Investigation of Rotational Skin Stretch for Proprioceptive Feedback with Application to Myoelectric Systems

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***Abstract*— We present a new wearable haptic device that provides a sense of position and motion by inducing rotational skin stretch on the user's skin. In the experiments described in this paper, the device was used to provide proprioceptive feedback from a virtual prosthetic arm controlled with myoelectric sensors on the bicep and tricep muscles in 15 able-bodied participants. Targeting errors in blind movements with the haptic device were compared to cases where no feedback and contralateral proprioception were provided. Average errors were lower with the device than with no feedback but larger than with contralateral proprioceptive feedback. Participants also had lower visual demand with the device than with no feedback while tracking a thirty degree moving range. The results indicate that the rotational skin stretch may ultimately be effective for proprioceptive feedback in myoelectric prostheses, particularly when vision is otherwise occupied.**

***Index Terms*— Haptic Displays, Myoelectric Prostheses, Proprioception.**

I. Introduction

RECENT advances in engineering and medicine have provided more sophisticated upper-limb prostheses as well as improved means of controlling them. However, most upper-limb prostheses lack direct sensory feedback. While still not approaching the richness of human sensory systems, the current state of technology is such that prostheses could be equipped with a variety of sensors to measure position and motion, forces, temperature and vibration. A more challenging problem, however, is to relay this information back to the amputee. Promising research is being done in the field of neural prosthetics that may allow direct stimulation of neurons in the central or peripheral nervous system (e.g. [1], [2]). However, these approaches are invasive, and more research and development will be needed before this becomes a viable option.

Wearable haptic displays can provide a non-invasive approach to communicating information from sensors in a prosthetic limb to the amputee. As early as the 1960's, a number of researchers were evaluating tactile displays to provide users a sense of grip force in myoelectric hands. These included cutaneous [3]–[5] and implanted [6], [7] single electrode displays as well as vibrotactile displays [8]. Mann [9] implemented a vibrotactile display of position proprioception in a myoelectric elbow joint. The prevailing opinion at this time was that the feedback would be beneficial if not critical for the advancement of myoelectric prostheses. While these studies found benefits to the feedback, it was not widely adopted clinically. It is not clear why amputees and clinicians did not respond to this concept, but there are a number of potential contributing factors. First, the feedback modalities used can be subjectively unpleasant [10], [11]. There may also have been some aversion to adding complexity in the prosthesis without providing a more significant functional benefit. Most traditional myoelectric prostheses have high-impedance actuators in the joints which, coupled with the noise in the myoelectric signal (MES), may limit the user's ability to control force even if they have some afferent feedback. Amputees also have some limited sense of force in the prosthesis through vision (deformation) and reaction forces at the socket interface. Recently, as more sophisticated arms and hands are being created, haptic force feedback is again being evaluated, and may be better accepted clinically as amputees adopt more technologically advanced prostheses. Cipriani et al. implemented vibration feedback of grip force in an anthropomorphic hand that was myoelectrically controlled and found qualitative benefits in unimpaired subjects [12].

Less work has focused on haptic proprioceptive feedback for prostheses. As there does not seem to be a consistent definition of proprioception (or kinesthesia) in the literature, we will use the term proprioception to describe the gross perception of relative position and motion of the body segments not due to vision, touch or the organs of equilibrium [13]. In the absence of proprioception, amputees can rely on vision to position the arm. When visual feedback is not available, users can potentially rely on internal models of the arm and controller to perform some tasks with open-loop strategies. This has led some to question the value of proprioceptive feedback in prostheses (e.g. see [14]). In fact the role of proprioceptive feedback in human motor control is not completely understood. The substantial delays associated with the feedback (50-250ms, depending on the modality [15]) indicate that it is likely not being used for real-time control for most motions at preferred speeds. Many motions seem to be performed in feed-forward fashion based on an internal model of the body [15]. How this idea translates to myoelectric prostheses, where arm dynamics and control parameters are quite different (along with typical movement speeds) is also not well understood. However,

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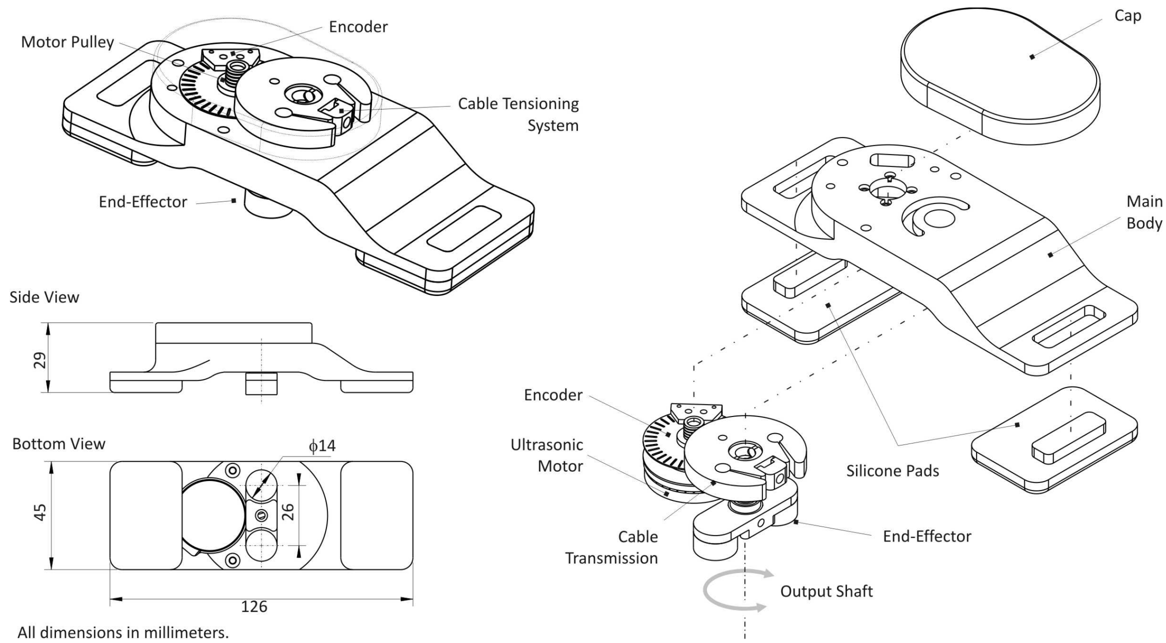

Fig. 1. Wearable skin stretch device: assembly layout and main dimensions.

there is evidence suggesting that feedback would be useful. Proprioceptive or visual feedback could be used to construct an internal model of the system, which could improve long-term performance. Visual servoing consumes a large portion of the user's visual capacity that could be devoted to other simultaneous tasks. High visual demand has been cited as a drawback of myoelectric arms by amputees [16]. Additionally, some benefits of proprioception have been shown even when vision is present [17]. The addition of proprioceptive feedback creates a closed-loop system that may make the prosthesis a more intuitive extension of the user. This is evidenced by the fact that body-controlled prostheses (including some with external power, e.g. [18]–[22]) have proven popular in part due to the fact that they take advantage of the principle of extended physiological proprioception (EPP) as first described by Simpson [23]. Myoelectric prostheses can provide more intuitive and less cumbersome control schemes but lack this element of feedback, which some amputees have cited as a reason for rejection of the prosthesis [24]. In the present studies, we aim to show quantitative benefits of proprioceptive haptic feedback in blind targeting accuracy and a reduction in visual demand for a continuous movement task. The visual demand measurement is an attempt to quantify how often a user must glance at the arm in order to complete the task.

In the present work, we use a new wearable haptic display (Figure 1) that produces rotational skin stretch on the non-glabrous skin. The use of this type of feedback for displaying position and motion information is motivated by findings that skin stretch contributes to kinesthesia at a number of joints [25]–[27]. We therefore hypothesize that the feedback is more intuitive than vibrotactile or electrotactile information for conveying joint position and velocity, even if the feedback is not physiologically accurate. In addition, this type of feedback can simultaneously display position and motion and is bi-directional (it can rotate in two directions). In previous studies, we compared a benchtop version of the skin stretch device used in this paper to a single vibration tactor and no feedback in a blind closed-loop positioning task of a virtual object controlled with a force sensor. Participants had smallest position and velocity errors with skin stretch feedback [28]. In that study, we found it necessary to give the virtual object non-trivial dynamics in order to show any benefit to the relatively coarse feedback provided by either vibration or skin stretch. If the object had simple mass and damping, subjects quickly learned open-loop strategies that allowed accurate positioning. For myoelectric prostheses, we hypothesize that the inconsistent and noisy characteristics of the MES, coupled with the nonlinear joint dynamics (due to static friction), will make open-loop strategies less effective and the haptic display more beneficial.

We have also recently evaluated some psychophysical properties of the skin stretch device [29]. We found that inexperienced subjects had occasional difficulty interpreting the position of the device when it was moved autonomously, with direction errors being one of the primary symptoms. However, when users could control the motion of the device, they were able to reach targets with much greater accuracy. This result suggests that the device is most beneficial in applications where there is a closed-loop connection between the afferent sensation produced by the display and an efferent command generated by the user. The current application is an example of such a situation.

In the following sections, we evaluate the wearable skin stretch device with unimpaired subjects controlling a virtual prosthetic elbow joint with EMG electrodes on the bicep and tricep muscles. The experimental setup approximates a transhumeral amputation as the elbow joint is effectively fixed by having participants hold a stationary handle with their hand.

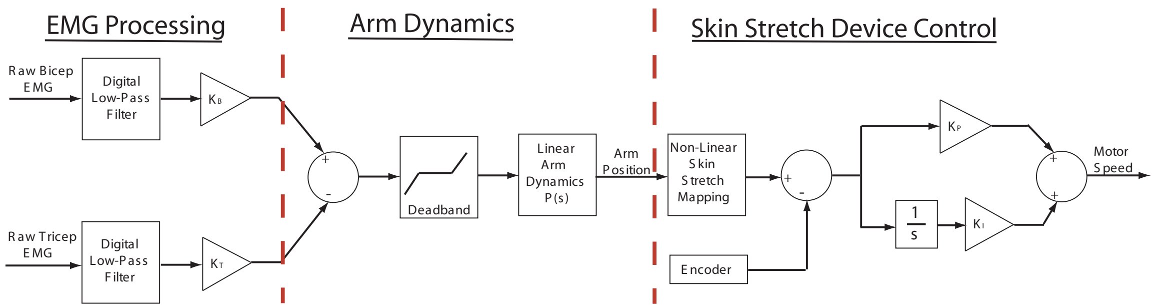

Fig. 2. Schematic representation of the primary control elements in the system.

We use a blind targeting task to compare their accuracy with the haptic device to the cases where no feedback is provided (open-loop movements) and where proprioception is provided on the contralateral arm. These two comparison cases establish what we consider the worst (no feedback) and best (contralateral proprioception) afferent cases for blind targeting while maintaining the salient efferent characteristics of myoelectric control. In addition, we compare the visual demand required to roughly position the virtual elbow continuously in a visual occlusion study with skin stretch compared to no feedback.

II. Methods

In this section, we first describe the skin stretch device used in the present studies. We then describe the control of the virtual arm, including EMG data processing steps and the control of the feedback devices. Finally, we describe the calibration procedure and the experimental protocol for two user studies.

A. Feedback Device

The haptic feedback evaluated in this study was applied as rotational skin stretch, using a rotating end-effector with two circular contact pads (d = 1.4 cm) spaced 2.6 cm apart, attached to the skin with double-sided tape. The pad dimensions, spacing, contact pressure and range of rotation are based on the results of experiments performed with an earlier, highly adjustable benchtop device [28]. The new version of the device can be strapped to a user's arm or leg using velcro belts (Figure 1). It is designed so that the end-effector pads depress the skin 1-2 mm, resulting in a normal force of 1-2 N.

In most applications we envision the device being used in position control mode rather than with a controlled torque. Therefore, a non-backdriveable system is preferred as it does not require continuous power to hold a fixed orientation. In the wearable device, rotation of the end effector is produced by an ultrasonic motor (Shinsei Motors, USR30-B3). The advantages of this actuator for the present application are that it is lightweight, is not backdriveable and produces a relatively high torque for its size (Table I) [30]. For the purposes of testing, it has the additional advantage that it produces no vibrations in the frequency range that humans can perceive. When it is used in combination with a smooth capstan/cable transmission with a 6:1 speed ratio, there are no vibrations that could confound the effects of skin stretch.

The body of the device was manufactured using Shape Deposition Manufacturing [31] to allow materials of various properties to be integrated (i.e. soft pads for skin contact and rigid structures for mounting the mechanical components). An unpackaged optical encoder provides position feedback. The design requirements and specifications of the device are shown in Table II.

TABLE I
Ultrasonic Motor Specifications

	Motor specifications
Diameter	30 mm
Thickness	9 mm
Maximum torque	0.1 Nm
Holding torque	0.1 Nm
Weight	20 g
Maximum speed	250 rpm
Minimum speed	30 rpm
Driving frequency	50 kHz
Excitation voltage	12 V
Max dynamic current	0.5 A
Max static current	20 mA

TABLE II
Wearable Skin Stretch Device Characteristics

	Design Requirements	Device Specifications
Size	small	29 x 45 x 126 mm
Max Torque	0.2 Nm	0.6 Nm
Speed Range	< 200 deg/s	15-150 deg/s
Weight	< 200 g	115 g
Sensor Resolution	0.5 deg	0.05 deg

B. Control System

We used a simple controller that captures the salient characteristics of many myoelectric prostheses and was easy for the subjects to learn to use. Figure 2 shows a schematic representation of the primary control elements of the experimental setup. The control elements can be divided into three sub-sections; EMG data collection and processing, virtual prosthetic arm dynamics, and control of the skin stretch device. Each of these is discussed in the following subsections. Matlab's xPC realtime target was used to perform the data collection and closed loop control at 1 kHz.

1) EMG Data Processing: Differential EMG electrodes with pre-amplification (Delsys 2.1) were used to collect the

MES from the bicep and tricep muscles. A second stage handheld amplifier with gain of 1000 (Delsys Bagnoli 2) was used for further signal amplification. Each EMG channel was sampled at 1 kHz, the DC bias was subtracted (found during an initial calibration procedure described below), and a low-pass Butterworth filter with a cut-off frequency of 6 Hz was applied. Gains were then applied to each channel before sending the signal to the virtual arm. These gains varied between subjects and were determined as part of the calibration procedure.

2) Virtual Prosthetic Arm Dynamics: After the gains were applied to the filtered MES, the tricep signal was subtracted from the bicep signal to create a virtual torque that was sent to the arm. This virtual torque is analogous to the current sent to the motor in a prosthetic limb (multiplied by the torque constant of the motor). A deadband was used to simulate static friction in the arm so that it was necessary to apply a minimum torque of 0.5 units for the arm to move. The torque was saturated at 5 units. Units are defined such that one unit of torque produces an angular acceleration of one rad/sec^2 when applied to one unit of inertia (in the absence of friction).

After applying the non-linear elements, the torque was sent to the linear dynamics of the virtual arm, which were modeled as a simple inertia and viscous damping such that the transfer function of arm position θ to torque T is:

$$\frac{\theta(s)}{T(s)} = \frac{1}{Js^2 + bs} \tag{1}$$

where J is the inertia of the arm and b is the viscous damping. In the present case we used values of $J = 0.5$ and $b = 2.5$. On the virtual display, an arm angle of zero was represented by a horizontal line. The arm position was limited to $\pm$ 60 degrees, corresponding to a total of 120 degrees of elbow flexion.

3) Skin Stretch Device Control: For the experimental trials where skin stretch feedback was provided, our intent was to provide the user with an absolute sense of the virtual arm's position. When the arm's position was zero (horizontal), zero skin stretch was provided, and when it was at its limits (plus or minus 60 degrees), the skin stretch device was also rotated to its maximum value, which was limited to $\pm$ 40 degrees in this case. However, in pilot studies we found that a linear mapping of arm position to skin stretch resulted in difficulty in detecting positions around zero. We therefore implemented a nonlinear mapping by fitting a fifth order polynomial that was constrained to pass through zero and the endpoints, had a slope of 1 near the origin and a slope of $1/2$ near the limits (Figure 3). While more work remains to determine an optimal mapping, pilot subjects were better able to use the nonlinear mapping than a linear one in closed-loop targeting studies. Any future improvements in the device design and mapping should improve performance compared to the results presented here.

After the desired position of the skin stretch device was determined by evaluating the polynomial in Figure 3 at the current virtual arm position, it was compared to the encoder reading of the device to calculate the position error. A proportional plus integral (PI) controller was then used to set the speed of the motor based on this position error. The proportional gain (K_p) was 0.75 and the integral gain (K_i) was 0.1. A small driver board provided by the manufacturer (Shinsei Motors, D6030) was used to control the motor. In addition to the analog speed channel, which was specified as described above, two digital lines were used to control the direction of rotation. Due to these direction changes and the fact that the motor has a minimum speed at which it can rotate, determined by the resonance of the piezo elements, chatter around the desired position was sometimes encountered when the arm moved very slowly. We therefore implemented a deadband of 0.1 degrees on the position error. The position error due to the controller was always less than 0.8 degrees.

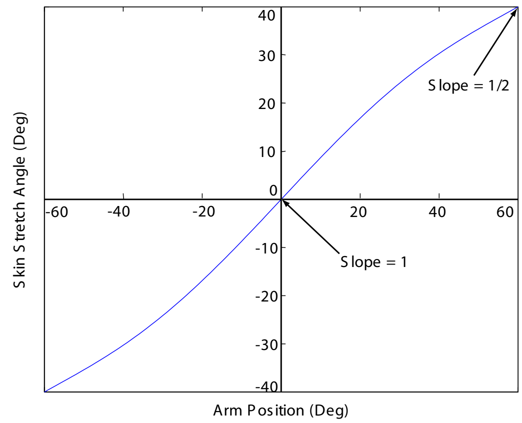

Fig. 3. Non-linear mapping of virtual arm angle to skin stretch rotation. Mapping is a fifth-order polynomial with a slope of one near the origin and $1/2$ near the limits.

C. Experimental Setup

1) Setup Overview: The experimental setup is shown in Figure 4. The participant was seated in front of a computer display. EMG electrodes were placed on the right bicep and tricep muscles near the center of the muscle body with the differential electrodes aligned approximately with the fiber direction. A reference electrode was placed on the back of the right hand. The skin stretch device was placed on the back of the right upper arm, proximal to the elbow, near the tricep electrode. The participant gripped a handle with their right hand. A button was placed on the handle so it could be easily pushed with the thumb. The setup was designed to simulate a transhumeral amputation as the elbow joint was effectively fixed. Participants could produce the MES by pulling up or pushing down on the handle. Pulling up approximates the bicep acting in isolation, and pushing down approximates the tricep in isolation.

In the no feedback case, the skin stretch device was turned off. In trials where contralateral proprioception was provided, the left arm of the participant was placed in a single degree of freedom manipulandum that could position the elbow joint at a desired angle in the horizontal plane. The elbow angle was set to match the angle of the virtual arm with zero degrees of elbow flexion corresponding to minus 60 degrees of rotation of the virtual arm, and 120 degrees of elbow flexion corresponding to 60 degrees of rotation of the virtual arm (Figure 5).

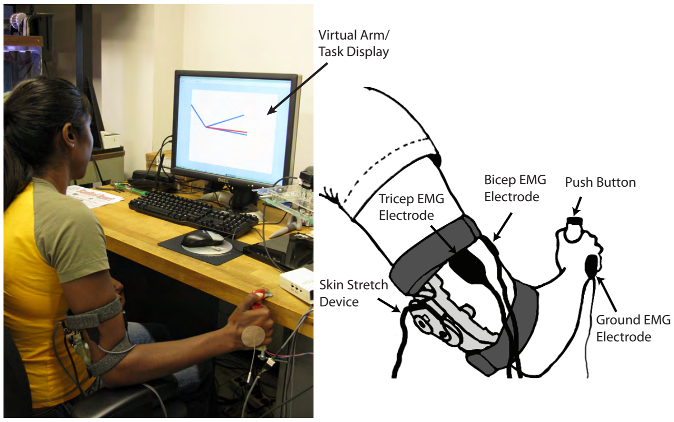

Fig. 4. Experimental setup. Right pane shows schematic view of the primary input and feedback elements.

The elbow manipulandum used a DC motor (Maxon RE025) with a 13.5:1 capstan cable drive transmission. A PID current controller on the desired elbow angle was implemented that produced dynamic tracking errors less than 2.0 degrees and static position errors of less than 0.21 degrees in all trials. A screen was placed above the manipulandum so the arm motion was not visible to the subject.

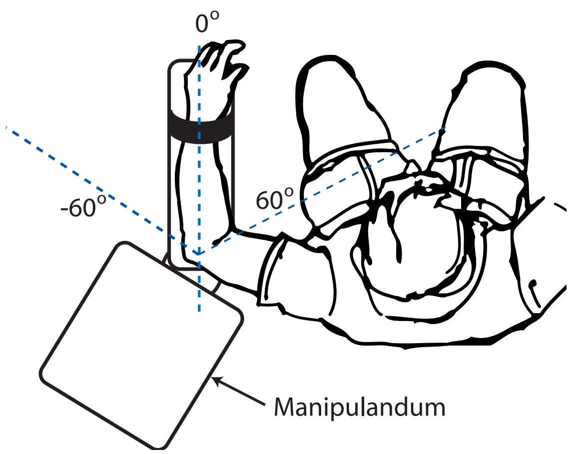

Fig. 5. Schematic view of elbow manipulandum used to provide contralateral proprioceptive feedback.

2) Calibration: Prior to the experiment, the participants underwent two phases of calibration. First, they were asked to grip the handle but otherwise relax the arm. EMG activity from both muscles was recorded for five seconds, and the mean of this resting data was subtracted from each channel during later trials. Second, the participant was asked to pull up and push down on the handle while EMG signals were recorded. They were not instructed to use maximal force but a large force that would be comfortable to apply during the experiment. The respective gains for the bicep and tricep muscles were calculated by dividing the maximum filtered EMG signal recorded during the calibration by a constant, in this case 5. The participants were then allowed to move the virtual arm on the screen with visual feedback to determine if the calibration was satisfactory. The calibration was considered satisfactory if the participant had some control of the speed of the arm in both directions and approximately equal effort was required to move the arm in both directions. The calibration was repeated as necessary until the participant was satisfied with the controller. This took no longer than four attempts. In a few cases during the experiment, some drift was encountered due to a change in the DC bias. In such cases the calibration was repeated as needed.

3) Participants: Fifteen volunteers participated in the studies. Nine were male, six female, ages 19-49, mean age 27.3. Nine of the fifteen participants (6 male, 3 female) completed the targeting trial with contralateral proprioception. These trials were added to the study after six participants had already completed the skin stretch and no feedback trials in an attempt to provide a best-case baseline for performance. The studies took a total of 45-120 minutes to complete, including setup and calibration. The study was approved by Stanford's Institutional Review Board.

D. Targeting Study

The subjects first participated in a targeting accuracy study where no real-time vision feedback was provided. Prior to each set of trials a brief training period was provided. Each participant was allowed to move the arm through the workspace with visual feedback for two minutes and then was given 10 practice trials, identical to the experimental trials. In the no feedback case, the training allowed them to learn open-loop strategies, and in the haptic and contralateral proprioception feedback cases it allowed them to learn the mapping of arm position to feedback sensations. In each experimental trial, the starting position of the arm was displayed on the screen chosen from [-60,0,60] degrees. An instruction to move $\pm$ 30, 60 or 90 degrees was also displayed. Movements were constrained such the desired ending position was either -30, 0 or 30 degrees. This resulted in a total of eight starting/ending combinations, each of which was repeated four times in random order for a total of 32 trials per participant. A sample starting screen is shown in Figure 6.

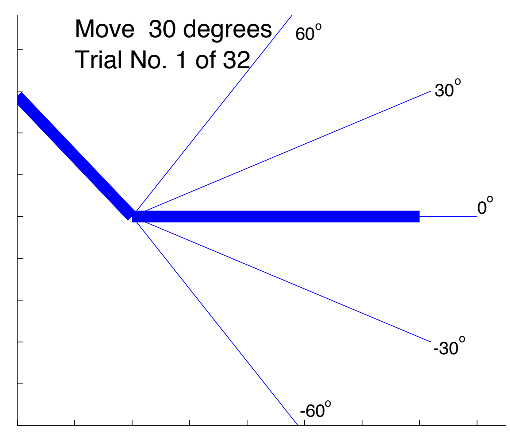

Fig. 6. Visual display of virtual arm at beginning of a sample trial with starting position equal to zero degrees. The thick diagonal line represents the upper arm and does not move.

After this screen appeared, the participant moved the virtual arm without seeing the motion of the arm on the screen. The participant was instructed to press the button when they felt the target was reached. After the trial, the participant was shown the actual and desired ending position of the arm. The 32 trials were done with both skin stretch feedback and no feedback. Seven participants did the no feedback trials first; eight did the skin stretch trials first. The nine participants who completed the trials with contralateral proprioception did these last to avoid biasing the primary cases of interest (skin stretch and no feedback).

E. Visual Occlusion Study

In the second study we assessed participants' visual demand with and without the skin stretch device using the visual occlusion method [32], [33]. In this case, the task was to keep the virtual arm in a desired range that moved in a quasi-random fashion. Specifically, the participant was instructed to keep the virtual arm in a 30 degree window, displayed on the screen as two lines that moved together (thick outer lines in Figure 7).

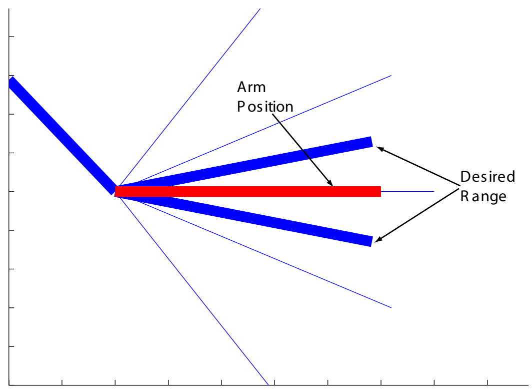

Fig. 7. Visual display of virtual arm at beginning of visual occlusion trial. Thick outer lines represent the desired range, thick horizontal line in the center is the virtual arm which is only visible for one second after a button press.

The desired arm position range was generated by creating a desired position and drawing lines (limits) at $\pm$ 15 degrees from this position. The desired position was determined in pilot trials such that the task was relatively easy with constant visual feedback and very difficult without. A sum of 11 sine waves that appeared random to the participants was chosen with amplitudes and frequencies based on guidelines for human tracking in [34] and [35]. The frequencies of the sine waves (ω_k) were relative prime multiples of the fundamental frequency 0.003 Hz using the prime multipliers 2, 3, 5, 7, 11, 17, 23, 37, 59, 87, and 131. The amplitudes (A_k) were defined with an exponential function of frequency as follows

$$A_k = 0.3 \sum_{k=1}^{11} e^{-0.12(k-1)} \tag{2}$$

The scaling (0.3) and decay rate (0.12) values were determined such that sufficient coverage of the workspace was generated while not exceeding the range. The phase angles of the sine waves were randomized before each trial to prevent the participants from memorizing the inputs. A sample of the desired range and the virtual arm's position for a trial are shown in Figure 8.

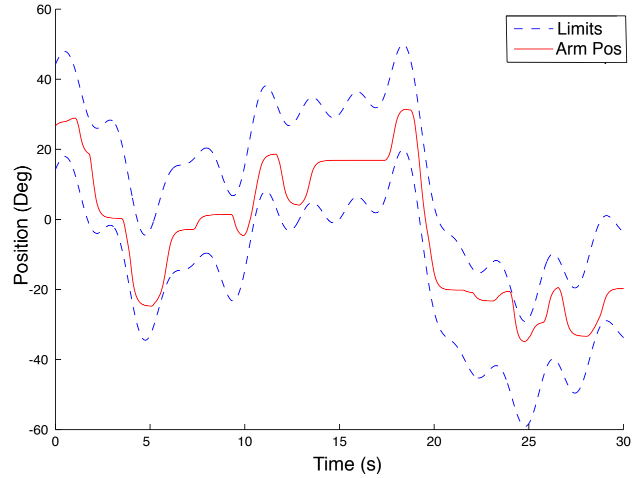

Fig. 8. Virtual arm position and task position limits for 30 seconds of a sample trial with skin stretch feedback.

Visual feedback of the virtual arm's position was provided for one second each time the participant pressed the button on the handle. Participants were instructed to press the button whenever they felt visual feedback was required to complete the task. If the arm left the desired range, the trial ended and restarted. The trial was successful if the participant kept the arm in the desired range for the duration of the trial (40 seconds). Trials were repeated until the participant had three successful trials for the no feedback and skin stretch cases. The order of the trials (no feedback and skin stretch) was randomized. Prior to the experiment, participants were given one full practice trial with constant vision feedback for each feedback case. Over the course of the trials (including failures) participants were able to determine an appropriate strategy of button press frequency that allowed them to complete the task.

For each successful trial the average visual demand was calculated. The instantaneous visual demand (VisDem) can be calculated at each button press using the following relation

$$VisDem = \frac{1.0}{t_i - t_{i-1}} \tag{3}$$

where the numerator is the duration of the feedback (1 sec.) and t_i and t_{i-1} are the times corresponding to the current and previous button presses. The average visual demand is calculated by averaging these values over the trial, or equivalently, by dividing the number of button presses by the duration of the trial. The average visual demand is the fraction of time that visual demand was provided over the course of the trial. It is a measure of how often the participant felt they needed to glimpse at the arm to complete the task. In the data analysis, the average over all three successful trials was used to compare the haptic and no feedback cases. As a secondary measure, the number of task failures before completing the three trials was also recorded.

III. Results

A. Targeting Study

In the targeting study, the primary measure of interest is the average targeting error (across the 32 trials). Only absolute values are reported as the errors were approximately symmetric (i.e. subjects were not consistently under- or over-shooting the target). A box plot of the average position errors for each of the participants is shown in Figure 9. Repeated measure analysis of variance (ANOVA) with pairwise comparisons was performed for this performance measure. For all 15 subjects a repeated measure ANOVA with two levels (no feedback and skin stretch) was performed. Significant main effects were found due to both feedback condition, $F(1, 14) = 36$, $p = 3.3 \cdot 10^{-5}$ and participant $F(14, 14) = 544$, $p = 1.3 \cdot 10^{-12}$. Errors with skin stretch were significantly lower compared to errors with no feedback with $p = 3.3 \cdot 10^{-5}$. For the nine subjects who participated in the contralateral proprioception feedback case a separate repeated measure ANOVA with three levels was performed. Significant main effects due to feedback condition, $F(2, 8) = 29$, $p = 6.6 \cdot 10^{-4}$, and participant $F(8, 8) = 57$, $p = 6.8 \cdot 10^{-5}$ were again found. Post-hoc pairwise comparisons with Bonferroni correction found significant differences between all feedback cases with skin stretch lower than no feedback with $p = 0.015$ and contralateral proprioception lower than skin stretch with $p = 2.9 \cdot 10^{-3}$.

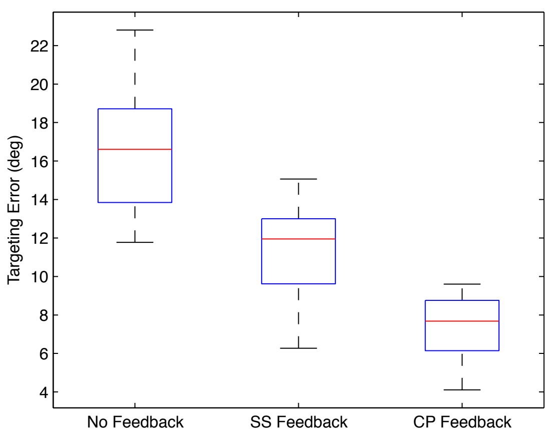

Fig. 9. Box plot of average targeting errors for participants with no feedback and skin stretch (SS) feedback (n=15) and contralateral proprioceptive (CP) feedback (n=9).

While variations between participants were significant, trends were consistent across participants as all 15 had lower average errors with skin stretch than no feedback, and all nine participants who completed the contralateral proprioception trial had lowest errors in that case.

B. Visual Occlusion Study

The primary measure in the visual demand study was the average visual demand over the three successful trials for each feedback condition. A box plot of these data is shown in Figure 10. While there was significant participant-to-participant variation in the visual demand in both feedback cases, participants generally had lower visual demand with the haptic device. With skin stretch feedback, the average visual demand was reduced 23 percent (from 0.53 with no feedback, to 0.41). A repeated measure ANOVA was performed for average visual demand over the three trials. Significant main effects were found due to both feedback condition, $F(1, 14) = 31$, $p = 7.2 \cdot 10^{-5}$ and participant $F(14, 14) = 184$, $p = 1.9 \cdot 10^{-9}$. Visual demand was significantly lower with skin stretch compared to the no feedback case with $p = 7.2 \cdot 10^{-5}$.

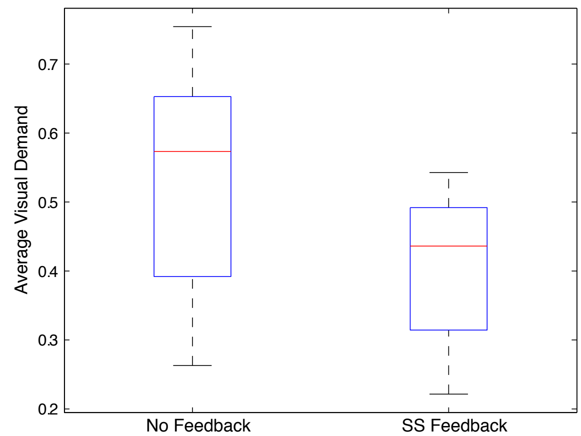

Fig. 10. Box plot of average visual demand over three successful trials for all 15 participants with no feedback and skin stretch (SS) feedback.

Participants also had fewer failures before completing the three trials with skin stretch feedback. The mean number of failures before completing the no feedback trials was 12.2 (range 5-24), and with skin stretch the mean was 8.2 (range 2-21). The difference in means was significant in a paired t-test with $p < 1 \cdot 10^{-6}$.

IV. Discussion

Participants performed better in both user studies with the skin stretch device than with no feedback. This indicates that the device may have value as a proprioceptive feedback device in myoelectric prostheses. When no feedback was provided, participants were forced to rely on open-loop strategies for controlling the arm's position. Presumably due to the non-linearities in the virtual arm model and the noisy nature of the EMG signal, this was difficult, making even a relatively coarse sense of position feedback useful. Targeting errors with skin stretch were approaching, but still significantly larger than errors with proprioceptive feedback from the contralateral arm, which we consider to be the best case scenario for blind targeting as it relies on intact physiological sensors while maintaining the other aspects of the experiment. In addition to reducing targeting errors, skin stretch feedback reduced the visual demand required to track a moving angular window. The lower visual demand indicates that providing a rough sense of position could allow the arm to be used (if position accuracy is not critical) while vision is devoted elsewhere, making the prosthetic arm a more natural extension of the person. The addition of haptic feedback could effectively close the control loop and require only occasional glances from the user, reducing the need for visual attention.

In both studies, only a short training period was provided. Additional training may improve performance with no feedback and with the haptic device; however, we expect minimal training benefit in the contralateral proprioception case. Though we did not directly assess the effects of training as uniform training was provided in all cases, a simple regression analysis was performed to determine if subjects' performance improved over the time. In the targeting study, we fit a regression line to the absolute value of the errors versus trial number across all participants and found a significant negative slope for the skin stretch case ($p = 0.034$), providing evidence that errors decreased as the trial progressed. However, when no feedback is provided, the slope is not significantly different from zero. Individually, four of fifteen participants had significant negative slopes with skin stretch while two had significant slopes with no feedback. While future studies are needed to evaluate the effects of training, this indicates that participants could improve with more training, particularly with the haptic device.

While the device used in this study is wearable, it is not yet portable. We are developing devices that are smaller and lighter, and that can be worn on more parts of the body. We envision future devices that include on-board power, processing and wireless communication for portable use. All of the primary components of the device, including the motor, driving electronics, and sensors are amenable to portable use. The device consumes power only while changing its position. For powered prosthetic arms, it may be possible to integrate the device into the electronics of the limb, minimally increasing the size, weight and power consumption. Placement of the haptic device should be customizable depending on the wearer, as with most prosthetic components. In some cases, it may even be possible to integrate the device into the prosthetic socket directly, eliminating the need for straps. In the studies described here, in which the user wore the device for no more than 2 hours, negligible slipping between the device and the skin was observed. Future studies will evaluate whether the current attachment method is adequate for longer term use (i.e. a full day). Due to the relatively small amount of rotation applied in the study, subjects did not complain of skin irritation or fatigue despite nearly continuous use for up to 30 minutes at a time. While increasing the range of rotation may improve performance, it is unclear how this will affect comfort. We expect that each user will have a different preferred range.

This paper has presented a proof of principle for rotational skin stretch as a means of providing proprioceptive feedback in applications, such as prosthetic limbs, where it is otherwise absent. The results of the experiments indicate that rotational skin stretch can be used for closed-loop positioning in vision-impaired circumstances. Looking ahead, we acknowledge some differences between the experiments described in this paper and clinical applications, which may affect the performance of skin stretch device. First, in amputees both the MES and feedback properties may differ from unimpaired users. When the muscles on the residual limb are activated, more significant conformal changes may result, which could affect the performance of the skin stretch device in some cases. While subjects in this study did not have a gross sense of joint motion, it is possible that they had some low-level afferent feedback from tendons and other structures that amputees would lack. The effect of these factors is unclear and will be evaluated in future studies. Another difference between the present case and the proposed application is the virtual arm and controller dynamics. These were not specifically designed to match any specific prosthetic arm but were intentionally simplified so that participants could learn to use the virtual arm with minimal training. In reality, prosthetic arm dynamics would likely include more significant non-linearities, control irregularities, and exogenous disturbances that would make the arm more difficult to control in an open-loop fashion (which would increase targeting errors in the absence of feedback). We conjecture that the addition of proprioceptive feedback would be more useful when controlling an arm with more realistic dynamics. The fact that it was beneficial even with these relatively simple dynamics (while still maintaining the basic properties of the MES) is encouraging. There is also the potential for improving the quality of the feedback. The device configuration and mapping used in this study were effective but likely non-optimal. We are evaluating potential improvements in these areas that may increase performance.

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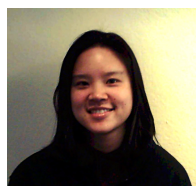

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