2020

Providing Non-Invasive Sensory Feedback for Transradial Prosthetic Hand Users

Benjamin Stephens-Fripp

Providing Non-Invasive Sensory Feedback for Transradial Prosthetic Hand Users

Benjamin Stephens-Fripp

This thesis is presented as part of the requirement for the conferral of the degree:
Doctor of Philosophy

This research has been conducted with the support of the Australian Government Research Training Program Scholarship

University of Wollongong
School of Mechanical, Materials, Mechatronic and Biomedical Engineering

April 2020

Abstract

Currently there is a large rejection rate and dissatisfaction with prosthetic hands. One primary reason for the rejection of the prosthetic hands is that there is no or negligibly small feedback or tactile sensation from the prosthetic hand to the user, making the prosthetic device less functional. This lack of feedback requires significant reliance on visual information from the user in order to do basic gestures and daily activities, and therefore, can lead to significant cognitive effort. In addition to reducing the need for visual attention, sensory feedback has been shown to increase embodiment and reduce the occurrence of phantom limb pain.

This thesis examines the application of mechanotactile stimulation and electrotactile stimulation to communicate to prosthetic hand users their level of grasping force being applied to objects. The focus is on those with transradial upper limb loss, providing up to three channels of information to represent the grasping force from three different fingers (thumb, pointer and remaining three fingers).

In this thesis, an alternate method to apply mechanotactile stimulation is developed and tested, which applies a combination of vertical pressure and transversal skin stretch to help aid in recognition of stimulations or sensory feedback. This technique has been characterised to determine the optimum direction of the skin stretch and the recognition rate of six grip combinations at two different strength levels. Further, to enable a reliable method of communicating the level of grasping force, just noticeable difference and the relationship between the applied stimulation and perceived intensity for this mechanotactile device is determined.

A novel method of creating 3D printed, reusable and flexible electrodes for electrotactile stimulation is presented and its performance was experimentally verified by comparing with disposable electrodes that are typically used in current prosthetic hands. Further, a comparison was conducted, on both the qualitative and quantitative performance of two differently sized concentric electrodes and the dual separated electrodes for various

psychophysical properties. These results have demonstrated the advantages of the concentric electrodes over dual separated electrodes, and provided the reasoning and justification for the use of concentric electrodes in electrotactile stimulation for sensory feedback.

Current literature on the application of non-invasive sensory feedback typically applies stimulation to either the upper arm region or lower arm region, with minimal information available on the impact of the location on recognition rate and sensitivity. In this thesis, it is demonstrated that there is no statistically significant difference in sensitivity between the two regions of the arm, and is shown that there is no statistically significant difference between the two locations in the recognition of three channels of information from mechanotactile and electrotactile stimulation. These data allow for sensory feedback to be applied to the upper arm, without any significant reduction in performance, and leaving the forearm region for EMG control and remove the need to modify any existing prosthetic sockets.

A large amount of literature examining the use of non-invasive stimulation for sensory feedback as part of the control loop uses either able-bodied subjects or requires amputees to quickly adapt to a new prosthetic hand. However, in this thesis, data is presented from experiments with five transradial prosthetic hand users with their existing myoelectric prosthetic hand, moving a sensorised object both with and without sensory feedback. All five subjects tested were able to recognise and utilise the sensory feedback, either in the modality matched form of mechanotactile stimulation or sensory substitution form of electrotactile stimulation, to reduce their maximum and average gripping forces. Further, all five subjects rated the comfort of both stimulation methods very high, and the feedback increased their perceived confidence in being able to control gripping force.

Acknowledgments

I would firstly like to thank my two supervisors Prof Gursel Alici and Dr Rahim Mutlu for all your help, guidance, assistance and support during this time. Both of you have been always willing to provide me with anything I needed and were very generous with your time. I have learnt a lot from both of you about research and I am truly appreciative. I am particularly thankful to Prof Gursel Alici for the reassurance throughout the course and many discussions and advice and developing me in my research career.

I would like to thank my fellow team members in the AMBER Group. You have made my time at UOW more enjoyable and you were always willing to help whenever it was needed.

I would also like to express my gratitude to Dr Greg Bowring and Melissa Leong, at the Prince of Wales Hospital in Sydney, who offered valuable insight into my work and connected me with existing prosthetic limb users.

Lastly, I would like to thank my family for their support and encouragement. To my parents, you have always supported me and encouraged me to achieve my best. To my kids Archer, Xander and Eloise thank you for your happy smiles that warm my heart and your patience with me and my work. My deepest gratitude to my wife Jodie for your friendship, your encouragement to pursue my passions, and for your continual support.

Certification

I Benjamin Stephens-Fripp declare that this thesis submitted in fulfilment of the requirements for the conferral of the degree Doctor of Philosophy, from the University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. This document has not been submitted for qualifications at any other institution.

Benjamin Stephens-Fripp
25 April 2020

List of Publications Throughout PhD Course

Walker, M., Goddard, E., **Stephens-Fripp, B**. and Alici, G. "A Survey on Towards including end users in the design of prosthetic hands: Ethical analysis of a survey of Australians with upper-limb difference". *Science and Engineering Ethics*. *Vol 26, 981–1007 (2020). https://doi.org/10.1007/s11948-019-00168-2*

Stephens-Fripp, B., Mutlu, R., Alici, G. "A Comparison of Recognition and Sensitivity in the Upper Arm and Lower Arm to Mechanotactile Stimulation". *IEEE Transactions on Medical Bionics and Robotics Vol. 2, no. 1, pp. 76-85, Feb. 2020.* DOI: 10.1109/TMRB.2019.2956231

Stephens-Fripp B., Wallace, E., Searle, T., and Alici G., "Design of a Sensorised Object to Test Sensory Feedback for Prosthetic Hands ", in *IEEE/ASME International Conference on Advanced Intelligent Mechatronics,* pp.157-162, Hong Kong, July 2019.

Young, S., **Stephens-Fripp B.**, Gillet, A., Zhou, H., and Alici G., " Pattern Recognition for Prosthetic Hand User's Intentions using EMG Data and Machine Learning Techniques" *in IEEE/ASME International Conference on Advanced Intelligent Mechatronics,* pp.544-549, Hong Kong, July 2019.

Stephens-Fripp B., Walker M., Goddard E., and Alici G., "A Survey on What Australian's with Upper Limb Difference Want in a Prosthesis: Justification for using Soft Robotics and Additive Manufacturing for Customised Prosthetic Hands," Disability and Rehabilitation: Assistive Technology (IIDT), DOI:10.1080/17483107.2019.1580777, February 2019.

Stephens-Fripp B., Mutlu R., and Alici G., "Using Vibration Motors to Create Tactile Apparent Movement for Transradial Prosthetic Sensory Feedback," in 2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob), 2018, pp. 213-218.

Stephens-Fripp B., Mutlu R., and Alici G., "Applying Mechanical Pressure and Skin Stretch Simultaneously for Sensory Feedback in Prosthetic Hands," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*, 2018, pp. 230-235.

Stephens-Fripp B., Sencadas V., Mutlu R., and Alici G., "Reusable Flexible Concentric Electrodes Coated With a Conductive Graphene Ink for Electrotactile Stimulation," (in English), *Frontiers in Bioengineering and Biotechnology,* Original Research vol. 6, no. 179, 2018-December-03 2018.

Stephens-Fripp B., Alici G., and Mutlu R., "A Review of Non-Invasive Sensory Feedback Methods for Transradial Prosthetic Hands," *IEEE Access,* vol. 6, pp. 6878-6899, 2018.

Stephens-Fripp, B., Mutlu, R., Alici, G. "A Comparison Between Separated Electrodes and Concentric Electrodes for Electrotactile Stimulation". *IEEE Transactions on Medical Bionics and Robotics,* (Under Review).

Stephens-Fripp, B., Leong, M., Bowring, G., Mutlu, R., Alici, G. "Non-Invasive Sensory Feedback to Enhance Gripping Force Control of Myoelectric Transradial Prosthetic Hands". *Disability and Rehabilitation,* (Under Review).

List of Names or Abbreviations

AR	Augmented Reality
CNS	Central Nervous System
DOF	Degree of Freedom
DT	Detectable Threshold
ECG	Electrocardiogram
EEG	Electroencephalogram
EMG	Electromyography
FMRI	Functional Magnetic Resonance Imaging
FSR	Force Sensitive Resistor
HSD	Honestly Significant Difference
JND	Just Noticeable Difference
MRI	Magnetic Resonance Imaging.
NSF	Non-Somatotopical feedback
PNS	Peripheral Nervous System
PT	Pain Threshold
PW	Pulse Width
SD	Standard Deviation
SE	Standard Error
SF	Somatotopical Feedback
TENS	Transcutaneous Electrical Nerve Stimulation

TABLE OF CONTENTS

List of Figures

List of Tables

Chapter 1

Introduction

1.1 Problem statement and Rationale

Tactile information is required for correction and control of object grasps and manipulations as vision alone does not provide enough of the information required [1]. Prosthetic users have also shown a strong desire to decrease the need for visual attention to perform functions [2, 3]. Prosthetic hand rejection rates are estimated to be as high as 40% [4]. Some of user's reasons for rejection or not wearing a prosthetic device are that they believe it is more functional and easier to receive sensory feedback through their stump without using the prosthetic hand [4]. Sensory feedback is also important for prosthetic devices as it can provide users with a sense of embodiment in their prosthesis [5-7].

Body-powered prosthetic limbs can transmit a limited amount of sensory feedback through cable tension. However, with myoelectric prosthetic devices, this indirect feedback pathway no longer exists [8]. This problem was identified early on in the Boston Arm prosthetic [9] where the authors introduced vibration feedback to give the user proprioceptive information on the elbow joint of an EMG controlled prosthetic device resulting in a performance comparable to that of the cable driven prosthesis. Sensory feedback from the nerves within our hands provides feedback on our grasp, contact surface and its roughness and shape, and grasp stability [1]. Biological skin detects these features through four different types of mechanoreceptors in our skin [10], as shown in Figure 1.1. In a simplified overview of a biological feedback system, action potentials are then sent through our Peripheral Nervous System (PNS) to transmit this information to our Central Nervous System (CNS) for decision-making. However, as shown in Figure 1.2, the feedback loop for a prosthetic device differs from our own biological feedback system. A combination of sensors is required in prosthetic devices to match the range of signals detected by our mechanoreceptors in our skin. The signals from these sensors require signal processing to encode them into a form that the user

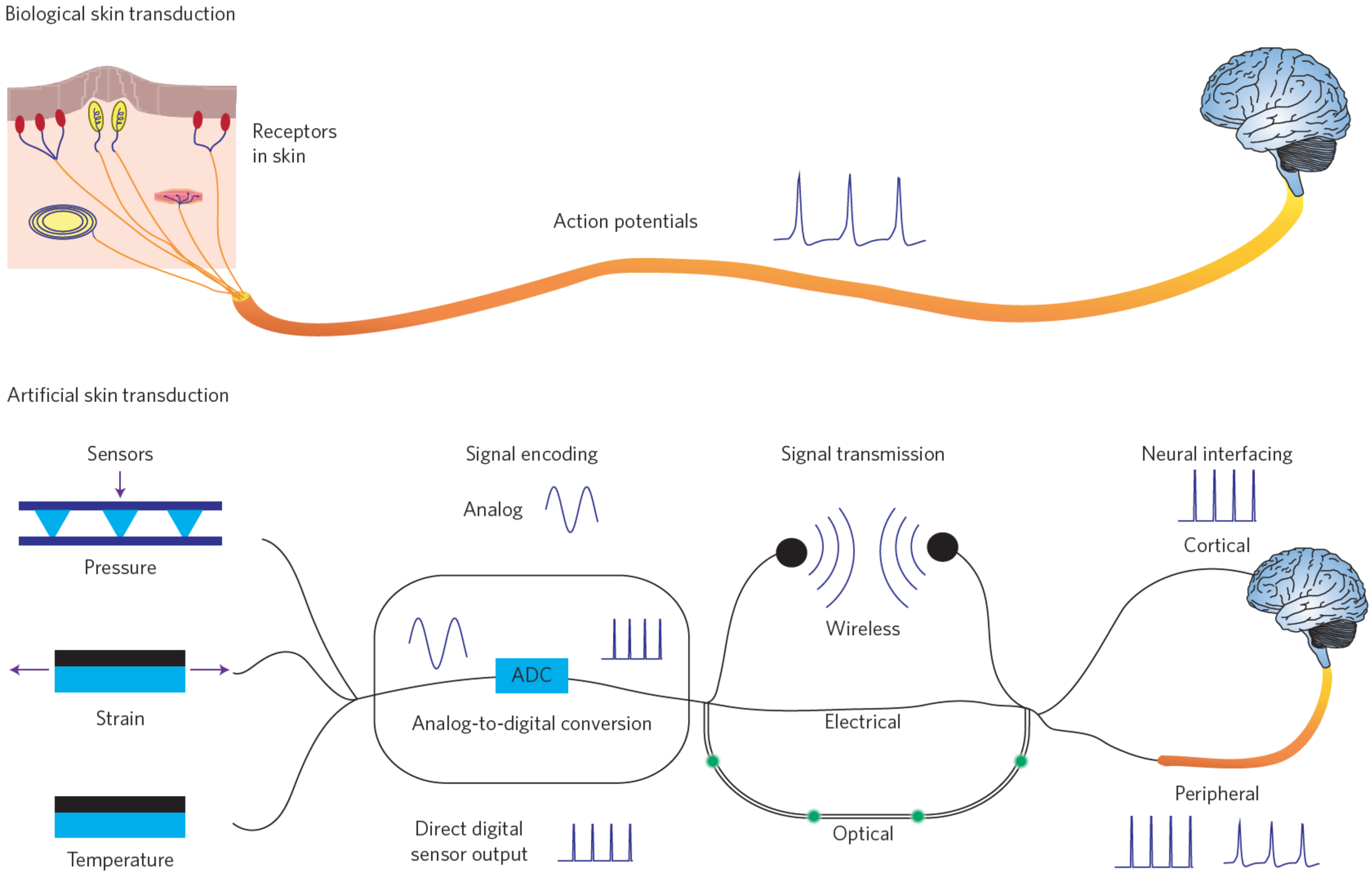


Figure 1.1 - Sensory feedback in biological skin vs artificial skin *[10]*

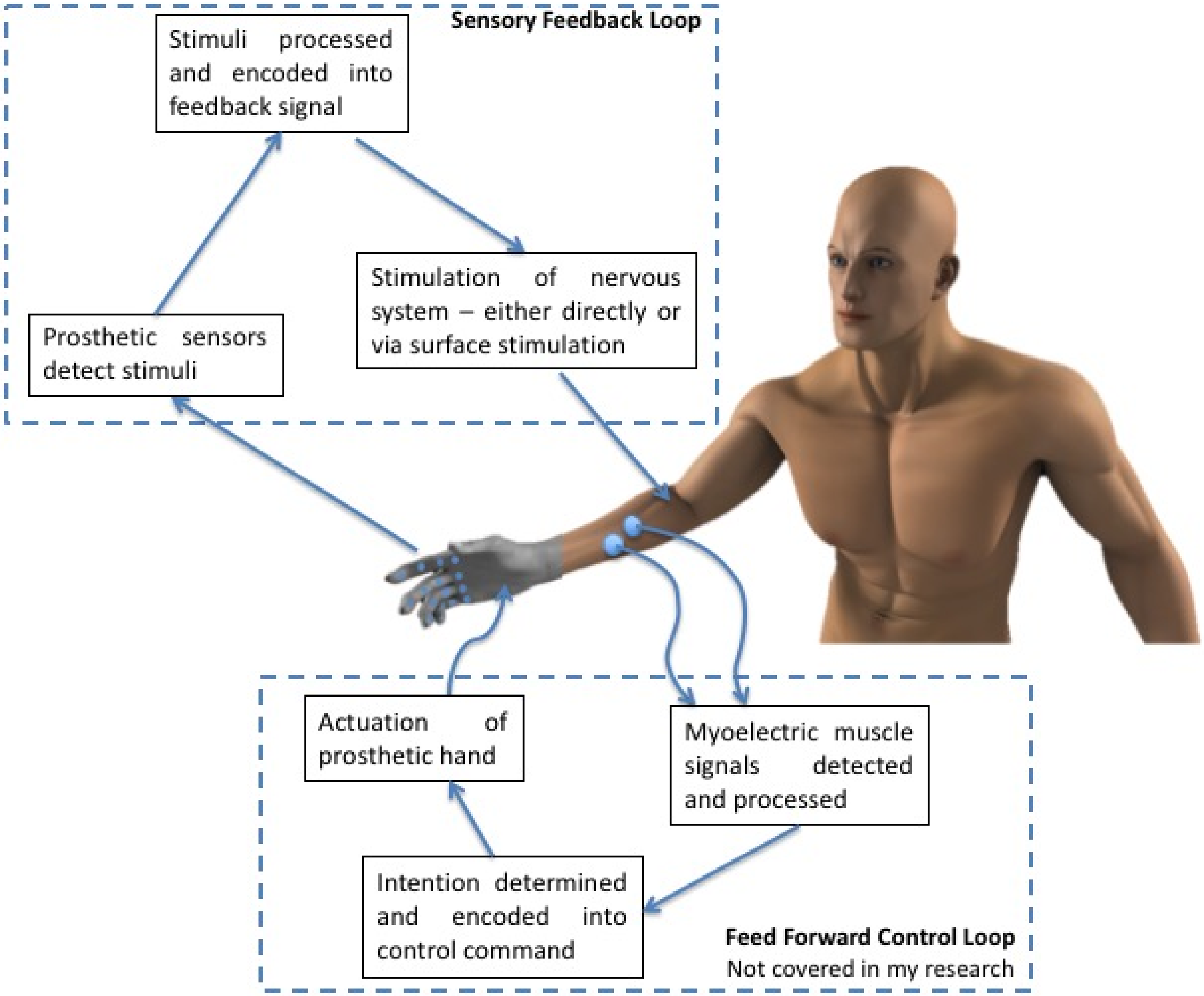


Figure 1.2 - Sensory Feedback and Feed Forward Control Loop

can understand. This encoded information is then sent to the CNS, either by direct stimulation of the PNS [11, 12] or CNS [13, 14] using electrode arrays as shown in Figure 1.1, or via activation of the mechanoreceptors at a location somewhere on the body.

Sensory feedback for prosthetic devices can be provided by applying a sensation to a different area of the body to represent the stimuli detected by the hand. This, however, requires the user to associate this sensation with the stimuli being detected. Having the feedback somatotopically and modality matched makes the feedback feel more natural and potentially easier to understand. In modality matched feedback, the stimulus is perceived as the same method of stimulation. For example, a pressing force on the finger is perceived by a feeling of pressure [15, 16]. An example of a non-modality matched feedback uses vibration on the skin to represent the detected pressure on a finger. Modality matching in non-invasive feedback can be achieved through mechanotactile feedback for grasping force, temperature feedback for temperature and vibrotactile feedback to communicate surface vibrations. In addition, electrotactile feedback can be used to create modality matched sensations by varying the stimulation waveform properties to create the feeling of either vibration, tapping and/or pressure/touch [17]. In somatotopical feedback, when the prosthetic pointer finger detects pressure, the communicated sensation is detected by the brain at the pointer finger. Although the invasive methods of targeted reinnervation [18] and nerve electrode interfaces [11, 12] communicate through somatotopical feedback, non-invasive methods can also apply mechanotactile, electrotactile, vibrotactile or temperature feedback to phantom hand maps [19-24] to produce somatotopical feedback. However, a recent study by Wijk et al. [25] has demonstrated our ability to, over time, associate sensations on predefined locations of our forearm with individual fingers, which is beneficial for non-somatotopical forms of sensory feedback.

Within literature, there are currently survey papers that have reviewed the methods deployed in sensory feedback, which have various degrees of invasiveness. A few surveys have examined the role of implants into the CNS [13, 14]. These methods, however, require a high level of invasiveness as subjects are required to undergo brain surgery to place the appropriate implant. Recent developments have also been made with direct nerve stimulation, which relies upon implants within the PNS. Normann and Fernandez’s review paper [26] focused on the variety of nerve arrays available and their use within control and feedback in prosthetic hands. Nghiem et al. [27] also provided a comprehensive overview of current types of feedback methods and prosthetic hands on market, with a large focus on direct nerve stimulation through the PNS. There have been recent studies which have demonstrated the longer-term stability of electrodes [28]. Further, using direct nerve stimulation has shown potential in communicating proprioceptive and grasping force information simultaneously [29].

Although the work involving PNS electrodes have shown some satisfactory early results [11, 12, 30-32], they are still in an early stage of development with limited numbers of test subjects in the laboratory testing that has been undertaken. In addition, at present there remains a reluctance among prosthetic hand users to undergo surgery for PNS electrodes [33]. To take advantage of the full potential of neural interfaces, current amputation techniques may also need to be changed [34].

The focus of this thesis is, therefore, on non-invasive methods (those not requiring surgery) for sensory feedback, and therefore excluded recent advances in sensory feedback that require surgery. Even though sensory perception can be communicated via non-invasive methods once a patient has undergone targeted reinnervation [18], these approaches are not considered in this thesis as patients are still required to undergo surgery in preparation. However, these techniques could also be potentially applied to those who have undergone targeted reinnervation [35].

In a recent review conducted by Benz et al. [33], prosthetic hand users felt a strong need for their prosthetic devices to be lightweight, as the weight of their current prosthetic hand leads to fatigue in the arm, shoulder and back. The users also raised concerns about their limited functionality and difficulty in performing precise tasks. In addition to the requirement of low weight, Cipriani et al. [36] have also suggested that transradial prosthetic devices need to be low in their power consumption so that they can be used all day, and have a low cost. Peerdeman et al. [37] developed a survey, which examined the requirements for feedback (and control) from a combination of interviews with professionals who regularly interacted with users (occupational therapists, physiotherapists etc.) and existing literature surveys. As a result, they produced the following feedback priorities, in hierarchical importance;

1) Continuous and proportional feedback on grasping force should be provided
2) Position feedback should be provided to user
3) Interpretation of stimulation used for feedback should be easy and intuitive
4) Feedback should be unobtrusive to user and others
5) The intensity and location of the feedback stimulation should be adjusted for each user

Cordella et al. [38] have also reported that future prosthetics should integrate tactile sensing, decrease the need of visual attention, increase the dexterity of the hand and number of grasp types.

1.2 Aim of This Thesis

The aim of this thesis is to develop non-invasive feedback methods that can communicate grasping force for three channels of information, relating to the grasping force on the thumb, pointer and the remaining three fingers. A mechanotactile method and an electrotactile stimulation method for sensory feedback are developed and presented, and their sensory feedback performance has been evaluated with the experimental data from able-bodied subjects to recognise the three channels of information separately and simultaneously. Finally, the effectiveness of both feedback methods in reducing grasping force of existing myoelectric prosthetic users when picking up a fragile object was demonstrated. This was realised with five prosthetic hands users. The scope of our work is limited to sensory feedback for transradial amputees.

1.3 Ethics

For each of the experiments undertaken within this thesis, written informed consent was obtained from all individuals participating, and ethical approval was obtained from the University of Wollongong Human Research Ethics Committee.

1.4 Statistical Analysis

The Statistical analysis presented within this thesis is performed using SPSS (IBM SPSS V24, IBM Armonk NY).

1.5 Participant Disclaimer

Some experiments, such as the orientation of the mechanotactile feedback (section 3.3.2) used the exact same group of participants for different tests, resulting in a repeated measures statistical analysis undertaken. Unless specified, the other experiments did not use the exact same participants as each other. However, some participants assisted in multiple experiments, which brought with them some previous knowledge of the device and experiment which has a potential to impact the results.

1.6 Principal Contributions

The principal contributions of this thesis are:

a) Development of an effective method of mechanotactile feedback which provides a combination of perpendicular pressure and skin stretch to improve touch recognition for sensory feedback. The performance of using this technique was demonstrated for the three channels of information, resulting in a high level of recognition for six different grips at two different intensity levels. Just noticeable difference results suggest that there are 12 discrete steps of recognisable intensity levels which do not statistically differ over the full stimulation range and are independent of the location of the stimulation on upper arm and lower arm.

b) Development of an effective approach to producing 3D printable reusable electrodes, which are also flexible and can conform to the profile of the human arm, for electrotactile simulation for sensory feedback. The resulting electrode was characterised, and its performance was measured. The manufacturing technique was shown to produce electrodes with a comparable performance to disposable electrodes and allow to customise the electrodes for the required shape, size and purpose.

c) Comparison of the impact of the electrode geometry on just noticeable difference, dynamic range, localisation, intensity, comfort of stimulation and type of induced sensation for electrotactile simulation for sensory feedback. These results demonstrate the advantages of the proposed concentric electrodes, particularly for stimulation on multiple arm locations simultaneously.

d) A comparison between the recognition and sensitivity of the upper arm and lower arm to mechanotactile stimulation and electrotactile stimulation. These results provide justification for applying sensory feedback stimulation to the upper arm, without any significant reduction in performance, to eliminate need for alterations to the prosthetic socket of the myoelectric prosthetic devices and leave the lower arm region for the EMG sensors used in controlling myoelectric prosthetic devices.

e) Development of a model relating the applied stimulation to the perceived intensity for the mechanotactile stimulation and the electrotactile stimulation. This model based on a linear relationship estimates the perceived intensity to be accurately applied to the user's arm, as per grasping force between a prosthetic hand and an object.

f) Testing the proposed mechanotactile method and electrotactile sensory feedback with five existing transradial myoelectric prosthetic hand users to evaluate the effectiveness of both feedback methods in reducing the grasping force the hand users are applying on an object, and improving the intuitive control of their myoelectric prosthetic hand. A purpose-built force measurement cube was used to measure the grasping force of existing prosthetic devices without any modifications required either in the socket or the prosthetic hand. The subjects effectively used the sensory feedback information to reduce their grasping force when gripping the force measurement cube attached with a range of mass

1.7 Organisation of Thesis

Chapter 2 presents a literature review on sensory feedback, with a focus on non-invasive sensory feedback methods for transradial amputees. These studies are analysed and a discussion on the research gaps in the current literature is provided.

Chapter 3 presents the design and performance evaluation of the mechanotactile sensory feedback method, based on the experimental data obtained from able-bodied subjects.

Chapter 4 describes an alternative method of developing reusable 3D printable concentric electrodes for use in electrotactile stimulation for sensory feedback. Further, this chapter presents a performance evaluation and comparison of the electrotactile sensory feedback through concentric electrodes and separated electrodes.

In Chapter 5, the upper arm was used as an alternative stimulation region to compare against the forearm, with performance comparisons made for the recognition rate, and sensitivity of electrotactile and mechanotactile sensory feedback methods to the stimulation site.

Chapter 6 details the experimental results from five myoelectric prosthetic hand users to determine how they benefitted the electrotactile and mechanotactile feedback to adjust the gripping force they applied while picking and placing the force measurement cube loaded with a range of mass.

Chapter 7 presents conclusions and recommendations for future research.

Chapter 2

Literature Review

2.1 Introduction

This chapter provides a comprehensive overview of different methods used and the recent developments in providing non-invasive sensory feedback for transradial prosthetic hands that exist within current literature. In addition, the challenges and opportunities associated with the non-invasive sensory feedback methods are discussed. The scope and constraints placed on the literature review are described in Section 2.2. Section 2.3 presents an overview of the various non-invasive stimulation methods. The use of these techniques applied to the phantom hand map and in hybrid stimulation techniques are detailed Section 2.4 and Section 2.5, respectively. A discussion on the common trends and gaps within the literature is presented in Section 2.6 and the existing gap that this thesis has focused on is outlined in Section 2.7.

2.2 Scope of Literature Review

This review is limited to non-invasive methods (those not requiring invasive operations such as surgery), and therefore does not discuss recent advances in sensory feedback that require surgery. Even though sensory perception can be communicated via non-invasive methods once a patient has undergone targeted reinnervation [18], these approaches are not discussed as part of this review as patients are still required to undergo surgery in preparation.

When conducting a systematic search of the literature, the following restrictions were, therefore, placed on studies to be included in this review;

- Focus on full hand prosthetic devices, not partial hand amputees, with the emphasis being on transradial amputees (amputation through the forearm).
- Focus on feedback methods to the user, not the sensors used to detect information within the prosthetic hand.
- Feedback to include the user as part of the feedback loop. Studies where the hand creates its own feedback loop without involving the user (such as camera to

automatically recognise appropriate grip [39], or automatically adjusting grip when slip occurs [40, 41]) are not included.

2.3 Non-Invasive Stimulation Methods

There are a variety of feedback methods that currently have been deployed within literature including the use of temperature [42, 43], vibration [44-54], mechanical pressure and skin stretching [15, 16, 55-60], electrotactile stimulation [61-74], audio feedback [75-77], and augmented reality [78, 79]. A mind map of the different feedback methods is shown in Figure 2.1. Some of these stimulation techniques have been explored [80-85]; whereas electrotactile, vibration and mechanical pressure have also been applied to phantom limb stimulation [19-24]. Each of these methods are discussed separately, with an assessment of the methodologies used and any challenges and opportunities that are involved in each technique. Studies with limited subjects and/or a lack of performance metrics have still been included to give an insight into the different approaches currently being explored within this area.

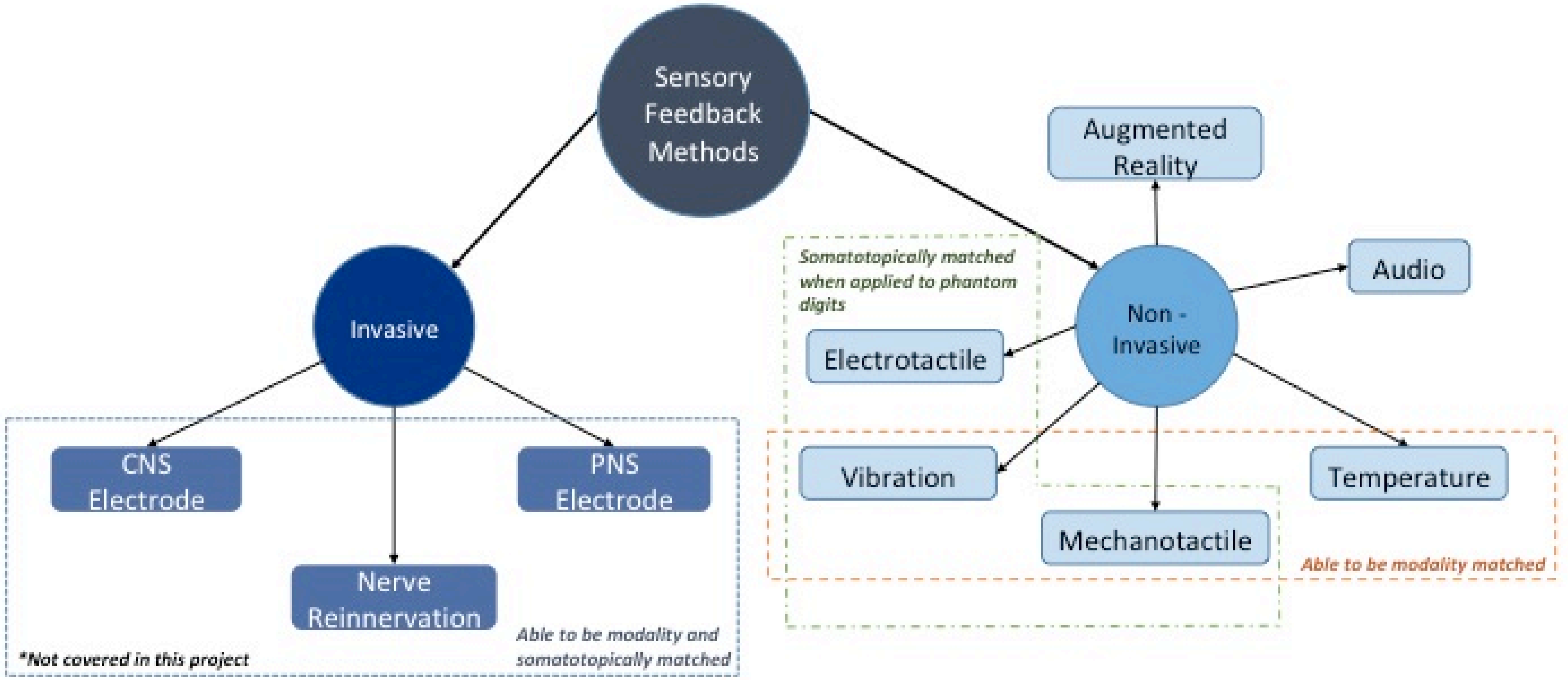


Figure 2.1 - Mind Map of Feedback Methods for prosthetic hands

2.3.1 Vibrational Feedback

Vibrational feedback typically uses small commercially available vibration motors, which are applied to the skin surface and activate the Pacinian corpuscle mechanoreceptors in the skin. These are usually small and light weight, as shown in Figure 2.2. The user learns to associate the vibration at that site with one of the senses from their prosthetic hand. Vibration has typically been used to communicate grasping force, however, a few studies have examined its role in communicating proprioceptive information [45, 53, 86], and some hybrid systems have used vibration to provide modality matched feedback on texture information [83, 87]. These studies only contain preliminary testing and further investigation into this form of modality matched feedback is required. Using vibration as a source of force feedback has been demonstrated to have improvements over using vision alone as a feedback tool [44, 47, 49], but some literature suggests that this benefit is only visible during inadequate feedforward control [88]. However, the drawbacks of using vibration include: an extra delay of approximately 400ms to begin generating vibration and a limited bandwidth being available [89]. In addition, it has also been suggested that perception of vibrational frequency can be affected by how tightly a vibration motor is attached [90], which raises difficulties in predictive and reliable sensory feedback.

(a) (b)

Figure 2.2 - Examples of Vibrators used in Vibrotactile feedback (a) Spatially Placed Vibrators [44] © 2016 IEEE; (b) Coin Vibration Motors [46] © 2016 IEEE.

The use of three vibration feedback devices to communicate grasping force and grasping angle (separately) from a prosthetic hand to its user was examined by Yamada et al. [44]. They concluded that by incorporating vibration feedback, there was a reduction in cognitive load (also known as cognitive strain or mental effort), required to pick up objects compared to using visual feedback alone, however, this was not consistent across all the subjects. Deploying vibrotactile stimulation has also been shown to provide an amputee with a higher sense of embodiment in their prosthesis [5] when undertaking an experiment modelled with the rubber hand illusion. However, vibrational feedback requires users to undergo training in order to develop the full benefit [91]. Ninu et al. [45] examined the performance of vibrational feedback on the forearm to help improve grips for picking up objects. This study examined 13 subjects (11 able-bodied subjects and two amputees), using a commercially available myoelectrically controlled prosthetic arm. The authors used a constant frequency with varying amplitude to communicate velocity of the closing hand, and simultaneously modulated the amplitude and frequency of vibrations to the grasping force. The researchers demonstrated that using vibrotactile feedback to communicate hand velocity, point of contact and grasping force without visual feedback was enough information for the subjects to pick up objects. However, they also noted that the hand velocity was the most important feature and the addition of grasping force feedback had a minimal effect. Other studies have also demonstrated that the use of vibrotactile feedback results in an improvement in grasping objects [92-94]. Nabeel [46] developed a pressure sensor that could be applied to the finger tip of any prosthesis and implemented a vibration feedback system to the forearm of the user. Their test was only conducted on one amputee, who, however, recognised the improvements as a result. The authors also suggested that more training would be required to increase its performance.

Rosenbau-Chau et al. [47] demonstrated that recognition of grip force could be improved by using vibrotactile feedback, however, the impact was large for some users and not for others. The feedback system had three stages of force; low, medium and high; represented by differing pulse frequencies and strengths. They proposed that by incorporating more than three stages of feedback, the system could become more unreliable. The effectiveness of sinusoidal, sawtooth and square vibrational waves on amputees with upper limb prosthetic devices was examined and sinusoidal waveform performed the best. The proximal region of the residual limb was determined to be the most comfortable by the subjects and achieved the highest accuracy. Desensitisation occurred after 66 seconds and the authors proposed to use a series of pulses, rather than continuous vibrations, to achieve a higher success rate and reduce desensitisation. They also concluded that training increased the success of vibrotactile feedback. This research group also examined the effect of varying pulse frequency in vibrotactile feedback to communicate grasping force [48]. The six subjects overall had positive responses to the use of vibrational feedback, with one subject commenting that he enjoyed shaking his 5-year-old granddaughter's hand knowing that he was not squeezing too tight.

Clemente et al. [49] also demonstrated a practical method of using vibrational feedback to control grasping force. The researchers placed pressure sensor thimbles on an existing prosthetic hand and used a cuff on the upper arm to provide vibrotactile feedback to the subjects for a period of 60ms when the hand either made or broke contact with an object. Their data showed that the subjects using vibrotactile feedback achieved a higher success rate picking up blocks without breaking than those only using visual feedback. The subjects maintained this performance whilst using this prosthetic hand with vibration feedback at home over a period of four weeks. Hanif and Cranny [50] demonstrated the use of intermittent vibrational pulses as a possible method to communicate different surface textures. The feedback system detected different surface textures using a piezoelectric sensor at the fingertip and sent vibrational frequencies corresponding with each of the four surfaces. They only demonstrated the production of differing frequencies visually, as the method was not tested on any subjects and their perception of these varying vibrational frequencies.

Li et al. [51] examined the use of vibrators on a sports glove on the other hand to provide force feedback from the prosthetic device. This enabled the user to identify the level of force on the back of the corresponding finger on the other hand quickly. Each vibrator had three different intensities to represent either a soft, medium or a hard level of force being applied to the prosthetic device. Their results showed that users quickly learnt how to interpret the vibrations, and their performance in picking up objects improved as a result. However, it may be not as effective outside of the laboratory when two hands are required to complete tasks.

Raveh et al. [52] examined the effect of vibrotactile feedback on the visual attention required in performing tasks with a prosthetic hand. Subjects drove a simulated car whilst performing basic tasks with their myoelectric controlled hand. Their data showed no improvement in the required visual attention to complete basic tasks. However, their subjects were new to myoelectric control, received minimal training on vibrotactile feedback and the system only used vibration feedback to communicate contact. The authors hypothesised that the subjects may not have had enough time to begin to trust the feedback and, therefore, still felt they needed to rely on visual cues.

Hasson and Manczurowsky [53] examined the effect of vibrational feedback on providing position and velocity proprioception information. They only tested moving a virtual arm to a target position, not in grasping objects. However, their results showed no improvement from vibration feedback.

Witteveen et al. [86] also compared using vibrotactile feedback to communicate grasping force and the amount of hand closure. Both forms of vibration feedback improved performance in grasping objects, however, there was no significant difference between the two different approaches.

Vibrational feedback offers an affordable and lightweight system of feedback that users prefer it over electrotactile feedback [95]. One limitation, however, is the delay in stimulation and since the feedback delay can decrease embodiment [96, 97]. This may attribute towards some of the negative results.

A comparison summary of the different studies using vibrational feedback are shown in Table 2.1.

Table 2.1 - Comparison of Vibrotactile Feedback Studies

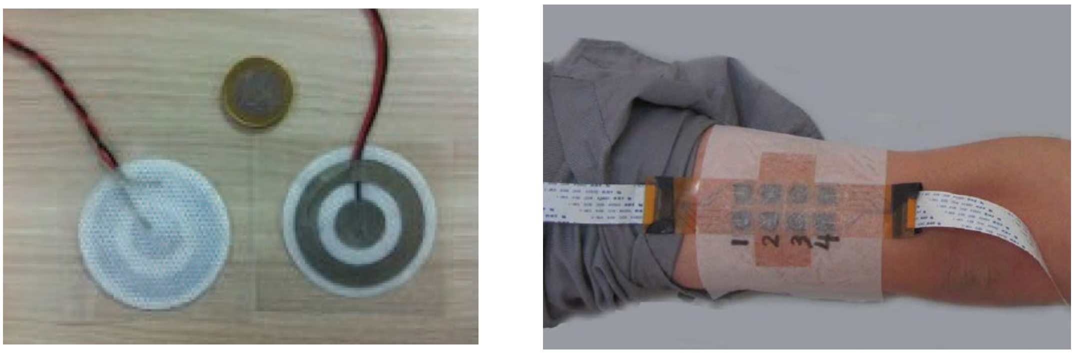
Reference	Type of Hand	Location	Number of subjects/ Number of Amputees	No feedback channels and Sensor	Range and number of feedback levels	Performance
Yamada et al. [44] 2016	Myoelectrically controlled 1-DOF robotic hand gripper	3 on bicep – 1 for each level	5 / 0	1 - Single force sensor for grasping force OR Potentiometer for aperture angle	PWM range matched to strength of grasping force PWM range matched to aperture angle	3 Subjects demonstrated 10% lower cognitive load from vibrotactile feedback on grasping force 4 Subjects demonstrated a lower cognitive load (10-40%) from vibrotactile feedback on aperture angle
Ninu et al. [45] 2014	Myoelectrically controlled Gripper	1 on forearm	13 / 2	2 - Single Force Sensor and Velocity sensor	Varied Amplitude to match closing velocity Varied Frequency and amplitude simultaneously proportionally to grasping force	Performance in achieving desired grasping force for Low and High Force levels: Visual Hand feedback – 76% & 52% Velocity and Contact Vibration feedback (No visual) – 74% & 33% Velocity, Force &Contact Vibration Feedback (No visual) – 84% & 53% No visual or Vibration Feedback – 19% & 22%
Nabeel [46] 2016	Body powered prosthetic hand	2 on the forearm	7 / 1	1 - Single Force sensor	PWM range corresponding to sensor values 0-255	94% of able bodied subjects could use feedback to determine whether bottle was half or completely full of water

Rosenbau-Chau et al. [47] 2016	Myoelectrically controlled Robotic Hand (opens and closes)	2 on the forearm below the elbow	6 / 6	1 - Single force sensor on thumb	Varying Pulse rate and Frequency to induce Light, Medium and Strong.	Vibrational feedback improved grip force accuracy by 129% for light grip force, 21% for medium grip force. No statistical improvement for strong grip force
Chaubey et al [48] 2014	Myoelectrically controlled Robotic Hand (opens and closes)	12 locations on biceps (1 activated at a time	7 / 7	1 - Pressure sensor on target object	Linearly mapped PW to pressure signal input	Vibrational feedback significantly improved grasping force error at 60% maximum force but not at 80% maximum force
Clemente et al. [49] 2016	Myoelectrically controlled Robotic Hand (opens and closes)	2 within a cuff on biceps	5 / 5	1 - Pressure sensor on thumb and index finger	60ms length vibration when hand made or broke contact with object	Less blocks were broken with vibrotactile feedback on compared to no vibrotactile feedback ($p<0.001$)
Hanif and Cranny [50] 2016	N/A - Computer Simulation	N/A -Computer simulation	0 / 0	1 - 1 piezoelectric sensor at fingertip	Changed length of on and off pulses to represent roughness	N/A – No performance measures listed
Li et al. [51] 2016	N/A – simulated sensations for perception test	5 Vibrators, 1 on the back of each finger of the opposite hand mounted in a sports glove	5 / 0	5 (1 each finger) – simulated sensations for perception test of forces on individual fingers	3 Values for each finger – Strong Medium and Weak	N/A – No performance measures listed
Raveh et al. [52] 2017	Myoelectrically controlled artificial hand	8 Vibrators wrapped around the forearm	43 / 0	1 - 2 Force Sensors to determine force	Full strength to indicate contact pressure above	No statistical difference in visual demand when using vibrotactile feedback to communicate contact of object

					predefined threshold, otherwise off	
Hasson and Manczurowsky [53]	Virtual Arm, EMG controlled angle	1 Vibrator on Forearm	9 / 0 (9 in each of the 3 groups, 27 total)	1 – Calculated position of Arm OR Calculated velocity of arm	Amplitude modulated to Velocity OR Amplitude modulated to Position	No significant improvement resulting from velocity based vibrotactile feedback or position based vibrotactile feedback in achieving desired arm position
Walker et al. 2015 [54]	Simulation of holding an object, controlled by a stylus	1 vibrator on bicep	23 / 0	2 – Force Feedback on stylus and objects slipping acceleration through vibration	Vibration mapped to objects acceleration due to slip	Recovery of slipping objects - Visual feedback only 90% - No feedback 42% - Vibrotactile feedback 80%
Witteveen et al. 2015[83]	Computer simulated hand controlled through mouse scrolling	An array of 8 vibrators for aperture 1 Vibrator on forearm for force	10 / 10	1 – Hand Aperture OR Grasping Force	Position of tactor activated representing hand opening / 8 different levels of intensity represent grasping forces	No significant differences between performance in grasping objects when using either Hand Aperture Feedback OR Grasping Force Feedback

2.3.2 Electrotactile Feedback

Electrotactile stimulation for sensory feedback contains no moving parts and has an efficient power consumption. Multiple features can be easily and reliably controlled including the intensity, pulse width, frequency and location of stimulation (with multiple electrodes), which leads to a higher bandwidth being available [98]. The electrodes are slim and lightweight, shown in Figure 2.3, and electrotactile stimulation is safe and comfortable to use. However, each person's minimum sensation threshold and pain threshold is different and the perception of electrotactile information changes with the placement of the electrodes [66], with movements as small as 1mm having an influence [99]. In addition, skin conditions can also influence the comfort and dynamic range of electrotactile stimulation [99].



(a) (b)

Figure 2.3 - Examples of Electrotactile Electrodes: (a) Concentric Electrodes *[61]*, (b) four pairs of electrodes *[65]* © 2016 IEEE.

Not only does this mean that re-calibration of thresholds are required every time electrodes are placed on the user; but that the pulse width, frequency and amplitude may need readjusting to achieve the same perception each time. In addition, potential problems arise from interference between myoelectric sensors for control and electrotactile stimulation, however, this has begun to be addressed within literature [72-74].

Electrotactile stimulation induces a sensation by directly stimulating the primary myelinated afferent nerves in the dermis [100]. Concentric electrodes limit the current spread and can increase localisation and discernibility of the induced sensation [98, 100] and can reduce the resulting noise on the EMG used for myoelectric control [73]. Despite their advantages, only approximately half of the electrotactile feedback systems examined use them [63, 67, 69, 71-74], which may impact upon their performance.

A few studies have demonstrated the benefit of using electrotactile feedback, such as [61]. The authors used a constant 100Hz frequency and 3mA intensity sent to electrodes on the dorsal side of the forearm to communicate the force applied to a joystick controlled robotic hand. The Pulse Width (PW), however, was varied from 20% above their sensation threshold to 20% below their pain threshold to communicate the force level detected on the robotic hand by a pressure sensor. Their research indicated that training with electrotactile feedback helped improve the user's recognition of grip strength when picking up a variety of objects. Isakovic et al. [62] also demonstrated that using electrotactile feedback helped users learn to regulate myoelectric control of grasping force quicker. Schweisfurth et al. [63] showed that using electrotactile stimulations to feedback the EMG control signals outperformed force feedback in achieving a target initial grasping force. In EMG feedback, the processed myoelectric control signal was sent to the subject via electrotactile stimulation from beginning of trial to 0.35 seconds after contact with the object. In force feedback, the system detected the grasping force by a pressure sensor on the prosthetic finger, and then sent an electrotactile signal corresponding to this level of pressure from contact until 0.35 seconds after contact. The range of pressures was matched to a varying amplitude and PW of the stimulation current, up to 90% of the pain value. The subjects achieved closer to a target force when receiving electrotactile feedback based on EMG control signals than electrotactile feedback based on grasping force.

Shi and Shen [64] demonstrated the effect of varying intensity, frequency, PW on electrical stimulation and the effect on subject's perception. The authors individually varied the PW, frequency and amplitude, and applied these stimulation currents through 9mm diameter electrodes to the subject's arm. The data showed that pulse width could be varied within 0.2-20ms; intensity within 0.2mA-3mA; and frequency within 45-70Hz. These ranges delivered an appropriate level of feeling in the subject and proportionally increased grades of intensities felt by the subject.

The work by Xu et al. [65] compared communication of pressure, slip, and pressure with slip information through electrotactile stimulation, with visual feedback of lights representing the sensors information, and no feedback. They tested 12 subjects, with six of them being amputees, using a simulated environment gripping and picking up objects. Four pairs of electrodes placed on the forearm, shown in Figure 2.3b, was used to deliver the electrotactile feedback. The frequency was set to a constant value of 100Hz, and the PW was regulated from 0μs to 500μs to communicate any detected changes in grasping force. To communicate slip, the authors sent the electrotactile stimulation through a sequence of the four available pairs of electrodes (1-2-3-4-1 etc.), where the time interval between changing electrode pairs represented the amount of measured slip in the hand grasp, ranging from 20ms to 500ms. The data showed that pressure + slip feedback through electrotactile feedback performed the best out of sensory feedback methods, however, visual feedback outperformed all of them in grasping failure rate and ability to keep the grasping force as constant as possible. The authors also identified a performance difference between amputees and able-bodied test subjects, but they also recognised that their able-bodied subjects used their dominant hand and were younger than their amputee subjects.

Although there has been success in incorporating one feedback channel with electrotactile communication for one grasp, prosthetic devices often control more than one grasp. Therefore, more than one feedback channel is beneficial when closing the loop in feedback control with the user. Choi et al. [66] demonstrated that subjects could distinguish two channels of electrotactile feedback on their biceps. However, they did not connect the system to any sensors but instead showed that users could distinguish between the two channels. They also demonstrated that better recognition was achieved when using intermittent stimulation on both channels (switching between the 2), rather than both channels being on at the same time, resulted in better recognition.

Patel et al. [67] used four electrotactile feedback sensors to map the configurations of a 4-Degree of Freedom (DOF) prosthetic hand. They maintained a constant PW and intensity but varied the frequency. Four channels of feedback were used on the subjects to help them either control individual finger flexion, or different hand grasps, with myoelectric control. However, tests were only conducted on able-bodied patients, with feedback being on the opposite arm to the myoelectric sensing. Patel et al. used multiple electrotactile channels to communicate proprioception whereas Pamungkas and Ward [68] demonstrated the potential of using six electrotactile feedback channels for force feedback. Six electrotactile locations were used to communicate information from pressure sensors contained on a glove controlled robotic hand. Five of the locations were used to communicate force acting on the prosthetic fingers, and the other location was used to communicate the force acting on the palm. For each finger, three frequencies (100Hz, 60Hz and 30Hz) were used to represent the force on each phalange, and 20Hz was used for the palm. Only the highest pressure value from each finger was sent to the fingers' corresponding electrode to avoid confusion from multiple frequency signals. Their data showed that the subject learnt how to use the feedback appropriately to pick up a range of objects, as they had more success when alternating between picking up heavy and light objects. Their one subject also stated that they preferred electrotactile feedback to only using visual feedback when operating the robotic hand.

Strbac et al. [69] demonstrated a different electrode design that enabled users to distinguish up to 16 stimulation locations, with up to five different frequencies at once, to provide multiple levels of feedback. Test results from a small number of able-bodied and amputee subjects demonstrated that six electrodes with four different frequency signals could be identified with more than 90% accuracy by the subjects after minimal training. The highest number of channels recognised was from one able-bodied subject identifying all 16 pads after two hours of reinforced learning. Six amputees also recognised eight different stimulation patterns that corresponded to different movements, with an average accuracy of 86%. The authors stated that their next development was to integrate this approach into the prosthetic socket connection with an automatic calibration (minimum amplitude set at just above recognition and maximum just below maximum pain threshold), but this is yet to appear in any published literature. They also noticed that there was a large difference between individual user's performances, indicating that this approach could work well for some but not others. Although this study only used simulated signal patterns instead of feedback from sensors, it demonstrated the potential of using a multichannel electrotactile feedback as a potential interface for prosthetic hands.

A human hand does not contain pressure sensors, which communicate isolated forces back to the user, rather, nerves are embedded throughout the whole skin and each translates a different feeling to the brain. Franceschi [70] investigated possibilities of communicating information from artificial skin by translating information from 64 pressure sensors into 32 electrotactile electrodes on the subject's arm. They only conducted tests on able-bodied subjects and the users could detect movement directions easily, but had trouble determining individual positions. Hartmann et al. [71] also demonstrated that the recognition of simple movement patterns using electrotactile arrays could be learnt by able bodied subjects through training. This opens future possibilities to be explored that could provide the prosthetic user with richer sensory feedback.

Surface electrodes are predominantly used for myoelectric control of prosthetic devices. One problem that arises is the interaction of the electrotactile stimulation with the myoelectric surface electrodes. In experiments, by using myoelectric control on the opposite arm to the one being stimulated, this effect is sometimes avoided, but in practical applications interference needs to be addressed. One approach undertaken is time-division multiplexing for myoelectric control and electrotactile stimulation [72]. The system constantly switches between myoelectric control and electrotactile stimulation so that the two are never occurring at the same time, with a minimal reduction in performance. Other studies have reduced noise interference through redesigned electrodes. Jiang et al. [73] demonstrated a specially designed electrode for electrotactile stimulation that, in combination with signal processing and optimisation of the stimulation waveform, limited the noise interference from electrotactile stimulation feedback with the myoelectric control. Xu et al. [74] produced a new flexible electrode design that incorporated stimulation and EMG recording at the one site simultaneously without interference. Their redesigned electrodes were used to control the robotic hand and transmit electrotactile stimulation for sensory feedback. The electrotactile stimulations were proportional to grasping force and they resulted in a lower error rate when picking up a plastic bottle. Xu et al. also demonstrated the use of tactile funneling illusion in proprioceptive feedback, whereby stimulation was perceived at a location between two electrodes, depending upon the intensity of each of the corresponding electrode. The higher the ratio of intensity of one electrode in the pair, the closer the perceived stimulation will be towards that electrode.

Electrotactile feedback shows potential for a quick and easily controllable method of feedback that users can identify multiple sites of feedback at once. However, currently this sensation is often referred to as a tingling feeling and occasional feeling of touch. Further research is required to be undertaken on the particular waveform characteristics to improve the induced sensation to the subject to achieve a more natural feeling of pressure, as has been demonstrated in direct nerve stimulation [101]. Additional care and analysis is also required to ensure that minimal interference occurs with the EMG interface used for myoelectric control, so it does not significantly impact the control of the prosthetic device

A comparison summary of the different studies using electrotactile feedback are shown in Table 2.2.

Table 2.2 - Comparison of Electrotactile Feedback Studies.

Reference	**Type of Hand**	**Location**	**Number of subjects**	**No of feedback channels & Sensor**	**Range and number of feedback levels**	**Performance**
Jorgovanovic et al. [61] 2014	Joystick controlled 1-DOF gripping simulation	2 bipolar electrodes on dorsal side of forearm	10 / 0	1 - Simulated force	PWM to correspond to grasping force, increments of 50μ from 20% above minimum sensation to 20% below pain threshold	Success picking up objects: - 72% with feedback - 40% without feedback
Isakovic et al. [62] 2016	Myoelectrically controlled 2-DOF prosthetic hand	Electrode array – 16 cathodes and one anode	3 / 3	1 - Grasping force	6 discrete Force levels represented by different combinations of electrodes being activated	94% accuracy in recognition of 6 discrete force levels Reduction of error from 24.4% to 15.6% when using feedback
Schweisfurth et al. [63] 2016	Myoelectrically controlled 2-DOF prosthetic hand, but only 1 movement was used	4 electrodes on forearm	11 / 1	1 - Grasping force	8, 4 electrodes each with 2 frequency options	EMG feedback resulted in 21% lower error than force feedback
Shi and Shen [64] 2015	N/A – just rating feelings from feedback method	1 stimulation electrode on wrist	1 / 0	1 - No sensor – testing sensations	Intensity ranging from 0-3mA, .1ma increment; Frequency ranging 1Hz-100HZ, 5HZ increment;	No quantitative measurement - Increasing electrotactile sensation can be brought on by increasing amplitude, frequency or pulse width

					Pulse width 1ms – 50ms, 1ms increment	
Xu et al. [65] 2016	Simulated Hand	4 pairs of electrodes on biceps. 1 pair used at a time	12 / 6	2 - Slip sensor and/or Grasping force	Pressure feedback: PWM 0µs to 500µs Slip feedback: Time between switching pairs of electrodes used 20ms-500ms	Pressure, Slip, Pressure +Slip feedback outperformed no feedback ($p<0.05$) in achieving desired grasping force and grasping time. However, Visual Feedback outperformed all tactile feedback methods in achieving desired grasping force and outperformed pressure as well as slip feedback in time of grasping force ($P<0.05$)
Choi et al. [66] 2016	N/A – simulated sensations for perception test	2 pairs of electrodes, 1 stimulating electrode on either side of upper arm between biceps and triceps	10 / 0	3 - simulated sensations for perception test	4 different levels on 2 channels – resulting in 15 different stimuli across the 2 channels. An additional on/off state was communicated for the thumb through offset pulses	2 channel stimulation resulted in recognition accuracy of 52.9% for simultaneous stimulation and 73.8% for intermittent stimulation
Patel et al. [67] 2016	Myoelectrically controlled simulated prosthetic hand with 4-DOF and Myoelectrically	4 concentric electrodes on forearm	9 / 0 (Virtual Finger) 8 / 0 (Virtual Grasp)	4 - Simulated thumb flexion, thumb opposition, index flexion and middle/ring/litter finger flexion	Linearly mapped frequency from 3 to 30Hz to represent flexion level	Finger flexion recognition – 94% Grasp pattern recognition – 79%

	controlled Robotic hand with 4-DOF		11 / 0 (Robotic Finger)			
Pamungkas and Ward [68] 2015	Data glove controlled humanoid robotic hand	6 electrodes on forearm	1 / 0	6 - Pressure force on each of the 5 fingers and on the palm	4 intensity ranges (Zero, light, light, medium, high) corresponding to change in intensity	No Measurements listed
Strbac et al. [69] 2016	N/A – simulated sensations for perception test	16 electrodes on flexible cuff placed on forearm	16 / 6	8 different patterns used as channels - simulated sensations for perception test	Tested 3, 4, 5, & 6 different frequency intervals	The concentric electrode pattern had a recognition rate of 99%, 95%, 80% and 74% for 3, 4, 5 &6 different frequency levels respectively
Franceschi [70] 2015	N/A – simulated sensations for perception test	32 channel electrode array placed on forearm	5 / 0	10 different movement patterns on sensors - Array of 60 pressure sensors	10 different movement patterns – on/off no in between	Direction recognition ~ 90% Orientation recognition ~70% Position recognition ~ 60% (Measurements approximated from graph)
Hartmann et al. [71] 2014	N/A – simulated sensations for perception test	8 electrodes placed on the forearm	2 / 0	8 different locations - Array of 60 pressure sensors	Intensity of stimulation used to help provide location	Subjects could recognise each of the 8 locations with 92% accuracy
Dosen et al. [72] 2014	Myoelectrically controlled simulation	1 concentric electrode on the forearm	9 / 0	1 - Simulation error	Intensity proportional to error amplitude	RMS tracking error increases from ~13% for normal feedback to ~21% with a 100ms delay. Overshoot increased from ~13% for normal feedback to ~27% feedback with 100ms delay.

						(Measurements approximated from graph)
Jiang et al. [73] 2014	6 EMG electrodes to detect noise	1 stimulated electrode on upper arm	1 / 0	1 - Constant Simulation current	On and off value, compared noise from 6 different types of EMG electrodes	Filtering increases Signal to Noise ratio from 15dB to 43dB
Xu et al. [74] 2016	Myoelectrically controlled virtual arm to move elbow joint	2 electrodes on biceps	1 / 0	1 - Position of simulated elbow joint	1 pressure sensor - Intensity of stimulation proportional to gripping force OR Virtual Arm angle mapped to varying intensity of 2 electrodes	No measurements given

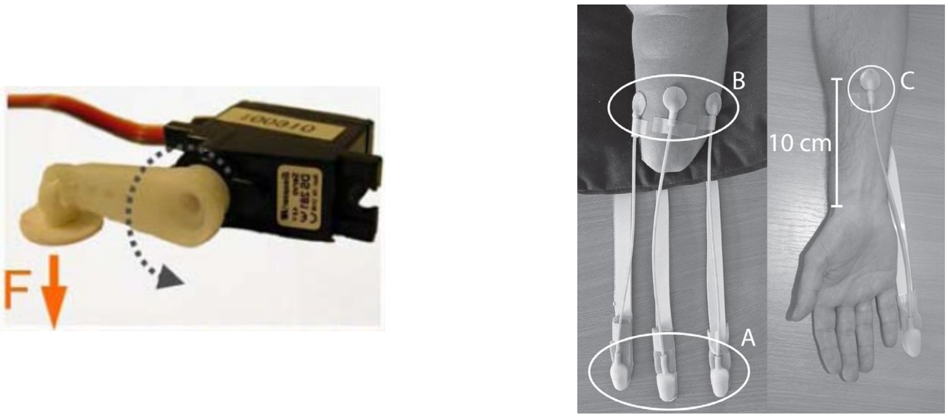
2.3.3 Mechanotactile Pressure Feedback

Preliminary tests conducted by Aziziaghdam et al. [55] showed that an object could be identified as either hard or soft from the acceleration response obtained whilst tapping an object. Pressure feedback on the clavicle bone could then be used to communicate this acceleration profile to the user. Some other studies have examined the role of wearable haptic devices on feedback. Fallahian [102] demonstrated improvement of fine grasp control using a small mechanical servo on the upper arm of one amputee picking up fragile object with their myoelectric prothesis. Morita et al. [56] used a winding belt motor on the upper arm to communicate grasping force feedback of a myoelectric controlled prosthetic hand. The speed of winding also gave the user an indication of the hardness of the object. Casini et al. [57] demonstrated the application of distributed haptic force from a combination a pressure and skin stretch via a cuff on the bicep, as shown in Figure 2.4, to help a user determine an object as hard, medium or soft Godfrey et al. [15] also examined the use of a feedback band around the arm to provide information to users on grasping force. However, although a trend was observed in grasping force modulation, this was not statistically significant compared to visual feedback. Also, shown in Figure 2.4, all these haptic feedback devices were quite large and provided unnecessary bulk to prosthetic devices.

Figure 2.4 - Pressure feedback cuff *[57]* © 2015 IEEE.

Schoep et al. [103] developed an alternative approach using two mechanical tactors applying normal forces to the skin driving through gears and a cable system. Their setup was tested on one transhumeral amputee operating a myoelectric prosthesis whilst picking up an object with embedded load cells. Using feedback, the subject was able to correctly identify the two feedback locations and incorporate the feedback to reduce the maximum force and average grasping force. However, they took longer to complete the task.

Antfolk et al. [16] demonstrated the use of five servo controlled mechanical pressure devices, shown in Figure 2.5a. This allowed the user to recognise touch within individual digits and three levels of pressure feedback. The authors noticed, however, that it was not helpful for improving grip recognition, but they suggested more training was necessary to overcome confusion between neighbouring areas. Antfolk et al. also suggested the use of improved actuators and placing them on the phantom hand map to further improve results. The use of silicon bulbs, shown in Figure 2.5b, has been shown as a novel way to apply mechanotactile feedback [58] to communicate touch and levels of grasping pressure. Three silicon bulbs were attached to the user's forearm and they recognised three distinct zones and up to two levels of force. The authors, however, recognised that the ideal location for the bulbs was within the phantom digit zones and they had positive feedback from a pilot test on one amputee with distinct phantom digit locations.



(a) (b)

Figure 2.5 - (a) Mechanical Pressure Feedback device *[19]* © 2013 IEEE, (b) Silicon Bulb Mechanical Feedback *[58]*

Akhtar et al. [59] explored the use of linear skin stretch on the forearm to provide feedback on the flexion of fingers. As one of the three motors for thumb, index, remaining three fingers, respectively, drives the tendon in the corresponding finger, it pulls a contact pad attached to the forearm to increase the skin tension. Subjects described this as comfortable over the whole experiment and the data indicated an improved grasp recognition whilst using the feedback. However, testing was only conducted on able-bodied subjects and the contacts pad required tape or adhesive glue to attach to the skin.

Bark et al. [104] examined the use of rotational skin stretch for proprioceptive feedback. Although subjects had trouble with using absolute position sensing, the authors concluded that rotational skin stretch had some benefit for proprioceptive information when controlling movement, for an EMG controlled prosthetic hand. This would, however, only be suitable for feedback with 1-DOF. Wheeler et al. [60] then investigated its application to proprioceptive feedback of an elbow of a myoelectric transhumeral prosthesis. The authors found that the use of the rotation skin feedback resulted in a lower target error and visual demand.

Battaglia et al. [105] used skin stretch from a rotating mechanical rocker on the bicep of the arm to communicate proprioception information for a 1-DOF hand. Using this feedback, 18 healthy subjects were able to discriminate between different spherical sizes with an average accuracy of 73.3%. Rossi et al. [106] also provided proprioception information for a 1-DOF hand through the use of a haptic device encompassing a wheel rolling up and down the user's forearm. Their data from 16 able-bodied undertaking one of three different testing conditions (no haptic feedback, linearly mapped feedback, logarithmically mapped feedback subjects demonstrated an improvement in distinguishing between four different diameters. Five subjects receiving logarithmically mapped feedback had the highest success rate, achieving an average of 75%. Further, Rossie et al. undertook testing on one amputee who achieved a success rate of 90% receiving logarithmically mapping when receiving information on their residual limb. However, when testing the feedback device on the upper arm whilst connected to the prosthetic, the amputee subject described the experience as uncomfortable and the testing was stopped.

A comparison summary of the different studies using mechanotactile feedback are shown Table 2.3.

Table 2.3 - Comparison of Mechanotactile feedback.

Reference	Type of Hand	Location	Number of subjects/Amputees	No of feedback channels & Sensor	Range and number of feedback levels	Performance
Aziziaghdam et al. [55] 2014	N/A Simulated sensations from tapping mechanism for perception test	Mechanical Actuator on Clavicle Bone	1 / 0	1 - Acceleration on tapping mechanism to simulate tapping finger on object	3 - 1 for hard, semi hard and soft	No performance measurement
Morita et al. [56] 2014	Myoelectrically controlled prosthetic hand with thumb and only 1 finger	Mechanically winding belt on bicep	5 / 0	1 - Pressure and displacement of finger to calculate hardness	Speed of winding corresponds to hardness	Hardness sensitivity of 0.59N/mm
Casini et al. [57] 2015	Robotic hand – SoftHand Pro	Pressure and skin stretch cuff worn on bicep	1 / 0	1 - difference in current to close hand compared to look up table	3 levels of hardness	100% accuracy in distinguishing between 3 levels of hardness
Godfrey et al. [15] 2016	Robotic Hand – SoftHand Pro	Pressure and skin stretch cuff worn on bicep	6 / 0	1 - Estimation on force based on current drawn	5 levels of tightness mapped to grasping force	Measurements only displayed in graphical form
Antfolk et al. [16] 2013	N/A only tested recognition of sensations	5 servo motor controlled actuators on forearm	10 / 5	Up to 5 - Pressure sensor from prosthetic hand	Up to 3 levels of pressure	(Amputee and Able Bodied) Localisation: 75.2% & 89.6%; Pressure level: 91.7% & 98.1%; Grip recognition: 58.7 & 68.0%
Antfolk et al. [58] 2012	N/A only tested recognition of sensations	Bulbs attached to the forearm	32 / 12	Up to 3 - simulated sensations for perception test	2 levels of pressure	Pressure: 90% & 80%; Localisation: 96%

Akhtar et al. [59] 2014	Myoelectrically controlled prosthetic hand	Contact pads on forearm	5 / 0	3 - Driven by motors that drive thumb, index and middle fingers	Range of 13mm of movement to represent fingers range of motion	Single Finger identification error: NF – 17.75%, VT – 8.58%, Skin stretch – 9.79%
Wheeler et al. [60] 2010	Myoelectrically controlled virtual arm	Rotational Skin Stretch on back of triceps	15 / 0	1 – rotational angle of elbow	±60° of elbow range corresponds to ±45° skin rotation	Error rate only displayed in graphical form; 23% reduction in visual demand using skin stretch device
Battaglia et al. [105] 2017	Myoelectrically controlled SoftHand	Rocker on the bicep	18/0	1 – aperture of hand grip	Hand opening linearly mapped to 0-60° rocker rotation	Discrimination of different sized spheres with an accuracy of 73.3%
Rossi et al. [106]	Underactuated prosthetic device (not connected)	Wheel on forearm	16 (split into 3 test) /1	1 – aperture of hand grip	Hand opening mapped (linearly or logarithmically) to when on 40mm path	Average discrimination accuracy of 75% from 4 different diameter objects
Schoep et al. [103]	Myoelectrically controlled prosthesis	2 mechanical tactors on upper arm	1 transhumeral amputee	2 Force on thumb and pointer	Force on sensors was exponentially mapped to the tactor position.	Reduction in maximum and average grasping force
Fallahian [102]	Myoelectrically controlled prosthesis	1 Mechanical Crank on upper arm	1 transradial amputee	1 Grasping force on FSR	Force up to level of crushing object was mapped to comfortable sensation range	Reduction in breakages of fragile test objects

2.3.4 Temperature feedback

Sensory feedback has mainly been deployed to communicate and identify a prosthetic device's gripping force and the flexion of its fingers [37]. Temperature, however, provides users with extra information about their environment, and potential dangers or warnings that involve heat. Producing heat on the upper arm to correspond with temperature detected at the prosthetic hand was the only method of temperature feedback found within literature. Cho et al. [42] used a disguised temperature sensor in a prosthetic hand to sense temperature and wirelessly transmit the measured temperature range. The corresponding temperature was then communicated to the subject via a Peltier element on their opposite hand. The subjects distinguished between high, warm and cold temperature setting with reasonable accuracy, however, it drew upon a large amount of power. Ueda and Ishii [43] also examined the use of temperature feedback via a Peltier element. However, they developed a prediction algorithm based upon initial measurements to speed up their response times. This resulted in a quicker response time when providing temperature information to the subject. Although these results are positive, with the desire for minimal weight and power consumption in prosthetic devices, and a higher need for other sensations sent to the user, this feedback method may not be deeply investigated until further advances are made with force and proprioceptive feedback. A potential focus of research would be to incorporate temperature feedback with another feedback method so that they can occur simultaneously, since it is not a priority to occur by itself.

A comparison summary of the different studies using temperature feedback are shown in Table 2.4.

Table 2.4 - Comparison of Temperature Feedback.

Reference	Type of Hand	Location	Number of subjects / Number of Amputees	No of feedback channels & Sensor	Range and number of feedback levels	Performance
Cho et al. [42] 2007	Externally driven prosthetic hand (Myoelectric controls bypassed)	Peltier element placed on users left hand	6 / 0	1 - Temperature Sensor on prosthetic finger	3 Temperature values – Hot, Mild and Cold	Temperature recognition of 3 temperature ranges with an accuracy of 96.7%
Ueda and Ishii [43] 2016	Myoelectrically controlled prosthetic hand with thumb and only 1 finger	Peltier element placed on user's bicep	10 / 0	1 - Temperature Sensor on prosthetic finger	5 Temperature values - Hot, Lukewarm, not much, a little cold, cold	Temperature recognition of 5 temperature ranges with an accuracy of 88%

2.3.5 Audio Feedback

Wilson and Diren [75] demonstrated the potential of deploying audio to communicate sensory feedback from a prosthesis. They examined the test subject's ability to interpret modulation of two audio channels to control a computer simulation. Their data showed that the subject could interpret two channels, but there was a 602ms delay and the audio feedback resulted in a high cognitive load. The subjects accurately completed the simulation and their success improved with training, although they rated two frequencies playing simultaneously as difficult to interpret. Gibson and Artemiadis [76] showed that a subject could use auditory feedback alone to pick up objects with a robotic hand. Within their study, the variance in volume represented the level of grasping force and the varying frequency corresponded with the location of two different regions of the hand. After training, subjects incorporated feedback to pick up and identify objects. In another approach, Gonzalez et al. [77] utilised triads to communicate the movement of a robotic hand. The sound of cello corresponded to the force on the thumb and a piano sound represented the force on index finger. The subjects were also able to use the audio feedback to help improve their movements and control when grasping objects. Each of these audio feedback experiments was conducted within the laboratory, and given their high cognitive load required, further investigation is required to determine their effectiveness whilst background noise is occurring.

A comparison summary of the different studies using audio feedback are shown in Table 2.5

Table 2.5 - Comparison of Audio Feedback.

Reference	Type of Hand	Location	Number of subjects / Number of Amputees	No of feedback channels & Sensor	Range and number of feedback levels	Performance
Wilson and Diren [75] 2016	N/A – Sensations simulated for perception test	Headphones	8 / 0	2 - Sensations simulated for perception test	Range of Frequency from 300-3400Hz, Amplitude from 50-65dB and beat frequency 0-15Hz	Frequency and beat modulation resulted in a mean squared error of 0.0406 and delay of 522ms for frequency, and a mean squared error of 0.0658 and a delay of 602ms for the beat frequency channel
Gibson and Artemiadis [76] 2014	5 Fingered Myoelectrically controlled prosthesis	Headphones	12 / 0	2 - Pressure Sensor on prosthetic fingers and position of robotic hands	Amplitude corresponded to grasping force, 2 different frequencies used to represent 2 different hand locations	3 groups of 4 subjects with their own individual mappings of frequencies to hand locations. They identified objects with and accuracy of 83%, 87% and 100% respectively.
Gonzalez et al. [77] 2012	Tendon driven robot hand	Headphones	8 / 0	3 - Pressure Sensor on prosthetic fingers and position of robotic hands	8 different piano triads to recognise different hand configurations. Amplitude corresponded to grasping force	Subjects achieved a lower duration completing tasks with audio feedback (37.52s vs 43.67s) and used a lower grip force (0.17V vs 0.25V)

2.3.6 Augmented Reality

Markovic et al. [78] used Google glasses to communicate the aperture angle, contact time, grasping force and EMG strength for sensory feedback of a prosthetic hand to its user. Subjects used the visual feedback to improve their task performance when moving objects that required various strengths without breaking them. The subjects noted, however, that they typically only glanced at the information and did not use EMG strength signals.

Clemente et al. [79] also examined the use of augmented reality (AR) for sensory feedback for prosthetic devices. They communicated information through an ellipse, with the axis lengths corresponding to grasping force and angle of grasp closure onto the user's AR glasses. The authors changed the proportions of the grip force and grip closure feedback and examined if the users changed their movements accordingly. The data indicated that the subjects relied on the force feedback but not the closure feedback, however, in the tasks they were constantly looking at the objects, so the grip closure information was redundant. The grasp angle feedback may only become important when doing tasks without looking at the hand as closely. Although there was a lower variability in initial grip force using the feedback, there was a significant increase in the duration of time required to pick up the object. This suggests that although performance repeatability can be increased with augmented feedback, it increases the cognitive load required from the user.

A comparison summary of the different studies using augmented reality feedback are shown in Table 2.6.

Table 2.6 - Summary of Augmented Reality Feedback.

Reference	**Type of Hand**	**Location**	**Number of subjects / Number of Amputees**	**No of feedback channels & Sensor**	**Range and number of feedback levels**	**Performance**
Markovic et al. [78] 2017	Myoelectrically controlled prosthetic hand	Graphical feedback displayed in Google glasses	20 / 0	4 - Aperture angle, Pressure sensor (contact), Pressure Sensor (Grasping Force), EMG sensors	Hand aperture on a linear scale, Contact with object displayed as on/off, Grasping force and EMG force on a linear scale	The improvement in speed and accuracy of grasping from using augmented reality feedback compared to no augmented reality was statistically significant.
Clemente et al. [79] 2017	Myoelectrically controlled prosthetic hand	Feedback in ellipse form displayed in Google glasses	8 / 0	2 – Pressure sensor (force), potentiometer (angle)	Horizontal axis of ellipse representing grip closure Vertical axis of ellipse representing grasping force	Smaller variability in initial grip force with feedback provided. Significantly larger duration in picking up the object with feedback provided

2.3.7 Stimulation of Phantom Hand

Amputees can not only experience phantom limb pain, but also experience phantom limb sensations as explored in [107]. Amputees can have locations known as phantom digits that, when touched, trigger a sensation that corresponds in their brain to touching their missing finger. Phantom digits provide a pathway for a natural and efficient communication for a variety of sensations that would not require any training. However, these phantom digit locations are not located in all amputees and their location and size can vary amongst individuals, as shown in Figure 2.6. Wang et al. [108] suggested that the distribution of phantom digits is located along the stump nerves. This approach, therefore, cannot be applied uniformly to all patients, as it is unsuitable for those without phantom digits. It will also require individual customisation for those who possess them, however, prosthetic sockets are customised to each individual and mapping stimulators to phantom digits could potentially be part of this process. D'Alonzo et al. [5] were able to demonstrate that by stimulating phantom digit locations during a rubber hand experiment, they were able to promote a sense of self attribution with the rubber hand.

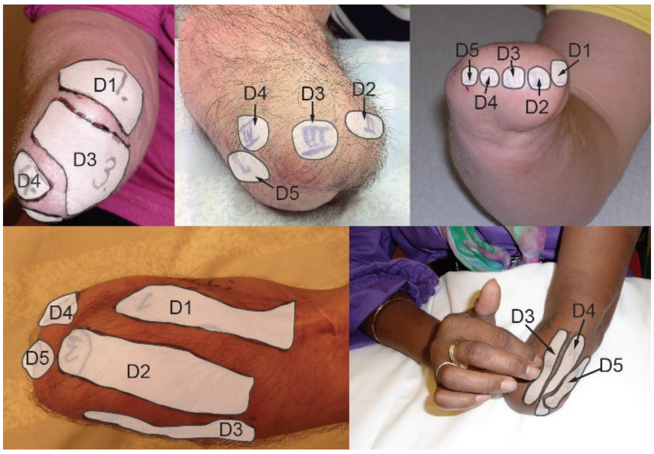


Figure 2.6 - Examples of Phantom Hand Maps and their corresponding Phantom digit locations *[107]*

Ehrsson et al. [6] examined 18 amputees, out of which 12 had a phantom hand map. These 12 subjects underwent a human rubber hand illusion test whilst their phantom digit locations were stimulated. Their experimental data showed that stimulating these sites induced a sense of ownership with the prosthetic hand. In addition, another study [7] examined two amputees undergoing a functional MRI scan whilst completing the rubber hand illusion test. The MRI scans showed that stimulating these phantom locations activated the corresponding finger location within the brain.

Antfolk et al. [19] examined multi-site stimulation through vibrotactile and mechanotactile feedback with amputees that had complete phantom hand maps. They found that those with a complete phantom hand map recognised multiple sites of feedback with a higher success rate than those who had an incomplete or no phantom hand map. Zhang et al. [20] demonstrated that using Somatotopical (phantom digits) Feedback (SF) outperformed Non-Somatotopical feedback (NF) on the upper arm in electrotactile stimulation feedback. The SF was faster in response time (600ms), had a lower cognitive workload and achieved a higher recognition rate. One channel of feedback resulted in similar recognition rates for NF and SF; however, three channel SF performed as effectively as one channel of NF. Five feedback channels in SF performed marginally lower and was equivalent to the three channels of NF; although the authors suggested that interference and crossovers with the different electrodes due to their size may have affected the performance of the five channel SF feedback. Zhang et al. also recommended to combine SF and NF for those who do not have complete mapping and/or have limited stump size to place the electrodes.

Li et al. [24] examined the effect of electrode size and spacing on stimulating a phantom hand map with Transcutaneous Electrical Nerve Stimulation (TENS.) They demonstrated that the bigger electrode, the wider range of sensations produced. However, a higher current is then required and further space between electrodes is needed. They concluded that having an electrode sizing of 5-7mm was a good compromise based on their preliminary investigations.

TENS can induce sensations in these phantom digit locations for all fingers [21]. This study demonstrated the effect of varying pulse width, frequency and current density, and their corresponding sensation induced. The feelings of pressure, pressure + vibration, vibration, tingling and numbness in the corresponding finger location were induced through TENS applied to the phantom digit location. Liu et al. conducted a follow up study to show that these signals could be induced by pressing on a tactile sensor on each prosthetic finger [22]. Chai et al. [23] went on demonstrating that these sensations were stable for an 11-month period for nine amputees. Testing was only conducted using one electrode and further investigation was required on simultaneous stimulation of multiple electrodes. Furthermore, a thorough investigation into creating sensations that correspond to varying levels of grasping force has not yet been reported in published literature. Although initial data suggests that variations in the TENS PW, amplitude and frequency ranges, could induce varying intensity of sensations [21].

A comparison summary of the different studies using phantom limb stimulation for sensory feedback are shown in Table 2.7

Table 2.7 - Comparison of Phantom Limb Stimulation.

Reference	Type of Hand	Location	Number of subjects / Number of amputees	No of feedback channels & Sensor	Range and number of feedback levels	Performance
Antfolk et al. [19] 2013	N/A – simulated sensations for perception test	On the forearm, up to 5 vibrotactile or 5 mechanotactile	8 / 8	5 – simulated sensations for perception test	Only on and off values were used	Complete Phantom Map: Mechanotactile – 100%, Vibrotactile – 91% Partial Hand Map: Mechanotactile – 61%, Vibrotactile – 49%
Zhang et al. [20] 2015	N/A – simulated sensations for perception test	Up to 5 electrodes On Phantom digits for SF and Upper arm for NF stimulations	7 / 7	1, 3, and 5 channels tested – simulated sensations for perception test	Changed frequency from 1-75Hz	Position: SF 97%, NF 90% Strength: SF 86%, NF 80%
Chai et al. [21] 2013	N/A – simulated sensations for perception test	1 Stimulation electrode on user's phantom digits	2 / 2	5 (only 1 tested at a time) – simulated sensations for perception test	Current: 0 to Upper limit (UL), .125mA increment, PW: 20µs to UL, 10µs increment Frequency: 1Hz to UL, 10Hz increment	Measurements displayed in graphical form
Liu et al. [22] 2015	N/A – simulated sensations for	1 Stimulation electrode On user's phantom digits	2 / 2	5 (only 1 tested at a time) - Pressure sensors to	Current varied proportional to pressure, from 0mA to 25mA	Measurements displayed in graphical form

	perception test			detect force on prosthetic finger		
Chai et al. [23] 2015	N/A – simulated sensations for perception test	1 stimulation electrode On user's phantom digits	19 / 11	5 (only 1 tested at a time) – simulated sensations for perception test	Current: 0 to Upper limit (UL), .125mA increment, PW: 20µs to UL, 10µs increment Frequency: 1Hz to UL, 10Hz increment	Measurements displayed in graphical form
Li et al. [24] 2015	N/A – simulated sensations for perception test	2 electrodes placed on PTP area	6 / 6	2 – simulated sensations for perception test	On and off value	(electrode size – discrimination distance) Parallel electrode: 12mm–39.0mm, 9mm-36.1, 7mm-31.3mm, 5mm-27.2mm Perpendicular electrode: 12mm-36.1mm, 9mm-33.5mm, 7mm-29.1mm, 5mm-26.5mm

2.3.8 Combining modalities: Hybrid tactile feedback methods

The literature discussed thus far has only communicated one type of sensation at a time, this can often lead to an ability to only communicate one sensation at a time. A few studies have examined the potential of using multiple feedback methods simultaneously. This may be to improve the recognition rates and/or range of one type of stimuli, or create the ability to communicate two different stimuli simultaneously.

D'Alonzo et al. [80] demonstrated that subjects could identify nine levels of stimulation through a hybrid feedback of electrotactile or vibrotactile stimulation, shown in Figure 2.7, compared with either mode in isolation. These same authors also went on showing that subjects could identify patterns from four stimulation devices, that used a combination of electrotactile and vibrotactile stimulation, with a higher accuracy than similar sized vibrotactile devices [81]. However, testing was only conducted on able-bodied subjects. D'Alonzo et al. suggested that their results were limited by the size of electrodes and the performance may improve if their size was reduced. Combining mechanical pressure and vibration has also been explored [82], but only an experimental prototype was built, without any testing performed on subjects. The device also appears very bulky.

Jimenez and Fishel [83] examined a prosthetic finger with a temperature, vibration and force sensor incorporated for sensory feedback. The weight of an object was translated into squeezing pressure on the arm, the temperature was produced on the bicep of the arm and surface textures were communicated through vibration feedback. The subject accurately perceived the mass, temperature and roughness of the objects but each modality was only tested one at a time. The subject also suggested that the vibrational feedback mechanism was too distracting. Li et al. [84] also presented a new design for a feedback mechanism that combined vibrational feedback with mechanical pressure into a small, lightweight and power efficient module that can be used as part of arrays. However, at the time of preparation of this chapter, there was no literature on the testing of this system on a person.

Motamedi et al. [85] examined the perception of pressure and vibration feedback at the same time. They found that pressure by itself was perceived with the highest accuracy, followed by pressure and vibration at the same location, pressure and vibration at different locations and lastly vibration by itself performed the weakest.

Hybrid tactile feedback systems are still in an early stage of development, with half of the studies examined only displaying a prototype without undertaking any experimentation. Further testing is therefore, not only required to be undertaken to determine a person's ability to recognise two different feedback systems simultaneously, but to also examine the effect on the cognitive load. More experimental data on recognition rates and cognitive load could help determine if hybrid tactile feedback systems can be successfully incorporated into a feedback loop to improve the user's control and embodiment with their prosthetic hand.

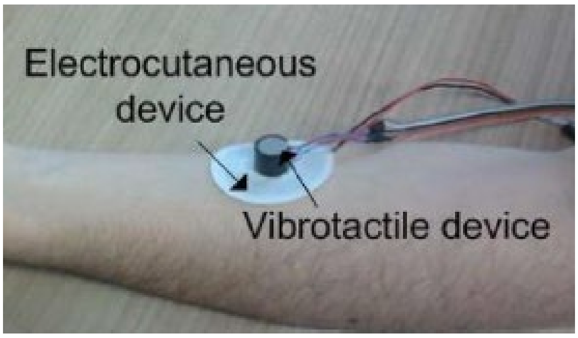


Figure 2.7 - Combination of Electrotactile and Vibrotactile feedback [80] © 2014 IEEE.

A comparison summary of the different studies using hybrid stimulation techniques are shown in Table 2.8

Table 2.8 - Comparison of Hybrid Stimulation Techniques.

Reference	Type of Hand	Location	Number of subjects / Number of amputees	No of feedback channels & Sensor	Range and number of feedback levels	Performance
D'Alonzo et al. [80] 2014	N/A – Sensations simulated for perception test	1 Electrotactile/Vibrotactile combination stimulator on the Forearm	10 / 0	1 – Sensations simulated for perception test	9 levels of intensity	Recognition of 9 levels using hybrid setup – 56% & 72%, vibrotactile only – 29%, electrotactile only 44%
D'alonzo et al. [81] 2014	N/A – Sensations simulated for perception test	A combination of 3 electrotactile stimulators and 2 vibrotactile stimulators spread across 3 locations on the forearm	10 / 0	1– Sensations simulated for perception test	5 different single channels (representing each finger), 5 different grasp patterns	Single Finger: Hybrid – 98%, Electrotactile-94%, Vibrotactile 1 – 89%, Vibrotactile 2 – 73% Pattern: Hybrid – 77%, Electrotactile-79%, Vibrotactile 1 – 77%, Vibrotactile 2 – 69%
Clemente et al. [82] 2014	No Testing conducted – just prototype built			2 - Contact made/break & grasping force	5 levels of pressure Vibration frequency range from 5Hz to 200Hz	No performance measurement as no testing undertaken
Jimenez and Fishel [83] 2014	Robotic Gripper	Force Tactor, Vibration Tactor and Temperature Tactor - all on bicep	1 / 1	3, only 1 tested at a time: Temperature, Force and Vibration sensor	Temperature range +- 3C Vibration varied amplitude	Measurements only displayed in graphical form

					Pressure 0-200kpA of air muscle pressure	
Li et al. [84] 2016	No Testing conducted – just prototype built			2 channels of 15 actuators - No Testing, just prototype built	Max Vibration 240Hz, Max Pressure 4.4N	No performance measurement as no testing undertaken
Motamedi et al. [85] 2017	N/A – Sensations simulated for perception test	Applied to forearm. Normal stress and Vibration applied at same location OR 6cm away from each other	14 / 0	1 channel of feedback	3 values of normal stress, 3 values of vibration feedback	Measurements only displayed in graphical form

2.4 Discussion and Summary

Each of the sensory feedback methods has been successful in providing extra information to the prosthetic user, often enabling them to make better decisions in the control and use of their prosthetic hand. Although some studies included subjects' reflections on their use of the prosthetic device with sensory feedback at home [49, 58], the majority of testing, however, has been completed under laboratory conditions, often involving an external computer. During simulated sensation testing, all concentration is on perception of the sensation. However, during everyday tasks, perception requires detection and understanding whilst undertaking other tasks, thus minimisation of cognitive load becomes more important. To use these feedback methods within a real-life context, thorough testing outside the laboratory (such as home, outside, office, restaurant etc.) is required to examine success rates with the normal background noise and distractions that occur in everyday environments. For example, will audio feedback be able to be heard as easily with background noise, or will vibrational feedback be able to be felt whilst undertaking everyday tasks?

A large amount of testing was completed on the dominant arm of able-bodied subjects. However, when this same feedback is fed to the forearm of an amputee, the perception, sensitisation and response can be different.

Both electrotactile stimulation and vibrotactile stimulation suffer from the disadvantage that perception can not only vary between people, but also by the location of applied stimulation. This may affect the practicality of systems for use day after day. There has also been no examination on whether repeated application produces the same results. Vibrotactile feedback is dependent upon the pressure of the tactor against the skin, and the tactor reapplication by the user therefore may not result in consistent sensations. In addition, when using multiple vibration tactors or electrotactile electrodes, electrode locations may affect their repeatability. Recalibration may be required each time the user places it on, and moving locations may impact the cognitive load required in using the device. Further research into these areas is required.

Another challenge is to communicate the location of the feedback. Within current literature, most studies only communicate the force that represents one location on the digit. When grasping an object, however, subjects may want to feel the difference between force on the fingertip and force on the inside of the finger. Vibrotactile and electrotactile arrays appear to be one potential solution to this problem.

There is a large amount of different approaches to test sensory feedback methods. Some studies have only tested simulations to ensure correct perception, whilst others have incorporated a myoelectric controlled prosthetic hand. There are also variances within the number of degrees of freedom employed, the number of channels and levels of feedback, as well as the type of sensation being communicated. These differences can make a performance comparison between studies difficult. However, in addition, it also appears that different approaches may be required for different prosthetic users [69] and for different prosthetic hands. For example, if a prosthetic hand only contains a simple grasping motion, then using a pressure cuff or single vibration motor could be well suited. Although current pressure cuffs are quite bulky, the winding belt mechanisms provide a simple and easy to learn feedback device for single DOF devices. However, if feedback is required for all five fingers, then an approach of using phantom digits or electrotactile stimulation could be better suited. Commercial prosthetic hands are further developing their dexterity and degrees of freedom [109] and will therefore require multiple channels of feedback. Additionally, a recent literature analysis by Cordella et al. [38] identified that increasing the dexterity and degrees of freedom in the prosthetic hand is a high priority. Initial results for vibrotactile and electrotactile arrays have shown some successes as users have been able to identify locations and movements. However, more research should be undertaken to connect them with a prosthetic hand through sensory feedback.

Comparative testing is required to compare the effectiveness in improving control and user comfort when using the various methods. This testing would be required to be specific for each type of prosthetic hand. For example, one set of experiments on feedback mechanisms for a 1-DOF hand and then another series of tests for a 3-DOF hand, as they may not produce the same result. These would need to incorporate not only grasping performance, but also measures from the subjects on areas such as: comfort, ease of use and cognitive load.

Electrotactile stimulation of the phantom hand [20-24] has shown some potential for sensory feedback in a multiple DOF system. Current literature suggests that by stimulating the phantom digits, it can provide up to five separate somatotopically matched feedback pathways that feel natural to the user. By using electrotactile stimulation, it provides a lightweight, low-power, larger bandwidth mechanism that can be easily controlled. However, phantom hand maps are not located on every amputee, and their location and number of digits appear to be unique to each person. Initial testing has only stimulated one site at a time, and no testing has been reported on stimulating multiple phantom digits at once. Graczyk et al. [12] has reported a predictable linear relationship between perceived intensity, amplitude, frequency and pulse rate in intraneural stimulation. Further testing is required to determine if this same relationship exists within phantom digit stimulation.

As previously discussed, the top two feedback priorities for prosthetic hand users are force and proprioceptive feedback. Initial research on proprioceptive feedback has had mixed results. Hasson and Manczurowsky [53] concluded that providing position information through vibrotactile feedback did not result in any improvement. Blank et al. [110] concluded from their data that proprioceptive feedback alone improved the performance of a 1-DOF grasping task when no visual cues were available. When visual cues were available, however, the feedback only improved tasks with a moderate level of difficulty. The authors suggested that for precise tasks, other tactile cues were required as well. Pistohl et al. [111] also examined the role of proprioceptive feedback. Subjects controlled a cursor with EMG on one arm and fed proprioceptive information to the other user's arm using a robotic manipulator. The proprioceptive information was beneficial to the user when no visual information was available, but did not benefit the user when visual information was available. However, both Bark et al. [104] and Wheeler et al. [60] concluded that rotational skin stretch had some benefit in providing proprioceptive feedback, but only for 1-DOF actuator such as an elbow joint. Similarly [105] also demonstrated success in providing proprioceptive information for a 1-DOF hand. Further research is therefore required to provide proprioceptive information for hands with multiple degrees of actuation in the fingers.

At present, the majority of literature has focused on using feedback to send one sensation at a time. Using a single method to communicate more than one sensation may be difficult for the user to understand or result in a high cognitive load for the user. An effective approach could be to use multiple feedback methods to communicate combinations, with each feedback method communicating a different sensation, either simultaneously or by constantly switching between the two modalities running concurrently. There have been some contradicting results on a person's ability to understand multiple sensory feedback cues. Ajoudani et al. [87] demonstrated multiple cues being used successfully, with mechanical pressure cuff to communicate pressure forces and vibrational feedback to communicate texture information. However, in a study undertaken by Kim and Colgate [18], their subject showed a lower performance picking up a virtual object when receiving shear forces through vibrations at the same time as receiving pressure feedback on grasping force, although this experiment was only performed with one subject with five sets of trials. Other multimodal feedback systems [12, 20-24, 80-84] have shown capability, with initial testing demonstrating that users could distinguish multiple channels of information sent simultaneously. This could provide a method that allows for multiple channels of information to be provided back to the user to make informed controlling decisions on their prosthetic hand.

Both electrotactile stimulation of the phantom hand and multimodal sensory feedback are only at initial stages of testing, with only simulated perception (pre-generated feedback values) being examined, rather than feedback based of information detected from sensors embedded in the prosthetic hand. Further testing is required to determine whether these feedback mechanisms improve the user's ability to take part in the control loop.

Examination of effectiveness of sensory feedback techniques needs to progress away from being done in isolation from the control system. In the case of electrotactile sensory feedback, interference may occur and compromises may need to be made in the feedback or control system's performance to enable them to work together at the same time, as reported in [72]. In addition, as shown in Figure 2.8, it may be optimal for two sensory feedback loops to exist, one to the controller and one to the user. This is because currently there are limited pathways to effectively transmit all stimulations back to the user. Too much information may cognitively overload them or incorporate too long of a delay. Instead when minor alterations are required, such as during an object slipping, a higher performance may result from the prosthetic controller regulating the constant grasp rather than incorporating the user. However, further testing in this area is required to ensure the correct balance is achieved for improving grasping performance, user comfort, cognitive load and embodiment.

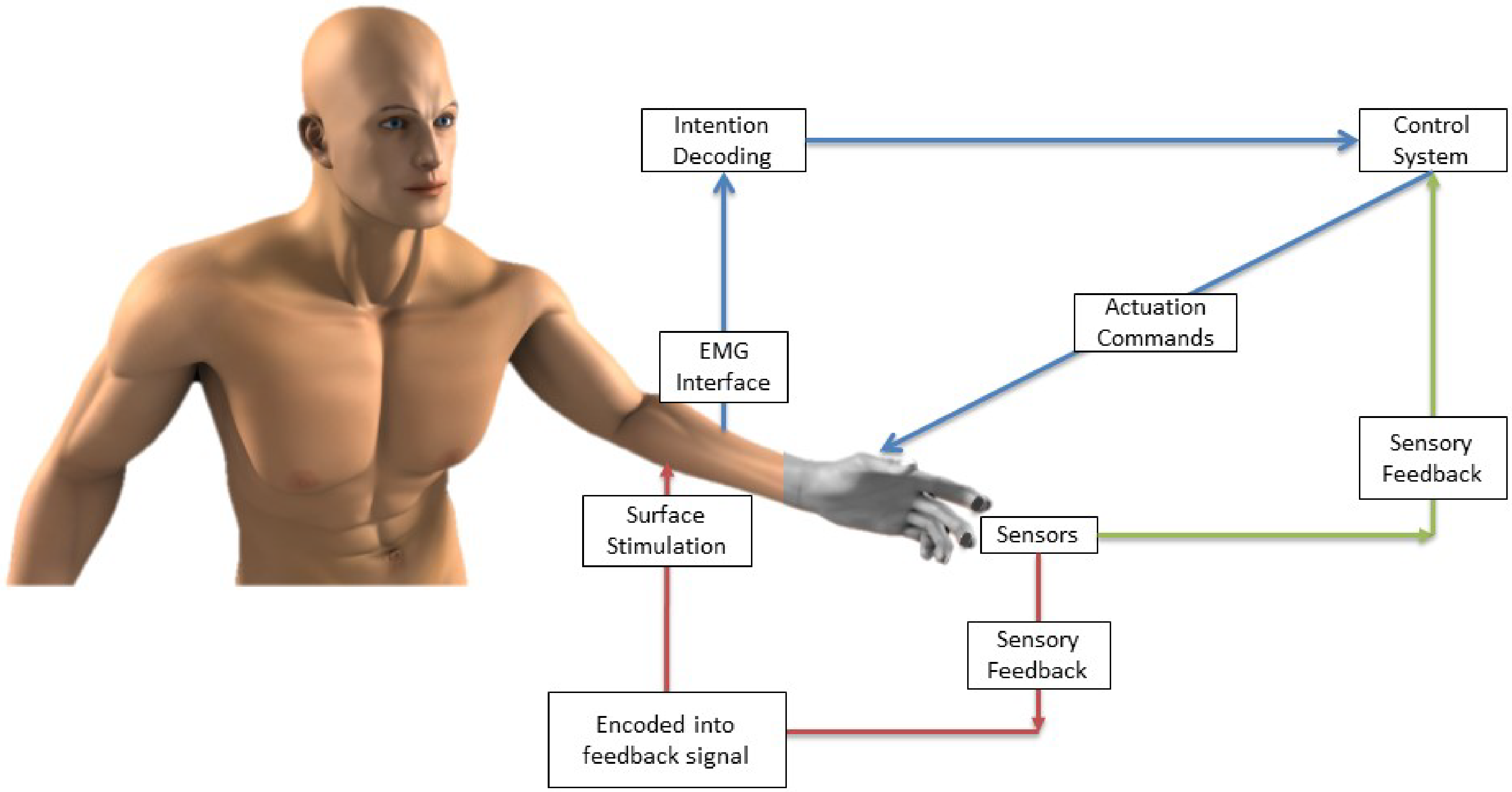


Figure 2.8 - Multiple Sensory Feedback loops (adapted from *[38]*)

Although there are a few longitudinal studies that examine the use of sensory feedback over a longer period [49, 91, 112], these mainly repeat the testing at discrete intervals over a few days or weeks. However, further analysis should be done on whether performance is maintained when consistently using the sensory feedback throughout the day over a few weeks, similarly to the work done by Clemente et al. [49]. Potentially, over time, the nervous system could become desensitised to the stimulation site, resulting in a higher cognitive load required to focus on the stimulations. If such a problem exists, stimulation sites may need to be moved up and down the arm to reduce the chance of desensitisation. Longitudinal studies are also required to examine the impact of the training and adaptation to using sensory feedback. Chai et al. [113] demonstrated that subjects were able to improve their recognition rate of electrotactile feedback on non-phantom digit sites over a three day period to a performance comparable to phantom digit sites. Stepp [91] et al. showed that incorporating vibrational feedback, subjects continued to increase in performance over an eight day period and they still saw a reduction in performance when the feedback was removed on day 8. However, recently, Strbac et al. [112] demonstrated that sensory feedback was greatly beneficial in the beginning of using the prosthetic device and learning to reliably manipulate the grasping force though their EMG control. However, overtime the user tended to rely more on feedforward control and their understanding of the relationship between EMG commands and resulting grasping force. Further investigation is therefore required to determine the role of sensory feedback long term and on its role in learning EMG control.

In addition, studies currently examine how sensory feedback assists a user in picking up objects, but no testing on holding these objects for longer periods has been conducted to date. For example, how does the feedback mechanisms work in assisting the user to hold a cup of coffee over the time it takes to drink it? The constant feedback over time, may be helpful, or it may be distracting for the user and the feedback may need to be also incorporated into the control mechanisms to successfully hold objects. Further, perception of stimulation may be altered when a muscle is activated compared to at rest.

The speed in communicating sensations has not been widely reported on when examining the performance of a sensory feedback system. A healthy peripheral nervous system can take approximately 14-28ms to deliver tactile information [1]. As a result, it was suggested by Antfolk et al. [114] that any surface stimulation for sensory feedback should be communicated in small percentage of that amount (3-5ms) in order to have a minimal impact on the overall travel time. Additionally, the timing delay between visual and tactile information can impact the sense of body ownership in the prosthetic device. A rubber hand illusion test performed by Shimadi et al. [97] and an Functional Magnetic Resonance Imaging (FMRI) study on body ownership by Bekrater-Bodmann et al. [115] showed that 0-300ms delay occurred no loss in body ownership. This FMRI study also showed significant disconnect between visual information and tactile information when there was a separation of more than 600ms. However, a further refinement study by Ismail and Shimadi [96] suggest that the feedback delay should be less than 200ms to maximise sense of body ownership. Therefore, timing becomes very crucial when considering the method of feedback. This gives an advantage to using electrical stimulation and may limit the effectiveness of mechanotactile systems. This effect of timing may also explain some of the conflicting results of techniques such as vibrotactile feedback. Although it can be as low as 10ms to detect vibration [5], it can be up to 400ms to reach the desired vibration level and frequency [89]. However, although only mentioned in a vibrotactile study by Hasson and Manczurowky [53], haptic drivers can be implemented to decrease start up times of vibration motors.

Although invasive methods show promise for providing a richer sensory feedback experience in the long term, non-invasive methods provide an opportunity to benefit users whilst more invasive methods are still being developed. In addition, not all users will be willing to undergo further surgery [33] and may instead opt for the non-invasive feedback option. Particularly within laboratory conditions, various approaches to providing sensory feedback through non-invasive methods show promise. A focus, therefore, for the immediate future should therefore be placed on implementing a simple feedback strategy that can be practically used at home every day so that prosthetic users can begin to take advantage of the benefits that sensory feedback could provide them.

This thesis, therefore, focuses on

- Communicating the grasping force of up to three channels of information, representing force on thumb, pointer and the remaining three fingers finger.
- Developing simple and easy to understand interfaces that can be incorporated into any myoelectrically controlled prosthesis.
- Developing analysing a mechanotactile approach for providing three stimulation channels
- Provide an analysis on the best electrode arrangement used for electrotactile stimulation and determine the recognition rate when being used for three stimulation channels
- Provide a comparison of the two commonly used stimulation sites for non-invasive feedback, the upper and lower arm regions
- Measure the ability existing myoelectric prosthesis using their existing prosthetic device to incorporate electrotactile and mechanotactile feedback into controlling their grasping force.

-

Chapter 3

Mechanotactile Stimulation for Sensory Feedback

3.1 Introduction

Mechanotactile information can be easier to discriminate than vibrotactile information [116]. Although mechanotactile feedback can affect EMG measurements in and around the same location, this interference can be easily filtered out using a high pass filter [117]. There have been multiple approaches undertaken to use mechanotactile devices to communicate sensory feedback for prosthetics [118]. However, these methods only provided one channel feedback to the user and were bulky. This chapter focusses on the development and characterisation of a mechanotactile feedback device for three channels of grasping force information, as this is currently the highest priority for prosthetic hand users [37].

In this chapter, the potential for recognising three channels of information through this feedback mechanism is demonstrated. Following this, the relationship between the applied stimulation and perceived intensity is determined, as well as the Just Noticeable Difference (JND) across the stimulus ranges, so that known levels of perceived intensity can be accurately induced on the subject's forearm. This perceived intensity will correspond to the level of grasping force applied on the objects handled by a prosthetic hand.

Since previously published literature has shown that a delay of greater than 300ms can decrease embodiment with sensory feedback [96, 97], the time taken to reach maximum displacement is measured as detailed in subsection 3.3.1. Subsection 3.3.2 compares the recognition rates of subjects with three different orientations of the mechanical cranks; transversally, longitudinally and diagonally to the arm as demonstrated in Figure 3.4; to determine which direction the shear stress/translational skin stretch is more easily perceived on the human forearm.

Section 3.3.3 determines the smallest perceivable difference in stimulation that test subjects can correctly identify. This is done using a Two-alternate force choice method to determine the JND for three reference stimulations of 10^0, 15^0 and 20^0. The relationship between the applied stimulation and the perceived intensity across the range of stimulations is then determined. This will follow two techniques suggested by Stevens [119], as detailed in subsection 3.3.4. Finally, the results are presented and discussed in Section 3.4.

3.2 Background

Wearable haptic devices have had some previous success in sensory feedback with winding belts being used to feedback information on grasping force [15, 56], and the hardness of the object [57], through changing pressure and skin stretch on the bicep. Similarly, a rocker design has been used to communicate proprioceptive information through skin stretch [105], however, it also only communicates one degree of actuation.

In this chapter, an improved method of mechanotactile feedback to that used by Antfolk is proposed, by using three servo controlled mechanical cranks which combine vertical pressure with linear skin stretch when providing sensory feedback. The number of feedback channels were limited to 3; to represent the movement of the thumb, the pointer finger and the remaining three fingers. When testing only single site stimulation, Antfolk et al. [120] reported an average discrimination rate of 97% for three feedback channels using mechanotactile devices, compared to an average discrimination rate of 82% for five DOF. Prosthetic hands with three degrees of freedom are one common approach taken [58, 59, 77]. The grasping taxonomy used by Vergara et al. [121] to record the usage frequency of different grasps also does not require independent movement of the ring and little fingers.

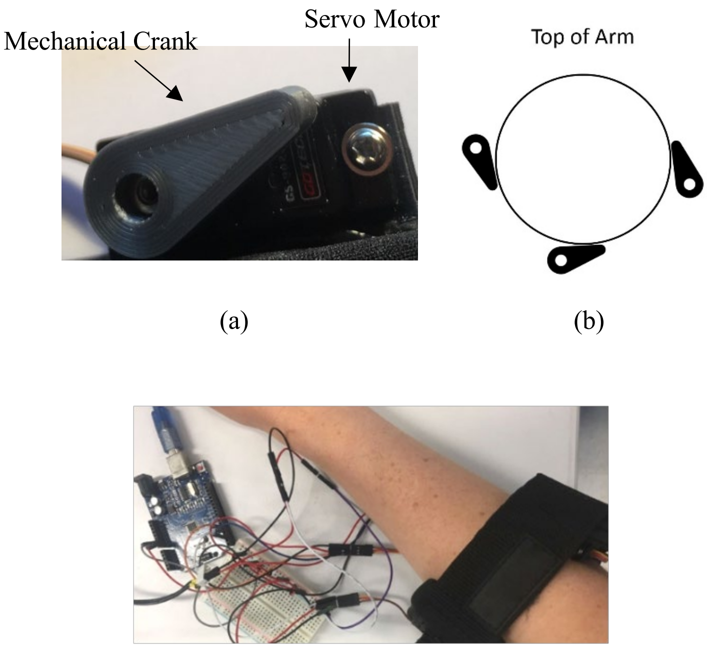
Previous literature has primarily focused on recognition rates of various mechanotactile stimulation methods and their improvement on grasping. However, no methodology has yet been developed to accurately and consistently induce a known level of sensation on the user from mechanotactile stimulation. Previous literature on mechanotactile feedback has tested the recognition of a discrete levels of force (ranging from two – five levels)

[118]. However, there has been no investigation to determine the amount of distinct intensity levels that can be consistently recognised by test subjects. Since Weber's Law [122] predicts that as the intensity of stimulation is increased, the smallest perceivable change will also increase accordingly, these perceivable changes need to be examined across multiple reference values and stimulation ranges.

The work by Antfolk et al [117] determined a simplified relationship between the servo rotation angle and the force applied. However, there is no literature on examining the relationship between the applied stimulation and perceived intensity for mechanotactile stimulation. Steven's Power Law [123] predicts that as the applied stimulation is increased, the corresponding increase in sensation evoked by the stimulus will follow a power law. Therefore, to provide an accurate representation of the level of grasping force through haptic stimulation on the forearm, a model needs to be established between the applied stimulation and the perceived intensity of the subject, which is described in subsections 3.3.4. and 3.4.4.

3.3 Method

The proposed mechanical crank feedback system shown in Figure 3.1 consists of three servo-motors, controlled via a microcontroller with a LabVIEW Interface. For the optimum crank orientation and timing experiments, Gotek micro servo motors were used. However, for all remaining experiments Saxon SH-1350 servo motors were used, as the Gotek micro servo motors became noisy over time and which may have impact experimental test results in perceived stimulation. The Saxon motors have the same rated speed (0.11 secs/60°) as the Gotek motors. The mechanical cranks were custom 3D printed to match the length of the motor, with a depth of 5mm. A surfboard leash cuff (Smart Leash Co.) was used to hold them firmly against the user's skin. To minimise the impact of variation of forces as a result from self-grounding, the leash cuff was always ensured to be applied firmly, whilst still being comfortable. The servos were mounted to a 3D-printed frame, which was then attached to the cuff.

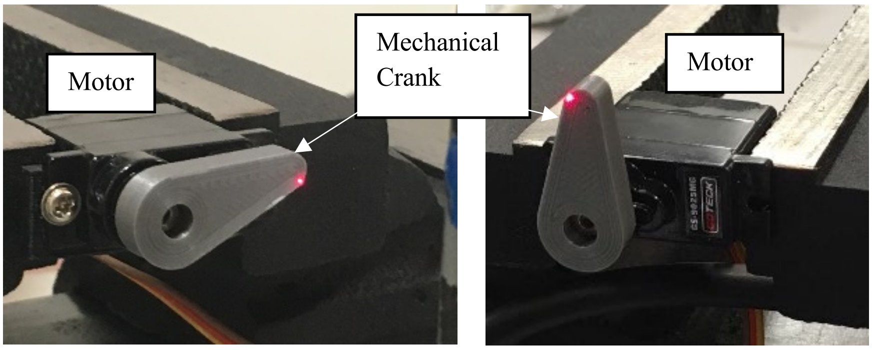


(c)

Figure 3.1 - Mechanical Crank Feedback: (a) Mechanical Crank, (b) Crank location on cross-section of arm and, (c) Placement on arm.

3.3.1 Time of Movement

To measure the time taken to begin activation of the feedback mechanism, as well as the time to complete the movement, a mechanical crank attached to a servo motor was fixed into place and its movements detected through use of two laser triangulation sensors (Micro-Epsilon optoNCDT1700). The laser one detected the initial movement time when the trailing edge began moving, as shown in Figure 3.2; and the finished movement was measured from the detection of the leading edge reaching the maximum displacement detected by laser 2, shown in Figure 3.2b. A LabVIEW interface was used to control the servomotor, via a microcontroller, and operate the millisecond precision timer. A flowchart of its process is shown in Figure 3.3, which was repeated ten times.



(a) (b)

Figure 3.2 - Mechanical Crank Timing Experiment Setup; (a) Measuring starting movement from trailing edge, (b) Measuring finished movement by detecting leading edge
.

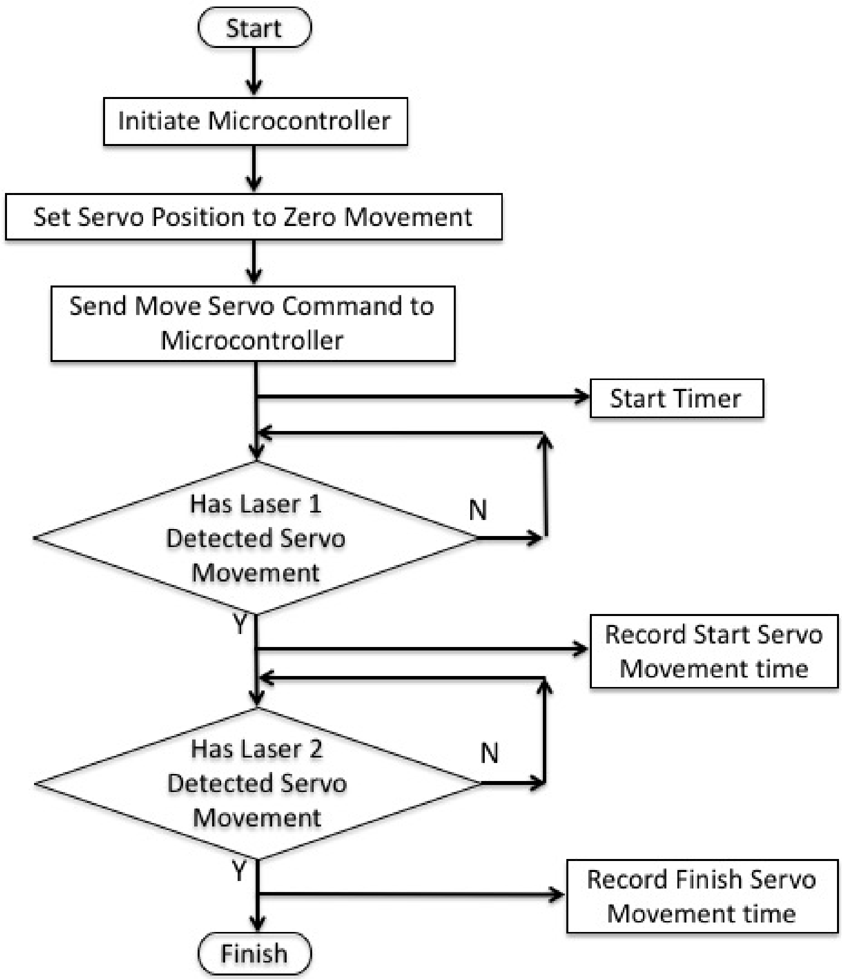
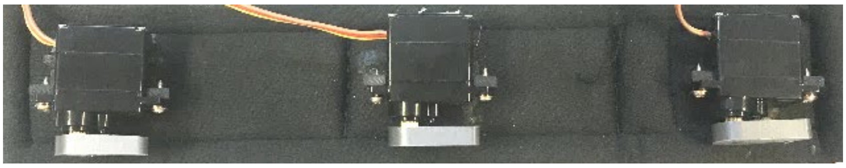


Figure 3.3 - Laser Timing Flowchart.

3.3.2 Optimum Crank Orientation

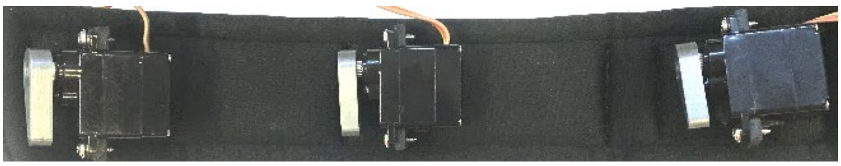
The range of movement of the crank for each user was determined through a calibration routine, where the system slowly increased the range of movement, resetting back to the zero position each time, to determine the largest crank movement comfortable. The user indicated when it was no longer comfortable, and the last comfortable movement was set as the maximum displacement for the user. Pilot testing of this experiment demonstrated that individual users had different comfort tolerance with the mechanical cranks, and differences existed between the comfort levels across the three stimulation sites and different orientations. Therefore, to increase the comfort level for the test subjects and to help increase perception recognition, all three mechanical crank stimulation sites were calibrated separately for each individual user and for each orientation tested.

3 orientations of crank movement to the forearm were compared: longitudinally, transversally and diagonally at an angle of 45 degrees, as shown in Figure 3.4. Performance was measured by the accuracy in recognition of grip patterns and intensity of pressure based on the amount of crank rotation.



(a)

(b)



(c)

Figure 3.4 - Mechanical Crank Orientations; (a) Transversal, (b) Diagonal and, (c) Longitudinal.

Recognition of six different grip patterns, shown in Figure 3.5, was tested: thumb only, pointer only, pistol grip (closing remaining three fingers only), fine grip (closing thumb and pointer), tool grip (closing thumb and remaining three fingers) and power grip (closing all fingers). The three motors correspond to the movement of the thumb, pointer and remaining three fingers, respectively. These are commonly used grip patterns to test sensory feedback [59, 124]. Each of these grips were tested in the fully closed position, represented by maximum comfortable crank displacement of the servo; or half-closed position, represented by 50% of the maximum comfortable angular displacement.

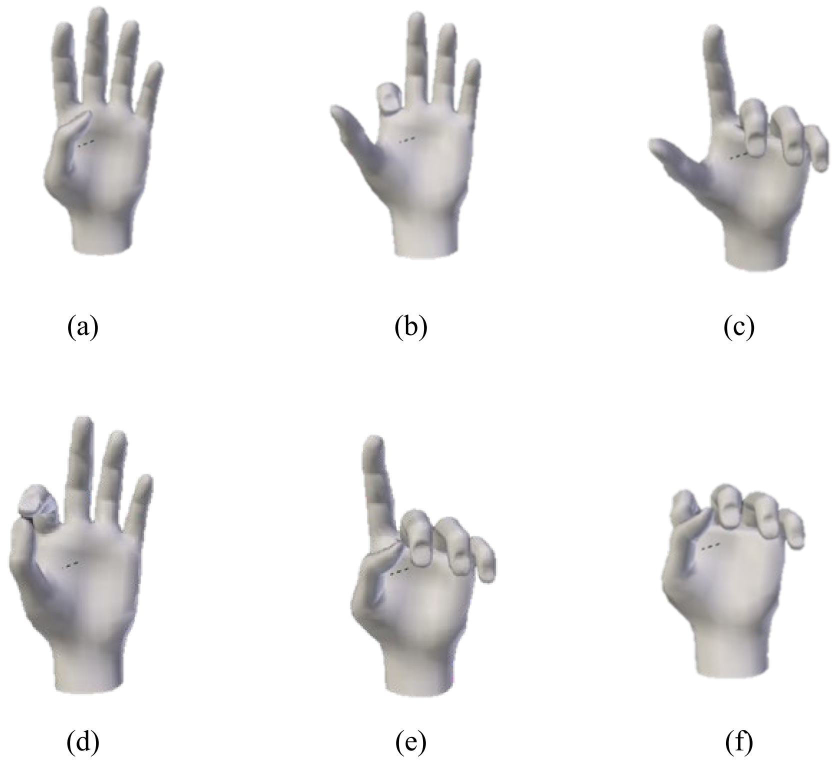


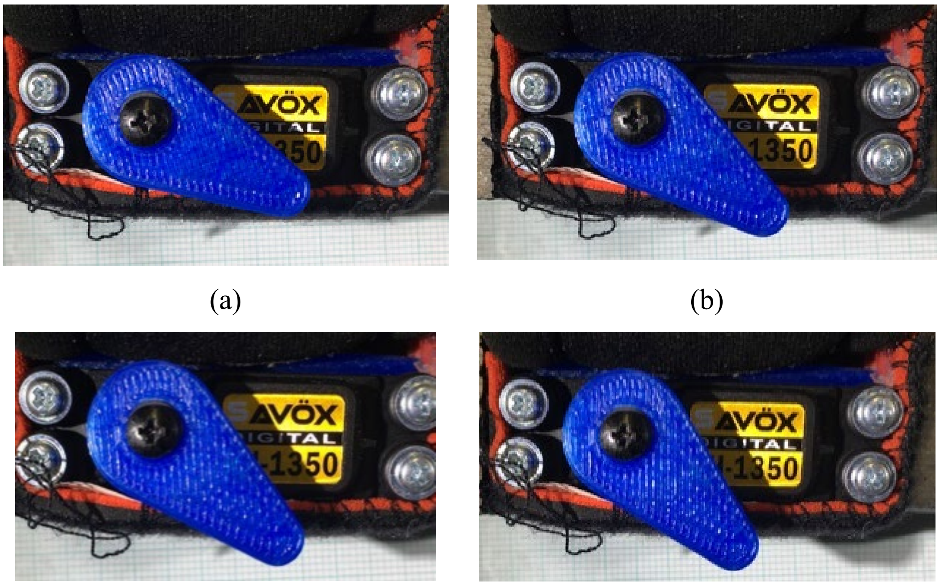
Figure 3.5 - Hand Grips: (a) Thumb Only, (b) Pointer only, (c) Pistol Grip, (d) Fine Grip, (e) Tool Grip and, (f) Power Grip.

In the training phase, each of the six finger movements was demonstrated to the user at the maximum displacement. The movement was communicated to the user prior to commencing sensory feedback, both verbally and visually with a picture of the corresponding grip. The crank stayed in the maximum displacement for a period of 800ms before returning to zero displacement, where there was a pause of five seconds before the next movement took place. After six movements, a 20-second-long break occurred before repeating all the grips at 50% displacement. A 2-minute break then occurred prior to the commencement of the testing phase. This short training period was used to demonstrate that due to intuitive nature of understanding the communicated feedback, extensive training is not required to achieve successful results.

In the testing phase, a randomised order of the six movements with three repetitions was developed, resulting in a total of 18 movements. Half of these movements were randomly assigned as the maximum displacement and the other half were assigned 50% displacement. Each test subject had their own randomised movement and strength combinations, presented to them in their own randomised order. The grips were held at the displacement for 800ms before returning to zero displacement. There was at least a 5-second pause between each movement for the subject to communicate the perceived movement. The subject could verbally tell the grip perceived or could choose the grip picture in a chart corresponding to those shown in Figure 3.5 This process was repeated for the two other crank orientations, with a 5-minute break in between each orientation test. A total of 18 subjects was tested, consisting of 16 males and two females, with a mean age of 32.7 years ± 7.1 (S.D) and no physical or cognitive impairment. The order of the orientation tested was changed for each subject to prevent the effect of additional training influencing the results. In total, the six different combinations of the testing orders were repeated three times across the 18 subjects.

3.3.3 Just Noticeable Difference

To determine the smallest perceivable change in stimulation, a two-alternate force-choice method was employed to determine the JND. This technique sends pairs of stimulation to the test subject; (R) & (R ± δx); where R is a reference value, and ± δx is a small increase/decrease in the stimulation value; and the subject is required to pick which stimulation is larger. In previous experiments [125], a 25 degree rotation was found to be comfortable for all 18 subjects, therefore this value was chosen as the upper limit of this experiment. In order to determine the JND at different points in the range of motion available, testing was conducted for three reference points of 10, 15 and 20 degrees. For each of these reference points, recognition of a difference of 0.5, 1, 1.5, 2, 3, 4 and 5 degrees was tested. 0.5 and 1.5-degree differences were added to improve the reliability of the psychometric curve after the first trial runs did not contain results close enough to the guessing rate asymptote.



(a) (b) (c) (d)

Figure 3.6 - Mechanical Crank Positions: a) 0° rotation, b) 10° rotation, c) 15° rotation, d) 20° rotation.

The mechanical crank was rotated to the reference point for one second, followed by one second of no pressure/crank rotation. The mechanical crank was then rotated to the second position which was a slightly different position before no rotation/pressure was applied. The test subject was then required to say which one felt stronger. Each of the pairs was used four times, twice with the larger rotation first (E.g. a ten degree rotation followed by a five degree rotation), and twice with the larger rotation second (E.g. a five degree rotation followed by a ten degree rotation). By repeating the tests in a reverse order, it minimised any impact of potential bias that would have occurred if the subjects regularly guessed either the first value or second value when they were unsure. This resulted in 56 test values being used for each reference level on each subject. A total of 168 pairs were presented to the subject in a randomised order, performed once on the underside (i.e. ventral region) of the arm and once on the outside (i.e. ulnar region) of the arm, as shown in Figure 3.7. These two sites were chosen as they represent the two main compositions found on the forearm, bony region (outside location) and a soft tissue region (underside location).

The psychometric functions were fitted with a logistic sigmoid using the psnigifit toolbox v4.0 for Matlab which implements the maximum-likelihood method as described in [126].

This curve was used to determine the JND threshold, taken as the midpoint between the lower and upper asymptotes.

(a) (b)

Figure 3.7 - Tested Stimulation Locations: (a) ventral region (under) of lower arm, (b) ulnar region (outer) of lower arm.

A short 30-second break occurred every 30 trials, and a 2-minute break occurred between the two locations. These breaks were taken to minimise any desensitisation from stimulation, and to reduce the effect of cognitive overloading from the concentration required. Subjects were able to take any additional rest breaks as desired.

Testing was undertaken by ten able-bodied subjects (two females, eight males) with a mean age of 27.1 years ± 3.7. (S.D).

3.3.4 Perceived Intensity

Stevens [119] previously proposed two methods of magnitude estimation to determine a relationship between applied stimulation and the subject's perceived intensity. In both methods, pairs of different stimulation strength levels are presented to the subject and the subjects then identify the ratio of the increase in perceived intensity. In the first method, the standard they refer to is only presented at the start, and then subjects continue to give feedback in comparison to their previous presented stimulation. In the second method, the stimulus is presented to the subject in every stimulation pair. In this chapter, both methods will be undertaken to determine the relationship between applied stimulation of our mechanotactile stimulation on two locations of the forearm, and then compare the results obtained from both methods. In both methods, the number of stimulations were chosen to keep the test session under ten minutes, as recommended by Stevens [119].

a) Method one – Standard presented once

In the experiment using the first method, subjects were given a reference stimulation (10 degrees) for one second and then were asked to assign a number to rate the feeling on the intensity induced, similarly to the method used by Graczyk et al. [12] in their intraneural stimulation study. Mechanotactile stimulation was then applied at this same level for one second, followed by one second of no stimulation, and then stimulation was applied to the subject at another random level of rotation for one second. The subject was then asked to assign a number to rate the feeling of intensity, using the previous stimulation value as a reference. For example, if the first stimulation was rated an intensity value of four, and the second stimulation felt twice as strong, they were instructed to assign it a value of eight. Subjects were encouraged to use decimals/fractions as required. These instructions were used to ensure they understood to use a ratio scale. This process was then repeated for the next stimulation value, where the rotation pair consisted of the last rated stimulation presented first followed by a new stimulation value.

This process was performed in four rounds per location; where each round contained 12 values between 2 and 24 degrees inclusive, separated by two degrees; and each value after the 12-degree reference value was presented in a random order. These rotation values were chosen so that they could be separated by the average JND resulting from all of the test subjects in the experimental part outlined in subsection 3.3.3. This process was repeated for 48 stimulations per location (underneath the forearm and outside of forearm) per subject. A 30-second break occurred after each round and a 2-minute break between the two locations. To ensure the stimulation values were between the minimum detectable threshold and maximum comfortable level for each subject, the lowest value and highest value were presented to the user before each round began. The intensity values given from the subjects were normalised by dividing them by the mean intensity value for that round. This allowed us to collate the individual results into a group data set and to determine the relationship between increasing rotation and increase in stimulation across the group of subjects rather than individually.

Testing was undertaken by ten able-bodied subjects (three females, seven males) with a mean age of 25.8 years ± 5.5 (S.D). Subjects wore noise-cancelling headphones with pink noise to prevent any impact from the motor's noise.

b) Method two– Standard presented every time

In the experiment using the second method, subjects had control of the graphical user interface as shown in Figure 3.8. When subjects pressed the “Intensity 10 Standard” button, they received the standard stimulation (12 degrees rotation) that they were told was assigned an intensity of ten for one second. When subjects pressed the “Stimulation to rate” button, they received another stimulation to compare to the standard for one second. The subject was then asked to assign a number to rate the feeling of intensity, using the standard stimulation as a reference. Again, if the first stimulation was rated an intensity value of four, and the second stimulation felt twice as strong, they were instructed to assign it a value of eight. Subjects were once again encouraged to use decimals/fractions as required. They were able to go back and forth and receive either of the two stimulations as required. Once they determined the intensity, they entered into the “Perceived intensity” text box and press next round. The round number would then increase, the perceived intensity returns to zero and the next stimulation value will be loaded. This process was then repeated for the next stimulation value, with the “intensity 10 standard” stimulation of 12 degrees staying the same throughout the whole experiment.

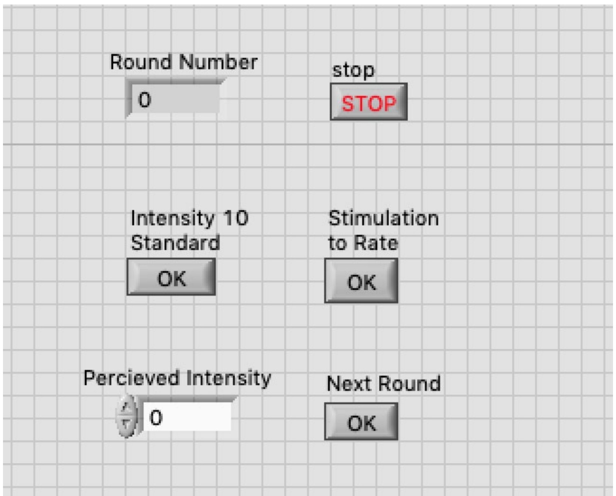


Figure 3.8 - Magnitude Estimation GUI

The stimulations of [4, 6, 8, 9, 16, 18, 20, 24] were tested against the standard to represent the ratios of $[\frac{1}{3}, \frac{1}{2}, \frac{2}{3}, \frac{3}{4}, 1\frac{1}{2}, 1\frac{1}{2}, 1\frac{2}{3}, and\ 2]$ respectively. Each test session consisted of each of these values tested five times in a random order, resulting in 40 stimulation pairs per test. Subjects were encouraged to take a 30-second break every ten rounds to reduce any possible impact of desensitisation and concentration fatigue.

Testing was undertaken by ten able-bodied subjects (four females, six males) with a mean age of 29.8 years ± 4.4 (S.D). Subjects wore noise-cancelling headphones with pink noise to prevent any impact from the motor's noise.

3.4 Results and Discussion

3.4.1 Time of Movement

An average time of 53.4ms ± 9.5ms (S.D.) was recorded for the servo to begin movement. This time consists of the time taken for the microcontroller to process and send the command (measured at 22ms), as well as start-up time of the motor to drive dynamics and stiction. An average time of 162.4ms ± 6.6ms was recorded for the full servo movement from when the command was sent, which is lower than 300ms proposed in the literature.

3.4.2 Recognition Rate

a) Grip Only

The average recognition rates for the different orientations are shown in Table 3.1 and Figure 3.9. A repeated measures ANOVA with a Greenhouse-Geisser correction determined that the difference between the mean recognition accuracy of the three different orientations was statistically significant ($F(1.552,26.387) = 4.970$, $p=0.021$). Post hoc tests using the Bonferroni correction revealed that longitudinal orientation (88.0% ± 6.9%) produced an increase in performance against transversal orientation (78.4% ± 10.4%) with a statistical significance of $p=0.006$; and an improved recognition rate compared to diagonal orientation (78.4% ± 15.7%) with a statistical significance of $p=0.035$. The difference in performance between transversal and diagonal orientation was not significant ($p=1.000$). A confusion matrix for grip recognition from all orientations combined and from the best performing orientation (longitudinal) are shown in Figure 3.10 and Figure 3.11, respectively.

Table 3.1 - Recognition Rate of Grip Only

Orientation	Average % Recognition ± SD
Longitudinal	88.0% ± 6.9%
Transversal	78.4% ± 10.4%
Diagonal	78.4% ± 15.7%

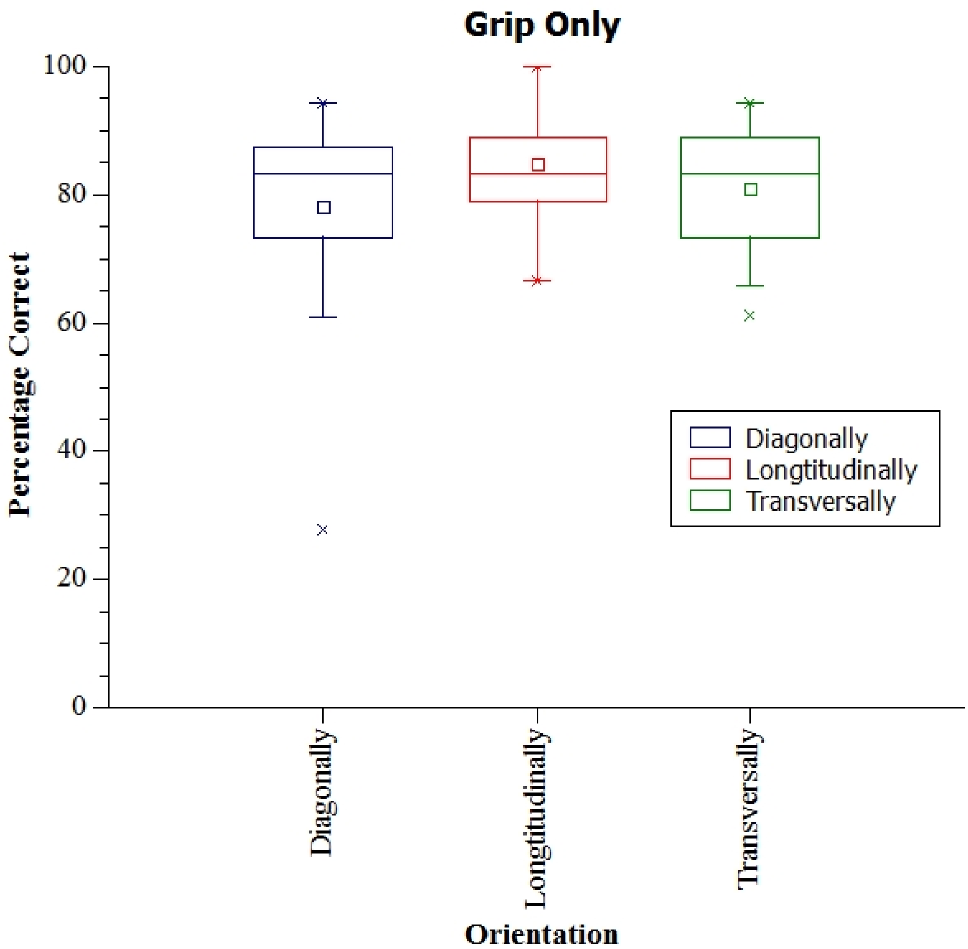


Figure 3.9 - Box Plot: Recognition Rate of Grip only

While normal and shear pressures are induced in each crank orientation, shear stress/tangential skin stretch appears to be interpreted easier when applied longitudinally to the human arm as it results in the highest recognition rate. This thesis postulates that this direction is more intuitive due to the natural biological mechanisms behind proprioception using the skin stretch around the nearby joints [127].

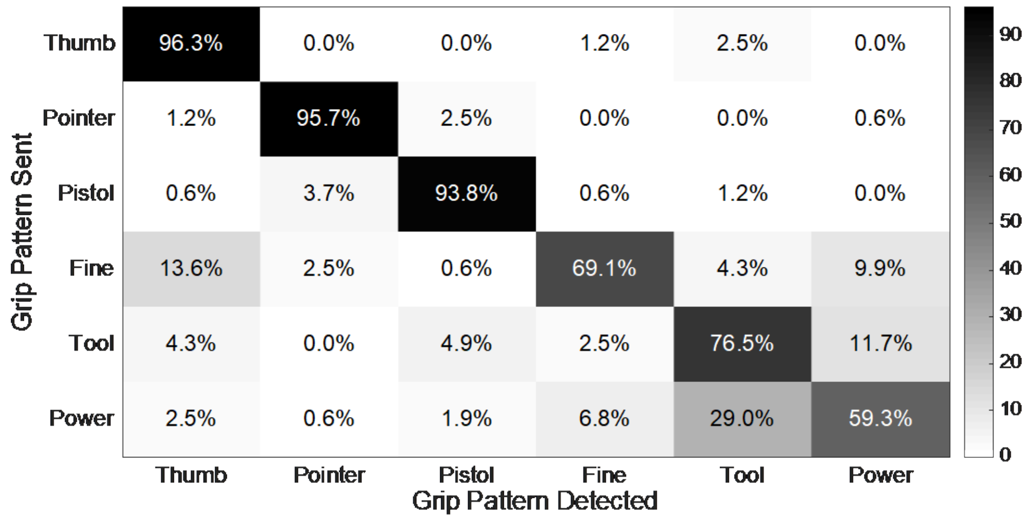


Figure 3.10 - Confusion Matrix for Grip from all orientations. Row labels identify the applied stimulation grip pattern; Column labels identify the percieved stimulation grip pattern; Values represent the percentage of times the applied stimulation grip pattern was percieved that way.

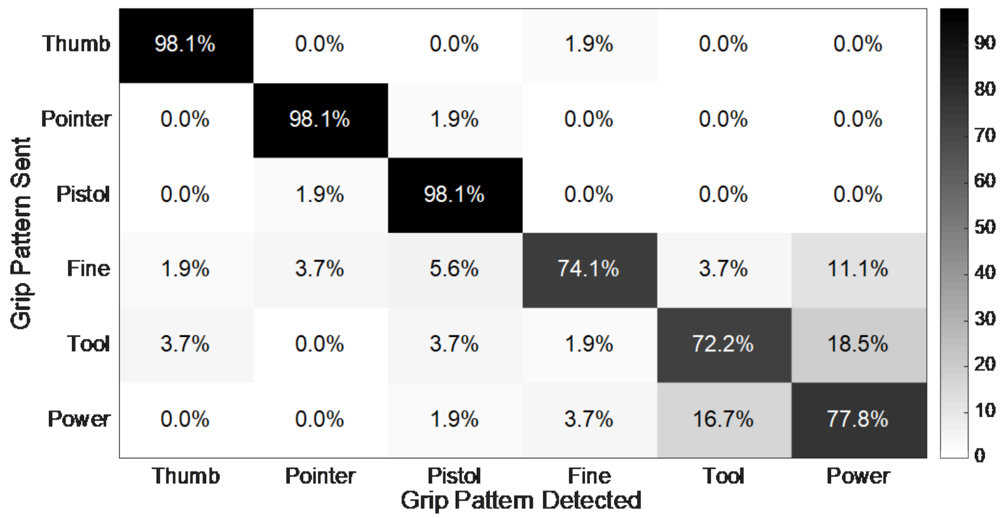


Figure 3.11 - Confusion matrix of Grip for Vertical Orientation. Row labels identify the applied stimulation grip pattern; Column labels identify the perceived stimulation grip pattern; The values represent the percentage of times the applied stimulation grip pattern was perceived that way.

c) Grip and Intensity Combined

In section 3.4.2 a) and b), the accuracy was based off the participants ability to only recognise the grip pattern (i.e. combination of motors used) or the intensity (amount of motor rotation), respectively. However, in this section, the response was only recorded correct if the participant responded with the correct intensity (high or low) and the correct grip pattern (thumb, pointer, pistol, fine, tool, and power). The average recognition rates for the different orientations are shown in Table 3.2 and Figure 3.12. A repeated measures ANOVA with a Greenhouse-Geisser correction determined that the difference between the mean recognition accuracy of the three different orientations were statistically significant ($F(1.580,26.865)=7.284$ $p=0.005$). Post hoc tests using the Bonferroni correction revealed that longitudinal orientation (80.9% ± 11.6%) produced an increase in performance against transversal orientation (68.2% ± 13.7%) with a statistical significance of $p=0.009$; and an improved recognition rate compared to diagonal orientation (69.8 ± 16.3%) with a statistical significance of $p=0.002$. The difference in performance between transversal and diagonal orientation was not significant ($p=1.000$).

Table 3.2 - Recognition Rate of Grip and Intensity Combined.

Orientation	Average % Recognition ± SD
Longitudinal	80.9% ± 11.6%
Transversal	68.2% ± 13.7%
Diagonal	69.8 ± 16.3%

Considering the small training time, with only one demonstration of each grip at both force levels, subjects achieved a high recognition rate of both grip and force levels. The training also only incorporated visual pictures and verbal labels of grips. Although there were promising results with minimal training, increased learning time with a visualisation of a prosthetic hand moving, either real or virtual reality, could still help further increase the accuracy. Some testing subjects used their previous prediction to help determine what grip and/or intensity the next stimulation was, without knowing whether their previous prediction was correct, which sometimes resulted in multiple incorrect recognitions. In real world situations, however, subjects would incorporate visual feedback as a truth basis for continual learning to help improve their recognition rates.

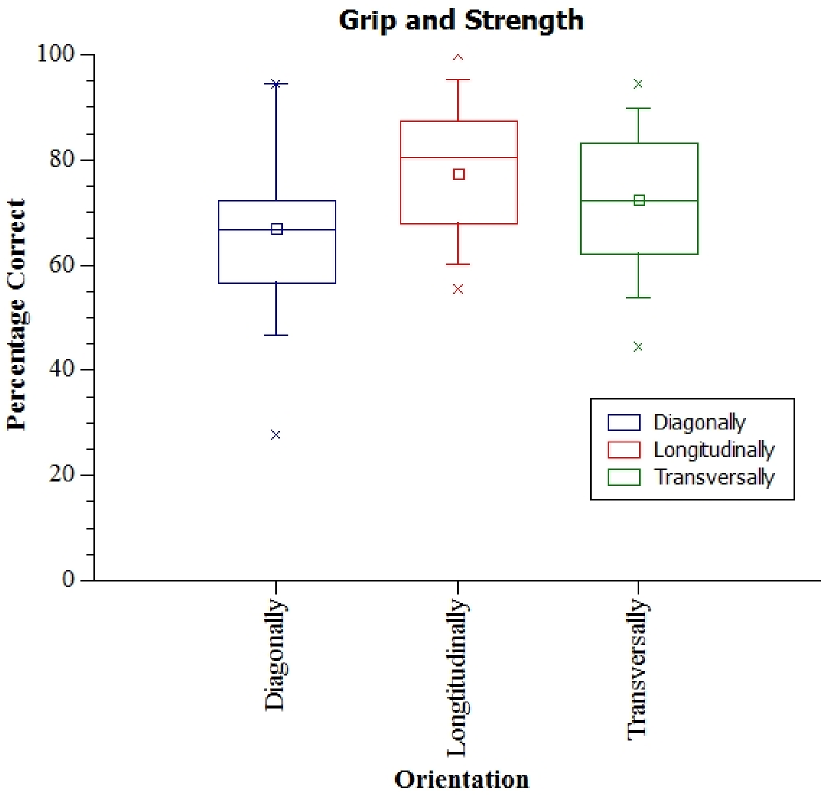


Figure 3.12 - Box Plot: Recognition Rate of Grip and Intensity Combined.

An analysis was performed to determine if there was any significant impact on the order of testing, independently of the orientation they used. A repeated measures ANOVA with a Greenhouse-Geisser correction determined that the mean recognition performance that contained no statistically significant difference for the order of testing for grip only ($F(1.605,27.279)=1.728$, $p=0.200$). However, there was a small statistically significant difference between order of testing when examining grip and intensity combined ($F(1.879,31.935)=3.927$, $p=0.32$). Post hoc tests using the Bonferroni correction revealed that the second trial (77.5% ± 12.8%) produced an increase in performance against the first trial (67.0% ± 16.1%) with a statistical significance of only 0.042, but no statistical difference compared to the third trial (72.5% ± 13.0%) with $p=0.577$. The first and third trial showed also showed no significant difference ($p=0.447$).

These results are an improvement upon the results reported by Antfolk et al. [16], who achieved an average accuracy of 68% for their able bodied subjects. In their study, five out of ten of their subjects were amputees, however, they noted that there was no statistical difference between able bodied subjects and amputees for the grip recognition and distinguished level of touch experiments. Our experimental evaluation tested recognition of a larger number of grip patterns, examining six grip patterns at two different force levels, totalling 12 different possible options; compared to Antfolk et al.'s testing of three different grips, with only one grip containing three different force levels, totalling five different grip options. Therefore, since our lowest result was comparable to the previously obtained results, whilst incorporating twice as many grip options, this result demonstrates the benefit of using the skin stretch action when applying pressure through the use of the mechanical crank. Further, our results indicate that this skin stretch is most effective when applied longitudinally to the human arm.

As shown in the confusion matrices (Figure 3.10 and Figure 3.11), errors were made when multiple motors are activated at once (Fine, Tool and Power Grip). Currently, the motors and cranks rest on the skin when no movement occurs. This may make it difficult to distinguish between when a crank is moving against your skin and when the motor/crank is pulled against you from movement of another crank. Adding a layer of padding underneath the motors, with gaps for the crank to go through, could improve the comfort level and help reduce false detections. Verbal feedback from the subjects was that the crank on the middle motor, corresponding to the pointer finger, was the hardest to detect when multiple motors were activated. Although individually calibrating each crank aimed to reduce any difference in perception between the motors, it could be further improved by operating the cranks using a constant force feedback method where an intensity is communicated by the crank supplying a corresponding force, rather than the currently utilised method of intensity corresponding to crank displacement.

Within this experiment, each person used the same armband with the same spacing, however, there were large variances in the size of the subject's arms. Further improvements could be made in comfort and recognition rate by using different armbands specific to the size of the subject's arm. In addition, further improvements could be achieved by each servo motor being attached to their own separate armband, so that when one motor activates it does not unintendedly pull another motor into the skin by stretching the armband.

3.4.3 Just Noticeable Difference

Each subject had their own psychometric function fitted to their data and a mean and median of each of the individual subject's JND was calculated. In addition, due to the low number of stimulation per subject, as performed in [128] the data for the ten subjects will be combined together for a group psychometric curve for both the underside and outside location, as shown in Figure 3.14. A summary of the JND results attained for each of the ten subjects is shown in Figure 3.13 and Table 3.3.

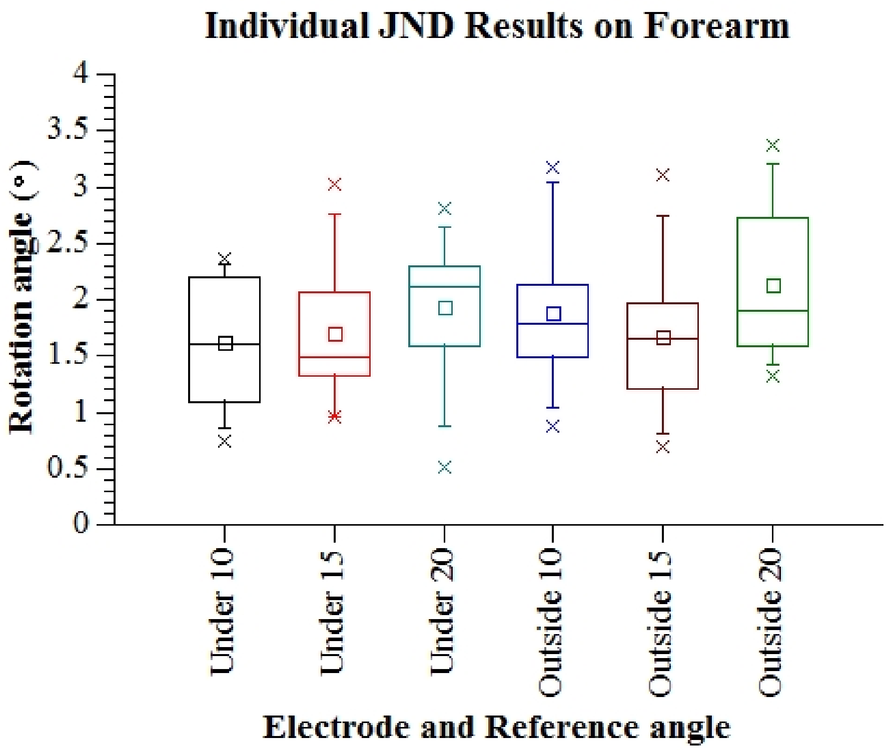


Figure 3.13 - Summary of JND results attained from the ten subjects

To determine if either the reference angle or location on the forearm made an impact on the JND, a nested and repeated measures ANOVA was applied. Both the location and reference angles met the assumption of sphericity using the Grenhouse-Giesser estimate of sphericity (ε=0.1.000 and ε=0.835, p=0.477, respectively). The analysis, however, showed there was no significant differences in the JND obtained at any of the three reference levels [F(2,18)=2.630, p=0.1] or for the twolocations tested on the forearm [F(1,9)=0.773, p=0.402].

Table 3.3 - Determined JND Values. Combined Group JND and confidence Interval was calculated based of fitting a psychometric curve to the all the data combined from the ten subjects.

Location	**Reference Angle (°)**	**Mean Individual JND (°)**	**SD Individual JND (°)**	**Median Individual JND (°)**	**Combined Group JND (°)**	**Combined Group JND Confidence Interval (°)**
Forearm Under	10	1.62	0.60	1.61	1.54	1.12-1.97
	15	1.71	0.66	1.49	1.53	0.97-1.99
	20	1.93	0.66	2.12	1.68	1.07-2.23
Forearm Outer	10	1.89	0.70	1.78	1.69	1.06-2.30
	15	1.67	0.70	1.65	1.59	1.02-2.11
	20	2.12	0.71	1.90	2.01	1.46-2.55

As shown in Figure 3.14a and Figure 3.14b, the psychometric curves are near identical for the three reference values. For the outside forearm location, although there is a difference between the threshold values, these differences are not statistically different, as shown in the large overlap of the 95% confidence intervals for the three reference values shown in Figure 3.14d. The observed differences do not follow the trend of increasing JND from increased stimulation reference values. Instead, the JND is not statistically different across the full range of motion. It is postulated that this is because as rotation angle is increased, not only is there an increase in pressure applied normally to the skin surface, but also an increase in the transversal force to the skin surface as a result of the skin stretch. However, these results should be repeated with a larger amount of stimulation values and more subjects to have a higher statistical confidence.

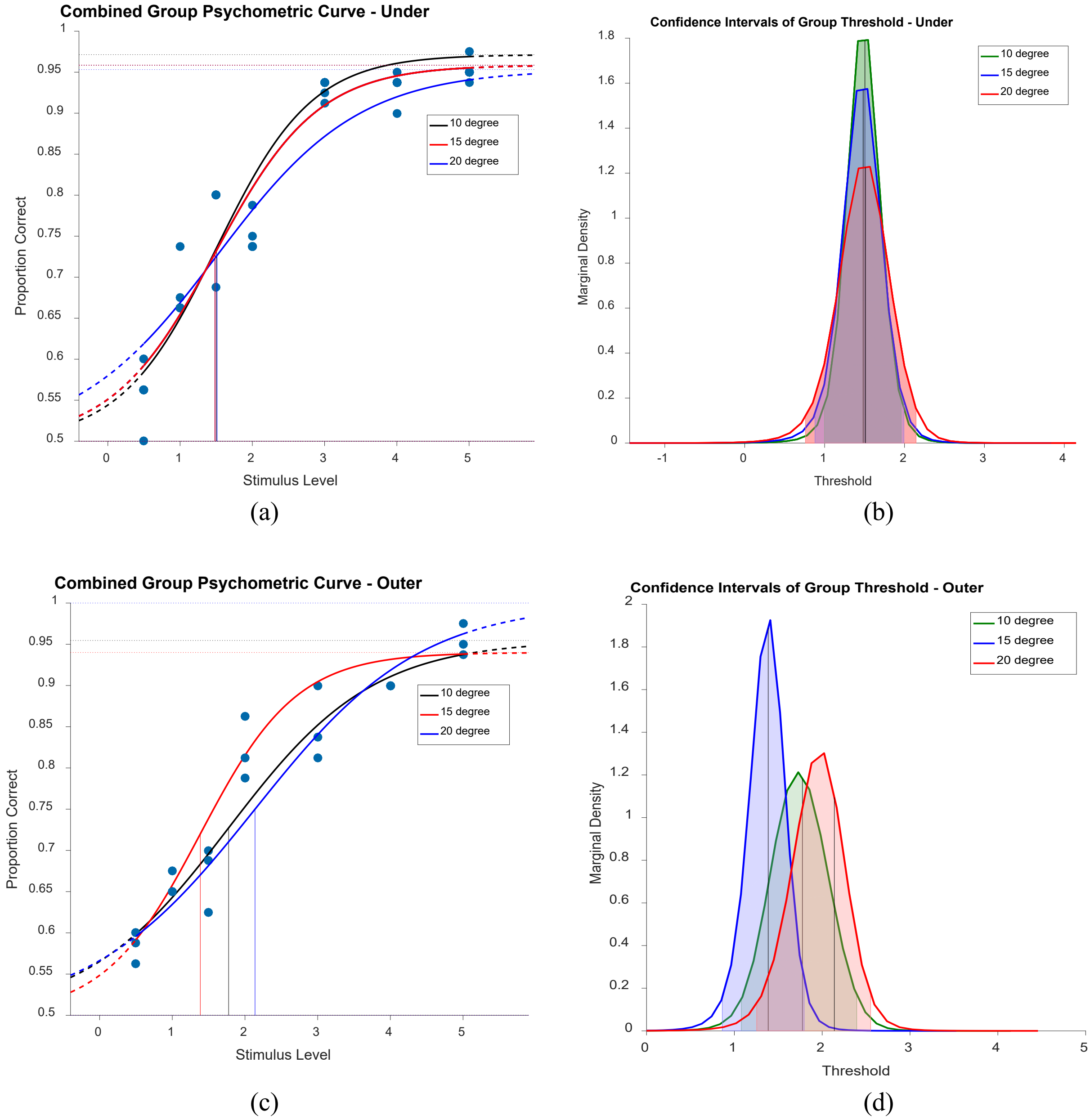
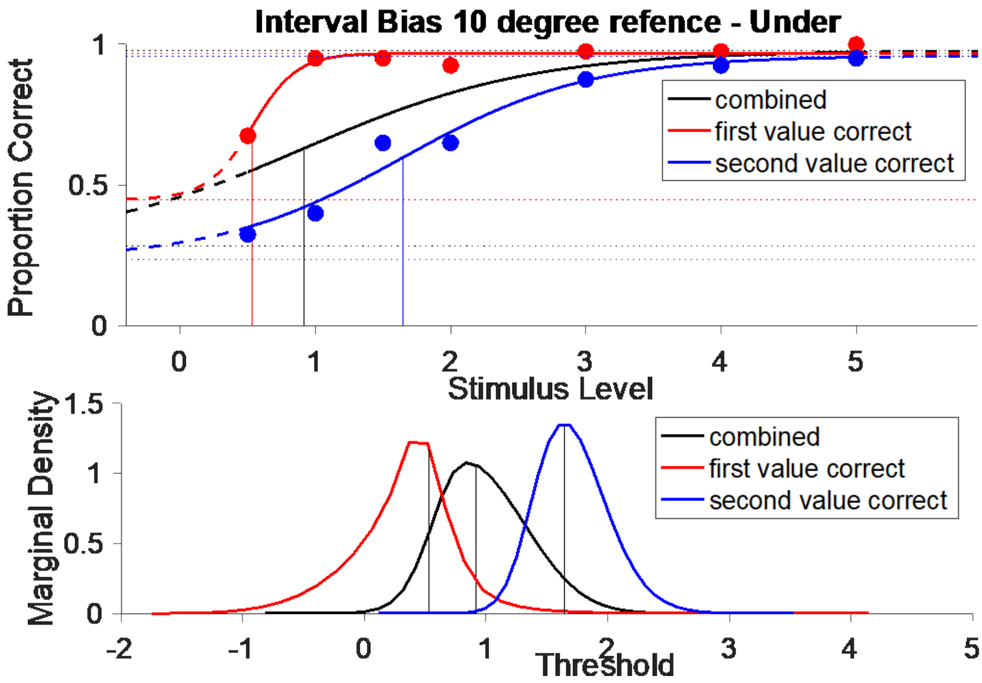
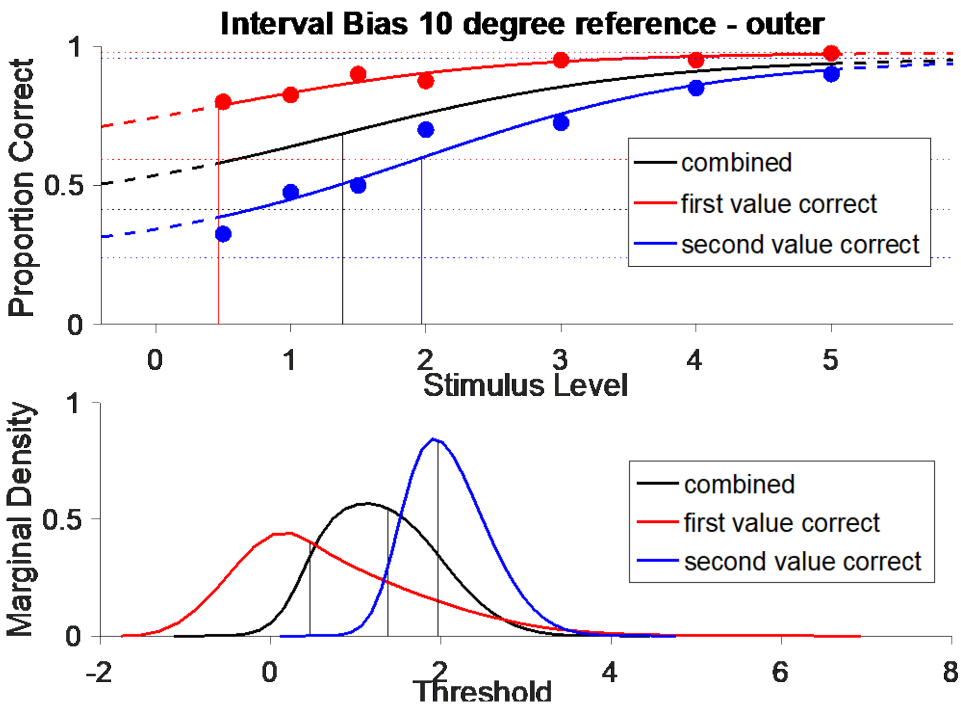


Figure 3.14 - Combined Group Psychometric Curve

To examine the difference in perception of the stimulation when applied as the first pair or second pair, the combined group data was split up into two groups, one where the first stimulation applied is the correct choice (higher value), and the one group where the second stimulation applied is the correct choice.



(a)



(b)

Figure 3.15 - Interval Bias for the different reference stimuli at the twodifferent locations

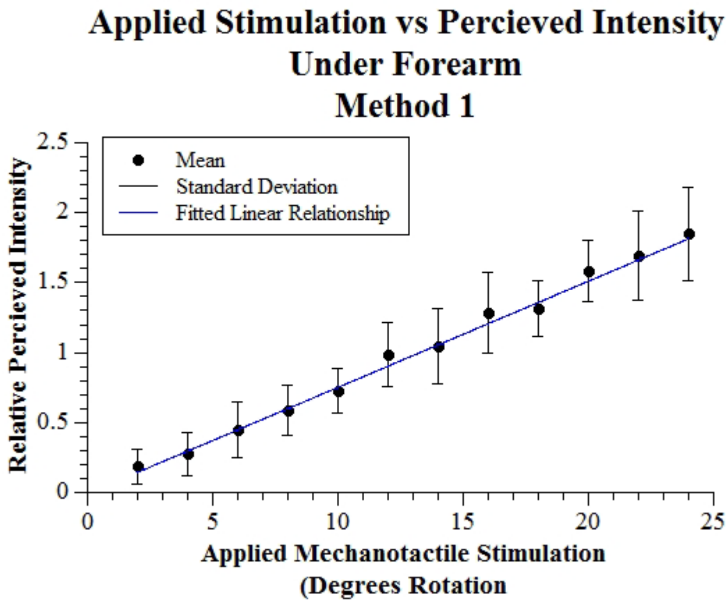
At the 10-degree reference level, shown in Figure 3.15a and Figure 3.15b, there is a large discrepancy between the JND thresholds when going from a higher value to a lower value, compared to when the stimulations are increasing in intensity. This suggests a significant bias in these tests when subjects are presented with the larger stimulation first. However, for the 15-degree and 20-degree references on both the underneath and outside forearm locations, the results from two different groups have moved back closer to each other. Since it is not consistent across all the reference levels, it suggests that this cannot be due to ten subjects defaulting to guess the first stimulation when they are unsure, nor from the recency effect. Instead, the authors postulate that it may be as a result of desensitisation, even though there is a one second break in between the stimulations, the second stimulation does not feel as strong. However, as the reference level is increased, this desensitisation effect is reduced, perhaps as a result of increasing transversal force\skin stretch occurring at higher levels of rotation. Since every pair combination was tested twice with the larger rotation first, and twice with the largest rotation second, this bias should have no impact upon the overall results. However, when this feedback mechanism is used in the context of prosthetic feedback, this effect may change the perception of stimulation.

3.4.4 Perceived Intensity

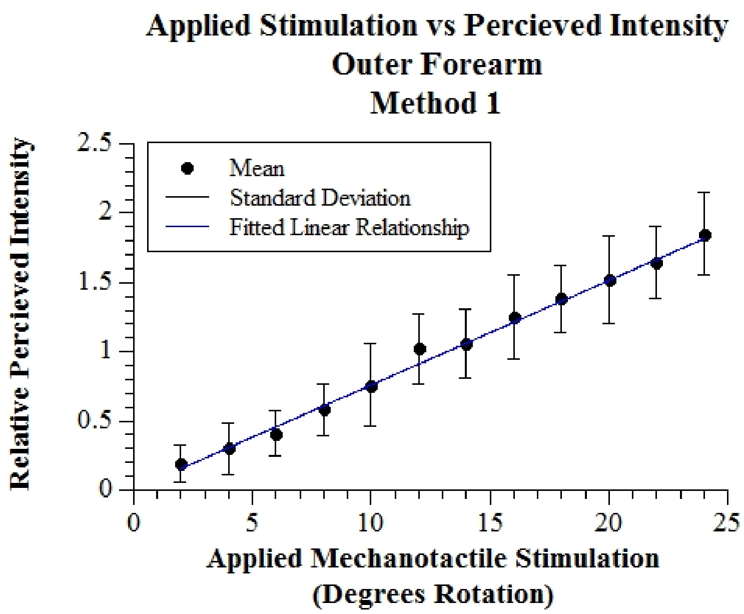
a) Method One – Standard Presented Once

The normalised intensity results for all individuals are pooled together and are displayed in Figure 3.18. The linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the underneath location with an R^2 value of 0.924. Individually each subject achieved an R^2 value of 0.960±0.026. A mixed model linear analysis was performed to determine the coefficients whilst taking into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of the five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 0.076 ± 0.0021 (S.E) [$t(109.000)= 36.721$, $p<0.001$].

A linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the underneath location with an R^2 value of 0.803. Individually the subjects achieved an R^2 value of 0.820 ± 0.087. Mixed model linear analysis was performed to determine the coefficients and take into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of the five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 0.075 ± 0.0020 (S.E) [t(103.855)= 37.115, p<0.001].



(a)



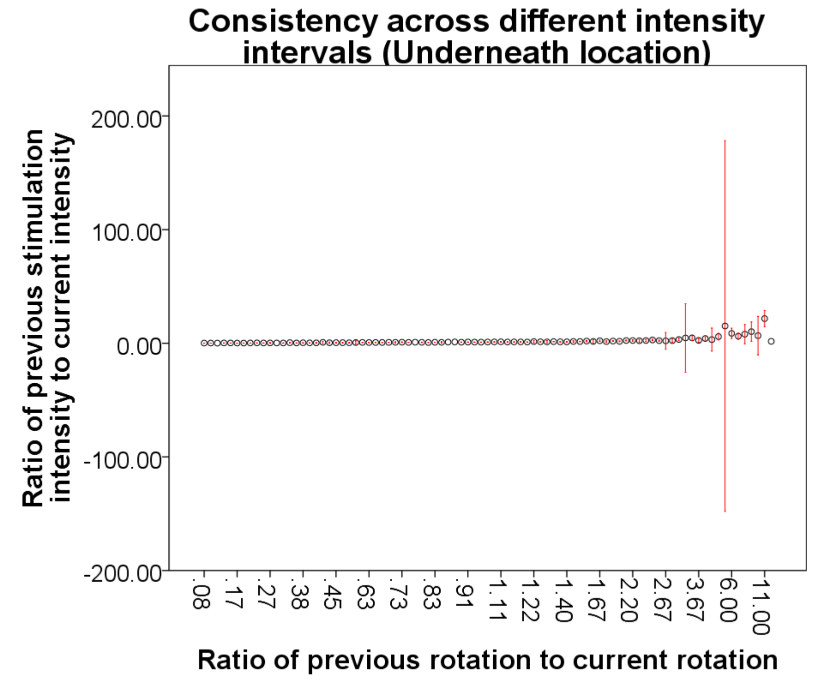
(b)

Figure 3.16 - The relationship between the crank rotation angle and it's induced perceived intensity on the subjects forearm using Method One: (a) underneath forearm location (b) outside forearm location

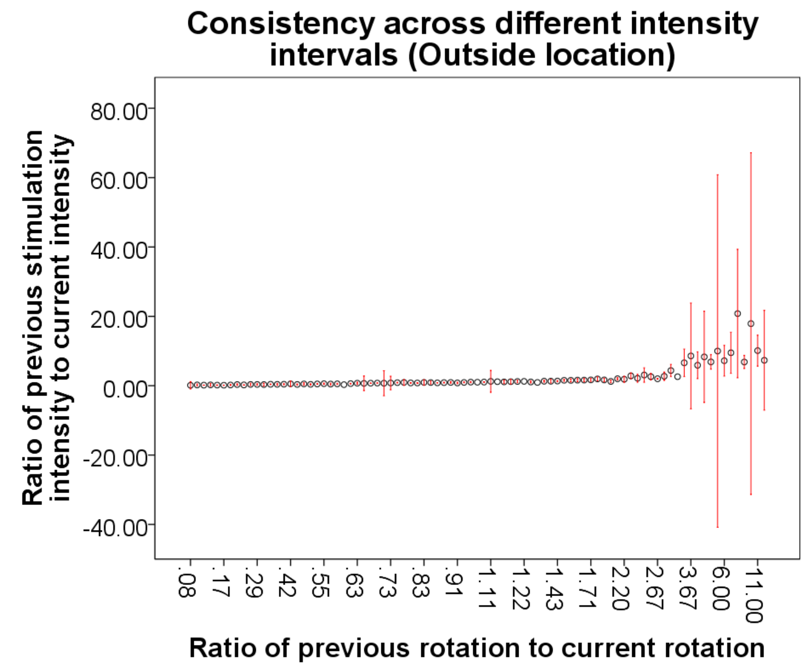
At both the underneath and outer locations, there was minimal reduction in R^2 values when the results were pooled together indicating this relationship is not subject specific and can be approximated across a whole population. Further a repeated t-test was conducted of the slopes generated from the individual regression analysis which showed no significant difference between the two locations [$t(9)=-0.384$, $p=0.710$].

As evidenced by the strong linear relationship at both locations, this stimulation pattern has suggested that this relationship follows Steven's Power Law with a power of one and a coefficient of 0.0755. It is postulated that the rotation angle does not correspond proportionally to the applied stimulation, but instead the increasing rotation produces both an increased normal force and increased tangential force to the skin.

Due to the nature of the test, it is inevitable to have inherent inaccuracies and variations. Although subjects were encouraged to use decimals, they tended to round off to the nearest whole or ½ number because it would be difficult for someone to confidently tell the difference between say 1.5 and 1.7. This, however, means that since the crank rotations were not in whole number ratio intervals, it created variation in the normalised intensity. As shown in Figure 3.17, the biggest variation is introduced when going from a small number to a higher number that is more than 3.5 times bigger than the first value. When reducing the amount of rotation by a large amount, subjects tended to be able to remember/recognise the smallest values. However, when comparing with a much larger rotation, they may have been distracted by the fact that it was significantly large. For the purpose of this experiment, large discrete jumps in grasping force were tested. However, in a real-world scenario, a hand will typically increase its force applied to an object as it closes, rather than undergoing a large discrete jump in the applied force. This will result in a more continual and steady increase in applied stimulation, albeit quick, which may alleviate issues with this. However, it is still important to identify as it would have impacted upon the correlation results obtained. It has, however, been suggested that the rate of force may impact upon the perceived intensity [129], which will need further investigation if various speeds of application are utilised in a feedback scenario.



(a)



(b)

Figure 3.17 - Consistency in proportionality from the different jumps of interval rations

d) Method Two– Standard presented every time

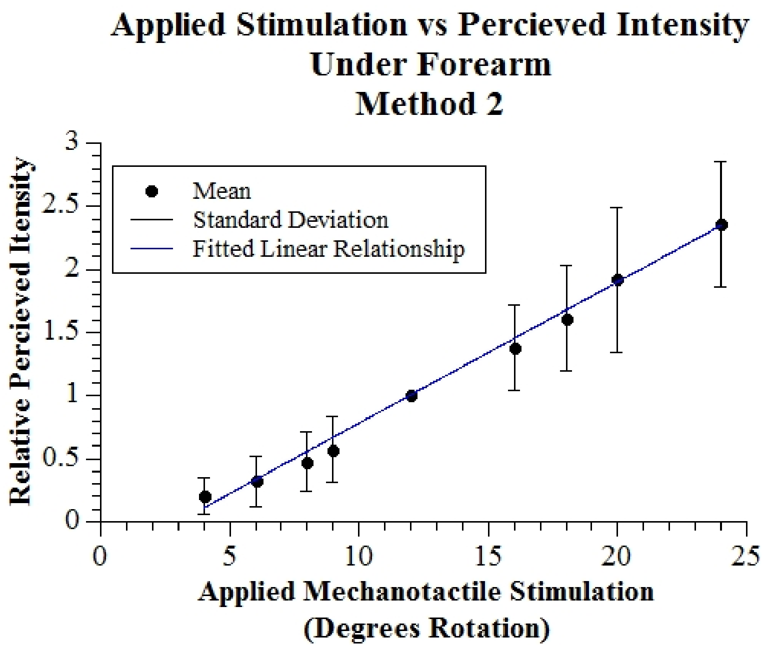
The normalised intensity results for all individuals are pooled together and are displayed in Figure 3.18. The intensity values were divided by ten for the results to be relative to the standard stimulation so that a comparison can be made to method 1.

The linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the underneath location with an R^2 value of 0.812. Individually each subject achieved an R^2 value of 0.902 ± 0.031. A mixed model linear analysis was performed to determine the coefficients and take into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of the five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 0.112 ± 0.0096 (S.E) [$t(8.776)= 11.595$, $p<0.001$].

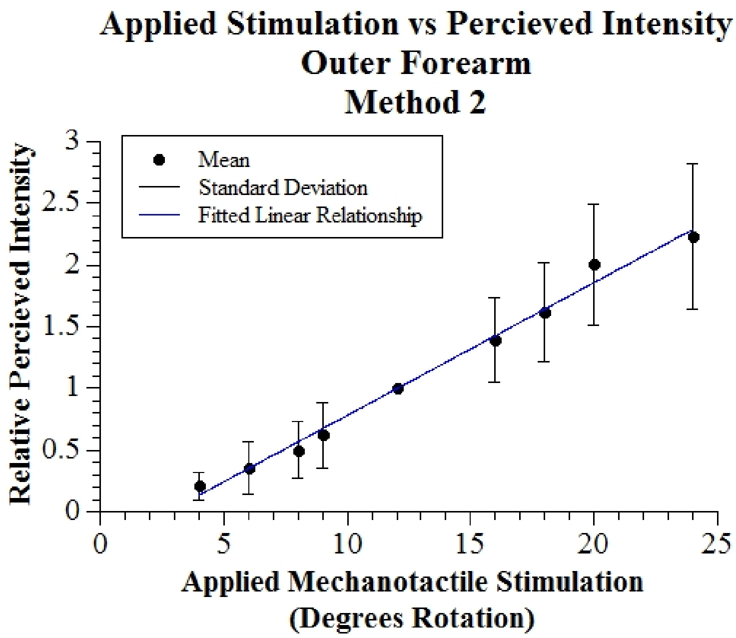
The linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the outside location with an R^2 value of 0.801. Individually each subject achieved an R^2 value of 0.914 ± 0.028. A mixed model linear analysis was performed to determine the coefficients and take into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of the five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 0.108 ± 0.010 (S.E) [$t(8.891)= 10.45$, $p<0.001$].

At both the underneath and outer locations, there was minimal reduction in R^2 values when the results were pooled together indicating this relationship is not subject specific and can be approximated across a whole population. Further a repeated t-test was conducted of the slopes generated from the individual regression analysis which showed no significant difference between the two locations [$t(9) = 0.513$, $p = 0.639$].

As evidenced by the strong linear relationship at both locations, this stimulation pattern has suggested that this relationship follows Steven's Power Law with a power of one and an average coefficient of 1.10. It is postulated that the rotation angle does not correspond proportionally to the applied stimulation, but instead the increasing rotation produces both an increased normal force and increased tangential force to the skin.



(a)



(b)

Figure 3.18 - The relationship between the crank rotation angle and it's induced perceived intensity on the subjects forearm using Method Two: (a) underneath forearm location (b) outside forearm location.

e) Comparing Method One and Method Two

Since our previous tests demonstrated no statistical differences between the outside and underneath locations, the results were pooled together to compare the results of the two methods together. Independent t-tests were performed on the R^2 measurements and the Relative Standard Error measurements. These showed that there were no statistically significant differences between the relative standard errors in the two techniques [t(38)=-4.11, p=0.683], however, that was a statistically significance in the R^2 mean from method one to method two of 0.07085[t(38=-4.131, p<0.001].

Similar to method one, inaccuracies in the method will exist due to rounding off estimations. However, in method one, the biggest variations occurred with the large differences in stimulation pairs, which were removed with the standard always being in the middle.

In addition, the informal observations found that some subjects struggled with understanding of the methodology technique one with the standard only being provided at the beginning of the round, which did not appear to occur with technique two where the standard was presented for every stimulation.

It is important to note that the proportional increase for the two locations tested is the same, however, they may still have different sensitivities; i.e. an initial rotation of four degrees may feel different levels of intensity on the different location. However, the ratio increase from this is the same; once the locations are calibrated to the same stimulation intensity, they will both increase and decrease at the same rate.

3.5 Summary

The mechanotactile stimulation method presented in this chapter has demonstrated to be an effective and low-cost approach that could be used in grasping force feedbac
k for a prosthetic hand with three channels of feedback. With a short training period, recognition rates of up to 80% were achieved with six different grip patterns at two different intensity levels. This approach has the advantage of being easily applied, removed, adjustable location, only adds minimal bulk and has a maximum delay time of 162ms.

In achieving the similar results as Antfolk et al. [16] with more than twice as many grip options, this stimulation method has demonstrated the benefit of combining skin stretch with the vertical pressure. The skin stretch was also demonstrated to result in a better result when applied longitudinally to the forearm, shown by the statistically significant improvements in recognition rate compared to the other orientations.

The results obtained from ten subjects show a high level of discrimination in the ranges of 1.4-2.1 degrees for three reference stimulation values of 10^0, 15^0 and 20^0 at the ventral and ulnar regions of the forearm, as shown in Table 3.3. These JND values appear consistent across the stimulation ranges and do not statistically differ from the results between the two locations tested on the forearm.

A very strong linear relationship is obtained between the applied rotation and the perceived intensity level of stimulation, instead of following Steven's Power Law. This strong linear relationship provides a methodology to consistently induce a desired level of sensation on the users' forearm. This relationship is shown to be consistent between forearm locations and between subjects, suggesting that it is not subject-dependent.

Chapter 4

Development of Flexible Concentric Electrodes and their Characterisation

4.1 Introduction

Currently, disposable electrodes are the main type of electrodes used in sensory feedback research. Although some flexible electrode arrays have been developed [62, 69], no reusable concentric electrodes have been found in the current literature. Although commercial dry electrodes are used in TENS stimulation for physiotherapy and pain relief purposes, they are typically larger in size and are used in sensory feedback, particularly when with multiple channels. In this chapter, the process of coating a 3D printed flexible substrate with a thin layer of conductive graphene ink is presented, to create a low-cost reusable flexible electrode that can be used in the application of electrotactile stimulation without the need for additional adhesive. In addition, this chapter also aims to provide a comparison between the two electrode arrangements used for electrotactile stimulation, the concentric electrode and separated dual electrodes. Similar to the analysis undertaken by Geng et al. [130] who compared subdermal and transcutaneous electrodes, qualitative and quantitative psychophysical properties are measured and compared. To this end, the dynamic range and JND of transmitted currents are determined; and the comfort, spread, intensity and type sensation induced, and the resulting EMG interference from electrical stimulation through the different electrode arrangements are ascertained and analysed.

There are two main electrode arrangements used in sensory feedback literature, disposable concentric electrodes [63, 67, 69, 71-74] and disposable separated electrodes [64-66, 68]. Concentric electrodes are often used instead of separated anode and cathode electrodes as literature suggests that they can minimise electric current spread and improve localisation of the induced sensation [98, 118, 131], decrease in crosstalk with the EMG signal [73] that is often used for prosthesis control [132]. Although there are some techniques to avoid minimising or ignoring the cross talk [72-74], reducing the interference will help assist this process. As prostheses move towards control over multiple digits [133], providing multiple channels of sensory feedback will become vital and an electrode's localisation is therefore an important performance characteristic. In addition to this, having the stimulation localised to a contained area, can result in higher consistency of sensation, which improves a user`s ability to interpret feedback [132].

Comfort of sensory feedback is also a priority for users [37] and is important for its acceptance, particularly when it has been reported that vibrotactile stimulation is preferred over electrotactile feedback from dual separated electrodes [95]. Although these resulting improvements are suggested within literature, there is no previously reported study, which compares both qualitatively and quantitatively, the effect of different electrode arrangements on the performance of the electrotactile stimulation.

4.2 Background

Of non-invasive sensory feedback techniques currently being researched [118], electrotactile feedback shows a high potential due to its higher bandwidth [98] and lower power requirements than mechanotactile feedback [19, 87, 105, 125] and vibrotactile feedback [5, 45, 48, 94, 134]. Further, it has a potential for a higher available bandwidth to communicate information [98] due to the multiple parameters of pulse width, frequency, amplitude and location of stimulation being available for reliable manipulation. Multiple studies have also shown the potential of incorporating electrotactile feedback with myoelectrically controlled prosthetic devices [62, 63, 72].

Flexible reusable electrodes have been previously developed [135-137] for applications such as electroencephalogram (EEG), electrocardiogram (ECG) and EMG. These electrodes are typically smaller in size to offer higher resolution in signal recognition. However, using electrical stimulation for the purpose of sensory feedback requires larger electrodes to produce a comfortable sensation [24, 138, 139]. In addition, the high impedance value for electrical stimulation [140], or the conductive material based on sputtered metals reduce it stretchability [135], and therefore loses its flexibility when a larger surface area is required.

Due to the wastage of materials, cost, skin irritation and signal degradation over time resulting from the use of disposable electrodes, several studies have researched viable reusable replacement electrodes. Polymers mixed with either silver microparticles or carbon additives have been considered for their application in ECG and EEG recordings [141, 142]. Rubber and fabric-based materials have also been examined for creating flexible reusable electrodes in EMG signal detection [143]. Krachunov and Casson used 3D printing to create rigid dry EEG electrodes and painted them with a silver coating to increase their conductivity. For electrical stimulation on the forearm, however, flexibility is important to conform to the surface of the arm and skin.

4.3 Electrode Development

The basic electrode structure was 3D printed (Flashforge Inventor) using a commercially available thermoplastic poly(urethane) (TPU) known as Ninjaflex (NinjaTek, USA) material in three sections; inner electrode, separator, and outer electrode; as shown in Figure 4.1a. The print was performed using a layer height of 0.18mm, a fill density of 35%, an overlap of 30% and three perimeter shells. The inner electrode has a diameter of 15mm. This is to match the size of the disposable electrode to produce an equivalent current density. The separator has a width of 5mm, and the outer electrode has an inner diameter of 20mm and an outer diameter of 35mm. This outer electrode size was chosen from initial testing of the electrotactile stimulation to produce a comfortable sensation. Flexibility of the 3D printed electrode is demonstrated by clamping across the electrode and twisting it, as shown in Figure 1d, and the electrode can undergo high deformations with no permanent damage to either the structure or its conductivity. Due to this flexibility of the base material, the components compress as they are pushed together and stay connected without the need for any adhesives. This allows for easy disassembly and reassembly for cleaning and sterilization. All three sections have a 3mm thickness to provide an effective compression fit when pushed together. The inner and outer electrodes have a knob on top (4mm x 2mm x 3mm high, Figure 4.1a to d) to allow easy attachments to the electrical stimulator and other measurement and testing devices.

4.4 Electrode Characterisation

To enable easy application to the human arm, an off the shelf conductive TENS adhesive (TAC GEL) was applied to the bottom surface of both the graphene coated sections of the electrode. This enables the electrodes to stick to the arm without the need for tape. However, demonstration testing was conducted both with and without the conductive gel to compare the electrode performance. The chemical analysis and characterization is described in [144].

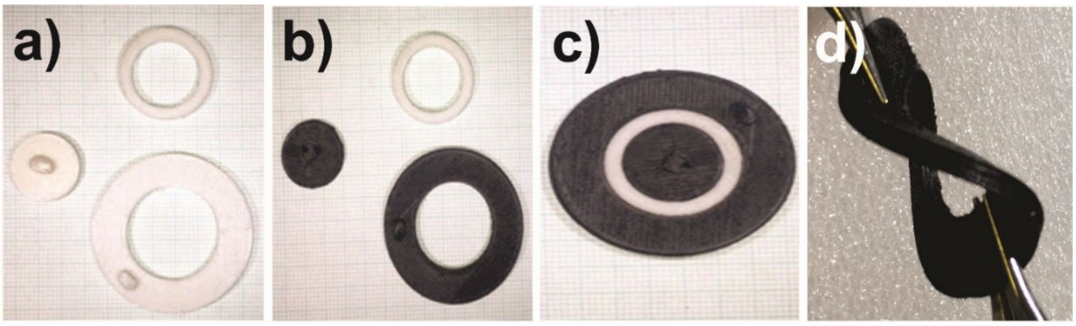


Figure 4.1 - Electrode (on 1mm grid paper): (a) 3D Printed Uncoated Electrode components, (b) Coated Electrode Components, (c) Assembled Electrode, (d) Demonstration of Electrode's Flexibility

4.4.1 Sheet Resistance

The sheet resistance was measured by a 4-point probe system (Jandel RN3) using a square array probe with 0.635mm spacing. ten readings were taken, measuring both the forward and reverse current from five different locations, and the average sheet resistance was calculated across these ten samples. The average sheet resistance of the graphene coated electrode across the ten readings taken was determined to be 903.5 ± 262.15 Ω/□.

Since the sheet resistance is a characteristic used to compare the conductivity of thin materials, it would be invalid to measure the conductive of the material in the disposable electrode due to its large thickness. Therefore, the conductivities of the disposable electrode and graphene-based electrode were compared using the impedance measurements, as shown in Section 4.5.

4.4.2 Scratch Test

To ensure robust adhesion of the graphene coating, a scratch test was performed. This was conducted by scraping the electrode with a pointy hook tool, shown in Figure 4.2, followed by a pair of tweezers. After both scraping sessions, no marks or damage was visible on the electrode and no change in impedance was recorded.

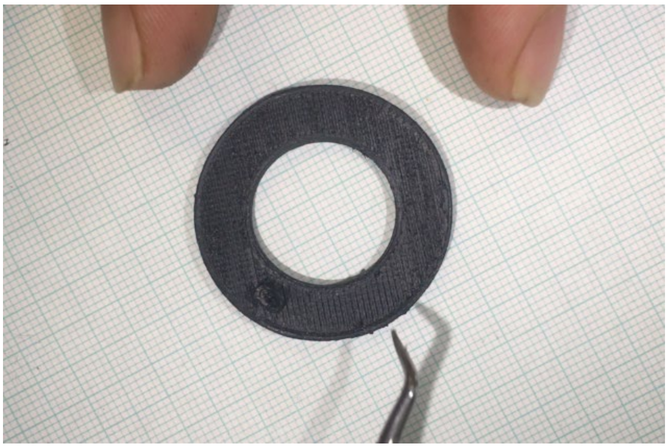


Figure 4.2 - Scratch test performed with hook tool.

4.4.3 Environmental and Financial Cost

In addition to providing more versatility in custom electrode design, this electrode design has potential to result in a financial saving and a significant reduction of the environmental impact of regularly using disposable electrodes.

In this analysis, the calculations are based on a batch of ten concentric electrodes being produced at once, which in addition to resting and drying time, requires two hours of ink preparation and roughly ten minutes to spray. This equates to approximately 13 minutes of preparation time per electrode, which would reduce when making a larger batch as there would be a minimal increase in ink preparation time. Table 4.1 outlines the costs of the materials required for both printing the base material and the ink coating. This does not consider the cost of equipment required for ink preparation/spraying or 3D printing. The largest cost of the electrode is from the Ninjaflex filament, which could be reduced in size, particularly in developing thin flexible electrodes to be embedded in a fabric.

Based on the durability of the electrode demonstrated in the scratch test, and the known flexible properties of the Ninjaflex materials, a 1-year life-time is estimated for the custom printed flexible concentric electrode. Further analysis and testing is required to determine any reduction in performance or durability over longer periods of time and repeated use. Within this period of time, using one pair of disposable electrodes per day would result in a total use of 730 electrodes. At an approximate costing of $1.30 per a disposable electrode [58], using the concentric electrode proposed in this study would result in a significant saving both financially and environmentally as a result of the reduction in waste produced.

Table 4.1 - Material costs of concentric 3D printed electrode for a batch of ten concentric electrodes.

Material & Price	Amount Required	Price per batch ($)
SEBS – $0.5 / 1kg	0.5g	0.0025
Toluene - $73.5 / L	3mL	0.2205
Graphene - $50 / 5g	20mg	0.2
Ninja Flex $93 / 750g	3.3g per concentric electrode	4.11
Total material cost per batch		$4.53
Total material cost per electrode		$0.45

4.4.4 Impedance Measurements

Impedance measurements were taken using an MFIA Impedance Analyzer (Zurich Instruments) from 1kHz to 1MHz. Due to the different locations that result from placing a concentric electrode (Figure 4.3c) compared to disposable electrode pairs (Figure 4.3a), it would be invalid to compare impedances between the two. Therefore, an additional test was conducted using dual graphene electrode pairs (Figure 4.3b). This was to enable a comparison based on the material properties and the electrode geometrical configuration.

five different electrode combinations were, therefore, tested for comparison: 15mm disposable electrode pairs (Figure 4.3a); dual 15mm graphene covered electrode pairs (Figure 4.3b), tested dry and with conductive adhesive; graphene coated concentric electrode (Figure 4.3c), tested dry and with conductive adhesive.

Typical pulse width range used in electrotactile stimulation for prosthetic sensory feedback ranges from as low as 50μs up a value of 500μs [65]. Therefore, the frequency band of interest is 1kHz – 10kHz. As shown in Figure 4.4, although the disposable electrode's impedance values are slightly higher, the graphene-coated electrodes are comparable within this frequency range. In addition, the concentric configuration (Figure 4.3c) also slightly reduced the impedance of the electrode; however, this would be largely due to the fact that the current flows through a smaller distance within the body.

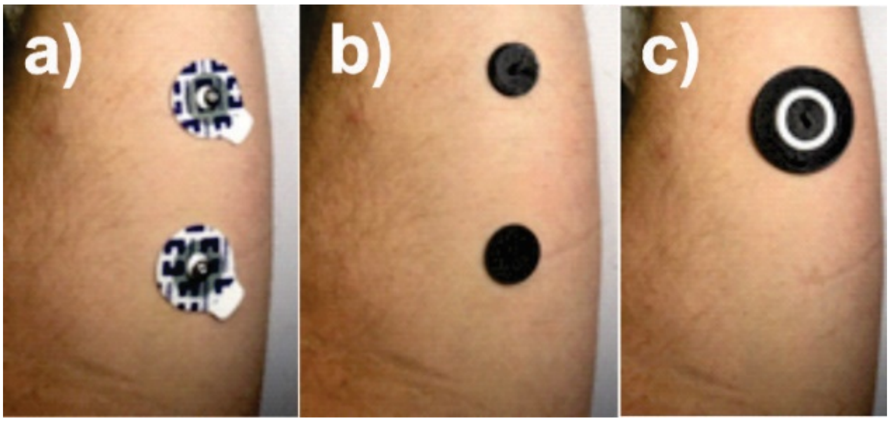


Figure 4.3 - Positioning of electrodes for impedance test: (a) Dual Disposable Electrodes, (b) Dual 15mm Graphene Coated Electrodes (shown here with adhesive), and (c) Concentric Graphene Coated Electrodes (shown here with adhesive).

Testing in this study was conducted at a pulse width of 100μs which corresponds to a frequency of 5 kHz. At this frequency, the corresponding impedance is ~3.2kΩ for the disposable electrodes, ~6.2kΩ for the concentric graphene electrodes, and ~ 8kΩ for the dual graphene electrodes.

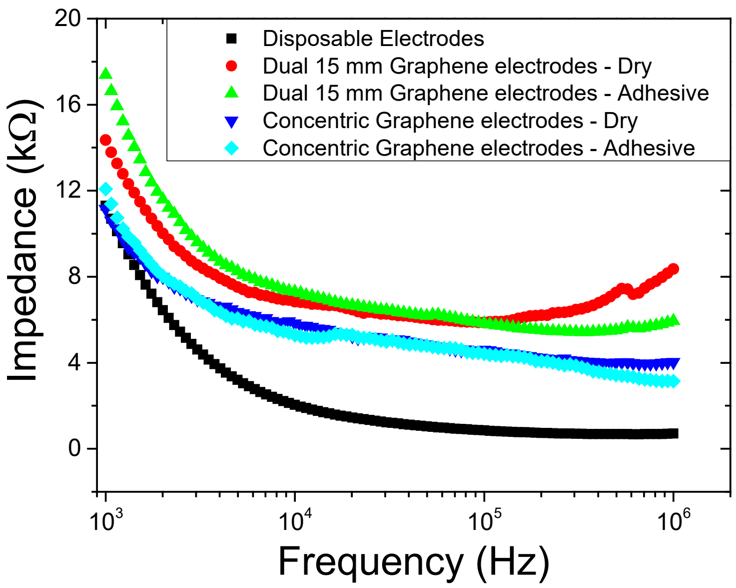


Figure 4.4 - Impedance measurement from 1 kHz to 1 MHz

4.4.5 Application Demonstration

Stimulation was provided through a BioPac constant current linear isolated stimulator (STMISOLA) controlled through a Biopac MP36 data acquisition system with a sampling frequency of 100 kHz for the stimulator. Stimulation was provided through a biphasic square wave with a pulse width of 100 μs, frequency of 10 Hz and an inter-pulse delay of 100 μs.

Although the stimulator produces square waves, due to the capacitance of the skin and the electrode, the transmitted waveforms have an associated rise time and do not form perfect square waves. To view these current waveforms flowing through skin, the transmitted current was recorded using a National Instruments Current Input Module (NI-9203) through a LabVIEW interface with a sampling rate of 200 kHz. A constant current biphasic square wave with a peak current of 4 mA was used for the electrotactile stimulation to ensure that it was within the comfortable and recognisable range. The pulse width, frequency and inter-pulse delay were left at 100 μs, 100 Hz and 100 μs, respectively.

A single pulse for each electrode pair is shown in Figure 4.5, with their associated rise times averaged from five sequential pulses. Although the disposable electrode pair has a slightly lower time, all electrodes produce comparable wave forms with comparable rise times. It is also worth noting that since the current input module had a maximum sampling rate of 200 kHz, it was only able to take a current reading every 5μs.

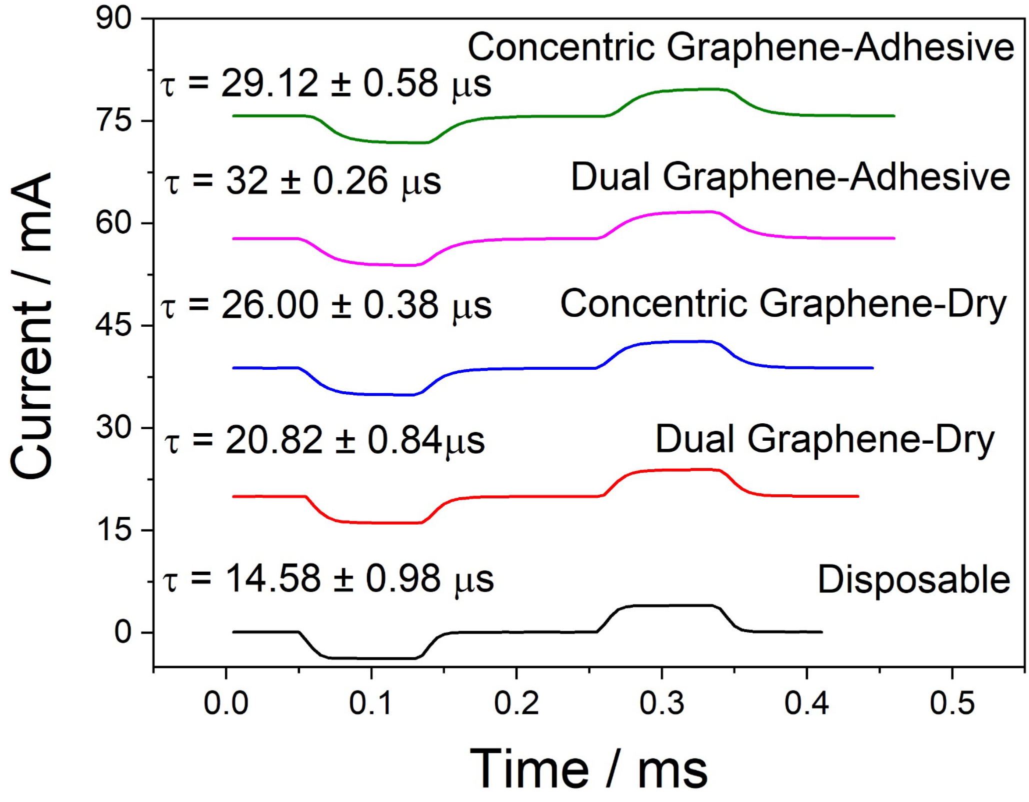


Figure 4.5 - Measuring Current from TENS Stimulation through various electrodes (Amplitude - 4mA, Frequency - 100Hz, Pulse Width - 100μs, InterPulse Delay - 100μs).

4.5 Electrode Geometry Comparison

4.5.1 Methods

Graphene coated electrodes [144] were used as they have been previously demonstrated to provide a reusable and flexible electrode with comparable performance to the disposable electrode [144] and allow us to create and examine different electrode sizes. . For the dual separated electrodes, two 15mm circular electrodes were used to be the same surface area as typically used in sensory feedback studies [64-66, 68]. A 40mm concentric electrode was used to be the same size as disposable electrodes used within literature [63, 67, 71, 72]. The inner concentric electrode and disposable separated electrodes have an area of

176mm^2. The outer electrode of the large concentric pair has an area of 942mm^2. It has been previously suggested that a large ratio of the outer electrode to the inner electrode should be used to maximise localisation [100]. Furthermore, larger electrodes have been shown to produce a more comfortable and a larger variety of sensation types [24, 138, 139]. To examine this in more detail, a smaller outer diameter of 35mm resulting in an area of 648mm^2 was also included in this study to determine if a smaller electrode can be used without any significant reduction in comfort, localisation and type of sensations (pressure, vibration, pain etc.) produced. There are many combinations of electrode sizes that could be tested to optimise the different desired characteristics, however, this lies beyond the scope of this study and instead analysis is limited to these three electrodes, with the main goal to examine the difference between the concentric and separated electrode configurations.

The electrodes were placed in the middle of the dorsal side of the subject's dominant forearm, approximately one-third of distance from the elbow to the hand, as shown in Figure 4.6. For each subject, the placement was marked with an "x" on their skin to ensure identical placement of the centre of the three electrodes tested. The second electrode for the disposable separated electrode was placed near the elbow, as shown in Figure 4.6a. An off the shelf conductive adhesive (TAC GEL) was used on the conductive sections of the concentric electrodes to secure the electrode to each subject's arm.

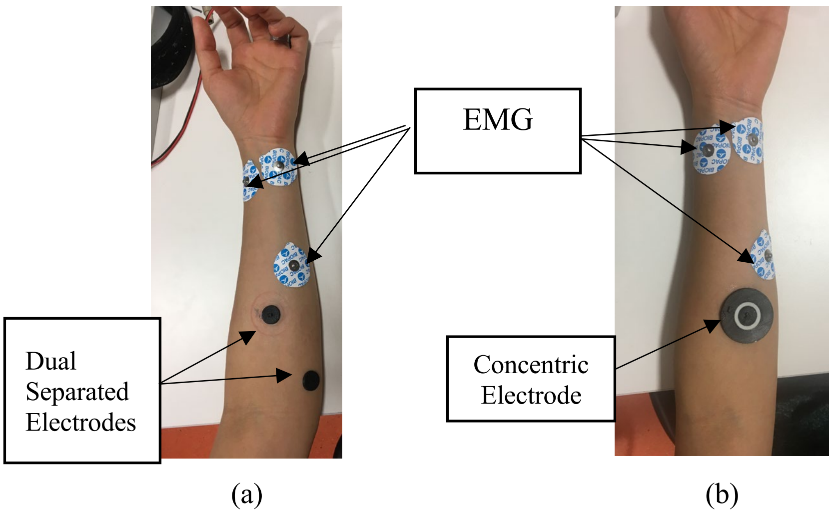


Figure 4.6 - Electrode placement, shown on a left handed participant: (a) Dual separated electrodes (b) Concentric electrode

Electrical stimulation was delivered by a constant current neurostimulator (Inomed ISIS) and controlled through a .NET API. Cathodic biphasic pulses were used with a pulse width of 100µs. A frequency range of 0-100Hz [131] has been proposed to be the most suitable for electrotactile stimulation. Therefore, three frequencies, 10Hz, 50Hz and 100Hz, were used to span this range.

Since the three electrodes have different surface areas resulting in different overall current densities, identically delivered current levels can result in different perceived levels of intensities. To compensate for this impact, each subject and frequency had their stimulation currents determined as a percentage of the difference between their detectable threshold (DT) and pain threshold (PT) for that electrode and frequency combination, as determined in subsection 4.5.1.a). Three current levels were chosen, corresponding to 25%, 50%, and 75% of the difference between the DT and PT for the electrode and frequency combination for that subject (resulting in nine stimulation values per electrode per test subject). A statistical analysis was performed based on the average of these three current values for each individual.

The study consisted of four different experimental blocks; Tolerable current range of comfortably perceivable current levels (as outlined in 4.6.2), Perception of induced sensation (outlined in subsection 4.6.3), JND (as outlined in subsection 4.6.4), and induced EMG interference (as outlined in subsection 4.6.5). To ensure there were no changes in electrode placement, all four experimental blocks were performed on one electrode prior to proceeding to the next electrode. Further, the EMG electrodes remained in place for the all three electrodes' testing and a 5-minute break was given between the electrode tests to reduce any impact of fatigue or desensitisation. The order of each electrode tested was altered between subjects to eliminate any impact of desensitisation, with the six possible order combinations for the three electrodes tested twice in total. For this experiment evaluation, 12 subjects were tested, consisting of nine male and three females, with an average age of 27.2 ± 5.7 years. Each subject's experiment was performed within a 1.5 hour session.

a) Dynamic range of the comfortably perceivable current levels

Using the Staircase Method [145], current was increased in intervals of 0.1 mA until sensation was detected, then decreased by 0.01mA disappeared and then increased by 0.01mA until the sensation re-appeared. This point was recorded as the DT.

Similarly, for the PT measurements, starting from the DT, the current was increased in intervals of 0.1mA until the user stated that it was no longer comfortable, and this was set as the PT.

b) Perception of Induced Sensation

Using the subject's nine stimulation values as determined in subsection 4.6.2, the stimulation was provided to the subject for a period of two seconds. This period was chosen to allow the participants enough time to focus on the stimulation whilst minimising any impact of desensitisation from longer stimulations.

For each of the tested stimulations, subjects were asked to select the appropriate quality of sensation, rate the intensity and comfort, and mark the location of the perceived sensation on a provided grid (Figure 4.7). For the quality of sensation, subjects were asked to select from 12 predefined descriptors [130]; pressure, tap, vibration, tingling, pinprick, itch, pinch, pain, warm, cold, movement, or muscle twitch. When rating the intensity, the subjects were required to rate the intensity on a scale of 0 – 10; and for grading the comfort they used a scale of 1-7 where 1 represented very comfortable, 4 neutral, and 7 represented very uncomfortable.

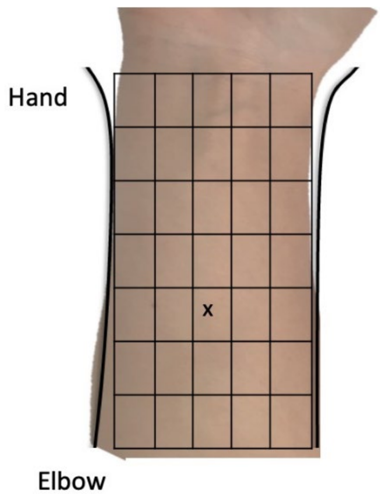


Figure 4.7 - Participants response sheet for localisation experiment

To obtain a representation of the area/s on the arm where a sensation was induced, the subjects were asked to mark all areas on the arm where stimulation occurred by using the image shown in Figure 4.7. The 'x' represents the location of the dual electrode on the centre of the forearm, or the centre electrode for the concentric electrodes. Subjects were instructed to indicate the areas stimulated using two different relative intensity levels, corresponding to a large or small sensation felt in that marked region, which was taken account in the analysis shown in 4.5.2.c).

c) JND

A two-alternate force choice method (2AFC) [122] was used to determine the JND of electrical stimulation through each of the electrodes. Pairs of one second stimulations were sent to the subject separated by a two second period of no stimulation to avoid any possible effects of desensitisation. Each stimulus pair consisted of a reference value (R) and small increase/decrease of the reference value ($R \pm \delta x$). The subject was required to identify the stronger of the two stimulation values received. The JND was only examined at 50Hz. The reference current was determined for each subject and each electrode, corresponding to 50% of the difference between the DT and PT for that subject and electrode combination. Although these current levels are different for each electrode, they will provide a better comparison of the JND across the acceptable values for the subjects. Stimulation pairs with differences of 0.1, 0.3, 0.5, 0.7, 0.9, 1.1 and 1.3mA to the reference value, tested both as an increase and decrease, resulting in 14 different pairs per reference value. A total of 112 stimulations pairs were tested for the reference value per electrode per subject; consisting of each pair tested eight times with the larger amplitude first (E.g. 4.3mA followed by 4.2mA), and eight times with the larger amplitude second (E.g. 4.2mA followed by 4.3mA), providing 16 stimulation pairs for each tested difference level.

112 test pairs for each electrode were presented to each subject in a random order and the subjects were given a 2-minute break every 28 test pairs.

Each subject had their own psychometric function fitted to their data and a mean of each of the individual subject's JND was calculated, which was used to determine if there was any statistically significant difference between the JND values for each electrode. The psychometric functions were fitted using the psnigifit toolbox v4.0 for Matlab

(MathWorks) [126], which produced a JND threshold equal to the midway point between the lower and upper asymptotes, with the lapse rate set at 0.02.

d) EMG Interference

EMG electrodes were attached to the dominant forearm of the subjects, as shown in Figure 4.8. A Biopac MP36 data acquisition system was used to record the EMG data with a sampling rate of 25,000 Hz. Although this is well above the sampling rates commonly used in EMG for prosthetic control [146], it was required to meet Nyquist sampling theorem for the small pulse width used (100µs) in the stimulation.
Each subject was asked to demonstrate a strong muscle contractions to produce reference EMG data levels for comparison. The subjects were then provided with three electrical stimulations, for each frequency with the current pulse amplitude set to 50% of the difference between the subjects' PT and DT for that electrode and frequency combination. The average height of the resulting peaks of each spike induced in the EMG signal was determined for each frequency. To ensure the measurement was based on the peaks from the electrical stimulation and not from background EMG noise, only the peaks with an absolute value above 30% of the signal's maximum value were included.

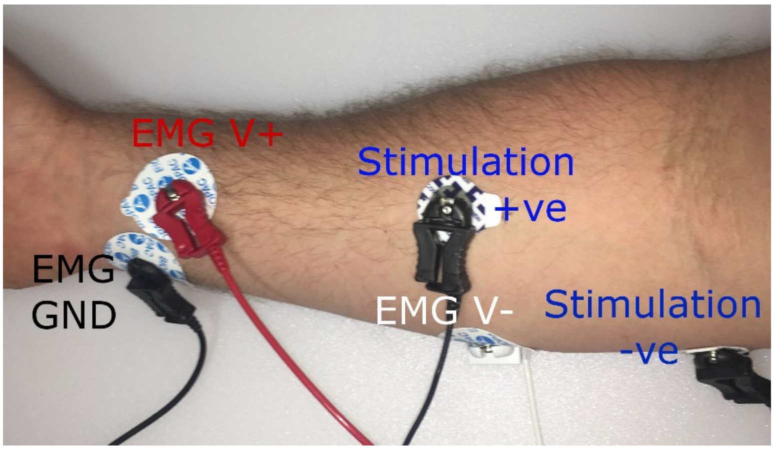


Figure 4.8 - EMG interference recording setup

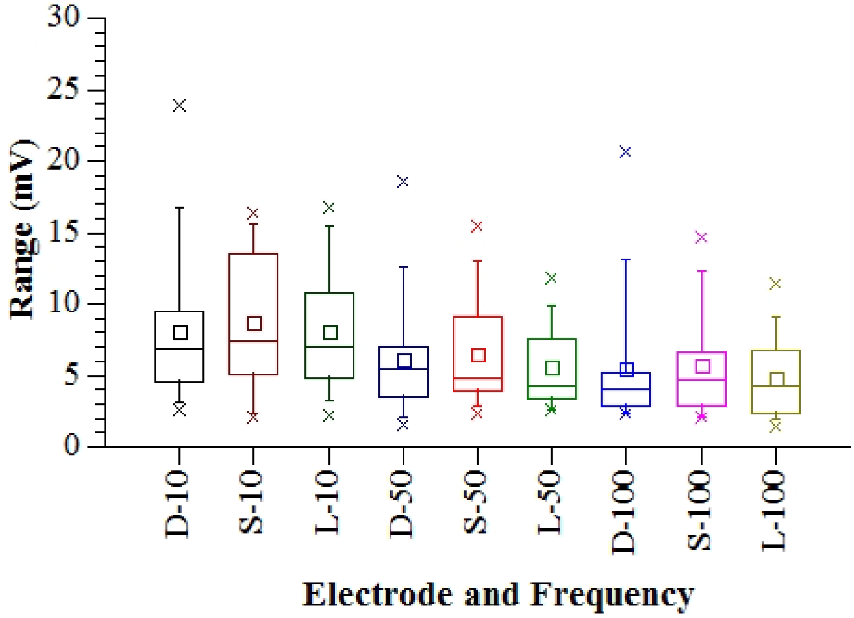
4.5.2 Comparison Results and Discussion

a) Dynamic Range

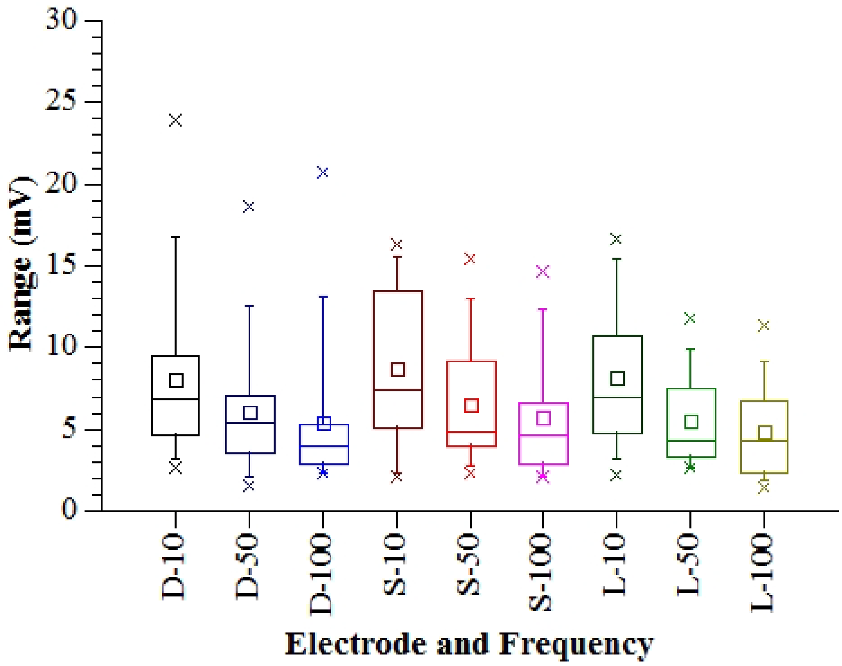
The average detectable threshold, pain threshold, range of current and dynamic range over the 12 subjects are shown in Table 4.2 and Figure 4.9. To determine the impact of frequency and the type of electrode on the dynamic range, a nested and repeated measures ANOVA was applied. Using the Greenhouse-Geisser estimate, the impact of the electrode met the assumption of sphericity ($\varepsilon=0.820$, $p=0.107$). However, the impact of the frequency did not meet the assumption of sphericity ($\varepsilon=0.734$, $p=0.036$), and a Greenhouse-Geisser adjustment was therefore used. The analysis showed that there was no significant impact of the electrode on the range of acceptable currents [$F(2,22)=0.451$, $p=0.643$] but there was a statistically significant impact of the frequency on the range [$F(2,22)=0.451$, $p=0.643$]. Post hoc tests using a Bonferroni correction showed that there was a statistically significant difference between the current range of 10Hz to both the 50 Hz (mean=2.26mA, $p=0.002$), and to the 100 Hz (mean – 0.567mA, $p=0.001$). The mean difference between the current range of 50Hz and of 100Hz is also approaching significance (mean = 0.711mA, $p=0.066$). As shown in Table 4.2, there are large variances for each measurement which demonstrates the large variability of inter-subject thresholds. This variance was overcome by each subject using their own current levels based off their own individual DT and PT for all subsequent tests. In addition, all subjects underwent all tests which allowed for repeated measures statistical analysis.

Table 4.2 - Average acceptable currents

	Frequency (Hz)	**Range (PT-DT) (mA)**
Dual Separated	10	8.0 ± 5.7
	50	6.1 ± 4.4
	100	5.4 ± 5.0
Small Concentric	10	8.7 ± 5.0
	50	6.5 ± 3.9
	100	5.7 ± 3.8
Large Concentric	10	8.1 ± 4.5
	50	5.5 ±2.9
	100	4.9 ± 2.9



(a)



(b)

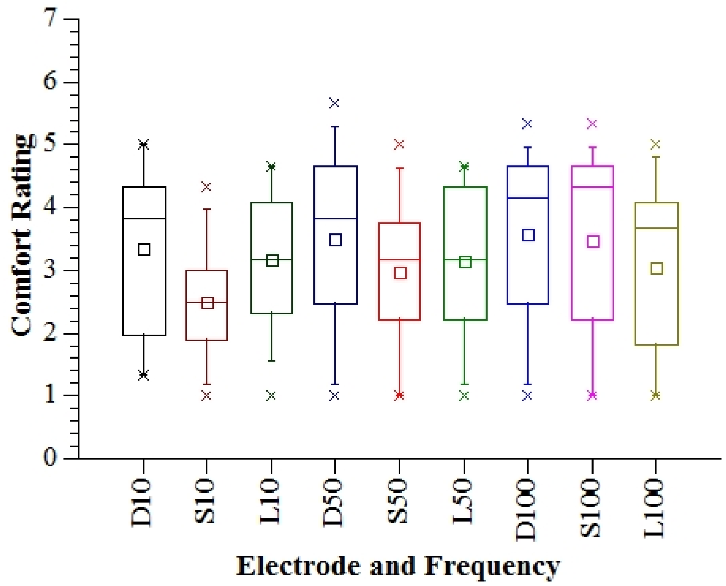
Figure 4.9 - Range of acceptable current. (a) Grouped by Electrode, (b) Grouped by Frequency). The small square represents the mean, the box contains the middle two quartiles, the whiskers correspond to the 5th-95th percentile, and the cross marks indicate maximum and minimum values. D – Dual Separated; S – Small Concentric, L – Large Concentric.

b) Perception of Induced Sensation

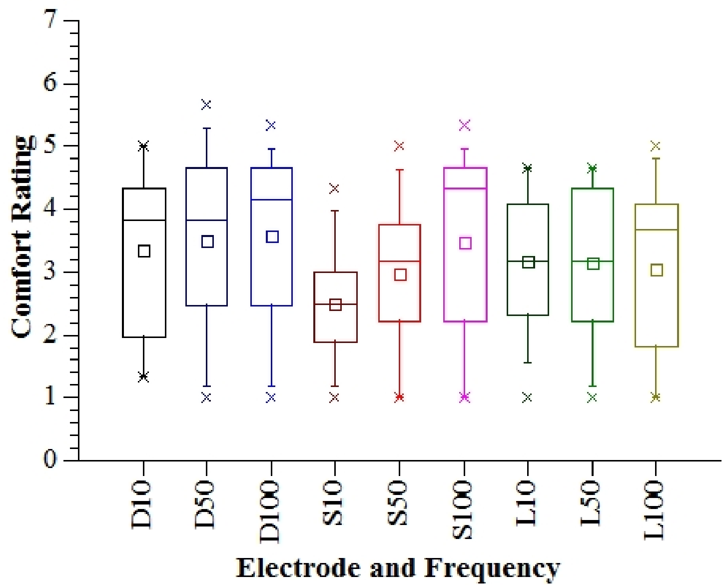
i. Comfort of Sensation

The average comfort rating given by 12 subjects for all three frequencies is shown in Figure 4.10, where the results were supplied on a scale of 1-7, where 1 represents very comfortable and 7 represents very uncomfortable. To determine the impact of frequency and the type of electrode on the perceived comfort, a nested and repeated measures ANOVA was applied. Using the Greenhouse-Geisser estimate, the impact of the electrode on the perceived comfort did meet the assumption of sphericity ($\varepsilon=0.840$, $p=0.129$). The analysis showed that the impact of the electrode type on the perceived comfort level was approaching significance [$F(2,22)=3.420$, $p=0.051$]. Post hoc tests using a Bonferroni correction, however, showed that the biggest difference came between the small concentric electrode and the dual separated electrode. However, this means that a difference of 0.5 still did not reach significance ($p=0.099$). In addition, the mean difference between the large concentric electrode and other electrodes were not significant, ($p=0.968$, $p=0.473$ for the small and dual electrodes respectively.). Further investigation is required with an even smaller concentric electrodes, and larger sample sizes are, therefore, required to determine if there is a statistically significant difference in the comfort levels.

When analysing the impact of frequency on the perceived comfort, it met the assumption of sphericity using the Grenhouse-Giesser estimate of sphericity ($\varepsilon=806$, $p=0.092$). The analysis showed that the effect of the frequency on the perceived comfort was not significant $F(2,22)=1.115$, $p=0.346$. Post hoc tests using a Bonferroni correction showed that the mean difference of 0.311 between 10Hz and 50Hz was approaching significance ($p=0.052$) and the mean difference of 0.6 between 10Hz and 100Hz was significant ($p=0.011$). The mean difference of 0.289 between 50Hz and 100Hz was also approaching significance ($p=0.071$).



(a)



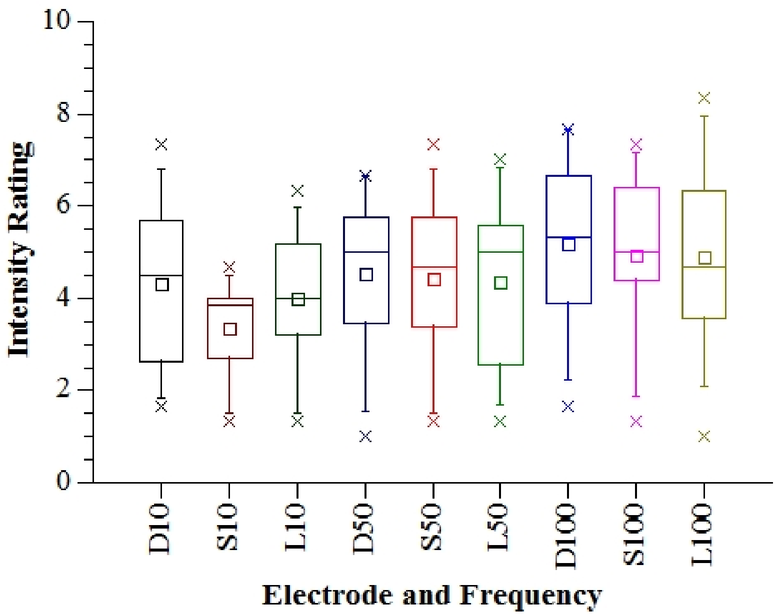
(b)

Figure 4.10 - Comparison of the comfort levels with the electrodes over the three frequencies. Rating scale from 1 - very comfortable to 7 - very uncomfortable. (Grouped by Electrode, (b) Grouped by Frequency. The small square represents the mean, the box contains the middle two quartiles, the whiskers correspond to the 5th-95th percentile, and the cross marks indicate the maximum and minimum values. D – Dual Separated; S – Small Concentric, L – Large Concentric.

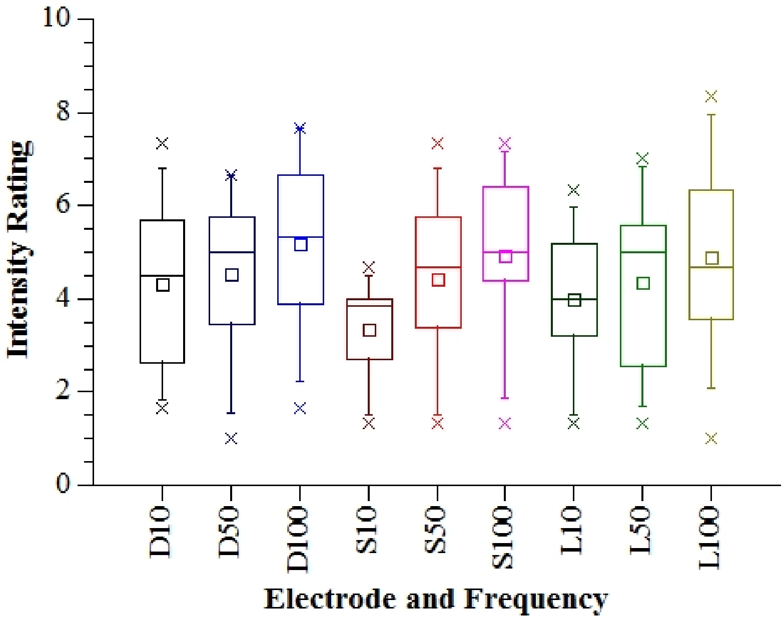
ii. Intensity of Sensation

For each of the nine stimulations, subjects rated the intensity of their stimulations from 0 to 10. These results were grouped together with the corresponding stimulation frequency and electrodes, as shown in Figure 4.11. As expected, the increase in frequency results in an increase in stimulation, as shown in Figure 4.11. To determine the impact of frequency and the type of electrode on the perceived intensity, a nested and repeated measures ANOVA was applied. The perceived intensity met the assumption of sphericity using the Grenhouse-Giesser estimate of sphericity (ε=1.000, p=0.911). The analysis showed there was not a significant impact of the electrode type on the perceived intensity [$F(2,22)=1.198$, $p=0.321$].

Using the Greenhouse-Geisser estimate, the impact of the frequency on the perceived intensity did meet the assumption of sphericity (ε=0.0.758, p=0.052). The analysis showed that the effect of the frequency on the perceived intensity was significant $F(2, 22)=5.977$, $p=0.008$. Post hoc tests using a Bonferroni correction failed to show any significant pairwise differences, (p=0.294 and p=1.00). However, since the ANOVA showed a difference and the Bonferroni correction is very conservative, the analysis was repeated with a Least Significance Difference (LSD) correction on the post hoc tests. These results showed that the mean differences for 100Hz, 10Hz and 50 Hz were significant ((mean = 1.102, p=0.02) and (mean = 0.556 p=0.03), respectively). This data matches previously reported results that a higher frequency for the same current level results in a higher perceived intensity level [147].



(a)



(b)

Figure 4.11 - Comparison of the intensity levels with the electrodes over the three frequencies (scale 0-10). (a) Grouped by Electrode, (b) Grouped by Frequency. The small square represents the mean, the box contains the middle two quartiles, the whiskers correspond to the 5th-95th percentile, and the cross marks represent the maximum and minimum values.

iii. Location of Sensation

Figure 4.12 shows the results for the distribution of sensation induced by the electrical stimulation with the three frequencies combined. If a subject identified an area with half strength, it was given half the weighting in calculating the distribution of the intensity. This intensity distribution is shown in Figure 4.12, where 100% represents the stimulation always being felt in that location and 0 representing it never being felt in that location. Two of the subjects were left handed, and their grids were reversed to correspond with the same orientation as the other subjects. As shown in Figure 4.12b and c, both concentric electrodes have an extremely high chance of stimulation being induced on the location of the electrode, with a small probability of sensation also being induced in the surrounding areas. However, as shown in Figure 4.12b and c, both concentric electrodes have an extremely high chance of stimulation being induced on the location of the electrode, with a small probability of the sensation also being induced in the surrounding areas. However, as shown in Figure 4.12a, the dual separated electrodes do not consistently stimulate the centre forearm location and there is a high probability of the sensation being felt at the second electrode's location, which is closer to the elbow. Although this may not be an issue for communicating one channel of information, if multiple electrodes are used in the same region (e.g. on the same forearm), then this spreading of signal may result in additional confusion when locating and interpreting the source of the stimulation.

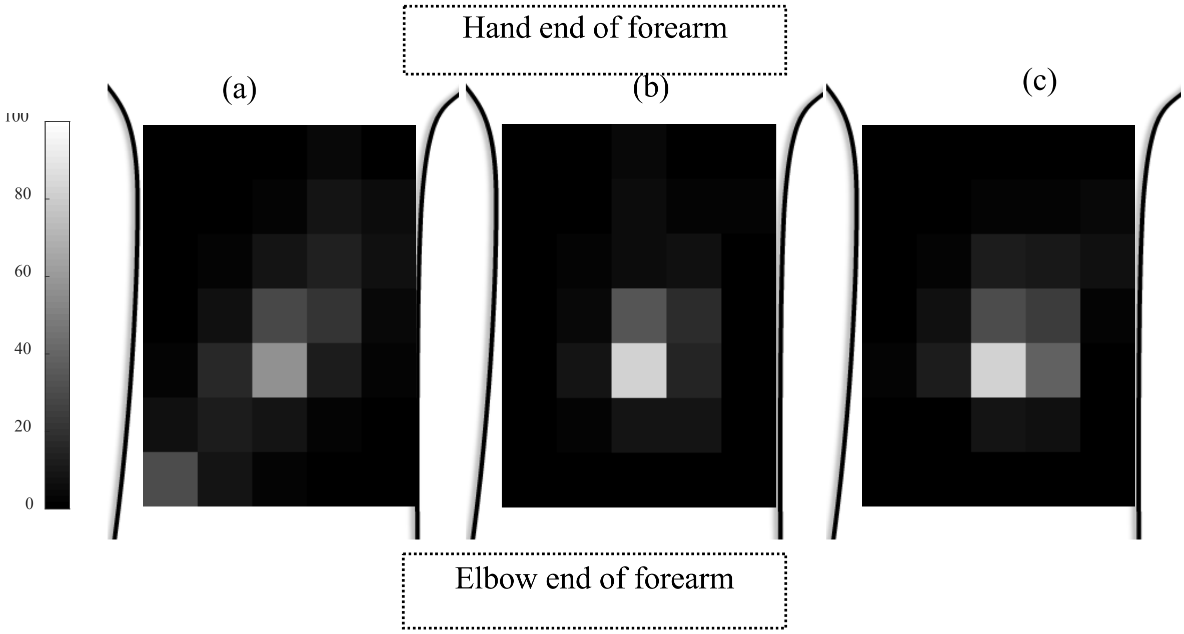


Figure 4.12 - Distribution percentage of perceived stimulation in locations across the forearm (a) Dual Separated Electrode Pair, (b) Small Concentric Electrode, (c) Large Concentric Electrode.

iv. Type of Sensation

A visualisation of the probability of the different types of sensations induced for the different frequencies on the three different electrodes is shown in Figure 4.13.

To better compare the type of sensations induced from the three electrodes, Figure 4.14 shows the difference in probabilities for each of the sensations for the three different frequencies. An alternate colour map was used for Figure 4.14 to easily distinguish the differences due to the range of values obtained. When it indicates a positive value, it represents the first electrode (e.g. small concentric in Figure 4.14a) f inducing that sensation more times. When the graph indicates a negative value, the second electrode induced that sensation a higher number of times (e.g. disposable separated electrode in Figure 4.14a inducing that sensation.

As can be seen, the dual separated electrode has a higher probability of inducing a pin-prick sensation and a slightly higher probability of inducing a pinch sensation on the subject. The subjects in the experiment undertaken by Geng et al. [130] also reported the sensation of pin prick being induced when using the separated surface electrodes. In addition, it has been previously reported that concentric electrodes result in inducing a lower amount of pain sensations [148]. The authors of [148] postulate that this is due to the edges of a concentric electrode, as there are lower current densities around the edges of the electrode [99] which is correlated with a lower chance of inducing pin-prick sensation [149]. When looking at whether the electrodes produced any of the three uncomfortable/undesired stimulations (pinprick, pinch and pain), the dual separated electrodes resulted in these sensations in 25.0% of all of the stimulations delivered, compared to the 11.1% and 12.0% for the small and large concentric electrodes, respectively.

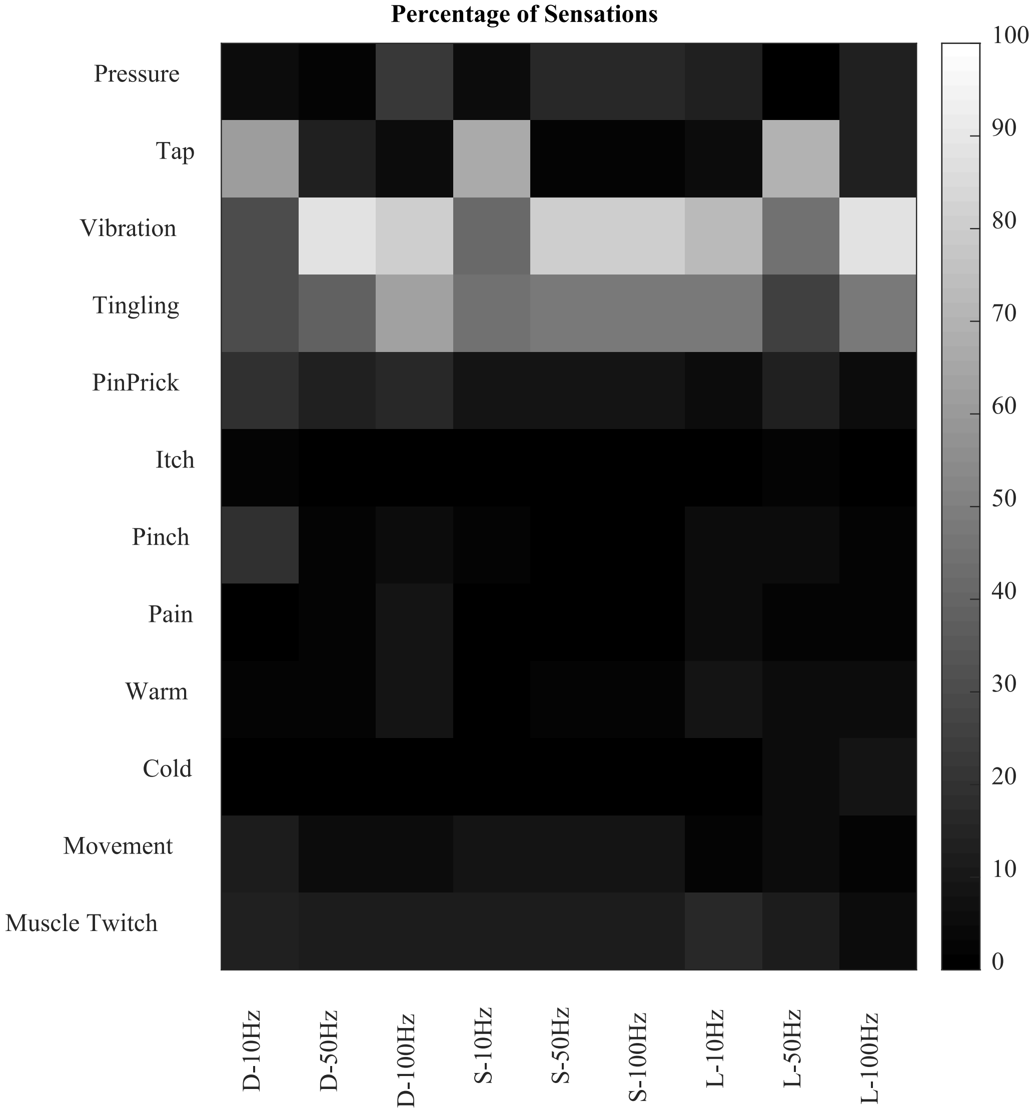


Figure 4.13 - Number of times sensations felt by the user for the three frequencies on all electrodes. D – Dual separated electrodes, S – Small concentric electrode, L – Large concentric electrode (27 total stimulations, each stimulation could produce more than one sensation).

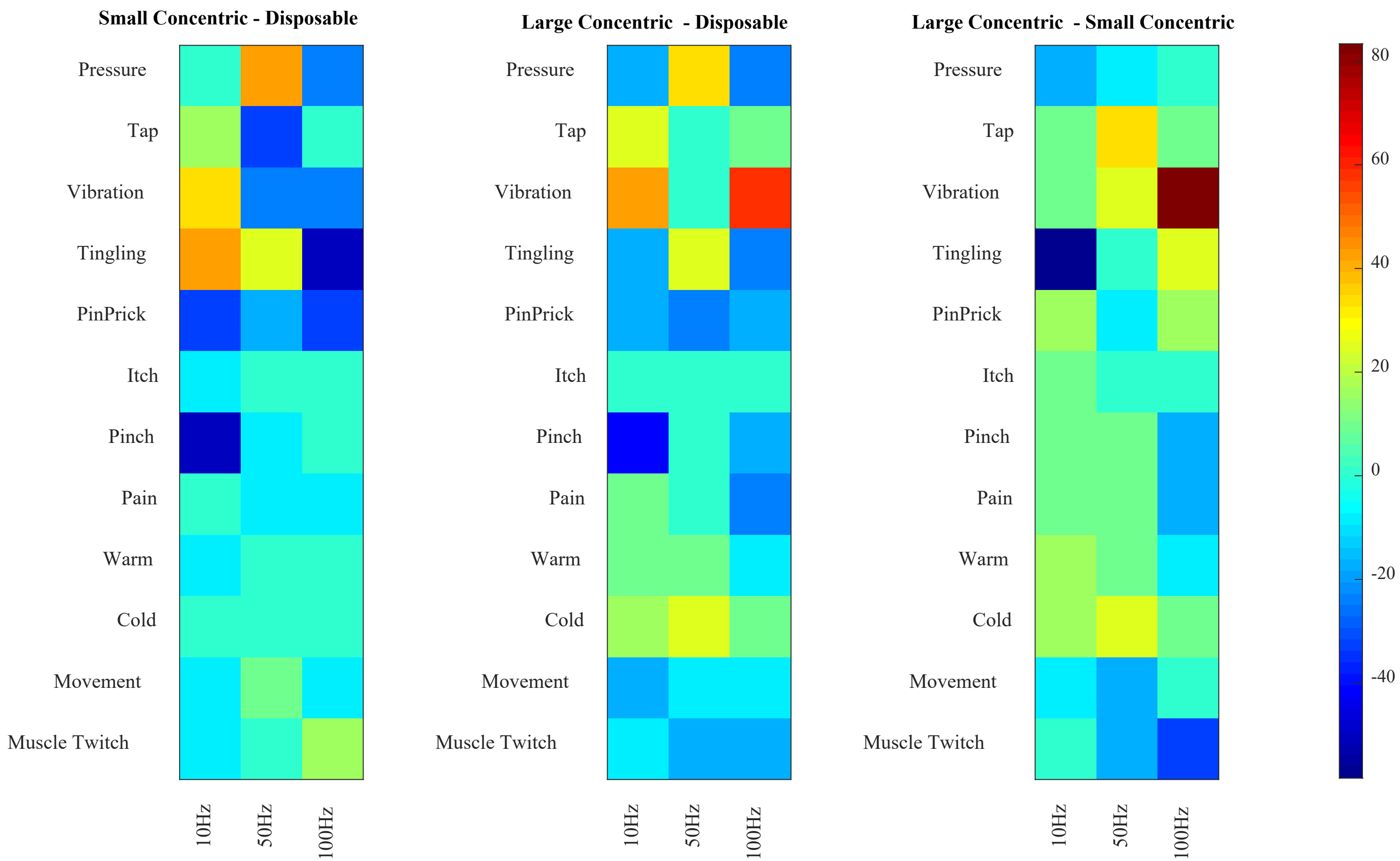


Figure 4.14 - Difference in probability of sensations induced by the different electrodes

c) JND

The average JND data for the different electrodes are shown in Table 4.3,presenting the JNDs calculated individually. Statistical analysis was performed on the individual JND results using a repeated measures ANOVA to look at the impact of the electrode type on the JND. Using the Greenhouse-Geisser estimate, the impact of the electrode on the JND did not meet the assumption of sphericity (ε=.808, p=0.094). Therefore, a Greenhouse-Geisser adjustment was used. The analysis showed that three electrodes did not have any statistically significant differences between their JNDs [F(2,22)=0.677, p=0.518].

Table 4.3 - JND Threshold Values

Electrode	Mean Individual JND (mA)
Dual Separated	0.44 ± 0.21
Small Concentric	0.46 ± 0.32
Large Concentric	0.54 ± 0.28

d) EMG Interference

Figure 4.15 shows the average peaks for the 12 subjects from the three different electrodes recorded at three different frequencies. A nested and repeated measures ANOVA was used to examine the statistical significance of the different results. Using the Greenhouse-Geisser estimate, impact of the electrode on the EMG interference did not meet the assumption of sphericity (ε=0.563, p=0.003), and the impact of the frequency was close to not meeting the assumption of sphericity (ε=0.563, p=0.089), therefore, a Greenhouse-Geisser adjustment was used for both the electrode and the frequency statistical tests. The analysis showed that there was a significant impact of the electrode type and frequency of stimulation on the induced EMG interference [F(1.126,10.137)=21.093, p=0.001] and [F(1.376,12.384)=57.733, p<0.001], respectively. Post hoc tests using a Bonferroni correction showed that the mean difference between the average peak interference from the disposable separated electrodes compared to the small and large concentric electrodes was significant [(39.124mV, p=.002) and (32.192mV, p=0.007), respectively].

Post-hoc tests using a Bonferroni correction showed that the mean differences between the various frequencies were significant [(10Hz and 50Hz: 3.508mV, p<0.01), (10Hz and

100Hz: 5.530mV, p<0.01), (50Hz and 100Hz: 2.022, p=0.002)]. This result aligns with the prediction previously made in literature [73]. However, in our tests, an amplitude of 50% between the PT and DT was used, which resulted in differing current levels. Since increasing frequency causes an increase in perceived intensity, as the frequency increased so was the current level. Therefore, these results will need to be repeated at identical current levels to determine if the small reduction in EMG interference is a direct result of increasing stimulation frequency, or just indirectly from the associate decrease in the stimulation current.

Even though the smaller concentric electrode produced the smallest amount of EMG interference, it is still significant when compared to the level of EMG produced by a large muscle contraction. This will be needed to be taken into account when using EMG and electrotactile feedback and will need to incorporate techniques such as time-division multiplexing [72], filtering [150] or placing the electrodes on a different body location to minimise the impact. However, any reduction in noise from electrotactile stimulation through careful electrode selection may further enhance the chosen technique.

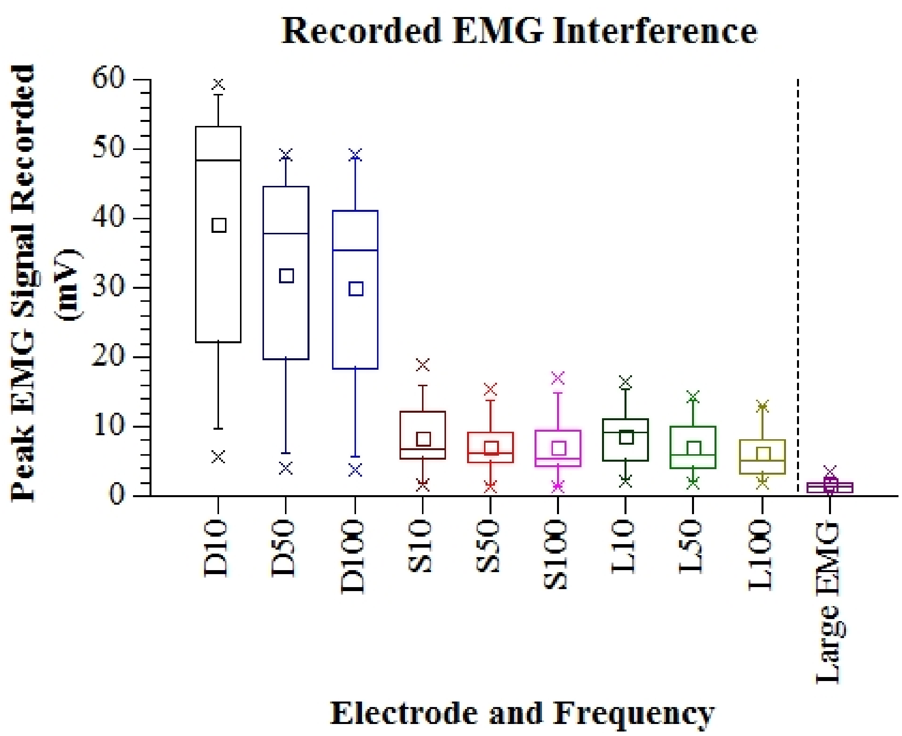


Figure 4.15 - Average peak EMG interference. . The small square represents the mean, the box contains the middles two quartiles, the whiskers correspond to the, Whiskers 5th-95th percentile, and the cross marks maximum and minimum values.

4.6 Summary

In this chapter, the reusable flexible electrodes for electrotactile stimulation to provide sensory feedback to amputees using prosthetic devices was presented. These affordable electrodes offer a more environmentally friendly option for a long-term use. These electrodes demonstrated a higher, but a comparable impedance to that of the disposable electrodes. The higher impedance resulted in a higher voltage required to maintain the desired current. Although this would increase the power consumption, with an effective duty cycle of 2% (for the 100μs pulse width used) this increase would be minimal.

Although the addition of conductive adhesive to the flexible electrodes made it easier to stay attached for testing purposes, there was no noticeable difference in performance between the graphene electrodes used dry or with the conductive adhesive. This suggests that the electrodes can be built into wearable fabrics. Removing the adhesive, that is often used in disposable and reusable electrodes, could reduce the level of irritation on the skin and reduction in performance over time [151, 152]. In addition, the movement of electrodes resulting in changing impedance levels is no longer a significant issue due to recent developments in electrotactile stimulators being able to compensate for this changing impedance [153].

Further testing is required to determine the optimum geometry and sizing of these electrodes. Since they are manufactured using additive manufacturing, they can be designed to match the curvature of different arm sizes. The electrodes tested in this chapter had a 3mm thickness to enable a stable or tight fit between the concentric components of the electrodes. However, if they were instead built into a fabric, this tight fit requiring a reasonable thickness would no longer be required. This would enable the electrodes to be printed significantly thinner, which would further increase their flexibility for a better conformance to the surface of the human arm. It must be noted that only two sizes of concentric electrodes were tested in this chapter. More experimentation is required to determine the impact of reducing or increasing the size of the outer electrode on the performance of the electrodes. In addition, the inner electrode size was kept consistent with the size of the disposable separated electrode. Therefore, further investigation is required to determine the effect of the size of the inner electrode on electrical stimulation, and overall performance of the concentric electrodes.

This chapter has presented the results from psychophysical experiments to compare the performance of two different electrode arrangements. The data presented demonstrate that the concentric electrodes can result in a reduction in uncomfortable sensations (pinprick, pinch or pain) being produced. Comfort of sensory feedback is a priority for users [37] and is important for its acceptance..

Further, there is an increase in the localisation of the area where the sensation is induced, which is particularly valuable when more than one channel/location of electrotactile stimulation is being used concurrently. As prostheses move towards control over multiple digits [133], providing multiple channels of sensory feedback will become vital and an electrode's localisation is therefore an important performance characteristic. In addition to this, a better localisation can result in a higher consistency of sensation, which improves a user `s ability to interpret feedback [132].

The different electrodes resulted in tolerable current ranges and JND that were not significantly different between the electrode geometries, however, bigger electrode sizes are required to determine if they are statistically comparable to each other. The concentric electrodes also resulted in lower induced EMG interference, but the interference produced was still larger than the EMG signal detected from a muscle contraction. This result aligns with the prediction previously made in literature [73].

Within the two concentric electrode sizes tested in this chapter, there was no statistical difference between the small and large concentric electrodes for comfort and perception of intensity. The larger size, however, resulted in a higher level of induced EMG noise. The optimum electrode size may also depend upon the application. For example, the smaller electrode may allow for better identification and recognition when using multiple stimulation sites simultaneously due to the smaller region being stimulated. The smaller concentric electrode is, therefore, used in the electrotactile stimulation experiments presented in Chapter 5 and Chapter 6.

Chapter 5

Comparison of Upper Arm and Lower Arm for Application of Sensory Feedback

5.1 Introduction

The upper arm has the potential to minimise interaction with the EMG interface, remove the need to interfere with existing sockets and provide a greater surface area for transradial amputees. In addition, it provides a potential feedback site for above elbow amputees. Fontana et al. [154] demonstrated that there was a similar recognition of vibrotactile stimulation on the upper arm compared to the lower arm due to their similar density of mechanoreceptors [155]. In addition, Stepp and Matsuoka [156] reported that for vibrotactile stimulation, the stimulation site has a minor effect on tactile feedback during object manipulation. This finding was obtained when the user received enough training with vibrotactile feedback. However, there have been no similar studies performed for mechanotactile stimulation and electrotactile stimulation applied to the upper arm and lower arm. Previous non-invasive sensory feedback methods consisting of mechanotactile, vibrotactile, and electrotactile stimulation have been applied to the upper [15, 44, 48, 49, 54, 56, 57, 60, 65, 66, 73, 74, 83, 105] arm and lower arm [16, 45-47, 52, 53, 58, 59, 61-64, 67-72, 80, 81, 85, 86, 125] regions.

This chapter aims to compare the sensitivity of the upper and lower arm through JND measurements using the mechanotactile stimulation device presented in Chapter 3, and shown in Figure 5.2. Sensitivity is important as small increments in sensory feedback stimulation levels are required to improve the fine control of grasping [157]. Further, the accuracy of recognition of our three channels of stimulation to these two arm regions through the use of the mechanotactile stimulation and electrotactile stimulation will be examined.

5.2 Methods

5.2.1 Sensitivity - JND

To determine the smallest perceivable change in stimulation, a 2-alternate force-choice method was employed to determine the JND. This technique sends pairs of stimulation to the test subject; (R) & (R ± δx); where R is a reference value, and ± δx is a small increase/decrease in the stimulation value; and the subject is required to pick which stimulation is larger.

In Chapter 3, no statistical difference was found between the JND at the three tested reference angles, therefore only the middle reference level (15° rotation) was examined, as shown in Figure 5.2b. In the previous JND measurements presented in Chapter 3, measurements were taken once on the underside (i.e. ventral region) of the lower arm and once on the outside (i.e. ulnar region) of the lower arm, as shown in Figure 5.1a-b. For an effective comparison, measurements on the upper arm were therefore similarly undertaken at two stimulation sites of the upper arm; the medial proximal triceps region (referred to under), and lateral proximal triceps region (referred to as outer), shown in Figure 5.1c-d. Once again, differences of 0.5, 1, 1.5, 2, 3, 4 and 5 degrees rotation were tested. These differences formed stimulation pairs consisting of the reference value (R i.e.15°) and a value with a small offset to the stimulation (R ± δx).

The mechanical crank was rotated to the reference point for one second, followed by one second of no pressure/crank rotation. The mechanical crank was then rotated to the second position with a slight change in rotation from the first stimulation. The test subject was then required to say which stimulation felt stronger. Each of the pairs was used four times, twice with the larger rotation first (E.g. a 15-degree rotation followed by a 10-degree rotation), and twice with the larger rotation second (E.g. a ten degree rotation followed by a 15 degree rotation). By repeating the tests in a reverse order, it minimised any impact of potential bias that would have occurred if the subjects regularly guessed either the first or second value when they were unsure. This resulted in 56 stimulation pairs being used, with eight values for each stimulation difference tested (4 above and four below). These 56 pairs were presented to the subject in a randomised order once on the underside (i.e. medial proximal triceps region) of the upper arm and once on the outside (i.e. lateral proximal triceps region) of the upper arm.

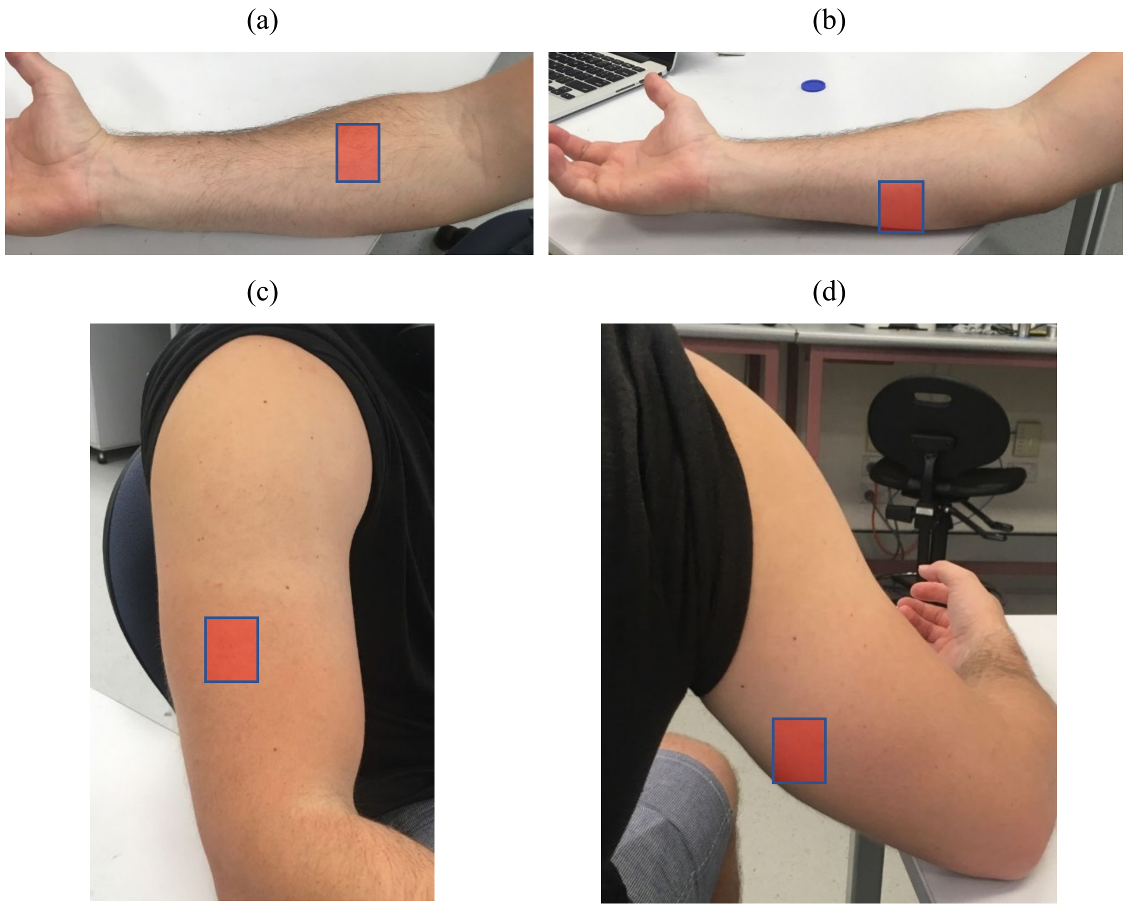


Figure 5.1 - Tested Stimulation Locations: (a) ventral region (under) of lower arm, (b) ulnar region (outer) of lower arm, (c) medial proximal triceps (under) of upper arm, (d) lateral proximal triceps (outer) of upper arm.

Figure 5.2 - Mechanical Crank Positions: a) 0° rotation, b) 15° rotation

A short 30-second break occurred every 30 trials, and a 2-minute break occurred between the two locations. These breaks were taken to minimise any desensitisation from stimulation, and to reduce the effect of cognitive overloading from the concentration required. Subjects were able to take any additional rest breaks as desired.

Psychometric functions were fitted with a logistic sigmoid using the psnigifit toolbox v4.0 for Matlab which implements the maximum-likelihood method as described in [126]. This curve was used to determine the JND threshold, taken as the midpoint between the lower and upper asymptotes. The JND thresholds were determined individually, with the mean, median and S.D of the results from the ten subjects calculated. Further, as performed in [128], due to the lower number of samples used a combined group threshold was determined based of a psychometric curve using all ten subjects' data in the one dataset.

10 subjects were tested, consisting of six males and four females with a mean age of 27.8 years ± 4.5 (S.D), with no physical or cognitive impairment. The order of the stimulation sites was alternated for each subject to minimise any impact of training and/or fatigue influencing the results.

5.2.2 Three Channel Recognition

Similar to the experiment presented in Chapter 3, to the number of stimulation channels was limited to 3; to represent the movement of the thumb, the pointer finger and the remaining three fingers. Similarly, the same six grip patterns used in Chapter 3 were used in this experiment shown in Figure 5.3.

Since previous literature has shown that training can improve recognition rate [25], a short training period was used to examine the intuitive nature of understanding the multichannel mechanotactile and electrotactile stimulation.

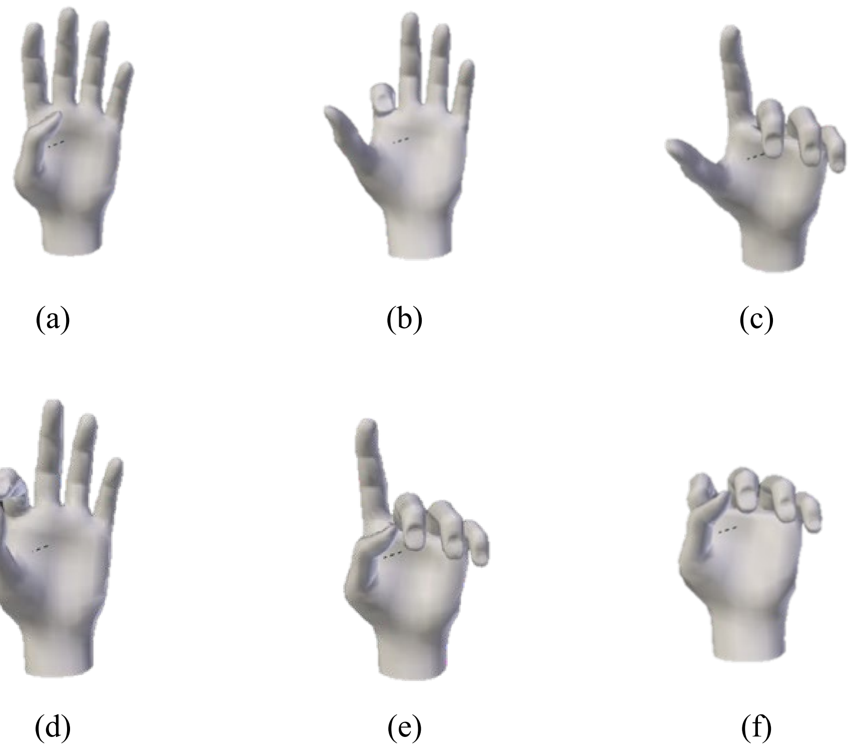


Figure 5.3 - Hand Grips: (a) Thumb Only, (b) Pointer only, (c) Pistol Grip, (d) Fine Grip, (e) Tool Grip and (f) Power Grip

a) Mechanotactile Stimulation

For the mechanotactile recognition experiment, three motors spaced at a distance of 90mm were used as shown in Figure 5.4. When attaching the mechanotactile device to the arm, the leash was attached 2/3 up the lower arm, and ½ way up the upper arm, as shown in Figure 5.5. The mechanotactile device was placed on the right arm of all subjects and was placed so that the middle motor could be in the lateral centre of the underneath (ventral region) side of the lower arm and back (medial proximal tricep region) of the upper arm, as shown in Figure 5.1.

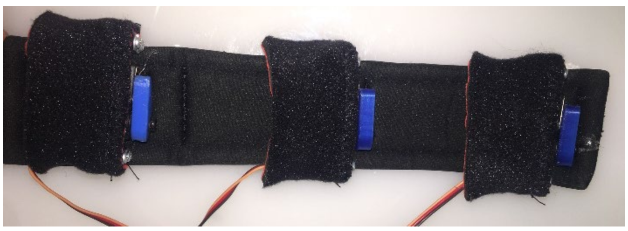


Figure 5.4 - Mechanical crank setup with three mechanical motors on leash cuff

In the experiments presented in Chapter 3, all 18 subjects were comfortable with receiving a stimulation of 24^0 rotation. Therefore, this was the starting level of rotation to represent our strong level, and 12^0 rotation representing our weak level. To ensure that the three motors were at the same perceived intensity, the weak and strong rotation levels were calibrated. In the calibration process, the designated weak level of rotation was applied to each of the three motors for one second, one at a time in succession, with a 1-second break in between each stimulation. The subject then recommended any increases or decreases required to the amount of rotation for the weak level for each motor to ensure that they felt the same intensity. This process was repeated until the subject felt all three motors at the same intensity. This calibration process was then repeated for the rotation representing the strong level of stimulation.

In the training period, each of the six motor combinations (representing the six different grip patterns), were demonstrated to the subject once at the rotation level representing the weak level of stimulation and once at the rotation level representing the strong level of stimulation. For each of these movements, they were told which motors would be active and their level of stimulation (strong or weak) immediately prior to the movement taking place. Within each movement, the crank applied the stimulation for a period of one second, before returning to the rest position (0° rotation). A 3-second break occurred between each movement. After all 12 possible movements were communicated, a 30-second break occurred prior to the commencement of the testing phase.

Each subject had two rounds of testing, performed once on the upper arm and once on the lower arm. For each stimulation site, each subject received a training period followed by the testing phase. The testing phase used a randomised order of the 12 movements (6 grips at two strengths) with two repetitions each, resulting in a total of 24 movements for each site. Each grip stimulation was held at the displacement for one second before returning to zero displacement. There was at least a 5-second pause between each movement for the subject to communicate the perceived movement. The subject was required to communicate what combination of sites the stimulation was applied (i.e. the grip) and whether it was applied at the stronger or weaker strength level. During the testing phase, subjects wore noise cancelling headphones with pink noise playing, to avoid the sound of any motor movement impacting their responses.

(a) (b)

Figure 5.5 - Mechanotactile Feedback System; (a) Placement on Lower Arm and (b) Placement on Upper Arm

10 subjects were tested, consisting of six males and four females with a mean age of 28.5 years ± 4.3 (SD), and with no physical or cognitive impairment. The order of the locations tested (upper and lower arm) was alternated for each subject to prevent the effect of additional training or fatigue influencing the results.

b) Electrotactile Stimulation

The electrotactile stimulation was delivered by a constant current neurostimulator (Inomed ISIS) and controlled through a LabView interface. Cathodic biphasic pulses were used with a pulse width set at 100 μs and a pulse frequency of 50Hz. The 35mm diameter concentric electrode detailed in Chapter 4 was used as the stimulating electrode. Similar to the mechanotactile setup, the electrodes were placed 2/3 up the lower arm and ½ way along the upper arm, as shown in Figure 5.6. The middle electrode was placed on the lateral centre of the underneath (ventral region) side of the lower arm and back (medial proximal tricep region) of the upper arm. The remaining two electrodes were placed midway up the sides of the lower arm/upper arm to ensure equal spacing and they were at the same lateral position.

In the calibration process, since perceived magnitude is dependent upon the minimal Detectable Threshold (DT) [99], first the DT of each stimulation site is determined for each subject. For each electrode, the current was sent continuously, beginning at 0.5mA and slowly increasing by 0.5mA until the subject was able to perceive the electrical stimulation. It was then slowly decreased by 0.5mA to find the point the sensation disappeared. The lowest detectable stimulation value was then set as the DT. This process was repeated for the other two sites.

In Chapter 4, for the ten subjects tested, there was an average of 5.59mA difference between the DT and the Pain Threshold (PT). Therefore, to ensure that the PT region was avoided whilst obtaining two distinct but recognisable magnitudes of stimulation, the initial weak level was set to 1.5mA above the DT and the initial strong level to 4mA above the DT.

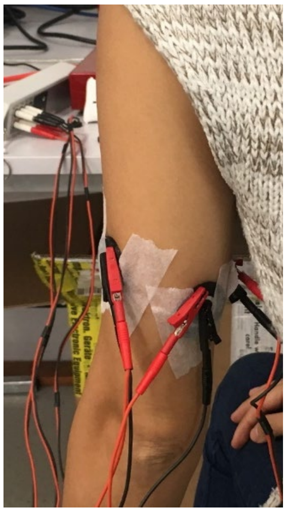
To ensure that all the three sites perceived the same level of intensity, the designated weak level of current was applied to each of the three electrodes placed at the three stimulation sites for one second, one at a time in succession, with a 1-second break in between each stimulation. The subject then recommended any increases or decreases required to the level of current for the weak level for each electrode to ensure they felt the same intensity. This process was repeated until the subject felt all three sites at the same intensity. This calibration process was then repeated for the current level representing the strong level of stimulation.

In the training period, each of the six combinations of electrodes being stimulated (representing the six different grip patterns), were demonstrated to the subject once at the current level representing the weak level of stimulation and once at the current level representing the strong level of stimulation. For each of these stimulations, they were told which electrodes would be active and their level of stimulation (strong or weak) immediately prior to the stimulation taking place. In each stimulation, the current was applied for a period of one second, before switching off the electrical stimulation. A 3-second break occurred between each stimulation. After all 12 possible stimulations were communicated, a 30-second break occurred prior to the commencement of the testing phase.

Each subject had two rounds of testing, performed once on the upper arm and once on the lower arm. The testing phase round used a randomised order of the 12 stimulations (six grips at two strengths) with two repetitions each, resulting in a total of 24 stimulations. Each grip stimulation was applied for one second before being turned off. There was at least a 5-second pause between each movement for the subject to communicate the perceived stimulation. The subject was required to communicate what combination of sites the stimulation was applied (i.e. the grip) and whether it was applied at the stronger or weaker strength level.

Ten subjects were tested, consisting of six males and four females with a mean age of 28.1 years ± 4.0 (SD), and with no physical or cognitive impairment. The order of the locations tested (upper and lower arm) was alternated for each subject to prevent the effect of additional training or fatigue influencing the results.

(a)



(b)

Figure 5.6 - Placement of Electrodes: (a) Lower and (b) Upper Arm

5.3 Results

5.3.1 Sensitivity

A summary of the JND results attained for each of the ten subjects is shown in Figure 5.7 and Table 5.1. However, due to the small amount of tested values for each reference value per individual subject, in addition to the results for each individual, all results are pooled together to form a combined group psychometric curve for both the underneath and outside locations, as shown in Figure 5.8.

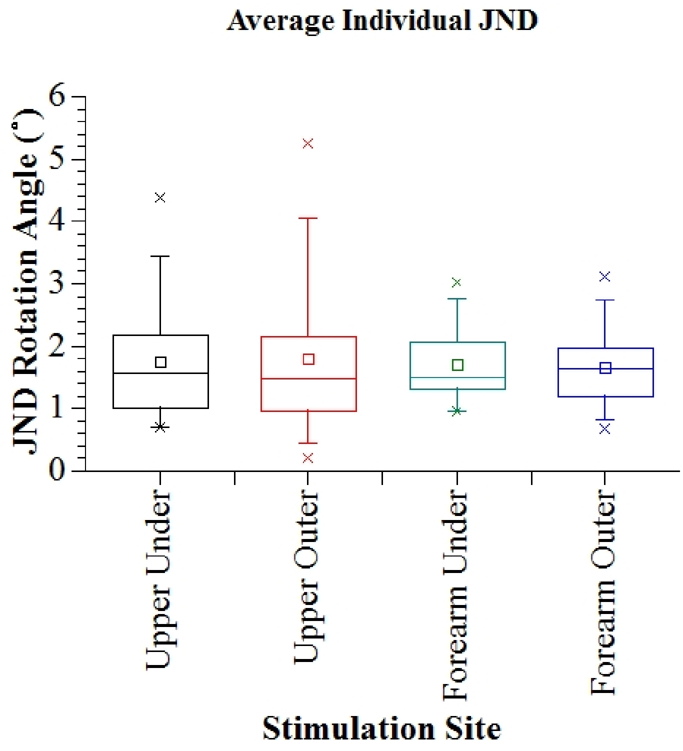
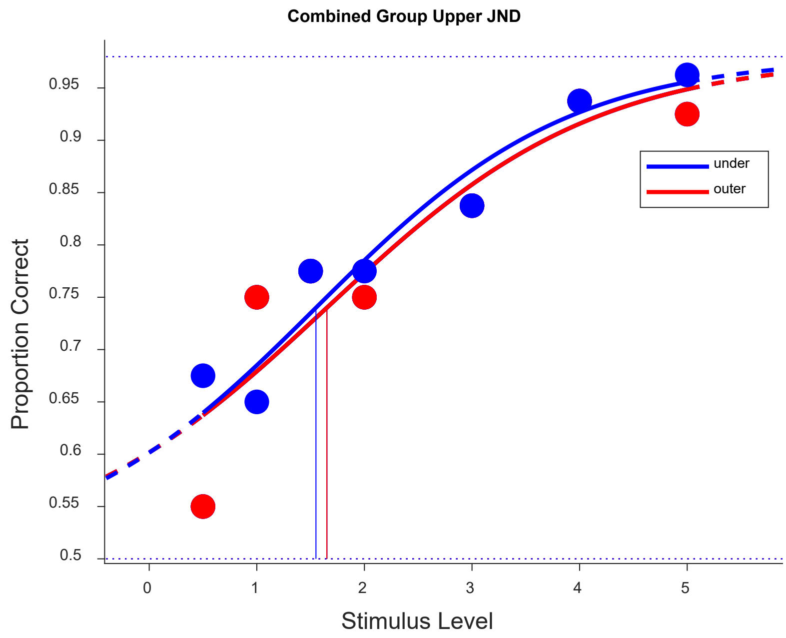


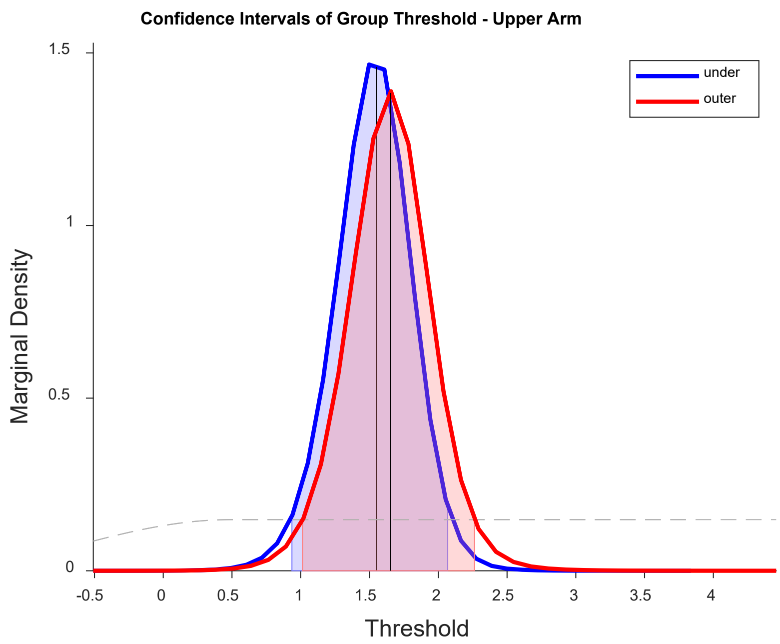
Figure 5.7 - Average Individual Mechanotactile JND Rotation Angle

Similarly for the lower arm, the individual mean, S.D. and median are displayed in Table 5.1. In addition, it also shows the threshold calculated off the combined group data. A repeated t-test showed that there was no statistically significant difference between the individual JND obtained from the two stimulation sites on the upper arm [$t(9) = 0.228$, $p=0.825$]. This also corresponds with the psychometric curves and confidence thresholds from the combined group data, shown in Figure 5.8a, and Figure 5.8b, respectively, which are very similar with a large amount of overlap with the confidence interval regions.

An ANOVA was applied to determine if there was any statistically significant difference between the two stimulation sites on the lower arm and two stimulation sites on the upper arm using the 15-degree stimulus reference. There was again no statistically significant difference between the mean JND of the four tested stimulation sites $F(3,36) =0.71$, $p = 0.975$.



(a)



(b)

Figure 5.8 - Upper Arm JND from Combined Data. a) Psychometric Curve and b) Confidence Intervals for JND Threshold

Table 5.1 - Determined JND Values. Compbined Group JND and Confidence Interval was calculated based of fitting a psychometric curve to the all the data combined from ten subjects

Location	Reference Angle (°)	Mean Individual JND (°)	SD Individual JND (°)	Median Individual JND (°)	Combined Group JND (°)	Combined Group JND Confidence Interval (°)
Lower Arm Under	15	1.71	0.66	1.49	1.53	0.97-1.99
Lower Arm Outer	15	1.67	0.70	1.65	1.59	1.02-2.11
Upper Arm Under	15	1.84	1.07	1.83	1.66	1.00-2.29
Upper Arm Outer	15	1.81	1.40	1.48	1.65	1.01-2.26

5.3.2 Recognition of three Channels of Stimulation

a) Mechanotactile Stimulation

Figure 5.9 presents the recognition rates for the two different locations. It is broken down into three measurements: Grip Only – measuring the accuracy in identifying which motors were active; Strength Only – measuring the identifying whether it was at the strong or weak intensity; Grip and Strength – measuring the accuracy of indenting which motors were active and the strength correctly. Further confusion matrices of the grip recognition for the lower arm and upper arm locations are shown in Figure 5.10a and Figure 5.10b, respectively.

Repeated measured t-tests were performed and found that there was no statistically significant difference for either grip only [t(9) = 0.0.497, p = 0.631], strength only [t(9) = -1.695, p = 0.124] or both grip and strength [t(9) = -0.422, p = 0.683].

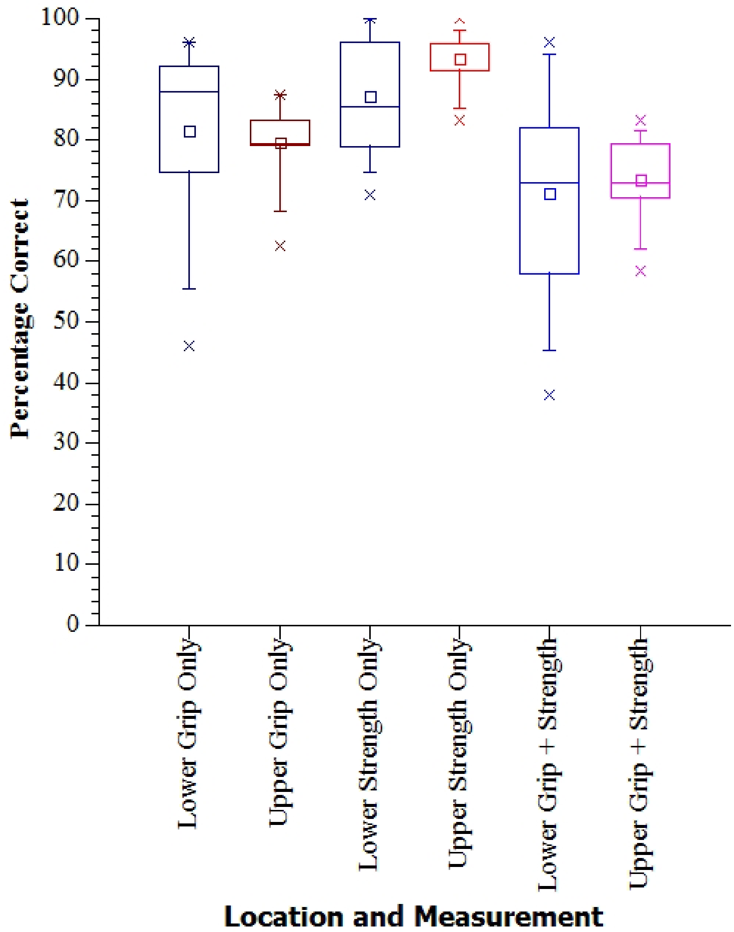
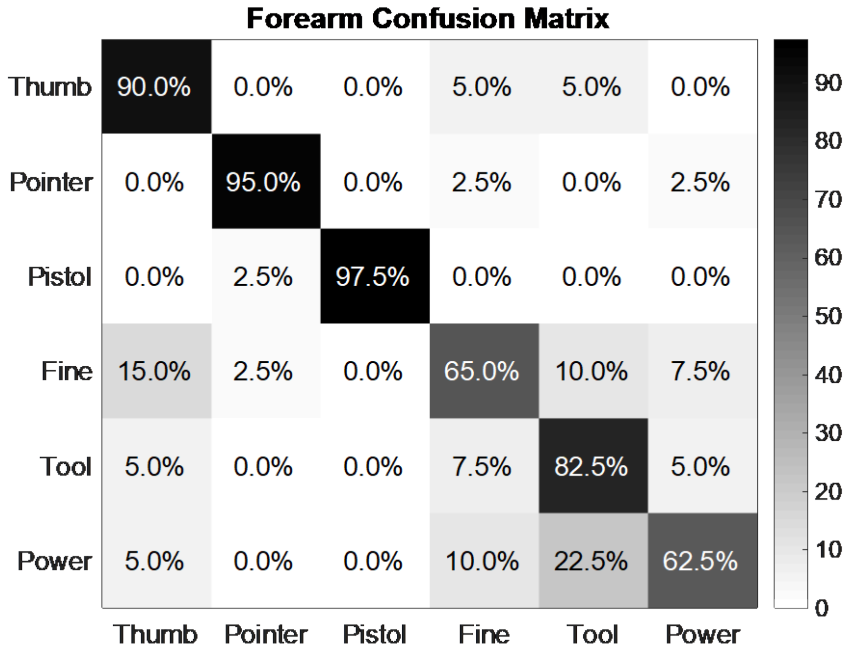
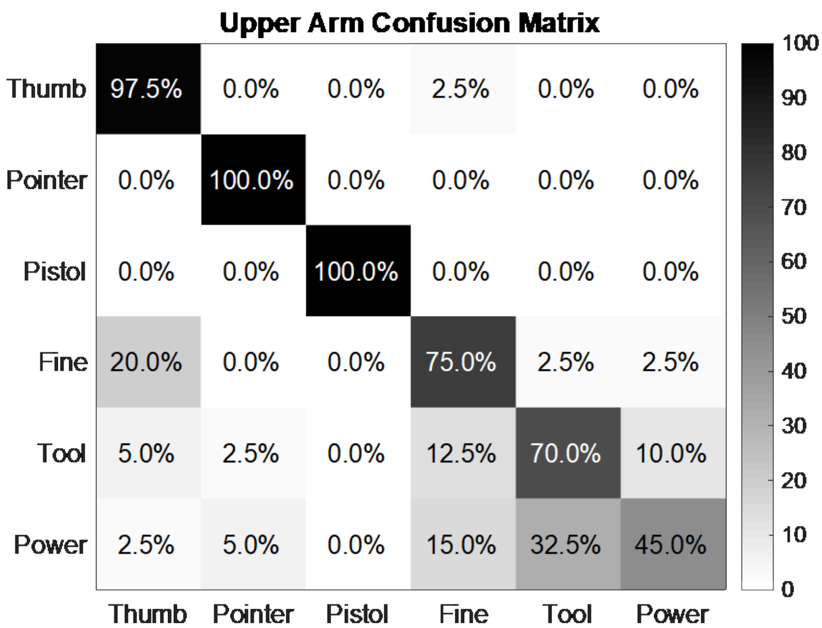


Figure 5.9 - Average Mechanotactile Recognition Rates on Upper and Lower Arm

The confusion matrices in Figure 5.10 show that for both locations, the recognition rate for one motor (thumb, pointer and pistol grips) achieved a high level of accuracy. In the upper arm, there is a large amount of confusion when attempting to correctly identify the power grip. Although these results are lower than the results in Chapter 3, it is anticipated that they can be improved with training [25]. In addition, since the main purpose of this experiment was to compare the recognition rate for the two locations, the distance between the three motor locations were kept consistent. However, optimisation of the place for both the location and the individual arm size may help improve recognition further.



(a)



(b)

Figure 5.10 - Confusion Matrices of Mechanotactile Grip Recognition. The Rows represent the applied grip pattern and the columns represent the percieved grip pattern.

b) Electrotactile Stimulation

Figure 5.11 presents the recognition rates for two different locations. It is broken down into three measurements: Grip Only – measuring the accuracy in identifying which motors were active; Strength Only – measuring the identifying whether it was at the strong or weak intensity; Grip and Strength – measuring the accuracy of indenting which motors were active and the strength correctly. Further confusion matrices of the grip recognition for the lower arm and upper arm location are shown in Figure 5.12a, and Figure 5.12b, respectively.

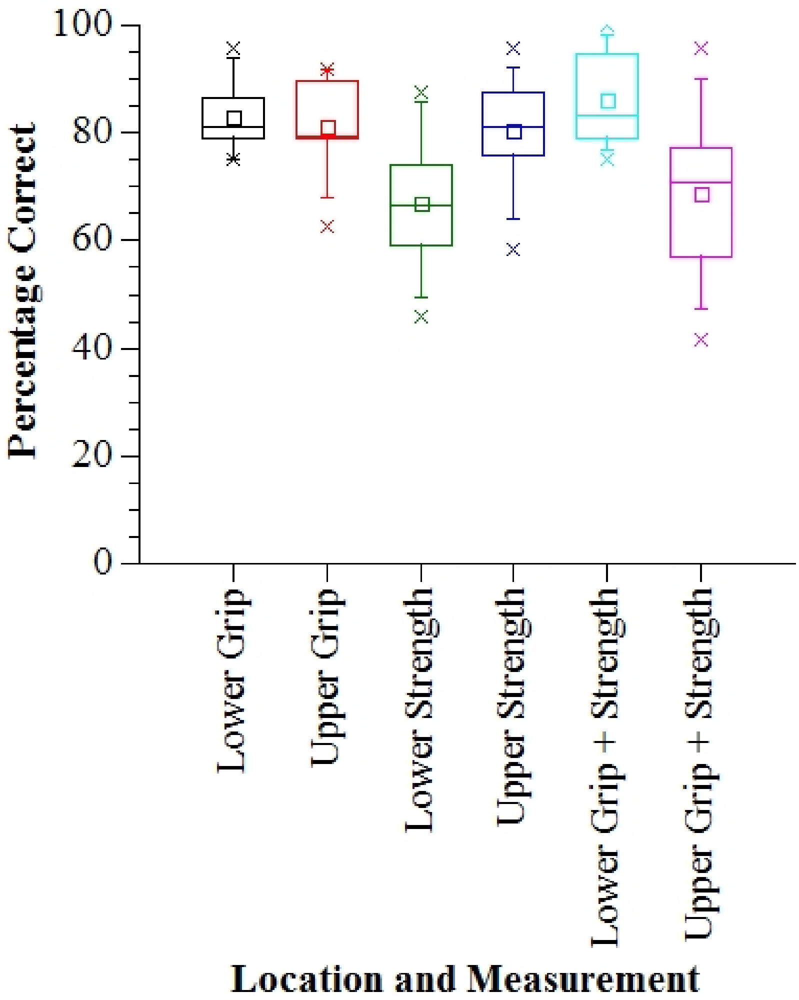
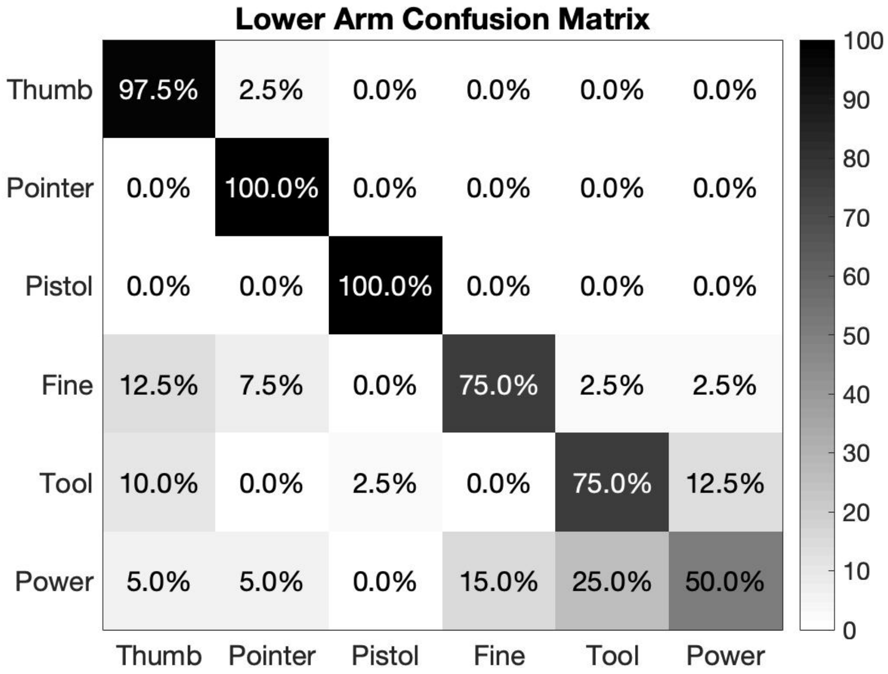


Figure 5.11 - Average Electrotactile Recognition Rates on Upper and Lower Arm

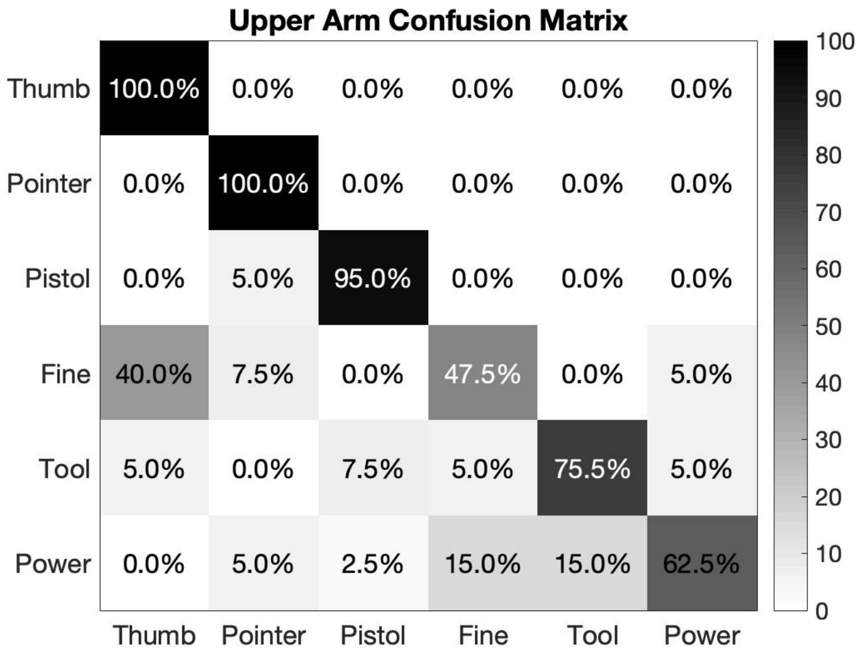
Repeated measured t-tests were performed and found that there was no statistically significant difference for either grip only [$t(9) = 0.786$, $p=0.452$], strength only [$t(9) = -1.650$, $p=0.133$] or both grip and strength [$t(9) = -0.371$, $p=0.719$].

The confusion matrices in Figure 5.12 show that for both locations, the recognition rate for one motor (thumb, pointer and pistol grips) achieved a high level of accuracy. In both the lower and upper arms, there is a large amount of confusion when attempting to correctly identify the power grip. In addition, the fine grip appeared difficult to correctly identify in the upper arm. Again, it is anticipated that these results can be improved with training [25].To be consistent with the mechanotactile stimulation, the three concentric electrodes were placed in a straight line. An improvement in recognition rates may be seen, however, if the electrodes are offset from each other creating a large spatial distance between them. Since there was no difference in accuracy between the lower and upper arms, placing the electrodes on the upper arm provides more surface area to spread out the electrodes. Further experimentation is therefore required to determine the optimum placement of electrodes to maximise recognition rate. In addition, using smaller electrodes may result in an increased accuracy, but as discussed in Chapter 4, may affect comfort of stimulation. Similarly, further experimentation is required to examine the impact of the size of the electrodes on these two factors to determine the optimum size.

An ANOVA comparison was run between the four measurements from the two stimulation techniques on the two different locations. There was no statistically significant difference between the four measurements for the grip recognition [$F(3,36) = 0.175$, $p=0.913$] or the grip and strength recognition [$F(3,36)=0.378$, $p=0.769$]. However, there was a statistically significant difference detected in the four measurements of the strength recognition performance [$F(3,36=3.400$, $p=0.028$]. Post hoc tests using a Tukey Honestly Significant Difference (HSD) correction revealed only one statistically significant difference, a 12.1% increase in recognition rate of the mechanotactile stimulation strength when applied on the upper arm compared to the lower arm location ($p=0.015$).



(a)



(b)

Figure 5.12 - Confusion Matrices of Electrotactile Grip Recognition

5.4 Upper Arm Perceived Intensity

Since was no statistical difference between the recognition rate and sensitivity between the lower and upper arms, the upper arm will be the site of stimulation used in our prosthetic users in Chapter 6, as this removes the need to modify their socket. The relationship between applied stimulation on the back of the upper arm (triceps region) will therefore be examined, from both mechanotactile and electrotactile stimulation, using the Method two for the magnitude estimation presented in Chapter 3.

Through a graphical interface as detailed in Chapter 3, subjects were able to choose when to receive the standard stimulation and the comparison stimulation. The standard stimulation represented an intensity of 10. The subject was then asked to assign a number to rate the feeling of intensity, using the standard stimulation as a reference. Again, if the first stimulation was rated an intensity value of four, and the second stimulation felt twice as strong, they were instructed to assign it a value of 8. Subjects were once again encouraged to use decimals/fractions as required. They were able to go back and forth and receive either of the two stimulations as required. Once they determined the intensity, they entered into the value text box and press next round. The round number would then increase, and the next stimulation value will be loaded.

The comparison values were chosen to represent the ratios of $[\frac{1}{3}, \frac{1}{2}, \frac{2}{3}, \frac{3}{4}, 1\frac{1}{2}, 1\frac{1}{2}, 1\frac{2}{3}, and\ 2]$ respectively from the standard stimulation. Each test session consisted of each of these values tested five times in a random order, resulting in of 40 stimulation pairs per test. This process was then repeated for the next stimulation value, with the "intensity 10 standard" stimulation staying the same throughout the whole experiment.

Subjects were encouraged to take a 30 second break every ten rounds to reduce any possible impact of desensitisation and concentration fatigue. They then received a 5-minute break before undergoing the test on the other stimulation (mechanotactile or electrotactile) technique.

Testing was undertaken by ten able-bodied subjects (four females, six males) with a mean age of 27.6 years ± 4.3 (S.D). Subjects wore noise-cancelling headphones with pink noise during the mechanotactile stimulation to prevent any impact from the motor's noise. The order mechanotactile stimulation and electrotactile stimuation was alternated for each subject to minimise any impacts on fatigue and concentration fatigue.

5.4.1 Mechanotactile

The standard "10" intensity was a rotation of 12 degrees and the stimulations of [4, 6, 8, 9, 16, 18, 20, 24] were tested against the standard to represent the ratios of $[\frac{1}{3}, \frac{1}{2}, \frac{2}{3}, \frac{3}{4}, 1\frac{1}{2}, 1\frac{1}{2}, 1\frac{2}{3}, and\ 2]$ respectively.

The normalised intensity results for all individuals were pooled together and are displayed in Figure 5.13. The linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the underneath location with an R^2 value of 0.852. Individually each subject achieved an R^2 value of 0.896 ± 0.047, showing minimal reduction in the relationship suggesting the relationship is not subject specific, similar to the lower arm location. A mixed model linear analysis was performed to determine the coefficients and take into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 0.895 ± 0.053 (S.E) [t(8.381)= 16.970, p<0.001].

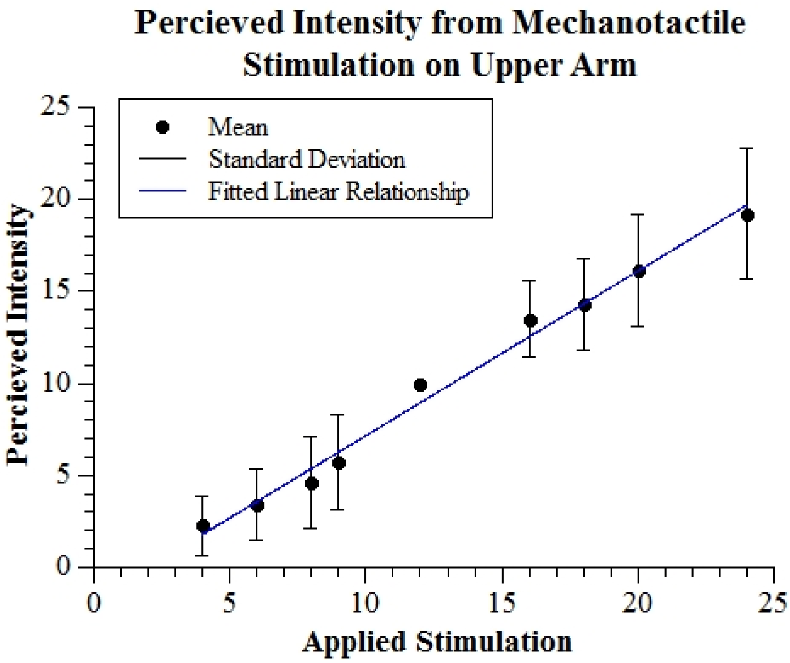


Figure 5.13 - Relative change in perceived stimulation from mechanotactile stimulation on upper arm

5.4.2 Electrotactile

The electrotactile stimulation was delivered by a constant current neurostimulator (Inomed ISIS) and controlled through a LabView interface. Cathodic biphasic pulses were used with a pulse width set at 100 µs and a pulse frequency of 50Hz. The 35mm diameter concentric electrode detailed in Chapter 4, was used as the stimulating electrode.

Since perceived intensity is related to the current applied above a minimum threshold [12], each test subject firstly determined their minimum detectable current. The current was increased with increments of 0.1mA until the subject recognised the stimulation. It was then decreased by 0.1mA until the sensation disappeared, then increased by 0.1mA until it was detected again. This second detected level was recorded as the minimum detectable current. The lowest current level used in the process was set at 0.4mA above the minimum detectable level, to ensure it was easily recognised. The current was then slowly increased to ensure that the subject was able to detect the highest current level, which was 4.8mA above the minimum detectable current.

The standard "10" intensity was a current of 2.4mA above the DT, and the stimulations of [0.8, 1.2, 1.6, 1.8, 3.2, 3.6, 4, 4.8]mA above the DT were tested against the standard to represent the ratios of $[\frac{1}{3}, \frac{1}{2}, \frac{2}{3}, \frac{3}{4}, 1\frac{1}{2}, 1\frac{1}{2}, 1\frac{2}{3}, and\ 2]$, respectively.

The normalised intensity results for all individuals were pooled together and are displayed in Figure 5.14. The linear regression analysis revealed that there is a strong positive association between the rotation angle and the normalised mean intensity at the underneath location with an R^2 value of 0.717. Individually each subject achieved an R^2 value of 0.793 ± 0.081, again showing minimal reduction in the relationship suggesting the relationship is not subject specific. A mixed model linear analysis was performed to determine the coefficients and take into account the repeated measurements on multiple subjects. In the mixed model analysis, an average of five values given for the intensity of the repeated stimulation values was used for each subject. The output of the mixed model determined a slope of 3.227 ± 0.30 (S.E) [$t(8.798)= 10.937$, $p<0.001$].

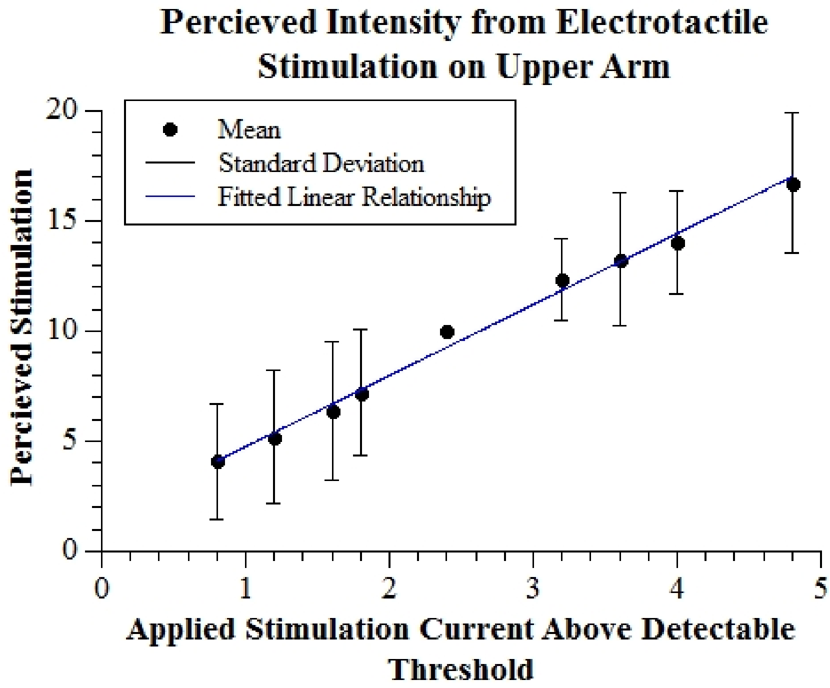


Figure 5.14 - Relative change in perceived stimulation from electrotactile stimulation on upper arm

5.4.3 Comparing Upper Mechanotactile and Upper Electrotactile Stimulation

A repeated measures t-test was performed on the individual R^2 values and the individual relative standard errors for the electrotactile stimulation and mechanotactile stimulation. A repeated measures t-test showed that the there was an increase of 0.104 of the R^2 value from the electrotactile stimulation test to the mechanotactile stimulation test which is statistically significant [$t(9)=3.197$, $p=0.012$]. However, a repeated measures t-test also demonstrated that there was no statistically significant difference between the relative standard errors between the mechanotactile stimulation and electrotactile stimulation applied on the upper arm [$t(9)=0.752$, $p=0.471$].

The experiments undertaken by Hartman et al. [158] also demonstrated the difficulty for individuals to consistently and accurately identify electrotactile intensity levels from changing the current level of the pulses, but requires training to learn to interpret the stimulation level correctly. Further experimentation is required to see if training can continue to improve the consistency of recognition of the electrotactile and mechanotactile stimulation to improve the resulting model matching the applied stimulation to the perceived intensity. One contributing factor to the difficulties in the recognition of electrotactile stimulation intensity may be as a result of the ability to jump to higher levels instantaneously. For mechanotactile stimulation, when you apply a high level of stimulation, it must increase to the desired stimulation level by moving through all of the intensity levels. In electrotactile stimulation, however, there is the ability to apply the desired level of current from the start of the stimulation. Both the increasing nature of the mechanotactile stimulation and the extra time required to reach a higher stimulation, may improve the subject's ability to consistently recognise stimulation levels. Further experimentation, however, is required to determine the impact of these two factors, and if simulating them in electrotactile stimulation improves consistency of electrotactile intensity perception.

5.5 Summary

This chapter verified that previously obtained results of sensitivity [154] for the upper and lower arms using mechanotactile stimulation, and the recognition rates of electrotactile and mechanotactile stimulations for both the upper and lower arm regions were measured and compared. The recognition rate of up to three channels of mechanotactile stimulation, and sensitivity to small stimulation changes of the upper and lower arms were not statistically different to each other. This allows either an additional or alternative site to be used instead of the lower arm region without any statistically significant loss in performance. This also allows a pathway for undertaking experimentation using sensory feedback with existing myoelectric prostheses without requiring modification to their existing prosthetic socket, as they typically encase the whole lower arm region.

The results obtained from ten able-bodied subjects show a high level of discrimination in the ranges of 1.53-1.65 degrees for mechanotactile stimulation at the ventral and ulnar region of the lower arm, as well as the medial and lateral proximal tricep regions of the upper arm. These JND values do not statistically differ between any of the four tested locations - two stimulation sites at the two regions (upper and lower arms) examined. This testing, however, was only performed on able-bodied subjects. In future work, this testing should be repeated upon people with upper limb difference to determine how their sensitivities compare of their residual limb to both to their own upper arms, and to these able-bodied results.

The average performance of 79.6%-82.9% was able to be achieved across the ten subjects for the four grip measurements recorded by the two stimulation techniques (mechanotactile and electrotactile) at the two locations (upper and lower arm) feedback. This was achieved with minimal training However, this can be improved with further training.

We have demonstrated that there is a linear relationship between applied stimulation and perceived intensity for both mechanotactile and electrotactile stimulation on the triceps region of the upper arm.

Chapter 6

Effect of Sensory Feedback on Controlling Grasping Force for Myoelectric Transradial Prosthetic Hand Users

6.1 Introduction

Out of the possible sensations, gripping force feedback is currently rated by users as the highest priority to incorporate into next-generation hand prosthesis [3, 37]. A few prior studies have examined the impact of sensory feedback on the ability to control gripping force [44, 45, 49, 78, 159] with existing prosthetic hands. Of these, three only tested on able bodied subjects controlling a myoelectric hand [44, 45, 159] and another study tested amputees not using their normal prosthesis [160]. Only two previous studies examined the impact of providing grasping force sensory information to myoelectric prosthetic users on controlling their grip with their existing prostheses, but focussed on vibrational feedback [49] and augmented reality [78]. This chapter examines the use of mechanotactile and electrotactile stimulation to provide non-invasive sensory feedback for people with upper limb difference in controlling grasping force with their existing myoelectric prosthetic hand. Since there is no statistical difference between accuracy and recognition of the upper arm and lower arm, as demonstrated in Chapter 5, the sensory feedback is provided to the back of the upper arm where the triceps brachii muscle is located. This will enable us to test the sensory feedback technique on existing prosthetic hand users without requiring any modification to their existing prosthesis socket.

In this chapter, firstly the device constructed to measure and record the grasping force from prosthetic hands is outlined in Section 6.2. Section 6.3 presents the experiments undertaken with five myoelectric prosthetic hand users to determine their ability to receive the sensory feedback on their upper arm in order to adjust the grasping force they are applying to a test object.

6.2 Force Measurement Device

6.2.1 Device Construction

The virtual egg test is one technique that has been used to examine the impact of sensory feedback on the ability to control the "pinch grip" of fragile objects [49, 78, 159, 161-165]. These tests either use mechanical or electronic means to determine if the user applies in excess of the predefined "safe" force to lift a test object without "breaking" it. However, this approach typically relies on either embedded sensors into the hand or Force Sensitive Resistors (FSR) placed on the inner surface of the top of the prosthetic fingers, since sensor surface curvature and compliance can impact any calibration performed on FSR's [166], it is impractical to apply them to various individual prosthetic hands for testing purposes. Alternatively, this test often involves users using a new hand they are not accustomed to. As an alternative, this thesis proposes a sensorised object in the form of a cube, or sometimes called "virtual egg", that can not only record and measure the force applied during a pinching grip but also can wirelessly transmit this force data to drive sensory feedback mechanisms. This allows the virtual egg test to be employed to evaluate the sensory feedback technique for transradial prosthetic hand users who generally has had successful integration and adaption with their existing prosthesis.

The force measurement cube is a hollow-shell design with an outer length of 44mm, and uniform wall thickness of 4mm. A cross-sectional cut was made at 40mm normal to the base to create the base (major part) and lid (minor part), as shown in Figure 6.1. 18 4.2mm x 2.2mm x 1.4mm holes are bored into the exposed cross-section of both the base and the lid to house N52 magnets (Neodymium Block Magnet, Frenergy Magnets) to allow a seamless assembly of the force measurement cube.

The cube itself was fabricated using additive manufacturing techniques (Mark Two Desktop Printer, Markforged) using continuous fibre printing Fiberglass. The pucks for each of the sensors were fabricated using PLA via Fused Filament Fabrication (Creator Pro, Flashforge).

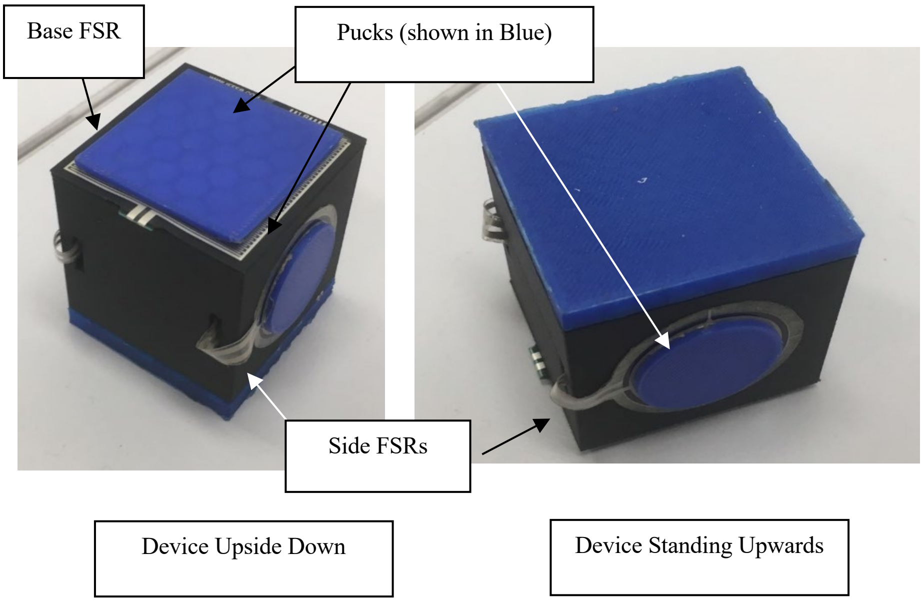


Figure 6.1 - The force measurement cube.

The two side sensors use a TekScan 401 FSR as they offer improved performance [167] in resolution, and repeatability against comparable off-the-shelf sensors by Interlink Electronics. Since we are interested in instantaneous measurements, the measurement issues such as baseline drift over time that exist with FSR's are not an issue. The base sensor is a square Flexiforce 401, sized to cover the whole face. This sensor is used to examine the lift off timing in relation to the "squeeze". This is especially important for the sensory feedback which has been shown to improve the coordination between application of gripping force and lifting force [49]. To keep the contact force area consistent, 1.5mm thick pucks were placed on each FSR, as shown in Figure 6.1. A 24mm diameter disc was used on the side FSRs and a 37mm square was placed on the base sensor.

A Beetle BLE microcontroller was used due to its small footprint with built-in low energy Bluetooth wireless transmission. Each FSR was connected to their own instrumental operational amplifier (AD623AN), as shown in Figure 6.2. The gain resistors to ensure the target force was spread across the possible voltage ratings without saturation. For the side sensors, the maximum target force is 60N which is in the high region of maximum grip forces applied by healthy hands [11]. The base sensor was used to measure the lift-off timing, so the target force was limited to a minimum value of 2.5N.

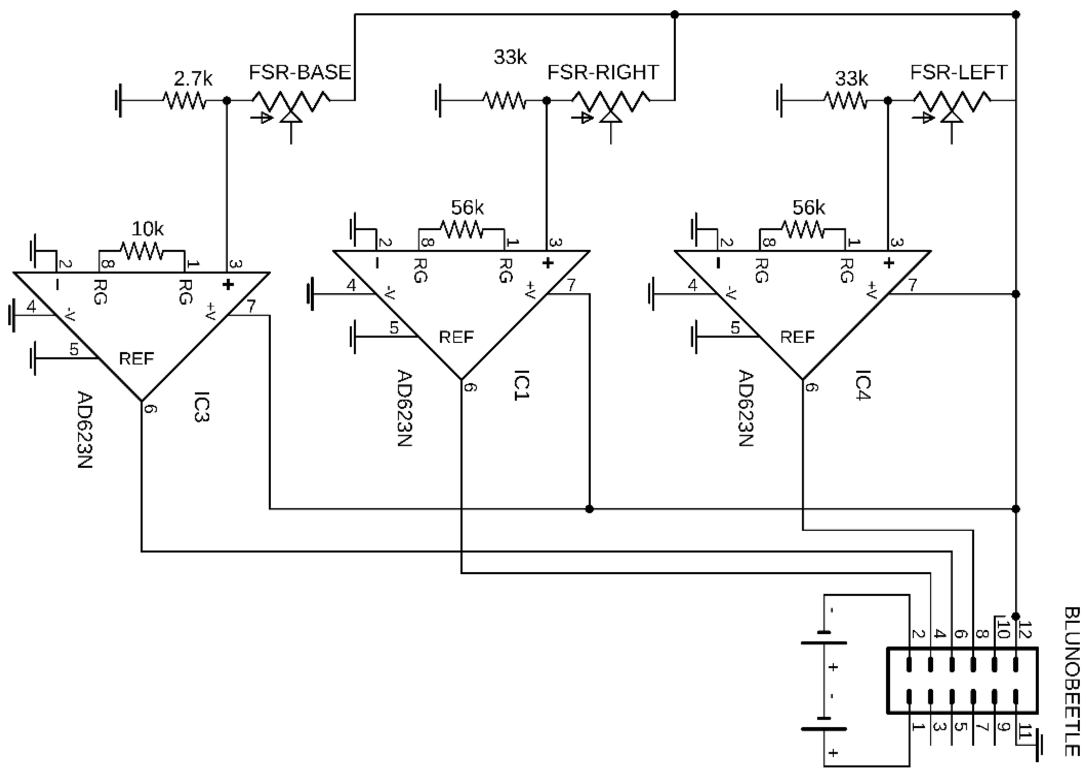
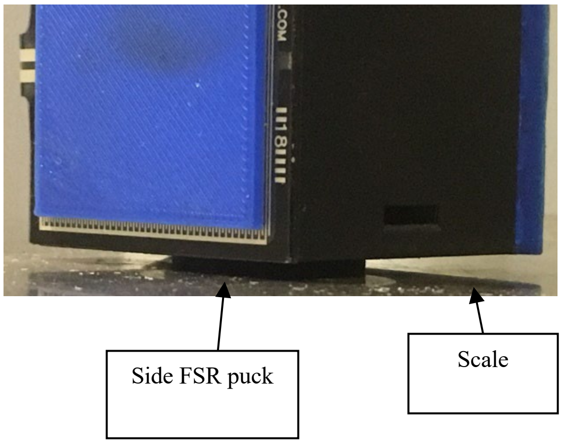


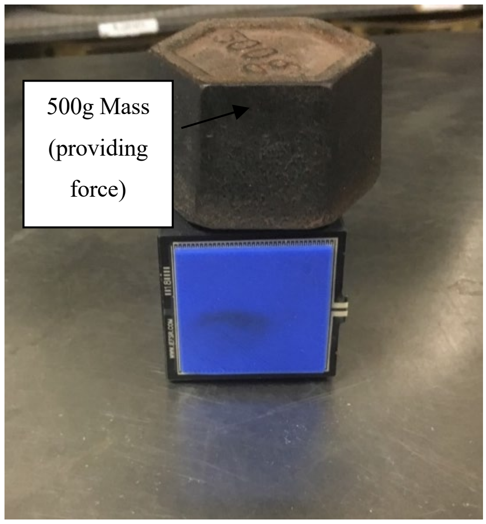
Figure 6.2 - The circuit diagram for the force measurement cube.

6.2.2 Calibration

A second-order polynomial was used for each sensor to model the relationship between the voltage measured at the analog pin, and the applied force. Since there is variability between individual sensors, this was repeated for each sensor separately. Weights were placed on top of the cube, as shown in Figure 6.3b, with the sensor being calibrated on the bottom against the scales (A&D GP-12K, 12kg capacity, 0,1g resolution). Since the cube has a mass of 74g, the sensor base was calibrated using these masses in grams (0, 74, 124, 174, 224, 274). This (274gram mass) was the limit prior to saturating the sensor. The side sensors were calibrated using theses masses in grams (0, 200, 500, 1000, 1500…up to a maximum of 6kg in 500g increments). This is approximately 60N [168], which is in the high region of maximum grip forces applied by healthy hands. This process was performed twice; ramping the weight up and down in order to minimize the impact of any hysteresis. The average of the four voltage values was used in the calibration curve calculation, and the standard deviation across these four voltage measurements are shown in Figure 6.4. Although FSR's have an approximately linear relationship with conductance, it is typical for strain sensors to not conform to a precise linear scale over a wide range of force [169] and as a result a lower residual error was achieved with a 2nd order polynomial compared to a linear relationship.

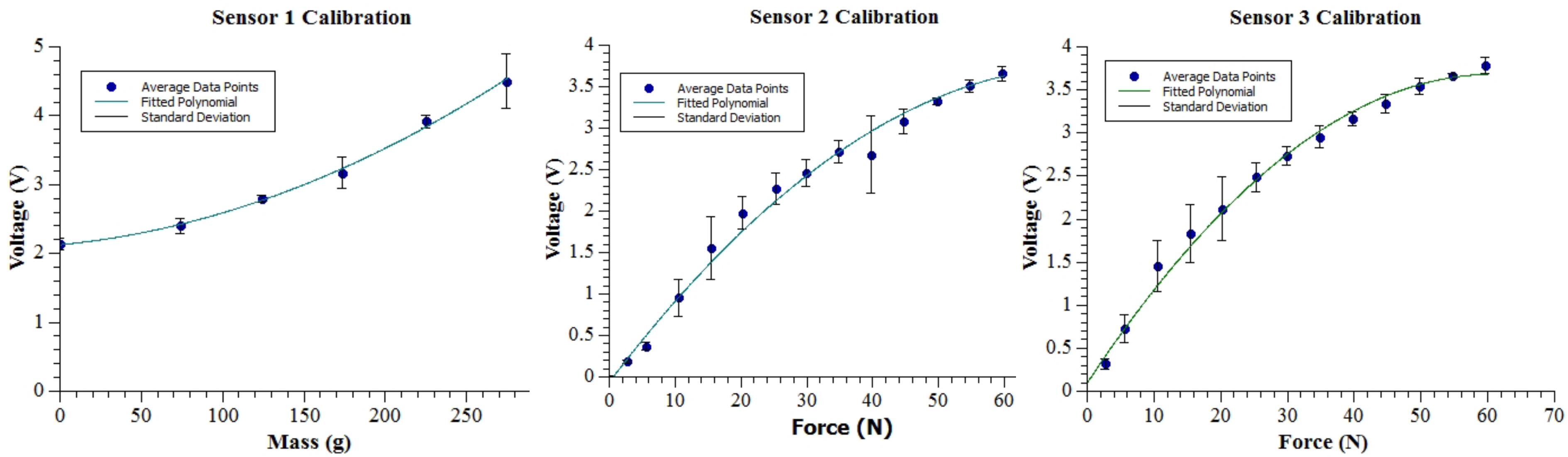


(a)



(b)

Figure 6.3 - The cube calibration setup: device placed on scaled on side sensor (a) close up showing contact only being made through contact puck, (b) Device on scales with 500g additional mass on top to provide calibrating force



(a) (b) (c)

Figure 6.4 - Calibration Curves: (a) Base FSR for Weight Measurement, (b) and (c) Side FSR for Squeezing Force

6.2.3 Device Demonstration

To demonstrate the recordings of the force measurement cube, a small mass 200g and another mass of 500g was placed on the cube as shown in Figure 6.5a. An abled-bodied subject was asked to lift the cube using a pinch grip, as shown in Figure 6.5b, place it on the other side of a fence-like obstacle, then lift it back over and place it in the original position. The subject was instructed to use as little strength as possible without dropping the object. To overcome some of inherent inaccuracies associated with FSR sensors, the average of the two side sensors measuring the grasping force was recorded as the resultant grip force. The average picking force for lifting the cube with a 200g mass and 500g mass are shown in Figure 6.6a and Figure 6.6b, respectively.

As shown in Figure 6.6, there is a clear increase in the pinch force used by the subject for lifting the object with a 500g mass compared to the 200g mass placed on top. As previously suggested [49], there is a correlation between the lifting force and the grasping force - as represented by the overlap in Figure 6 of the gipping force increases whilst the force recorded by the base sensor decreases.

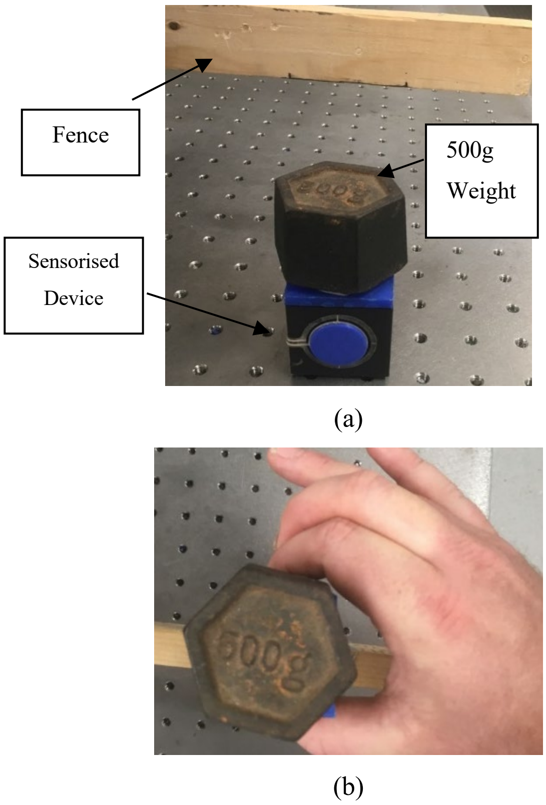
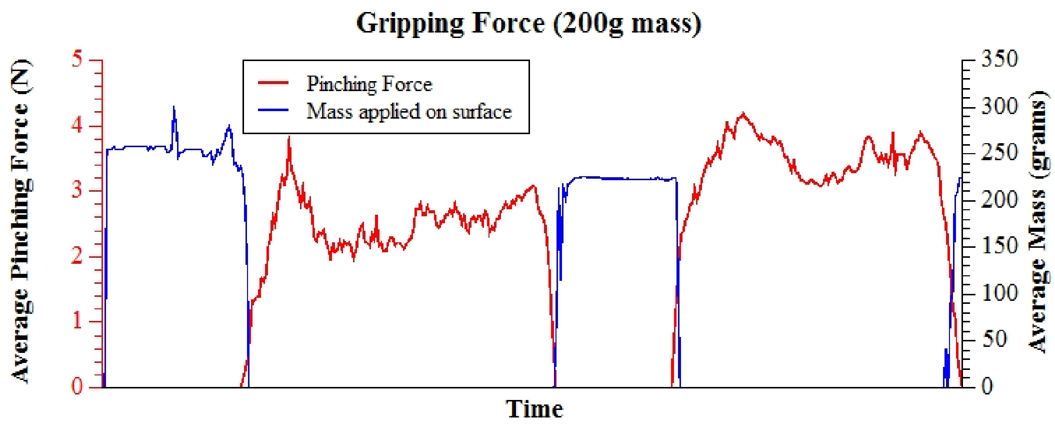
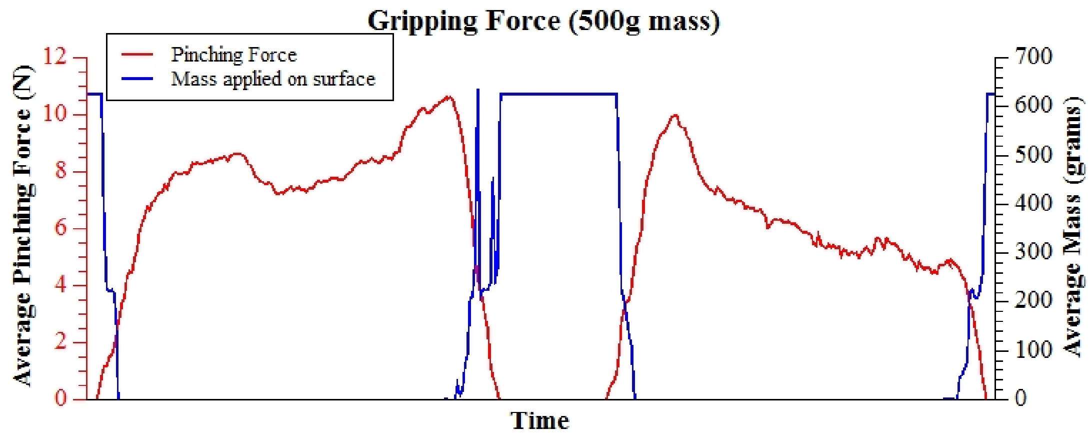


Figure 6.5 - Force Demonstration Setup: (a) Resting position (side view); (b) at the maximum height of the lift over the fence (top view)



(a)



(b)

Figure 6.6 - Demonstration Grasping Force Recordings: The average pinching force shown in red associated with the left axis, and the base sensor weight force shown in black correlating with the right axis.

The prototype used in this chapter relies on commercially available FSRs. These, however, are only supplied as set sizes and shapes, which limit the potential design. Even after improving the repeatability of these sensors by adjusting the contact area and compliance, they do not produce perfectly repeatable results. The impact of this was minimised through averaging the results from the two side force sensors. Although the current level of accuracy is sufficient to demonstrate the effectiveness of tactile sensory feedback techniques, custom made force sensors would provide better control over the size and shape of the force measurement device (e.g., the cube), and possibly produce more accurate force measurements.

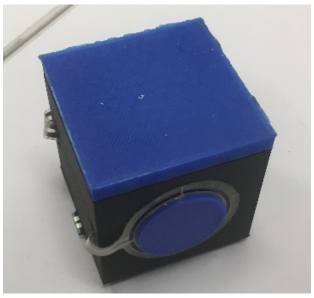
6.3 Testing on Existing Myoelectric Prosthetic Users

Five transradial amputees (three females and two males) with their existing myoelectric prosthesis were recruited to participate in this experiment. Their prosthesis they use is shown in Table 6.1. The feedback device, either mechanotactile or electrotactile, was placed on the triceps region of the upper arm, in the middle of the arm, so as to not interfere with the existing socket on the lower arm, as shown in Figure 6.8. Although integration of sensory feedback devices into the prosthetic socket may provide stable placement and minimise movement, this location was not used during this work to avoid making modifications to existing sockets and potentially impact their comfort and control of their prosthetic device.

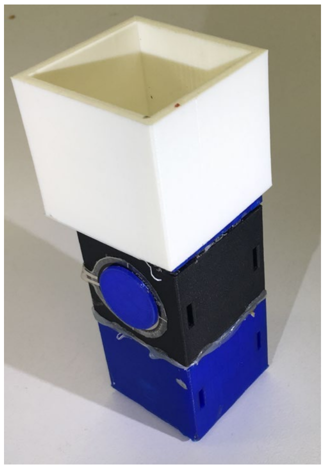
Table 6.1 - Prostheses worn by subjects

Subject	Prosthetic Device Used
1	Ottobock Variplus Rigid Grip
2	Ottobock Variplus Rigid Grip
3	Ottobock DMC Plus Rigid Grip
4	Motion Control Electric Terminal Device
5	Ottobock Variplus Rigid Grip

The force measurement cube, shown in Figure 6.7a, had an extra spacer block (blue) placed underneath for subjects 3-5 to make it easier to grip, as shown in Figure 6.7b. This extra spacer block was added due to feedback from subjects one and two, as they had difficulty in ensuring their thumb and pointer landed on the force measurement cube sensors. A 100g mass was placed in the bottom spacer to lower the centre of gravity, and an open cube (white) was mounted on top for housing additional masses.



(a)

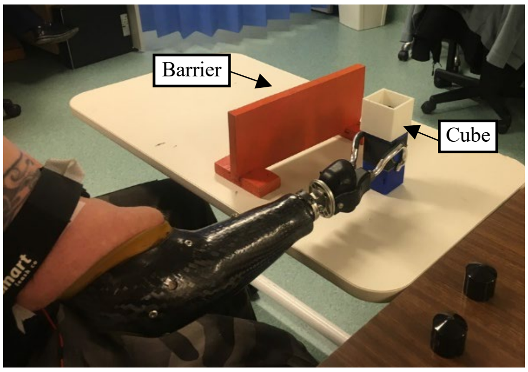


(b)

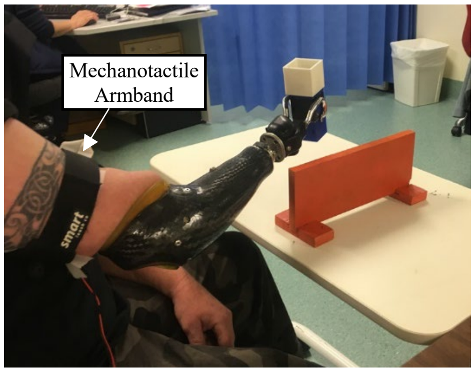
Figure 6.7 - Force measurement cube: (a) Original Design, (b) Modified Design to accommodate additional masses.

Subjects were required to lift the force measurement cube from one side of a 10cm high barrier to the other side, place on the table and release, as shown in Figure 6.8. This was repeated six times (three in each direction). Subjects were instructed to pretend the cube was a fragile object, and they were asked to attempt to move the object with the least amount of force without dropping the object. Since the focus of the study was on their gripping force, we wanted them to focus their attention on controlling their grasp. The subjects were informed that their speed to complete this task was not considered as part of this study, but they should still try to perform the task quickly. A trial period was performed with no feedback and a 100g mass in the force measurement cube to allow the subject to adjust to the scenario of gripping the object, which is the force measurement cube. The data from subjects one and two were the preliminary results with the first mass (200g). They conducted these preliminary experiments in four different rounds (trial movement (100g), feedback technique one (200g), no feedback (200g), feedback technique two (200g)). For subject one, their first technique was mechanotactile feedback and the second feedback was electrotactile. For subject two, their first technique was electrotactile feedback and the second feedback technique mechanotactile. The reason to have the ‘no feedback test’ in the middle of the experiments was to distinguish between any improvement in controlling the grasping force due to extra practice in performing the grasping movement, and any improvement due to the sensory feedback only.

To get a better insight into the effect of different masses on the grasping force, subjects 3-5 ran their experiments with masses of 200g, 300g, 400g, and 500g. However, subject four was only able to comfortably lift up to 300g. An additional no feedback round for subjects 3-5 was placed after the second stimulation session, to ensure the improvement in the second feedback method was not as a result of additional practice of performing the grasping movement.



(a)



(b)

Figure 6.8 - Experimental Setup (a) Subject grasping force measurement cube, (b) Subject lifting cube over barrier using mechanotactile stimulation device.

We obtained the following data to evaluate the effect of the sensory feedback on controlling the gripping force.

(i) the maximum gripping force applied over the whole movement (during initial grip, lifting object, and placing object), and

(ii) the average force applied over the lifting object phase.

For each mass and each feedback method, an average gripping force was calculated for all of the six test movements. Each session was designed to be completed in 45 minutes for each subject.

6.3.1 Electrotactile Feedback

The electrotactile stimulation was delivered by a constant current neurostimulator (Inomed ISIS) and controlled through a LabView interface, as shown in Figure 6.9, sending commands through a .NET APO. Cathodic biphasic pulses were used with a pulse width set at 100 μs and a pulse frequency of 50Hz, with the amplitude corresponding to the applied force proportional to the subject's limits determined in the calibration phase. The

35mm diameter concentric electrode detailed in Chapter 4, was used as the stimulating electrode.

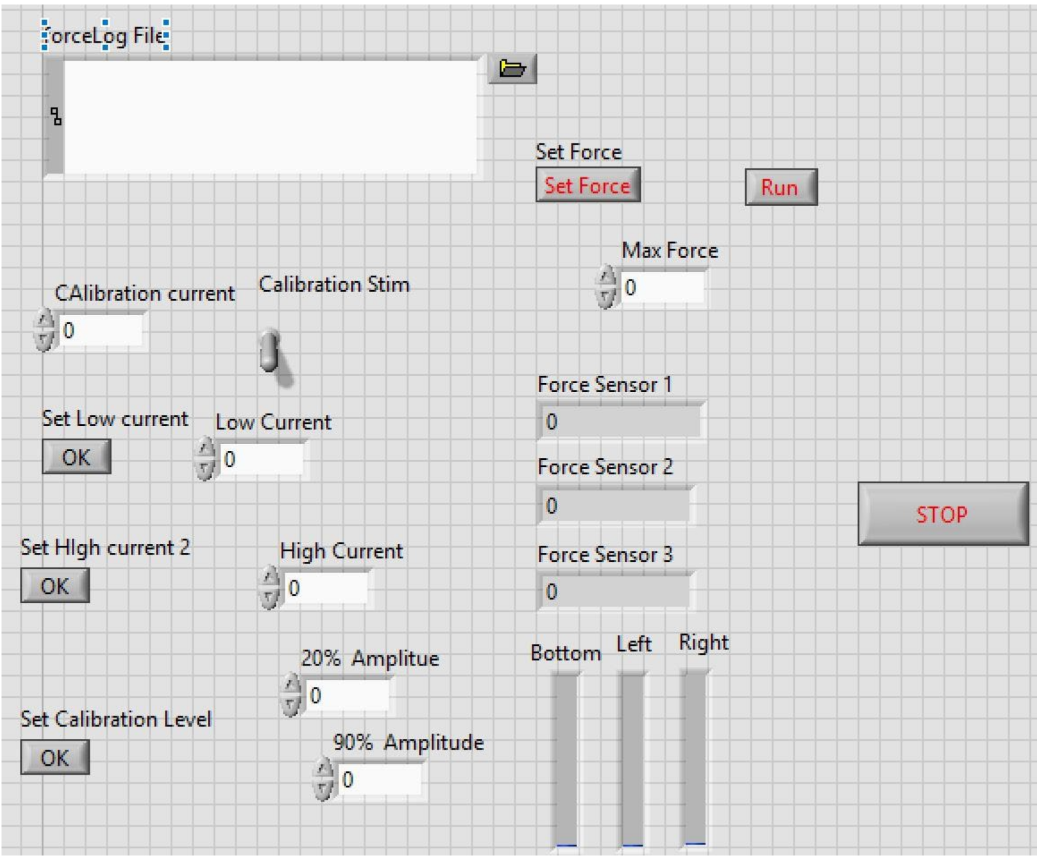


Figure 6.9 - Electrotactile LabVIEW Interface

In the calibration phase, the intensity was slowly turned up until the subject was able to recognise the sensation, which is their detection threshold (DT). The intensity was then turned up until the subject indicated it was starting to get uncomfortable, which is their pain threshold (PT). Their maximum applied current level was set at 90% between their DT and PT to ensure the stimulation always stayed within a comfortable range for the subjects. Similar to the approach taken in [61], the smallest applied current level to the subject was set at 20% between their DT and PT to ensure the minimum stimulation was easily detected. These values were selected to ensure that all stimulations were below an uncomfortable level and stayed at an easily recognisable level. The subject was then asked to squeeze the force measurement cube as hard as they can, and this measurement was recorded as their maximum gripping force. The subject's maximum gripping force was set to correspond to their highest current level, linearly decreasing to their minimum current which was set to correspond to the smallest force (0.2N) that can be detected by the cube [170].

6.3.2 Mechanotactile Feedback

Mechanotactile feedback was provided by the feedback system described in Chapter 3, with the mechanical crank operating longitudinally to the arm. The feedback device was placed on the triceps region (i.e. on the triceps brachii muscle) of the upper arm, so as not to interfere with the existing socket on the lower arm and eliminate alterations to the prosthesis socket. The crank was rested on the skin prior to rotation so that the subjects can detect the stimulation straight away as soon as the crank rotates to simultaneously apply the pressure and skin stretch. The maximum range of the crank rotation for each subject was determined through a calibration phase, where the crank slowly increased its rotation to determine the largest comfortable crank rotation, resetting back to zero position each time. The user was asked to indicate when it was no longer comfortable. The last comfortable movement was then set as the largest crank rotation for the user. The subject was then asked to squeeze the force measurement cube as hard as they can, and this measurement was recorded as their maximum gripping force. This force was then set to correspond to the 90% of their maximum comfortable crank rotation and linearly decreased to 20% of their maximum comfortable crank rotation, set to correspond to smallest detectable force by the force measurement cube (0.2N) [170]

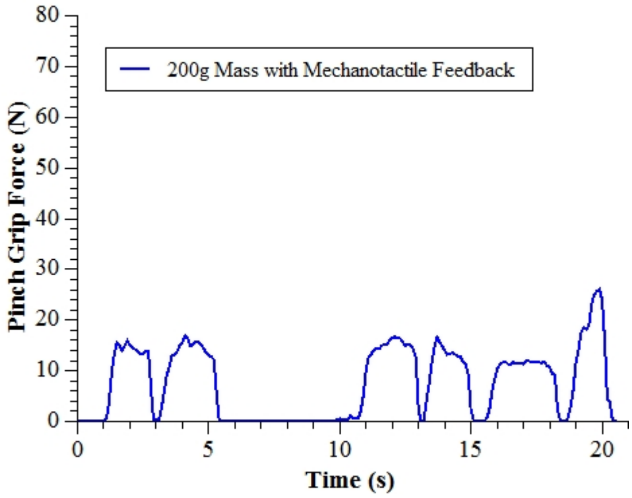
6.4 Experimental Results

6.4.1 Subject One

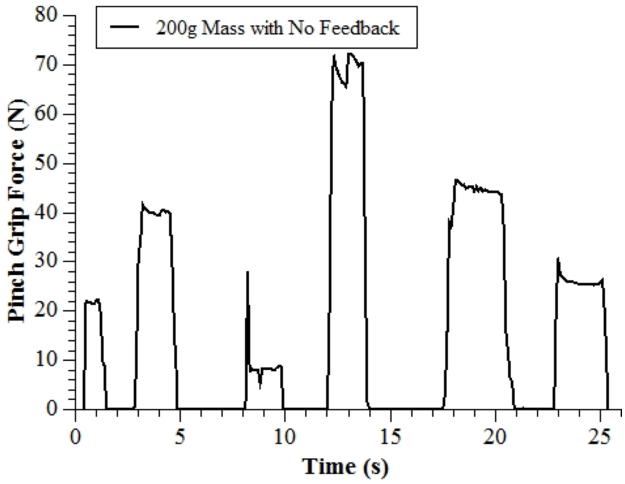
The results for three rounds of experiments with subject one are shown in Table 6.2. For subject one, there was a significant decrease in the maximum gripping force and average grip force when the feedback was on, compared to when the subject received no feedback, which also led to a much lower average grip force. Figure 6.10 shows the typical force data recorded during this experiment.

Table 6.2 - Subject One Results (200g) presented in the order of testing: with mechanotactile feedback, without any feedback, with electrotactile feedback.

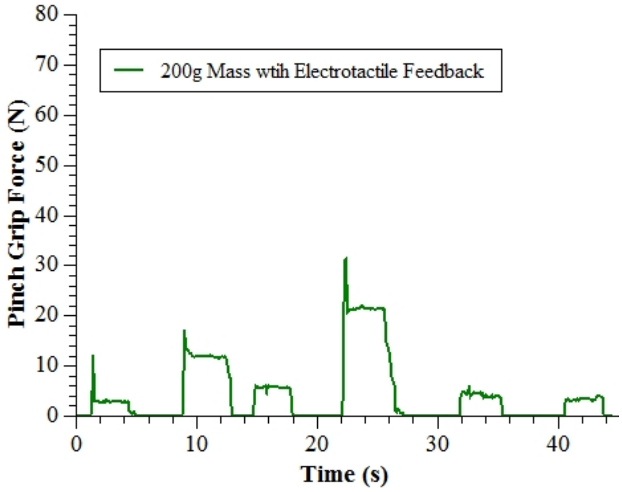
	Mechanotactile	No Feedback	Electrotactile
Maximum Grip (N)	15.0	41.0	11.4
Average Grip (N)	12.3	37.6	7.7



(a)



(b)



(c)

Figure 6.10 - Subject One Force Measurement Curves. (a) Using Mechanotactile Feedback, (b) Using No Sensory Feedback, (c) Using Electrotactile Feedback.

6.4.2 Subject Two

Unfortunately, during the final session of subject two, the timings for when the cube was lifted and when it was placed back on the table were not recorded due to a technical issue. Therefore, the average grip force under the mechanotactile stimulation was unable to be calculated. Subject 2's results are presented in Table 6.3.

Table 6.3 - Subject Two Results presented in the order of testing: with electrotactile feedback, without any feedback, with mechanotactile feedback.

	Electrotactile	**No Feedback**	**Mechanotactile**
Maximum Grip (N)	32.6	55.2	16.6
Average Grip (N)	30.6	52.2	N/A

Subject 2's results demonstrate the reduction in the maximum grip force during the two rounds they received feedback, with electrotactile stimulation and with mechanotactile stimulation, compared to when they used no feedback.

6.4.3 Subject Three

Subject three was able to successfully perform the testing with the four different masses. Their results are displayed in the Table 6.4 .

Subject three was able to use the electrotactile feedback to reduce their maximum gripping force during the 200g and 500g masses on the cube, but not during the 300g and 400g masses. These results, however, may not necessarily be as a result of being unable to use electrotactile feedback. Since for this subject, the electrotactile stimulation was the first session recorded, therefore, they still may be getting used to determining their level of control on the cube. Additionally, other than determining minimum and maximum levels of stimulation, this was the first time the subject had received electrotactile stimulation without any training. In a previous study by Hartman et al. [158], they demonstrated that changing the current level to control the perceived intensity of electrotactile stimulation was sometimes difficult for individuals to correctly identify the intended intensity level of electrotactile stimulation. This follows that training is required to learn to interpret the stimulation level correctly. It has also been suggested that incorporating training into the use of sensory feedback is important to improve the subject's ability to incorporate feedback into their prosthesis control [61, 156, 171].

Table 6.4 - Subject Three Results presented in the order of testing: with electrotactile feedback, first round of no feedback, with mechanotactile feedback, the final round of no feedback

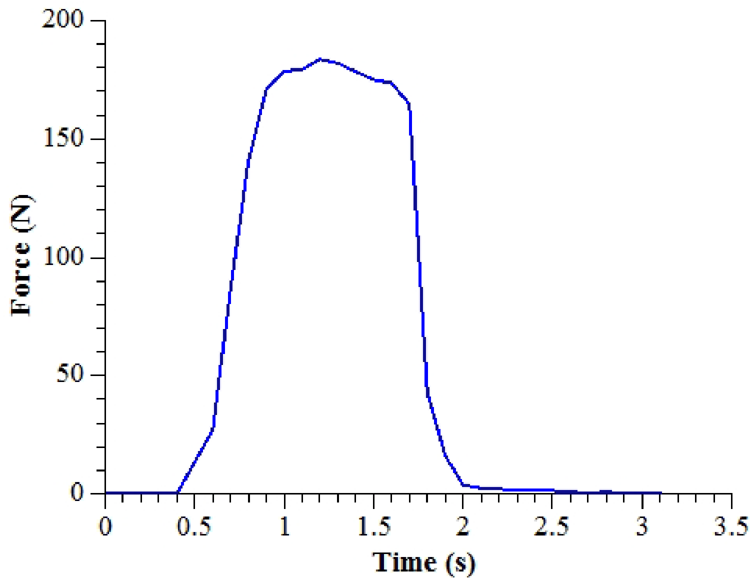
Feedback Method	200g	300g	400g	500g
	Maximum Grip (N)			
Electrotactile	31.6	50.8	40.7	50.1
First No Feedback	37.7	27.4	35.7	63.5
Mechanotactile	13.5	25.8	37.0	40.9
Final No Feedback	39.3	34.0	34.9	59.2
	Average Grip (N)			
Electrotactile	26.8	39.8	35.0	45.0
First No Feedback	35.1	25.2	31.9	50.5
Mechanotactile	10.2	23.8	35.2	37.3
Final No Feedback	34.3	32.3	32.0	55.1

6.4.4 Subject Four

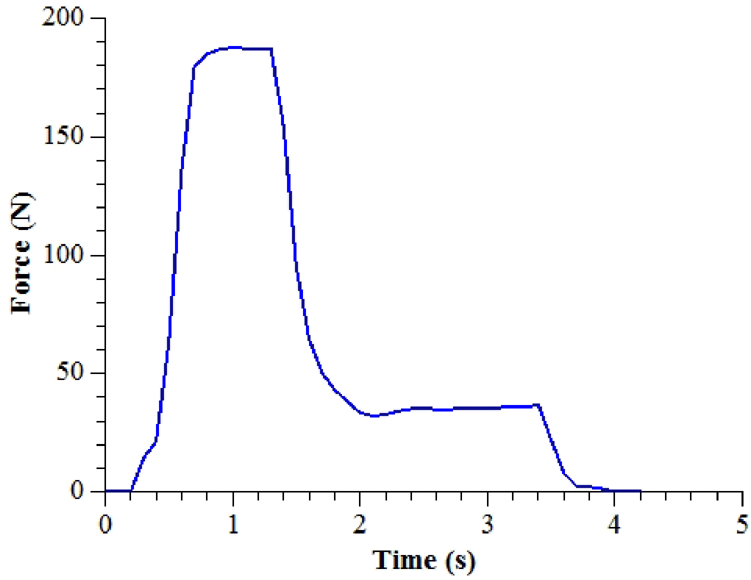
Unfortunately, subject four struggled to lift the heavier objects, so they only performed the tests using the 200g and 300g masses. As shown in Table 6.5 and Figure 6, they have a very large maximum gripping force, in a very short amount of time. However, this subject often went very quickly to maximum force, but with the sensory feedback, they were able to realise that they overshot the desired force and subsequently reduced the gripping force, as shown in Figure 6.11a-b. This is reflected in the lower average gripping force for both mechanotactile and electrotactile feedback methods in Table 6.5. Although subject four was very happy to take part in the experiment, they often found it difficult to control their prosthetic correctly under pressure. From the significant amount of force reached in a short period of time, they presumably sent a large close command through their proportional myoelectric control strategy, which caused the overshoot in the gripping force. This is a common phenomenon in the control of a myoelectric prosthesis that it is difficult and frustrating for users [2, 172] as the control strategies used in commercial prosthesis have not significantly changed over time [173, 174] and the current myoelectric control strategies are highly difficult to master [173]. In addition, fatigue, sweat, and electrode movement that would most likely occur during the testing process could also impact the myoelectric control.

Table 6.5 - Subject Four Results presented in the order of testing: with mechanotactile feedback, first round of no feedback, with electrotactile feedback, the final round of no feedback

Feedback Method	**200g**	**300g**
	Maximum Grip (N)	
Mechanotactile	169.9	184.0
First No Feedback	144.1	178.9
Electrotactile	154.9	177.9
Final No Feedback	156.9	144.8
	Average Grip (N)	
Mechanotactile	100.0	115.3
First No Feedback	115.0	153.5
Electrotactile	100.3	133.4
Final No Feedback	135.0	133.3
	Minimum Grip (N)	
Mechanotactile	44.9	64.7
First No Feedback	83.6	102.7
Electrotactile	22.2	41.7
Final No Feedback	75.9	109.2



(a)



(b)

Figure 6.11 - Sample Force Time Curves from Subject four for Mechanotactile Feedback: (a) No Feedback, (b) Mechanotactile Feedback – showing adjusting force halfway through the grip

Even if the sensory feedback is incorporated correctly, there is a significant delay involved in correcting or controlling the myoelectric movements. In addition to the delay in updating the level of intensity in sensory feedback due to our electrical stimulator's limitations, it has been shown that there can be a delay of up to 300-400ms from when a decision is made until actual movement is detected in prosthesis [175]. The combined effect of the user-unfriendly myoelectric control and delays from the stimulation and those inherent in myoelectric control can explain the overshoot followed by a reduction in force, shown in Figure 6.11. Even with these difficulties, as shown in Table 6.5, the average and minimum forces during gripping the objects are lower in both the mechanotactile and electrotactile feedback demonstrating that the subjects were able to recognise the sensory feedback, interpret and respond accordingly to adjust the gripping force. Since training has shown to improve the control performance of a prosthetic hand or a prosthetic hand user [61, 156, 171], this overshoot may reduce overtime.

6.4.5 Subject Five

Subject Five was able to successfully perform all four rounds of testing with four different masses. Their results are presented in Table 6.6, which shows that subject five was able to reduce their maximum grip force and average grip force using both mechanotactile stimulation and electrotactile stimulation. Only the 500g electrotactile feedback round had a larger value than the non-feedback round. As discussed above, this may be due to variability due to no prior training throughout six tests.

Table 6.6 - Subject Five Results presented in the order of testing: with electrotactile feedback, first round of no feedback, with mechanotactile feedback, the final round of no feedback

Feedback Method	200g	300g	400g	500g
	Maximum Grip (N)			
Electrotactile	7.4	6.9	5.6	12.4
First No Feedback	16.9	12.5	12.5	11.0
Mechanotactile	10.8	7.7	8.0	10.6
Final No Feedback	15.0	12.1	11.5	12.5
	Average Grip (N)			
Electrotactile	6.5	5.8	4.3	11.0
First No Feedback	15.4	11.0	11.4	9.5
Mechanotactile	9.5	6.4	6.4	9.0
Final No Feedback	13.6	10.4	10.4	11.2

6.4.6 Feedback from Subjects

In addition to the recordings, each subject was required to give an indication of the confidence in their ability to pick up the "fragile" object and the comfort for the two stimulation methods. The results are shown in Table 6.7. These confidence scores were obtained with no information provided to them on their experimental gripping force results

Table 6.7 - Subject Feedback on Grasping Control Experiment

Subject	**Confidence** (1-no confidence to 7-very confident, 4 Neutral)			**Comfort** (1-very uncomfortable to 7-very comfortable, 4 Neutral)	
	No Feedback	**Mechanotactile**	**Electrotactile**	**Mechanotactile**	**Electrotactile**
1	3	6	5.5	6	7
2	2	5	5	7	7
3	5	6	7	6	7
4	2	4	5	7	7
5	6	6	6	6	6

The first four subjects all found that sensory feedback increased their confidence in being able to pick up fragile objects. Interestingly subject three rated the confidence in electrotactile feedback the highest, even though it did not make a consistent positive impact in the electrotactile results shown in Table 6.4, possibly due to issues with learning how to interpret the signal correctly and required training to effectively control the prosthesis with this feedback. The first four subjects also found all of the stimulations comfortable, with two out of the four subjects having a slight preference for the electrotactile stimulation. All five subjects also found both stimulation methods comfortable, with two out of the five subjects having a slight preference for the electrotactile stimulation.

6.5 Summary

In this chapter, a force measurement device to measure the grasping force of prosthetic hands was presented and detailed.

The mechanotactile and electrotactile stimulation feedback methods were also tested on transradial amputees with their existing myoelectric prosthesis to help assist them in picking up a "fragile" object, the force measurement cube.

In this chapter, all five subjects were able to benefit the mechanotactile feedback to reduce their gripping force. A previous study [15] did not show any difference in gripping force using a mechanical cuff around the arm with five able-bodied subjects. However, this result in the literature may be due to the able-bodied subjects controlling a prosthetic hand and the difficulties associated with learning to use myoelectric control [173]. Our experiment removed this issue by evaluating the effect of sensory feedback on amputees' ability to control the gripping force using their existing myoelectric prosthetic device, which they have already learnt to use prior to the study.

Although the effect on reducing the gripping force was not as consistent from the use of electrotactile stimulation, it appeared to be rated slightly higher than mechanotactile feedback by some of our subjects in regard to comfort. Previous studies based on virtual reality [61] and without myoelectric control [176] have demonstrated that electrotactile feedback can improve gripping force. Electrotactile stimulation consumes less power than the motors required to drive the mechanotactile stimulation. However, complex circuity is required to produce consistent, safe and predictable electrical pulses with minimal effect due to electrode movement. While mechanotactile stimulation appears to be a more intuitive method of control, the electrotactile stimulation has the ability to alter the number of pulses, the pulse width and frequency which can give a greater range of possible sensations and types to communicate through one device.

The experiments undertaken by Hartman et al. [158] demonstrated the difficulty for individuals to consistently and accurately identify electrotactile intensity levels due to changing the current level of the pulses, but required training to learn to interpret the stimulation level correctly. Further experimentation is required to explore if training can improve the consistency of the electrotactile stimulation and mechanotactile stimulation for sensory feedback. One contributing factor to the difficulties in the recognition of electrotactile stimulation intensity may be due to applying the current levels instantaneously. For mechanotactile stimulation, when you apply a high level of stimulation, it must increase to the desired stimulation level by moving through some intensity levels. In electrotactile stimulation, however, the desired level of current is applied in a single step from the start of the stimulation. Both the increasing nature of the mechanotactile stimulation and the extra time required to reach a higher stimulation may have improved the subject's ability to consistently recognise stimulation levels. Further experimentation is required to determine the impact of these two factors. The idea of using a ramping signal (i.e. a linearly increasing current profile) to apply the electrotactile stimulation should be explored to evaluate whether this improves consistency in intensity perception; i.e. rather than using a single step in the current level to reach the desired stimulation level, gradually increasing/decreasing the current levels in smaller discrete steps to the desired level over a short period of time, similar to how the mechanotactile stimulation is applied.

Ninu et al. [45] suggested that some of their inconsistent results were due to poor controllability of the prosthetic hand. Since our aim of the sensory feedback is to improve overall control of the hand, it is important to recognize the limitations of the prosthetic device, its myoelectric electrode interface, and the mastery level of myoelectric control attained by the prosthetic hand user.

This chapter has shown that sensory feedback enhances prosthetic hand users' ability to control their gripping force and improves their self-confidence. Although the learning associated with a new prosthetic hand was removed in the experimental results presented in this chapter, these results demonstrate the importance of other factors that need to be considered when designing future experiments. All of the tests for each subject were conducted within a 45-minute session to minimise the time required of each subject. However, this may have resulted in fatigue for some of the subjects and therefore needs to be taken into account when planning future experiments. Due to the inherent delays in myoelectric control, and the difficulty in providing fine adjustments, sensory feedback delays need to be as minimal as possible to maximize the efficacy of sensory feedback. Further, future work should incorporate training with the amputees to refine their ability to recognise signals from the electrotactile stimulation and mechanotactile stimulation, and to practice the coordination of the EMG control and the sensory feedback. In addition, these tests were only conducted for a pinch grip with a square object, and therefore further experimentation is required with a variety of grips on objects with different shapes, textures, stiffness and sizes.

Chapter 7

Conclusions and Recommendations for Future Research

This thesis has examined both mechanotactile and electrotactile forms of non-invasive sensory feedback for transradial prosthetic hand users. A new mechanotactile feedback device and an alternative form of electrodes to be used in electrotactile feedback was presented and characterised. The performance of these non-invasive sensory feedback methods were measured and compared, both in the upper and lower regions of the human arm. Finally, their use in closed-loop feedback with existing myoelectric prosthetic users to improve the control of grasping force was demonstrated.

7.1 Conclusions

The following conclusions are drawn from the results presented in this thesis:

- Sensory feedback produced through mechanotactile stimulation from servomotors, meets the required timing limits to avoid any impact on embodiment from delays in stimulation. The presented mechanotactile device results in a statistically significant higher recognition rate when the movement is applied longitudinally to the arm rather than transversally or diagonally. Further, it was found to result in an average JND between 1.4^0 -2.1^0 of rotation without any statistically significant differences being measured across the range of motion or the location of the applied stimulation. This equates to under 8.5% of the acceptable range of motion for all subjects. A consistent linear relationship was also determined between the applied rotation of the mechanotactile stimulator crank to the perceived intensity of the test subject, and this relationship appeared to be subject and location independent.

- The manufacturing method of 3D printed flexible reusable concentric electrodes presented in this thesis was demonstrated to be robust, flexible, have a low environmental and financial cost, and showed a comparable impedance to that of disposable electrodes. Further, they were shown to have comparable performance without the conductive adhesive, opening up possibility for other forms being produced, such as fabric-embedded electrodes.

- The concentric electrode geometry arrangement outperformed the dual separated electrodes in a number of key performance indicators for use in sensory feedback. It was able to increase the comfort and localisation of the induced sensations whilst maintaining a comparable JND and dynamic range. Further, the concentric arrangement decreased the perceived intensity, proportion of uncomfortable induced sensation and resulted in a lower amount of EMG interference.

- There was no statistically significant difference found between the sensitivity of the upper arm and lower arm for mechanotactile stimulation. Further, there was no statistically significant difference in the recognition rate of three channels of applied stimulation (mechanotactile or electrotactile) for the upper or lower arm. This provides an alternate location for stimulation with more surface area and less modifications required to existing prostheses. In addition, it provides a pathway for stimulation in transhumeral prostheses.

- The average recognition rate for six different grip patterns with minimal training ranged from 79.6%-82.9% for mechanotactile and electrotactile stimulation at both the upper and lower regions of the arm. Further, there was no statistically significant difference between these two stimulation methods for accuracy of grip recognition.

- A linear relationship was determined between the current level (above the sensory threshold level) and the perceived intensity of electrotactile stimulation applied to the upper arm.

- Five amputee subjects tested with upper limb difference were able to recognise and utilise the non-invasive sensory feedback, in both the form of mechanotactile and electrotactile stimulation. The amputees were able to incorporate this information to reduce their applied grasping force, maximum and/average force, when lifting an object using the pinch grip. Further, four subjects rated the comfort of the stimulation very high, and there was an increase in their perceived confidence in being able to control their grasping force.

7.2 Recommendations for Future Work

Additional work is still required to develop a deeper understanding of non-invasive sensory feedback methods in order to successfully incorporate into commercial prosthesis for regular use. Possible directions for future research are;

- In this thesis, the mechanotactile feedback locations were fixed for all subjects in one position of the forearm or upper arm, in line with each other. The impact of varying the locations needs to be examined to see the impact on improvement of recognition rates. This includes varying the spacing between the motors, adjusting the transversal alignment to be offset from each other, and examining the impact of using more than three channels of stimulation. In addition, specially designed motors may result in a reduced size.

- Similarly, electrodes were placed in the same transversal line as each other. Varying their locations so they are offset to each other will create further spatial distance between them, possibly affecting the recognition rate and hence requires further investigation. In addition, the sizing of the concentric electrode requires optimisation for both grip recognition and comfort of electrotactile stimulation.

- All experimentation was conducted to determine accuracy with minimal training. However, more experimentation is required to determine the impact of regular training on recognition. In addition, all tests were conducted immediately after training and future work should examine performance of regular and repeated use of the sensory feedback. Unanswered questions include: how often is the recalibration required, how does performance compare after a significant break between stimulations, is the same site able to be repeatedly used for stimulation or is a "piano effect" required where the location of stimulation is regularly moved?

- The mechanotactile and electrotactile stimulation processes used within this work were in "proof of concept" form. However, to enable these to be used in a commercial product, further design work is required. An improved design for the "armband" attachment and encasing of the servomotors is required and the size of the mechanical crank should be optimised to improve recognition and comfort. Further, the electrodes in their current state would be impractical to attach and detach regularly. It is suggested to use the graphene production technique presented in this thesis to develop fabric based electrodes that stretch and can be held firm on to the stimulation surface.

- The work presented in this thesis only examined the impact of providing sensory feedback on grasping control when using fine grip for a short period of time. However, often objects require other grips, such as power grip, and maybe held for a long period of time. Future work will need to examine the role of the sensory feedback and interaction with the automatic hand control system when holding a grip for a long period of time. In addition, further experimentation is required in differentiating feedback from a fine grip and a power grip in the sensory feedback stimulation.

- The experimentation with existing myoelectric prostheses users only incorporated one channel of feedback, but there is demonstrated success in recognising up to three channels of stimulation successfully. Further work could examine the success of existing myoelectric prostheses users in recognising and utilising sensory feedback from up to three different stimulation sites simultaneously

- The electrotactile waveforms used in this study kept the frequency and pulse width constant, to only communicate one style of pressure. Further experimentation is required to examine the role of varying all multiple factors (frequency, pulse width, number of pulses, current amplitude) on the impact of perceived intensity and sensation. In addition, current stimulation produces vibration and tingling feelings and experimentation with varying these factors may result in a more natural feeling of pressure and therefore should be further explored.

- Feedback on the grasping force only was examined within this thesis as it is the highest priority of prostheses users. However, recognition of texture and slippage is an extension of this that may improve embodiment and control of prosthetic devices. Further experimentation of both recognition and incorporating this style of feedback is required in both able-bodied subjects and those with upper limb difference.

- Current mechanotactile stimulation was based on position control of the servo motor. However, future work could investigate the use of force control of the mechanical crank, so that a consistent force can be applied to the arm. This may result in an increased recognition of strength and grip, particularly when the arm muscles are no longer at rest.

- Repetition of the experiments examining the JND across the range of mechanotactile stimulation from the crank based device used in this thesis with a higher number of stimulation values and subjects is required to increase the statistical confidence in the JND being constant across the whole range due to the combination of the normal pressure and transversal skin stretch.

- Test results from one subject indicated that sensory feedback may be useful in training subjects to help control their level of grip force. Further exploration in this area is required, particularly to determine the length and regularity of required training.

- Further work is required to investigate the cognitive load required to use the sensory feedback from mechanotactile and electrotactile stimulation. The experiments undertaken in this thesis were under “ideal” conditions, where participants were only concentrating on the sensation. Future work could examine the impact on cognitive strain through the use of a dual-task method, where participants undertake a task, such as performing simple mathematical calculations, whilst using the prosthetic device with and without sensory feedback [44].

References

[1] R. S. Johansson and J. R. Flanagan, "Coding and use of tactile signals from the fingertips in object manipulation tasks," *Nature Reviews Neuroscience,* vol. 10, no. 5, pp. 345-59, May 2009.

[2] D. J. Atkins, D. C. Y. Heard, and W. H. Donovan, "Epidemiologic Overview of Individuals with Upper-Limb Loss and Their Reported Research Priorities," *JPO: Journal of Prosthetics and Orthotics,* vol. 8, no. 1, pp. 2-11, 1996.

[3] B. Stephens-Fripp, M. Jean Walker, E. Goddard, and G. Alici, "A Survey on What Australian's with Upper Limb Difference Want in a Prosthesis: Justification for using Soft Robotics and Additive Manufacturing for Customised Prosthetic Hands," *Disability and Rehabilitation: Assistive Technology (IIDT),* pp. 1-8, Mar 11 2019.

[4] E. A. Biddiss and T. T. Chau, "Upper limb prosthesis use and abandonment: A survey of the last 25 years," *Prosthetics and Orthotics International,* vol. 31, no. 3, pp. 236-257, 2007.

[5] M. D. Alonzo, F. Clemente, and C. Cipriani, "Vibrotactile Stimulation Promotes Embodiment of an Alien Hand in Amputees With Phantom Sensations," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 23, no. 3, pp. 450-457, 2015.

[6] H. H. Ehrsson, B. Rosen, A. Stockselius, C. Ragno, P. Kohler, and G. Lundborg, "Upper limb amputees can be induced to experience a rubber hand as their own," *Brain,* vol. 131, no. Pt 12, pp. 3443-52, Dec 2008.

[7] L. Schmalzl, A. Kalckert, C. Ragno, and H. H. Ehrsson, "Neural correlates of the rubber hand illusion in amputees: a report of two cases," *Neurocase,* vol. 20, no. 4, pp. 407-20, Aug 2014.

[8] J. D. Brown, T. S. Kunz, D. Gardner, M. K. Shelley, A. J. Davis, and R. B. Gillespie, "An Empirical Evaluation of Force Feedback in Body-Powered Prostheses," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 25, no. 3, pp. 215-226, Mar 2017.

[9] R. W. Mann and S. D. Reimers, "Kinesthetic Sensing for the EMG Controlled "Boston Arm"," *IEEE Transactions on Man-Machine Systems,* vol. 11, no. 1, pp. 110-115, 1970.

[10] A. Chortos, J. Liu, and Z. Bao, "Pursuing prosthetic electronic skin," *Nature Materials,* vol. 15, no. 9, pp. 937-50, Sep 2016.

[11] C. M. Oddo *et al.*, "Intraneural stimulation elicits discrimination of textural features by artificial fingertip in intact and amputee humans," *Elife,* vol. 5, p. e09148, Mar 08 2016.

[12] E. L. Graczyk, M. A. Schiefer, H. P. Saal, B. P. Delhaye, S. J. Bensmaia, and D. J. Tyler, "The neural basis of perceived intensity in natural and artificial touch," *Science Translational Medicine,* vol. 8, no. 362, pp. 362ra142-362ra142, 2016.

[13] G. W. Vidal, M. L. Rynes, Z. Kelliher, and S. J. Goodwin, "Review of Brain-Machine Interfaces Used in Neural Prosthetics with New Perspective on

Somatosensory Feedback through Method of Signal Breakdown," *Scientifica (Cairo),* vol. 2016, p. 8956432, 2016.

[14] D. Perruchoud, I. Pisotta, S. Carda, M. M. Murray, and S. Ionta, "Biomimetic rehabilitation engineering: the importance of somatosensory feedback for brain-machine interfaces," *Journal of Neural Engineering,* vol. 13, no. 4, p. 041001, Aug 2016.

[15] 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 *2016 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, 2016, pp. 5439-5442.

[16] C. Antfolk *et al.*, "Transfer of tactile input from an artificial hand to the forearm: experiments in amputees and able-bodied volunteers," *Disability and Rehabilitation: Assistive Technology,* vol. 8, no. 3, pp. 249-254, 2013/05/01 2013.

[17] K. A. Kaczmarek, J. G. Webster, P. Bach-y Rita, and W. J. Tompkins, "Electrotactile and vibrotactile displays for sensory substitution systems," *IEEE Transactions on Biomedical Engineering,* vol. 38, 1991.

[18] K. Kim and J. E. Colgate, "Haptic feedback enhances grip force control of sEMG-controlled prosthetic hands in targeted reinnervation amputees," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 20, no. 6, pp. 798-805, Nov 2012.

[19] 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 Transactions on Neural Systems and Rehabilitation Engineering,* vol. 21, no. 1, pp. 112-120, 2013.

[20] D. Zhang, H. Xu, P. B. Shull, J. Liu, and X. Zhu, "Somatotopical feedback versus non-somatotopical feedback for phantom digit sensation on amputees using electrotactile stimulation," *Journal of Neuroengineering and Rehabilitation,* vol. 12, p. 44, May 02 2015.

[21] G. H. Chai *et al.*, "Phantom finger perception evoked with transcutaneous electrical stimulation for sensory feedback of prosthetic hand," in *2013 6th International IEEE/EMBS Conference on Neural Engineering (NER)*, 2013, pp. 271-274.

[22] X. X. Liu, G. H. Chai, H. E. Qu, and N. Lan, "A sensory feedback system for prosthetic hand based on evoked tactile sensation," in *2015 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, 2015, pp. 2493-2496.

[23] G. Chai, X. Sui, S. Li, L. He, and N. Lan, "Characterization of evoked tactile sensation in forearm amputees with transcutaneous electrical nerve stimulation," *Journal of Neural Engineering,* vol. 12, no. 6, p. 066002, Dec 2015.

[24] P. Li, G. H. Chai, K. H. Zhu, N. Lan, and X. H. Sui, "Effects of electrode size and spacing on sensory modalities in the phantom thumb perception area for the forearm amputees," in *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, Milan, Italy, 2015, vol. 2015-November, pp. 3383-3386.

[25] U. Wijk, P. Svensson, C. Antfolk, I. K. Carlsson, A. Björkman, and B. Rosén, "Touch on predefined areas on the forearm can be associated with specific fingers: Towards a new principle for sensory feedback in hand prostheses," *Journal of Rehabilitation Medicine,* vol. 51, 2019.

[26] R. A. Normann and E. Fernandez, "Clinical applications of penetrating neural interfaces and Utah Electrode Array technologies," *Journal of Neural Engineering,* vol. 13, no. 6, p. 061003, Dec 2016.

[27] B. T. Nghiem *et al.*, "Providing a sense of touch to prosthetic hands," *Plastic and Reconstructive Surgery,* vol. 135, no. 6, pp. 1652-63, Jun 2015.

[28] F. M. Petrini *et al.*, "Six-Month Assessment of a Hand Prosthesis with Intraneural Tactile Feedback," (in eng), *Annals of Neurology,* vol. 85, no. 1, pp. 137-154, Jan 2019.

[29] E. D'Anna *et al.*, "A closed-loop hand prosthesis with simultaneous intraneural tactile and position feedback," *Science Robotics,* vol. 4, no. 27, p. eaau8892, 2019.

[30] M. Ortiz-Catalan, B. Hakansson, and R. Branemark, "An osseointegrated human-machine gateway for long-term sensory feedback and motor control of artificial limbs," *Science Translational Medicine,* vol. 6, no. 257, p. 257re6, Oct 08 2014.

[31] T. S. Davis *et al.*, "Restoring motor control and sensory feedback in people with upper extremity amputations using arrays of 96 microelectrodes implanted in the median and ulnar nerves," (in eng), *Journal of Neural Engineering,* vol. 13, no. 3, p. 036001, Jun 2016.

[32] J. A. George *et al.*, "Biomimetic sensory feedback through peripheral nerve stimulation improves dexterous use of a bionic hand," *Science Robotics,* vol. 4, no. 32, p. eaax2352, 2019.

[33] H. L. Benz, "Upper extremity prosthesis user perspectives on unmet needs and innovative technology," in *IEEE Engineering in Medicine and Biology Society (EMBC)*, 2016, vol. 2016-October, pp. 287-290.

[34] A. D. Roche, B. Lakey, I. Mendez, I. Vujaklija, D. Farina, and O. C. Aszmann, "Clinical Perspectives in Upper Limb Prostheses: An Update," *Current Surgery Reports,* vol. 7, no. 3, 2019.

[35] 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," in *Brain*, 2011.

[36] C. Cipriani, M. Controzzi, and M. C. Carrozza, "Objectives, criteria and methods for the design of the SmartHand transradial prosthesis," *Robotica,* vol. 28, no. 06, pp. 919-927, 2010.

[37] B. Peerdeman *et al.*, "Myoelectric forearm prostheses: State of the art from a user-centered perspective," *Journal of Rehabilitation Research and Development,* vol. 48, no. Number 6, pp. 719-738, 2011.

[38] F. Cordella *et al.*, "Literature Review on Needs of Upper Limb Prosthesis Users," *Frontiers in Neuroscience,* Review vol. 10, no. 209, 2016-May-12 2016.

[39] M. Markovic, S. Dosen, C. Cipriani, D. Popovic, and D. Farina, "Stereovision and augmented reality for closed-loop control of grasping in hand prostheses," *Journal of Neural Engineering,* vol. 11, no. 4, p. 046001, 2014 2014.

[40] G. Sriram, A. N. Jensen, and S. C. Chiu, "Slippage control for a smart prosthetic hand prototype via modified tactile sensory feedback," in *IEEE International Conference on Electro/Information Technology*, 2014, pp. 225-230.

[41] W. Shaw-Cortez, D. Oetomo, C. Manzie, and P. Choong, "Towards dynamic object manipulation with tactile sensing for prosthetic hands," in *2016 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, 2016, pp. 1164-1169.

[42] Y. Cho, K. Liang, F. Folowosele, B. Miller, and N. V. Thakor, "Wireless Temperature Sensing Cosmesis for Prosthesis," in *2007 IEEE 10th International Conference on Rehabilitation Robotics*, 2007, pp. 672-677.

[43] Y. Ueda and C. Ishii, "Development of a feedback device of temperature sensation for a myoelectric prosthetic hand by using Peltier element," in *2016 International Conference on Advanced Mechatronic Systems (ICAMechS)*, 2016, pp. 488-493.

[44] H. Yamada, Y. Yamanoi, K. Wakita, and R. Kato, "Investigation of a cognitive strain on hand grasping induced by sensory feedback for myoelectric hand," in *2016 IEEE International Conference on Robotics and Automation (ICRA)*, 2016, pp. 3549-3554.

[45] A. Ninu, S. Dosen, S. Muceli, F. Rattay, H. Dietl, and D. Farina, "Closed-Loop Control of Grasping With a Myoelectric Hand Prosthesis: Which Are the Relevant Feedback Variables for Force Control?," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 22, no. 5, pp. 1041-1052, 2014.

[46] M. Nabeel, "Vibrotactile stimulation for 3D printed prosthetic hand," in *Robotics and Artificial Intelligence (ICRAI), 2016 2nd International Conference on.*, 2016, pp. 202-207: IEEE.

[47] 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," *JPO: Journal of Prosthetics and Orthotics,* vol. 28, no. 4, pp. 136-144, 2016.

[48] P. Chaubey, T. Rosenbaum-Chou, W. Daly, and D. Boone, "Closed-Loop Vibratory Haptic Feedback in Upper-Limb Prosthetic Users," *JPO: Journal of Prosthetics and Orthotics,* vol. 26, no. 3, pp. 120-127, 2014.

[49] 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 Transactions on Neural Systems and Rehabilitation Engineering,* vol. 24, no. 12, pp. 1314-1322, 2016.

[50] N. Mohamad Hanif, P. Chappell, N. White, A. Cranny, and N. N. Nik Hashim, "Tactile to vibrotactile sensory feedback interface for prosthetic hand users," presented at the 2016 IEEE EMBS Conference on Biomedical Engineering and Sciences (IECBES), 2016. Available: http://eprints.soton.ac.uk/405015/

[51] T. Li, H. Huang, C. Antfolk, J. Justiz, and V. M. Koch, "Tactile display on the remaining hand for unilateral hand amputees," *Current Directions in Biomedical Engineering,* vol. 2, no. 1, 2016.

[52] E. Raveh, J. Friedman, and S. Portnoy, "Visuomotor behaviors and performance in a dual-task paradigm with and without vibrotactile feedback when using a myoelectric controlled hand," *Assistive Technology,* pp. 1-7, Jun 19 2017.

[53] C. J. Hasson and J. Manczurowsky, "Effects of kinematic vibrotactile feedback on learning to control a virtual prosthetic arm," *Journal of Neuroengineering and Rehabilitation,* vol. 12, p. 31, Mar 24 2015.

[54] J. M. Walker, A. A. Blank, P. A. Shewokis, and M. K. O'Malley, "Tactile feedback of object slip facilitates virtual object manipulation," *IEEE Transactions on Haptics,* vol. 8, no. 4, pp. 454-466, 2015.

[55] M. Aziziaghdam and E. Samur, "Providing contact sensory feedback for upper limb robotic prosthesis," in *2014 IEEE Haptics Symposium (HAPTICS)*, 2014, pp. 575-579.

[56] T. Morita, T. Kikuchi, and C. Ishii, "Development of Sensory Feedback Device for Myoelectric Prosthetic Hand to Provide Hardness of Objects to Users," *Journal of Robotics and Mechatronics,* vol. s28 no. 3, no. 361-370, 2016.

[57] S. Casini, M. Morvidoni, M. Bianchi, M. Catalano, G. Grioli, and A. Bicchi, "Design and realization of the CUFF - clenching upper-limb force feedback wearable device for distributed mechano-tactile stimulation of normal and tangential skin forces," in *2015 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, 2015, pp. 1186-1193.

[58] C. Antfolk, S.-O. Bjorkman A Fau - Frank, F. Frank So Fau - Sebelius, G. Sebelius F Fau - Lundborg, B. Lundborg G Fau - Rosen, and B. Rosen, "Sensory feedback from a prosthetic hand based on air-mediated pressure from the hand to the forearm skin," (in eng), *Journal of Rehabilitation Medicine,* no. 44, pp. 702-707, 2012.

[59] A. Akhtar, M. Nguyen, L. Wan, B. Boyce, P. Slade, and T. Bretl, "Passive mechanical skin stretch for multiple degree-of-freedom proprioception in a hand prosthesis," in *International Conference on Human Haptic Sensing and Touch Enabled Computer Applications*, 2014, pp. 120-128: Springer.

[60] J. Wheeler, K. Bark, J. Savall, and M. Cutkosky, "Investigation of Rotational Skin Stretch for Proprioceptive Feedback With Application to Myoelectric Systems," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 18, no. 1, pp. 58-66, 2010.

[61] N. Jorgovanovic, S. Dosen, D. J. Djozic, G. Krajoski, and D. Farina, "Virtual grasping: closed-loop force control using electrotactile feedback," *Computational and Mathematical Methods in Medicine,* vol. 2014, p. 120357, 2014.

[62] M. Isakovic *et al.*, "Electrotactile Feedback Improves Performance and Facilitates Learning in the Routine Grasping Task," (in eng), *European Journal of Translational Myology,* vol. 26, no. 3, p. 6069, Jun 13 2016.

[63] M. A. Schweisfurth, M. Markovic, S. Dosen, F. Teich, B. Graimann, and D. Farina, "Electrotactile EMG feedback improves the control of prosthesis grasping force," *Journal of Neural Engineering,* vol. 13, no. 5, p. 056010, Oct 2016.

[64] P. Shi and X. Shen, "Sensation Feedback and Muscle Response of Electrical Stimulation on the Upper Limb Skin: A Case Study," presented at the 2015 Seventh International Conference on Measuring Technology and Mechatronics Automation, Nanchang, China, 2015.

[65] H. Xu, D. Zhang, J. C. Huegel, W. Xu, and X. Zhu, "Effects of Different Tactile Feedback on Myoelectric Closed-Loop Control for Grasping Based on Electrotactile Stimulation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 24, no. 8, pp. 827-836, 2016.

[66] K. Choi, P. Kim, K. S. Kim, and S. Kim, "Two-channel electrotactile stimulation for sensory feedback of fingers of prosthesis," in *2016 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, 2016, pp. 1133-1138.

[67] G. K. Patel, S. Dosen, C. Castellini, and D. Farina, "Multichannel electrotactile feedback for simultaneous and proportional myoelectric control," *Journal of Neural Engineering,* vol. 13, no. 5, p. 056015, Oct 2016.

[68] D. Pamungkas and K. Ward, "Electro-tactile Feedback System for a Prosthetic Hand," presented at the International Conference on Machine Vision and Mechatronics in Practice, Twoowoomba, Australia, 2015.

[69] M. Strbac *et al.*, "Integrated and flexible multichannel interface for electrotactile stimulation," *Journal of Neural Engineering,* vol. 13, no. 4, p. 046014, Aug 2016.

[70] M. Franceschi, L. Seminara, L. Pinna, S. Dosen, D. Farina, and M. Valle, "Preliminary evaluation of the tactile feedback system based on artificial skin and electrotactile stimulation," in *2015 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, 2015, pp. 4554-4557.

[71] C. Hartmann *et al.*, "Towards prosthetic systems providing comprehensive tactile feedback for utility and embodiment," in *2014 IEEE Biomedical Circuits and Systems Conference (BioCAS) Proceedings*, 2014, pp. 620-623.

[72] S. Dosen, M.-C. Schaeffer, and D. Farina, "Time-division multiplexing for myoelectric closed-loop control using electrotactile feedback," *Journal of NeuroEngineering and Rehabilitation,* vol. 11, p. 138, 09/15 2014.

[73] L. Jiang, Q. Huang, J. Zhao, D. Yang, S. Fan, and H. Liu, "Noise cancellation for electrotactile sensory feedback of myoelectric forearm prostheses," in *2014 IEEE International Conference on Information and Automation (ICIA)*, 2014, pp. 1066-1071.

[74] B. Xu *et al.*, "An Epidermal Stimulation and Sensing Platform for Sensorimotor Prosthetic Control, Management of Lower Back Exertion, and Electrical Muscle Activation," *Advanced Materials,* vol. 28, no. 22, pp. 4462-71, Jun 2016.

[75] S. Wilson and S. Dirven, "Audio sensory substitution for human-in-the-loop force feedback of upper limb prosthetics," in *2016 23rd International Conference on Mechatronics and Machine Vision in Practice (M2VIP)*, 2016, pp. 1-6.

[76] A. Gibson and P. Artemiadis, "Object discrimination using optimized multi-frequency auditory cross-modal haptic feedback," in *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, 2014, pp. 6505-6508.

[77] J. Gonzalez, H. Suzuki, N. Natsumi, M. Sekine, and W. Yu, "Auditory display as a prosthetic hand sensory feedback for reaching and grasping tasks," in *2012 Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, 2012, pp. 1789-1792.

[78] M. Markovic, H. Karnal, B. Graimann, D. Farina, and S. Dosen, "GLIMPSE: Google Glass interface for sensory feedback in myoelectric hand prostheses," *Journal of Neural Engineering,* vol. 14, no. 3, p. 036007, Jun 2017.

[79] F. Clemente, S. Dosen, L. Lonini, M. Markovic, D. Farina, and C. Cipriani, "Humans Can Integrate Augmented Reality Feedback in Their Sensorimotor Control of a Robotic Hand," *IEEE Transactions on Human-Machine Systems,* vol. 47, no. 4, pp. 583-589, 2017.

[80] M. D'Alonzo, S. Dosen, C. Cipriani, and D. Farina, "HyVE: hybrid vibro-electrotactile stimulation for sensory feedback and substitution in rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 22, no. 2, pp. 290-301, Mar 2014.

[81] M. D'Alonzo, S. Dosen, C. Cipriani, and D. Farina, "HyVE-hybrid vibro-electrotactile stimulation-is an efficient approach to multi-channel sensory feedback," *IEEE Transactions on Haptics,* vol. 7, no. 2, pp. 181-90, Apr-Jun 2014.

[82] F. Clemente and C. Cipriani, "A novel device for multi-modal sensory feedback in hand prosthetics: Design and preliminary prototype," in *2014 IEEE Haptics Symposium (HAPTICS)*, 2014, pp. 569-573.

[83] M. C. Jimenez and J. A. Fishel, "Evaluation of force, vibration and thermal tactile feedback in prosthetic limbs," in *2014 IEEE Haptics Symposium (HAPTICS)*, 2014, pp. 437-441.

[84] T. Li, H. Huang, J. Justiz, and V. M. Koch, "A Miniature Multimodal Actuator for Effective Tactile Feedback: Design and Characterization," *Procedia Engineering,* vol. 168, pp. 1547-1550, 2016.

[85] M. Reza Motamedi, M. Otis, and V. Duchaine, "The Impact of Simultaneously Applying Normal Stress and Vibrotactile Stimulation for Feedback of Exteroceptive Information," *Journal of Biomechanical Engineering,* vol. 139, no. 6, Jun 01 2017.

[86] H. J. Witteveen, H. S. Rietman, and P. H. Veltink, "Vibrotactile grasping force and hand aperture feedback for myoelectric forearm prosthesis users," *Prosthetics and Orthototics International,* vol. 39, no. 3, pp. 204-12, Jun 2015.

[87] A. Ajoudani *et al.*, "Exploring teleimpedance and tactile feedback for intuitive control of the Pisa/IIT SoftHand," *IEEE Transactions on Haptics,* vol. 7, no. 2, pp. 203-15, Apr-Jun 2014.

[88] I. Saunders and S. Vijayakumar, "The role of feed-forward and feedback processes for closed-loop prosthesis control," *Journal of NeuroEngineering and Rehabilitation,* journal article vol. 8, no. 1, p. 60, October 27 2011.

[89] C. Cipriani, M. D'Alonzo, and M. C. Carrozza, "A Miniature Vibrotactile Sensory Substitution Device for Multifingered Hand Prosthetics," *IEEE Transactions on Biomedical Engineering,* vol. 59, no. 2, pp. 400-408, 2012.

[90] J. Cohen, M. Niwa, R. W. Lindeman, H. Noma, Y. Yanagida, and K. Hosaka, "A closed-loop tactor frequency control system for vibrotactile feedback," in *CHI'05 Extended Abstracts on Human Factors in Computing Systems*, 2005, pp. 1296-1299: ACM.

[91] C. E. Stepp, Y. An Q Fau - Matsuoka, and Y. Matsuoka, "Repeated training with augmentative vibrotactile feedback increases object manipulation performance," (in eng), *PLoS One,* vol. 7, no. 2, 20120302 DCOM- 20120827 2012.

[92] C. E. Stepp and Y. Matsuoka, "Vibrotactile Sensory Substitution for Object Manipulation: Amplitude Versus Pulse Train Frequency Modulation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 20, no. 1, pp. 31-37, 2012.

[93] C. Pylatiuk, A. Kargov, and S. Schulz, "Design and Evaluation of a Low-Cost Force Feedback System for Myoelectric Prosthetic Hands," *JPO: Journal of Prosthetics and Orthotics,* vol. 18, no. 2, pp. 57-61, 2006.

[94] A. Chatterjee, P. Chaubey, J. Martin, and N. Thakor, "Testing a Prosthetic Haptic Feedback Simulator With an Interactive Force Matching Task," *JPO: Journal of Prosthetics and Orthotics,* vol. 20, no. 2, pp. 27-34, 2008.

[95] H. J. Witteveen, E. A. Droog, J. S. Rietman, and P. H. Veltink, "Vibro- and electrotactile user feedback on hand opening for myoelectric forearm prostheses," (in eng), *IEEE Transactions on Biomedical Engineering,* vol. 59, no. 8, pp. 2219-26, Aug 2012.

[96] 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.

[97] S. Shimada, K. Fukuda, and K. Hiraki, "Rubber Hand Illusion under Delayed Visual Feedback," *PLOS ONE,* vol. 4, no. 7, p. e6185, 2009.

[98] A. Y. J. Szeto and F. A. Saunders, "Electrocutaneous Stimulation for Sensory Communication in Rehabilitation Engineering," *IEEE Transactions on Biomedical Engineering,* vol. BME-29, no. 4, pp. 300-308, 1982.

[99] K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita, and W. J. Tompkins, "Electrotactile and vibrotactile displays for sensory substitution systems," *IEEE Transactions on Biomedical Engineering,* vol. 38, no. 1, pp. 1-16, 1991.

[100] A. Y. Szeto and R. R. Riso, "Sensory Feedback Using Electrical Stimulation of the Tactile Sense," in *Rehabilitation Engineering*, J. H. L. R.V. Smith, Ed.: CRC Press, 1990, pp. 29-78.

[101] D. W. Tan, M. A. Schiefer, M. W. Keith, J. R. Anderson, J. Tyler, and D. J. Tyler, "A neural interface provides long-term stable natural touch perception," *Science Translational Medicine,* vol. 6, no. 257, p. 257ra138, Oct 08 2014.

[102] N. Fallahian, H. Saeedi, H. Mokhtarinia, and F. T. Ghomshe, "Sensory feedback add-on for upper-limb prostheses," *Prosthetics and Orthotics International,* vol. 0, no. 0, p. 0309364616677653, 2016.

[103] K. R. Schoepp, M. R. Dawson, J. S. Schofield, J. P. Carey, and J. S. Hebert, "Design and Integration of an Inexpensive Wearable Mechanotactile Feedback

System for Myoelectric Prostheses," *IEEE Journal Translational Engineering in Health and Medicine,* vol. 6, p. 2100711, 2018.

[104] K. Bark, J. Wheeler, P. Shull, J. Savall, and M. Cutkosky, "Rotational Skin Stretch Feedback: A Wearable Haptic Display for Motion," *IEEE Transactions on Haptics,* vol. 3, no. 3, pp. 166-176, Jul-Sep 2010.

[105] E. Battaglia, J. P. Clark, M. Bianchi, M. G. Catalano, A. Bicchi, and M. K. O'Malley, "The Rice Haptic Rocker: skin stretch haptic feedback with the Pisa/IIT SoftHand," in *World Haptics Conference (WHC), 2017 IEEE*, 2017, pp. 7-12: IEEE.

[106] M. Rossi, M. Bianchi, E. Battaglia, M. G. Catalano, and A. Bicchi, "HapPro: A Wearable Haptic Device for Proprioceptive Feedback," *IEEE Trans Biomed Eng,* vol. 66, no. 1, pp. 138-149, Jan 2019.

[107] A. Bjorkman, U. Wijk, C. Antfolk, I. Bjorkman-Burtscher, and B. Rosen, "Sensory Qualities of the Phantom Hand Map in the Residual Forearm of Amputees," *Journal of Rehabilitation Medicine,* Article vol. 48, no. 4, pp. 365-370, Apr 2016.

[108] H. Wang *et al.*, "Towards determining the afferent sites of perception feedback on residual arms of amputees with transcutaneous electrical stimulation," in *2015 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, 2015, pp. 3367-3370.

[109] J. Belter, J. Segil, A. Dollar, and R. Weir, "Mechanical design and performance specifications of anthropomorphic prosthetic hands: A review," (in English), *Journal of Rehabilitation Research and Development,* vol. 50, no. 5, pp. 599-618, 2013 2013.

[110] A. Blank, A. M. Okamura, and K. J. Kuchenbecker, "Identifying the role of proprioception in upper-limb prosthesis control," *ACM Transactions on Applied Perception,* vol. 7, no. 3, pp. 1-23, 2010.

[111] T. Pistohl, D. Joshi, G. Ganesh, A. Jackson, and K. Nazarpour, "Artificial proprioceptive feedback for myoelectric control," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 23, no. 3, pp. 498-507, May 2015.

[112] M. Strbac *et al.*, "Short- and Long-Term Learning of Feedforward Control of a Myoelectric Prosthesis with Sensory Feedback by Amputees," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* Jun 06 2017.

[113] 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 Transactions on Neural Systems and Rehabilitation Engineering,* vol. 25, no. 5, pp. 469-480, 2017.

[114] C. Antfolk, M. D'Alonzo, B. Rosén, G. Lundborg, F. Sebelius, and C. Cipriani, "Sensory feedback in upper limb prosthetics," (in English), *Expert Review of Medical Devices,* vol. 10, no. 1, pp. 45-54, Jan 2013 2013.

[115] R. Bekrater-Bodmann *et al.*, "The Importance of Synchrony and Temporal Order of Visual and Tactile Input for Illusory Limb Ownership Experiences – An fMRI Study Applying Virtual Reality," *PLOS ONE,* vol. 9, no. 1, p. e87013, 2014.

[116] 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 Transactions on Neural Systems and Rehabilitation Engineering,* vol. 21, no. 1, pp. 112-20, Jan 2013.

[117] 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," *BioMedical Engineering OnLine,* journal article vol. 9, no. 1, p. 50, 2010.

[118] B. Stephens-Fripp, G. Alici, and R. Mutlu, "A Review of Non-Invasive Sensory Feedback Methods for Transradial Prosthetic Hands," *IEEE Access,* vol. 6, pp. 6878-6899, 2018.

[119] S. S. Stevens, "The Direct Estimation of Sensory Magnitudes: Loudness," *The American Journal of Psychology,* vol. 69, no. 1, pp. 1-25, 1956.

[120] C. Antfolk, V. Kopta, J. Farserotu, J. D. Decotignie, and C. Enz, "The WiseSkin artificial skin for tactile prosthetics: A power budget investigation," in *2014 8th International Symposium on Medical Information and Communication Technology (ISMICT)*, 2014, pp. 1-4.

[121] M. Vergara, J. L. Sancho-Bru, V. Gracia-Ibanez, and A. Perez-Gonzalez, "An introductory study of common grasps used by adults during performance of activities of daily living," (in eng), *Journal of Hand Therapy,* vol. 27, no. 1545-004X (Electronic), pp. 225-234, 20140725 DCOM- 20150511 2014.

[122] G. Fechner, "Elements of Psychophysics. Adler HE, translator; Howes DH, Boring EG, editors," ed: New York: Holt, Rinehart & Winston, 1966.

[123] S. S. Stevens, "On the psychophysical law," *Psychological Review,* vol. 64, no. 3, pp. 153-181, 1957.

[124] T. A. Kuiken *et al.*, "Targeted Muscle Reinnervation for Real-Time Myoelectric Control of Multifunction Artificial Arms," *JAMA : the journal of the American Medical Association,* vol. 301, no. 6, pp. 619-628, 2009.

[125] B. Stephens-Fripp, R. Mutlu, and G. Alici, "Applying Mechanical Pressure and Skin Stretch Simultaneously for Sensory Feedback in Prosthetic Hands," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*, 2018, pp. 230-235.

[126] F. A. Wichmann and N. J. Hill, "The psychometric function: I. Fitting, sampling, and goodness of fit," *Perception & Psychophysics,* journal article vol. 63, no. 8, pp. 1293-1313, November 01 2001.

[127] J. P. Clark, S. Y. Kim, and M. K. O'Malley, "The Rice Haptic Rocker: Comparing Longitudinal and Lateral Upper-Limb Skin Stretch Perception," in *Haptics: Science, Technology, and Applications*(Lecture Notes in Computer Science, 2018, pp. 125-134.

[128] S. B. Schorr and A. M. Okamura, "Three-Dimensional Skin Deformation as Force Substitution: Wearable Device Design and Performance During Haptic Exploration of Virtual Environments," *IEEE Transactions on Haptics,* vol. 10, no. 3, pp. 418-430, Jul-Sep 2017.

[129] D. Poulos *et al.*, "The neural signal for the intensity of a tactile stimulus," *The Journal of Neuroscience,* vol. 4, no. 8, pp. 2016-2024, 1984.

[130] B. Geng, J. Dong, W. Jensen, S. Dosen, D. Farina, and E. N. Kamavuako, "Psychophysical Evaluation of Subdermal Electrical Stimulation in Relation to Prosthesis Sensory Feedback," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* pp. 1-1, 2018.

[131] A. Y. J. Szeto, J. Lyman, and R. E. Prior, "Electrocutaneous Pulse Rate and Pulse Width Psychometric Functions for Sensory Communications," *Human Factors,* vol. 21, no. 2, pp. 241-249, 1979.

[132] 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 Review of Medical Devices,* vol. 11, no. 5, pp. 499-511, Sep 2014.

[133] M. Atzori and H. Müller, "Control Capabilities of Myoelectric Robotic Prostheses by Hand Amputees: A Scientific Research and Market Overview," (in English), *Frontiers in Systems Neuroscience,* Review vol. 9, no. 162, 2015-November-30 2015.

[134] B. Stephens-Fripp, R. Mutlu, and G. Alici, "Using Vibration Motors to Create Tactile Apparent Movement for Transradial Prosthetic Sensory Feedback," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*, 2018, pp. 213-218.

[135] Y. Khan *et al.*, "Inkjet-Printed Flexible Gold Electrode Arrays for Bioelectronic Interfaces," *Advanced Functional Materials,* vol. 26, no. 7, pp. 1004-1013, 2016.

[136] T. Asai, N. Kanayama, S. Imaizumi, S. Koyama, and S. Kaganoi, "Development of Embodied Sense of Self Scale (ESSS): Exploring Everyday Experiences Induced by Anomalous Self-Representation," (in English), *Frontiers in Psychology,* Original Research vol. 7, no. 1005, 2016-July-05 2016.

[137] J. A. Rogers, T. Someya, and Y. Huang, "Materials and Mechanics for Stretchable Electronics," *Science,* vol. 327, no. 5973, pp. 1603-1607, 2010.

[138] A. Kuhn, T. Keller, M. Lawrence, and M. Morari, "The influence of electrode size on selectivity and comfort in transcutaneous electrical stimulation of the forearm," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 18, no. 3, pp. 255-62, Jun 2010.

[139] J. D. Gomez-Tames, J. Gonzalez, and W. Yu, "A simulation study: Effect of the inter-electrode distance, electrode size and shape in Transcutaneous Electrical Stimulation," in *2012 Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, 2012, pp. 3576-3579.

[140] P. Lepola *et al.*, "Screen-printed EEG electrode set for emergency use," *Sensors and Actuators A: Physical,* vol. 213, pp. 19-26, 2014.

[141] S. Flurin *et al.*, "Skin Conformal Polymer Electrodes for Clinical ECG and EEG Recordings," *Advanced Healthcare Materials,* vol. 7, no. 7, p. 1700994, 2018.

[142] Y.-H. Chen *et al.*, "Soft, Comfortable Polymer Dry Electrodes for High Quality ECG and EEG Recording," *Sensors,* vol. 14, no. 12, p. 23758, 2014.

[143] C. Pylatiuk *et al.*, "Comparison of surface EMG monitoring electrodes for long-term use in rehabilitation device control," in *2009 IEEE International Conference on Rehabilitation Robotics*, Kyoto, Japan, 2009, pp. 300-304.

[144] B. Stephens-Fripp, V. Sencadas, R. Mutlu, and G. Alici, "Reusable Flexible Concentric Electrodes Coated With a Conductive Graphene Ink for Electrotactile Stimulation," (in English), *Frontiers in Bioengineering and Biotechnology,* Original Research vol. 6, no. 179, 2018-December-03 2018.

[145] T. N. Cornsweet, "The Staircase-Method in Psychophysics," *The American Journal of Psychology,* vol. 75, no. 3, pp. 485-491, 1962.

[146] A. N. K. Phinyomark, R.; Scheme, E., "Feature Extraction and Selection for Myoelectric Control Based on Wearable EMG Sensors," *Sensors,* vol. 18, no. 5, 2018.

[147] C. L. Van Doren, "Contours of equal perceived amplitude and equal perceived frequency for electrocutaneous stimuli," *Perception & Psychophysics,* journal article vol. 59, no. 4, pp. 613-622, June 01 1997.

[148] F. Y. Oda H, "Concentric electrodes for producing acupuncture-like anesthetic effects.," *The Tohoku Journal of Experimental Medicine,* vol. 160, no. 3, pp. 169-175, 1990.

[149] J. L. Mason and N. A. M. MacKay, "Pain Sensations Associated with Electrocutaneous Stimulation," *IEEE Transactions on Biomedical Engineering,* vol. BME-23, no. 5, pp. 405-409, 1976.

[150] C. Frigo, M. Ferrarin, W. Frasson, E. Pavan, and R. Thorsen, "EMG signals detection and processing for on-line control of functional electrical stimulation," *Journal of Electromyography and Kinesiology,* vol. 10, no. 5, pp. 351-360, 2000/10/01/ 2000.

[151] Y. M. Chi, T. P. Jung, and G. Cauwenberghs, "Dry-Contact and Noncontact Biopotential Electrodes: Methodological Review," *IEEE Reviews in Biomedical Engineering,* vol. 3, pp. 106-119, 2010.

[152] S. Yao and Y. Zhu, "Nanomaterial-Enabled Dry Electrodes for Electrophysiological Sensing: A Review," *JOM, The Journal of The Minerals, Metals & Materials Society (TMS),* vol. 68, no. 4, pp. 1145-1155, April 01 2016.

[153] J. Cornman, A. Akhtar, and T. Bretl, "A portable, arbitrary waveform, multichannel constant current electrotactile stimulator," in *2017 8th International IEEE/EMBS Conference on Neural Engineering (NER)*, 2017, pp. 300-303.

[154] J. M. Fontana, R. O'Brien, E. Laciar, L. S. Maglione, and L. Molisani, "Vibrotactile Stimulation in the Upper-Arm for Restoring Individual Finger Sensations in Hand Prosthesis," *Journal of Medical and Biological Engineering,* vol. 38, no. 5, pp. 782-789, 2018.

[155] E. B. Goldstein, "The Cutaneous Sense," in *Sensory and Perception*, J.-D. H. a. J. A. Perkins, Ed. 8 ed.: Wadsworth, Cengage Learning, 2010, pp. 329-354.

[156] C. E. Stepp and Y. Matsuoka, "Object manipulation improvements due to single session training outweigh the differences among stimulation sites during vibrotactile feedback," (in eng), *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 19, no. 6, pp. 677-85, Dec 2011.

[157] M. Markovic, M. A. Schweisfurth, L. F. Engels, D. Farina, and S. Dosen, "Myocontrol is closed-loop control: incidental feedback is sufficient for scaling

the prosthesis force in routine grasping," *Journal of NeroEngineering and Rehabilitation,* vol. 15, no. 1, p. 81, Sep 3 2018.

[158] C. Hartmann, "Electrocutaneous stimulation to close the loop in myoelectric prosthesis control," Doctor rerum naturalium, der Georg-August University School of Science, der Georg-August-Universitat G¨ottingen, 2016.

[159] D. Blustein, A. Wilson, and J. Sensinger, "Assessing the quality of supplementary sensory feedback using the crossmodal congruency task," *Scientific Reports,* vol. 8, 2018.

[160] S. Cheng, A. Yi, U. Tan, and D. Zhang, "Closed-Loop System for Myoelectric Hand Control Based on Electrotactile Stimulation," in *2018 3rd International Conference on Advanced Robotics and Mechatronics (ICARM)*, 2018, pp. 486-490.

[161] R. S. Johansson and G. Westling, "Roles of glabrous skin receptors and sensorimotor memory in automatic control of precision grip when lifting rougher or more slippery objects," *Exp Brain Res,* vol. 56, 1984.

[162] R. S. Johansson, "Sensory Input and Control of Grip," in *Novartis Foundation Symposium 218 - Sensory Guidance of Movement*, pp. 45-63.

[163] M. Controzzi, F. Clemente, N. Pierotti, M. Bacchereti, C. Cipriani, and S. Superiore, "Evaluation of hand function trasporting fragile objects: the virtual eggs test," in *Myoelectric Control Symposium*, 2017.

[164] E. Mastinu *et al.*, "Grip control and motor coordination with implanted and surface electrodes while grasping with an osseointegrated prosthetic hand," *Journal of neuroengineering and rehabilitation,* vol. 16, no. 1, p. 49, 2019.

[165] A. W. Shehata, L. F. Engels, M. Controzzi, C. Cipriani, E. J. Scheme, and J. W. Sensinger, "Improving internal model strength and performance of prosthetic hands using augmented feedback," *Journal of neuroengineering and rehabilitation,* vol. 15, no. 1, p. 70, 2018.

[166] 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," (in eng), *Journal of Biomechanics,* vol. 49, no. 5, pp. 786-792, 2016.

[167] F. Vecchi, C. Freschi, S. Micera, A. Sabatini, P. Dario, and R. Sacchetti, "Experimental evaluation of two commercial force sensors for applications in biomechanics and motor control," in *5th Annual Conference of International Functional Electrical Stimulation Society*, 2019.

[168] T. Nilsen, C. S. Hermann M Fau - Eriksen, H. Eriksen Cs Fau - Dagfinrud, P. Dagfinrud H Fau - Mowinckel, I. Mowinckel P Fau - Kjeken, and I. Kjeken, "Grip force and pinch grip in an adult population: reference values and factors associated with grip force," (in eng), *Scandinavian Journal of Occupational Therapy,* no. 1651-2014 (Electronic).

[169] W. M. Murray and W. R. Miller, *The bonded electrical resistance strain gage: an introduction*. Oxford university press, 1992.

[170] B. Stephens-Fripp, E. Wallace, T. Searle and G.l Alici "Design of a Sensorised Object to Test Sensory Feedback for Prosthetic Hands," presented at the

IEEE/ASME International Conference on Advanced Intelligent Mechatronics, Hong Kong, China, 2019.

[171] C. E. Stepp, Q. An, and Y. Matsuoka, "Repeated training with augmentative vibrotactile feedback increases object manipulation performance," *PLoS One,* vol. 7, no. 2, p. e32743, 2012.

[172] A. D. Roche, H. Rehbaum, D. Farina, and O. C. Aszmann, "Prosthetic Myoelectric Control Strategies: A Clinical Perspective," *Current Surgery Reports,* vol. 2, no. 3, p. 44, 2014/01/25 2014.

[173] N. V. Iqbal, K. Subramaniam, and S. A. P, "A Review on Upper-Limb Myoelectric Prosthetic Control," *IETE Journal of Research,* vol. 64, no. 6, pp. 740-752, 2017.

[174] N. Jiang, S. Dosen, K. Muller, and D. Farina, "Myoelectric Control of Artificial Limbs; Is There a Need to Change Focus?," *IEEE Signal Processing Magazine,* vol. 29, no. 5, pp. 152-150, 2012.

[175] A. Chadwell, L. Kenney, S. Thies, A. Galpin, and J. Head, "The Reality of Myoelectric Prostheses: Understanding What Makes These Devices Difficult for Some Users to Control," *Frontiers in Neurorobotics,* vol. 10, p. 7, 2016.

[176] M. Zafar and C. L. V. Doren, "Effectiveness of supplemental grasp-force feedback in the presence of vision," *Medical and Biological Engineering and Computing,* vol. 38, 2000.