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Custom Force Sensor and Sensory Feedback System to Enable Grip Control of a Robotic Prosthetic Hand

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***Abstract*— Amputees living in the developing world can benefit greatly from a dexterous low-cost robotic prosthetic hand that can be controlled via electromyography (EMG). This research addresses part of the challenge of designing and constructing such a low-cost device. In particular, the development of novel and functionally suitable fingertip sensors is presented in this paper. The sensors allowed for the user with a trans-humeral amputation to intuitively control grip strength of the robotic prosthetic hand with the help of an EMG electrode placed on the bicep muscle, as well as, a haptic sensory feedback system. The fingertip sensors illustrated a stable linear relationship with force, an even sensitivity to force over the pulp of finger and the medial and lateral sides of the finger above the distal inter-phalangeal joint across the fingertip. Additionally, it had a low cost of construction ($1.00) and the ability to fit on curved surfaces. Two test subjects evaluated the performance of the sensors in combination with the haptic sensory feedback system. The use of the novel sensors allowed for the test subjects to discriminate the forces experienced by each finger when gripping objects of different shapes, with an accuracy of 80% and 73% accuracy respectively. Hence, the fingertip sensors along with haptic feedback can provide a possible solution for amputees to regain the sense a touch and at a low cost. This is a step towards a cost effective ($\pm\$150$), yet functional robotic prosthetic hand for amputees.**

I. INTRODUCTION

In the United States alone, it is predicted that limb loss will affect 2 million people by the year 2020 [1]. Such amputees can benefit greatly from a robotic prosthetic hand controlled by surface electromyography (sEMG) from the residual muscles on their arm, in order to regain some hand functionality [2]. This would allow the amputee or user to contract the muscle to initiate a hand movement or grasp with proportionate force to the strength of the muscular contraction. Sensors can then feed back information about the grip strength to the amputee [3].

However, widespread usage of the aforementioned EMG systems is limited due to the high financial cost of commercial myoelectric prosthetics such as the Michelangelo (OttoBock), i-limb (Touch Bionics) and the BeBionic 3 (RSLSteeper) which cost between $11000 and $75000 [4]. Additionally, these prosthetics do not have mechanisms of conveying afferent sensory information. Rather they utilize 'auto-grasp' control systems which removes sensing and actuation from the users control [4].

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The lack of user control is problematic since sensory perception plays an important role in object manipulation; without it users need to rely on visual feedback to use the prosthetic hand [5]. This makes it difficult for amputees to engage in daily tasks (such as grasping)[5][6]. Amputees also do not feel a sense of embodiment over the prosthetic hand due to the unintuitive mechanisms of control and feedback [5][6]. Consequently, the user either reduces use of the prosthetic or rejects it entirely [7]. Hence, the ability to sense stimuli, such as grip strength, using electronic sensors in order to provide real-time sensory feedback could allow amputees to regain natural and sub-conscious control over the prosthetic hand. Furthermore, this should be made possible at a low-cost to promote widespread adoption of the technology, especially in developing countries.

In addressing the problem of a lack of low-cost prosthetics and sensory feedback, this paper asks the questions: can low-cost custom force sensors coupled with a sensory haptic feedback system enable grip control of an existing robotic prototype prosthetic hand. Successful performance of the custom force sensors will be measured by similarity/improvement in voltage-force relationship compared to commercial FSRs, as well as, spatial performance of the sensor. The haptic feedback system will be assessed based on the ability to convey spatial and intensity information regarding the measured force. Ultimately, this preliminary research forms part of the broader research theme of designing a low-cost, yet highly functional EMG prosthetic for trans-humeral (above the elbow) amputees.

It has been found that piezoresistive sensors are best suited for high sensitivity applications (such as replicating human sensation on fingertips) and slip detection [8], which is needed to control grip. However, attaching sensors to fingertips and fingers can be a complex task due to the curvature of the fingers [9]. To address this challenge, piezoresistive sensors can be moulded around the finger and mimic the position and area of the finger. An example of this can be seen in Leong et al. [10] whereby a textile based sock was produced by using three fabric layers. The sensing data from the sock is mapped to six vibration motors to create a haptic sensory feedback system. It was further deduced that piezoresistive sensors are best suited for high sensitivity applications, piezoelectric for slip detection and capacitive sensors for high flexibility applications.

Sensory feedback from prosthetics can be subdivided into two types, depending if stimulation of the nerve is direct or indirect. Direct stimulation typically involves electro-tactile feedback to stimulate the remaining portion of the nerve [11].

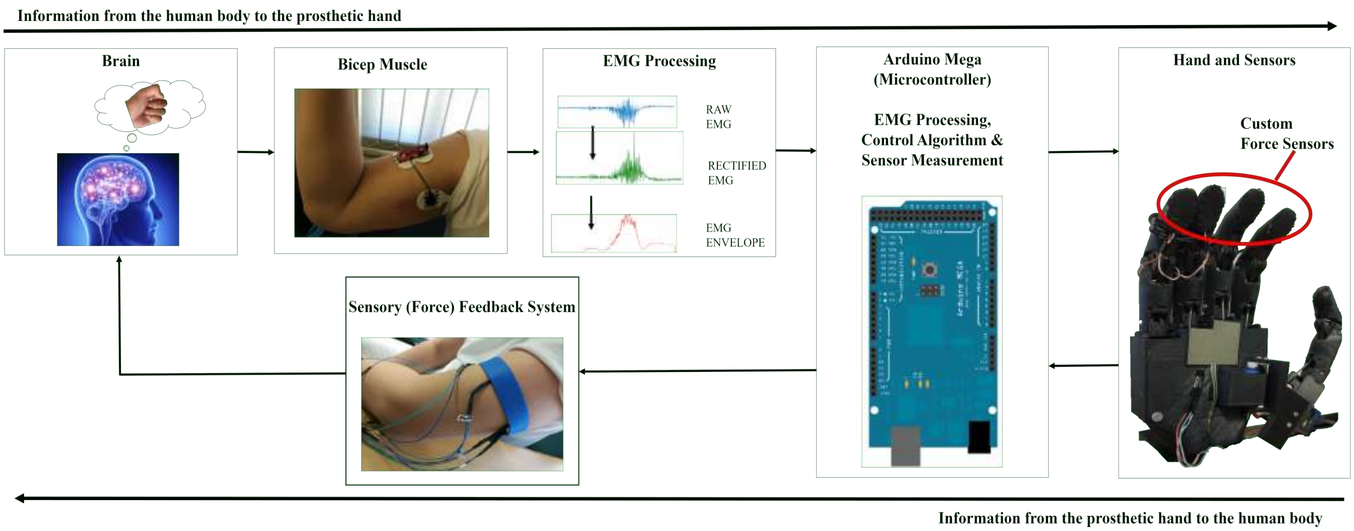

Fig. 1: High-Level System Overview

Numerous parameters of the stimulation are manipulated with varying results. Indirect stimulation conversely is a relatively cheap, easy and effective technique. Vibrotactile and Mechanotactile are typically associated with this method, with the bulk of the literature dealing with the former [4][5][10].

Vibro-tactile feedback makes use of vibration motors [4][10]. These motors evoke stimulation of the skin, with amplitude being modulated to represent intensity. In particular, the benefit of vibration feedback has been proven in terms of grasp success rate [5][12][13], i.e. how often the prosthetic hand successfully grasped an object using the feedback.

This paper is structured as follows: Section II presents a System Overview, whilst Section III provides a detailed System Design. Section IV outlines Testing and Results. Section V discusses the solution and presents recommendations and concluding remarks.

II. SYSTEM OVERVIEW

Fig. 1 shows the high-level system overview. Each subsystem in Fig. 1 is discussed in detail in Section III. Raw EMG data and the EMG envelope were collected and conditioned using the Myoware EMG sensor (Advancer Technologies LLC), which was placed on the bicep of the arm which underwent the amputation. The Arduino Mega microcontroller board (MCU) with Atmel ATmega640 sampled and processed the conditioned EMG envelope signal from the Myoware. The MCU actuated the servomotors of the prosthetic hand in proportion to the amplitude of the EMG envelope. The custom force sensors (referred to from here as low-cost piezoelectric force sensors: LCPFS) were placed on the tips of the fingers of the robotic hand. The sensory measurement of force was then analysed by the MCU and proportionally mapped, in real-time and in terms of intensity, to the corresponding haptic feedback. The vibrations from the haptic feedback sensed on the non-amputated arm of the user was processed by the user's brain, allowing the user to make decisions on how to vary the grip strength of the prosthetic hand. The placement of the system on a subject can be seen in Fig. 2.

III. SYSTEM DESIGN

A. EMG Signal Acquisition and Conditioning

Due to the trans-humeral (above the elbow) amputation, the Myoware EMG sensor was placed on the belly of the bicep muscle of the residual arm as illustrated in Fig. 2. It was used to acquire and condition the EMG signal used to control the prosthetic hand.

The Myoware is a low-cost ($38.00) and portable EMG sensor when compared to clinical EMG systems. The entire device was placed directly on the skin making it less prone to noise [14]. It performs conventional EMG conditioning including amplification and filtering (bandpass and notch filtering), full-wave rectification and low pass filtering, in order to obtain the EMG envelope.

The EMG envelope signal was then used to control the prosthetic hand. This signal was proportional to the strength of the muscular activity/contraction.

B. EMG Signal Processing and Grip Control

Using a sliding window, a Z-Score moving average of the EMG envelope signal was calculated as per Algorithm 1 [14].

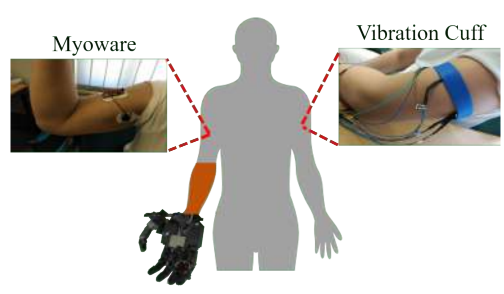

Fig. 2: Placement of the System on Test Subjects

This reduces the effects of sudden changes in the EMG signal brought on by muscle tremors [14]. The normalized EMG envelope was then mapped linearly to the servo angle ranging from 0 to 180°. This mapping performs interpolation and is based on Equation 1. Based on experiential testing, the maximum servo angle was mapped to 40% of the Maximum Voluntary Contraction (MVC), to prevent muscle fatigue.

Algorithm 1 Z-Score and Moving Average Algorithm

Window of size (N = 47) approximately obtaining samples every 4.7 ms : $\mathbf{x}$ = [x0,x1,x2...x48]
EMG sample: Store as x0 and slide each value forward
Calculate mean of window ($\bar{x}$) = $\frac{\sum_{n=0}^{N} x_n}{N+1}$
Calculate standard deviation (S) = $\sqrt{\frac{\sum_{n=0}^{N}(\bar{x}-x_n)^2}{N}}$
Calculate Z-Score (Z_n) for each x_n = $\frac{\bar{x}-x_n}{S}$
If Z_n is > 3 or < -3 then replace x_n with the mean
Calculate the moving average (x_{ma}) = $\frac{\sum_{n=0}^{N} x_n}{N+1}$

$$\text{map} = \frac{(\text{Input} - \text{minIn}) * (\text{maxOut} - \text{minOut})}{(\text{maxIn} - \text{minIn})} + \text{minOut} \quad (1)$$

Different window/buffer sizes (between 10 and 100 samples), as well as, Z-Score standard deviation criteria (where Zn = $\pm 1, \pm 2, \pm 3$) were evaluated, with quantitative response times. It was found that smaller windows and standard deviation criteria led to faster grip response but greater hand jitter and vice versa for larger window sizes. The choice of a 47-sample window (obtained every 4.7 ms) provided the best qualitative balance between smooth yet rapid transient response to the EMG signal. To ensure functionality for different users, a 10 s calibration procedure was incorporated, which measures the users baseline (relaxed EMG value) and MVC. The two values are stored in the MCU's EEPROM.

A standby mode was incorporated whereby the user locked the motors when a suitable, good quality grip was achieved. This aided with usability by reducing muscle exertion when holding objects for long periods of time, as well as, reducing power consumption. For testing of the sensors, the prosthetic hand was hard coded to perform power grips only.

C. Force Sensing:

The commercial Force Sensitive Resistor (FSR) 402 has been used for force measurement in some prosthetic hand applications reported in literature [15]. However, this type of sensor only measures force at the centre of pressure (CoP). Thus it is prone to unreliable results if forces are applied outside its CoP. Hence the novel LCPFSs were developed to mitigate this issue. Additionally, they can be fitted to curved surfaces, such as fingertips. An innovative feature of the LCPFS is that it can be made in any size and shape and can thus be incorporated to any prosthetic hand. Furthermore, similar commercial force sensors cost up to 7 times more than the sensor produced.

1) Low-cost Piezoelectric Force Sensor (LCPFS) Design: The LCPFSs shown in Fig. 3, were constructed using both pressure sensitive fabric (which decreases its resistance with increased pressure) and two layers of conductive fabric. Neoprene layers were added to increase grip/friction and to evenly distribute the force along the surface of the fingertip. It also keeps the actual piezoresistive fabric from wrinkling, which can otherwise introduce noise. The copper fabric layers were used as conductive layers across the pressure sensor fabric.

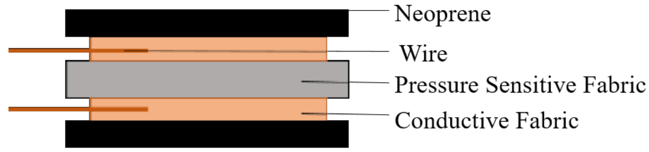

Fig. 3: Custom Force Sensor Design

Eeontex piezoresistive material, was used as the pressure sensitive fabric due to it having a high conductivity with a surface resistance of 2 Kohm/sq, a dynamic range of 5g to 100kg and data acquisition rate of 500 cycles per second [16]. A piezoresistive material was chosen due to its low cost, good sensitivity and low noise [17]. The layers were sewn together using non-conductive thread around the edges of the neoprene. The sensors were shaped to fit the pulp of the fingers.

2) Sensor Placement and Implementation: A colour transfer test (not outlined in this paper) was performed to determine optimal sensor placement on the prosthetic hand. This was done by allowing the prosthetic hand to grasp a variety of common objects; while colour from the object was transferred to areas of the hand that made contact with the objects. This confirmed that force was primarily exerted on the palm and fingertips (distal phalanges). Consequently, the LCPFSs were placed on the fingertips as shown in Fig. 1 whilst a square FSR was placed on the palm due to the flat surface of the latter. To obtain a quantifiable value of force, each sensor's linear regression equation (voltage-mass) was used to translate voltage to mass in the MCU, which was then converted to Newtons.

D. Force (Haptic) Feedback

The haptic feedback system, conveyed the measured force from the sensors and translated it into an afferent sensory feedback. Vibro-tactile feedback (vibration motors) were used as they are smaller, easier to use and lead to a cost effective feedback solution. The vibration motors were proportionally distributed in a circular cuff, which fits around the arm as shown in Figure 4. The vibration motor cuff was placed on the non-amputated arm to minimize vibrational noise in the EMG signal recorded from the amputated arm. Each force sensor was mapped to a corresponding vibration motor as depicted in Fig. 4. The vibration motors used in the vibro-tactile cuff were driven at a constant frequency of

490 Hz by a transistor (2N2222) driver circuit. It was used to supply sufficient current to the motors which had a high startup current of 120 mA. Each transistor was controlled by varying the duty cycle of an MCU pulse width modulation (PWM) output. Consequently, a higher intensity of force corresponded to a higher intensity of vibration.

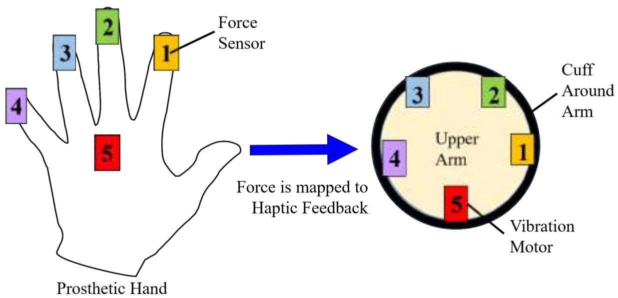

Fig. 4: Vibro-tactile Haptic Feedback Configuration

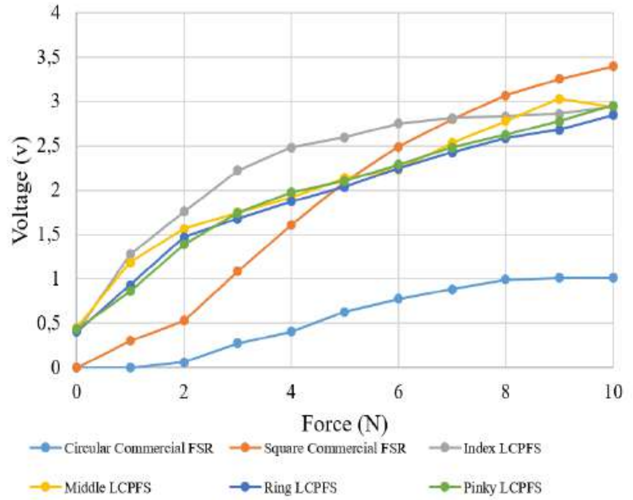

Fig. 5: Voltage vs Force Relationship FSR Comparison (LCPFS and Commercial)

IV. TESTING AND RESULTS

It is hypothesised that the novel LCPFSs have similar/improved performance to commercial FSRs in terms of linearity of mass-force relationship, as well as, spatial performance. It is also hypothesized that the sensory information conveyed by the haptic feedback system will adequately represent differences both spatially and in terms of intensity. The following tests were conducted to evaluate these hypotheses. User testing was only conducted on two patients due to the limitations of ethical clearance.

A. Sensor Testing

In order to validate the LCPFSs it was required that they be tested against commercial FSR's in terms of the force-voltage relationship. Furthermore, the spatial performance of the LCPFS needed to be quantified compared to the center of pressure performance of commercial FSRs.

1) FSR Comparison: The LCPFSs developed for each finger were compared to commercial circular and commercial square FSRs in terms of their voltage-force relationships. This was assessed by placing the sensors on a flat surface. Thereafter, circular lead masses (diameter=30mm) of between 100-1000g (in intervals of 100g) were placed flat on the center of the sensor and the voltage recorded for each applied mass (i.e. force applied). Three repetitions of each test were conducted and the results averaged. The comparative force-voltage relationships are shown in Fig. 5. It can be seen that all the FSRs approximate a linear relationship between voltage and force. The square FSR and LCPFSs have superior voltage resolution (3V) compared to the commercial circular FSR (1V). Furthermore, each LCPFS costs approximately \$1.00, which is cheaper then commercial FSRs (cost \$7 each).

2) Spatial Performance: The spatial performance of the LCPFSs was quantified to determine which areas of the sensor were sensitive to the applied forces and to what extent. In a similar manner as Section IV-A.1, the lead masses were placed on top of the three different parts of the sensor as shown in Fig. 6. The accuracy of force measurement was determined by translating the measured resistance of the sensor to measured force as a percentage of the actual applied force. A heatmap shown in Fig. 6 illustrates that the most of the area of the sensor (including the centre and side regions) transduces above 90 % of the applied force. This solves the problem of commercial FSR's which only measure force at the CoP. The bottom of the sensor had poor relative performance measuring only 40% of the actual force. This was likely attributed to the wires being located in this region.

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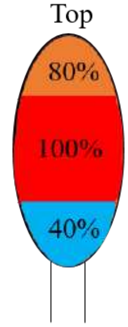

Fig. 6: Force Mapping Across Sensor, as percentage of actual applied force

B. Haptic Feedback

The performance of the vibro-tactile force feedback was assessed based on the test subject's ability to accurately interpret the force in terms of spatial origin and intensity. With Ethics approval from the University of Witwatersrand, two able-bodied male test subjects were used for testing, without both visual feedback (using a blindfold) and without auditory feedback (using white noise). Test subjects were trained prior to testing where the terminology used in the tests was clarified.

1) Spatial discrimination: In order, to validate the functionality of the haptic feedback system, the ability to convey spatial differences was tested. Each individual sensor (mounted on the hand) was pressed by squeezing it randomly to activate the motor. The vibration intensity was maintained at a constant 80% of the maximum vibration intensity, irrespective of the strength of the force applied to the sensor. This singular test was repeated twice for each sensor. Since adjacent sensors on adjacent fingers are mapped to adjacent vibration motors on the feedback cuff, the sensors on the regions of adjacent fingers (e.g. index and middle finger, middle and ring finger, ring and pinky finger etc) were pressed randomly and this was repeated twice. The individual finger and adjacent finger tests amounted to a total of 20 tests conducted per subject. The combined confusion matrix from both participants for real stimulus vs subject response is shown in Fig. 7. It highlights that subjects were able to spatially discriminate feedback using the haptic system proposed. Fig. 7 also shows that spatial misclassification only occurred for adjacent spatial regions. This is due to the vibrations propagating through the skin to adjacent regions of the skin or the motors being too close due to physical limitations and variations in arm circumference.

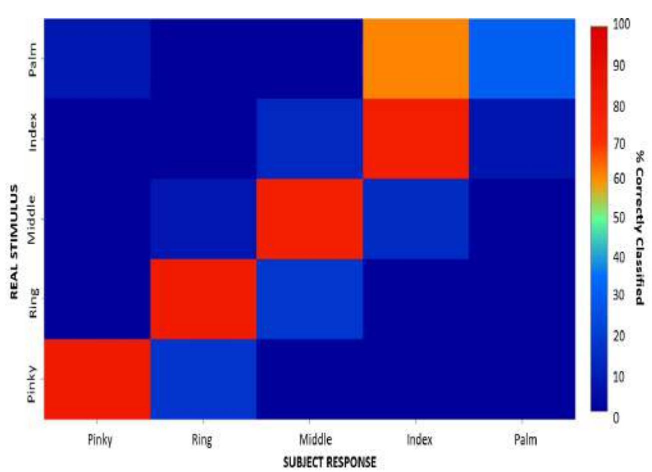

Fig. 7: Combined Spatial Confusion Matrix of Subject 1 and Subject 2 (where the y-axis is Actual Stimulus and the x-axis is Subject Response)

2) Intensity discrimination: The motors vibration was set in code and varied at random intensities between 0-100% of maximum vibration intensity (in 20% intervals). The spatial configuration was kept constant such that the subject was unaware of the new intensity. The test subjects recorded the perceived intensity as either light (corresponding to 20%), medium (corresponding to 40%), hard (corresponding to 60%), very hard (corresponding to greater than 80%). Both subjects obtained an 80% accuracy for the 20 tests. This indicates that the haptic feedback system allows for discrimination of different intensities.

3) Intensity variation: The subjects were introduced to the system as discussed in Section IV-B and the aim was to determine the ability of the system to represent intensity variation. The test was a combination of the two aforementioned tests. The subjects were exposed to an initial vibration. Thereafter, the vibration of a singular motor was either increased or decreased in code. The subjects were then asked to identify whether the intensity had increased or decreased in code and were scored on a pass-fail criteria. Five tests were conducted and both subjects had 100% accuracy at assessing the variation in intensity. This highlights that the system can represent variations in intensity which subjects can identify.

C. Integrated System

A "blind" integration of all the previous tests was then conducted to evaluate overall performance. It involved placing 10 objects in a random order onto the palm. The following objects were used where d is diameter, w is width and l is length: tennis ball (d: 68.6 mm), firm spongy stress ball (d: 68.6 mm), toy rugby ball (l: 120 mm, w: 72 mm), soft plastic cup (d: 95 mm, h: 125 mm), hard plastic cup (d: 105 mm, h: 95 mm), ballpoint pen (l: 88.9 mm), screwdriver (l: 150mm), earbud box(l: 72 mm, w: 72 mm, d: 28 mm), sauce bottle (d:82 mm, h: 228 mm), marble (d: 10 mm). The variety of differently shaped objects was used to test the feedback system rather than the strength of the hand, hence heavy objects were not used.

Upon feeling the object on the palm the user closed the hand until the grip "felt" secure. The subjected indicated three things: (i) Spatial discrimination: Which fingers were currently activated, (ii) Regional discrimination: Which regions of fingers (e.g. index and middle or middle and ring etc) were activated, (iii) Intensity discrimination: The perceived intensity of the grip force. The assessments of the test subjects were compared with measured values displayed to the tester. The unit tests were randomized and run 3 times per object. The test was only considered a pass, if both spatial and intensity were correctly identified. The accuracy are summarised in Fig. 8.

V. DISCUSSION

The LCPFSs have shown accurate proportional representation of forces applied to the curved surfaces of the fingertips of the prosthetic hand. The LCPFSs have also allowed for proportional sensory feedback of these forces via vibro-tactile haptic feedback, which the user can accurately decipher in terms of intensity and spatial distribution across the fingertips. This highlights the functionality of both the LCPFSs and the haptic feedback system. However, the performance was degraded on subjects with smaller arm

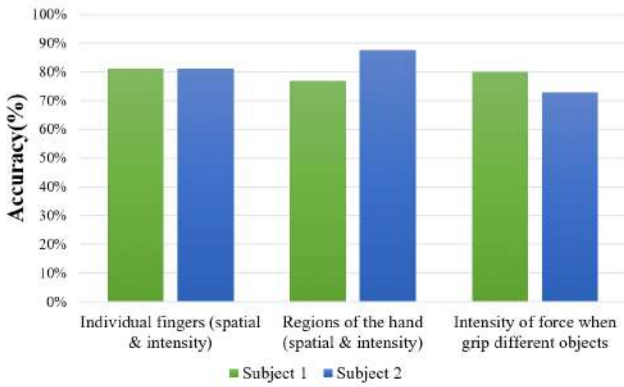

Fig. 8: System Integration Results

circumferences. This is due to feedback vibration motors being close together around the circumference, thereby impeding the spatial discrimination capabilities. As illustrated in Fig. 7 by the confusion matrix, this resulted in user misclassification to the adjacent finger.

Whilst, the benefit of the LCPFSs coupled with haptic feedback has shown to be potentially viable in this preliminary study, it is recommended that for a future study, amputees and a larger cohort of test subjects be used to ascertain statistical significance of the results.

The LCPFSs displayed superior characteristics over the commercial FSRs for this application as was outlined in Section IV-A.1. However, the no-load resistance is not infinite as with commercial FSRs. Therefore, it is imperative to mitigate this initial bias using the conditioning software in the micro-controller when using this sensor. Furthermore, the cost-effective nature of the LCPFSs ($1.00 vs $7.00 for commercial FSR's) means that it would be useful in low-cost prosthetic hand applications, as well as, other low-cost force sensing application domains, particularly in third world countries. In particular, the LCPFSs are ideal for sensing forces applied to prosthetic hands since they can be fashioned easily to different sizes and shapes to accommodate the complex areas of a prosthetic hand.

Moreover, the overall robotic prosthetic hand system (LCPFS, EMG and feedback) proposed fills the gap in the low-cost EMG prosthetic domain as follows: prosthetic hand structure and motors ($36); LCPFSs ($5); haptic feedback ($16); supporting electronics ($89). The total system costs approximately $150, which is significantly cheaper than commercial prosthetics which do not include a functional haptic feedback system. This highlights the potential practical translational aspects of the system, which will benefit the quality of life of amputees in developing countries through access to a cost effective, yet highly functional and dexterous robotic prosthetic hand.

In summary, low-cost piezoelectric force sensors (LCPFS) have been developed and tested to detect forces on the curved surfaces of the fingertips of a low-cost robotic prosthetic hand. They have enabled force haptic feedback to enable grip control of the prosthetic hand. The solution requires further development and testing, but is a step closer to realizing the aim of cost effective yet functional prosthetic hand for trans-humeral amputees.

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