Abstract—Despite advances in mechanical design, lower limb prostheses are still limited by their control. Amputees frequently list the desire for more complete control over their devices. Similarly, these devices are limited by their lack of sensory feedback. Bilateral teleoperation of prostheses has the potential to provide direct user control and sensory feedback without having to integrate these design needs into lower limb prostheses. Exoskeletons have recently developed as a technology that allows for forces to reliably be applied to users. Therefore, we propose the design of a wrist exoskeleton for the teleoperation of a lower limb prosthesis. The device measures wrist angle to control the angle of a slave device, and can apply forces to the wrist based on position deviations of the slave. Force feedback was modeled as a virtual wall at the locations of heel and toe contact. To validate the design of our device, and to examine if sensory feedback could be beneficial in teleoperating a slave device, we conducted a study of $N=6$ subjects. We examine if haptic feedback, with or without visual feedback, could improve position tracking in a point-to-point test from -10 degrees wrist flexion to 15 degrees wrist extension. We found that haptic feedback significantly reduced tracking error in the task compared to strictly visual feedback ($p<0.05$), while including visual feedback along with haptic feedback did not significantly affect tracking ($p=0.12$). This experiment indicates that haptic feedback could be useful in teleoperation of lower limb prostheses and it will inform the design of future teleoperation devices.

I. INTRODUCTION

Advanced lower limb prostheses have the potential to alleviate many of the co-morbidities associated with missing a limb, such as reduced mobility, joint problems, high metabolic cost of walking, and increased fall-risk [1], [2], [3]. Active prostheses, which use motors to inject energy to assist the user, have great potential to provide this higher functionality [4], [5]. Despite this potential, most amputees still use passive devices. Further advancements are still needed to make the active devices more ubiquitous.

One of the biggest complaints from users is the lack of intuitive control of their devices. Many active device controllers rely on state information from the device to predict what the prosthesis should do [6]. Some previous work has attempted to include the user in control, whether with or other forms of operation [7]. Other complaints from the user are a lack of haptic feedback that is not provided by the device. Despite some attempts in incorporating such feedback for control of prostheses, much of this work is targeted towards upper limb amputees, which have different needs than lower limb amputees. Feedback on forces on lower limb prostheses could allow users to better modulate ankle torques and improve walking outcomes.

II. PREVIOUS WORK

Active prostheses show potential in improving outcomes for lower limb amputees. Among the most promising of these indicators is the success of a commercial device, the iWalk BiOM. The BiOM uses a combination of passive components and an electric motor for series-elastic actuator configuration to improve performance. Previously, the BiOM was shown to reduce the metabolic cost of walking by 8% and increase self-selected walking speed by 23% for some amputees [9]. Other devices have used pneumatic actuation to reduce the metabolic cost of walking [10]. Despite the demonstrated
successes based on mechanical design advancements of the field, users still seek intuitive forms of control.

In lieu of direct control from the user, researchers have been using cues from the device and the user to try to predict intent and improve performance. Many devices run on simple control models, ranging from strictly angle-based controllers to models of how an intact limb would behave [9]. Some devices use electromyography (EMG), which measures muscle activations in the residual limb. One upper limb myoelectric prosthesis used EMG patterns to try to map desired simultaneous movements [7]. A similar strategy was used to classify EMG patterns in the thigh for control of a knee-ankle prosthesis for sit-to-stand transitions [11]. Similar to these neural signals, mechanical state indicators were used to train a controller to differentiate between intent to do level-ground walking, ascending stairs, or descending stairs [6], they rely on training the device with a large amount of previously collected data, and still lack the intuitiveness and complexity of direct control by the user.

Incorporation of sensory feedback in prostheses shows potential for improving device outcomes and user satisfaction. Recently, skin stretch feedback was used to communicate proprioception of a hand prosthesis [12]. Other methods of haptic feedback, such as vibrotactile and electrocutaneous stimulation, have also been used for proprioception in the upper limb [13], [14]. In the lower limb, a cuff around the calf with pneumatic actuators was used to relay foot pressure information; it is important to note that this study was not conducted with amputees [15]. More recently, pressure feedback was used to assist EMG control of an ankle prosthesis in a unloaded task [16]. These improved control outcomes from sensory feedback motivate the need for haptic feedback in devices designed for teleoperation of prostheses.

III. DESIGN

A. Wrist Exoskeleton

We designed a wrist exoskeleton that can measure the angular position of the wrist and apply torques to the wrist (Fig. 2). We wanted the device to be lightweight, comfortable, and to have high sensor resolution and high actuation capabilities. Therefore, we used an electromechanical actuator since we wanted precise angle measurements and accurate force feedback. Specifically, we chose a 20 Watt brushed DC servo motor from Maxon (model 118754). We placed the motor off board and used a Bowden cable transmission system to reduce worn mass and therefore increase comfort of the exoskeleton. The Bowden cable system was made up of two outer conduits and one inner wire which was wrapped around the drum of the motor and connected to either side of the exoskeleton on top and bottom of the hand. This allowed us to apply torques in either flexion or extension by applying a positive or negative force to the motor. The outer conduits terminated just before the motor drum on one end and just before the rigid linkage on the hand of the exoskeleton. The inner wire was wrapped around the motor drum three times to reduce slipping. The wire was also fed along a constant radius arc on the rigid link on the hand in order to ensure that a linear force was applied to the hand throughout the range of motion.

In order to measure the angle of the wrist, we used a HEDS-5540#A02 Optical Encoder that was mounted on the shaft of the motor; see the Integration section below for further details of the encoder). Lastly, we considered ease of manufacturing in the design by utilizing Birchwood and Duron to be used for the frame of the exoskeleton. We also used 3D printed materials to design the termination points for the outer conduit of the Bowden cable assembly. We used velcro straps to attach the exoskeleton to the forearm/hand of the user.

Fig. 2. Wrist Exoskeleton with labeled components. Wooden components keep the worn mass of the device low.

B. Model of Ankle Prosthesis

For the purpose of evaluating the wrist exoskeleton, we built an ankle prosthesis model to act as the slave in a teleoperation experiment (Fig. 3). This slave device was made from a modified Hapkit with a swapped component to resemble the toe of an ankle prosthesis from the Stanford Biomechatronics Lab [17]. The device uses an electromagnetic motor to actuate a 3D printed toe using a capstan drive. The sector pulley, which was connected to the toe on the other side of the joint, could contact the ground as well. This contact was considered to be the heel of the device. Position of the motor was measured using a magnet in the motor drum and a magnetoresistive (MR) sensor, which is mounted on a custom Hapkit Arduino board.

C. Integration

Both devices were actively controlled using one custom Hapkit board (Fig. 4). Pulse-width modulation (pwm) was used to drive both motors. Angle measurements of the wrist were calculated from the optical encoder on the motor driving the wrist exoskeleton, multiplied by the gear ratio from the wrist exoskeleton to the size of the motor drum. Angle measurements from the slave were calculated from a calibration curve of the Hapkit, based on the readings from the MR sensor on the slaves Hapkit board, connected by a wire.
Fig. 3. Slave device for teleoperation experiments. This device was made from a modified Hapkit, with a 3D printed component to resemble the toe (blue) of an ankle prosthesis. Heel contact was considered when the sector pulley (white) contacted the stand, as shown.

D. Control

A bilateral control architecture was designed for operation of the ankle prosthesis model using the wrist exoskeleton. The slave device tracked the angle of the wrist exoskeleton, scaled from the wrist to the range of motion of the slave by a scaling term $\kappa$. This closed-loop control was done using proportional-derivative (PD) control (1). To provide force feedback to the user, a proportional controller was used to drive the motor of the exoskeleton. This force feedback was based on the concept of a virtual wall. When the commanded position of the slave went beyond either of the slaves contact points, a motor command was sent proportional to the difference between the contact location and the wrist angle (2). This force feedback was only applied at the contact points, not in the free space, to avoid issues with instability and transparency associated with the poor resolution of the slave devices MR sensor.

\[
\begin{align*}
\tau_s &= k_{ps}(\kappa \theta_e - \theta_s) + k_{ds}(\kappa \dot{\theta}_e - \dot{\theta}_s) \\
\tau_e &= k_{pe}(\theta_{wall} - \kappa \theta_e)
\end{align*}
\]

IV. EXPERIMENTAL PROTOCOL

To evaluate the effectiveness of the exoskeleton in controlling position, we wanted to see if users could track a desired position with the slave device. We also wanted to validate the devices ability to apply force feedback. Therefore we designed a position tracking study with three conditions: visual feedback only (VF), haptic feedback only (HF), and both visual and haptic feedback (VHF).

6 subjects total (5 male, 1 female, ages 21-24) were included in the study. Subjects were instructed to use the wrist exoskeleton to move the slave device between two goal positions, 15 degrees and -10 degrees, which corresponded to toe and heel strike, respectively. In the test conditions where haptic feedback was applied, these angles also corresponded to the location of the virtual wall. The subjects were to move in time with a metronome, ticking at 1 Hz; at each tick, the user was to move from one goal position to another. The goal was to have the desired position of the slave device as close to the goal as possible; that is, if the user were to move the wrist exoskeleton beyond the contact point of the toe, this would accumulate position error, despite the toe of the slave stopping at the contact point. Haptic feedback for this study was the virtual wall. If there was no haptic feedback for a condition, no forces were applied. If the user had visual feedback for a condition, they could look at the slave device during the task; if not, they were instructed to keep their eyes closed.

At the beginning of each experiment, the user would put on the wrist exoskeleton, and we would explain the functionality of the device to them. Before the task began, the subject had 1 minute to familiarize themselves with the wrist exoskeleton and try controlling the slave device. The subjects were also introduced to the ticking of the metronome. During this training period, the experimental protocol was explained to them. Following training, the user then conducted the position tracking test in three conditions: visual feedback only, haptic feedback only, and both visual and haptic feedback. The ordering of these conditions was pseudo-random; each subject had a different ordering of conditions, and each possible order of the conditions was tested exactly once. For each condition, the user alternated between goal positions, alternating once per second, for a duration of 30 seconds. During the condition, the angle of the wrist exoskeleton at the time that corresponded to the metronome tick was recorded in Arduino, and transferred to MATLAB at the end of the trial. After all three conditions were tested, the user completed a brief qualitative survey about their perceived difficulty of the task and their preferred feedback condition for greatest accuracy.

Root-mean-square error (RMSE) between the goal positions and measured positions at the time of the metronome was calculated for each of the three conditions. Using RMSE to evaluate position tracking error ensures that an even distribution of overshooting and undershooting the desired
position would not result in an error of zero. An analysis of variance (ANOVA) was used to compare between conditions. For visualization, average measured position at each goal was also calculated.

V. RESULTS

The type of feedback applied significantly affected performance in the experimental task. RMSE values were recorded for each subject and each condition (Table 1, Fig. 5). To understand the distribution of errors among subjects, and see if the errors were biased to overshooting or undershooting the targets, the average angles for each goal location, for each subject and each condition, were calculated (Fig. 6).

TABLE I
RMSE FOR EACH FEEDBACK CONDITION

<table>
<thead>
<tr>
<th>Subject</th>
<th>VF</th>
<th>VHF</th>
<th>HF</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>7.26</td>
<td>1.28</td>
<td>0.58</td>
</tr>
<tr>
<td>2</td>
<td>5.71</td>
<td>0.77</td>
<td>2.75</td>
</tr>
<tr>
<td>3</td>
<td>4.22</td>
<td>0.63</td>
<td>3.47</td>
</tr>
<tr>
<td>4</td>
<td>5.07</td>
<td>0.42</td>
<td>3.09</td>
</tr>
<tr>
<td>5</td>
<td>4.33</td>
<td>3.20</td>
<td>2.99</td>
</tr>
<tr>
<td>6</td>
<td>6.22</td>
<td>2.43</td>
<td>2.86</td>
</tr>
</tbody>
</table>

Across all subjects, the RMSE for the VF condition was 5.48 degrees, the HF condition was 2.62 degrees, and the VHF condition was 1.46 degrees. A one-way analysis of variance (ANOVA) test indicated that the type of feedback was a significant factor that affected RMSE, with a p value less than 0.05 (p=0.00042).

To compare conditions, 2-tailed paired t-tests were calculated between each combination of conditions. The RMSE values for the VF condition were significantly larger than both the VHF condition and the HF condition (p<0.05 and p<0.01, respectively). This shows that including haptic feedback significantly increased performance in the experimental task. Interestingly, there was no significant difference between RMSE values of the HF and VHF condition (p=0.12). This indicates that inclusion of visual feedback did not significantly improve performance.

Responses to the qualitative questions following each experiment were also recorded. Despite having no noise-cancelling devices in the experiment, users reported that the sounds of the device did not consciously affect their control of the device. When asked what they thought which condition led to the most accurate tracking, four of six subjects said the combination of both visual and haptic feedback, while two subjects said they thought visual feedback alone was most effective. Five of six subjects said combination of haptic feedback and visual feedback would translate best to using the device during a walking task. One subject said strictly haptic feedback would be most effective for walking, citing that it would be distracting to watch one’s foot during walking.

VI. DISCUSSION

Our experiment demonstrated that a wrist exoskeleton could be successful in teleoperation tasks. Based on the significant improvement in position tracking as a result of the inclusion of haptic feedback, the device seems to successfully transmit forces back to the user that help them accurately determine the angle of the foot. Although the measured positions were not exactly the same as the goal positions, the average tracking error for the visual and haptic feedback condition seems satisfactory for a lower limb task. It is promising that the combination of haptic and visual feedback was not significantly greater than haptic feedback alone; this indicates that a successful control strategy for lower limb devices could use haptic feedback without requiring the user to be constantly looking at their prosthesis. Qualitative results indicate users are satisfied with their ability to control the slave device. Although our experiment differs from the designed use case of walking, we believe these successes indicate a similar device could be beneficial in teleoperation of a lower limb prosthesis.
Bilateral teleoperation of the device would be more effective if the device had force scaling from the slave to the wrist exoskeleton. This would allow more accurate feedback to the user, and was a common point of feedback from subjects. Force sensing would need to be implemented to reliably provide accurate force feedback. This could be done by incorporating strain gauges or load cells on the exoskeleton on the inner cable of the Bowden cable transmission. In other systems, it is possible to get a good approximation of applied force by measuring the amount of current drawn by the motor, and using the motor constant to calculate the applied torque. For this device, however, that approximation would not hold true. The friction in the Bowden cable of the device is both large and nonlinear, making it difficult to model. Much of the torque of the motor is likely working to overcome this friction. Therefore, any claim about how much force we were applying in absolute terms would be difficult to quantify, without force sensing.

The friction of the Bowden cable transmission also negatively affected the impedance in free-space. One user mentioned that the passive impedance of the device was too large to reliably feel the force feedback when extending the wrist. Reducing the impedance of the device would be necessary to avoid fatigue in subjects during use, especially in longer experiments such as walking trials. Future iterations of the wrist exoskeleton could reduce this impedance by using a different type of transmission than a bowden cable, such as a capstan drive or a friction drive.

VII. CONCLUSIONS

In summary, we designed and built a wrist exoskeleton to teleoperate an ankle prosthesis with position tracking from the exoskeleton to the prosthetic, as well as force feedback from the prosthetic to the exoskeleton. In addition, we ran an experiment that tested a user’s ability to move the prosthetic to a specific location with visual feedback, haptic feedback, and a combination of visual and haptic feedback. We saw a statistically significant improvement in accuracy with haptic feedback over strictly the visual feedback. However, between purely haptic feedback and a combination of haptic and visual feedback, we did not see statistically significant improvement in accuracy. In the future, we will redesign the wrist exoskeleton to utilize a capstan drive rather than a Bowden cable assembly, since we found that the high levels of friction in the Bowden cable were too much of a drawback to justify the lower worn mass of the design. Additionally, we intend to make our second iteration more compact and more compatible with multiple forearm/hand sizes. Following a second design, we will implement teleoperation with an ankle prostheses for walking experiments.

REFERENCES


