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Design and Integration of an Inexpensive Wearable Mechanotactile Feedback System for Myoelectric Prostheses

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ABSTRACT The aim of this paper was to demonstrate the functionality of an inexpensive mechanotactile sensory feedback system for transhumeral myoelectric prostheses. We summarize the development of a tactile-integrated prosthesis, including 1) evaluation of sensors that were retrofit onto existing commercial terminal devices; 2) design of two custom mechanotactile tactors that were integrated into a socket without compromising suction suspension; 3) design of a modular controller which translated sensor input to tactor output, was wirelessly adjusted, and fit within a prosthetic forearm; and 4) evaluation of the system with a single transhumeral participant. Prosthesis functionality was demonstrated over three test sessions; the participant was able to identify tactor stimulation location and demonstrated a reduction in grasp force with the mechanotactile stimulation. This system offers an inexpensive and modular solution for integration of a mechanotactile sensory feedback system into a prosthetic socket without compromising the suction seal. These principles can be applied in future studies to investigate the direct impact of sensory feedback on tangible outcomes for prosthetic users, thereby reducing barriers to clinical translation.

INDEX TERMS Medical robotics, prosthetics, haptic interfaces, force feedback.

I. INTRODUCTION

Upper limb loss can significantly reduce quality of life, leaving an amputee feeling less capable and independent. Myoelectric prosthetic devices are becoming more advanced, with major developments in control and functionality [1]. However, up to 23% of myoelectric prostheses are rejected, with sensory feedback often cited by amputees and clinicians as a desirable improvement [2], [3].

Various methods have been studied to provide sensory feedback to prosthetic limb users. The most commonly measured sensory inputs acting on the prosthesis include normal force (perpendicular to surface), shear force (slip), position, and temperature [4]. These inputs are translated to the user through one or more modalities including mechanotactile, vibratory, electrical, or auditory, where mechanical stimulation is delivered to the user via a device termed "tactor" [5]–[7]. Although mechanical vibration (vibrotactile) feedback is one of the most commonly used non-invasive feedback methods [8], matching the modality of the stimulus sensed on the prosthetic hand is thought to reduce the cognitive burden on the prosthetic user [5]–[7]. In-lab studies have demonstrated that sensory feedback may improve grip control [9]–[11], improve task performance [12], [13] especially in complex tasks [14], reduce error rates during movement and grasping tasks [15], [16], improve training of grasping force [17], increase embodiment [18], and reduce phantom pain [19]. However, others have suggested that laboratory conditions may not be directly applicable to real-life activities, as most research devices do not emulate realistic noisy conditions such as end-effector loading and socket fit challenges [7], [20], [21]. It is therefore recommended that sensory feedback systems be evaluated using fully fitted sockets closely resembling the systems that would be used in daily life [7], [20].

In an attempt to address these issues, researchers have developed low-cost wearable feedback systems. In the case of transradial amputations, systems have been developed to provide vibrational feedback against the skin in response to forces measured on the terminal device [22]–[24]. Tactors have been placed proximal to the prosthetic socket; in two studies the tactor was placed underneath the prosthetic liner [23], [24] and in one it was secured within an elastic cuff [22]. These studies have demonstrated potential for reducing grasp force [24], improving performance [22], and improving grip force accuracy at low force levels [23]. In a recent study [25], a wearable transradial mechanotactile feedback system held a lever-arm against the skin using a stretchy cuff proximal to the prosthetic socket. This allowed the participant to handle disposable glasses without visual or auditory cues.

Although these results are promising, further research is needed to better understand the impact of incorporating sensory feedback devices into fully-functioning prostheses. This impact is especially relevant in the case of transhumeral prostheses, where additional loss of the elbow joint further separates the user from indirect sensory clues coming from the prosthetic hand. Mounting tactors proximal to the prosthetic socket, as done with transradial systems, is generally not feasible for transhumeral prostheses due to the presence of the rigid socket covering the residual upper arm. Tactors would need to be integrated directly onto the prosthetic socket; however, placement is limited by the need to accommodate other components (such as electrodes) and consideration must be given to avoid compromising socket suspension or fit. Some desktop experiments have mounted tactors to a prosthetic socket with access to the user's limb through a hole [9], [26], [27], however these studies focused on liner-based tactor integration and did not assess the impact on socket fit. As transhumeral myoelectric prosthetic sockets may use suction as an adjunct suspension [28], holes in the socket could compromise the suction seal and result in slippage of the prosthesis relative to the limb. It is therefore desirable to investigate integration methods that could be incorporated into existing prostheses while allowing the socket to remain sealed from the external environment.

Our goal was to develop an inexpensive mechanotactile feedback system that could be easily fabricated and retrofit onto existing wearable transhumeral prostheses. This system would provide a laboratory-based platform for exploration of the benefits of sensory feedback in a functional task work space more closely representing clinical use. This type of testing platform in the lab could provide a clearer pathway from benchtop testing to clinical implementation.

This study presents the development and technical evaluation of a fully integrated mechanotactile feedback system for transhumeral prostheses (Fig. 1), where normal forces are sensed on the fingertips of the terminal device and communicated to the user via mechanotactile tactors mounted to the prosthetic socket. The associated electronic components are modular and adjustable, allowing for flexibility in the number of feedback channels and wireless adjustment of settings.

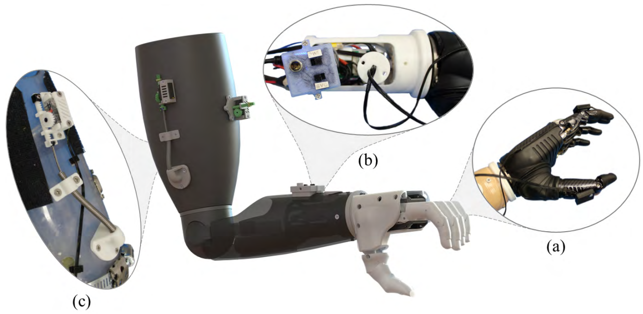

FIGURE 1. **Schematic of mechanotactile tactor system (a) sensors, (b) electronics, and (c) tactor.**

The paper is outlined as follows: Section II describes the design and evaluation of our system, including the selection and retrofitting of an appropriate force sensor, the design and integration of an inexpensive mechanotactile tactor, and the design of an integrated wireless electronics system. Section III describes the methods and results for evaluation of a case study of a transhumeral prosthesis user, followed by the discussion in Section IV.

II. SYSTEM DESCRIPTION AND EVALUATION

A. SENSING OF NORMAL FORCES

Most commercial prosthetic hands do not measure or provide access to internal force measurements. Therefore, a sensor must be retrofit to detect normal forces applied to the prosthetic fingertips, and ideally be applicable to a range of prosthetic prehensors. Many studies have utilized force sensitive resistors (FSR) to measure applied force [22]–[25], [29]–[32], due to their low cost and thin profile. Other sensors include subminiature load cells [18], strain gauges [9], [16], [33]–[37], and impedance-based sensors [38]. In a 2011 review paper, Chappell concluded that the FSR was the optimal sensor for use in prosthetic applications, given the existing state of sensor technology. However, it was also suggested that a force sensor based on capacitive effects would be more stable and accurate, given careful design of the electronics [4]. More recently, inexpensive capacitive sensors have come to market with similar dimensions to the FSR [39], warranting exploration into this alternative technology.

In this section, we describe the evaluation of three commercially available sensors, with the aim to instrument individual prosthetic digits, minimize cost, maximize accuracy, and avoid significant alterations to the prosthetic hand. We then discuss our method of integrating the sensors onto a commercial prosthetic device.

1) COMMERCIAL SENSOR EVALUATION

Similar to previous studies [40], a full factorial design of experiments was used to evaluate three force sensors

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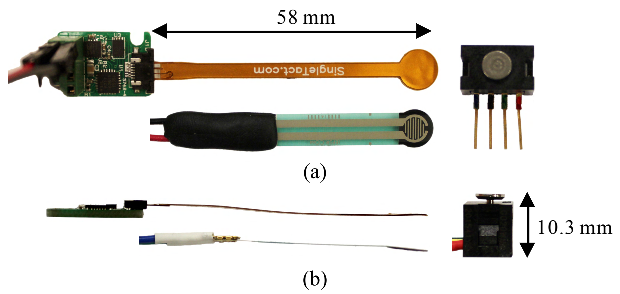

FIGURE 2. Evaluated sensors (a) top and (b) side, approx. to scale; top: SingleTact, bottom: FSR, right: load cell.

TABLE 1. Reported sensor characteristics.

Specification	SingleTact	FSR	Load Cell
Model	SingleTact S8-10N calibrated	Interlink FSR 400	Honeywell FSG020WNPB
Cost (USD)	$75	$6	$107
Height (mm)	Sensor 0.35 Electronics 3.5	0.35	Housing 9 Indenter 1.3
Weight (g)	1.8	< 1	< 1
Range (N)	0 to 10	0.2 to 20	0 to 20
Linearity (%)	< 2.0	N/a	0.5
Hysteresis (%)	< 4.0	10.0	N/a
Safe overload (N)	30	N/a	60

(Fig. 2 and Table 1), namely a capacitance sensor (SingleTact, S8-10N calibrated), an FSR (Interlink, FSR 400), and a subminiature load cell (Honeywell, FSG020WNPB). Three sensors of each model were evaluated, each loaded sinusoidally three times to 8.5 ± 0.3 N, modified from the ANSI/ISA 51.1 Standard [41]. Manipulated variables included contact material (stiff polylactic acid (PLA) vs. Alpha Classic®Gel, Ohio WillowWood) and loading rate (0.5 vs. 5 s period). Indenter curvature (flat vs. Ø 10 mm) was also manipulated for the SingleTact and FSR sensors to simulate real-world usage, where the sensors could be bent around a prosthetic fingertip. Sensors were calibrated based on manufacturer's recommendations [42]–[44]. Three root mean squared errors (RMSE) were determined for each trial; Self-RMSE: error relative to specific calibration conditions, Baseline-RMSE: error relative to baseline conditions (PLA, 5 s, flat), and Cross-RMSE: error relative to baseline conditions, of another sensor of the same model.

Statistical analysis was conducted using Excel (Microsoft, 2016) and Statistica (TIBCO Software Inc. 2016). Normality of the data was confirmed using a normal probability plot. Paired two-sample t-tests were used to evaluate differences between sensor types (FSR, SingleTact, and Load Cell) within each calibration condition (Self-RMSE, Baseline-RMSE, and Cross-RMSE), with $\alpha = 0.05$. General Linear Models (GLM) were developed for each sensor type to determine the effects of changing manipulated variables (contact material, loading rate, and indenter curvature for the FSR and SingleTact) on Baseline-RMSE.

Sample data from each sensor is provided (Fig. 3) as well as summaries of the error analysis (Table 2 and Fig. 4).

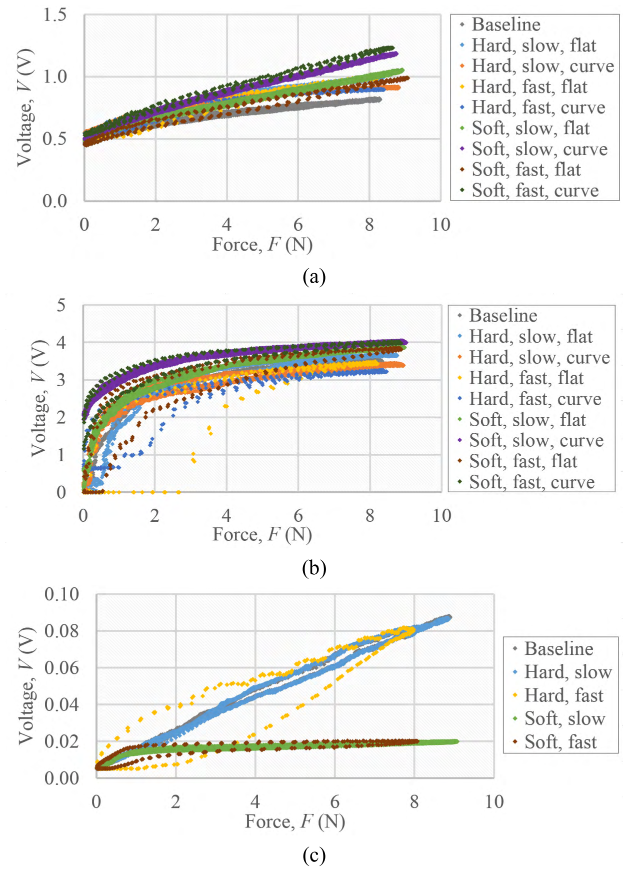

FIGURE 3. Calibration curves of (a) SingleTact, linear responses under different conditions, (b) FSR, non-linear responses with larger spread in data and (c) load cell, non-linear response under soft loading condition compared to hard, with hysteresis demonstrated under fast loading.

TABLE 2. Root mean square sensor errors (RMSE) under changing load rate and contact material.

Error (N)	SingleTact	FSR	Load Cell
Self-RMSE	0.5 ± 0.2	0.9 ± 0.3	1.1 ± 0.6
Baseline-RMSE	1.8 ± 1.0	2.7 ± 1.7	2.4 ± 1.8
Cross-RMSE	2.4 ± 1.8	5.0 ± 3.1	2.4 ± 1.8

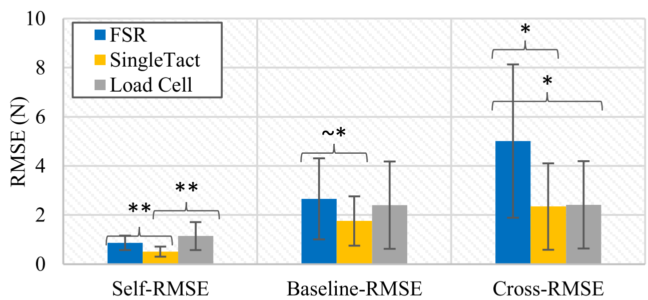

FIGURE 4. Mean absolute sensor errors under changing load rate and contact material, where error bars represent standard deviation, ** indicates significant differences between sensors $p < 0.01$, * indicates significant differences $p < 0.05$, and ∼* indicates approaching significance $0.05 < p < 0.07$.

Paired two-sample t-tests revealed that Self-RMSE was significantly lower for the SingleTact (M = 0.51, SD = 0.20) compared to the FSR (M = 0.87, SD = 0.30);

t(11) = 4.07, $p < 0.01$ and the load cell (M = 1.14, SD = 0.57); t(11) = 3.17, $p < 0.01$. Baseline-RMSE showed the FSR (M = 2.66, SD = 1.65) approached being significantly higher than the SingleTact (M = 1.75, SD = 1.01); t(11) = 2.05, $p = 0.06$. Cross-RMSE for the FSR (M = 5.01, SD = 3.12) was significantly higher than the SingleTact (M = 2.35, SD = 1.76); t(11) = 1.95, $p = 0.05$ as well as the load cell (M = 2.41, SD = 1.78); t(11) = 2.36, $p < 0.05$. The GLM for the FSR revealed a significant increase in Baseline-RSME resulting from changes in contact material (t(20) = 4.06, $p < 0.01$). For the SingleTact, changing contact material (t(18) = 3.30, $p < 0.01$) and indenter curvature (t(18) = 2.94, $p < 0.01$) significantly increased this error. This Baseline-RMSE was significantly increased in the load cell by changes to contact material (t(6) = 54.67, $p < 0.01$) and loading rate (t(6) = 3.73, $p < 0.01$).

Based on this evaluation, the FSR is only recommended for low-accuracy applications where low cost is a high priority. Consistent with the recommendations of Schofield *et al.*, each FSR should be calibrated in conditions as close to intended use as possible [40]; interacting with objects possessing different material properties will increase error. The load cell is recommended for higher-accuracy applications, where the sensor can be recessed within the fingertip, however this likely requires modification to the prosthetic digit. Calibrations from one load cell sensor can be applied to another without compromising accuracy, however changes in contact material and loading rate will increase error. Load cell accuracy could possibly be improved by amplifying the output signal to a higher voltage before reading it into a data acquisition system to improve the signal to noise ratio, or by mounting a larger diameter platform over the indenter to reduce the effect of compliant materials off-loading forces onto the sensor housing. The SingleTact is recommended when the fingertip cannot be substantially modified, and higher accuracy is desired than that provided by the FSR. Each SingleTact sensor should be calibrated individually, understanding that changing material properties and indenter curvature will increase error.

Some prosthetic hands may apply forces higher than the safe overload of the specific sensors evaluated in this study, however each sensor type is available with a larger range; these larger ranges may be appropriate for certain systems, understanding that there will be a trade-off in accuracy. Depending on the grasp configuration of the prosthetic device, contact may occur at different locations on the digit; since the sensors do not cover the entire digit, careful placement is needed to reduce off-loading of forces. Multiple sensors may be integrated onto a single digit to compensate for this issue, where multiple FSR or SingleTact sensors would be easier to integrate than load cells.

2) SENSOR INTEGRATION

To maximize accuracy of sensor readings and avoid permanent modification of the prosthetic digits, the SingleTact sensors were selected for use in our case study. The model

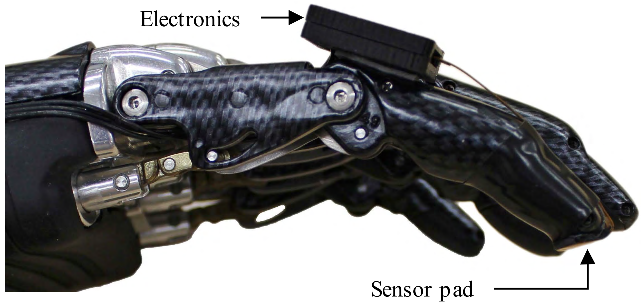

FIGURE 5. SingleTact integration onto prosthetic fingertip.

with a 10 N range was selected to ensure optimal resolution at the lower range of applied forces. Sensors were adhered to the prosthetic fingertips of the index and thumb (Fig. 5) using double-sided tape (Scotch, 3M 4011) and covered with a fingertip cut from a nitrile glove (MicroFlex, MK-296-XS). Electronics were housed within a 3D printed enclosure, attached to the dorsal side of the middle phalange using the same double-sided tape. Calibration of each individual sensor was conducted in these conditions prior to testing to ensure accurate force readings.

B. MECHANOTACTILE FEEDBACK

Various mechanotactile devices have been explored in the literature. Tactors typically include a motor, a mechanism to translate the force, and a head for skin contact. Mechanisms differ between studies; some use a lever-arm that rotates into position [25], [26], [30], [45], [46], others use a linkage [10], [18], [27], [47] or rack-and-pinion [9], [31] system to convert rotational motion to linear displacement. Maximum applied forces range from 3 to 13 N, with maximum excursion distances from 10 to 17 mm [9], [27], [30], [48]. Developing tactors with these outputs that are inexpensive and easy to manufacture would improve accessibility of these systems.

In this section, we present the design and evaluation of two mechanotactile tactors. We then present a method for integration that does not compromise the socket suction seal.

1) TACTOR DESIGNS

Two tactors were designed for attachment to a socket; a linear and a cable-driven tactor. Both models were driven using the same servo motor (Hitec, HS-35HD), and contacted the skin via an Ø 8 mm domed head similar to that suggested in [47] and used in [27]. Other components were 3D printed in PLA and attached using screws.

The linear tactor converted linear motion from a servo motor to rotational motion via rack and pinion gears, where the rack gear had a tactor head on the end which pushed onto the residual limb (Fig. 6 (a and b)). The cable-driven tactor used the same rack and pinion principle, however the motor and tactor head were connected via a constrained Bowden cable (Fig. 6 (c and d)). The cable was routed through a 90° insertion into the socket to allow pushing forces to drive

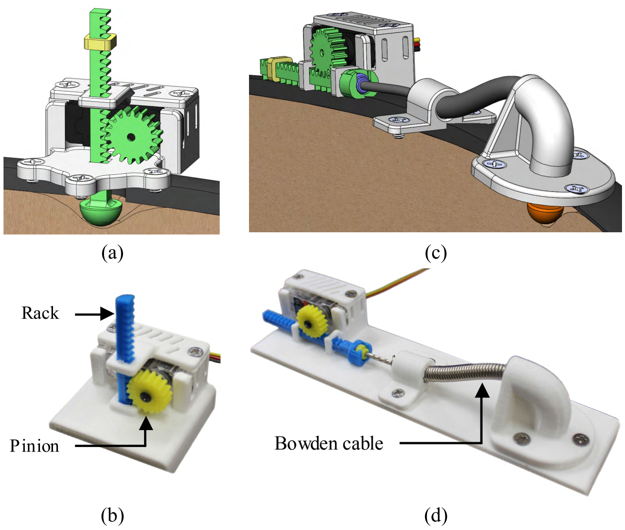

FIGURE 6. Linear (a) tactor design and (b) manufactured components, and cable-driven (c) tactor design and (d) manufactured components.

the tactor head. The cable could be routed along the surface of the socket around interfering components. This design minimized the vertical profile at the tactor head site, which could be useful in areas of the socket that are more difficult to access or prone to interfering with the user.

TABLE 3. Tactor dimensions and performance, mean ± standard deviation.

	Linear tactor	Cable-driven tactor	
		Housing	*Elbow*
Height (mm)	Housing: 14 Rack: 35	14	20.5
Width (mm)	29	44	27
Depth (mm)	35	22	29
		Min. length	*Max. length*
Mass (g)	9	20	36
Max. excursion (mm)	16.5	16.5	11.5
Peak applied force (N)	13.5 ± 1.6	7.9 ± 0.7	6.8 ± 0.9
Current at 5 N (mA)	114 ± 16	170 ± 16	195 ± 28
Delay (ms)	92 ± 16	105 ± 18	115 ± 18
Servo temp. after 20 min at 5 N (°C)	37.6 ± 0.3	N/a	43.4 ± 3.4

Note: min. length tactor was 80 mm long, max. length 220 mm long with 20 mm deviation from straight.

Table 3 summarizes the performance of the tactor models. Peak applied force was measured using a load cell (LCM703-5, Omega Engineering), maximum excursion using a caliper (150MM Digital Caliper, Sparkfun Electronics), current using a multimeter (Fluke 189, Fluke Corporation), delay between initial contact of the sensor and initial movement of the tactor head using a high-speed camera (iPhone SE, Apple Inc.) analyzed frame by frame (Premier Pro CC, Adobe Systems), and temperature using a thermocouple (Traceable Total-Range Thermometer, Control Company). The cable-driven tactor was larger and heavier overall compared to the linear tactor. The use of the Bowden cable reduced the peak applied force, drew more current for a given force, resulted in a longer delay, and reached a higher temperature under sustained loading. Longer cabling and deviations reduced the maximum excursion of the tactor head, likely due to increased frictional forces and bending of the cable. Material cost for both tactors was approximately $40 USD, not including manufacturing and assembly time.

2) TACTOR INTEGRATION AND SELECTION

3D printed plastic components were screwed in place onto the socket, with care taken to avoid creating holes through the socket wall that would compromise the suction seal within the socket. To maintain the seal while allowing the tactor head to extend through the hole in the socket, a flexible membrane was inserted to isolate the tactor from the internal socket pressure (Fig. 7). The membrane seal was constructed using a 0.2 mm thick nitrile fingertip cut from a glove (MicroFlex, MK-296-XS) wrapped around a nitrile O-ring (McMaster Carr, 4061T158) with edges coated in silicone (Momentive Performance Materials Ltd., Silicone II). The seal was of a shape and size such that it was long enough to deform, rather than stretch, when the tactor was depressed. Benchtop testing demonstrated that this membrane could withstand vacuum pressures greater than 60 kPa gauge. The membrane showed no signs of wear after being loaded by a tactor up to 5 N over a tissue analog (SynDaver, Basic Tissue Plate) for 120 cycles.

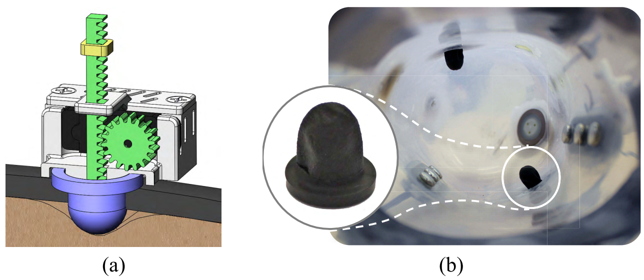

FIGURE 7. Tactor sealing (a) concept and (b) inside view of socket showing extended tactors, with seal superimposed.

Both tactors demonstrated similar performance, however the cable-driven tactor was selected for use in the case study to minimize the vertical profile of the tactor.

C. WEARABLE TACTOR CONTROLLER (WTC)

An electronic controller was required to map forces measured on the prosthetic fingertips to the tactors. Many reported systems in literature are tethered to a computer, limiting the user's ability to perform tasks unencumbered [9], [10], [13], [18], [26], [36], [49]–[51]. To facilitate functional evaluation of the feedback system, discrete on-board design and packaging of the controller was required.

A controller was designed that was untethered and battery powered allowing the user freedom of movement, lightweight and inexpensive to improve acceptance, reliable, and safe. The system was integrated with the existing electrical components, including the prosthetic hand, battery, and electrodes.

Wireless adjustment of tactor parameters was possible using the Graphical User Interface (GUI). Finally, the controller was integrated into the distal portion of a prosthetic forearm.

1) CONTROLLER DESIGN

The designed system (Fig. 8) could control up to eight individually mapped tactors, could be programmed via a microUSB connector, and had a built-in microSD card reader for onboard datalogging. It was comprised of four custom boards; the switch board with charging capabilities to provide reverse polarity protection for the battery, the voltage converter which passed the 7.4 V from the battery to the controller and stepped it down to 5 V to power the motors in the tactors, the controller which was responsible for the force mapping, and the adaptor which allowed the controller to be connected to different sensors and motors, giving the system some flexibility. The controller board employed the same ATmega32u4 (Atmel) 8-Bit microcontroller that is used in the Arduino Micro board (Arduino, A000053). An off-the-shelf Bluetooth board (SparkFun, Bluetooth Mate Gold, WRL-12580) provided wireless communication.

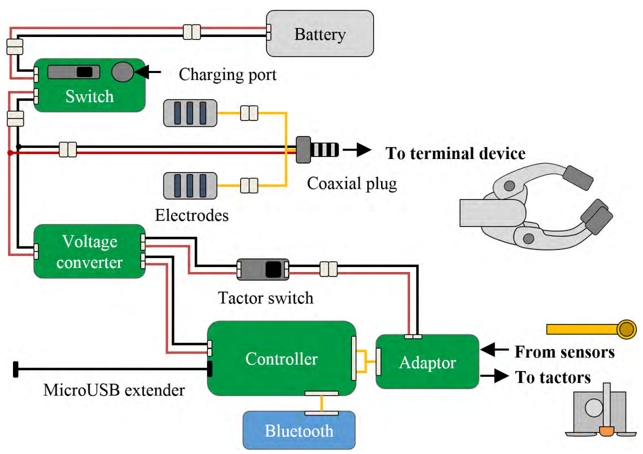

FIGURE 8. Wearable tactor controller schematic, where red wires indicate power, black ground, and yellow signal.

2) FIRMWARE AND SOFTWARE

Controller firmware was written using the Arduino Integrated Development Environment (Arduino, version 1.6.5), and was able to filter the sensor data using a moving average filter ($n = 10$), map to tactor output, receive tactor settings serially, save settings on-board, and send sensor information to the computer wirelessly. The microcontroller updated sensor values and sent motor position commands at a rate of ~50 Hz.

Exponential mapping between sensor readings and applied motor position commands was selected to increase differences in measurements in the fine-touch range (1), where S is the sensor reading (ranging from S_{min} to S_{max}), and P is the tactor position (ranging from P_{min} to P_{max}).

$$P_{exp} = P_{min} - (P_{min} - P_{max}) \left(\frac{S - S_{min}}{S_{max} - S_{min}} \right)^{0.5} \tag{1}$$

The GUI was written in Visual Studio (Microsoft, version 10.0.40219.1) using the C# language. The GUI allowed serial connection to the WTC to enable individual tactors, set tactor position ranges, adjust force mapping ranges, and display measured forces. Settings could be saved to the computer and tactor readings logged. Once settings were finalized the GUI could be closed and the Arduino software on the embedded controller would continue to run independently.

3) CONTROLLER INTEGRATION

A custom enclosure was designed to house the electronic components within the forearm of the prosthesis (Fig. 9). The enclosure had a diameter of 50 mm and length of 86 mm, which allowed it to slide into the empty forearm space available in most commercial transhumeral prostheses. The battery was mounted proximally to the electronics to reduce the torque on the residual limb. The estimated weight of the controller and enclosure was less than 100 g. Component cost was approximately $215 USD, excluding manufacturing and assembly time.

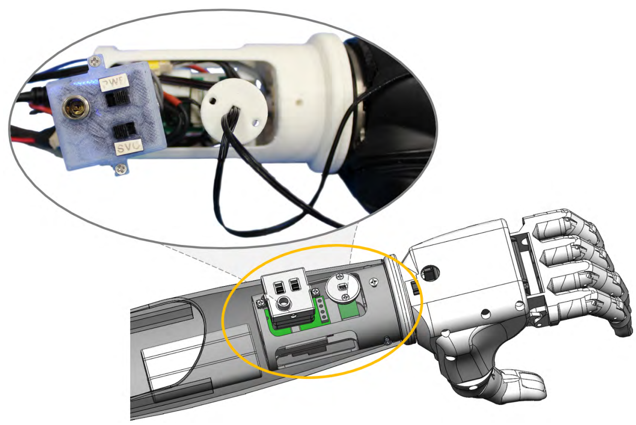

FIGURE 9. Electronics enclosure within prosthetic forearm solid model with manufactured enclosure and controls superimposed.

III. CASE STUDY

A. METHODS AND PROCEDURES

The tactor integrated prosthesis was evaluated with a single male transhumeral participant over three sessions (Fig. 10). Approval for this study was obtained from our institutional Health Research Ethics Board, and the participant provided written informed consent. The participant was an experienced myoelectric user, having used dual-site direct control of a myoelectric hand for over two years prior to the experiment. The commercial prosthetic components were matched to the participant's typical system and included a BeBionic hand (BeBionic, BBHMDLQD), hollow forearm (Hosmer, E400 Forearm Assembly), body powered elbow with lift assist (Hosmer, Lift Assist Unit), and standard check socket with two embedded electrodes (Motion Control, Triad Preamp System). The thermoplastic socket was custom made as per standard prosthetic practice, however

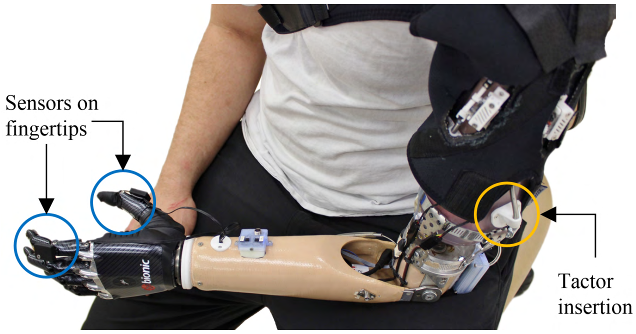

FIGURE 10. Participant wearing tactor integrated prosthesis; note second tactor head is on medial side of residual limb.

all other integrated components were inexpensive, modular, and not specific to the individual, which would thereby allow them to be applied to any transhumeral prosthesis. SingleTact sensors on the prosthetic thumb and index were mapped to two cable-driven tactors on the prosthetic socket. No specific training was provided.

1) SUCTION VALIDATION

Suction air pressure was monitored throughout testing using a pressure sensor (Freescale Semiconductor, MPXx6250A) to verify whether the system was sealed. Air pressure and fingertip sensor data was collected during repetitive grasping tasks while seated and while standing.

2) TACTOR DISCRIMINATION

Similar to the protocol in [30] and [45], we evaluated the ability to identify which finger was being stimulated through two sessions. A learning session was conducted where each of the prosthetic fingers was stimulated by the investigator for three seconds (index, thumb, or both), followed by verbal confirmation of the finger(s) being stimulated. This learning session was conducted with vision and hearing intact until the participant was comfortable with identification, which required three presentations of each finger combination. The test session was conducted with the participant's vision and hearing occluded. Finger stimulations were applied in a random order (including a "no stimulation" condition) four times each for a total of sixteen trials.

3) GRASPING FORCE

The participant was asked to move a cup instrumented with load cells (Omega, LC703-25), similar to devices used in [52] and [53] and an inertial measurement unit (Adafruit, BNO055). He was instructed to move the cup over a partition within a box without dropping or crushing the device. Instructions were kept simple to evaluate the naturally chosen grasp force with and without the sensory feedback, and a fixed grasp pattern was selected. The outer dimensions of the box were 76 by 36 cm, and the partition height was 15 cm. The participant repeated this movement 10 times in each testing block. There were two testing blocks for each feedback condition (on or off), presented in a randomized order for a total of 20 movements in each condition.

The following measurements were recorded: maximum and average force during grasp, tilt, and acceleration, as well as time to task completion. Statistical analysis was conducted using Excel (Microsoft, 2016). Paired two-sample t-tests were used to compare between conditions, with $\alpha = 0.05$.

4) SUBJECTIVE REPORT

The participant was asked to complete a survey by selecting responses on a Likert scale of 1 to 5, defined as: 1, not at all, to 5: extremely. The survey included questions on reliability of control and confidence in releasing objects with and without sensory feedback. Regarding the feedback system specifically, questions included naturalness, whether the feedback was distracting, and whether the system was desirable to incorporate into everyday life.

B. RESULTS

1) SUCTION VALIDATION

An initial testing session demonstrated that even slight leakages in the socket resulted in slippage such that the user needed to reseat the prosthesis every few minutes (Fig. 11 (a)). Once the socket was sealed, representative data from stationary, seated, and standing tasks was collected (Fig. 11 (b, c, and d, respectively)).

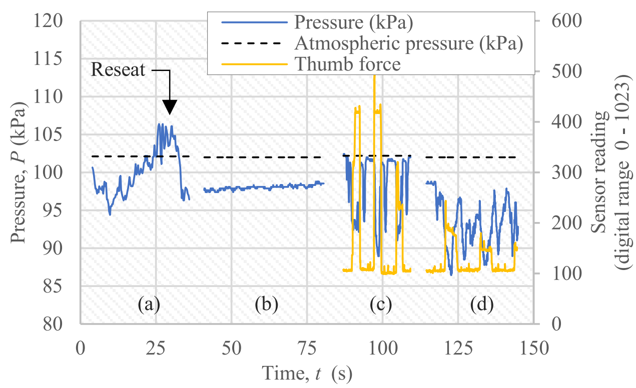

FIGURE 11. Suction pressure measurement (a) leaking, (b) sealed and stationary, (c) seated grasping activity, and (d) standing grasping activity.

Socket vacuum pressure was shown to fluctuate up to 10 kPa during grasp activities. These fluctuations did not compromise prosthetic function and were likely due to changes in limb volume during the muscle contraction required for myoelectric control. Pressures were lower during standing tasks where the prosthesis was suspended from the limb compared to seated tasks where it rested on the table. Air pressure was generally sub-atmospheric, with the main exceptions being during donning or reseating the socket onto the limb.

2) TACTOR DISCRIMINATION

The participant was able to identify which finger(s) were stimulated with 100% success.

3) GRASPING FORCE

Analysis indicated that maximum grasp force was significantly higher for the no-feedback case (M = 22.48, SD = 3.54) compared to the feedback case (M = 17.74, SD = 4.86); t(19) = 4.06, $p < 0.01$. Average grasp force was also higher for the no feedback case (M = 19.27, SD = 3.35) compared to the feedback case (M = 14.60, SD = 3.87); t(19) = 4.62, $p < 0.01$. The no-feedback condition resulted in significantly faster transfer times (M = 2.95, SD = 0.32) compared to the feedback condition (M = 3.30, SD = 0.27); t(19) = 3.59, $p < 0.01$. There were no significant differences in tilt or acceleration measurements (Fig. 12 and Table 4).

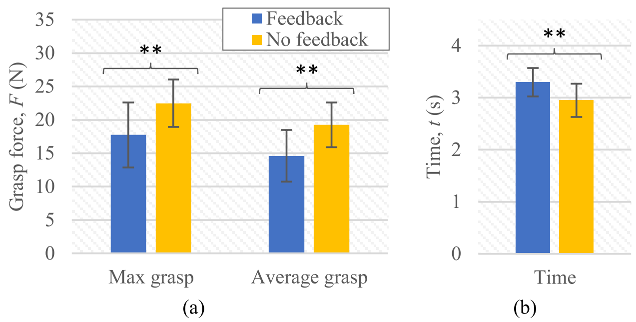

FIGURE 12. Graph of (a) grasp force maximum and average, and (b) grasp duration with and without feedback, where error bars represent standard deviation, ** indicates significant differences between feedback conditions $p < 0.01$.

TABLE 4. Average grasp force and time measurements during cup movement task, mean ± standard deviation.

	No feedback	Feedback
Maximum grasp (N)	22.5 ± 3.5	17.7 ± 4.9
Average grasp (N)	19.3 ± 3.3	14.6 ± 3.9
Maximum tilt (°)	20.1 ± 3.9	20.7 ± 3.4
Average tilt (°)	7.7 ± 1.3	7.9 ± 1.4
Maximum acceleration (G)	1.3 ± 0.4	1.4 ± 0.6
Average acceleration (G)	1.01 ± 0.01	1.01 ± 0.01
Transfer time (s)	3.0 ± 0.3	3.3 ± 0.3

4) SUBJECTIVE REPORT

Without feedback, the participant felt that he had "very" reliable control and was "extremely" confident in knowing if he was releasing an object. He commented that he used constant visual feedback to gauge his control and grip status, and tended to maximally contract his muscle sites to ensure adequate grasp. With feedback, responses to the questions did not change, however he rated that he felt "moderately" more confident in his grip using feedback than without. He rated the sensory feedback as "moderately" natural feeling. He found the stimulation "slightly" distracting, citing the noise of the motors in particular. He recommended further reduction in tactor size and noted that practice sessions may be useful to help adapt to the use of the feedback. Despite these limitations, the participant indicated that it was "extremely" desirable to incorporate the tactor system into his everyday life.

IV. DISCUSSION

A wearable and inexpensive mechanotactile feedback system for use in the laboratory was designed and retrofit onto existing transhumeral prosthetic components. Component costs for a two-channel tactor system were less than \$500 USD, excluding manufacturing and assembly time. Evaluation demonstrated that prosthesis functionality was not impeded by the integration of the sensors, tactors, and controller; specifically, suction suspension was maintained, and the participant was able to successfully manipulate objects in both standing and seated positions.

The SingleTact sensor was shown to produce lower errors than the FSR sensor used in many previous studies [22]–[25], [29]–[32], and was simple to retrofit onto a prosthetic hand. For each of the sensors evaluated in this study, calibrations should be performed as close to the intended use as possible to reduce error. The error in SingleTact readings was significantly increased with changes to contact material and indenter curvature. Future studies should investigate the use of a SingleTact sensor with a larger range, as some prosthetic applications may require measurements above the 10 N sensor evaluated in this study. It may be beneficial to compare the performance of this sensor to others such as strain gauges [9], [16], [33]–[37] or sensors specifically designed for prosthetic applications [38], [54], [55]. Long-term performance of these sensors should be quantified prior to any take-home studies, including investigation into coatings or covers, such as those employed by Rosenbaum-Chou *et al.* [23] and Clemente *et al.* [22].

Two mechanotactile tactors were designed using hobbyist servo motors and 3D printed parts. Tactor selection should weigh the benefits of each design. If height is not a concern, such as on the outer surface of the upper arm, the linear tactor is a better choice. However, when low profile is essential the cable-driven tactor should be used, with an understanding that the output forces will be lower and timing delays higher. It has been recommended that timing delays be shorter than 200 ms to elicit a sense of ownership and agency [56]; delays in both tactor models fall within this limit. Tactors were integrated onto a prosthetic socket using a custom nitrile seal that isolated the tactor from the internal socket pressure to maintain suction suspension; it was assumed that tactor parameters did not change with the addition of the seal given it was not stretched tight over the tactor head, however this might have decreased the maximum output force slightly. The next steps will involve evaluation of long-term performance of the tactors, leading to refinements towards a sensory integrated prosthesis robust enough for functional testing and home use. Improvements should include further iteration to reduce overall size and power consumption (as recommended in [5]), as well as noise. To reduce the potential for pinch points, moving components should be covered.

A custom controller was designed and fit within the prosthetic forearm. This controller allowed wireless adjustment of tactor settings, where parameters were saved on-board so that the system could be used independently of a computer. Exponential mapping from sensor reading to tactor output was employed, however future work should investigate the optimal method of conveying information to the prosthetic user. For example, moving towards a more bimodal sensory response with a peak force applied at the time of contact and release and a lower force maintenance phase, in order to reduce battery requirements. While this controller was designed to allow modularity of components, future refinement should simplify the controller onto a single board. Reduction in size may also allow for the controller to be applied in transradial prosthetic systems, where prosthetic forearm space is more limited. Exploration of alternate servo motors and microcontrollers to reduce delays in the system is also recommended.

The feedback system was integrated onto a transhumeral prosthesis for participant testing. Suction suspension was maintained, with peak vacuum pressures occurring during muscle contraction. Future work should investigate the change in socket pressure with a tactor integrated socket compared to a non-instrumented socket.

The participant was able to discriminate which prosthetic fingertip was stimulated with 100% accuracy, as demonstrated in previous studies [30], [45]. Sensory feedback significantly reduced applied grasping force during object manipulation, similar to previous findings [10], [24], [52]. It is likely that with the sensory feedback, the participant was able to recognize an adequate grasp, preventing the need for maximal contraction to ensure a secure grasp. This reduction in grip force may help to offset the increased power demands due to the incorporation of the tactors. Feedback increased the time required to move the object, also similar to previous findings [57], likely because the participant was experiencing different sensations than normal prosthesis operation. The introduction of sensation to object handling with a prosthesis is a novel skill for experienced myoelectric users, and it is likely that longer training times and accommodation to the sensation is required for the individual to fully integrate the sensation into their motor planning. In addition, there was no negative consequence for exceeding a force threshold in our tasks, so implementing a risk factor whereby the sensory input becomes more important to succeed at the task will be crucial for future evaluation of the impact of sensory feedback. The subjective preference towards the use of tactile feedback demonstrated here has also been documented in previous studies [11], [14], [22]–[24].

This case study has demonstrated the technical performance and functionality of a mechanotactile feedback system that has been integrated into a fully wearable transhumeral socket. This system allowed us to evaluate capabilities in an unrestricted laboratory environment, closer to being representative of a real-world setting, which is a key step towards adequate evaluation of novel feedback for myoelectric systems [7], [20]. This sensory feedback system can be retrofit in a modular and inexpensive way onto current prosthetic components, with the notation that a custom thermoplastic socket is still required for each participant as per standard clinical practice. Further research should explore integration of the feedback system with different prosthetic hands and evaluate functionality across a range of users. It would be valuable to investigate how the sensory feedback system may impact user performance over multiple training sessions, with comparison to other methods such as electrotactile feedback [58]. Studies have demonstrated that initial training with closed-loop feedback may improve the internal feed-forward model for grasping, thus precluding the need for continual feedback [14], [17]; it will be valuable to further explore these concepts in future trials. Outcome measurement should also take into account not just motor performance and simple rating scales of confidence, but the degree of visual attention required when manipulating objects [59], [60]. Our design and integration of a wearable tactor system can provide a low-cost opportunity for conducting this research.

V. CONCLUSION

This study demonstrated the functionality of an inexpensive wearable tactile-integrated prosthesis, with design principles that can be applied to a range of existing prosthetic systems for in-lab testing. The approach provides a platform for further investigation into the potential impact of sensory feedback in a functional task setting more representative of the real world than can be provided with traditional lab-based desktop testing. A SingleTact sensor was shown to be more accurate than commonly used FSRs for sensing normal forces. Two mechanotactile tactors were designed, with air-tight integration into a prosthetic socket demonstrated. A modular controller was developed that was housed within the prosthetic forearm, allowing for wireless adjustment of parameters and operation independent of an external computer. Finally, the tactile-integrated prosthesis was evaluated with a single transhumeral user to show that functionality of the system was maintained, identification of tactor stimulation location was possible, and grasp force was decreased with the application of sensory feedback. This work presents a successful step towards integration of a mechanotactile sensory feedback system into a prosthetic socket without compromising the suction seal. The principles of this tactile-integrated sensory feedback system can be applied in future studies to investigate the direct impact of sensory feedback on tangible outcomes for prosthesis users, thereby reducing barriers to clinical translation.

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REFERENCES

[1] C. Lake and R. Dodson, "Progressive upper limb prosthetics," *Phys. Med. Rehabil. Clin.*, vol. 17, no. 1, pp. 49–72, 2006.
[2] E. A. Biddiss and T. T. Chau, "Upper limb prosthesis use and abandonment: A survey of the last 25 years," *Prosthetics Orthotics Int.*, vol. 31, no. 3, pp. 236–257, 2007.
[3] E. Biddiss and T. Chau, "Upper-limb prosthetics: Critical factors in device abandonment," *Amer. J. Phys. Med. Rehabil.*, vol. 86, no. 12, pp. 977–987, 2007.
[4] P. H. Chappell, "Making sense of artificial hands," *J. Med. Eng. Technol.*, vol. 35, no. 1, pp. 1–18, 2011.
[5] C. Antfolk, M. D'Alonzo, B. Rosén, G. Lundborg, F. Sebelius, and C. Cipriani, "Sensory feedback in upper limb prosthetics," *Expert Rev. Med. Devices*, vol. 10, no. 1, pp. 45–54, 2013.
[6] J. S. Schofield, K. R. Evans, J. P. Carey, and J. S. Hebert, "Applications of sensory feedback in motorized upper extremity prosthesis: A review," *Expert Rev. Med. Devices*, vol. 11, no. 5, pp. 1–13, 2014.
[7] P. Svensson, U. Wijk, A. Bjorkman, and C. Antfolk, "A review of invasive and non-invasive sensory feedback in upper limb prostheses," *Expert Rev. Med. Devices*, vol. 14, no. 6, pp. 439–447, 2017.
[8] F. Cordella *et al.*, "Literature review on needs of upper limb prosthesis users," *Frontiers Neurosci.*, vol. 10, p. 209, May 2016.
[9] S. G. Meek, S. C. Jacobsen, and P. P. Goulding, "Extended physiologic taction: Design and evaluation of a proportional force feedback system," *J. Rehabil. Res. Develop.*, vol. 26, no. 3, pp. 53–62, Jan. 1989.
[10] K. Kim and J. E. Colgate, "Haptic feedback enhances grip force control of sEMG-controlled prosthetic hands in targeted reinnervation amputees," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 6, pp. 798–805, Nov. 2012.
[11] C. Cipriani, F. Zaccone, S. Micera, and M. C. Carrozza, "On the shared control of an EMG-controlled prosthetic hand: Analysis of user–prosthesis interaction," *IEEE Trans. Robot.*, vol. 24, no. 1, pp. 170–184, Feb. 2008.
[12] C. Tejeiro, C. E. Stepp, M. Malhotra, E. Rombokas, and Y. Matsuoka, "Comparison of remote pressure and vibrotactile feedback for prosthetic hand control," in *Proc. BIOROB*, Jun. 2012, pp. 521–525.
[13] J. D. Brown *et al.*, "Understanding the role of haptic feedback in a teleoperated/prosthetic grasp and lift task," in *Proc. World Haptics Conf.*, Apr. 2013, pp. 271–276.
[14] M. Markovic *et al.*, "The clinical relevance of advanced artificial feedback in the control of a multi-functional myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, pp. 28-1–28-18, 2018.
[15] J. Wheeler, K. Bark, J. Savall, and M. Cutkosky, "Investigation of rotational skin stretch for proprioceptive feedback with application to myoelectric systems," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 18, no. 1, pp. 58–66, Feb. 2010.
[16] A. Chatterjee, P. Chaubey, J. Martin, and N. Thakor, "Testing a prosthetic haptic feedback simulator with an interactive force matching task," *J. Prosthetics Orthotics*, vol. 20, no. 2, pp. 27–34, 2008.
[17] A. M. De Nunzio *et al.*, "Tactile feedback is an effective instrument for the training of grasping with a prosthesis at low- and medium-force levels," *Exp. Brain Res.*, vol. 235, no. 8, pp. 2547–2559, 2017.
[18] P. D. Marasco, K. Kim, J. E. Colgate, M. A. Peshkin, and T. A. Kuiken, "Robotic touch shifts perception of embodiment to a prosthesis in targeted reinnervation amputees," *Brain*, vol. 134, pp. 747–758, Mar. 2011.
[19] C. Dietrich *et al.*, "Sensory feedback prosthesis reduces phantom limb pain: Proof of a principle," *Neurosci. Lett.*, vol. 507, no. 2, pp. 97–100, Jan. 2012.
[20] I. Vujaklija *et al.*, "Translating research on myoelectric control into clinics—Are the performance assessment methods adequate?" *Frontiers Neurorobot.*, vol. 11, p. 7, Feb. 2017.
[21] W. Hill, Ø. Stavdahl, N. Hermansson, P. Kyberd, S. Swanson, and S. Hubbard, "Functional outcomes in the WHO-ICF model: Establishment of the upper limb prosthetic outcome measures group," *JPO, J. Prosthetics Orthotics*, vol. 21, no. 2, pp. 115–119, 2009.
[22] F. Clemente, M. D'Alonzo, M. Controzzi, B. B. Edin, and C. Cipriani, "Non-invasive, temporally discrete feedback of object contact and release improves grasp control of closed-loop myoelectric transradial prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 24, no. 12, pp. 1314–1322, Dec. 2016.
[23] T. Rosenbaum-Chou, W. Daly, R. Austin, P. Chaubey, and D. A. Boone, "Development and real world use of a vibratory haptic feedback system for upper-limb prosthetic users," *J. Prosthetics Orthotics*, vol. 28, no. 4, pp. 136–144, 2016.
[24] C. Pylatiuk, A. Kargov, and S. Schulz, "Design and evaluation of a low-cost force feedback system for myoelectric prosthetic hands," *J. Prosthetics Orthotics*, vol. 18, no. 2, pp. 57–61, 2006.
[25] N. Fallahian, H. Saeedi, H. Mokhtarinia, and F. T. Ghomshe, "Sensory feedback add-on for upper-limb prostheses," *Prosthetics Orthotics Int.*, vol. 41, no. 3, pp. 314–317, 2017.
[26] C. Antfolk *et al.*, "Artificial redirection of sensation from prosthetic fingers to the phantom hand map on transradial amputees: Vibrotactile versus mechanotactile sensory feedback," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 21, no. 1, pp. 112–120, 2013.
[27] R. S. Armiger *et al.*, "Enabling closed-loop control of the modular prosthetic limb through haptic feedback," *Johns Hopkins APL Tech. Dig.*, vol. 31, no. 4, pp. 345–353, 2013.
[28] N. G. C. (NGC), "VA/DoD clinical practice guideline for the management of upper extremity amputation rehabilitation," Dept. Veterans Affairs Dept. Defense, Washington, VA, USA, Tech. Rep. Version 1.0, 2014.
[29] I. Saunders and S. Vijayakumar, "The role of feed-forward and feedback processes for closed-loop prosthesis control," *J. Neuroeng. Rehabil.*, vol. 8, pp. 60-3–60-60, 2011.
[30] C. Antfolk, C. Balkenius, G. Lundborg, B. Rosén, and F. Sebelius, "A tactile display system for hand prostheses to discriminate pressure and individual finger localization," *Med. Biol. Eng.*, vol. 30, no. 6, pp. 355–360, 2010.
[31] J. S. Hebert *et al.*, "Novel targeted sensory reinnervation technique to restore functional hand sensation after transhumeral amputation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 4, pp. 765–773, Jul. 2013.
[32] M. Aziziaghdam and E. Samur, "Real-time contact sensory feedback for upper limb robotic prostheses," *IEEE/ASME Trans. Mechatronics*, vol. 22, no. 4, pp. 1786–1795, Aug. 2017.
[33] T. A. Rohland, "Sensory feedback for powered limb prostheses," *Med. Biol. Eng.*, vol. 13, no. 2, pp. 300–301, 1975.
[34] G. F. Shannon, "A myoelectrically-controlled prosthesis with sensory feedback," *Med. Biol. Eng. Comput.*, vol. 17, no. 1, pp. 73–80, 1979.
[35] R. B. Gillespie *et al.*, "Toward improved sensorimotor integration and learning using upper-limb prosthetic devices," in *Proc. Conf. IEEE Eng. Med. Biol. Soc.*, Aug./Sep. 2010, pp. 5077–5080.
[36] M. Štrbac *et al.*, "Short- and long-term learning of feedforward control of a myoelectric prosthesis with sensory feedback by amputees," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 11, pp. 2133–2145, Nov. 2017.
[37] C. Cipriani, M. Controzzi, and M. C. Carrozza, "The SmartHand transradial prosthesis," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, p. 29, May 2011.
[38] Z. Su, J. A. Fishel, T. Yamamoto, and G. E. Loeb, "Use of tactile feedback to control exploratory movements to characterize object compliance," *Frontiers Neurorobotics*, vol. 6, p. 7, Jul. 2012.
[39] *SingleTact User Manual*, Anonymous, Pressure Profile Syst., Inc., Los Angeles, CA, USA, 2016.
[40] J. S. Schofield, K. R. Evans, J. S. Hebert, P. D. Marasco, and J. P. Carey, "The effect of biomechanical variables on force sensitive resistor error: Implications for calibration and improved accuracy," *J. Biomech.*, vol. 49, no. 5, pp. 786–792, Mar. 2016.
[41] *ANSI/ISA Process Instrumentation Terminology*, ANSI/ISA Standard 51.1, 1979.
[42] Interlink Electronics. *FSR 400 Series Data Sheet*. Accessed: 2018. [Online]. Available: https://cdn.sparkfun.com/datasheets/Sensors/ForceFlex/2010-10-26-DataSheet-FSR400-Layout2.pdf
[43] SingleTact. *Miniature Force Sensors*. Accessed: 2018. [Online]. Available: https://www.singletact.com/SingleTact_Datasheet.pdf
[44] Honeywell. *FSG Series Force Sensor*. Accessed: 2018. [Online]. Available: https://sensing.honeywell.com/honeywell-sensing-force-sensors-fsg-product-sheet-008028-2-en.pdf
[45] C. Antfolk *et al.*, "Transfer of tactile input from an artificial hand to the forearm: Experiments in amputees and able-bodied volunteers," *Disab. Rehabil. Assist. Technol.*, vol. 8, no. 3, pp. 249–254, 2013.
[46] C. Antfolk, C. Balkenius, G. Lundborg, B. Rosén, and F. Sebelius, "Design and technical construction of a tactile display for sensory feedback in a hand prosthesis system," *Biomed. Eng. Online*, vol. 9, p. 50, Sep. 2010.
[47] K. Kim, J. E. Colgate, J. J. Santos-Munné, A. Makhlin, and M. A. Peshkin, "On the design of miniature haptic devices for upper extremity prosthetics," *IEEE/ASME Trans. Mechatronics*, vol. 15, no. 1, pp. 27–39, Feb. 2010.

[48] K. Kim, J. E. Colgate, M. A. Peshkin, J. J. Santos-Munné, and A. Makhlin, "A miniature tactor design for upper extremity prosthesis," in *Proc. Fbit*, Oct. 2007, pp. 537–542.

[49] P. E. Patterson and J. A. Katz, "Design and evaluation of a sensory feedback system that provides grasping pressure in a myoelectric hand," *J. Rehabil. Res. Develop.*, vol. 29, no. 1, pp. 1–8, Jan. 1992.

[50] C. Cipriani *et al.*, "A novel concept for a prosthetic hand with a bidirectional interface: A feasibility study," *IEEE Trans. Biomed. Eng.*, vol. 56, no. 11, pp. 2739–2743, Nov. 2009.

[51] D. D. Damian, M. Ludersdorfer, Y. Kim, A. H. Arieta, R. Pfeifer, and A. M. Okamura, "Wearable haptic device for cutaneous force and slip speed display," in *Proc. IEEE Int. Conf. Robot. Autom.*, May 2012, pp. 1038–1043.

[52] S. B. Godfrey, M. Bianchi, A. Bicchi, and M. Santello, "Influence of force feedback on grasp force modulation in prosthetic applications: A preliminary study," in *Proc. EMBC*, Aug. 2016, pp. 5439–5442.

[53] J. D. Brown *et al.*, "An exploration of grip force regulation with a low-impedance myoelectric prosthesis featuring referred haptic feedback," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, p. 104, 2015.

[54] J. Kim *et al.*, "Stretchable silicon nanoribbon electronics for skin prosthesis," *Nature Commun.*, vol. 5, p. 5747, Dec. 2014.

[55] J. Segil, R. Patel, Y. Xiong, M. Schmitt, R. Weir, and N. Correll, "Force sensing prosthetic finger tip using elastomer-embedded commodity infrared proximity sensor," in *Proc. MEC*, Fredericton, NB, Canada, 2017, p. 133.

[56] M. A. F. Ismail and S. Shimada, "'Robot' hand illusion under delayed visual feedback: Relationship between the senses of ownership and agency," *PLoS ONE*, vol. 11, no. 7, p. e0159619, 2016.

[57] H. J. B. Witteveen, E. A. Droog, J. S. Rietman, and P. H. Veltink, "Vibro- and electrotactile user feedback on hand opening for myoelectric forearm prostheses," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 8, pp. 2219–2226, Aug. 2012.

[58] G. Chai, D. Zhang, and X. Zhu, "Developing non-somatotopic phantom finger sensation to comparable levels of somatotopic sensation through user training with electrotactile stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 5, pp. 469–480, May 2017.

[59] J. S. Hebert *et al.*, "3D-gaze and movement: A novel metric of visual attention to measure upper limb prosthetic function," in *Proc. MEC17*, Fredericton, NB, Canada, 2017, p. 37.

[60] M. M. D. Sobuh *et al.*, "Visuomotor behaviours when using a myoelectric prosthesis," *J. NeuroEng. Rehabil.*, vol. 11, no. 1, p. 72, 2014.