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The Role of Auxiliary and Referred Haptic Feedback in Myoelectric Control

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***Abstract*— The use of haptic display to refer cues sensed electronically from a prosthetic terminal device promises to improve the function of myoelectrically controlled upper limb prostheses. This promise is often evaluated in experiments involving non-amputees, though the availability of additional haptic feedback from an intact hand (auxiliary feedback) may confound attempts to use non-amputees as stand-ins for amputees. In this paper we test the influence of auxiliary haptic feedback on myoelectric control performance by introducing various grasp conditions in a compensatory tracking task. We ask non-amputees to compensate for the motion of a random signal by producing myoelectric control signals with a hard object, soft object, or no object (requiring co-contraction) in their grasp. The error signal is displayed through a squeeze band worn about the upper arm or a visual display. Our results suggest that the main difference between tracking with haptic and visual feedback is low-frequency drift, and that auxiliary feedback does not substantially influence task performance. Despite the drift, our results show that participants are able to respond to cues presented through the squeeze band in the compensatory tracking task.**

I. Introduction

Able-bodied individuals have access to a rich array of haptic sensory feedback including grip force and proprioceptive cues to form and guide object manipulation strategies [1]. Traditional shoulder-drive body-powered prostheses cannot rival the sensing capabilities of the human hand, but a minimal level of force feedback is passed through the Bowden cable to the shoulder. In contrast, myoelectric prostheses do not yet provide haptic feedback, even in sophisticated devices such as the iLimb [2] and bebionic [3]. This means that myoelectric users must rely heavily on visual feedback.

Amputees who use myoelectric prostheses often identify the lack of sensory feedback from their devices as a shortcoming [4]. To address this need, efforts are underway to develop haptic display devices that refer force and motion signals sensed electronically at the terminal device to the residual limb. A number of studies have shown that referred haptic feedback can lead to performance improvements in certain manual tasks [5], [6]. In other studies, however, performance differences were not clear between the sensory feedback and no-feedback conditions [7], [8], [9] or differences only became evident for certain populations or when uncertainty was introduced into the feedback loop [10]. As Saunders and Vijayakumar argue, feedfoward control is often sufficient, and lack of feedback does not necessarily lead to deterioration in performance.

Many experiments on referred haptic feedback have involved able-bodied individuals rather than amputees [5]. But can an able-bodied person stand in for an amputee in experiments involving myoelectric control of prostheses? Every muscle in the intact body spans at least one joint, so myoelectric signals produced by a non-amputee are always accompanied by auxiliary haptic feedback, which we define as any information gathered from the intact wrist. This includes exteroceptive and proprioceptive cues such as force feedback, sense of effort, sense of tension, muscle length, and cutaneous cues. For an able-bodied individual, this auxiliary feedback supplements the referred sensory feedback provided by a haptic device, but it is not available for an amputee. The forearm muscles of a transradial amputee that are typically used for myoelectric control generally cannot perform mechanical work, as they do not span skeletal joints.

In this paper we seek to characterize the effect of the auxiliary feedback available to able-bodied persons when using myoelectric control to perform a simple manual task. We undertake an experiment in which we modulate the auxiliary feedback available by asking able-bodied participants to generate myoelectric signals alternately with no object (requiring co-contraction of antagonist muscles), a hard object, or a soft object to grasp. We presume that the mix of cutaneous, proprioceptive, and kinesthetic cues that accompany the generation of myoelectric signals varies significantly across these experimental conditions. To examine the ability of referred feedback to substitute for vision, we also ask our participants to perform the task with and without referred haptic and visual feedback.

A significant challenge to be addressed in the implementation of referred haptic feedback is the communication of terminal device aperture. Proprioceptive cues in an intact hand are certainly sufficient to substitute for vision in the control of grasp aperture. These cues are available to an able-bodied person if their hand moves while generating EMG signals during an experiment. Depending on the experimental task, there may be a direct relationship between auxiliary haptic feedback and the task variable displayed through referred haptic feedback. In such a case the auxiliary feedback could function as a confounding cue. We expect that the closest relationship would exist when myoelectric control (and its associated auxiliary cues) are used in a position-control task. That is, where the controlled variable is directly related to the myoelectric signal without involving an integration or differentiation operation. Thus we propose to

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test whether an able-bodied participant can serve as a proxy for an amputee by adopting a position-control task. We use proportional myoelectric control without first integrating the signal, invoking position control rather than rate control.

Our use of proportional myoelectric *position* control is a contrast to rate control, which is the standard control paradigm in commercial prostheses. Rate control involves differencing the EMG signals from two antagonist muscles; the degree to which the difference favors the flexor or extensor signal determines closing or opening speed, respectively. The wide adoption of rate control in commercial prostheses provides an easy solution to the signal extraction challenges that accompany surface EMG signal acquisition. In particular, unintended prosthesis movement can be prevented with the use of a deadzone, and an amputee can relax when not actively opening or closing the terminal device. Note, however, that with improving myoelectric technology and implanted sensors on the horizon, it becomes worthwhile to revisit position control as a feasible alternative. Proportional position control is actually a better analogue of how an intact hand functions; when carrying an object, the muscles remain engaged, and when setting an object down the muscles relax.

Experiments involving referred sensory feedback for prostheses often address the relative utility of various types of haptic devices and their capacity for supplying signals that substitute for visual feedback. Rather than comparing types of feedback here, we have chosen to employ a single haptic device in our experiment: a squeeze band that tightens around the participant's arm. We initially explored several other feedback devices including a vibrotactile array, a C2 tactor, a skin stretch device, and an exoskeleton that imposed a torque about the wearer's elbow. For our task, features of the haptic device including resolution, the availability of an absolute percept, and continuity of the signal proved to be important to success, and the squeeze band fulfilled these conditions.

To quantify the performance of a task involving referred haptic feedback on the availability of auxiliary sensory feedback in the most rigorous manner possible, we have adapted an experimental paradigm developed in the 1960s to characterize human aircraft piloting performance [11], [12]. In particular, we employ a compensatory tracking task in which the human operator only has access to the error signals (either through vision or through referred haptic feedback).

The value of this tracking task as opposed to manipulation tasks such as grasp and lift is the richness of the data set. The data recorded is continuous instead of discrete and may be evaluated both in the time and frequency domains. There exists a large body of literature related to analysis of tracking results, placing many techniques at our disposal [11], [13].

The piloting literature generally addresses tracking performance using two paradigms: pursuit tracking and compensatory tracking. In the pursuit tracking task the user sees two objects: a target and cursor, with the target displaced a distance r and the cursor a distance y from a fixed ground reference. The signal r is generated by an external agent while the user directly controls the cursor position y through a manual control interface. The user is asked to track the moving target. In the compensatory tracking task, the user sees a single object displaced a distance e from a fixed ground reference (a horizon). This distance is the error $e = r - y$ between the external signal r and the user-controlled signal y. The goal in the compensatory task is to minimize e, or to keep the cursor on the horizon, without direct knowledge of the values of r or y in a ground-fixed reference frame. A depiction of how these two tasks might appear is given in Fig. 1.

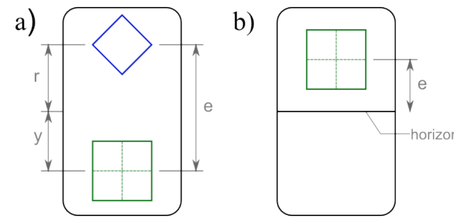

Fig. 1: Two paradigms for tracking tasks: a) pursuit tracking, in which the goal is to follow a target, and b) compensatory tracking, in which the goal is to keep the cursor on the horizon.

An advantage of employing proportional myoelectric position control in the compensatory tracking task is that human tracking performance is superior when the 'plant' under control is a constant (position control) rather than an integrator (rate control) [13], allowing tracking of higher frequencies with lower time delays. Further, the compensatory task itself ensures that any absolute cues that the participant receives come from auxiliary feedback and removes the ability of the participant to succeed without paying attention to the referred feedback by eliminating feed-forward control. Thus, we employ a compensatory tracking task to characterize the effects of auxiliary and referred haptic feedback in able-bodied individuals.

We hypothesized that participants would be able to perform the compensatory tracking task under myoelectric control with haptic feedback even in the absence of visual feedback, and that altering the levels of auxiliary feedback available would significantly change performance on a tracking task performed using myoelectric control.

II. Methods

A. Participant Population

Six able-bodied participants (mean age 26.2 years) were recruited from the community of University of Michigan graduate students and acquaintances. Three participants were female and three male. Five were right-handed and one left-handed. Before starting the study, each participant was consented according to a protocol approved by the Institutional Review Board of the University of Michigan. Participants were not compensated. Testing lasted about 1.25 hours.

B. Experimental Apparatus

EMG signals were measured from finger flexor muscles in the participants' non-dominant forearm. We asked participants to use their non-dominant arm given that most

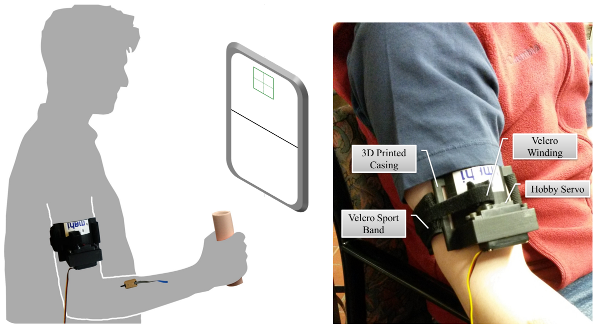

Fig. 2: Experimental setup and detail of haptic squeeze band display device.

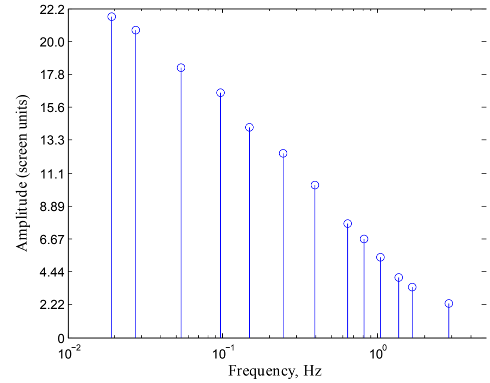

Fig. 4: Frequency makeup of the input signal. The signal is a sum of twelve sinusoids of different amplitudes.

amputees identify their amputated limb to be non-dominant. A commercial sensor was used for EMG signal detection (Otto Bock 13E200). This sensor featured internal EMG signal filtering, amplification, and rectification, such that the output signal was an analog 0-5V. An additional active first order analog lowpass filter with a cutoff frequency of 0.676Hz was used to smooth the EMG signal before it was acquired with a Sensoray S626 PCI Multifunction I/O Board.

Two types of displays were used: visual and haptic. The visual display included a reference line and a cursor on a computer screen, laid out as explained in Fig. 1b. The haptic display, worn on the upper arm, was a servo-driven "squeeze band" weighing 3.8oz. This band, shown in Fig. 2, included a HiTEC HS-485HB hobby servo motor with an adjustable stretchy velcro sport band. A 3D printed motor housing and pulley along with a velcro winding joined the servo to the band such that the servo's rotation wound the band up to tighten the strap around the arm.

A block diagram for the system is shown in Fig. 3. MATLAB/Simulink was used to perform signal generation, visual display, and data acquisition. An Arduino Mega 2560 microcontroller board was used to control the servo motor. The Simulink code produced an analog output on the Sensoray card that the Arduino converted into the PWM signal.

C. Experimental Design

In this experiment, able-bodied participants performed a compensatory tracking task, in which they were asked to use EMG control to cancel the effects of a pseudo-random signal composed of a sum of 12 sinusoids in order to maintain a setpoint. The sinusoids decreased in amplitude as they increased in frequency, as shown in Fig. 4 with amplitudes in "screen units" such that the whole display range for both visual and haptic is -100 to 100 (this unit convention will continue throughout). The procedure for each subject involved nine trials in a three-by-three experiment: tracking performance in three feedback conditions (visual only, visual and haptic, and haptic only) was tested against three types of grip. The three grip types provided different levels of auxiliary feedback: co-contraction lacked motion, squeezing a hard object (a PVC tube) provided only force feedback, and squeezing a compliant object (a block of foam) provided both force and proprioceptive feedback. Trials were grouped by grip type (co-contraction/hard object/soft object). The groups and trials within a group were performed in a different order for all subjects to reduce ordering and learning effects. Each trial lasted 3 min 7 sec.

An additional feedback condition not included in this 3x3 test matrix, that of no feedback, was tested for three participants, each with a different grip condition. No statistical analysis was performed on the no feedback condition.

The high gain of the commercial EMG sensor and the EMG levels found for all participants meant that the subjects all used the same gain for the object grip cases. The gain was increased by 76% for the co-contraction trials to avoid fatigue. If subjects expressed difficulty or fatigue during co-contraction training, the gain was increased by an additional 38% over the original grip gain.

D. Training

Before the training began, participants were first given the following mental image as an analog for the task that they

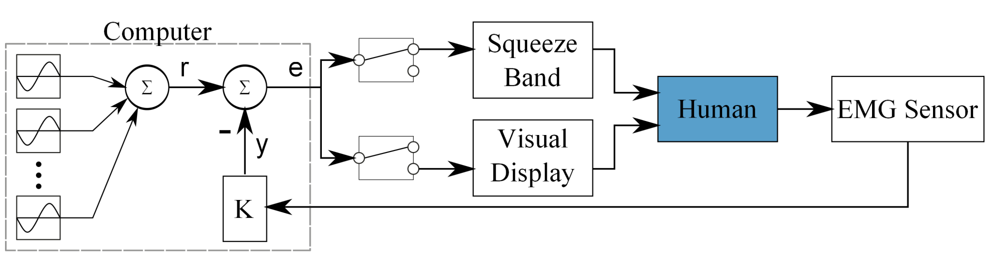

Fig. 3: Block diagram for the compensatory tracking task. Participants used myoelectric control to cancel a computer-generated signal, using the error displayed either visually or haptically.

would be performing: *imagine that instead of this motorized band, you have an inflated blood pressure cuff around your upper arm, and that it has two air bladders attached to it, one squeezed in your hand and one in mine. If I squeezed the bladder I was holding, you would need to relax your grip in order to maintain the same pressure in the arm cuff, and if I released, you would need to squeeze harder.*

After receiving this description, participants performed a training trial. The first third of the trial was performed with haptic feedback only, the second third with haptic and visual together, and the final third again with only haptic feedback. Participants were informed that at the beginning of the trial, the arm band would squeeze to a mid-level setpoint and hold it, and that after 8 seconds the computer would begin to squeeze and release the arm band in a slow predictable sine wave. Participants were instructed to try to cancel the computer's signal to maintain the initial "middle" squeeze level by gripping the object in their hand harder when the band loosened and releasing it when the band tightened.

In addition, participants were informed that for the visual feedback portion, the setpoint would appear as a horizontal line on the screen that they should try to stay as close to as possible. They were informed that the cursor moving up the screen corresponded to the band around their arm tightening, and vice-versa. The sinusoid frequency used for this initial training trial corresponded to the fourth-slowest sinusoid of 0.0961 Hz in the pseudo-random (sum of sinusoids) signal.

A second training run was performed using the same instructions and procedure as the first run, except that the computer-generated signal was the "random" summed sine wave used for the experimental trials instead of a single predictable sine wave.

Before the beginning of the co-contraction block, the participant was given a chance to learn how to control their finger flexor muscles through co-contraction, with both visual and haptic feedback on. This training lasted about 30s, or until the participant confirmed that they felt comfortable. To avoid excess fatigue, a full training run was not performed under co-contraction, since several of the subjects felt it to be much more tiring than the other conditions.

III. Results

Our participants were able to generate myoelectric signals that compensated for the pseudo-random signal generated by the computer, though differences did appear across display conditions. In particular, the haptic only case proved different from the cases with visual feedback, with higher errors characterized by a large amount of drift from the original setpoint. Throughout the results the three display conditions will be referred to as V (visual only), VH (visual and haptic), and H (haptic only). The three grip conditions will be referred to as co-C, hard (PVC tube), and soft (foam block).

Selected sample results for subject 6 are provided in Fig. 5 in the time and frequency domains. The left and center columns present results for the H and V display conditions. For comparison, the right column of Fig. 5 shows the performance of subject 6 when no feedback is provided. In the frequency domain, it should be noted that for the V and H conditions, the user's output EMG signal y is characterized by peaks of high magnitude that correspond

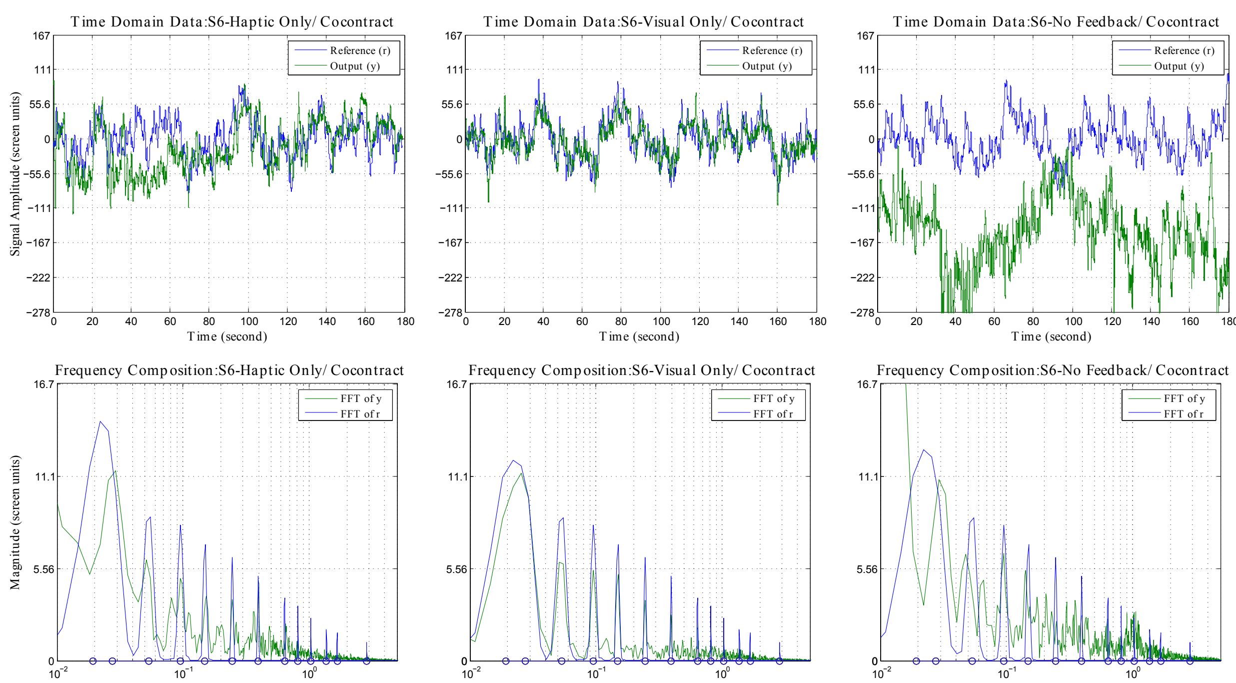

Fig. 5: Sample results for subject 6, under co-contraction. Values for r and y presented in (top) the time domain, and (bottom) the frequency domain. From left to right, the display conditions are haptic only, visual only, and no feedback. In the frequency-domain plots, the circles along the x-axis indicate the injected frequencies of r. For all plots, r is represented in blue, and the user-generated y is shown in green. HV condition not shown because of its similarity to the V condition.

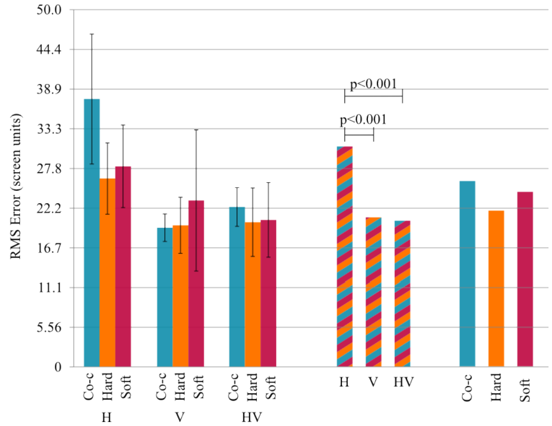

Fig. 6: Root mean square level of e, averaged over all subjects by trial (left 9 bars) and by condition (right 6 bars). Error measurement made with respect to the provided target.

to the frequencies injected in r, with some amount of noise between those frequencies at a much lower magnitude. In the no feedback condition (right column), however, the spectral power is more distributed, with large peaks between the frequencies that make up r. At this level of analysis, these sample results for subject 6 are typical of all subjects.

For all three participants who performed the no feedback condition, the frequency-domain results showed a distinct lack of correlation between the frequencies in r and y. Thus, these results illustrate that the condition amounts to random flexing of the muscle, since without visual or haptic feedback, there is no signal indication.

A measure of the error throughout the trial was taken using a root mean square level of e for each trial. The results provided in Fig. 6 are averaged across all six subjects. A 2-way ANOVA test indicated significant results by display type (H/V/HV), with $F(2,45) = 16.04, p < 0.001$. No significant effect from the grip type (co-C/hard/soft) was indicated, $F(2,45) = 2.40, p = 0.102$, and there was no indication of significant interaction effects, $F(4,45) = 2.37, p = 0.067$. A follow-up multiple comparison test revealed that the significant differences in RMS error between display conditions were between H and V display conditions, $p < 0.001$, and between H and HV, $p < 0.001$.

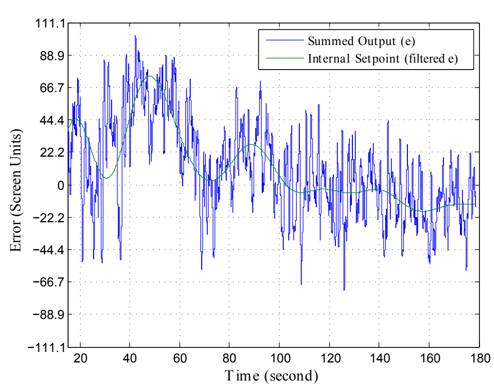

Fig. 7: Lowpass filtering of e to derive the subject's "internal setpoint". Example shown is co-C/H condition for subject 6.

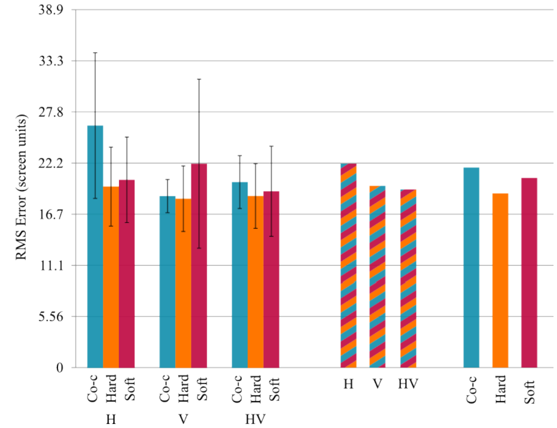

Fig. 8: Root mean square level of adjusted e, averaged over all subjects by trial (left 9 bars) and by condition (right 6 bars). Error measurement made with respect to internal setpoint determined by low-pass filtering the signal.

To achieve a measure of drift during the trials without visual feedback, a lowpass filter was applied to the recorded error signal in post-processing. The low-frequency filtered signal, essentially a copy of the participant's "internal setpoint", was subtracted off from the error signal e, and a new RMS level was calculated for the adjusted error. The digital filter used to adjust the drift out of the error in post-processing was a zero-phase digital filter based on the transfer function for a 12th order butterworth lowpass filter with a normalized cutoff frequency of 0.00008π radians per sample. A sample output for the filter is shown in Fig. 7, along with the original error signal.

The RMS error level for this adjusted error measurement is shown in Fig. 8. Based on a 2-way ANOVA, a marginally significant effect was found by feedback type, $F(2,45) = 3.32, p = 0.045$, but there was no significant difference by grip type or from an interaction effect. A multiple comparison test between the feedback types, however, yielded a $p = 0.067$ between H and V, $p = 0.089$ between H and HV, and $p = 0.990$ between V and HV.

The drift measurement is the change in RMS level from the raw measurement e (Fig. 6) to the RMS level of the adjusted error. The resulting drift measurements, averaged across all subjects, are shown in Fig. 9. It should be noted that the adjusted error is the difference between Fig. 6 and Fig. 9. An ANOVA test on the drift measure showed a significant effect by feedback type, $F(2,45) = 20.9, p < 0.001$, but no significant difference by grip type, $F(2,45) = 0.65, p = 0.526$, or significant interaction effect. A follow-up multiple comparison test revealed significant differences between haptic and visual feedback, $p < 0.001$, and between haptic and haptic/visual together, $p < 0.001$.

IV. Discussion and Conclusions

Our results indicate that the major performance difference between haptic feedback alone and the two conditions involving visual feedback is the drift of the subject's internal copy of the original setpoint (see Figs. 8 and 9). The lack

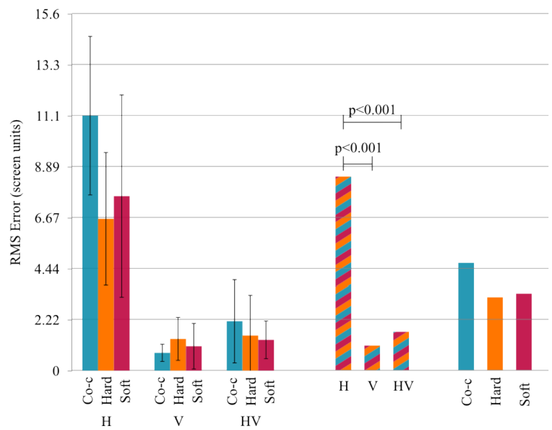

Fig. 9: Drift, as measured by the change in RMS level of e before and after adjustment with low-pass filter. Data averaged over all subjects by trial (left 9 bars) and by condition (right 6 bars).

of significant difference across condition for the corrected error RMS and the significant differences across condition for the change in RMS indicate that low-frequency drift is the primary determinant of performance changes with and without visual feedback. In other words, while participants accrued high absolute error over the course of the trials without visual feedback (Fig. 6), they were still reacting to the signal that they felt through the squeeze band. Practically, this means that haptic feedback of the type used here would be most useful over shorter periods, allowing the user to look away briefly during tasks, but might not support absolute positioning over periods of 10 seconds or more.

The absence of an effect by grip type on tracking ability came as a surprise to us, given the rigorous nature of the compensatory tracking test. However, it is not a discouraging outcome, as it implies that different levels of auxiliary feedback potentially do not play as much of a role as we had originally suspected in proportional EMG control. This is a positive finding for prosthetics research, because it indicates that amputees and able-bodied individuals may have comparable performance at EMG control tasks.

While the compensatory tracking task that we adopted cannot be considered a functional task, it did succeed in discriminating performance differences across the conditions involving referred haptic feedback, visual feedback, or both. Note that the tracking paradigm precluded the use feed-forward control, forcing participants to rely fully on real-time sensory feedback. This is distinctly illustrated by the nonsense results from the no feedback condition, since for the compensatory tracking task the reference signal r was not available without feedback. These trials amount to the participant randomly flexing the muscle without any indication of what (s)he should be doing.

The quantitative measures used to evaluate tracking performance here have all been derived from data in the time domain. The task, however, was chosen specifically because of the wide range of analysis techniques available, including examining the data in the frequency domain. The only frequency domain analysis so far has been a visual confirmation that the low frequencies in r are also present in y at higher amplitudes than the noise. Our next step is to extend the experiment to amputee subjects. So far, we have limited the work to characterizing EMG control by able-bodied subjects, and have seen indications that auxiliary feedback does not appear to play a role in signal tracking with EMG.

A major question not addressed in the current work is how proportional myoelectric position and rate control compare in the context of referred haptic feedback from a prosthetic device. It seems that since the feedback is in proportion to either grip force or position of the prosthetic hand, proportional position control would be the most intuitive and would therefore provide the largest advantage when paired with feedback. However, humans are capable of learning rate control quite well (i.e., steering a car). In future experiments we will explore whether haptic feedback paired with myoelectric rate control has the same utility as it does when paired with proportional myoelectric position control.

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