Vibrotactile Feedback of Pose Error Enhances Myoelectric Control of a Prosthetic Hand

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ABSTRACT

Advanced prosthetic hands offer the promise of great dexterity; however, myoelectric control techniques, successful with low degree-of-freedom prosthetics, are often set aside by amputees due to the lack of important sensations of touch and effort experienced in the interaction between prosthetic hand and task. In this paper, we explore the efficacy of various modalities of feedback (visual, tactile, visual and tactile, and none) conveying proprioceptive information, specifically the error in joint angles between a desired and actual pose of a virtual prosthetic hand. Our analysis of performance in achieving and maintaining a desired prosthetic hand pose indicates a significant effect of feedback condition, with visual and visual+tactile outperforming tactile alone and a no-feedback condition. Further, the combination of tactile and visual feedback does not seem to have significant drawbacks over visual feedback alone. For tasks that rely on proprioception in the absence of visual feedback, or when attention must be focused elsewhere, we see a performance benefit to the inclusion of tactile cueing, with no lags in reaction times or requirements for increased effort measured by muscle activation.

KEYWORDS: Human haptics, tactile devices and display, myoelectric prosthesis, virtual environment.

INDEX TERMS: Tactile and hand-based interfaces; External interfaces for robotics; Human performance: Medical applications

1 INTRODUCTION

Existing neuroprosthetic devices rely heavily on visual feedback to facilitate control of the prosthetic limb. The lack of functionality and lack of sensory feedback from current commercially available upper limb prostheses are significant factors in their non-use, especially for advanced myoelectric prostheses [1]. The availability of sensory feedback may be a key factor in functional outcomes for prosthesis users. Dudkiewicz et al. found that, for their surveyed population of amputees, above-elbow amputees with dominant hand amputation who used body-powered prostheses achieved the best functional outcomes [1], potentially due to access to kinesthetic sensory feedback transmitted by the cable mechanisms that actuate such prostheses.

Because of the lack of sensory feedback in advanced prosthetic devices such as those that are myoelectrically controlled, rather than body-powered, amputees rely on constant visual monitoring to substitute for the missing tactile and kinesthetic cues experienced by able-bodied individuals. It is well understood that sensory feedback from electronic touch and force sensors is required for fine control, especially in contact tasks and tasks involving discrimination of mechanical properties—where vision often breaks down [2][3][4][5]. So while prosthetic devices featuring sensing and multiple degrees of freedom of actuation have become more widely available, the means to effectively control and interface these new devices to the human nervous system still elude us.

Advanced techniques to directly couple a prosthetic interface to peripheral or central nerves via implantable electrodes have been proposed and studied; however, critical biocompatibility issues, surgical risks, and signal degradation impede translating these successes to an interface for chronic use [6]. Surgical techniques, such as targeted reinnervation, which involves the surgical re-implantation of nerve bundles to alternate muscle sites, appear to hold more immediate promise for chronic use. Targeted reinnervation, however, involves a substantial surgical intervention and seems to be indicated only for a small population of amputees [7]. In the meantime, less invasive technologies that offer control and sensory feedback must be exploited for interfacing between the amputee and the prosthetic device [8].

In this paper, we explore the use of tactile feedback to relay proprioceptive information regarding prosthetic hand pose to the same limb that generates the command signal to control the pose. Our goal is to achieve what is known in the prosthetics community as Extended Physiological Proprioception (EPP). When a prosthesis interface feeds back the mechanical response from the environment to the muscle that activated that response, then the brain seems to adopt that prosthesis as an extension of the body [8]. We hypothesize that feedback conveying proprioceptive information regarding the prosthetic limb can extend physiological perception of the prosthetic limb just as feedback regarding environmental response does. Sensory feedback is required for the user to build new internal models of the combined body and interface. Haptic feedback and proprioception in particular are critical in the development of internal models of the relationship between motor commands and their effects, and for obtaining real-time estimates of body states used for planning and control [9][10][11][12].

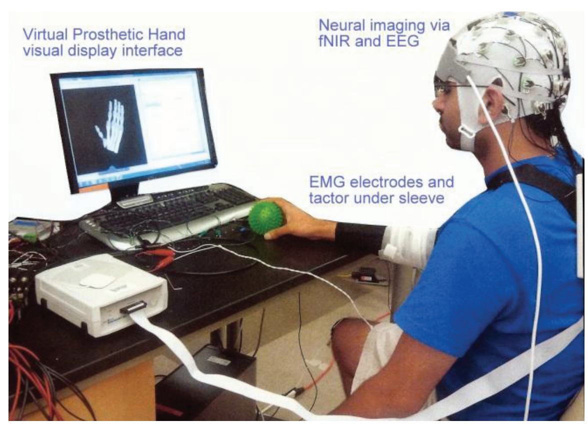

Figure 1. Experimental setup for analyzing effect of feedback modality on control of virtual prosthetic hand pose.

Vibrotactile devices offer numerous appealing attributes for sensory feedback in myoelectric prostheses, including their ease of application and non-invasive nature [13]. As a result, use of vibrotactile feedback for conveying prosthetic hand opening has

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been reported in the literature [14][15]. Witteveen et al. showed that vibrotactile feedback provided via an array of tactors was shown to improve control of a virtual object grasping task compared to a no feedback condition; however, control of the hand pose took place through the manipulation of a mouse scroll wheel [14]. Sergi et al. explored the use of vibrotactile feedback to guide forearm movements. In their work, visual cues indicating current pose and desired pose of the forearm were compared to vibrotactile cues indicating both the direction and magnitude of angular error between current and desired orientations, along with a combined visuotactile mode [15]. Their findings showed that the combined modality was more accurate than visual feedback alone for some poses, while tactile feedback alone was not significantly less effective than visual or visuotactile feedback across all poses.

Despite these studies which explored the utility of tactile feedback for conveying pose information, there are no reported studies that directly compare accurate control of hand pose for multiple feedback modalities in the presence of vibrotactile feedback for a myoelectrically controlled prosthesis scenario. In this paper, we report the results of a study comparing various modes of feedback (visual, tactile, and combined) regarding prosthetic hand pose to a user outfitted with sensors to enable myoelectric control of a virtual prosthetic hand.

2 Methods

To test the performance and quality of the tactile feedback combined with myoelectric control of a prosthesis, an experiment was designed in which non-amputees could interact with a virtual hand prosthesis to complete a simple pose-matching task (see Fig. 1). While the experimental subjects used were non-amputees, the sensors and actuators attached to the subjects were placed in a configuration that could be duplicated exactly on a transradial amputee, thus granting subjects no advantage over the intended users of prosthetic devices. While this experiment was undertaken with electroencephalograpy (EEG) and functional near-infrared spectroscopy (fNIR) sensors, these systems were largely independent from the tactile feedback data, and will not be discussed in detail.

2.1 Subjects

The experiment involved a total of seventeen non-amputee subjects (age 23 ± 5 yrs., 10 male, 4 female). All participants listened to a scripted description of the experiment objectives and methods prior to the start of the protocol, and provided informed consent. Data were discarded for three subjects who experienced prolonged equipment failure during evaluation. Thus, only data for 14 of the 17 subjects were analyzed. The study protocol was approved by the Institutional Review Boards of all collaborating institutions.

2.2 Experimental Setup

During the experiment, electromyography (EMG) signals were used to control a virtual prosthetic hand, and visual feedback (via computer monitor) and vibrotactile feedback (via a single tactor) were provided to the subject to indicate the difference in joint angles between the actual and desired hand configuration. Brain activity was measured using a 32 channel scalp electro-encephalography (EEG) with a BrainAmps DC amplifier and actiCAP active electrodes (Brain Products GmBH) and a 16-channel fNIR (fNIR Imager 1000) system (Fig. 1). Brain imaging results will not be presented in this paper. Visual and auditory signals were not blocked during any of the trials.

2.2.1 Control

Control of the virtual prosthesis was achieved through the use of Electromyography (EMG). Three two-inch square Austin Medical Equipment Electrotherapy Electrodes (item K220) were placed on the forearm of the subject in order to capture the activation of the Flexor carpi radialis muscle, which is activated by hand flexion. This muscle is among the nearest muscles to the hand that an amputee will have intact, and it provides a large surface for attaching electrodes. This makes the muscle a good choice for a clean EMG signal, despite the relatively weak signal obtained compared to other nearby muscles. Of the three electrodes, two are attached to skin above these muscles as a differential pair, with one electrode attached near where the muscles connect to the elbow, and the other electrode attached slightly farther down the forearm on the bulge of the muscle, as shown in Fig. 2. An athletic sleeve was pulled over the electrodes and compression tape was wound around the sleeve in order to prevent the electrodes from peeling off the skin for the duration of the experiment. The third electrode was connected to the kneecap, and was used as a reference ground signal. The kneecap works well as a reference ground signal because it is very bony, and has a large exposed surface.

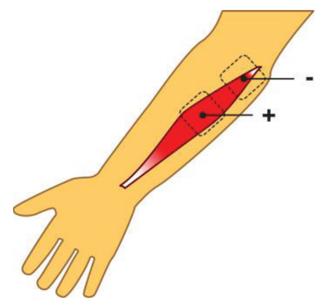

Figure 2. The Flexor carpi radialis muscle, with electrode placements on the skin of the right forearm.

The signals obtained from the electrodes were conditioned by a four stage filtering and amplification circuit. This circuit was necessary to filter out noise that was outside of the 20-400 Hz range of human EMG signals, as well as to amplify the EMG signal from its raw value of 0-1.0 mV to a more easily recordable value of 0-2.5 V. The first stage of the circuit was a differential amplification stage that amplified the difference between the signals from the forearm muscles with a gain of G = 73.6. The circuit then proceeded to the second stage, where the signal passed through an active second order low pass Bessel filter, which had a cutoff frequency of 380 Hz. In the third stage of the circuit, the signal passed through an active second order high pass Bessel filter, which had a cutoff frequency of 20 Hz. In the final stage of the circuit, the signal passed through another amplification stage with a gain of G = 23.5. Thus, the filtering circuit can be characterized as an amplifier with a gain of G = 2780 and a band pass filter with a pass band of 20-380 Hz.

After being acquired by the simulation computer, the filtered EMG signal was rectified by calculating the mean absolute value of the signal in overlapping 0.1 second intervals, each containing 100 samples. During calibration, a minimum threshold, maximum threshold, rest threshold, and extension value were set by the experimenter. The rectified signal was normalized on a scale between the minimum threshold and maximum threshold. The rest threshold and extension value are used to divide the region of the normalized signal into three areas: an extension region, which causes the virtual prosthetic hand to extend, a rest region, which causes the virtual hand to stay still, and a flexion region, which causes the virtual hand to flex, as illustrated in Fig. 3.

2.2.2 Simulation

The virtual prosthesis was simulated using MATLAB and a Simulink model adapted from Dalley et al. [16]. The position of the virtual prosthesis was represented as an interpolation between a series of states, illustrated in Fig. 4. Control from the EMG signal caused the position of the hand to advance toward the upper end of the series for flexion or the lower end for relaxation. The target positions were always discrete states on the interior of the series, allowing for overshoot and undershoot. The error value used to provide visual and tactile feedback was the distance on the hand position chart between the current hand position and the target hand position. At the beginning of each trial, the virtual prosthesis was reset to the opposition or reposition state according to the target position of the trial. The simulation ran at 1 KHz, with graphics updating at 15 Hz.

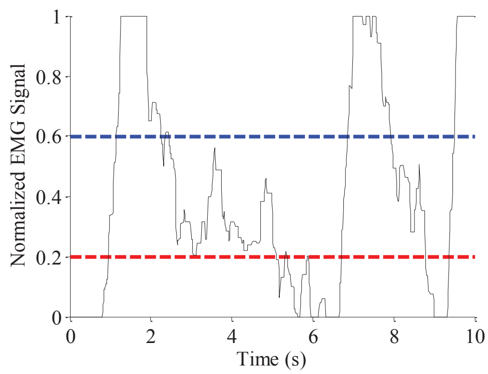

Figure 3. EMG signal control regions. The flexion threshold (upper dashed line) and extension threshold (lower dashed line) divide the normalized EMG signal scale into three regions representing flexion (top), rest (middle), and extension (bottom).

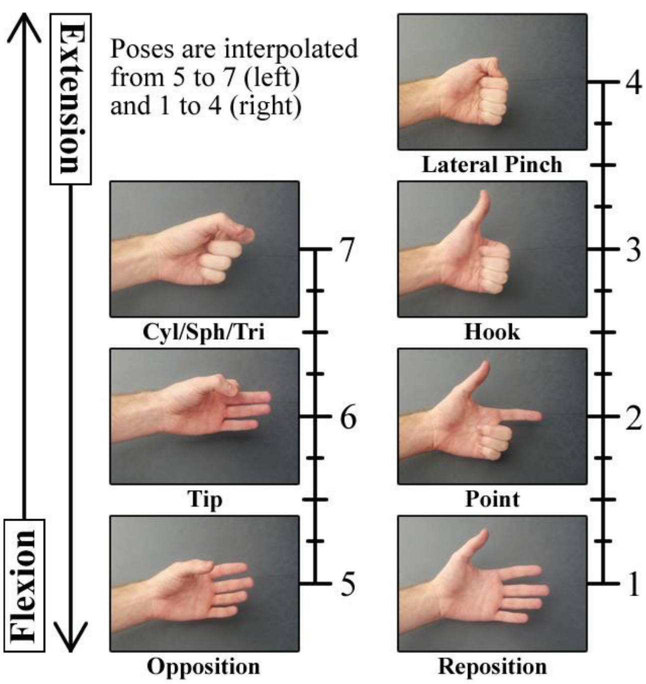

Figure 4. The hand position chart of the virtual prosthesis, adapted from Dalley et al. [16]. Intermediate positions are achieved by linearly interpolating joint angles.

2.2.3 Tactile Feedback

A C2 tactor (Engineering Acoustics, Inc.) was used to provide vibrotactile cues to the subjects. A single tactor was chosen instead of an array of tactors in order to prevent cognitive overloading, which would cause the subject to be unable to distinguish between the different strengths of feedback. The tactor was attached using a velcro mp3 player armband as shown in Fig. 5, and positioned directly over the subject's bicep, which was chosen both as the closest site to the forearm where the EMG control electrodes were attached without causing interference with the EMG signals. Additionally, the cues were positioned on the same arm as the EMG electrodes so that sensation and action were co-located, a configuration shown to be most effective for manual control of a prosthesis [17].

The vibrotactile cue was carefully designed considering human perceptual capabilities and prior psychophysical study results. The threshold for detection of vibrotactile stimulation is high at low frequencies and lowest at or around 250 Hz. Detection thresholds increase again with frequencies in excess of 250 Hz, reaching, at around 1000 Hz, a level comparable to that at the low end of the frequency spectrum [18]. Furthermore, the vibratory frequency range of Pacinian Corpuscles (PC) is 40-800 Hz with a maximum sensitivity near 300 Hz [19]. Additionally, psychophysical studies have shown that there is a loss of sensitivity on the skin after prolonged exposure to intense vibration [3]. This phenomenon, called adaptation, happens only if the frequency is kept constant. However, adaptation can be avoided by modifying the amplitude of vibration [18]. Moreover, the sensitivity of the skin to detect changes in amplitude of vibration increases if the amplitude changes proportional to the logarithm of the cue responsible for the change in the amplitude [20]. In other words, if the amplitude of vibration is a function of a variable x, then the sensitivity of the skin to detect changes in the amplitude increases if the amplitude is changed as a function of the logarithm of x. Therefore, the cue was created by multiplying a sine wave of constant frequency 250 Hz by a sawtooth function of constant frequency 10 Hz. The amplitude of the cue was exponentially proportional to the absolute value of the error in hand position e_{abs}, according to the following equation.

$$v = \sin(250t) * (10t)(0.5e^{e_{abs}}) \qquad (1)$$

This command signal allowed the voltage sent to the tactor amplifier to range from 0.6-3.7 V for the corresponding range of hand position errors possible in the experiment (0.2-2.0, corresponding to the pose classification method). This voltage range resulted in supplying the maximum recommended current for the C2 tactor when maximum error values were generated, producing a vibration amplitude of 580 μm. Conversely, a minimum error produced noticeably weaker vibration amplitudes of 200 μm. As long as the absolute value of the error was less than or equal to 0.2, the subject was considered to be on target, and the tactor voltage was set to zero so that no vibrotactile cue was generated.

Figure 5. The armband used to attach the tactor to the subject. This secured the tactor against the subject's bicep.

2.2.4 Visual Feedback

For the visual feedback modality, the subject viewed an animated 3D rendering of a prosthetic hand on the computer monitor. This rendering displayed the target position of the virtual prosthesis in blue overlaid with the actual position of the virtual prosthesis in white, and updated in real-time. As long as the absolute value of the error between the desired and actual pose was less than or equal to 0.2, or one fifth the distance between poses, the subject was considered to be on target, and the actual position of the virtual prosthesis was not rendered, in order to create equal conditions as the vibrotactile feedback.

2.3 Experimental Procedure

After attaching the EMG electrodes and tactor to the subject, the experimenter used a real-time display of the filtered and rectified EMG signal to calibrate the thresholds. The experimenter calibrated the minimum threshold to be just above the resting EMG value, and the maximum threshold to be at a level of grip strength the subject was comfortable maintaining. The rest threshold and extension value were set such that the subject could consistently grip at a lesser strength within the resting region. The experimenter read a description of the task to the subject, then initiated training. Training consisted of three trials under each of the four feedback conditions (no feedback, visual feedback, tactile feedback, and visual+tactile feedback).

Once the testing began, the system automatically progressed through all 60 trials over the span of 23 minutes. The experimenter did not act again until the experiment was complete and the data were saved. The 60 trials were divided into blocks of 15 trials for each feedback condition, with each block presented in a random order. Within each block of 15 trials, each of the three starting and target pose combinations (1→2, 1→3, and 5→6, with poses indicated by number in Fig. 4) were randomly presented five times. The user was made aware of what feedback condition they were experiencing as well as when they were in a rest period or in a trial using an on-screen text display.

2.4 Data Analysis and Statistical Methods

Due to unintentional shifting of the subject in their chair during the course of the experiment, there were some small intervals in which the EMG signal fluctuated greatly but momentarily, likely due to peeling of the adhesive in the electrode skin contact or reorientation of the subject's arm and muscles. In order to exclude control data that was not intended by the user, all trials in which the rectified EMG signal measured more than 1.5 V for more than one second were excluded from analysis. This excluded 43 of the 840 trials for the 14 subjects considered, amounting to 5.12% of the data. Of those trials excluded, 13 had the visual feedback condition, 14 had the tactile feedback condition, 13 had the visual and tactile feedback condition, and 3 had the no feedback condition. The largest number of trials excluded for a single subject was 8, thus no one subject lost a large amount of data.

Since there were missing data for one feedback condition for one subject, we imputed the data using the means of the rest of the sample based on the feedback condition and poses to account for the no feedback condition of this case. This resulted in our having a sample size of 11. Upon inspection of the data, we found that one subject (#5) appeared to be an outlier as represented by 1-3 SD greater than the mean for each feedback level. Therefore, we analyzed the full dataset and the reduced data set with the outlier removed. Results for the two analyses were similar, and so the full data set results are presented here.

The main effect for Feedback was tested using one-way repeated measures ANOVA, with Subject and Feedback designated as fixed effects factors. Geisser-Greenhouse (G-G) correction was used when violations of sphericity occurred in the omnibus tests. Tukey's post hoc tests were used to determine the locus of the main effects with a 0.05 significance criterion. Four feedback levels were defined for each participant: visual, tactile, visual + tactile and no feedback.

In separate analyses, the main effect of pose was analyzed using the same model as used for the Feedback main effect. As there were no significant effects of pose, only feedback condition is considered in the remainder of this paper. Dependent measures representing performance metrics include: control effort, response time (sec) and time on task (sec). Number Cruncher Statistical Software (NCSS) 2007 (www.ncss.com) was used for the statistical tests.

3 Results and Discussion

The goal of the experiment was to investigate the effect of different position error feedback conditions on the subject's ability to accurately position the virtual prosthetic hand in one of three desired poses using myoelectric control.

3.1 Assessing Task Performance

Performance was measured using the time on target metric, which was the total amount time during a 10 second trial in which the virtual prosthesis position error was less than 0.2. A higher time on target metric meant that the subject had matched the pose for a longer period of time, and thus performed the task better. The results, shown in Fig. 6, indicate the mean time on target to be significantly greater for visual and visual+tactile feedback than for the tactile feedback and no feedback conditions, with mean values of 2.30 seconds for visual feedback, 2.10 seconds for visual+tactile feedback, 0.77 seconds for tactile feedback, and 0.43 seconds for no feedback. There was a significant main effect of the factor feedback condition [$F(3, 39) = 20.67$, $p < 0.001$]. Post-hoc comparisons $q(39) = 3.79$, $p < 0.05$ indicate that the visual (mean $\pm$ SD; 2.30 $\pm$ 1.41s) and visual+tactile (2.10 + 1.49s) feedback conditions outperformed both the tactile(0.77 + 0.63s) feedback and no feedback(0.43 + 0.50s) conditions.

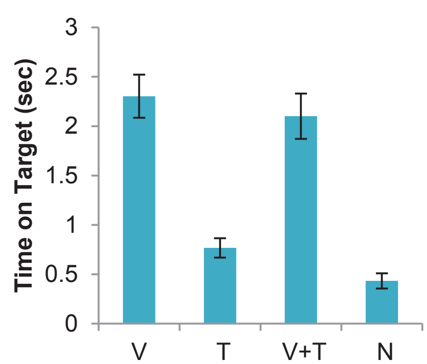

Figure 6. Time on target for the visual, tactile, visual+tactile, and no feedback conditions.

These findings are contrary to previous studies that showed that combined visual+tactile feedback was superior to visual feedback alone in accuracy of pose matching for some conditions of the study, and that tactile feedback alone was not significantly less effective in conveying error information for a pose matching task [15]. There are many possible reasons for our observed disparity in performance between visual and tactile feedback. First, visual feedback is familiar to the subject, while tactile feedback may not be, and may require active learning by the subject in order to be useful. Second, tactile sensing is much less acute than visual sensing, so tactile feedback provides less precise data than visual feedback. Finally, in our study, visual feedback provided information on both the magnitude and direction of the error, while tactile feedback provided only the magnitude, leading in some cases to the subject controlling the virtual prosthesis in the opposite direction of the target pose. In the future, tactile cues designed to convey both magnitude and direction of error will be evaluated, possibly using the approach of Sergi et al. [15].

3.2 Assessing Quality of Task Execution

In addition to assessing the ability of the subjects to match desired poses of the virtual prosthetic hand using myoelectric control, we sought to explore other aspects of task execution that might elucidate the quality of the control/feedback interface. We used two quality measures, response time and control effort.

The first measure of quality was the response time, which was calculated by measuring the time between when the subject suddenly is on target and when the subject control goes to the resting state. This represents the subject reacting to feedback being given that they are on target and stopping or changing the direction of motion in order to stay on target. The process for determining response time from trial data is illustrated in Fig. 7.

Response time could be calculated numerous times during a single trial, in which case the mean was used, or not be calculated at all, in which case no value was used for that trial. This metric was not considered for the no feedback condition, as there was no feedback event that the user received when on target. The results showed the mean response time to be faster for visual feedback than tactile feedback, with mean values of 0.75 seconds for visual feedback, 0.82 seconds for visual+tactile feedback, and 1.04 seconds for tactile feedback. However, there was not a significant main effect of the feedback condition [$F(3, 39) = 2.65$, $p = 0.105$]. This result may indicate that response times were relatively equal across all feedback conditions, thus visual feedback did not have a faster response time than tactile feedback, and perhaps for this application, tactile cues are as effective as visual cues for conveying the presence of error in desired pose. Such a finding indicates that designers of smart prosthetic systems could reduce visual load by conveying proprioceptive information via the tactile channel with no effect on performance in terms of reaction times. However, before we suggest implementation of such changes in the design of smart prosthetic systems, increased sample size and replication of these findings are necessary to reduce the possible influence of Type II errors.

The second measure of quality was the control effort, which was the total area under the curve of the absolute value of the EMG command. The units of this measure are seconds of effort, as the EMG command is a normalized value. In theory, more informative feedback systems should require less control effort, as they allow the subject to more accurately predict their motion. Surprisingly, the mean control effort was very similar across all feedback types, with values of 1.67 for visual feedback, 1.84 for tactile feedback, 1.79 for visual+tactile feedback, and 1.66 for no feedback. There was not a significant main effect of the factor feedback condition [$F(3, 39) < 1.0$, $p = 0.553$]. This result shows that while visual feedback may have provided increased performance to subjects completing the task compared to tactile feedback, it did not make the task significantly easier to perform. In other words, visual feedback allows for greater accuracy than tactile feedback, but both require a similar amount of effort. Additionally, the combined visual+tactile modality does not require greater effort over unimodal feedback conditions as measured by EMG in this manner. Additional work is needed regarding the impact of additional practice trials with the various feedback forms. Practice with additional informative feedback systems, e.g., tactile and visual + tactile, may result in enhanced prediction of motion and ease of use, as well as practice with different schedules of trials [21].

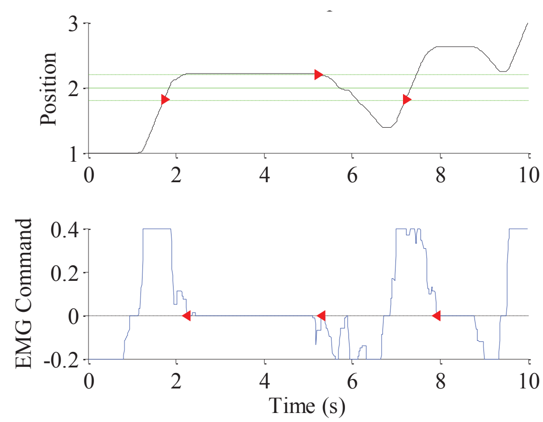

Figure 7. Response time for Subject 8, Trial 21. This trial supplied only tactile feedback, and had an average response time of 0.45 s. The response times are bounded by the right pointing triangles on the position graph and the left pointing triangles on the EMG command graph.

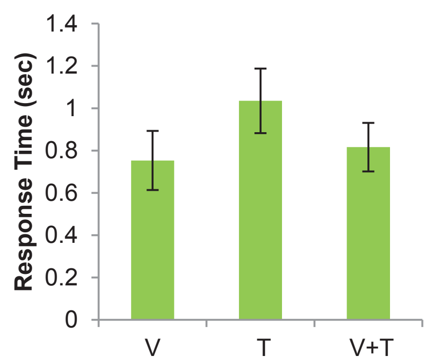

Figure 8. Response time (sec) for the visual, tactile, and visual+tactile feedback conditions.

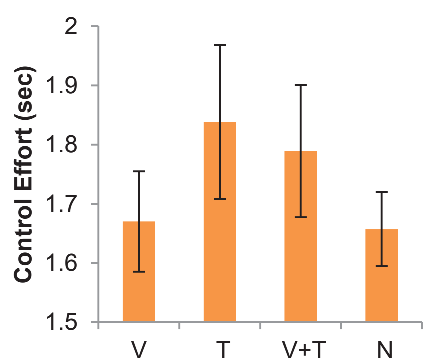

Figure 9. Control effort (seconds of effort) for the visual, tactile, visual+tactile, and no feedback conditions.

4 CONCLUSIONS

While prior studies have explored the utility of tactile feedback for conveying pose information, there existed no prior work that directly compared accurate control of hand pose for multiple feedback modalities in the presence of vibrotactile feedback for a myoelectrically controlled prosthesis scenario. Our results demonstrate that vibrotactile feedback is a viable method to convey hand position information, a finding which is in agreement with other studies [14]. For our experiment assessing the efficacy of various modalities of pose error feedback during myoelectric control of a virtual prosthetic hand, combined visual and vibrotactile cues outperformed the no feedback condition by a significant margin. While visual feedback was superior to vibrotactile feedback in terms of the performance of subjects on the pose matching task, the control efforts and response times for the two types of feedback were not significantly different. This finding is in agreement with a prior study concluding that combined visual+tactile feedback does not result in negative performance effects [15]. The combination of tactile and visual feedback does not seem to have significant benefits over visual feedback alone; however, for tasks that rely on proprioception in the absence of visual feedback, or when attention must be focused elsewhere, we see a benefit to the inclusion of tactile cueing. While we are encouraged with our preliminary findings, we are aware of the limitation of a small sample size on our findings and the resultant low statistical power.

Future experiments may include modifications to improve on the existing design of the tactile cue, such as increasing the range of amplitude for the vibration. Another promising modification would be to use two or more tactors in order to provide a directional (signed) cue, allowing more information to be conveyed. It would also be worth exploring the offloading of cognitive processes that might occur with the inclusion of vibrotactile feedback, as well as in cases where visual feedback is not reliable or is intermittent. To begin such an evaluation, the existing EEG and fNIR datasets from the present experiment will be analyzed. Further, while the participants in this study were not amputees, both the EMG control and the vibrotactile feedback system were fully independent of the hand, thus the results are expected to generalize directly to amputees.

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