectively represent the best possible performance given the experimental setup.

A two-way ANOVA was performed to evaluate the effects of haptic feedback and visual feedback on angle error across all test conditions for all subjects. From the results of this test,

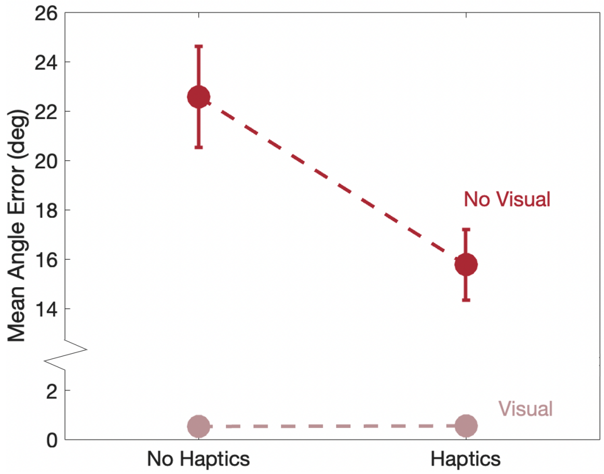

Fig. 6. The mean and standard deviation of angle error are compared for the four conditions of the study. When visual feedback is present, the error is minimal. Without visual feedback, haptic feedback results in significantly less error than without haptic feedback.

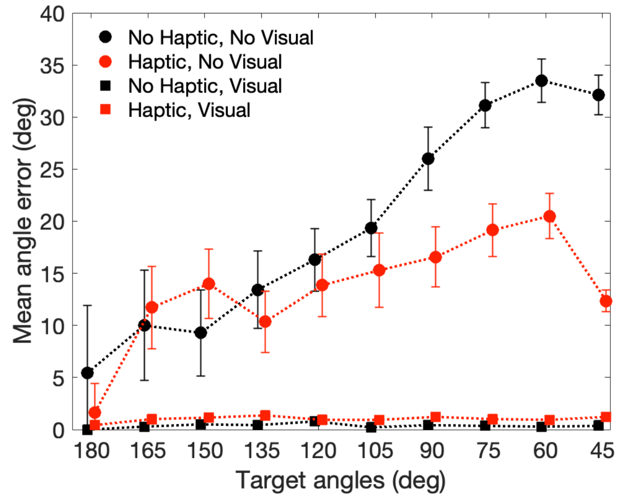

Fig. 7. Angle error versus target angle. For the cases with visual feedback (square data points), the error is slightly above zero due to the randomness we injected into the angle increment with each key press. For the case without vision and without haptic feedback (black circular data points), the error visibly increases with target angle as the arm becomes more flexed. For the case without vision and with haptic feedback (red circular data points), the error does not consistently increase with target angle as the arm becomes more flexed. The 180 degree target is the virtual arm fully extended, and the 45 degree target represents the maximum flex of the virtual arm in the study.

we observed that haptic feedback ($p < 0.05$), visual feedback ($p < 0.05$), and the interaction between haptic and visual feedback ($p < 0.05$) all had a statistically significant effect on absolute error in angle. Mean angle error was lower in the testing conditions with visual feedback (V), for both with haptic (H V) and without haptic feedback (nH V).

We also performed a within-subjects analysis. Grouping the test data by whether visual feedback was provided, we performed a one-way ANOVA with haptic feedback as the within-subjects factor variable and angle error as the dependent variable. Within the no visual feedback (nV) test conditions, we observed a statistically significant effect of haptic feedback on angle error ($p = 0.014$) but no such effect in the visual feedback (V) test conditions. To confirm the one-way ANOVA results, we conducted a pairwise comparison between haptic (H) and no haptic (nH) feedback test conditions for both visual (V) and no visual (nV) feedback groups. Similarly, this analysis showed that haptic feedback had a significant effect (adjusted $p = 0.014$) on absolute error in the no visual feedback (nV) group but not in the visual feedback group (V).

Figure IV-B shows the angle error as a function of target angle. In the testing conditions with no visual feedback (nH nV and H nV), we observed an overall decrease in mean angle error as the target angle increased. In the 45 degree to 135 degree target angle range, mean angle error was consistently higher in the nH V condition than in the H V condition. The relative flattening of the errors in the H nV condition indicates that participants' ability to achieve a desired target angle did not change substantially with target angle. This was somewhat surprising, given that sensitivity to changes in force decreases when force magnitude increases (per Weber's Law). We believe that the extensive training provided and the nature of the task (to achieve a target rather than to perform a two-alternative forced-choice task) enabled this result.

C. Actuator Force

In the testing conditions with haptic feedback (H nV and H V), a roughly linear relationship between target angle and measured actuator force was observed Figure 8. In the testing conditions without haptic feedback (nH nV and nH V), measured actuator force consistently stayed below 1N as expected. This confirmed that our mapping from virtual arm angle to actuator position was functioning as intended.

V. Conclusions and Future Work

In this study, we found that participants were able to learn a mapping between haptic feedback through deep pressure stimulation and target angles of a virtual arm. The use of a virtual arm was key in this study because our participants had intact proprioceptive sensation, and we seek to understand whether this form of haptic feedback will be effective in individuals with PIEZO2 loss of function, who do not have proprioception.

We now review our original research questions. First, we asked what range of forces applied to the forearm are noticeable and comfortable for participants. As shown in Figure 5, the group mean was 0.41 N for the minimum noticeable force and 6.42 N for the maximum comfortable force. Second, we asked whether participants were able to learn to map deep pressure applied to the forearm to the angles of a virtual elbow. This primary result, shown in Figure IV-B, was that after an extensive training session, participants were able to learn a mapping such that their performance with haptic feedback was statistically significantly better than without haptic feedback. The mean error over all participants and all target angles with haptic but not visual feedback was approximately 16 degrees,

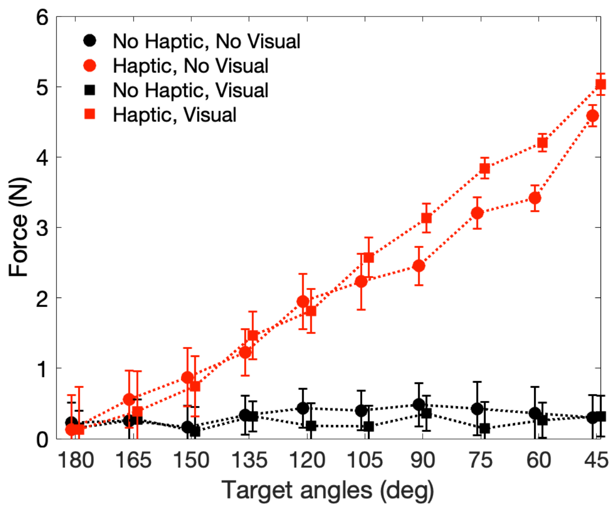

Fig. 8. Applied force versus target angle for the four conditions, averaged over all trials for all users. For the cases with no haptics (black data), the force is slightly above zero due to sensor noise and potentially light contact between the tactor and the skin. For the cases with haptic feedback (red data), the forces increased approximately linearly with angle as the arm becomes more flexed. The 180 degree target is the virtual arm fully extended, and the 45 degree target represents the maximum flex of the virtual arm in the study.

significantly less than the gold standard of near zero degrees when vision is available. Finally, we ask whether participants' accuracy would change with force (equivalent to virtual target elbow angle). With haptic feedback we did find some increase in error with flexion, but not to the extent that might be predicted if Weber's law (linearly decreasing sensitivity with stimulus intensity).

Our results are promising for deep pressure stimulation to be used a method for sensory substitution of proprioception for individuals with PIEZO2 loss of function, and possibly other scenarios such as amputation and sensory neuropathy. Once concern with the current approach is that the lower range of forces used by the healthy participants in our study will not be perceptible to future participants with PIEZO2 loss of function. This will decrease the range and likely the resolution with which forces can be displayed. Multiple tactors that effectively add to the force range may be required.

Our long-term engineering challenge is to generate an appropriate substituted feedback signal for individuals with PIEZO2 loss of function that is effective (i.e., replaces the missing proprioceptive sensation with an intact sensation in a manner that is useful for activities of daily living) and convenient (e.g. low-cost and wearable). Thus, our next step in the design of the device is to develop a soft pneumatic wearable haptic device that offers a smaller size and lower weight of the worn device. Eventually, we aim to develop multi-degree-of-freedom sensory prostheses to enable improvement in the performance of functional tasks for individuals with sensory loss.

Our long-term clinical challenge is to understand intact sensory abilities in individuals and populations with proprioceptive loss, as well as their ability to learn to use a sensory prosthesis. Our next step is to test in individuals with PIEZO2-LOF, and, if a one-degree-of-freedom elbow device is effective, we will proceed to encode multiple degrees of freedom of joint movement for both the upper and lower limbs. In addition to addressing a widespread medical problem, our work aims to provide neuroscientific insight into the role of proprioception in human motor control and how humans adapt to new sensorimotor scenarios.

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