Getting a Grip on the Impact of Incidental Feedback From Body-Powered and Myoelectric Prostheses

Michael A. Gonzalez, Christina Lee, Jiyeon Kang, *Member, IEEE*, R. Brent Gillespie, *Member, IEEE*, and Deanna H. Gates, *Member, IEEE*

***Abstract*—Sensory feedback from body-powered and myoelectric prostheses are limited, but in different ways. Currently, there are no empirical studies on how incidental feedback differs between body-powered and myoelectric prostheses, or how these differences impact grasping. Thus, the purpose of this study was to quantify differences in grasping performance between body-powered and myoelectric prosthesis users when presented with different forms of feedback. Nine adults with upper limb loss and nine without (acting as controls) completed two tasks in a virtual environment. In the first task, participants used visual, vibration, or force feedback to assist in matching target grasp apertures. In the second task, participants used either visual or force feedback to identify the stiffness of a virtual object. Participants using either prosthesis type improved their accuracy and reduced their variability compared to the no feedback condition when provided with any form of feedback ($p < 0.001$). However, participants using body-powered prostheses were significantly more accurate and less variable at matching grasp apertures than those using myoelectric prostheses across all feedback conditions. When identifying stiffness, body-powered prosthesis users were more accurate using force feedback (64% compared to myoelectric users' 39%) while myoelectric users were more accurate using visual feedback (65% compared to body-powered users' 53%). This study supports previous findings that body-powered prosthesis users receive limited force and proprioceptive feedback, while myoelectric prosthesis users receive almost no force or proprioceptive feedback from their device. This work can inform future supplemental feedback that enhances rather than reproduces existing incidental feedback.**

***Index Terms*—Feedback, human-machine interaction, prosthesis, sensing.**

I. Introduction

Analogues to the mechanoceptors and proprioceptors of the natural hand do not exist in conventional body-powered or myoelectric prostheses. This denies prosthesis users access to the reliable sensory feedback and subconscious feedback processing available to an anatomical limb when they perform tasks with their device. Thus prosthesis users must grasp and manipulate objects without the benefit of the reliable, native sensory feedback and subconscious feedback processing that usually guides such tasks [1], [2]. While native sensory feedback from the anatomical limb is absent, prosthesis users still have access to natural cues from their prosthesis. This incidental feedback includes visual, auditory, and socket-transmitted loads or vibrations [3]–[5]. However the relative availability and utility of incidental feedback between body-powered and myoelectric prosthesis users has only been characterized through anecdotal evidence [6].

The feedback available from a body-powered or myoelectric prosthesis is determined, in part, by each device's respective means of control and actuation. Body-powered prostheses are actuated by a Bowden cable that ties movements of the scapula to opening or closing movements of a terminal device. This direct connection between shoulder and prosthesis movement provides users with a sense of prosthesis configuration, also called extended physiological proprioception (EPP). EPP is thought to support the process by which a prosthesis becomes an extension of the user's body [3]. EPP is most effective in a system that couples user motion to prosthesis motion with limited friction or slack, such as the Bowden cable [7]. Evidence also suggests that prosthesis users can use tension in the Bowden cable to detect resistance forces when grasping an object with the prosthetic end effector, though this has only been demonstrated in healthy individuals using a voluntary-close body-powered prosthetic emulator [4], [8]. In contrast, myoelectric prostheses are driven by a motor that is controlled by electromyographic (EMG) signals detected from the residual limb. For most myoelectric devices, the magnitude of the EMG signal maps proportionally to the speed of terminal device movement. Incidental feedback from myoelectric devices includes motor sounds and vibrations or forces

Manuscript received April 8, 2021; revised August 25, 2021; accepted August 27, 2021. Date of publication September 13, 2021; date of current version September 21, 2021. This work was supported by the Office of the Assistant Secretary of Defense for Health Affairs through the Orthotics and Prosthetics Outcomes Research Program under Award W81XWH-16-1-06548. *(Corresponding author: Deanna H. Gates.)*

This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the University of Michigan's Medical School Institutional Review Board.

Michael A. Gonzalez is with the Robotics Institute, University of Michigan, Ann Arbor, MI 48105 USA (e-mail: magonzo@umich.edu).

Christina Lee is with the Department of Biomedical Engineering, University of Michigan, Ann Arbor, MI 48105 USA (e-mail: chslee@umich.edu).

Jiyeon Kang is with the Department of Mechanical and Aerospace Engineering, University at Buffalo—SUNY, Buffalo, NY 14260 USA (e-mail: jiyeonk@buffalo.edu).

R. Brent Gillespie is with the Department of Mechanical Engineering, University of Michigan, Ann Arbor, MI 48105 USA (e-mail: brentg@umich.edu).

Deanna H. Gates is with the School of Kinesiology, University of Michigan, Ann Arbor, MI 48109 USA (e-mail: gatesd@umich.edu).

Digital Object Identifier 10.1109/TNSRE.2021.3111741

transmitted through the socket [3], [5]. A sense of contraction or effort from the residual muscles that generate EMG signals can also be considered incidental feedback, though the relationship between effort and motion may not be as predictable in myoelectric devices compared to body-powered devices. All of these signals can be used to guide control actions, or be used to internalize the mapping from muscle activation to movement speed and thereby establish predictive control over the device [5], [9]. While some commercially available myoelectric devices have built-in force sensors to prevent a held object from dropping or slipping (e.g. i-Limb, bebionic), such inner loop control actions or grasping force information are not explicitly relayed back to the user.

To address the lack of native sensory feedback from either type of modern prosthesis, researchers have developed various means for providing prosthesis users with supplemental feedback, which describes any artificial exteroception or proprioception provided to an individual. Specifically, researchers have studied how the addition of supplemental feedback can improve accuracy for prosthesis users in grasping and manipulation tasks. The most common method for delivering information about grasp aperture or grip force is through sensory substitution [10], wherein sensory qualities are encoded into a signal that is delivered through an alternate feedback modality. The feedback modalities used in sensory substitution for grasp aperture and grip force predominantly include electrotactile [11], [12], vibrotactile [13]–[18], audio [18], [19], and visual feedback [13], [17]. When sensory substitution was used to provide feedback regarding the pose of a virtual or emulated prosthesis, individuals more accurately modulated their EMG levels [16], moved their hand into a target pose [19], [20], and differentiated object sizes [21] compared to conditions without additional sensory feedback. Providing grip force feedback to participants allowed them to more accurately modulate grip force [13] and identify object stiffnesses [4], [21], and more quickly detect and stop objects from slipping [22], compared to when no feedback was given. However, there exist examples of feedback modality combinations performing more poorly than individual modalities [16] or feedback modalities providing limited benefit for certain tasks [21]. Thus it is critical to study the contexts under which sensory substitution should be implemented in a prosthesis.

The development of effective supplemental feedback in prosthetic devices is also dependent on quantifying the incidental feedback already available to users of commercially available prostheses. By quantifying incidental feedback, it may be possible to establish a baseline for the information that prosthesis users receive. Supplemental feedback can then be applied to augment, rather than conflict with, the existing feedback. The utility of incidental feedback has been characterized through myoelectric control experiments, however these studies involved either individuals without limb loss using a prosthetic emulator [5], [14] or individuals with amputation using equipment provided by the study [9]. In fact, a recent literature review found that no studies of prosthetic sensation measured the performance of individuals with limb loss using their own device [10]. These findings are valuable for providing comparisons across feedback modalities and sensory substitution methods, but it may be difficult to generalize these findings to a population of prosthesis users who have more experience with their devices. Additionally, it is important to evaluate feedback in contexts that will be relevant to prosthesis users. Many studies have highlighted the importance of stiffness and force identification in manipulation tasks [21], [23]–[25], as these skills address the concerns of an individual crushing or dropping an object with their prosthesis. Pose matching tasks are also relevant, though less common [16], [20], as they quantify a prosthesis users' ability to match their hand aperture to the size of the object to initiate a grasp. While force feedback is typically delivered continuously [21], [23]–[25], hand aperture can be delivered either continuously [16], [17], [20], [21] or discretely [9], [26], with discrete methods typically denoting the contact events for grasping or releasing an object. Together, the context of a task and the type of prosthesis an individual is using may affect the utility of the feedback being delivered, or how available a particular feedback modality is to the user at all.

The purpose of this study was to quantify the utility of incidental feedback in the context of grasping and manipulation tasks performed by persons using their own body-powered or myoelectric prostheses. Throughout the study, participants interacted with a haptic device that could simulate physical objects. The study consisted of a grasp aperture matching task [8] in which we rendered force, vibration, or visual feedback at two levels to represent discrete contact events at the boundaries of a virtual object, and a stiffness identification task in which we provided graded force or visual feedback to simulate the properties of a physical object. We hypothesized that when participants attempt to match a target aperture using discrete contact events, performance would be consistent across feedback modalities (visual, vibration, or force) and be higher for body-powered prosthesis users than myoelectric users, on average. We expect these differences in performance to be greatest when force feedback was provided, because force feedback is purportedly less available through myoelectric than body-powered prostheses. Similarly, we expected myoelectric users would be less accurate compared to body-powered users at stiffness identification when presented with continuous force feedback but similarly accurate when presented with continuous visual feedback. We also recruited persons without limb loss to participate, to provide a performance baseline.

II. Methods

A. Participants

Nine adults with unilateral, transradial amputation (4F/5M, age 48.0 $\pm$ 16.2 years) and nine age- and sex-matched controls without amputation (4F/5M, age 44.7 $\pm$ 14.7 years) participated (Table I). Participants were recruited through an online database (umclinicalstudies.org) and through prosthetist referral. Potential participants were excluded if they were 18 or younger, had any history of neurological disorders or orthopedic conditions affecting their upper limbs (excluding the residual limb of participants with amputation), or if they had any significant self-reported visual or hearing impairments that would prevent them from completing the study protocol. All participants provided their written informed consent prior

TABLE I
PARTICIPANT DETAILS

Participant	Age	Sex	Cause of limb loss	Time since amputation (years)	Prosthesis type		Device experience (years)
P1	32	M	Acquired	1.2	Body-powered	Voluntary-open	0.8
					Myoelectric	Multi-DOF Hand	0.4
P2	52	F	Acquired	0.8	Body-powered	Voluntary-open	0.6
P3	55	F	Congenital	N/A	Myoelectric	1 DoF Hand	33.0
P4	66	F	Congenital	N/A	Myoelectric	Multi-DOF Hand	0.5
P5	26	F	Congenital	N/A	Myoelectric	Multi-DOF Hand	0.8
P6	29	M	Acquired	2.5	Body-powered	Voluntary-open	2.0
P7	46	M	Acquired	10.0	Body-powered	Voluntary-open	7.0
P8	54	M	Acquired	24.8	Body-powered	Voluntary-close	23.0
					Myoelectric	1 DoF Hand	23.0
P9	72	M	Acquired	0.9	Body-powered	Voluntary-open	0.3

*Device experience refers specifically to the time an individual has been using their current type of device (e.g. A 1 DoF Hand versus a multi-DoF hand) rather than overall prosthetic experience

to participation in this study whose protocol was approved by the University of Michigan's Medical School Institutional Review Board.

B. Experimental Protocol

All participants completed two experiments designed to assess different components of grasping. The first assessed each participant's ability to match a target grasp aperture, while the second assessed their ability to distinguish object stiffness. In both experiments, participants interacted with a custom device [8] that was designed to provide force and vibration through two motorized hand guides. The two experiments were performed either on different days, or on the same day with an extended break between, depending on participant schedule and fatigue. Each experiment took about one hour to complete per limb. Participants with amputation used their own prostheses. Three participants used body-powered prostheses, three used myoelectric prostheses, and two used both devices (Table I). Prosthesis users completed each experimental protocol with their intact hand and with the terminal device of their prosthesis or prostheses.

1) Grasp Aperture Matching: Participants placed their hand or prosthesis's terminal device into the two hand guides (Fig. 1A). Seat height and arm rest height were then adjusted to support the weight of the arm and ensure a comfortable hand posture. A shroud was placed over the device during the experiment to prevent the participant from seeing their hand (Fig. 1B). For individuals using a prosthesis, researchers notified participants to re-position their prosthesis if it slipped out of the device.

Participants were instructed to match the aperture of their hand to a target aperture shown on a computer monitor (Fig 1B). These targets consisted of small (5 mm), medium (11 mm), and large (17 mm) grasp apertures, presented in random order. The experimental device provided a stimulus if participants closed their grasp narrower than the target. This stimulus was removed if the participants then returned their grasp to be wider than the target. Participants were instructed to find the threshold where feedback switched on/off and to hold their hand position until the task timed out at 7 s. Targets shown on screen were matched one-to-one with the physical distances participants needed to match using the hand guides.

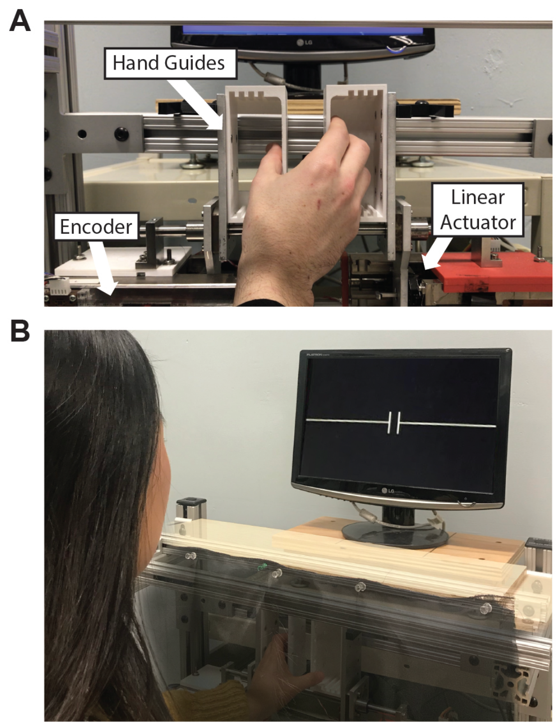

Fig. 1. Haptic object device setup. A) Participants placed their thumb and other four fingers, or the two ends of their terminal device, in either hand guide. Hand guides were attached to linear actuators which provided force and vibration feedback. Encoders relayed the hand guide position to the computer in real time. B) Participants were seated in front of a computer monitor that provided trial start cues, end cues, and visual feedback. A shroud (shown with increased transparency in this image) covered the device and the participant's hand during all trials.

The binary 'too narrow' or 'too wide' feedback was chosen to allow direct comparison of the different forms of incidental feedback.

Participants completed the aperture matching task under five conditions. These included a visual condition, a vibration condition, a low level force condition, a high level force condition, and a no feedback condition. These feedback modalities were chosen to represent the incidental feedback commonly available to individuals, such as visual and force feedback, and one of the most prevalent modalities for sensory substitution, vibrotactile feedback [10]. In the visual feedback condition, participants were presented with an empty box on the monitor if their grasp was too wide and a filled box when their grasp was too narrow. For vibration feedback, we vibrated the hand guides (magnitude: 0.5 N, frequency: 80 Hz) when the grasp was too narrow. Force feedback conditions were implemented using a ramp function over a 5 mm window, centered at the target, from 0 N to 5 or 30 N, for low or high force respectively. Once a grasp was more narrow than the window, force was constant at the high or low value, depending on the condition. Participants were allowed to practice each feedback condition at each aperture size prior to data collection until they were comfortable. Finally, in the no feedback condition, participants were shown the target aperture briefly and then told to match the target without any additional cues. We did not, however, control for other incidental feedback such as auditory cues from the experimental device or myoelectric motors, or forces generated between the prosthetic socket and residual limb. Ordering of the feedback conditions was pseudo-randomized, such that the no feedback condition was never performed first.

For each feedback condition, participants were presented with each of the three target widths 10 times, for a total of 30 trials per condition. They had a minimum of 4 s of rest between each trial, with 10 s of rest every 10 trials. Longer breaks were given between different feedback conditions. Participants with a prosthesis completed each condition first with their intact hand, followed by their prosthetic hand. Those without limb loss completed each condition first with their dominant hand, followed by their non-dominant hand.

2) Stiffness Identification: In this task, participants were asked to identify the stiffness of simulated springs as low, medium, or high using only force feedback or only visual feedback.

During force feedback trials the device simulated virtual springs without damping so that force output was only dependent on position and a selected spring constant. The force F commanded to the linear motors was

$$F = -k_s * (d_0 - d), \tag{1}$$

where d_0 was the fully-open hand guide position, d was the current hand guide position, and k_s was the spring stiffness. The values of k_s were chosen to be distinguishable (accuracy greater than 33.3%, or random chance) but not trivially so (accuracy $<$ 100%), according to pilot test results. This resulted in virtual springs of low, medium, and high stiffness of 200 N/m, 550 N/m, and 1500 N/m, respectively.

For the visual condition, participants observed a virtual object on the computer monitor that would deform horizontally as the participant closed the hand guides. Objects of differing stiffnesses were represented by differing rates of deformation. The medium spring deformed in 1:1 scaling with the distance between the hand guides (1 cm of hand guide compression corresponded to 1 cm of virtual object compression). The soft spring deformed more quickly (1:2.5), while the hard spring deformed more slowly (1:0.4). The visual scaling values were chosen through pilot testing. A constant 2 N force was applied outward to assist with opening the hand guides to facilitate multiple probes of the virtual object, regardless of the visually presented stiffness.

For each feedback condition, the virtual objects were presented randomly over 30 trials, with each stiffness presented 10 times. The participants started each trial with hand guides in their furthest separated positions, and had up to 20 seconds to probe each virtual object before making their identification.

To familiarize participants with the protocol, we first conducted a short training session using physical objects. Participants were presented with sets of foam blocks of 'Low', 'Medium', and 'High' stiffnesses. Each set also contained blocks of three different sizes. To familiarize participants with identifying stiffness using only visual information, the experimenter compressed each of the blocks with approximately the same amount of force and had participants attempt to identify the object stiffness without physical interaction. Participants were then allowed to manipulate the blocks themselves. To simulate identifying stiffness with only force feedback, participants were asked to close their eyes as the researchers placed a random block in their intact hand and asked them to report the block's perceived stiffness (Low, Medium, or High).

C. Data Analysis

Grasp aperture was measured from linear optical encoders at 1 kHz using a data acquisition card. Aperture error was calculated as the average difference between the target aperture and the hand position during the last 1 s of each trial. Thus, positive errors represent grasps which were too narrow. Error variability for each condition was quantified as the standard deviation of the aperture error across trials. We also took the average absolute difference between the target aperture and the hand position during the last 1 s of each trial, referred to here as absolute aperture error. For the stiffness identification task, researchers recorded participant responses via keyboard press. Accuracy was the percentage of the total responses that were correct for each condition. We also created confusion matrices of the presented stiffnesses versus the participants' stiffness identifications.

We excluded trials in which the participant's prosthesis slipped out of the hand guides, the participant needed to remove their hand due to discomfort, or the participant had difficulty opening or closing their prosthesis. This process excluded 2% (n = 96) of grasp aperture matching task trials, 90% of which were during trials completed with a prosthesis. We also excluded 1% (n = 13) of stiffness identification task trials, 85% of which were during trials completed with a prosthesis. Additionally, due to errors in data collection during the grasp aperture matching task, P6 is missing data for the High Force condition and P8 is missing data for their completion of the High Force and no feedback conditions using their myoelectric prosthesis.

D. Statistical Analysis

The primary dependent measures for the grasp aperture matching task were error, absolute error, and error variability. For the stiffness identification task, the primary dependent measure was identification accuracy. We first tested for differences in all outcomes between anatomical limbs using linear mixed-effect models in which limb (dominant and non-dominant for control participants, intact and prosthetic for prosthesis users) was a fixed factor and participant was a random factor. As there were no significant differences between limbs, we combined dominant, non-dominant and intact limbs into an 'anatomical limb' group.

We tested for differences between limbs and feedback types using a series of linear mixed models in which limb (anatomical, body-powered, myoelectric) and feedback type were predictor variables and participants was a random factor. The grasp aperture matching tasks had 5 levels for feedback (No Feedback, Visual, Vibration, Low Force, and High Force) while the stiffness identification task had two (Visual and Force). Significant main effects and interactions were explored using estimated marginal means with a Sidak correction for multiple comparisons. All statistical analyses were performed using SPSS 27 (IBM Corp., Armonk, N.Y., USA), with $\alpha = 0.05$. We calculated effect sizes (Hedges' g) for pairwise comparisons between participants using body-powered and those using myoelectric prostheses within each feedback condition. Effect sizes were considered small ($g \geq 0.2$), medium ($g \geq 0.5$), or large ($g \geq 0.8$) [27].

III. Results

A. Grasp Aperture Matching

Grasp aperture error was not affected by limb ($p = 0.607$) or feedback type ($p = 0.067$) (Fig. 2A), nor were there any significant interactions ($p = 0.137$). Differences in grasp aperture error between participants using body-powered versus myoelectric prostheses had small to large effect sizes across conditions (Visual: $g = 0.90$, Vibration: $g = 0.84$, Low Force: $g = 0.24$, High Force: $g = 0.75$, No Feedback: $g = 0.47$).

There were significant main effects of limb ($p < 0.001$) and feedback type ($p < 0.001$) on absolute aperture error (Fig. 2B). Participants had larger absolute errors when using either a body-powered ($p < 0.001$) or myoelectric ($p < 0.001$) prosthesis compared to an anatomical limb. Participants using body-powered prostheses had lower errors than those using myoelectric prostheses ($p < 0.001$). There were small to large effect sizes for the differences between prostheses across feedback conditions (Visual: $g = 1.13$, Vibration: $g = 0.73$, Low Force: $g = 1.23$, High Force: $g = 0.63$, No Feedback: $g = 0.07$). Participants had greater absolute error when grasping with no feedback compared to grasping during any feedback condition ($p < 0.001$), regardless of which limb was used.

Error variability was affected by both limb ($p < 0.001$) and feedback type ($p < 0.001$) (Fig. 2C). Participants had greater error variability when using either a body-powered ($p < 0.001$) or myoelectric ($p < 0.001$) prosthesis compared to an anatomical limb. Participants using a body-powered prosthesis were less variable than those using a myoelectric prosthesis ($p < 0.001$). These differences varied across feedback conditions with small to medium effect sizes (Visual: $g = 0.74$, Vibration: $g = 0.66$, Low Force: $g = 0.21$, High Force: $g = 0.79$, No Feedback: $g = 0.24$). Participants had greater error variability while grasping with no feedback compared to grasping during any feedback condition ($p < 0.005$).

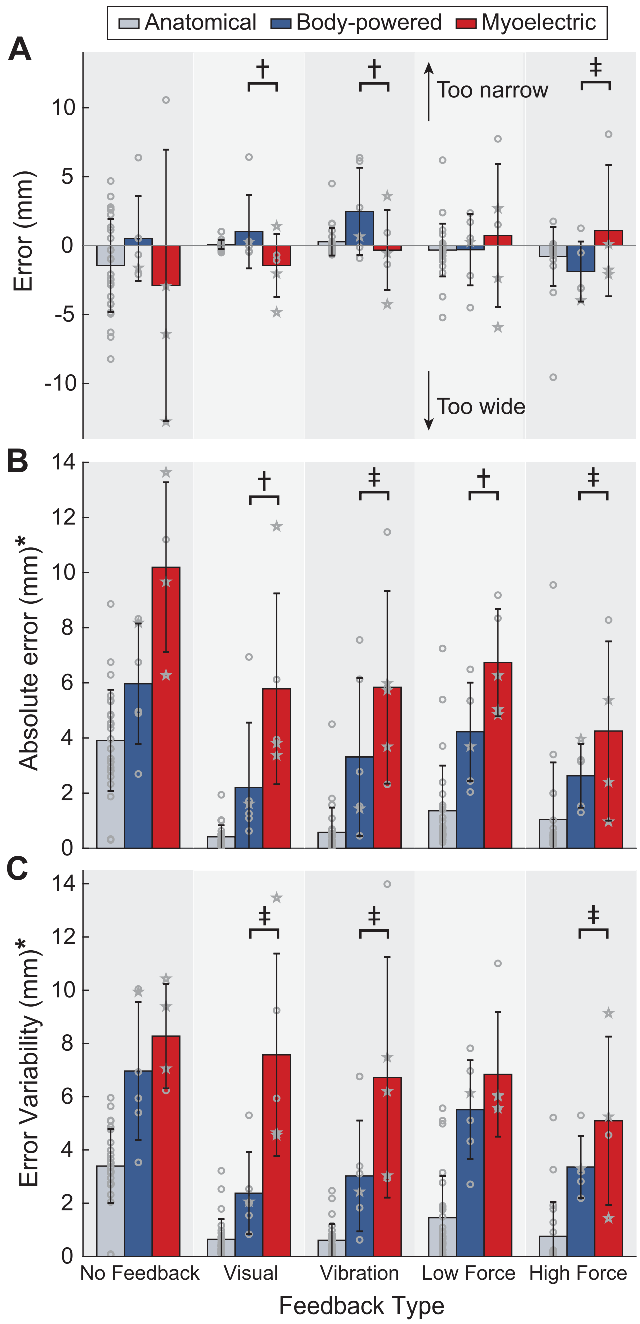

Fig. 2. A) Error, B) absolute error, and C) error variability between the hand/prosthesis aperture and target aperture for each feedback condition. Bars represent the average across the group, while individual averages are shown as points. Error bars are one standard deviation. Data for voluntary-close body-powered devices and multi-articulated myoelectric hands are indicated by stars. All other data are shown as open circles. '*' indicates a significant main effect of limb. Large and medium effect sizes for the pairwise comparison between body-powered and myoelectric prostheses are denoted by '†' and '‡', respectively.

B. Stiffness Identification

Participants using their anatomical limbs most frequently confused similar stiffnesses (i.e. confusing low and medium) (Fig. 3A). Prostheses users had more variable responses, confusing both similar and dissimilar stiffnesses (i.e. confusing low with high). There was a significant main effect of limb type ($p < 0.001$) and a significant limb $\times$ feedback type interaction ($p < 0.001$) for stiffness identification accuracy (Fig. 3B). Participants using their anatomical limb could identify stiffness more accurately across both feedback conditions (73.1%) compared to those using either a body-powered (58.3%; $p = 0.006$) or a myoelectric (51.6%; $p < 0.001$) prosthesis. When presented with force feedback, participants using body-powered prostheses could identify stiffness more accurately (64.0%) than those using a myoelectric prosthesis (39.3%; $g = 1.32$). In contrast, when presented with visual feedback, participants using myoelectric prostheses identified stiffness more accurately (65.3%) than those using body-powered prostheses (53.3%; $g = 0.79$).

IV. Discussion

This study quantified the incidental feedback available to individuals using their own body-powered or myoelectric prostheses by having participants complete both a grasp aperture matching task and a stiffness identification task under various feedback conditions. When matching grasp apertures, limb type had a greater effect on performance than feedback type. Those using body-powered prostheses were more accurate than those using myoelectric prostheses, regardless of feedback type. Regardless of limb type, all supplemental feedback improved performance compared to the no feedback condition. When identifying stiffnesses using visual cues alone, all limb types had similar accuracy. Predictably, when participants identified stiffness using force cues alone there was a large limb type effect. Those using anatomical limbs identified stiffness most accurately and those using a myoelectric prosthesis identified stiffness least accurately. These results suggest that individuals using either prosthesis type are able to use any feedback available to detect a contact event, but myoelectric prostheses do not provide sufficient force feedback to assess stiffness.

Regardless of whether visual, vibrotactile, or force feedback was provided, participants using myoelectric prostheses were less accurate and more variable at matching grasp apertures than those using body-powered prostheses. However, most participants were able to use any feedback modality to improve their performance relative to the no feedback condition. The parity across feedback conditions may be due to the way in which the feedback was provided. Feedback was binary, indicating that participants were either "too wide" or "too narrow". While this does not necessarily represent natural grasping in an anatomical hand, discrete cues have been used in upper limb prosthetics research [9], [26] and enable us to make more direct comparisons between feedback types. In fact, one study demonstrated that discrete vibrotactile feedback was utilized by participants even when continuous auditory feedback was available [18]. However, it is unclear in such studies if differences in performance are due to differences in the presentation of feedback or due to differences in the feedback modality. For example, in many studies that provide natural visual feedback, the visual feedback condition typically gives participants real-time information on hand position, hand velocity, and target position [16], [20]. Other feedback modalities, typically vibrotactile [16] or force feedback [20], are typically scaled to end effector pose or to the error between the current end effector pose and a target pose. Therefore differences in performance may be combinations of differences in feedback modality as well as differences in the amount of information or noise present in a modality by nature of how it is presented.

Participants using myoelectric devices were not less accurate at matching grasp apertures when using force feedback compared to other feedback modalities as hypothesized. Anecdotally, myoelectric devices do not provide force feedback of grip force [6], [28] so when we provided force feedback to individuals using their myoelectric device to match grasp apertures we expected little to no increase in accuracy compared to when they received no feedback. Surprisingly, myoelectric prosthesis users made use of any type of feedback, including force feedback, to improve grasp aperture matching performance relative to no feedback. When asked after completing the experiment, participants reported feeling interaction forces between their residual limb and prosthetic socket, particularly for medium and large apertures. Socket interaction forces have been discussed previously as one of many incidental cues available to body-powered and myoelectric prosthesis users [3]. However, the utility of these socket interaction forces in functional tasks has not been widely quantified in the literature.

Participant ability to identify object stiffness was impacted by limb type in the force feedback condition, but did not significantly vary between limb types in the visual feedback condition. Participants, regardless of limb type, were moderately accurate ($\sim$60% accurate) when identifying an object's stiffness using visual feedback alone. Under the visual feedback condition, large misclassifications (mistaking low and high stiffnesses) were also uncommon. A previous study of healthy individuals using prosthetic emulators found that participants could only identify three objects stiffness at 43% accuracy with visual feedback alone [4]. However, these individuals were manipulating physical sponges which would deform less consistently and with greater nonlinearity than our virtually-rendered, ideal springs.

Notably, identifying stiffness using visual feedback alone was the only condition in which participants using myoelectric prostheses were nominally more accurate than any other group. Previous evidence shows that prosthesis use, in general, demands a greater reliance on visual feedback compared to anatomical limb use [3], [6]. However, no particular advantage in the use of visual feedback has been empirically shown for either body-powered or myoelectric prostheses when grasping objects. While our results may indicate that those using myoelectric prostheses are marginally more sensitive to visual feedback for identifying object stiffness, our experimental setup obscures the view of the end effector. This is relevant as body-powered end effectors typically obscure grasped objects less than myoelectric end effectors [6]. It should also be noted that during the visual condition, we still applied a retraction force to facilitate multiple probes of the virtual object within a trial. While we explicitly instructed participants to only use visual feedback to identify stiffness in the visual condition, it is possible that the retraction force was interpreted as a cue by some participants.

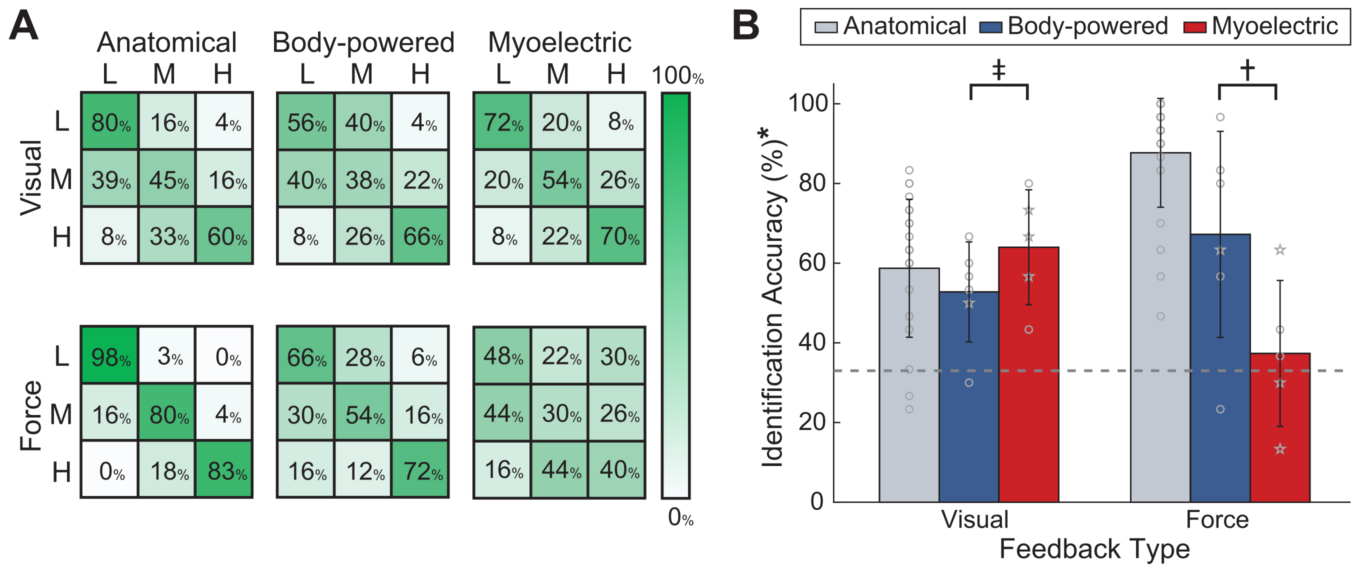

Fig. 3. A) Confusion matrices for the stiffness identification task for each type of limb. The y-axis is the presented stiffness (L: Low, M: Medium, H: High) and the x-axis is the participant-identified stiffness. Perfect accuracy would be indicated by a 100% along the diagonal. B) The average identification accuracy for the stiffness identification task. Bars represent the average across the group, while individual averages are shown as points. Error bars are one standard deviation. Data for voluntary-close body-powered devices and multi-articulated myoelectric hands are indicated by indicated by stars. All other data are shown as open circles. The dotted line marks 33%, or random chance. There was a significant limb effect across conditions, indicated by a '*' on the y-axis label, and a significant limb $\times$ feedback type interaction ($p < 0.001$). '†' and '‡' indicate large and medium effect sizes for the pairwise comparison between body-powered and myoelectric prostheses, respectively.

In contrast to the visual-only feedback condition, when participants used only force feedback to identify stiffness there were differences in performance between limb types. As expected, individuals using body-powered prostheses were more accurate than those using myoelectric prostheses, and both were less accurate than those using their anatomical limbs. This agrees both with anecdotal evidence [6], [28] as well as with previous work in which healthy individuals were provided with force feedback through prosthetic emulators [4]. Notably, those using myoelectric prostheses performed close to chance when determining stiffness using force feedback. This indicates that identifications were made almost randomly, which is supported by the nearly even distribution of percentages in the myoelectric force confusion matrix. Participants also did not report being able to take cues from socket forces, as they were able to do in the grasp aperture matching task. More than any other result, the inability of those using myoelectric prostheses to detect object stiffness indicates a gap in the functionality of current myoelectric devices that needs to be addressed.

We found the utility of feedback modalities in this study to vary across both prosthesis type and the task in which the feedback was presented. Primarily, participants using myoelectric prostheses were able to use force feedback to improve their grasp aperture error, but not to identify object stiffnesses. This supports previous findings that discrete cues can facilitate more accurate and reliable grasping in prosthesis users [9], [22], [26]. However, stiffness is a relationship between displacement and force, and this continuous relationship cannot be represented through discrete cues alone. When an individual uses a common myoelectric prosthesis, they may have unreliable control over their hand position due to noise in their muscle signals or a discrete set of possible hand positions. In contrast, body-powered prostheses and anatomical hands have no inherent noise that would impact their control, and are both analog in nature. The less reliable feed-forward control of myoelectric devices necessitates more use of feedback to close the control loop [14]. Thus, even if individuals using a myoelectric prosthesis could feel socket forces, they may also require some sense of hand position or velocity to accurately discern stiffness.

This study has several limitations. First, the small sample size limits the degree to which we can generalize our findings. Given the large number of conditions, we did not have the degrees of freedom for post-hoc pairwise comparisons. Instead, we calculated effect sizes for these comparisons. These can be used to power future studies comparing these feedback types directly. The small sample we recruited was also quite heterogeneous in the types of prostheses they used, their cause of limb deficiency, age, and their prosthetic experience. Each of these factors may affect neuroplasticity [29]–[31], and, correspondingly, their sensitivity to feedback. Additionally, the participants in this study may not represent the amputee population in terms of prosthetic experience. At the clinic we recruited from, the standard of practice is to first prescribe a body-powered prosthesis and then, if deemed medically necessary and covered by insurance, a myoelectric device. For this reason, all participants who used a myoelectric prosthesis in the study have some experience with a body-powered prosthesis, but not vice versa. Therefore we cannot make generalizations to "myoelectric users" or "body-powered users". Finally, our study was limited in its ability to decouple how feed-forward control and sensory feedback each contributed to task performance, particularly in the grasp aperture matching task. The resolution of possible hand positions for a myoelectric device may be significantly less than that of a body-powered prosthesis which has a continuous range of positions. Future work might eliminate, or at least control for, differences in feed-forward control in order to better isolate the impact of feedback on prosthetic function.

V. Conclusion

This work directly compares the utility of different feedback modalities for body-powered and myoelectric prosthesis users across two functional tasks. This work also supports previous anecdotal evidence [6] that individuals using a body-powered prosthesis receive more incidental haptic feedback and a greater degree of proprioception than individuals using a myoelectric prosthesis. When presented with the exact same force feedback, whether that was discrete or continuous, individuals using a body-powered prosthesis were more accurate and less variable than those using a myoelectric prosthesis. Of note, however, myoelectric prosthesis users were able to use incidental force feedback to improve their performance compared to when they had no feedback available when matching grasp apertures. We hope our findings can lead to novel prosthetic designs that augment a user's existing incidental feedback with supplemental force feedback or more explicit proprioceptive cues. In doing so, we may be able to reduce user dependence on visual feedback and ultimately improve prosthetic function.

Acknowledgment

The authors would like to thank John Busch for designing and fabricating the experimental apparatus.

References

[1] R. S. Johansson and J. R. Flanagan, "Coding and use of tactile signals from the fingertips in object manipulation tasks," *Nature Rev. Neurosci.*, vol. 10, no. 5, p. 345, 2009.

[2] J. W. Sensinger and S. Dosen, "A review of sensory feedback in upper-limb prostheses from the perspective of human motor control," *Frontiers Neurosci.*, vol. 14, p. 345, Jun. 2020.

[3] D. S. Childress, "Closed-loop control in prosthetic systems: Historical perspective," *Ann. Biomed. Eng.*, vol. 8, nos. 4–6, pp. 293–303, Jul. 1980.

[4] J. D. Brown, T. S. Kunz, D. Gardner, M. K. Shelley, A. J. Davis, and R. B. Gillespie, "An empirical evaluation of force feedback in body-powered prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 3, pp. 215–226, Mar. 2017.

[5] M. A. Wilke, C. Niethammer, B. Meyer, D. Farina, and S. Dosen, "Psychometric characterization of incidental feedback sources during grasping with a hand prosthesis," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, pp. 1–13, Dec. 2019.

[6] S. L. Carey, D. J. Lura, and M. J. Highsmith, "Differences in myoelectric and body-powered upper-limb prostheses: Systematic literature review," *JPO J. Prosthetics Orthotics*, vol. 29, no. 4S, pp. P4–P16, Oct. 2017.

[7] T. R. Farrell, R. F. Weir, C. W. Heckathorne, and D. S. Childress, "The effects of static friction and backlash on extended physiological proprioception control of a powered prosthesis," *J. Rehabil. Res. Develop.*, vol. 42, no. 3, p. 327, 2004.

[8] J. Kang, M. A. Gonzalez, R. B. Gillespie, and D. H. Gates, "A haptic object to quantify the effect of feedback modality on prosthetic grasping," *IEEE Robot. Autom. Lett.*, vol. 4, no. 2, pp. 1101–1108, Apr. 2019.

[9] M. Markovic *et al.*, "The clinical relevance of advanced artificial feedback in the control of a multi-functional myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, pp. 1–15, Dec. 2018.

[10] C. Antfolk, M. D'Alonzo, B. Rosén, G. Lundborg, F. Sebelius, and C. Cipriani, "Sensory feedback in upper limb prosthetics," *Expert Rev. Med. Devices*, vol. 10, no. 1, pp. 45–54, Jan. 2013.

[11] G. Wang, X. Zhang, J. Zhang, and W. Gruver, "Gripping force sensory feedback for a myoelectrically controlled forearm prosthesis," in *Proc. IEEE Int. Conf. Syst., Man Cybern. Intell. Syst. 21st Century*, vol. 1, Oct. 1995, pp. 501–504.

[12] M. Isaković *et al.*, "Electrotactile feedback improves performance and facilitates learning in the routine grasping task," *Eur. J. Transl. Myol.*, vol. 26, no. 3, Jun. 2016.

[13] A. Chatterjee, P. Chaubey, J. Martin, and N. V. Thakor, "Testing a prosthetic haptic feedback simulator with an interactive force matching task," *J. Prosthetics Orthotics*, vol. 20, no. 2, pp. 27–34, 2008.

[14] I. Saunders and S. Vijayakumar, "The role of feed-forward and feedback processes for closed-loop prosthesis control," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, p. 60, Oct. 2011.

[15] C. Antfolk *et al.*, "Transfer of tactile input from an artificial hand to the forearm: Experiments in amputees and able-bodied volunteers," *Disab. Rehabil., Assistive Technol.*, vol. 8, no. 3, pp. 249–254, May 2013.

[16] R. Christiansen, J. L. Contreras-Vidal, R. B. Gillespie, P. A. Shewokis, and M. K. O'Malley, "Vibrotactile feedback of pose error enhances myoelectric control of a prosthetic hand," in *Proc. World Haptics Conf. (WHC)*, Apr. 2013, pp. 531–536.

[17] A. Ninu, S. Dosen, S. Muceli, F. Rattay, H. Dietl, and D. Farina, "Closed-loop control of grasping with a myoelectric hand prosthesis: Which are the relevant feedback variables for force control?" *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 5, pp. 1041–1052, Sep. 2014.

[18] L. F. Engels, A. W. Shehata, E. J. Scheme, J. W. Sensinger, and C. Cipriani, "When less is more–discrete tactile feedback dominates continuous audio biofeedback in the integrated percept while controlling a myoelectric prosthetic hand," *Frontiers Neurosci.*, vol. 13, p. 578, Jun. 2019.

[19] J. Gonzalez, H. Suzuki, N. Natsumi, M. Sekine, and W. Yu, "Auditory display as a prosthetic hand sensory feedback for reaching and grasping tasks," in *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, Aug. 2012, pp. 1789–1792.

[20] A. Blank, A. M. Okamura, and K. J. Kuchenbecker, "Identifying the role of proprioception in upper-limb prosthesis control: Studies on targeted motion," *ACM Trans. Appl. Perception*, vol. 7, no. 3, p. 15, 2010.

[21] M. A. Schiefer, E. L. Graczyk, S. M. Sidik, D. W. Tan, and D. J. Tyler, "Artificial tactile and proprioceptive feedback improves performance and confidence on object identification tasks," *PLoS ONE*, vol. 13, no. 12, Dec. 2018, Art. no. e0207659.

[22] M. Aboseria, F. Clemente, L. F. Engels, and C. Cipriani, "Discrete vibro-tactile feedback prevents object slippage in hand prostheses more intuitively than other modalities," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 8, pp. 1577–1584, Aug. 2018.

[23] R. B. Gillespie *et al.*, "Toward improved sensorimotor integration and learning using upper-limb prosthetic devices," in *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol.*, Aug. 2010, pp. 5077–5080.

[24] H. J. B. Witteveen, F. Luft, J. S. Rietman, and P. H. Veltink, "Stiffness feedback for myoelectric forearm prostheses using vibrotactile stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 1, pp. 53–61, Jan. 2014.

[25] N. Thomas, G. Ung, C. McGarvey, and J. D. Brown, "Comparison of vibrotactile and joint-torque feedback in a myoelectric upper-limb prosthesis," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, Dec. 2019.

[26] F. Clemente, M. D'Alonzo, M. Controzzi, B. B. Edin, and C. Cipriani, "Non-invasive, temporally discrete feedback of object contact and release improves grasp control of closed-loop myoelectric transradial prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 24, no. 12, pp. 1314–1322, Dec. 2015.

[27] J. Cohen, *Statistical Power Analysis for the Behavioral Sciences*. New York, NY, USA: Academic, 2013.

[28] F. Cordella *et al.*, "Literature review on needs of upper limb prosthesis users," *Frontiers Neurosci.*, vol. 10, p. 209, May 2016.

[29] G. D. Pino, E. Guglielmelli, and P. M. Rossini, "Neuroplasticity in amputees: Main implications on bidirectional interfacing of cybernetic hand prostheses," *Progr. Neurobiol.*, vol. 88, no. 2, pp. 114–126, Jun. 2009.

[30] H. Flor and M. Diers, "Sensorimotor training and cortical reorganization," *NeuroRehabilitation*, vol. 25, no. 1, pp. 19–27, Aug. 2009.

[31] M. Lotze, W. Grodd, N. Birbaumer, M. Erb, E. Huse, and H. Flor, "Does use of a myoelectric prosthesis prevent cortical reorganization and phantom limb pain?" *Nature Neurosci.*, vol. 2, pp. 501–502, Jun. 1999.