Evoked Tactile Feedback and Control Scheme on Functional Utility of Prosthetic Hand

Luis Vargas, He Huang, Yong Zhu, and Xiaogang Hu

***Abstract*—Fine manual control relies on intricate action-perception coupling to effectively interact with objects. Here, we evaluated how electrically evoked artificial tactile sensation can be integrated into the functional utility of a prosthetic hand. Using different myoelectric-control strategies, participants performed a modified box-and-block task using a prosthetic hand. Transcutaneous nerve stimulation was employed to elicit somatotopic fingertip tactile feedback reflecting prosthetic fingertip forces. This feedback was evoked using an electrode grid placed along the participants' upper arm targeting the median and ulnar nerve bundles. Myoelectric signals from the finger flexor and extensor controlled the prosthetic joint velocity or position. Participants lifted, held, and transported cubes of varying weights using their minimum grip forces. The results showed that participants exerted lower forces and presented lower number of failed trials (prematurely dropped objects) when feedback was provided with respect to without feedback. We also found that position control required more flexor muscle activation compared with velocity control when tactile feedback was provided. Our findings reveal that non-invasively evoked tactile feedback could be used to effectively enable human-in-the-loop control of a prosthetic hand. The outcomes can provide a platform to characterize the action-perception couplings during prosthetic control, in order to improve user experience and system functionality.**

***Index Terms*—Tactile sensation, transcutaneous nerve stimulation, prosthetic hand, functional task, haptic feedback.**

I. Introduction

CURRENT developments in robotic prostheses have led to dexterous robotic limbs that can mimic essential traits of human motions [1]. Although many systems can imitate motor functions, such as individuated finger movements, the lack of sensory feedback critically limits their controllability [2]. During daily activities, we rely on sensory feedback to achieve basic tasks; fingertip tactile feedback provides crucial insight into object interactions, allowing for dexterous hand control without the need for auditory or visual cues. Tactile cues are depicted using mechanoreceptors embedded within our skin [3]. Intricate percepts are elicited as our mechanoreceptors identify a stimuli based on its intensity, frequency, and location [4]. Contrarily, arm amputations lead to a loss of both motor and sensory functions. Prosthetic users often rely on visual cues to operate their devices, leading to cumbersome control and low user confidence [2]. Prior work reported that visual feedback alone led to functional deficits during prosthetic use [5]. In addition, the lack of sensory feedback is considered one of the primary causes of prosthesis abandonment [6].

Prior studies have elicited artificial percepts that can resemble somatosensory cues [7]. Non-invasive methods employ mechanical or electrical stimulation on the skin surface. These strategies elicit somatotopic or non-somatotopic sensations that require users to associate elicited percepts to finger force or joint angle information [8]. Non-somatotopic percepts are categorized as sensory substitutional [9] or location-mismatched [10] based on whether the evoked percept resembles the sensation modality and location experienced by the prosthetic hand. Although non-somatotopic approaches can provide users with informative percepts, its locational dissimilarity can often lead to increased cognitive load during stimuli interpretation [11]. In contrast, somatotopic feedback, with matched modality and location of percepts experienced by the prosthetic hand, can lead to greater accuracy [12] when discerning the location or intensity of a sensation. Somatotopic percepts have been elicited via invasive stimulation of the peripheral or central nervous system [13], [14]. Although these approaches have improved prosthetic control, the invasive nature limits wide applications. Somatotopic percepts can also be elicited non-invasively by activating either the phantom finger map via mechanical or electrical stimulation [15], [16] or the sensory axons in the nerve trunk via transcutaneous nerve stimulation [17], [18]. Although object property recognition has been achieved via transcutaneous nerve stimulation [19], [20], the functional utility of this feedback approach during object manipulation has yet to be evaluated.

To enable human-in-the-loop control of a prosthesis, understanding the interaction between sensory feedback and prosthetic control is essential due to the intricate action-perception coupling [21]. Different myoelectric controllers are available for continuous prosthesis control [22]. Common controllers map the level of muscle activation to either the position or velocity of the prosthetic joint. During velocity control, users often experience greater movement stability as the muscles are relaxed to maintain a given force level or joint angle. Alternatively, position controllers may be more intuitive due to the similarity

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to the control of biological hand; however, this control strategy requires a sustained activation level to uphold the prosthetic state. Using invasive tactile feedback, an earlier study [23] found that position control resulted in improved object size recognition due to the intrinsic proprioceptive feedback from muscle activation. Despite these research efforts, few studies [24], [25] have evaluated how non-invasive feedback and different types of controllers affect the functional utility of prosthetic hands.

Accordingly, the purpose of the current study was to assess the functional performance of a modified box-and-block task during human-in-the-loop control of a prosthetic hand. Specifically, able-bodied participants controlled a prosthetic hand using one of two myoelectric controllers (position or velocity control), with or without fingertip tactile feedback. Tactile sensation was evoked by stimulating the median and ulnar nerves using an electrode grid placed along the upper arm. As distinct electrode pairs were selected, a selective set of axons can be recruited, which can produce localized sensation on the hand [26], [27]. The current amplitude modulation can produce graded sensations that resemble real-time fingertip forces. A prosthetic hand was myoelectrically controlled using either position or velocity control by mapping the electromyography (EMG) amplitude to either a joint position or joint velocity. This unique myoelectric control and tactile feedback platform allowed us to assess the integrative role of somatotopic tactile sensation during functional prosthetic use when different myoelectric controllers are used. The outcomes provided evidence on the effective utility of tactile feedback for object manipulation requiring fine control of prostheses. The developed stimulation approach has the potential to improve user experience and prosthesis utility. Due to its low risk, the non-invasive nature promotes the application of this system to a greater number of individuals.

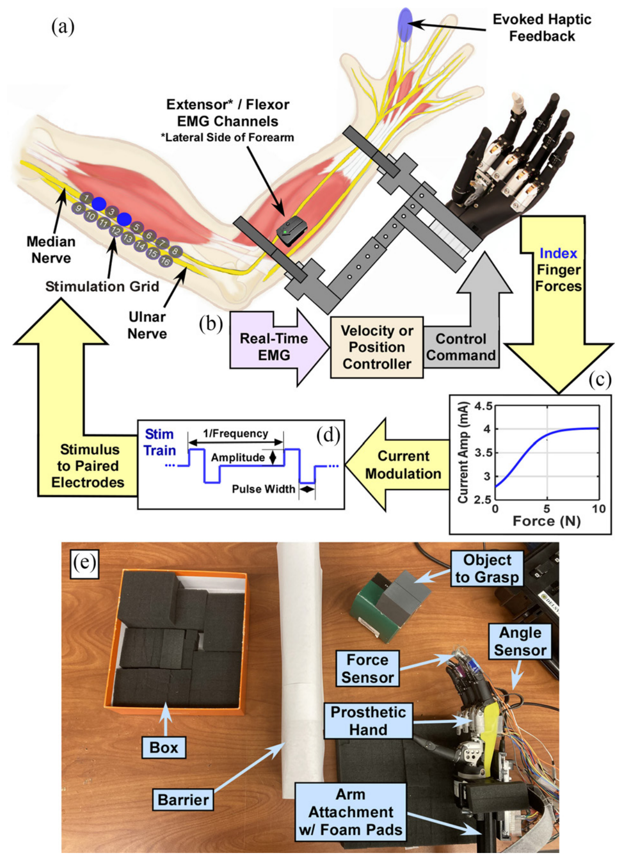


Fig. 1. Diagram illustrating the location of the EMG channels and 2x8 grid along the participant's arm (a). EMG recordings and fingertip forces were used to provide closed-loop control of the prosthesis. EMG recordings (b) were used to control the position or velocity of the prosthetic index finger, while prosthetic finger forces were used to elicit haptic feedback (c, d). The current amplitude for the biphasic stimulation train (d) was determined by transforming the forces using a participant-specific sigmoid function (c). The setup for the modified box and block task (e).

II. Materials and Methods

A. Participants

Eight able-bodied individuals (3 Female, 28.8 ± 4.3 years of age) participated in this study. Each participant provided informed consent via protocols authorized by the Institutional Review Board of the University of North Carolina at Chapel Hill (Approval #: 16-1852). All individuals had no prior experience operating a prosthetic hand nor utilizing the employed sensory feedback approach.

B. Experimental Setup

Participants were seated with their right arm placed atop of a table. The medial and lateral skin surface of their forearm and upper arm were cleaned using alcohol pads. To elicit sensory percepts resembling fingertip forces, transcutaneous nerve stimulation was delivered via a 2x8 electrode grid, composed of 1-cm Ag/AgCl gel-based electrodes placed below the short head of the biceps brachii (Fig. 1). This location was selected because the median and ulnar nerve bundles are close to the skin surface. As a result, electrical stimulation delivered to distinct electrode pairs created unique electric fields that activated selective sets of sensory axons, eliciting distinct sensations on the hand [26]. Stable electrode-skin contact was ensured by supplying mild inward pressure using a plastic vice. A single pair was selected for each participant to elicit modulated fingertip tactile feedback.

Stimulation paradigms and location (electrode pair) were controlled via a custom MATLAB script (v2017b, MathWorks Inc). The script communicated with a switch matrix (Agilent Technologies) to link the cathode and anode of an electrical stimulator (STG4008, Multichannel System) to an electrode pair. The stimulator delivered a biphasic, charge-balanced, square-wave current with a fixed frequency of 150 Hz and a pulse width of 200 µs, based on earlier studies [19], [20], [26].

The stimulation amplitude was modulated based on the force recorded from the index fingertip of the i-limb prosthetic hand (Ossur). Grasp forces were recorded via a force sensitive resistor (FSR) placed at the prosthetic index fingertip. The FSR sensor recorded the normal force during the grasping of the object. The FSR was calibrated using standard weights to account for nonlinearity. Electrical stimulation was controlled in real-time based on the interaction forces between the prosthetic hand and the object (Fig. 1(c) and (d)). Stimulation amplitudes were selected by transforming the recorded force using a participant-specific sigmoid function. The function was constructed using the allowable

TABLE I
ELECTRODE PAIRS AND SENSATIONS ELICITED FOR INDIVIDUAL PARTICIPANTS

Subject #	Evoked Sensation Region (I: Index, M: Middle, R: Ring, T: Thumb)	Electrode Pair (Cathode-Anode)	Sensory Threshold (mA)	Just Below Motor Threshold (mA)
1	M & R	3-5	3.2	4.1
2	I & M	3-5	1.4	2.3
3	I & M	4-6	2.9	4.6
4	I & M	3-7	3.5	4.9
5	I	6-13	1.6	2.1
6	I & R	11-13	2.8	3.6
7	M	3-5	3.4	4.0
8	I, M, & T	5-12	5	6.7

stimulation range, minimum and maximum force, and steepness value [18]. The stimulation range was identified for each participant and was bounded by the Sensory Threshold and Just Below Motor Threshold (Table I). The Sensory Threshold is defined as the stimulation amplitude required for the participant to first perceive sensory information in the palm side of the fingers, while the Motor Threshold is the stimulation amplitude that results in hand movement/twitching. The Sensory and Just Below Motor Thresholds were identified three times, with the average as the lower and upper bound of the sigmoid function, respectively. The flattening of the sigmoid function ensured participant safety and limited the possibility of inducing muscle activation.

The prosthesis was fixed to an arm attachment that was later strapped to each participant (Fig. 1(a)). Foam pads were placed between the forearm and the arm attachment to minimize discomfort. The arm attachment allowed each participant to transport the hand location within the experimental space using their arm, while the prosthetic hand grasping motions and grip forces were myoelectrically-controlled using two electromyography (EMG) electrodes (Delsys Trigno). The two electrodes recorded the activation of the flexor digitorum superficialis and extensor digitorum communis, after skin preparation using alcohol pads. The electrode positions were selected through palpitation of the anterior and posterior forearm muscles, respectively. The EMG signals were sampled at 5000 Hz with a bandpass filter of 20–450 Hz and a gain of 300. The EMG signals were processed to detect the user intent. The EMG signals were first processed by taking the mean absolute values (MAV) of the signal using a 25 sample (5 ms) window with a step size of 1 sample. The MAV values were then smoothed using a 200-ms moving window with a 100-ms overlap to determine the activation level from the two EMG channels. For each participant, the activation levels were normalized using the maximum voluntary contraction (MVC) for each channel. Using these two values, a linear regression function was calculated such that the relative normalized activations levels were proportional to the velocity or position of the prosthetic joint. A maximum joint position and velocity could be achieved by producing 50% MVC of the participants. This upper threshold was selected to minimize potential fatigue. A lower threshold of 2% MVC was used in the controller to initiate the movement of the prosthetic hand. This value was selected during preliminary testing, such that no premature movement of the prosthetic joint was observed and no obvious delay in the initiation of prosthetic movement was noticed by the participants.

Two control schemes, position- and velocity-based control, were employed to evaluate the prosthetic use. Position-based control associated the prosthetic's index finger position to a given level of activation, e.g., higher flexor activation relative to extensor activation results in a greater joint flexion angle. The reference joint angle regression function used in position control was calculated as:

$$\begin{aligned} &Joint\ Angle \\ &= (A_{Max} - A_{Min}) * \left(\frac{EMG_{Flex}}{0.5 * MVC_{Flex}} - \frac{EMG_{Ext}}{0.5 * MVC_{Ext}} \right), \end{aligned}$$

where A_{Min} and A_{Max} are the minimum and maximum joint angles, EMG_{Flex} and EMG_{Ext} are the activation levels for the flexors and extensors, respectively, and *MVC* is the MVC for the two EMG channels. In contrast, velocity-based control mapped the joint speed to the relative activation level of flexors or extensors. The reference joint speed regression function for velocity control was calculated as:

$$Joint\ Speed\ = (V_{Max} - V_{Min})\ * \frac{EMG_{Dir}}{0.5 * MVC_{Dir}},$$

where V_{Min} and V_{Max} are the minimum and maximum joint speeds, and *Dir* is *Flex* or *Ext*, whichever has the greater normalized activation level. The approach ensured a short latency when switching movement directions. The metacarpophalangeal (MCP) joint angle from the prosthetic index finger was recorded in real-time using an external 1-axis flex sensor and was integrated into our custom proportional-derivative (PD) controller to monitor the position or velocity of the finger. Control commands for the prosthetic hand were sent through a MATLAB interface and were updated at 40 Hz. The controller's reference position or velocity was calculated and updated at 10 Hz. The joint angles ranged from 0° to 85°, or from a fully open to fully closed finger position. Note that the participants used electrical stimulation to modulate the grip force, and used visual feedback to modulate the joint angle.

C. Procedures

Once the setup was complete, pre-experimental preparations were performed. First, stimulation was delivered to different electrode pairs to pinpoint a pair that elicited sensation on the participant's index finger. If index finger sensation could not be evoked, a pair inducing middle finger sensation was used. Once a pair was identified, the stimulation range was identified, and the sigmoid function was constructed (Table I). Second, the MVC of the flexor and extensor muscles were recorded using the two EMG electrodes. These participant-specific values were sent to the myoelectric controller. Finally, participants were given 2–4 minutes per control scheme to practice moving the prosthetic hand and flexing and extending its index finger with the arm attachment strapped to their forearm.

The main experiment assessed the participant's ability to complete a modified box-and-block task using the prosthetic hand (Fig. 1(e)). The modified box-and-block task required

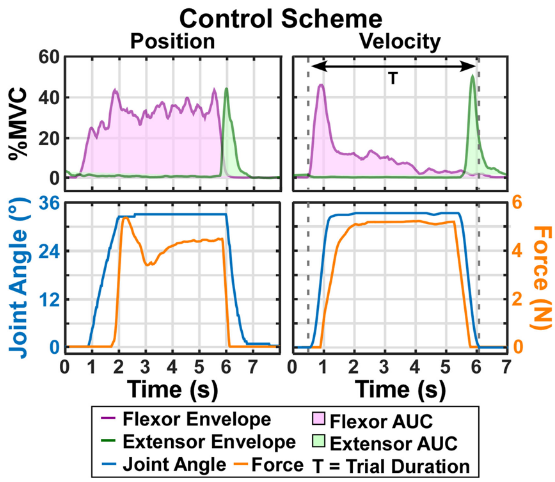


Fig. 2. Exemplar EMG envelopes produced when operating the hand using position and velocity control schemes. The purple and green traces correspond to the flexor and extensor EMG, respectively. The joint angle (blue) and force (orange) recorded from the sensorized prosthesis is displayed for the given trial. The area under the curve (AUC) utilized to quantify the effort is denoted, as well as the parameters utilized to summarize the data.

participants to grasp, lift, hold, and release a 5-cm cube. Three same-sized and visually identical cubes but with varying weight were employed to evaluate if the control of the hand could be regulated based on the object's perceived weight. The three cubes were categorized as light (50 g), medium (100 g), or heavy (150 g) weights. These weights were selected in preliminary testing requiring a range of grip forces. Namely, the low coefficient of friction between the objects and the prosthetic fingertip led to slippage when using heavier weights.

Four experimental conditions, i.e., 2 control schemes $\times$ 2 sensory feedback status (on vs. off), were tested in a pseudo-randomized order across participants. For each participant, blocks utilizing position- or velocity-control schemes were interleaved, meaning that blocks with the same controller were not performed consecutively to minimize the potential for learning effect. Prior to each block, participants were informed of the control scheme and feedback status to be used in the upcoming trials. In addition, participants were given the opportunity to practice controlling the prosthesis by transferring each of the three weights two times each. After 3–5 minutes of training, the cubes were retrieved and hidden out of the participant's line of sight. For each block, the participant was asked to grasp the 3 cubes four times each, resulting in 12 total trials. The cubes were presented to the participants in a random order across blocks and participants. For each trial, the experimenter placed a random cube in front of the participant. Participants were instructed to grasp and lift the cube using the minimum amount of force possible. Once lifted, participants were asked to hold the cube for at least two seconds, timed using a 1 Hz metronome. The participants then had to move the cube over a barrier and release it into a box. The trial was concluded once the prosthetic hand returned to its fully open position. If the cube dropped prior to reaching the box, the trial was identified as a failure and was repeated. The block ended after 12 successful trials (4 trials $\times$ 3 objects) were completed. Fig. 2 depicts exemplar force and joint angle traces and EMG envelopes in a representative position- and velocity-control trial. To reduce the potential for fatigue over time, a 1–2-minute rest between trials and a 5-minute rest between blocks was provided as a precautionary measure. The rest period also allowed time for the experimenter to set up the next trial or block. The supplementary attached video shows the control of the prosthetic hand when performing the modified box-and-block task. The grip forces with and without tactile feedback were compared in each control scheme.

D. Data Processing

First, the number of failed trials (premature drops of the object) was quantified to assess the reliability of each feedback and control scheme condition across participants. The number of failed trials was evaluated per experimental condition and object weight for each participant. Next, the force recordings and EMG envelopes were analyzed to quantify the effectiveness of sensory feedback in each experimental condition. For the force data, the average force level and force variability were quantified during the holding phase of the task. The force variability was calculated as the standard deviation of the forces during the holding period. Only the success trials were used for the force level and force variability calculation. We then determined if different levels of effort were required to complete the tasks. The relative difference in effort was quantified by the Area Under the Curve (AUC) normalized by the MVC for both the flexor and extensor EMG envelopes (Fig. 2) [28]. For each muscle, the resulting AUC values were then normalized by the trial duration to quantify the effort required for each experimental condition [29]. For each participant, the effort was then averaged among trials of the same experimental condition as shown in:

$$\overline{Effort_X} = \frac{1}{n} * \sum_{i=1}^{n} \frac{AUC_{X_i}}{T_i},$$

where X is a position- or velocity-control condition, T is the trial duration (the time between the initiation of the flexor activation, i.e., flexion activation surpassed 2% MVC, and the time when the prosthetic hand returned to its fully open position following object release), i is the trial number, and n is the number of trials in each condition. Ratios of the computed average effort values were calculated to compare the impact of feedback conditions and control scheme conditions separately. These ratios were then transformed using a logarithmic transformation for normal distribution, as shown in:

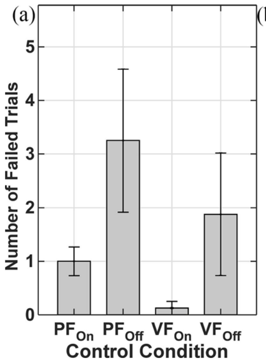
$$\log\left(Ratio\ of\ Effort\right) = \log\frac{\overline{Effort_A}}{\overline{Effort_B}},$$

where A and B correspond to two conditions (feedback status or control scheme). A positive log ratio indicates that $Effort_A$ is larger than $Effort_B$, and vice versa.

E. Statistical Analysis

For each performance metric, the Shapiro-Wilk Test was performed to assess its normality. The evaluations determined that all metrics were normally distributed across participants, except the failure outcomes. For the failure outcomes, the Friedman Test

(a)



(b)

Participant #	PF_{On} L	PF_{On} M	PF_{On} H	PF_{Off} L	PF_{Off} M	PF_{Off} H	VF_{On} L	VF_{On} M	VF_{On} H	VF_{Off} L	VF_{Off} M	VF_{Off} H	Total
1	1		1	2	2	1				4		2	13
2			2		1	4						1	8
3			1	1		1							3
4													0
5				1		2							3
6		1											1
7			1	2	6	3				1	7		20
8			1					1					2
Total	1	1	6	6	9	11		1		5	7	3	

Fig. 3. Diagram illustrating the average number of failed trials (a) and their distribution across participants and object weights (b) for each condition. 'PFOn' signifies position control with feedback, 'PFOff' signifies position control without feedback, 'VFOn' signifies velocity control with feedback, and 'VFOff' signifies velocity control without feedback. 'L', 'M', and 'H' denote the light, medium, and heavy object, respectively.

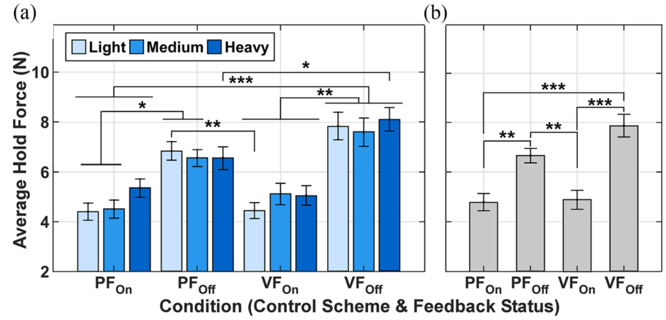


Fig. 4. The average force exerted for each experimental condition across all participants are reported for individual (a) and all (b) weights. The error bars indicate standard error, while 'PFOn' signifies position control with feedback, 'PFOff' signifies position control without feedback, 'VFOn' signifies velocity control with feedback, and 'VFOff' signifies velocity control without feedback. Significance is shown with '$*$' denoting p<0.05, '$**$' denoting p<0.01, and '$***$' denoting p<0.001 when comparing across conditions or object weights.

was performed to discern differences across the four conditions. ANOVA tests were performed to discern differences in other metrics (i.e., average hold force and force variability) across experimental conditions or object weights. When needed, multi-comparison correction was performed using Tukey's Honestly Significant Difference (HSD) test. Lastly, one sample *t*-tests were performed to determine if the logarithmically transformed ratios were significantly greater than or less than 0 (i.e., log (1)), indicating whether the effort required for each experimental condition were significantly different from one another.

III. Results

We first evaluated the functional task performance of the modified box-and-block task. Experimental conditions varied based on the control scheme (position vs. velocity) and status of the feedback (on vs. off). These conditions were labeled with 'PF_{On}' denoting position-control with feedback, 'PF_{Off}' denoting position-control without feedback, 'VF_{On}' denoting velocity-control with feedback, and 'VF_{Off}' denoting velocity-control without feedback. First, we assessed the number of failed trials (premature drops of the object) for each condition. Fig. 3 depicts the average number of failed trials across all participants, as well as the distribution of the failed trials in terms of participant # and object weight. Overall, the results showed that tactile feedback reduced the total number of failed trials. This trend was consistent for both position- and velocity-control. In addition, position-control led to a greater number of errors compared to velocity-control when comparing conditions with the same feedback status (e.g., PF_{Off} vs. VF_{Off}). When assessing the failure distribution across object weights, PF_{On} appeared to have the largest number of failed trials when handling the heavy object. Alternatively, the distribution was more uniform across object weights for PF_{Off} and VF_{Off}. In terms of participant distribution, 7 out of the 8 participants had at least a single failed trial; however, a relatively large variability in the number of failed trials per participant was observed. As a result, no significant difference was found across conditions and across participants.

To quantify the object-prosthetic interaction, the average force level during the object holding phase for each object weight and condition is depicted in Fig. 4(a). The results showed that tactile feedback reduced the average hold force for all weights when using either control scheme. Force reductions ranged from 18.1% to 43.7% when comparing conditions with/without feedback. Specifically, the average hold forces exerted in PF_{On} and VF_{On} conditions were significantly less than those in the VF_{Off} condition (p<0.01, power>0.8) for all individual object weights. Contrarily, although an average force reduction of 26.6% and 28.3% was observed when comparing PF_{Off} to VF_{On} and PF_{On}, respectively, only a subset of the three weights was significantly different (p<0.05, power>0.8). When comparing across control schemes with the same feedback status, no significant difference was observed across VF_{On} and PF_{On} conditions; however, the hold forces for the heavy cube in PF_{Off} were significantly less than those in VF_{Off} (p<0.05, power = 0.75). The force applied when holding distinct object weights was not significantly different during any of the four conditions; however, when comparing the heavy objects to the medium and light objects for PF_{On}, an average force reduction of 16.9% was observed. Similarly, an average force reduction of 12.6% was observed when comparing the heavy and medium objects to the light objects for VF_{On}. Fig. 4(b) depicts the average hold force for each condition when compiling all object weights. The results showed that the average hold force across all weights for PF_{On} and VF_{On} was significantly less than PF_{Off} and VF_{Off} (p<0.01, power>0.8).

We then assessed the force variability for each experimental condition (Fig. 5). Force variability was calculated as the standard deviation of the force during the object holding period. The results showed that less force variability was observed in conditions with tactile feedback. Specifically, a significantly higher force variability was observed when comparing PF_{Off} to PF_{On} (p<0.05, power>0.8) and VF_{On} (p<0.01, power>0.8). When assessing the variability for individual object weights (Fig. 5(a)), we found that the variability in PF_{Off} was significantly larger than those in PF_{On} during heavy object holding (p<0.05, power>0.8) and VF_{On} during heavy or light object

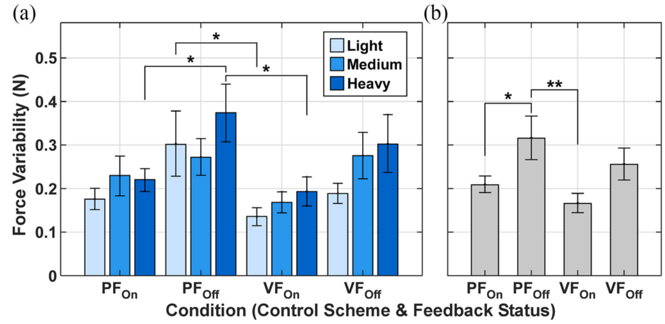


Fig. 5. Force variability at holding phase for individual (a) and all (b) weights for each condition. The error bars depict standard error, while 'PFOn' denotes position control with feedback, 'PFOff' denotes position control without feedback, 'VFOn' denotes velocity control with feedback, and 'VFOff' denotes velocity control without feedback. Significance is shown with '∗' denoting p<0.05, '∗∗' denoting p<0.01, when comparing across conditions.

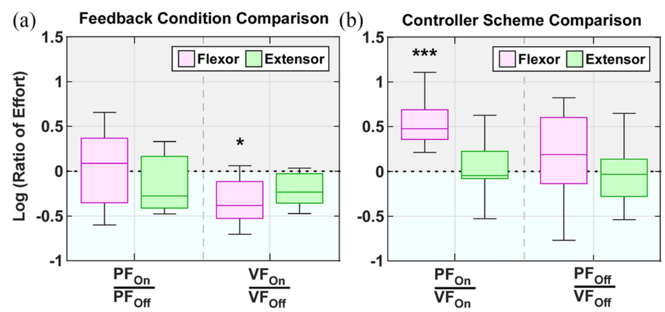


Fig. 6. Box plots showing the relative difference in effort, across feedback status (a) and across controller schemes (b). 'PFOn' signifies position control with feedback, 'PFOff' signifies position control without feedback, 'VFOn' signifies velocity control with feedback, and 'VFOff' signifies velocity control without feedback. Significance is shown with '∗' denoting p<0.05, and '∗∗∗' denoting p<0.001 when comparing to zero, i.e log (1).

holding ($p<0.05$, power>0.8); however, all other comparisons were not significant.

Lastly, Fig. 6 shows the relative difference in effort (muscle activation) required to complete the modified box-and-block task when comparing across conditions. In Fig. 6(a) it can be observed that no significant difference in flexor effort is present between PF_{Off} and PF_{On}; however, flexor effort during velocity control reduced significantly when feedback was on ($p<0.05$, power>0.8). In addition, a significant reduction in flexor effort was found when comparing PF_{On} with VF_{On} ($p<0.001$, power>0.8). Contrarily, without feedback, the flexor effort was not significantly different across PF_{Off} and VF_{Off} trials ($p>0.05$). Lastly, across all comparisons, no significant difference was found in the extensor effort ($p>0.05$).

IV. Discussion

This study determined how non-invasive tactile feedback and distinct myoelectric control strategies impacted the functional utility of a prosthetic hand. Our findings showed that participants could effectively use the evoked fingertip tactile feedback with a higher success rate in transferring objects of various weights, when both position- and velocity-control schemes were used. We also found a lower grip force and force variability when feedback was provided. These results reveal that non-invasive somatotopic tactile feedback can enable bidirectional closed-loop control of a prosthetic hand. The outcomes can provide insight into the intricate action-perception coupling during human-robot interactions, potentially improving the utility and dexterous control of assistive devices.

The total number of failed trials diminished when feedback was provided. This trend was observed during both position- and velocity-control. The results suggest that the participants reliably perceived that a stable grasp had been made. In contrast, without tactile feedback, only visual feedback could be used to discern the interaction. Similar findings have been reported in previous studies, where feedback led to improved object transfer performance and reduced failure rates [23], [30], [31]. When evaluating across experimental conditions with the same feedback status, the total number of failures was higher during position-control compared with velocity-control. The difference between control strategies may be caused by the reduced grasp force variability during position-control, leading to a smaller number of failed trials. Although two participants described position control as allowing for finer control of the prosthesis motion, velocity control does not require sustained muscle activation to maintain a given position. During the experiment, the participants were instructed to exert the minimal force possible, and, as a result, the failed trials were partly due to a lack of information on the grip forces. In addition, the force traces in the failed trials were not included when assessing the force level and force variability in the holding phase. Prior work has shown that position control leads to greater joint angle variability compared with velocity control [32]. This is due to the stochastic nature of the EMG signals and the need for sustained muscle activation during position control. This could potentially cause the finger to extend unintentionally leading to a dropping of the object. A reduction (numerically) of force variability was observed in the feedback-on condition when compared with the feedback-off condition; however, the reduction was only significant in certain conditions. This could potentially explain why the number of failed trials was greater when feedback was off but was not insignificantly different from the feedback-on condition. Further investigation is needed to quantify the mechanisms behind these failed trials to better understand how sensory feedback and controller design can be improved.

The average steady state holding force reduced when tactile feedback was provided when using both position- and velocity-control modes. On average, the tactile feedback reduced the exerted force in a given control scheme by approximately 33.1%. The reduction in grip force with tactile feedback has been shown in prior work [33]–[35]; however, conflicting results have been reported as well [36]. These different outcomes are likely due to differences in the device control scheme, feedback properties, and manipulated objects. For example, one study showed that a two-channel differential force controller led to no difference in the grip force with and without vibratory feedback [36]. The controller required healthy individuals to use their intact hand to press on force sensors to control a prosthetic hand. Greater controllability is likely achieved than myoelectric control strategies, which can potentially explain the lack of difference in grip force across conditions. Both invasive [23], [34], [35] and non-invasive [17], [33] electrostimulation tactile feedback strategies have shown reduced grasp forces when using myoelectric control. A study also showed that non-somatotopic modality-matched feedback lead to slight improvement during grasps; however, the outcomes were inconsistent due to excessive data variation [37]. In addition, the feedback modality can affect the perception of the stimuli. Studies have shown that the intensity and location of somatotopic feedback can be perceived more accurately during single and dual tasks, compared to sensory substitutional techniques [12]. Lastly, the objects tested may also play a role as well. In the current study, visually identical rigid cubes were used during the experiment, in order to limit visual cues about the grip force. Visual and auditory cues were not removed to replicate an ecological setting in which the prosthetic device would typically be used. In the future, it is necessary to explicitly evaluate the contribution of visual and auditory cues during object manipulations.

When evaluating across object weights, no significant differences were observed for any of the experimental conditions. Prior work has shown that force could be regulated based on object weight [36]. Specifically, participants could produce different force levels when grasping a light (150 g) or a heavy (300 g) object. In the current study, some differences, though not significant, in grip force were observed with tactile feedback; however, many of the participants reported that they could not easily distinguish the three objects due to the weight of the sensorized prosthetic hand (595 g) and the arm attachment (205 g). Other factors may include the relative difference in object weight, the stimulation range for each participant, and the speed of the prosthetic hand. In the current study, the object weights increased in a step of 50 g. In the prior study, the difference between the two objects was 150 g [36], making the two objects easier to distinguish. Next, a small electrical stimulation range can reduce the number of perceived stimulation levels and in turn the resolution of force amplitude perception. Adjusting the stimulation waveform through reductions in pulse width and the addition of an interphasic delay may potentially improve the resolution of the tactile perception [38]. In addition, employing a more biomimetic approach to associating the waveform to a given force response may also enhance the stimulus resolution [31]. Lastly, the prosthesis joint speed likely affected performance. Reducing the maximum speed allows for more time to perceive the stimuli during active prosthetic control [19], potentially leading to greater differences in force exerted across object weights.

The effort (muscle activation) required to complete the task was also influenced by the tactile feedback and control scheme. Overall, the effort required for position control was greater than velocity control when tactile feedback was provided. This is due to the nature of the two controllers. The tactile feedback also reduced the flexor activation when using the velocity controller. Participants felt more confident after object contact was made, and the flexor muscle were relaxed sooner, accounting for the difference across the two feedback conditions. When feedback was provided during position control, no significant difference was observed for either muscle. The sustained muscle activation during position control may explain the lack of significance. In addition, the prosthesis was controlled using the difference between the two EMG channels. As a result, a certain level of sustained co-contraction of the extensor during grasping would require greater levels of flexor activation as well. To account for potential co-contraction, a single channel EMG electrode placed on the finger flexor could be used to operate the prosthesis; however, the prosthesis response would likely be reduced for hand opening. Alternatively, a holding state could potentially be employed in the position controller to reduce the need for sustained muscle activation, limiting user effort [37].

Prior work has shown that relatively localized percepts can be elicited [26], including a single quadrant along the fingertip. The increase in sensory region evoked in the current study is likely caused by the increase in current amplitude. As the stimulation amplitude is increased to modulate the sensation intensity, the area of the evoked percept region is increased as well due to additional axonal recruitment [17]. During the stimulation grid exploration, we were less concerned with identifying a pair that evoked sensation only in the index finger, and the exploration stopped as soon as sensation was identifiable in the index or middle finger. It is possible that more extensive grid exploration could allow us to identify localized sensation only in the index finger. Alternatively, a smaller electrode grid may lead to more selective axon recruitment, due to a more localized electric field, which in turn could lead to single finger sensations. Additionally, a more complex stimulation pattern may lead to more selective axon recruitment for localized sensation [38], [39].

V. Conclusion

The non-invasive nature of this closed-loop evaluation platform allows for use with different sensorized prosthetic hands. In addition, joint angle recordings could also be used to elicit proprioceptive feedback regarding a prosthetic's joint position. Artificial proprioceptive feedback can improve the controllability of the system [34]. In addition, the use of tactile and proprioceptive feedback allows for an all-encompassing narrative of a prosthetic's movement, limiting user reliance on visual cues. Information pertaining to tangential forces and object slippage may also improve the observed outcomes in this study. This approach can also be implemented to provide feedback to individuals with sensory deficits or for use with teleoperated robotic devices. Sensory feedback can be used by persons with sensory impairments to enhance their limb control, potentially improving their overall quality of life. Future evaluations will assess the role this approach plays on these different applications.

Prior work showed that the haptic percepts elicited via transcutaneous electrical nerve stimulation are similar across individuals with and without arm amputations [27]. As a result, it is believed that the results observed in neurologically intact participants may depict those expected in amputees. It is possible that the haptic feedback elicited in able-bodied participants is likely correlated to those experienced in amputees. Prior work showed that similar success rates are achieved during object recognition tasks across individuals with and without amputation [25]. Nonetheless, future tests will assess how amputees incorporate tactile feedback during functional utility of their prostheses. Subjective analysis through questionnaires can also be used to better understand the differences across conditions from a user's perspective.

Overall, our study shows that tactile feedback delivered via transcutaneous nerve stimulation could be effectively integrated into closed-loop control of a prosthetic hand, enabling the human-in-the-loop control of assistive robots. Greater success in object manipulation and a better control of grip force were observed with tactile feedback when employing both myoelectric controllers. The outcomes can help support the system robustness and facilitate functional utility of prosthetic hands. Our findings also illustrate the differences in the intricate action-perception coupling involved in distinct controllers with an assistive device. The outcomes also help us to understand the interactions between the motor and sensory modules of current prosthetic systems during functional use, which can provide guidance for prosthesis design in order to further improve user confidence and experience in the future.

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