Vibrotactile Sensory Substitution for Electromyographic Control of Object Manipulation

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***Abstract*—It has been shown that incorporating augmentative vibrotactile feedback can improve performance of a virtual object manipulation task using finger movement. Vibrotactile sensory substitution for prosthetic applications, however, will necessarily not involve actual finger movement for control. Here we study the utility of such feedback when using myoelectric (EMG) signals for control, and demonstrate task improvement and learning for a force-motion task in a virtual environment. Using vibrotactile feedback, a group of unimpaired participants ($N = 10$) were able to increase performance in a single session. We go on to study the feasibility of this method for two prosthetic hand users, one of whom had targeted muscle reinnervation allowing the augmentative feedback to be perceived as if it were on the absent hand.**

***Index Terms*—Electromyographic-controlled prosthesis, haptic display, sensory feedback, upper extremity prosthesis.**

I. INTRODUCTION

WHILE prosthetic hands typically provide little direct sensory feedback of forces encountered in the environment, closed-loop prosthetic control can improve performance and help compensate for control uncertainty [1]. Force feedback in particular has been identified as a priority for increasing the usability of myoelectric prostheses [2]. In addition, there is evidence that incorporation of haptic feedback in prostheses can result in increased self-attribution of the body part [3]. Prosthetic users currently must rely on indirect means of determining how much force is being applied to an object, such as vision or simply experience and practice. The use of visual feedback for estimating manipulation forces requires constant attention, suffers from occlusion, and is unreliable for inelastic objects or in the dark. While some users might be able to use the vibrations caused by the motors of their prosthetic hand as a cue, intentionally providing augmentative sensory feedback has the potential to improve prosthesis control by allowing the user to directly sense task-relevant forces.

It is technologically possible to sense fingertip forces in real time, with many groups successfully integrating these technologies into prosthetic hands, e.g., the cybernetic hand [4], [5]. These capabilities have yet to be widely adopted in commercial products, in part because it is not yet clear how to provide feedback to users so they can optimally integrate the information for sensorimotor control. Though there is progress for a variety of feedback paradigms ranging in invasiveness and expressiveness [6], there is an immediate role for noninvasive methods. We suggest that vibrotactile stimulation is cheap, noninvasive, and could be easily implemented into existing prosthetic technologies as augmentative sensory feedback of fingertip forces.

Previous applications of vibrotactile feedback for object manipulation have produced inconsistent outcomes, but results have been difficult to compare due to the studies having different types of stimulation or different tasks and performance measures. While some studies have observed promising results for vibrotactile stimulation [7], [8], others have not [9], but the studies have varied widely in task and stimulation details. We have previously developed an experimental paradigm that specifically concerns object manipulation rather than grasping, and has allowed us to study the effects of vibrotactile feedback with larger numbers of unimpaired users. These studies establish the usefulness of amplitude-modulated vibrotactile feedback for a motion-force manipulation task. Users could improve performance using their intact finger to control the virtual prosthetic, with improved proficiency in the presence of vibrotactile force feedback [10], [11].

Here we extend our paradigm to electromyographic (EMG) control of the same virtual task, demonstrating the utility of augmentative vibrotactile feedback for EMG control in ten healthy individuals. Under EMG control, the activation of groups of muscles is used as a control input to the virtual prosthesis instead of finger movements, and is therefore usable by individuals with limb loss. We present pilot data of two participants with limb loss successfully using the system. The focus of this study is to characterize the improvement in performance for EMG control when given augmentative vibrotactile feedback. We hypothesized that users would experience performance gains for the EMG control task with feedback, as was previously observed for participants using actual finger movement. We expected the

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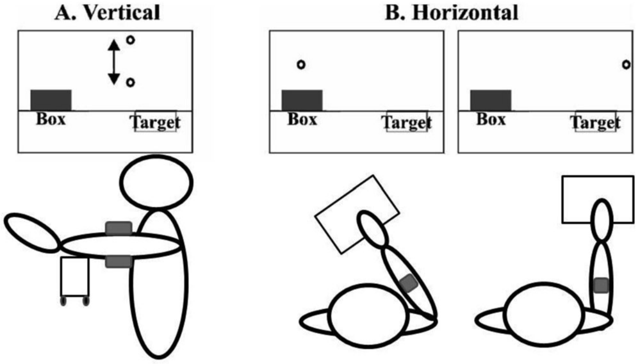

Fig. 1. Experimental apparatus. The virtual environment simulates a box sliding across a flat table. Panel (A) depicts participant control of the vertical location of a virtual finger in the virtual environment. The two squares on the forearm indicate EMG electrode placement. Panel (B) depicts the side-to-side motion of the arm or residual limb necessary to move the finger horizontally in the virtual workspace. The square on the upper segment of the arm indicates the placement of the magnetometer sensor of arm angle. Unimpaired participants rested their wrist on a small cart to prevent fatigue.

augmentative feedback to allow more dexterity, manifesting as faster and more complete task performance.

II. VIRTUAL MYOELECTRIC PROSTHESIS MANIPULATION TASK

A. *Virtual Environment*

Participants interacted with a virtual environment through EMG and horizontal arm motion. Their goal was to apply appropriate normal force to a virtual object to allow for its translation, and to drag it horizontally to a target without breaking it. Task success was measured by speed of translation and distance traveled before breaking the object. This task was specifically chosen due to the known difficulties of prosthetic hand users with appropriately applying normal force to delicate objects, such as picking up and manipulating a disposable plastic cup, and is the same task used in our previous studies using vibrotactile feedback to augment finger control of a virtual object manipulation task [10], [11]. This is more demanding than a gripping task in that it requires the precise application of forces while moving during contact.

A video monitor was positioned at 45° toward the participant, with a mirror placed between the virtual environment and the monitor to permit reflection of images from the monitor to the user. The virtual environment was programmed in C++, with graphics driven by OpenGL. Participants sat in front of the projection system with their arm or residual limb extended parallel to the floor. Control participants used a rolling support under their wrist to allow horizontal movement across the workspace without experiencing fatigue from holding the weight of their arm (see Fig. 1, lower).

During the task, users received visual feedback consisting of a real-time depiction of the location of the virtual "finger" in the virtual environment, and the current position of the box (Fig. 1, upper). Finger position was indicated by a small sphere. The finger location was occluded during penetration of the box, and deformations of the box were not shown visually. This approximates the real-life visual feedback of task performance available to users of prosthetic limbs during object manipulation. Many delicate objects do not visibly distort under force, e.g., an egg. Vibrotactile stimulation, when activated, provided a perception of the force resulting from the stiffness of the box as the finger moved vertically into it.

During interaction, one of two possible virtual objects was located at the left end of the workspace (Fig. 1, upper). The upper surface of these objects (referred to as boxes) had distinct stiffness characteristics. Box stiffness functions were scaled versions of a fit to the force–displacement curve acquired empirically by pushing on a disposable plastic cup. Box stiffness profiles were indicated by box color, red or blue, but participants were not told what the colors meant, just that the boxes could have different properties. The stiffness characteristics of each box were

$$F_{\text{blue}} = \left\{ \begin{array}{ll} (0.34\ \text{N/cm})x, & \text{if } x < 1.7\ \text{cm} \\ (4.65\ \text{N/cm}^2)x^2 - (14.33\ \text{N/cm})x & \text{if } x < 1.7\ \text{cm} \\ \quad + 11.55, & \end{array} \right\}$$

$$F_{\text{red}} = \left\{ \begin{array}{ll} (0.68\ \text{N/cm})x, & \text{if } x < 1.7\ \text{cm} \\ (9.59\ \text{N/cm}^2)x^2 - (30.67\ \text{N/cm})x & \text{if } x < 1.7\ \text{cm} \\ \quad + 25.56, & \end{array} \right\}$$

where x is the displacement of the finger into the box in the normal direction in centimeters, and F_{blue} and F_{red} are the virtual normal forces of the blue and red boxes, respectively. In short, the boxes mimic the scaled stiffness of a plastic cup, with the blue box less stiff than the red.

The virtual normal force required to overcome friction to translate each box, F_{move}, was 1.2 times the virtual force at the displacement of 1.7 cm. The virtual force threshold to "break" each box, F_{break}, was defined as 0.75 N greater than F_{move}. Virtual normal force applied to the box between F_{move} and F_{break} allowed the participant to slide the object to a target located 30 cm to the right of the workspace. The difference in stiffness between the two boxes resulted in different allowable displacements of the virtual finger during motion (1.6 mm for the red box and 2.7 mm for the blue box) in the direction of the virtual normal force. Thus, for most users, the red box was more difficult to move successfully.

B. *Electromyographic Control for Vertical Finger Movement*

Control of the finger in the vertical direction of the virtual environment was determined via two surface EMG signals acquired at 1000 Hz. The y-direction velocity was defined as the difference between the root-mean-square (RMS) of the two signals, multiplied by a subject-specific scaling factor. EMG signals were differenced such that greater muscle activity for wrist extension caused an increase in the upward velocity of the finger in the virtual environment, and greater wrist flexion caused an increase in the downward velocity. The RMS was calculated in 300-mS (300 samples) windows with 90% overlap. The scaling factor was chosen to account for individual variations in voltage measured at the skin surface due to differences in muscle activation, electrode contact, and subdermal fat. A brief calibration process using maximum voluntary wrist flexion and extension is used to set the scaling factors, as in [12]:

$$s = \frac{2.1}{2}(\max(\mathrm{EMG}_{\mathrm{extention}}) + \max(\mathrm{EMG}_{\mathrm{flexion}})).$$

The scalar value 2.1 was used for all subjects, and has been developed empirically over the course of preliminary experimentation before the present study to provide a virtual finger velocity suitable for completing the box-sliding task.

Surface EMG was recorded using a 2-channel Bagnoli system (Delsys Inc., Boston, MA, USA) with two Delsys 2.1 differential surface electrodes placed on the forearm parallel to the underlying muscle fibers of the extensor carpi radialis (wrist extensor) and palmaris longus (wrist flexor). Each electrode consisted of two 10-mm silver bars with an inter-bar distance of 10 mm. The arm of each participant was prepared for electrode placement by cleaning the skin surface with an alcohol pad and "peeling" (exfoliation) with tape to reduce electrode–skin impedance, noise, dc voltages, and motion artifacts. A ground electrode was placed on the superior aspect of the participant's right shoulder. EMG signals were preamplified and filtered using the Delsys Bagnoli system set to a gain of 1000, with a band-pass filter having roll-off frequencies of 20 and 450 Hz.

C. Magnetometer Control for Horizontal Finger Movement

The horizontal position of the virtual finger was controlled by the angle of the upper arm of the user in the axial plane. The user sat upright with shoulder in flexion, with adduction and abduction of the shoulder changing the shoulder angle in the axial plane. This was measured using a Honeywell HMC6352 electronic compass, providing a heading measurement at 20 Hz with 0.5° resolution. This magnetometer was strapped comfortably onto the upper arm of the participant. The compass data were read using a Microchip PIC18F14K22 microcontroller over an I2C bus, and translated into a 10-bit pulse-width-modulated signal with duty cycle proportional to the measured angle, suitable for reading by the control computer using a data acquisition card. Electronic compass angle was averaged over 30-mS windows and scaled such that real-world arm angles spanning the comfortable range correspond to the horizontal width of the virtual environment. As illustrated in Fig. 1, when the user's arm was extended straight forward, the virtual finger was situated at the rightmost side of the virtual environment. Additional low-pass horizontal dynamics of the virtual environment were achieved by filtering the input signal via a moving average filter with cutoff frequency of 26 Hz.

D. Vibrotactile Feedback

Vibrotactile stimulation at 250 Hz was provided using a C2 tactor (Engineering Acoustics, Inc.) with a 0.3-in diameter skin contactor. The tactor was secured to the right lateral upper arm and with an elasticized cloth bandage. A 250-Hz carrier frequency was used since human glabrous skin [13] has been shown to be maximally sensitive to vibrotactile stimulation at this frequency [14], [15]. During interaction with the virtual environment, increases in virtual normal force were translated linearly into increases in the amplitude of continuous vibrotactile stimulation. The maximum amplitude of vibration, corresponding to breaking the box, was approximately 400 μm. Participants wore noise-canceling headphones (Bose, Framingham, MA, USA) during experimentation to present the stimuli for the cognitive task, and to provide low-level masking noise. Vibrotactile feedback was provided at 250 Hz, which is in the range of human hearing. The masking noise and noise-canceling headphones were used to ensure that participants were not using any auditory feedback from the tactor to complete the motor task.

III. EXPERIMENTAL DESIGN AND ANALYSIS

A. Participants

Control participants were 10 adults (all right-handed; 6 male, 4 female; mean age = 22.8 years, SD = 1.9 years). Participants with limb loss, referred to here as P1 and P2, were two male adults of ages 23 and 53, respectively. Controls reported normal hand function, with no complaints related to their hands. Participant P1 had amputation of the left arm above the elbow, and targeted reinnervation of distal nerves to the residual limb [16]. Participant P2 had amputation of the right arm below elbow and a Medtronics implant for phantom limb clench implanted in right pectoral, innervating the right bicep. Both P1 and P2 had experience operating myoelectric prostheses. Informed consent was obtained from all participants in compliance with the Institutional Review Board of the University of Washington.

B. Cognitive Load

An auditory 2-back test [19] was used as a simultaneous cognitive load during the virtual task. Participants were presented with a 1-Hz auditory stream of 16 random digits via their headphones. During this stream, they were instructed to respond verbally to identify any numbers repeated with only one intervening number. To ensure that participants understood this task, participants first practiced 20 sets of this task without the motor task prior to experimentation. During interaction with the virtual environment, participants were asked to complete the cognitive task while simultaneously completing the motor task. Completion of the entire 16 digits of the cognitive task was not achieved if the box was broken in less than 16 s.

C. Experimental Protocol

Over approximately 2 h (including breaks), each participant completed 80 trials of interaction with the system, organized into five blocks of 16 trials each. Each block was randomized within block by feedback condition (vision alone or vision + vibrotactile stimulation), by box (blue, red), and by cognitive load (ON, OFF). Participants were encouraged to take breaks between any trials to avoid fatigue. Participants generally took one or two breaks during the experiment.

D. Analysis

Performance variables were box displacement (distance toward the target that the box translated during the trial) and

TABLE I
REPEATED MEASURES ANOVA OF BOX DISPLACEMENT FOR CONTROL SUBJECTS. DEGREES OF FREEDOM (DOF) F-STATISTIC, AND P-VALUE

Factor	**DF**	**F**	***P***
Box	1	16.65	<0.001
Cognitive Task	1	0.30	0.583
Feedback	1	32.47	<0.001
Block	4	17.14	<0.001
Cognitive Task × Feedback	1	2.52	0.113
Cognitive Task × Block	4	0.92	0.452
Feedback × Block	4	4.09	0.003

TABLE II
REPEATED MEASURES ANOVA OF BOX VELOCITY FOR CONTROL SUBJECTS

Factor	**DF**	**F**	***P***
Box	1	20.18	<0.001
Cognitive Task	1	4.81	0.029
Feedback	1	16.66	<0.001
Block	4	14.26	<0.001
Cognitive Task × Feedback	1	1.16	0.282
Cognitive Task × Block	4	0.94	0.438
Feedback × Block	4	3.90	0.004

average box velocity (box displacement normalized by trial duration). Each trial, then, provides a single measurement of each performance variable under a particular combination of the factors. Data analysis to determine the performance variables for each trial was performed using MATLAB (Mathworks, Natick, MA, USA), and statistical analysis was performed using Minitab Statistical Software (Minitab Inc., State College, PA, USA). For data from unimpaired participants, a 4 factor repeated measures analysis of variance (ANOVA) was performed to assess the effects of feedback (vision or vision plus vibrotactile), cognitive task (ON or OFF), presentation order (block), and box (red or blue), as well as the interactions of feedback × block, cognitive task × feedback, and block × cognitive task on box displacement and velocity. Post hoc two-sided Tukey's simultaneous tests were used when appropriate. All statistical analyses were performed using an alpha level of 0.05 for significance. Due to the small sample and disparate subject background, statistical analysis was not performed on the data from the two participants with limb loss.

E. Experimental Differences for Participants With Limb Loss

The virtual environment for participants with limb loss was identical to the control protocol, and the virtual myoelectric prosthesis was operated in the same way, with adjustments for EMG sensor and tactor placement. EMG placement was chosen to match the locations used for their myoelectric prostheses, which in both participants P1 and P2 were extensor and flexor muscles of the distal segment of the residual limb. For P1, this was left residual biceps and triceps brachii and the C2 tactor was placed at the target site of the targeted muscle reinnervation (TMR), on left lateral upper arm. Tactile stimulation of this area of skin corresponded to the sensation of touch along the thumb and index finger of the absent hand, according to self-report. For P2, EMG sensor placement was on right residual extensor carpi radialis and palmaris longus muscles, with tactor placement on right lateral upper arm.

IV. RESULTS

A. Control Participants

Table I shows the 4-factor ANOVA significant effects for displacement, and Table II shows those for average velocity. For displacement, significant effects were found for box, block, and feedback. Velocity showed significant effects for all four factors. Results from Tukey simultaneous tests show that trials in which participants were given visual and vibrotactile feedback had significantly increased displacement and velocity than trials using vision alone. Trials using the blue box (less stiff, therefore greater finger penetration can occur before breaking) had significantly increased displacement and velocity than trials with the red box. The trials with a simultaneous cognitive task had significantly decreased velocity. Performance as a function of block is illustrated in Fig. 2 and generally increased throughout the experiment, indicating a significant effect of learning within the session. Box displacement and velocity are both improved when the users have access to vibrotactile force feedback (Fig. 2 gray versus black circles; bars indicate 95% confidence intervals).

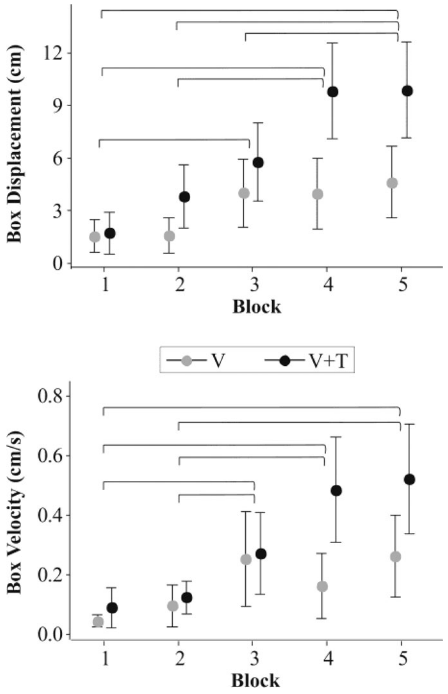

Fig. 2. Control subjects increase box displacement (above) and velocity (below) as the session progressed. Statistical significance between blocks is denoted by brackets. Means denoted by circles for vision only (gray) and vision plus vibrotactile feedback (black). Bars indicate 95% confidence interval. The performance advantage for vibrotactile feedback is especially apparent in blocks 4 and 5, and Tukey simultaneous tests show a statistically significant improvement for vibrotactile feedback relative to vision alone.

No significant interactions of cognitive task × feedback or block × cognitive task were observed, but both displacement and velocity show a significant interaction of feedback × block. This

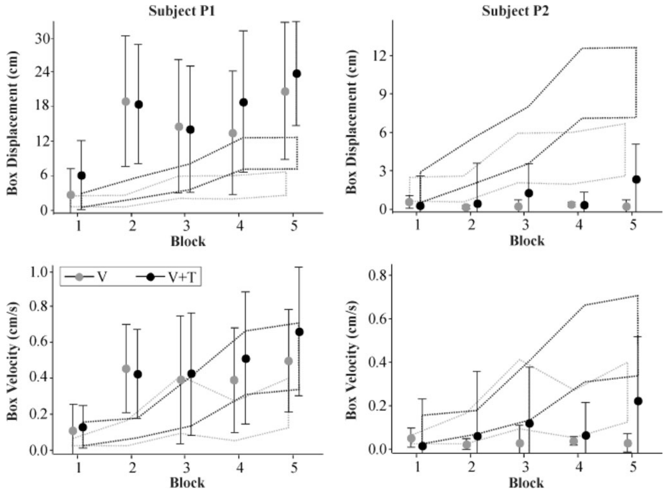

Fig. 3. Results for participants with limb loss using vision only (gray) and vision plus vibrotactile feedback (black). The left column depicts performance for participant P1, the right depicts performance for P2. The 95% confidence intervals for the means using vision alone (gray) and vision plus vibrotactile (black) for the controls are plotted as dotted lines behind the points for reference. The confidence intervals for P1 and P2 incorporate only the trials for a single user for that block, while those for the unimpaired participants incorporate the trials of ten users. Note the difference in y-axis scale for the left column (P1) and the right (P2).

indicates that the improvement due to vision plus vibrotactile feedback relative to vision-only increased with practice during the session.

B. Participants With Limb Loss

Performance measures were the same as the control group, but without statistical significance analysis. As presented in Fig. 3, left column, participant P1 learned to improve both displacement and velocity, with box displacement already exceeding the performance of the control group by block 2. Performance is clearly improved from initial performance by both measures, and improvement due to the presence of vibrotactile feedback is present but less pronounced.

Participant P2 had a more difficult time with the task, and showed a stronger difference between visual-only and visual + vibrotactile feedback conditions. The presence of vibrotactile feedback allowed P2 to perform with more success, resulting in slightly greater means and much greater variance as shown in Fig. 3, right column. The effect of learning over the course of the session was less clear than for P1, but over the session learning was exhibited.

Results in general for the participants with limb loss resemble those for the control group, with performance being improved with vibrotactile feedback relative to with vision only and worse for the more difficult box.

V. DISCUSSION

A. Vibrotactile Feedback Improves Performance

For unimpaired participants, the presence of vibrotactile feedback improved both the total displacement achieved and the velocity of box movement relative to using visual feedback alone. For one participant with limb loss, P1, vibrotactile appears to have slightly increased performance, and for the other, P2, the presence of vibrotactile feedback appears to have salvaged the task from being near-impossible. Displacement and velocity were improved, and P2 was able to move the box to the end of the possible range several times. This establishes the utility of the sensory replacement method used for this experiment and suggests future study into the feasibility of integrating vibrotactile force feedback into realistic contexts for prosthetics.

B. Effects of the Cognitive Task

Feedback having low cognitive demand would not be affected by a distractor task. Though some participants reported having difficulty with the cognitive task in general, its effect on task performance was not as pronounced as the effects of which box (red—more stiff, blue—less stiff) or the feedback paradigm. We observed no statistically significant effect of the cognitive load task on box displacement, but we did find significantly reduced box velocity, perhaps indicating that participants employ a slower strategy to compensate for the distraction. Previous experiments using this vibrotactile force feedback, with finger movement control instead of EMG, have shown either lower velocities [10] or a decrease in both displacement and velocity in the presence of the distracting cognitive task [17].

C. Benefit of Feedback Increases With Training

Performance improved with practice for both feedback modalities over approximately 2 h of the experiment. This is borne out by the significant effects of block on both displacement (Table I) and velocity (Table II). Therefore, the effect of learning on task performance was significant. As discussed in Section V-A, the presence of feedback was also significant. Interestingly, however, it appears that the benefit of feedback increased as learning progressed throughout the session. There was a significant interaction for feedback × block for both displacement and velocity, indicating that participants learned to use the vibrotactile feedback with increasing skill as the session progressed. By the end of the session there was a marked difference in performance with and without the vibrotactile force feedback, with the limit of single-session learning appearing earlier for vision-only. Previous work suggests that continued practice over multiple sessions would continue this trend [17]. Informal post-experiment reports of subjective task difficulty indicated that participants preferred the vision plus vibrotactile feedback.

D. Evidence of Utility of Vibrotactile Feedback

Previous application of vibrotactile feedback to EMG control of an anthropomorphic hand by Cipriani *et al*. showed conflicting results [18]. In their study, 14 young healthy participants learned to control a robotic arm with vibrotactile stimulation provided using a quasi-linear mapping of prosthetic hand closure force to the frequency of vibrotactile stimulation (ranging from 10 to 240 Hz). This previous work did not find objective improvement in participants' ability to perform the study tasks, but found statistically significant benefit of feedback via the subjective opinions of the users. Our results differ in that we did find statistically significantly improved objective motor performance when participants utilized the vibrotactile stimulation, which may be a result of the different methodologies employed by the previous study. In particular, the work of Cipriani *et al*. required multidimensional control using more complicated EMG control schemes than our current work. In addition, the mapping between force and vibrotactile stimulation differed from our approach, and we have shown previously that this mapping can greatly affect usability [10].

E. Qualitatively Different Strategies

Participants employed two kinds of qualitatively different strategies for sliding the boxes. Some preferred to lower the virtual finger onto the box and deliberately slide it across the workspace, constantly adjusting force to appropriate levels, while rarely stopping and repositioning the finger. This strategy was more difficult to employ for vision-only feedback, because box movement was a cue for when adequate force was being applied, but there was no warning before box breakage. Others focused on initially achieving just the minimum force necessary for box movement, then rapidly moving the box in a tapping motion while lifting the finger from the box. Therefore, the virtual finger moved in oval trajectories, intermittently moving the box, lifting off, and circling back. This tapping strategy was employed by one of the participants with limb loss, P1, resulting in very high performance compared to the control subjects. Participant P1's performance for the final block more than doubles the average displacement of the control group, while also achieving a slightly faster velocity (see Figs. 2 and 3).

F. Participants With Limb Loss and Limitations of the Pilot Study

The performance of the users with limb loss suggests that further experiments are warranted, but it remains to be seen if the increased performance observed for unimpaired subjects will necessarily translate into gains for users of myoelectric prostheses. While differences in motivation, experience with similar tasks, and constraints for EMG sensor placement make it difficult to directly compare outcomes, this pilot study shows that the use of vibrotactile force feedback has potential for sensory feedback for prosthetic applications.

Fig. 3 shows the performance of participants P1 and P2 juxtaposed with the combined performance of the control participants. As in Fig. 2, the gray and black circles indicate mean performance with and without vibrotactile feedback, with the bars indicating the 95% confidence interval. These comparisons show that while both participants improved performance, the learning and performance ceiling of P1 was faster and higher than the other participants, but with a less profound impact of vibrotactile feedback. From this pilot data, it is difficult to determine what factors contribute to this effect. It may indicate that vibrotactile feedback would not increase the asymptotically learned performance of P1, or it could simply be a consequence of details like electrode and tactor placement. Though previous work has shown that differences in tactor placement site are outweighed by single-session training effects [11], it is still an active area of research to understand how this interacts with TMR nerve transfer surgery. For P1 the target site of TMR was on the residual limb, relatively near the muscle group chosen for EMG sensor placement.

As shown in Fig. 3, right column, performance for participant P2 was poor compared to the control participants, though somewhat improved by the vibrotactile feedback. Without feedback, P2 was mostly unable to move the box without breaking it, with a slight learning effect through the session. With vibrotactile feedback, however, P2 was better able to move the boxes, though still with great variability. Unique factors potentially contributing to this performance include age difference (P2: 53 years, P1: 23 years, controls: mean 22.8, SD 1.9 years) and the presence of an implanted device for ameliorating phantom limb clench innervating right bicep brachii. P2 reported that the vibrotactile feedback subjectively made the task less difficult, but that the EMG sensor location chosen for the experiment was not a good choice in retrospect due to fatigue.

One limitation of this experimental paradigm is that performance is evaluated on the same task being learned. In the future, a more general exploration of the performance limitations of augmentative vibrotactile feedback could include novel force-sensitive task constraints, analysis of the sensitivity of this paradigm to sensor and tactor placement, and a full study on a larger number of users with limb loss.

VI. CONCLUSION

Using a virtual myoelectric prosthesis, we characterized the effects of visual and visual + vibrotactile haptic feedback on the object manipulation task of sliding a fragile box across a surface. We found that augmentative vibrotactile feedback can aid force-sensitive manipulation, and that this improvement increases with training. We demonstrate the applicability of the technique for two users with limb loss. These results are promising for future application for force-feedback prosthetics.

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