Understanding the Role of Haptic Feedback in a Teleoperated/Prosthetic Grasp and Lift Task

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Abstract

Achieving dexterous volitional control of an upper-limb prosthetic device will require multimodal sensory feedback that goes beyond vision. Haptic display is well-positioned to provide this additional sensory information. Haptic display, however, includes a diverse set of modalities that encode information differently. We have begun to make a comparison between two of these modalities, force feedback spanning the elbow, and amplitude-modulated vibrotactile feedback, based on performance in a functional grasp and lift task. In randomly ordered trials, we assessed the performance of N=11 participants (8 able-bodied, 3 amputee) attempting to grasp and lift an object using an EMG controlled gripper under three feedback conditions (no feedback, vibrotactile feedback, and force feedback), and two object weights that were undetectable by vision. Preliminary results indicate differences between able-bodied and amputee participants in coordination of grasp and lift forces. In addition, both force feedback and vibrotactile feedback contribute to significantly better task performance (fewer slips) and better adaptation following an unpredicted weight change. This suggests that the development and utilization of internal models for predictive control is more intuitive in the presence of haptic feedback.

Keywords: human-machine interface, prosthetics, sensory substitution, grasp and lift

1 Introduction

Given recent advances in actuator and sensor technology, upper-limb prosthetic development has seen an explosion in innovation, moving devices closer to the physiological form and function of the natural limbs they are replacing. Unfortunately, the ability to accurately and efficiently control the additional degrees of freedom lags significantly behind. In the intact limb, dexterous control relies on efferent pathways carrying user intent to the neuromuscular system and afferent pathways bringing feedback (both anticipated and unanticipated) to the central nervous system (CNS). This sensory information is presumed to be used by the CNS to develop and refine internal models of the limb and the environment it is operating within for dexterous control [7, 8].

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For upper-limb amputees, all efferent and afferent pathways abruptly end at the most distal point in the residual limb. Prosthetic limbs can therefore provide an artificial conduit through which these efferent and afferent signals can be restored. In terms of interpreting user intent, myoelectric sensors and force-sensing resistors provide alternatives. In addition, emerging technologies like targeted muscle reinnervation move closer to detecting user intent directly from the efferent command [13, 16]. Vision currently serves as the primary afferent signal. But alone, vision cannot provide the information needed for robust efficient volitional control. What lacks therefore, is adequate multimodal (vision, tactile, audition, and proprioceptive) afferent pathways capable of providing feedback regarding the device's interaction with the environment. Although there are several novel pursuits in the development of control and feedback technologies that can directly interface with the peripheral nerves of the residual limb, these technologies are still many years from being fully realized [1, 19]. What then can be done in the short term for amputees to improve the functionality and utility of current devices?

Waiting to be developed are a suite of haptic feedback technologies that can be used to non-invasively provide sensory feedback to amputees wearing prosthetic devices. Indeed, work has already been conducted on a few of these haptic feedback modalities, including: vibrotactile feedback [4, 5], skin stretch feedback [12, 21], and force feedback [9, 17]. Since all of these technologies hold promise, a comparison should be made to determine which modality provides the most utility to the amputee. It could be that the ideal modality depends on the task for which it is being used.

A good place to begin these comparisons would be with a functional task that has applications to an amputee's activities of daily living. Performing this task should rely heavily on multimodal sensory feedback, especially haptic feedback. The task should provide a rich data set from which comparisons could be made, both between modalities and with the preexisting literature. Ideally, benchmark comparisons could be made between amputees and able-bodied individuals performing the same task.

In this paper, we attempt to explore whether the choice of haptic feedback modality has an effect on performance during a specific type of object manipulation task, a grasp and lift task. The choice of grasp and lift as the functional task allows for an in-depth look at the impact of sensory feedback on task performance.

In healthy individuals, grasp and lift involves fine coordination of the grasp and lift forces. This careful coordination involves a parallel increase in the grip and load force prior to lifting. During the lifting phase, the object reaches its desired height in a critically damped fashion. These grip and load forces also scale with object weight. In addition, the grip force and load force rates are single-peaked, bell shaped, and scale with the load force needed to lift the object. This behavior suggests that the movements involved in task completion are planned, and planning involves using internal representations of the object's weight and other properties [7, 8, 10].

As a demonstration of this claim, when the weight of the object is experimentally manipulated to be lighter or heavier than expected, the resulting lifting phase occurs sooner than expected or not at all. The error arising from this mismatch between actual and predicted sensory information causes an increase or decrease in the fingertip forces, and an update of the internal representation of the object [7, 8, 10]. This updated representation is then used to ensure efficient coordination of grasp and lift forces on subsequent attempts.

For subjects with impaired sensory feedback, the grip force and load force are less correlated, resulting in elevated grip forces prior to the onset of movement. This safety-factor behavior is similar to results seen in healthy subjects with temporarily anesthetized fingers [15].

Cutaneous afferents from the fingers play four important roles during manipulation: maintaining a background level in grip force, detecting incipient slip, modulating the grip/force ratio based on object friction, and updating internal representations of the object properties. These roles are superior to vision and proprioceptive afferents in terms of providing information about mechanical events at the fingertips [11]. The ideal prosthesis should, therefore, be able to substitute the information coming from the entire suite of tactile afferents (FA-I, FA-II, SA-I, SA-II) that innervate the inside of the hand, as well as provide proprioceptive information regarding the configuration of the limb.

Given the complexity of such a task, it seems more fitting to start with two very important signals: slip-detection and grip force. Development of slip detection feedback systems is well underway [6, 20], and different haptic modalities have been used to provide grip force [4, 9, 14, 18]. In particular, when these tasks involve grasp and lift of a weighted object, performance with closed-loop control (providing haptic feedback) was better than performance with open-loop control (no haptic feedback) [14], especially in the presence of feedforward uncertainty [18].

In our previous work [2, 9], we found that force feedback spanning the elbow provides sufficient grip information to identify an object by its stiffness. We then demonstrated [3] through a simple haptic interface that the force and motion cues used to identify stiffness are best interpreted when the exploratory action and resulting force feedback are co-located at the same point of contact. We argue that only force feedback has the ability to co-locate force and action, and all other haptic modalities operate in a non co-located fashion.

In prosthetics, the use of haptic feedback involves some form of sensory substitution, and real-life tasks are more complex than object identification. How then do these various haptic feedback modalities compare in a grasp and lift task, where sensory substitution is necessary for the development of internal models of the prosthetic limb and object being manipulated?

We start with a comparison between two widely used haptic feedback modalities. Using an EMG controlled gripper, we will test both able-bodied and amputee participants with and without haptic feedback. Haptic feedback will either be force feedback or vibrotactile feedback. Visual and auditory feedback will be allowed in all conditions, making the no haptic feedback case similar to operation of current prosthetic devices. We will use different object weights, and will randomize the presentation of the weight and feedback to participants without notification.

1.1 Hypothesis

We expect to see better coordination of grasp and lift forces with haptic feedback than without haptic feedback. In particular, we expect that there will be more failed attempts (object slips) without haptic feedback than with haptic feedback, especially when there is a weight change. In terms of a comparison between the two types of haptic feedback, we expect force feedback to perform better than vibrotactile feedback.

2 Methods

2.1 Experimental Setup

Our experimental apparatus consisted of a motorized elbow brace, a motorized gripper, an instrumented object, and a vibrotactile display.

The motorized elbow brace is used to provide force feedback in the form of an extension moment spanning the elbow joint. It consisted of a right-handed Aircast Mayo Clinic Elbow Brace that has been customized with an attached motorized capstan drive. The DC motor used in the capstan drive is a Maxon RE 30 (60W) and was powered by a 24V power supply (TDK-Lambda ZWS150PAF) and H-Bridge amplifier (Advanced Motion Control 12A8). The motor was equipped with a rotary encoder on the motor shaft (Maxon 1024 CPR) and brace shaft (US Digital, 2500 CPR). The motorized brace was capable of delivering 0.15Nm of torque. Participants' arms were secured in the elbow brace through four velcro straps. For amputee participants, custom orthotic inserts were used in addition to the velcro straps. The width of the brace could also be adjusted. In operation, the motorized brace produced an extension moment about the elbow proportional to the measured grip force (Figure 1).

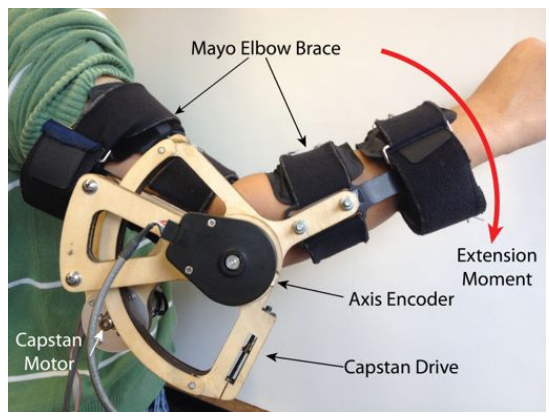


Figure 1: Motorized elbow brace with right-handed Aircast Mayo Clinic elbow brace and motorized capstan drive. Brace produced an extension moment about the elbow proportional to the forces measures by the gripper.

The motorized gripper was driven by a capstan drive powered by the same setup as the motorized brace. In addition, the gripper was equipped with a 5kg-capacity beam load cell (Transducer Techniques LSP-5). The gripper was capable of delivering 7.75N of force at the tip of the end-effector. The gripper was hand held about a foam grip for able-bodied participants, and mounted to the motorized elbow brace for amputee participants. In operation the gripper was position-controlled from EMG signals derived from the flexor carpi radialis muscle in the forearm (Figure 2).

The vibrotactile display consisted of an Engineering Acoustics Inc. C2 tactor driven through a 12-17V 8.5A 110W power supply and H-Bridge amplifier (LOGOSOL DC Servo Amplifier LS-5Y-12-DE). The tactor was held in place using an off-the-shelf mp3 player sports arm band. In operation, the tactor's vibration amplitude (T_c) was exponentially proportional to the measured grip force and driven according to Equation (1) with $T_{c_{amp}} = 0.1sec$, $T_{c_{freq}} = 250Hz$, and $T_{c_{ref}} = \frac{|Grip\ Force|}{Maximum\ Grip\ Force}$.

$$T_c = 0.5 \cdot e^{2T_{c_{ref}}} \cdot \frac{t}{T_{c_{amp}}} \cdot \sin\left(2\pi T_{c_{freq}} t\right) \quad (1)$$

The instrumented object was a custom, ABS plastic, 3D printed device with a hinged side, removable drawer for inserting a weight,

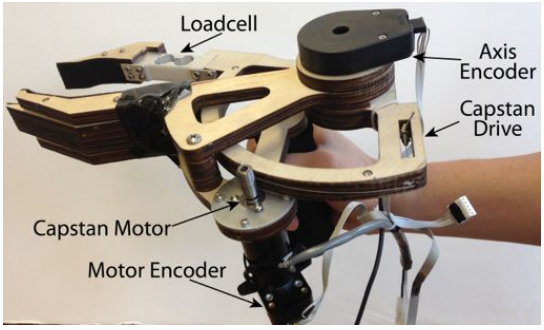


Figure 2: Motorized gripper with powered capstan drive and load cell attached. Encoders on axis and motor shaft.

Figure 3: C2 tactor inside mp3 sports band.

and rubber grips for grasping. The hinged side was held in place by a magnet on a moving slide. The magnet and slide were adjustable to vary the gripping force required to "break" the object, by overcoming the magnetic force. The object had two infrared distance sensors (Sharp 2D120X) to measure vertical position, which were used in conjunction with a 1-lb capacity force plate (AMTI HE6X6-1) to measure the vertical load. A piece of white card stock was attached to the top of the force plate to allow for more accurate position readings from the encoders (Figure 4).

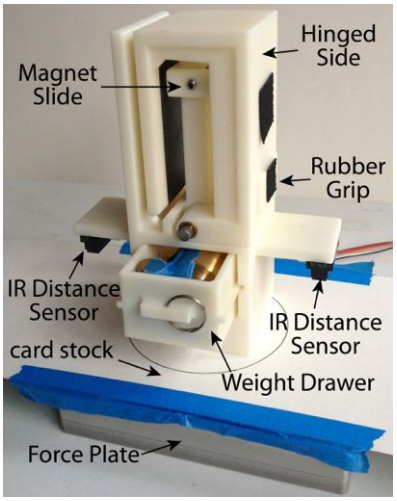


Figure 4: Testing object with attached optical encoders and removable weight drawer atop force plate with white card stock on top for optical encoder readings.

Control signals were taken from two EMG surface electrodes placed on the flexor carpi radialis muscle with a reference electrode on the clavicle bone. A custom conditioning circuit provided full-wave rectification, low-pass filtering with a 3.4Hz cutoff frequency, and variable amplification of the raw signal. Although the electrodes had adhesive backing, a compression sleeve was used to keep the electrodes from coming loose during the experiment.

The entire system was controlled by a Sensoray 626 PCI card installed in a Dell OptiPlex 7010 series desktop running Microsoft Visual C++ 2010 Express Edition.

Brain imaging was also measured using 32 channel scalp electroencephalography (EEG) with a BrainAmps DC amplifier and actiCAP active electrodes (Brain Products GmbH) and a 16-channel fNIR (fNIR Imager 1000) system (Figure 5). Brain imaging results will not be presented in this paper.

(a) Able-Bodied Participant (b) Amputee Participant

Figure 5: Testing setup for both able-bodied and amputee participants. Participants are wearing motorized elbow brace, motorized gripper, vibrotactile display, and brain imaging sensors.

2.2 Experimental Protocol

We tested N=11 male participants, eight able-bodied (mean age 26.6) and three trans-radial amputees (mean age 53.3). The first participant withdrew from the study prior to testing, so results are only given for the remaining 10 participants. Prior to starting the study, each participant was consented according to the protocol approved by the University of Michigan's Institutional Review Board, then given an overview of the study. Participants were not compensated, and testing lasted two hours.

2.2.1 Training

Due to the design of the brace, the right arm was used for experimentation. The vibrotactile display was placed on top of the biceps muscle of the right arm and secured with the velcro straps of the sports band. The motorized exoskeleton was placed around the upper and lower portions of the right arm with the elbow joint in line with the brace's axis of rotation. The brace was secured with velcro straps.The EMG gain was adjusted to ensure the participant's EMG signal was in the 0-5V range. This was checked using an oscilloscope. The EMG control gains and biases were adjusted so that the participant could open and close the gripper, as well as grasp and lift the instrumented object at the maximum weight three successive times. The force feedback and vibrotactile feedback gains were adjusted until they were independently recognized by the participant when grasping an object similar in size to the test object. The subject was then fitted with the fNIR and EEG systems.

2.2.2 Testing

The test consisted of 144 trials broken into four blocks of 36 trials each. There were three conditions being tested: vibrotactile feedback, force feedback, and no feedback. Visual and auditory signals

were not blocked during any of the trials. The weight and condition were arranged based on a stratified randomization on two factors. There were two weights being used, 340g (drawer empty) and 590g (250g weight in drawer). During each block, there were 12 trials of each condition and 18 presentations of each weight placed randomly throughout. The magnet and slide were set at the maximum level to ensure the object didn't break under normal gripping conditions.

Each trial lasted 10 seconds. During the trial, the participant was instructed to start from a rest position (holding the gripper in their hand), close and open the gripper, then reach, grasp, and lift the instrumented object, and place it back on the force plate. Subjects were instructed to grab the object at the rubber grips and lift it a few inches off the force plate before returning it. After the 10 seconds were up, the tester would remove the object from the force plate and change weights (if necessary) behind a cardboard curtain before replacing the object on the force plate for the next trial. This was also timed at 10 seconds (for participant two, 15 seconds was used). The participant was not aware of weight or condition changes prior to grasping and lifting the object. A timer with bell chimes kept track of timing. A break lasting a minimum of three minutes was taken after each block of 36 trials. Prior to starting each block, the control signals and feedback actuators were checked to ensure signal fidelity.

There were a few notable changes to the protocol made for amputee participants. The first was that participants were not required to close the gripper at the beginning of each trial prior to grasping and lifting the object. Also, for amputee participants we only included the first two blocks. This was based on results from the able-bodied participants. The last change was that we tested one amputee participant with a flexion moment force feedback as opposed to an extension moment.

2.3 Metrics

Our task performance metrics consisted of percent slip by condition overall and percent slip by condition after a weight transition.

2.4 Statistical Analysis

Linear mixed models using a restricted maximum likelihood estimate for variance components was used for analyses. Within the model, subjects were a random effect while fixed effects included participant type (able-bodied or amputee), block, object weight, feedback condition, and protocol (accounting for the force feedback protocol change for the last amputee participant). Bonferroni adjustments were applied to estimated means to control for Type I errors and a significance criterion of 0.05 was used for all tests. To determine the effect of feedback with vibrotactile and force feedback as one condition while no haptic feedback was the other condition, a separate linear mixed model was calculated including the subject factor as a random effect.

3 Results

Some of our participants encountered issues with the EEG and fNIR head gear. As a result, of our ten participants, only four completed all four blocks of the experiment. Five completed three blocks, and the remaining five only completed two blocks. This last group includes all amputee participants.

Figure 6 shows the grip force, load force, and position traces for all no feedback trials in the first block. The traces in Figure 6a are from an able-bodied participant, while the traces in Figure 6b are from an amputee participant. These do not include trials where there was a weight transition, or trials where the object slipped. T=0 represents the time point at which the load force is 5% of its max.

For the light to heavy transition (Figure 7), we analyzed how often the object slipped for the five participants who completed at least three blocks. For the first and second trial after the transition, there is no significant difference by condition on percent slip (Figure 7a,7b). For the third trial after the transition, there were significantly more slips in the no feedback condition (M=25%,SE=5.72%) that either the vibrotactile feedback (M=8%,SE=5.72%) or force feedback condition (M=8%,SE=5.72%) ($p<0.05$) (Figure 7c).

For the overall performance (Figure 8), we analyzed how often the object slipped for all participants in the first two blocks. We found that there were significantly more slips in the no feedback condition (M=43.22%, SE=4.06%) than either vibrotactile condition (M=34.24%, SE=4.06%) or the force feedback condition (M=33.55%, SE=4.06%) ($p<0.05$). An additional analysis treating the vibrotactile and force feedback condition as one haptic feedback condition was performed. The results show that there were significantly more slips in the no feedback condition (M=43.22%, SE=3.5%) than in the combined haptic feedback condition (M=33.9%, SE=3.5%) ($p<0.01$).

The current results do not show any significant differences in percentage slip between block, participant type (able-bodied or amputee), or protocol.

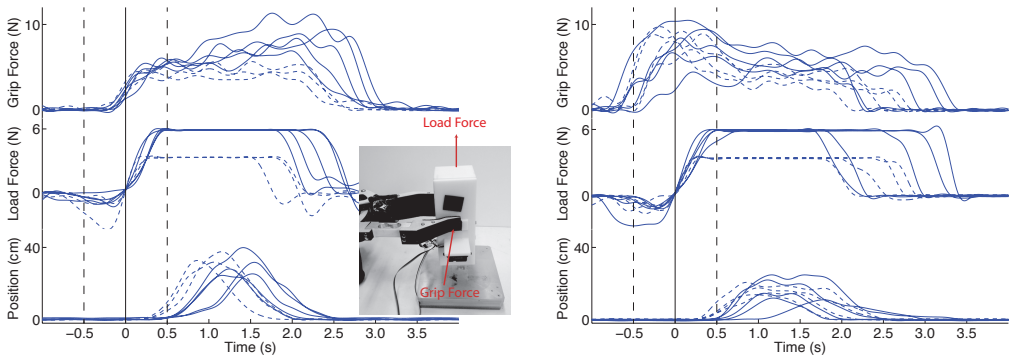
4 Discussion

In this current study, we have set out to understand the utility of haptic feedback on a grasp and lift task with a prosthetic gripper. We have chosen the grasp and lift task because of its reliance on the use of sensory afferents to develop internal representations of the object being lifted and the surrounding environment. In healthy individuals, these representations are used for predictive control of the limb, allowing for grip and load forces to be controlled in a highly coordinated manner. In addition, these representations are used to compute errors between predicted and actual sensory afferents, which in turn update the representations themselves. In impaired individuals with reduced sensory afferents and efferents due to amputation or engineered through the use of a teleoperated/prosthetic gripper, the ability to form and update these representations seems limited. We believe that the addition of haptic feedback will provide greater assistance in this aim. Not all haptic display modalities are created equal, and our ultimate aim is to determine which modality provides the most utility. We are interested in the coordination between grip and load forces, adaptation to changes in object weight, and overall performance in the manual grasp and lift task. Although we have only begun to tease out the results of the current experiment, our initial findings suggest that there may be differences between able-bodied and amputee participants in the coordination and timing of grip and load forces, and the use of haptic feedback produces significantly fewer slips during weight change adaptation, and significantly fewer slips overall.

4.1 Grasp and Lift Force Coordination

In our time series analysis, we get a sense of how participants approached the manual grasp and lift task in terms of coordinating grasp and lift forces for objects with accurately predicted weight. For both the amputee and able-bodied participant, there was a parallel increase in the grip and load forces prior to object lift-off. This is consistent with results found in the natural limb [10]. In addition, the grip and load force traces separate visually by object weight. This is most evident in the load plots with few exceptions. There are no noticeable differences in the object position by weight. This result, although expected, confirms that the weight of the object did not inhibit the participant's ability to perform the task.

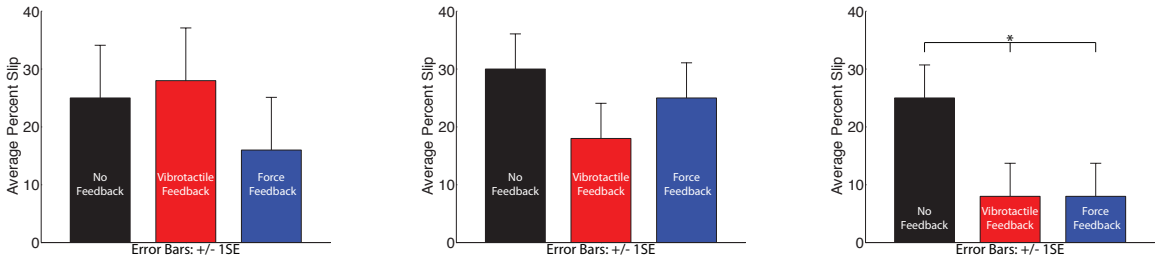
Of more interest is the difference in grip force coordination between the able-bodied and amputee participant. For the able-bodied participant, the grip force began to increase just before the $t = 0s$ time point. For both the heavy and light object, the grip force followed the same pattern, increasing to roughly the same level at $t = 0.5s$, then either increasing or staying the same if the object was heavy or light, respectively. In this manner, it seems as though



(a) Able-Bodied Participant

(b) Amputee Participant

Figure 6: Sample time-domain plots: (a) Able-bodied (b) Amputee participant. All traces are for successful (no slip) trials not immediately following a weight transition in the no feedback condition. Dashed lines (- -) represent light objects, solid lines (-) represent heavy objects. T=0 represents the time point at which the load force was 5% of its max.

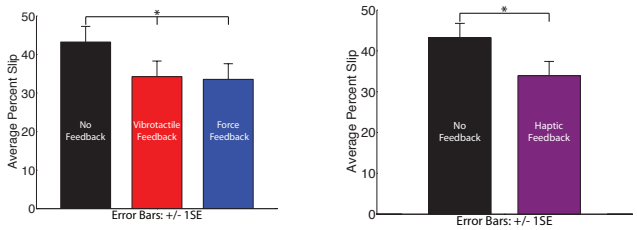


(a) Light Heavy Transition

(b) Light Heavy Transition + 1

(c) Light Heavy Transition + 2

Figure 7: (a) Average percent slip by condition for first trial following light to heavy weight transition for all subjects who completed 3 blocks. (b) Average percent slip by condition for second trial following light to heavy weight transition for all subjects who completed 3 blocks. (c) Average percent slip by condition for third trial following light to heavy weight transition for all subjects who completed 3 blocks.



(a) Uncombined

(b) Combined Haptic Feedback

Figure 8: Average percent slip by condition in blocks 1&2 (first 72 trials). (a) Uncombined refers to conditions no feedback, vibrotactile feedback, and force feedback. (b) Combined refers to conditions no feedback and vibrotactile+force feedback.

the participant had a targeted "checkpoint" grip force level at lift-off from which further adjustments could be made, based on object weight.

For the amputee participant, the grip force began to increase before the $t = -0.5s$ time point. Once initiated, the grip force increased to a level at $t = 0s$ higher than that observed in the able-bodied participant. At $t = 0s$, the grip force decreased to a level similar to that used by the able-bodied participant. This level was subsequently adjusted based on object weight. In this manner, the participant appeared to initially use a grip force that was at a "safety factor" level. This is similar to findings from impaired individuals and those with anesthetized digits [15].

Perhaps these trends provide insight into the generalizable manner in which able-bodied and amputee participants approach the task, or perhaps these differences are participant dependent. At present, they can only motivate further investigation.

4.2 Weight Transition

Important in the transition from a light object to a heavy object is how quickly participants adapt to the new weight. We believe this adaptation involves a revision of the internal representation of the object to one heavier than the current representation predicts, and updating the necessary motor commands. The revision and update are based solely on the multimodal sensory feedback provided. In the absence of haptic feedback, the object weight, which is detected in hand holding the gripper or through the rigid brace, along with visual monitoring of object liftoff are all that are present. When grip force is provided via haptic feedback, there is additional information available that could be used to control grip and load force coordination. In addition, we acknowledge that there were possibly other cues available, such as the manner in which the presenter held the object, but these types of confounding factors are present in all three conditions. Therefore, we would expect to see a difference in grip and load force coordination with and without haptic feedback. In particular, if the haptic feedback is useful, there should be improvements in grip and load force coordination with each successive presentation of the object, after the transition occurs.

A broader analysis of adaptation can be made by assessing how

successful the grasp and lift task is after the transition. How well the absence or addition of haptic feedback helps in adaptation to a change in weight should be reflected in how often the object slips during a grasp and lift attempt. Our results from five able-bodied participants in the first three blocks show that the addition of haptic feedback (vibrotactile or force) does significantly reduce how often slips occur by the second presentation of the heavy weight after the transition (transition + 2). This is compared to the transition and first presentation after the transition (transition + 1). This finding serves as evidence suggesting that the haptic feedback does provide better adaptation than no haptic feedback. It can be concluded then that sensory information is being used to update and maintain internal representations of the object used for volitional control.

4.3 Overall Performance

The fact that there were more slips overall without haptic feedback alludes to the potential utility haptic feedback could provide to amputees wearing active prostheses. This finding is also supported by findings that grip force control is enhanced by haptic feedback for surface EMG controlled hands in targeted reinnervation amputees [12]. In addition, the promising results in the area of adaptation suggest that this sensory information is in some way being incorporated in the brain's motor schema and is being used for predictive control. This finding combined with the fact that there were no differences between able-bodied and amputee participants in terms of overall performance suggest how broad of an impact this could have on limb use and functionality.

At this point, we acknowledge that we have not been able to differentiate between or compare the individual impact of force feedback versus vibrotactile feedback on a grasp and lift task. However, at the same time we understand that oftentimes these are differences that are more evident in cognitive processing than manual task performance. Fortunately, we have brain imaging data accompanying this current experiment that has yet to be fully analyzed. In addition, we acknowledge that we have only compared one type of force feedback, and one specific type of vibrotactile feedback. Yet, our current attempt has been to begin to unravel and understand the effects of haptics feedback in the use of a teleoperated/prosthetic gripper. For indeed if the trends presented in this brief analysis hold true for the larger amputee population, future prosthetic devices would benefit from the incorporation of some form of haptic feedback.

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