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Touching on elements for a non-invasive sensory feedback system for use in a prosthetic hand

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Doctoral Dissertation

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Abstract

Hand amputation results in the loss of motor and sensory functions, impacting activities of daily life and quality of life. Commercially available prosthetic hands restore the motor function but lack sensory feedback, which is crucial to receive information about the prosthesis state in real-time when interacting with the external environment. As a supplement to the missing sensory feedback, the amputee needs to rely on visual and audio cues to operate the prosthetic hand, which can be mentally demanding. This thesis revolves around finding potential solutions to contribute to an intuitive non-invasive sensory feedback system that could be cognitively less burdensome and enhance the sense of embodiment (the feeling that an artificial limb belongs to one's own body), increasing acceptance of wearing a prosthesis.

A sensory feedback system contains sensors to detect signals applied to the prosthetics. The signals are encoded via signal processing to resemble the detected sensation delivered by actuators on the skin.

There is a challenge in implementing commercial sensors in a prosthetic finger. Due to the prosthetic finger's curvature and the fact that some prosthetic hands use a covering rubber glove, the sensor response would be inaccurate. This thesis shows that a pneumatic touch sensor integrated into a rubber glove eliminates these errors. This sensor provides a consistent reading independent of the incident angle of stimulus, has a sensitivity of 0.82 kPa/N, a hysteresis error of 2.39±0.17%, and a linearity error of 2.95±0.40%.

For intuitive tactile stimulation, it has been suggested that the feedback stimulus should be modality-matched with the intention to provide a sensation that can be easily associated with the real touch on the prosthetic hand, e.g., pressure on the prosthetic finger should provide pressure on the residual limb. A stimulus should also be spatially matched (e.g., position, size, and shape). Electrotactile stimulation has the ability to provide various sensations due to it having several adjustable parameters. Therefore, this type of stimulus is a good candidate for discrimination of textures. A microphone can detect texture-elicited vibrations to be processed, and by varying, e.g., the median frequency of the electrical stimulation, the signal can be presented on the skin. Participants in a study using electrotactile feedback showed a median accuracy of 85% in differentiating between four textures.

During active exploration, electrotactile and vibrotactile feedback provide spatially matched modality stimulations, providing continuous feedback and providing a displaced sensation or a sensation dispatched on a larger area. Evaluating commonly used stimulation modalities using the Rubber Hand Illusion, modalities which resemble the intended sensation provide a more vivid illusion of ownership for the rubber hand.

For a potentially more intuitive sensory feedback, the stimulation can be somatotopically matched, where the stimulus is experienced as being applied on a site corresponding to their missing hand. This is possible for amputees who experience referred sensation on their residual stump. However, not all amputees experience referred sensations. Nonetheless, after a structured training period, it is possible to learn to associate touch with specific fingers, and the effect persisted after two weeks. This effect was evaluated on participants with intact limbs, so it remains to evaluate this effect for amputees.

In conclusion, this thesis proposes suggestions on sensory feedback systems that could be helpful in future prosthetic hands to (1) reduce their complexity and (2) enhance the sense of body ownership to enhance the overall sense of embodiment as an addition to an intuitive control system.

Key words: Sensory feedback, amputation, referred sensation, non-invasive feedback, upper limb prosthetics, body ownership

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To my family and friends

Ever tried? Ever failed? Doesn't matter.
Try again. Fail again. Learn and improve.
- Samuel Beckett

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Cover illustration
Illustration of a phantom hand with the skin's mechanoreceptors and technical solutions for supplementary sensory feedback with a microcontroller board in the background.

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List of papers

Included

I. **Touch on predefined areas on the forearm can be associated with specific fingers: towards a new principle for sensory feedback in hand prosthesis**
U. Wijk*, **P. Svensson***, C. Antfolk, IK. Carlsson, A. Björkman, B. Rosén
Published in: Journal of Rehabilitation Medicine. 2019;51(3):209-16.
* Authors 1 and 2 share first authorship

Author's contribution: part of planning the experiments, developed all the hardware and software, part of carrying out the experiments, part of the data analysis, part of writing.

II. **Characterization of pneumatic touch sensors for a prosthetic hand**
P. Svensson, C. Antfolk, N. Malešević , F. Sebelius
Published in: IEEE Sensors Journal. 2020 Apr 10;20(16):9518-27.

Author's contribution: planned the experiments, assembled the experimental setup, carried out the experiment, major part of the data analysis and writing.

III. **Electrotactile feedback for the discrimination of different surface textures using a microphone**
P. Svensson, C. Antfolk, A. Björkman, N. Malešević
Published in: Sensors. 2021 Jan;21(10):3384.

Author's contribution: part of planning the experiment, assembled the experimental setup and recruited participants, designed and implemented the algorithm, carried out the experiment, data analysis and writing.

IV. **The Rubber Hand Illusion evaluated using different modalities**
P. Svensson, N. Malešević, A. Björkman, U. Wijk, C. Antfolk
Manuscript

Author's contribution: planned the experiment, developed all the hardware and software, recruited participants for the experiment, carried out the experiment, data analysis and writing.

Related

Papers listed below have been presented at conferences and are not included in this thesis.

I. **A review of invasive and non-invasive sensory feedback in upper limb prostheses**
P. Svensson*, U. Wijk*, A. Björkman, C. Antfolk
Published in: Expert Review of Medical Devices. 2017 May;14(6):439-447.
* Authors 1 and 2 share first authorship

II. **Electrotactile feedback for the discrimination of different surfaces textures: Pilot study**
P. Svensson, C. Antfolk, A. Björkman, N. Malešević
Published in: Artifical Organs, Abstracts from the IFESS 2021 conferences. 2022 Feb

III. **Multi-modality Sensory Feedback**
P. Svensson, Huaiqi Huang, C. Antfolk
Conference abstract for oral presentation in ISPO 2017

Contents

Acronyms and Abbreviations

ADL Activity of daily living

AR Augmented reality

CNS Central nervous system

DIP Distal interphalangeal

DOF Degree of freedom

EMG Electromyography

FINE Flat interface nerve electrode

FIR Finite impulse response

fMRI Functional magnetic resonance imaging

LIFE Longitudinal intra-fascicular electrode

M1 Primary motor cortex

MCP Metacarpophalangeal

PHM Phantom hand map

PIP Proximal inter-phalangeal

PLP Phantom limb pain

PNS Peripheral nervous system

RLP Residual limb pain

S1 Primary sensory cortex

SMA Shape memory alloy

TIME Transverse intra-fascicular multichannel electrode

TMR Targeted muscle reinnervation

TSR Targeted sensory reinnervation

VR Virtual reality

Chapter 1

Introduction

Touch plays a crucial role in socially connecting with others and one's surroundings. It is believed to be the first sense to develop when a child is born and the last sense to function when people die at an advanced age. Touch is described as *"...the core of sentience, the foundation for communication with the world around us, and probably the single sense that is as old as life itself"* [1]. It has also been stated that touch might be the most basic form of communication [2].

After an amputation, the hand's functional capabilities and sensory connection are severed, causing a devastating burden for the individual. From 2015 to 2020, approximately 42.3 per 100,000 inhabitants in Sweden underwent an upper limb amputation, of whom 40.3 underwent wrist and hand level amputation, which is the most common amputation level in Sweden. In comparison, 0.7 persons per 100,000 inhabitants had forearm amputations and 1.3 per 100,000 inhabitants had upper arm and shoulder amputations [3]. In the International Classification of Diseases and Related Health Problems 11th revision (ICD-11), amputation is categorised as traumatic and congenital: traumatic amputation of wrist or hand, forearm, and shoulder or upper arm (NC59, NC38, and NC18) as well as upper limb reduction defects (LB99) [4]. There are various causes of limb loss resulting from traumatic injury or planned surgical intervention to hinder the spread of infectious disease. An amputation can also be congenital, meaning that the person was born without a limb.

A higher amputation level leads to a greater decrease in the functionality of the hand, thus increasing the difficulty of performing basic to high-level daily living activities [5]. Upper limb amputees tend to have higher levels of activity restriction and more significant body image disturbance than lower limb amputees [6]. Many advanced prosthetic hands can to some extent replace the lost motor function by enabling multiple grasp possibilities and high degrees of freedom (DoF). However, the prosthesis user needs to rely on his/her vision or the sound made by the prosthesis to compensate for the lack of sensation to perceive information about grip force and the position of the prosthetic fingers. However, it takes many years and a large amount of training before the amputee learns to use vision and sound as a substitute when controlling the prosthesis [7].

Tactile afferents most likely play a significant role in adapting the grip when handling objects, which was observed during local anaesthesia of the thumb and index finger [8]. Hence, the hypothesis that adding sensory feedback to prosthetic hands would improve the motor control of the prosthesis is a natural conclusion. Sensory feedback has been especially shown to be of assistance, especially in complex/demanding tasks, as opposed to routine/simple grasping tasks [9].

Several surveys also found sensory feedback to be a desired feature of electric prostheses [7, 10–12]. However, in a study of the use of electrotactile feedback to aid amputees with force control and effects of short- and long-term learning, it was shown that the benefit of feedback became redundant with training [13]. This suggest that feedback is only beneficial in an initial phase of prosthesis use for learning the feedforward control [13]. Other studies have found that no objective improvement was observed during grasp performance with sensory feedback, however, from a subjective perspective, the users reported that such feedback was helpful [9, 14].

1.1 Aim and scope

This thesis explores the possibilities of providing a straightforward and intuitive interpretation of sensory feedback.

The following research questions are addressed in this thesis:

1. Can touch on predefined areas on the forearm be associated with specific fingers after a training period?

2. Does a simple pneumatic sensor meet the characterisation criteria for implementation in a prosthetic hand?

3. Could the information recorded by a microphone be processed to provide the user with an intuitive sensation about the characteristic of a structure through electrical stimulation?

4. Which sensory substitution technique gives the strongest feeling of body ownership?

This thesis's scope focuses on non-invasive sensory feedback, and the experiments were conducted primarily on participants with intact limbs to evaluate whether the studies could be applicable for amputees. Lastly, the systems in Papers I and III were in "proof of concept" form; hence, the systems were not sized to fit in commercial prosthetic hands or sockets, but the size can be minimised by replacing the larger motors with smaller ones and by customising the microcontroller board.

1.2 Thesis outline

Chapter 2 serves as background and explains the somatosensory system and complications after an amputation. **Chapter 3** provides an overview of commercially available prostheses, recent research prototypes, and a brief introduction on how prostheses are controlled. **Chapter 4** provides an overview of commercial sensor technology that is commonly used in research on prosthetic hands and reviews some sensory feedback systems that have been used in research. **Chapter 5** gives a summary of the included papers. **Chapter 6** discusses and evaluates the main findings of the papers.

1.3 Ethics

In Paper I, the study was approved by Lund's regional ethics review board (Dnr 2012/778). In Paper II, a sensor was characterised and therefore no participants were included in this study. Paper III was approved by the Swedish Ethical Review Authority (DNR 2020-03937). The last study is presented as a manuscript and was approved by the Swedish Ethical Review Authority (DNR 2021-03630).

Chapter 2

The somatosensory system

The somatosensory system is divided into two subsystems that detect 1) mechanical stimuli (e.g., light touch, vibration, pressure) and 2) temperature and painful stimuli. Together, the subsystems convey information from the skin, muscles, and joints about the external environment through touch, external and internal forces that act upon the body (body position and movement), and also detect harmful situations. External stimuli activate a diverse population of cutaneous mechanoreceptors (sensory neurons) that innervate the skin. Several ascending parallel pathways carry the mechanosensory information through the spinal cord, brainstem, and thalamus to reach the primary somatosensory cortex (S1) [15].

Signals from the outside world make their way to the brain using specific routes that can be defined using the following sequence (Fig. 2.1). It starts at the receptor level, where receptors (exteroceptors and proprioceptors) respond to a stimulus. For example, exploring the external environment with the hand, stimulation is applied to the superficial skin activating cutaneous receptors in the skin. The cells carry sensory information through the dorsal root to the dorsal column of the cervical spinal cord (which handles sensory information from the upper limb). The dorsal column is divided into two sections, the fasciculus cuneatus and fasciculus gracilis, where the former is responsible for carrying sensory information from the upper limbs. The cells in the spinal cord are called first-order neurons. The first-order neurons will travel up to the brain stem, which contains three parts: the medulla, pons, and midbrain. In the medulla, the first-order neurons will synapse with the second-order neurons. After travelling through the brain stem, the second-order neurons will synapse with the third-order neurons in the thalamus, which carries the sensory information to the primary sensory cortex (S1) [16]. S1 is divided into different regions, which correspond to the different parts of the body. In this case, the sensory information is conveyed to the hand area in S1. In order to create relevant information about the stimulus, the sensation properties come from a population of receptors and not only from a single sensory neuron. The information is then processed in the cerebral cortex, which forms a percept [17].

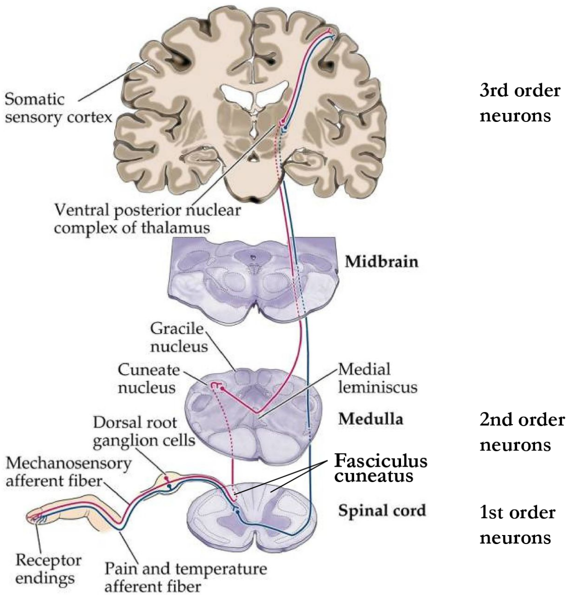

Figure 2.1: Mechanosensory pathway. (Edited picture from Neilson [18], ©2016 BMJ Publishing Group Ltd.)

2.1 Somatosensory receptors

The somatosensory system comprises cutaneous/tactile and kinesthetic submodalities, where the former receive sensory input from receptors embedded in the skin. In contrast, receptors (proprioceptors) within muscles, tendons, and joints send sensory information about the body's position and movement by sensing how stretched an organ is. The cutaneous receptors are categorised according to stimulus type: mechanoreceptors, thermoreceptors, nociceptors, thermal stimuli, and pain (free nerve endings). Most of the free nerve endings are unmyelinated, which means the nerve impulses are conducted slowly, compared to myelinated fibres. Furthermore, some nerve endings are wrapped around hair follicles to detect light touch and are activated when the hair bends [16].

There are different encapsulated cutaneous mechanoreceptors: Merkel's discs, Meissner's corpuscles, Pacinian corpuscles, and Ruffini's corpuscles (Fig. 2.2a). The receptors are embedded in the two layers of the skin, the epidermis (outer layer) and dermis (underlying layer). On the palmar surface of the hand (hairless/glabrous skin), the epidermis layer is folded into an array of ridges and has a greater tactile sensitivity than hairy skin [17]. These ridges amplify the signals to the mechanoreceptors, which are densely packed in the glabrous skin [19]. Pacinian corpuscles are the largest of the receptors and are located deep in the dermis. Ruffini's corpuscles are the second largest and can be found in the dermis in both hairy and glabrous skin, and can also be found in ligaments and tendons. Meissner's corpuscles are located just beneath the epidermis and can be found on the ridges of the glabrous skin. Merkel's discs are located within the epidermis [20]. Meissner's corpuscles account for about 40% of the cutaneous receptors in the hand, Pacinian corpuscles 10–15%, Merkel's discs 25%, and Ruffini's corpuscles 20%, where the slowly adapting receptors are particularly dense in the fingertips [15].

The receptors have different physical properties that determine the type of mechanical pressure the receptors are most susceptible to and are categorised according to the adaption rate and the size of the receptive field (Fig. 2.2b): type I (small receptive fields), Type II (large receptive fields), rapidly adapting (RA, no static response), and slowly adapting (SA, continues firing in response to a constant stimulus) [21]. The RA receptors adapt rapidly to the onset and offset of stimulation during a grasping task (vertical impact) and become silent during continuous deformation (steady-state). In other words, RA receptors sense motion on the skin, including vertical impact and lateral motion (such as stroking, rubbing, or palpation). SA receptors are activated and fire action potentials during static force [17].

Pacinian and Ruffini's corpuscles, located deep in the skin, have a larger receptive field than Meissner's corpuscles and Merkel's discs, which are located in the superficial layer of the skin. Comparing the receptive fields between the mechanoreceptors with a rapid indentation (0.5 mm) on the fingerpad, Pacinian corpuscles have the largest receptive field area (covering the entire finger or hand), followed by Ruffini's corpuscles (60 mm^2), Meissner's corpuscles (22 mm^2), and lastly Merkel's discs (9 mm^2) [17]. Mechanoreceptors with a large receptive field give a more intense response when a contact point is applied to the centre of the receptive field and decreases as the contact point moves away from the centre [2]. Those mechanoreceptors can respond to stimulation applied remotely from the receptive field centre. Stimulating an area larger than a receptor's receptive field will also activate the adjacent receptors; the total number of activated receptors depends on the size of the stimulation [17]. The size of the receptive field and the density of receptors varies throughout the body. There is a higher density of receptors in the fingertips, where the receptive field is also smaller and becomes sparser, moving in the distal-proximal direction of the finger. The receptors' receptive field in the fingertips and palm is 1–2 mm and 5–10 mm in diameter, respectively. The numerous and smaller receptors in the fingertips provide a means for better spatial acuity, the ability to discriminate two stimuli close in space. When two stimuli are too close together, the stimuli will blend and be felt as if they were one stimulus. The two-point discrimination is smallest on the fingertips, discriminating two-point stimuli 2 mm apart, while on the forearm the distance needs to be at least 40 mm between the two stimuli [15].

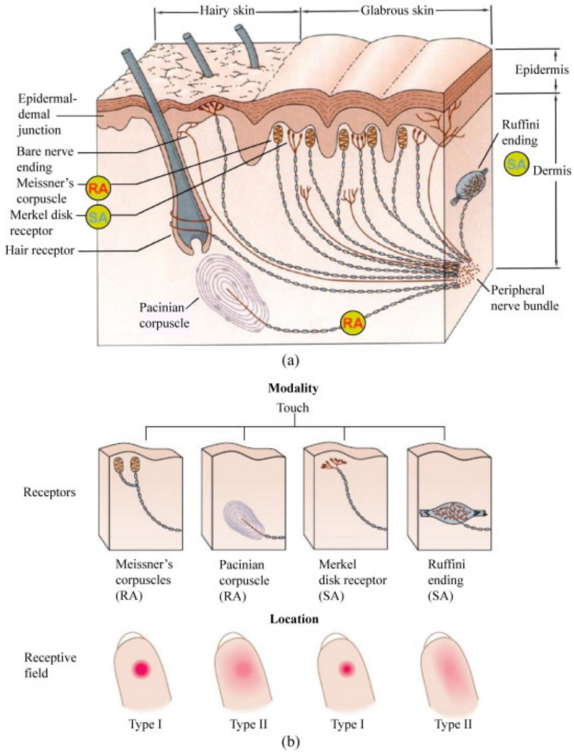

Figure 2.2: a) Skin structure: epidermis and dermis. Mechanoreceptors can be seen in different layers. b) Receptive field. Type I mechanoreceptors (Meissner's corpuscles and Merkel's discs) have small receptive fields, while type II (Pacinian corpuscles and Ruffini's corpuscles) have large receptive fields. (Reused picture from Ding and Bhushan [22], ©2016 Elsevier Inc.)

To summarise, Meissner's corpuscles (RA-I) are sensitive to dynamic skin deformation, making them susceptible to 30–50 Hz (skin moving across textured objects). The rapidly adapting feature makes Meissner's corpuscles insensitive to continuous deformation and an ineffective for detecting local spatial discontinuities. Similar to Meissner's corpuscles, Pacinian corpuscles (RA-II) respond to dynamic mechanical stimuli but have a lower threshold and adapt more rapidly than Meissner's corpuscles. Pacinian corpuscles only respond to high frequencies (250–350 Hz). These features allow Pacinian corpuscles to discriminate fine textures or other stimuli that produce high-frequency skin vibrations. Pacinian corpuscles are excited when in contact or breaking contact with an object. Both Merkel's discs (SA-I) and Ruffini's corpuscles (SA-II) are slowly adapting receptors. Stimulating Merkel's discs produces a light touch sensation and detects static deformation from edges, shapes, and rough textures. Ruffini's corpuscles detect cutaneous stretching from movements in digits or limbs [15]. Ruffini's corpuscles can also respond to a tangential shear strain on the skin during object manipulation.

2.2 The central nervous system

The central nervous system consists of the brain and the spinal cord. The nerve impulses from the receptors travel in afferent nerve fibres and form dorsal roots in the spinal cord. The neurons' axons synapse with the dorsal horn in the grey matter when entering the spinal cord. The information reaches the thalamus from the spinal cord, which contains a large number of nuclei. Each nucleus receives and projects fibres to a specific region of the cortex. The thalamus can be seen as a gateway to the cerebral cortex and a relay station. The sensory information received from the body is filtered in the thalamus, and the selected information travels to the primary somatosensory cortex [15,23]. The cortex processes the specific stimulus's localisation and generates the perception.

The entire body is spatially represented in the cortex, creating a somatotopic map. The proportions of each representation depend on the receptor density in the skin of each represented area [16]. Therefore, the hand, face, mouth, and foot representations are larger than the trunk and legs. This neurological map of the anatomical portions of the body is referred to as the sensory homunculus (Fig. 2.3). The map is constantly changing and is different for each individual; a violinist could have more extensive fingertip representations than a trumpet player who would have more significant lip representation in the cortex [24]. A sensory signal from the left side of the body decussates (crosses over) in the medulla and enters the right hemisphere (the contralateral cerebral cortex), and vice versa [25].

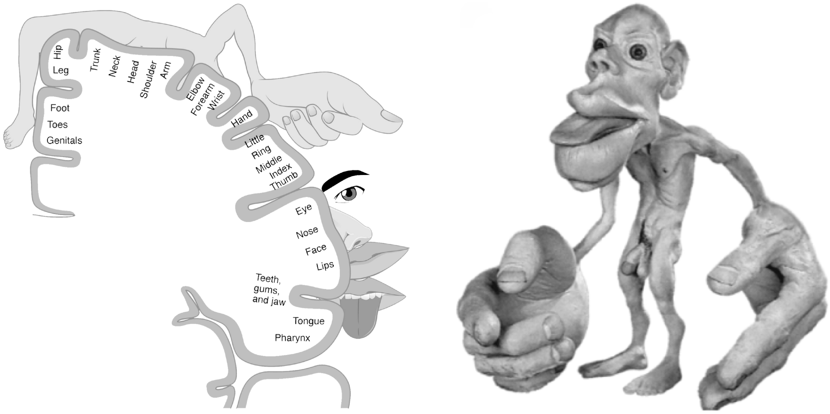

Figure 2.3: Somatosensory system located in the postcentral gyrus of the parietal lobe. From the drawn plane in the somatosensory cortex, a picture of the somatotopic representation of body parts is seen. The sensory cortex represents areas occupied by the body parts and is based on a neurological map (photo from OpenStax College, distributed under a CC BY 3.0 license), which can be illustrated as a caricature of a human (picture from Kasumyan [26], ©2011 Springer Nature).

2.3 Amputation consequences

After limb amputation, several physical and psychosocial challenges confront the individual, such as disruptions in physical functioning, changes in lifestyle (occupational and social), pain, adapting to prosthetic use, and alterations in body image [27].

2.3.1 Amputation levels

The upper extremities are composed of several components, e.g., neurovascular bundles, muscles, and bones [28]. During surgery for a traumatic amputation, it is of utmost importance that the surgeon chooses a surgical technique that would fit the patient's need, such as to help the post-operation healing, aim for an aesthetic appearance, and to increase the possibilities of wearing a prosthetic device [29]. An amputation can be performed on different levels (Fig. 2.4). Metacarpal amputation is the most distal amputation, which removes the digits (phalanges) or the metacarpal bones. The next level is a transcarpal amputation, which amputates the radius and the metacarpal bones and sections the ulnar and median nerves. This type of amputation preserves the extension and flexion of the wrist.

Wrist disarticulation removes all the carpal bones. Bony protrusions can be minimised by resecting the radial and ulnar styloid to make prosthetic use more comfortable [28]. The hand's function, such as enabling the pronation-supination of the forearm, can be restored by a special socket that reproduces the wrist joint anatomy [2].

The forearm has twenty muscles divided into two groups: the intrinsic muscles pronate and supinate the radius and ulna, and the extrinsic muscles flex and extend the hand's digits. The forearm has three primary nerves: the median, ulnar, and radial nerves. A transradial amputation (forearm level/below elbow) tries to preserve forearm lenght and could therefore provide for more movement, such as pronation and supination [28].

In elbow disarticulation, the elbow is separated, meaning the entire lower arm is removed, whereas the humerus is left, which preserves the humerus rotation [28]. Below elbow amputations are classified into different lengths depending on the residual limb length. The rotation function is proportional to the residual limb length; a short residual limb is often limited in power and motion [2].

An amputation that is performed above elbow level is called a transhumeral amputation, where the length of the residual limb is preserved as much as possible for a better prosthetic fit (minimum of 5 to 7 cm). If less than 30% of the residual limb remains, then the amputation is treated as a shoulder-disarticulation [2]. Myoplasty is used to detach part of a muscle, which can be done on the biceps and triceps to provide a more robust prosthetic control and ensure stronger myoelectric signals [28].

During shoulder disarticulation, the entire arm is removed. However, if possible, the surgeon will leave as much of the humerus intact to better fit the socket and the residual limb. A higher-level amputation is called a forequarter; in some cases, it has been necessary to remove the clavicle and scapula [2].

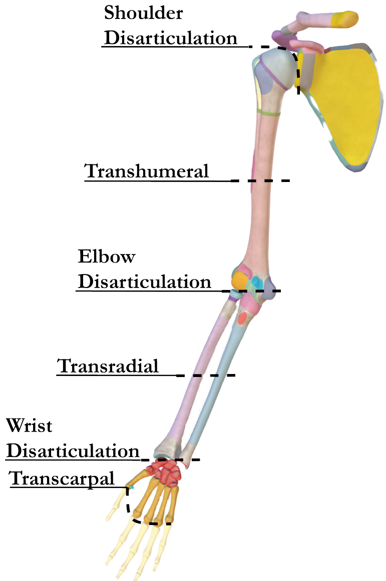

Figure 2.4: Amputation levels in upper limb amputees. (Figure from 3D models, ©2021 BioDigital Inc.)

2.3.2 Physical changes

Somatosensory receptors

After limb amputation, receptors in the extremities are lost, causing damage to the peripheral first-order afferents and leaving the second-order afferents intact. If damage occurs on the nerve fibres, the entire neuron may die if the damage is severe. However, if the cell body is left undamaged, the peripheral nerve fibres can regenerate at approximately 1.5 mm/day. Surviving Schwann cells form a regeneration tube that guides the nerve fibre sprouts from the injury to find their way to their original target across the injury gap [16].

Central nervous system

The brain can rewire the nervous system according to its activation history to adapt the nervous system to an altered environment. This phenomenon is called neuroplasticity [2] and is common post-amputation. It is mentioned that sensory deprivation and altered use are the main factors in inducing brain plasticity after an amputation [30]. Studies have shown a reorganisation in S1, where the somatotopically adjacent face territory has progressed into the deafferented territory [31]. Other studies identified only small partial shifts in the lip territory, contralateral to the missing hand, that moves towards, but does not invade, the deafferented S1 [32]. Valyear et al. [33] performed functional magnetic resonance imaging (fMRI) on unilateral upper extremity amputees as well as intact controls in order to map the sensory cortical representation on the intact hand and lower face. They showed great activity in S1 of the amputated limb during cutaneous stimulation of the intact hand, but not when stimulating the lower face. Furthermore, Makin et al. [30] studied plasticity using fMRI on both traumatic and congenital unilateral amputees and intact controls. Their results showed that the amputees who relied more on their intact hand had increased representation of the ipsilateral intact limb in the deprived cortex.

Pain

Different amputation-related pains have been observed in upper limb amputees, such as phantom limb pain (PLP, which will be addressed in section 2.4.1), residual limb pain (RLP), back pain, neck pain, and pain in the non-amputated limb. PLP is the most commonly experienced pain (experienced by 79% [34], 83% [35] of individuals), RLP (71% [34], 61% [35]), back pain (52% [34], 64% [35]), neck pain (43% [35]) and pain in the non-amputated limb (33% [34, 35]). RLP is a musculoskeletal pain that affects bones, joints, ligaments, tendons, or muscles due to remodelling scars, tissues, fascia, and muscles. Other complications that might occur are oedema (swelling caused by a build-up of fluid in body tissue), contracture formation, body asymmetry, skin breakdown, aesthetic acceptability of the residual limb, and the formation of neuromas [28]. However, a post-amputation process involving rehabilitation and regular check-ups might help the amputee to overcome and avoid certain complications.

A cross-sectional survey of 914 participants, of which 10.9% had upper limb amputations, found that 64% of the amputees experienced back pain [34]. However, there is a lack of information regarding whether the amputees regularly perform physical activities, which could have an impact on their back pain, as it is known that specific strength exercises can alleviate back pain [36]. In the case of a unilateral limb loss, one can assume that there could be an asymmetric load on the back, meaning that one side of the back is overworked, resulting in a strained muscle, while the other side could become weaker. Wearing a prosthesis could alleviate the asymmetrical load on the back.

2.3.3 Psychosocial challenges

Body image is an individual's perception of their own body and is a central component of one's self-concept. After an amputation, changes in body image are a common issue when adjusting to limb loss and accepting a prosthesis [27]. Social experience plays a crucial role in forming and building a new body image and self-concept [37]. Some individuals manage their altered body and disability with transient distress, while others develop a negative attitude toward themselves, lasting into the long-term. For individuals with congenital limb absence, the absent limb fully integrates into one's body image, as the limb is absent during self-concept formation. Instead, introducing prosthetics to these individuals might trigger a body image crisis, since wearing a prosthesis might be interpreted as an attack on a well-established healthy body image. Pain is another factor that has an impact on body image, since pain could be a constant reminder of the disability [27]. Pucher et al. [38] found a connection between PLP and body image, where individuals who adjusted to the limb loss had significantly lower levels of PLP than individuals who did not cope with their limb loss. Wearing a prosthesis could help with the adjustment of body image.However, the individual needs to adapt to two distinct aspects of body image: with and without a prosthesis. The former is often experienced in a social environment and the latter in a private environment. If the functionality and aesthetics of the prosthesis do not meet the threshold of the required satisfaction, there can be difficulties in body image adjustment. It has been shown that lower limb amputation is more socially accepted since wearing a leg prosthesis is more discreet and can be concealed, compared to an upper limb prosthesis. The social experience is also crucial when forming a new body image and self-concept. If individuals with limb deficits experience social discomfort in how others view them during the initial period following amputation, the experience can be internalised [27].

2.4 Phantom limb sensations (PLS)

After amputation, some patients perceive sensation as originating from the missing body part. Such sensations are commonly referred to as phantom limb sensations (PLS) [39]. Amputees experience PLS very differently, across the whole spectrum of somatosensory sensations, including pain, and some amputees experience being able to control the motion of the phantom limb [39]. Amputees can also experience sensations from some parts of the phantom limb (e.g., a specific finger or the palm). Phantom limbs can also assume different postures, such as the normal habitual posture or an abnormal posture that is anatomically impossible. Telescoping is another common sensation, where the phantom limb is perceived to be shorter or longer than an intact limb [40]. For some amputees, stimulating different areas on the residual limb elicits sensations in the phantom hand. These sensations are known as referred sensations and can be either non-painful or painful (Fig. 2.5) [41]. A possible explanation for the occurrence of PLS is due to cortical circuit rearrangement [31], where adjacent areas projects to the cortical territory of afferents from the amputated hand. It has also been shown that by touching the face, mis-sensations have been felt on the residual limb [20, 42].

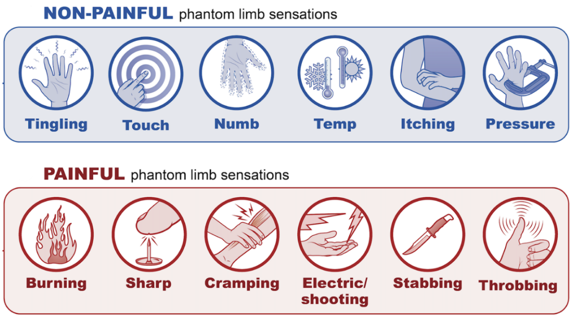

Figure 2.5: Non-painful and painful PLS (reused from Schone et al. [41], ©2022 Journal of Neurology, Neurosurgery & Psychiatry. CC-BY-NC 4.0 license)

2.4.1 Phantom limb pain (PLP)

PLP can be categorised into two groups, neuropathic PLP and nociceptive PLP; the latter is clinically termed neuroma pain, and refers to pain which arises during stimulation of nociceptive fibres [43] and the former is commonly related to non-nociceptive pain and is less understood than neuroma pain. Neuropathic phantom limb pain can be defined as: "PLP is pain perceived as arising from the missing limb due to sources other than stimulation of nociceptive neurons that are used to innervate the missing limb [43]." At the time of writing, the theory of PLP is still confounding and the mechanisms are not well understood. PLP is described as abnormal pain caused by a pathological nervous system, with incongruous explanations compared to neuroma pain caused by activated nociceptive fibres. Gaining an understanding of the different types of pain aids in finding a suitable treatment for the patient [44].

A common theory posits that PLP is positively related to a reorganisation in the primary somatosensory cortex in traumatic patients with PLP, compared to traumatic patients without PLP and congenital amputees [45]. Furthermore, greater remapping of the cortex showed a positive relationship to a high pain intensity [46].

In a study by [47], unilateral amputees executed movements with their intact hands and lips and imagined a fist movement with their phantom hand. The results demonstrated that patients who suffered PLP showed a shift of the lip area into the deafferented hand area in the contralateral primary motor cortex (M1) and S1. Furthermore, there was activation in the cortical mouth representation during the imagined movement of the phantom hand. This phenomenon was probably due to the overlap of the representation of the hand, arm, and mouth. Instead of looking at neighbouring representations of intact body parts, Makin et al. [48] studied the cortical representation of the missing hand. The results showed that persistent PLP is associated with more local activity and structural integrity within the cortical representation of the missing limb. They also propose that a combination of sensory deprivation and the experience of PLP could result in cortical changes. Moreover, a disruption of inter-regional connectivity, meaning a lingering inadequacy of co-activation between the cortex representing the intact and absent limbs, might contribute to PLP. Stochastic entanglement is another theory for the genesis of PLP, where somatosensory and motor deprivation create a chaotic network state where sensorimotor processing and pain perception entangle [43]. A cohort study showed that amputees with higher amputation levels experienced more frequent bouts of PLP [49].

Several studies have shown some correlation between changes in the primary somatosensory cortex and reduction of PLP. However, it is not clear whether the changes in S1 are the cause of the PLP or a consequence thereof, and there is no solid evidence that can highlight the relationship between cortical reorganisation, PLP, and preserved hand representation [50].

Treatment to ameliorate PLP

Different methods have been used to treat or prevent PLP, such as drugs, surgery, and treatments developed at the hand of hypotheses based on plasticity. Pharmacological approaches seem to be more successful in alleviating acute neuroma pain than treating PLP.

To treat neuropathic pain, hypotheses based on plasticity have suggested reversing cortical reorganisation by restoring motor control and sensory feedback. Both visualisation and proprioception are important cues to reduce PLP. Visualisation therapies are common treatments [51] which manipulate visual feedback in such a manner that an amputee is led to believe that they see movement of the absent limb, creating kinesthetic illusion. Mirror therapy or virtual reality (VR) therapy are therapies that use visual feedback. Ramachandran and Rogers-Ramachandran [52] introduced a new device named *virtual reality box* with which to study how visual input affects phantom sensations, and subsequently noticed pain relief for some patients. This concept was adopted and is used in mirror therapy for treating PLP (Fig. 2.6a). In mirror therapy, the amputee is exposed to visual feedback using a mirror. The amputee sees the reflection of the contralateral intact limb giving the visual illusion that the amputated limb exists. Furthermore, mirror therapy is limited to unilateral amputees, and not every amputee is susceptible to conventional PLP treatments.

Augmented reality (AR) or VR can overcome limitations in mirror therapy. Activating muscles in the residual limb in conjunction with visualisation therapies has been proposed as an effective treatment [53]. This is possible with AR, which has been used in motor execution therapy to create the illusion of a restored hand that responds directly to myoelectric signals from the residual muscles. The phantom motions can be further facilitated in gamified sessions, where the individual can control a car by myoelectric pattern recognition (Fig. 2.6b) [54]. In a case study, a chronic PLP patient experienced a gradually reduced pain when receiving a treatment based on phantom motor execution using AR, while other classic treatments did not decrease his pain level. This treatment has also demonstrated a reduction in PLP in a more extensive study with 14 patients, where the improvements in pain relief remained six months after the patients' last treatment [55].

It is proposed that enlarging the residual limb representation in the cortex, with motor control and sensory feedback, could reduce PLP. It has been suggested that it is possible to avoid evoking PLP if cortical reorganization happens in a gradual and functionally manner [43]. Another study proposed that motor control and sensory feedback would be the cause of reduced PLP [56]. In this way, repeatedly activating sensorimotor circuitry without activating pain could lead to a disentanglement and weaken the connection between the sensorimotor and pain circuitry. Furthermore, treatments involving physiological motor control and somatosensory feedback would be effective without including visual feedback [43].

Frequent and extensive use of a myoelectric prosthesis is positively related to the reduction of PLP, suggesting the effect is caused by ongoing muscle training, and visual feedback [57]. Eight forearm amputees with PLP used a myoelectric prosthesis for two weeks. They received somatosensory feedback of the grip strength, which was realised by delivering electrical stimulation on their residual limb. After extensive use of the prosthesis with the somatosensory feedback system, all amputees reported a reduction in PLP and an improvement in prosthesis motor control [56].

To my knowledge, there are no long-term studies with follow-ups to see whether the treatments persist in the long-term. It is therefore unknown whether PLP is kept at a low level, worsens, or improves with time after treatment.

Targeted muscle reinnervation (TMR) is a surgical method where nerves from the residual limb are redirected to other denervated muscles that have lost their biomechanical function after an amputation [58]. The redirected nerves can re-grow and reinnervate the new targeted muscle. TMR has been shown to reduce PLP and RLP when comparing pain in TMR amputees to non-TMR amputees. This technique also prevents the forming of neuromas [59].

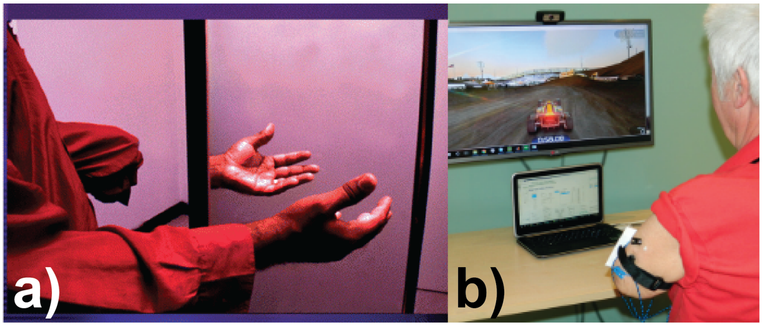

Figure 2.6: Different methods to alleviate PLP. a) Mirror therapy (reused from Ramachandran and Altschuler [60], ©2009 Oxford University Press), b) Motor execution therapy in AR (reused from Ortiz-Catalan et al. [55], ©2016 Elsevier Inc.)

2.4.2 Phantom hand map/referred sensations

Referred sensations are non-painful sensations in specific areas of the phantom hand that occur when cutaneously stimulating areas on the residual limb [61]. Referred phantom sensation is commonly referred to as a phantom hand map (PHM) [62–64].

It remains unknown why some amputees experience referred sensations, and the cause for the projected sensations on a PHM is also unknown. Björkman et al. [61] used functional magnetic resonance imaging (fMRI) to measure brain activity in eight amputees with a PHM and compared the results to individuals with an intact limb. The result showed that the location of the finger-specific area in S1 corresponded well with the intact fingers' area when applying cutaneous stimulation on the area of referred sensation on the residual limb. The perceived sensations can therefore be seen as somatotopically matched. They also showed an enlarged activation in S1 in amputees, indicating that the enlarged area also includes adjacent cortical hand and forearm areas. Ramachandran and Rogers-Ramachandran [42, 62] suggest that the referred sensation is a cause of cortical reinnervation, i.e. if another surrounding body site receives tactile input, the sensation would be misinterpreted as arising on the missing limb. Therefore, applying cutaneous stimulation to the face elicits referred sensations on the residual limb since the cortical representation of the face shares a direct border in S1 with upper limb representations [62].

The most common method to localise the phantom digits of a PHM is by touching areas on the residual limb and marking the referred sensations corresponding to specific amputated digits or areas of the hand. Thereafter, the traced referred areas are confirmed by the amputee [61, 64, 65]. Figure 2.7 shows a PHM for two amputees. It has been observed that the finger representation of a PHM, induced by transcutaneous electrical nerve stimulation (TENS), is more detailed for amputees with a more distal amputation level [66]. Furthermore, it has been shown that the vividness of PHM is more stable, or increased, for upper limb amputees wearing a functional prosthesis compared to a cosmetic one [67].

Stimulating a PHM provides the amputee with feedback that can be seen as somatotopically matched, which has also been shown to improve the performance in discriminating the location for electrotactile stimulation and reduce the cognitive load compared to when provided with non-somatotopic feedback [65]. The referred sensations could also be modality-specific, meaning that warmth and cold applied to the face yields the same perception on the phantom-specific fingers [68].

Figure 2.7: Phantom hand maps, where the location and area of the map for each digit or part of the hand varies for each individual. (Reused from Björkman [63], ©2012 Journal of Rehabilitation Medicine, CC-BY-NC 4.0 license.)

Chapter 3
Prosthetic hands

An individual with acquired limb loss or congenital limb deficiency has limited motor and sensory function in the residual limb. However, some of the hand's functionality and appearance can be restored by using a prosthetic limb. The investigation of myoelectric controlled prostheses started in Germany in 1945, and due to World War II, American scientists and engineers started to become deeply involved in finding solutions to replace lost limbs [69].

Despite the advancements in prosthetics, the abandonment rate is high for both paediatric and adult populations (35–45% and 23–26%, respectively). The reasons for abandonment are commonly due to discomfort, dissatisfaction with the appearance, lack of functionality, and lack of sensory feedback [70]. Even if upper limb amputees report satisfaction with the usefulness of their prosthesis, and display good skills in handling ADL tasks with their myoelectric prostheses, the actual use in everyday life is not more than half of ADL tasks [71]. In Austria, the abandonment rates reached 50% and showed no significant change in prosthetic abandonment throughout the advancements in prosthetics from 1996 to 2006. The advancements mentioned are related to comfort, weight, motor control, and functionality, by customising prosthetic sockets, adding lightweight lithium batteries, developing dexterous hands, and optimising control strategies [49]. The discomfort of a prosthetic fit often depends on the socket, which has come to improve over the years because of enhanced modern materials and improved designs to custom-fit the socket to the residual limb's anatomy [72]. Solutions such as 3D-printed sockets, where an optical scanner is used to scan the residual limb, have been introduced as an alternative to the traditional sockets made in clinics to achieve a more customised fitting [73]. Besides prosthesis technology advancements, well-structured prosthetic training that is tailored to the patient has been suggested in order to attain a higher acceptance rate in wearing prostheses [49].

It is also suggested that the prosthetic arm can help to maintain body posture and therefore reduce future neck and back pain [7]. Imaizumi et al. [74] examined body posture in two groups of amputees: those who frequently use their prosthesis and those who rarely use their prosthesis. Amputees frequently wearing their prosthesis showed a stabilised body posture; removing the prosthesis increased postural sway. In contrast, if rare users are fitted with a prosthesis, the postural sway is more significant than when not wearing a prosthesis. Due to the habitual wearing of a prosthesis, the prosthesis starts to be recognised as one's body part and becomes embodied. It is therefore suggested that an embodied prosthesis stabilises body posture while an unembodied prosthesis perturbs posture [74].

A case study was used to perform gait analysis for a patient with multi-limb amputation (left knee disarticulation and left transhumeral amputation). The gait mechanics were compared when wearing an upper limb prosthesis and not wearing a prosthesis. The amputee showed an improvement in gait, increased confidence during ambulation, and improved trunk lateral flexion symmetry when wearing an upper limb prosthesis [75]. For future studies, it would be interesting to measure repetitive stress on joints, neck, back, hips, and knees when not using a prosthetic limb for a long period of time. It would also be interesting to investigate whether wearing a prosthetic hand could avoid future wear-out pain, since the reduced weight on one side should make muscles on one side of the body weaker or tensed due to of overload.

Upper limb prostheses can be categorised as passive/cosmetic prostheses (Fig. 3.1a), active hands, and body-powered (Fig. 3.1b,c) and electrically powered prostheses (Fig. 3.3). Passive/cosmetic prostheses are lightweight and restore the aesthetics of the missing hand. There are passive prostheses that are functional, meaning that they facilitate specific activities [76]. Furthermore, passive prosthesis aids as support in bilateral activities. Passive/cosmetic prostheses are most commonly used by amputees who underwent a metacarpal amputation and those who did not work [77]. The simplest active prostheses, the body-powered prostheses, contain a wearable harness and cables and tend to be very durable. The harness captures the movement from the upper arm, shoulder, or chest and transfers the movements to a Bowden cable which opens and closes a hand or hook. During manipulation, the users experience proprioceptive feedback [78], which is one of the benefits of the body-powered prosthesis. Furthermore, it is cheaper than alternative active prostheses and simple to operate. It has also been shown that the user receives more feedback from a body-powered than a myoelectric prosthesis in terms of grasping force, where the user can detect resistance forces in the Bowden cable. Furthermore, the direct connection between the movements of actuating body part and the prosthesis gives a sense that the prosthesis is an extension of one's own body [79]. However, a body-powered prosthesis can bring discomfort to the axilla as well as skin irritation caused by sores after prolonged use and perspiration which can be seen as a cause for rejection [10]. Operating a myoelectric prosthesis requires less expenditure of energy from the user than a body-powered prosthesis and can provide more DoFs [76]. Electrically powered hands are promising candidates for transitioning into closely imitating a real hand, both in motor control and sensory feedback.

Figure 3.1: Examples of passive and simple active, cable-controlled prosthetic hands. a) Passive prosthetic hand (photo: Otto Bock [80]), b) cable activated Movo Hook 2 (photo: Otto Bock [80]), c) cable activated voluntary opening system hand (photo: Otto Bock, [80]).

3.1 Hardware design

Some improvements are still required in myoelectric hands if they are to imitate a human hand as much as possible. Most of the desired criteria from a user perspective are to make the prosthetic hand comfortable, lightweight and to provide sensory feedback [81].

Prostheses are progressing in terms of functionality by adding more DoF and at the same time keeping prosthetic hands to a similar size and weight as a human hand. Increasing DoF would require more motors, sensors, and additional electrical components, thus increasing the prosthetic hand's weight and, potentially, the size.

The underactuated mechanism addresses the increased number of actuators. This mechanism allows fewer actuators than the prostheses' DoFs, where the phalanges of the fingers are coupled using a tendon-driven mechanism or linkage-driven [82]. The linkage-driven mechanism links multiple joints together, where only one motor is required to produce flexion and extension in three-finger joints. The common linkage-driven mechanisms are comprised of bar linkages, which bend the finger in a curling motion similar to a finger's trajectory. This mechanism produces high grasping force; however, when a phalange touches an object, the finger motion will stop and prevents the fingers from fully enclosing objects with different shapes. The tendon-driven mechanism is created by passive components using cables and torsion springs (acting like tendons) to flex and extend fingers, which enables an adaptive grasp of arbitrarily shaped objects [83]. Prosthetic hands with an adaptive grasp would reach more contact points on the object and prevent slipping [84], and the force would be distributed between the fingers, and the object [85]. The tendon-driven mechanism uses a simpler structure than the linkage-driven mechanism, it is also lighter, but the linkage-driven mechanism exerts higher forces and is more durable.

Even if using a underactuated mechanism to reduce the weight by reducing the number of motors, Belter et al. [85] show that the weight of the prosthetic hand correlates with the number of joints, since the coupling of multiple finger joints would require additional material.

Usually, a myoelectric prosthetic device needs to be charged at least once a day, and most individuals employ their prosthesis for at least eight hours [86]. Implementing more features, such as a sensory feedback system, consumes even more power. The battery life should therefore be considered when developing prostheses, in order to make them last for a full day without requiring charge. A review article mentions several studies for lower limb prosthetic devices on how to harvest the biomechanical energy from human movement to be converted into electrical energy for the prosthetic device [87]. This approach could be interesting if the energy produced by human motions, e.g. during arm swing, could be enough for powering prosthetic hands or recharging the battery.

3.2 Motor control

The human hand has 27 DoF, which is more dexterous compared to commercially available prosthetic hands, which have 1 to 11 DoF [85]. A direct current (DC) motor is the most common actuator for controlling the fingers of a prosthetic limb. The small size of DC motors makes it possible to implement several motors to provide high DoF. In research, the most dexterous prosthetic hands have 16 DoF. However, having numerous actuators reduces the grasping force since a more dexterous hand requires smaller motors to accommodate the hand's space constraints compared to less dexterous ones, which can use larger and more powerful motors [85]. A human hand has an average precision grasping force of 95.6 N and a power grasp can reach up to 400 N, but only 68 N is required to carry out ADL [88]. It is suggested that the grip force for prosthetic hands should be at least 45 N [89].

There are different approaches to controlling a myoelectric prosthesis. The simplest and most commonly used technology is dual-site systems. Two electromyography (EMG) sensors are placed on different muscles to detect electric signals from the muscles. The EMG sensor amplifies, filter and rectifies the signal [2]. The sensors are usually placed on two antagonist muscles. The two different muscle movements are mapped to an open and close action of the prosthetic hand (Fig. 3.2a).

Most prostheses have proportional myoelectric control, where the opening and closing speed is proportional to the muscle signal's amplitude. With some prosthetics, the user can switch between different grip patterns by, e.g., co-contraction (activating both muscles simultaneously) [90]. This simple control system is limited by only generating three signals. Furthermore, the contact between the sensor and the skin is crucial for the sensor to detect myoelectric signals, the user needs to generate a clear muscle signal (activate specific muscles), and the response to the user is slow when switching between grip patterns [90].

A potentially more intuitive control system is myoelectric pattern recognition, which uses multiple sensors, thus allowing the user to move their hand naturally. Placing multiple sensors around the circumference of a limb, a resultant EMG vector can be calculated from the combination of each signal from the sensors. Specific gestures will produce a characteristic resultant EMG vector [2]. After processing the EMG vector, it is fed into a classifier to identify the user's intended movement and then translated into corresponding prosthetic movements. However, the number of sensors requires more space and has to fit on the residual limb. For higher levels of amputation (above elbow), some muscles that are crucial for generating enough distinctive muscle patterns that can be differentiated from each other are lost [90]. However, this can be resolved by TMR [81], which is a surgical technique used for amputees with transhumeral or shoulder disarticulation amputations. The chest is a typical area for reinnervation because of its flat and large area, which is suitable for recording electrodes, and because there is enough space for a dense array of electrodes [2]. The reinnervated muscle works as an amplifier for motor signals that can improve motor control of a prosthetic hand.

Reading muscle signals on the skin with EMG electrodes has some uncertainty related to skin-electrode problems. Such problems could be electrode conductivity changes (perspiration), spatial changes (electrodes move on the skin), or user pattern variation (contraction intensity fluctuations) [91]. Crosstalk between muscles is another influential factor that affects the reading of surface EMG, where the sensor can react to surrounding muscles when placed on the targeted muscle. Crosstalk can be reduced by removing subcutaneous fat so as to decrease the distance between the source of the signal and the electrode, which decreases the spatial filtering of the EMG signal [92]. An invasive approach can be considered with implantable sensors to remove skin-electrode problems. The electrodes can be implanted either 1) on the muscle (epimysial), 2) within the muscle (intramuscular), 3) within a nerve (intraneural), 4) by enclosing the nerve (extraneural) or lastly within the brain or spinal cord for direct control by the CNS. Compared to brain recordings, the signals recorded from peripheral nerves are not as stable since the peripheral nerves are close to moving tissues such as muscles and tendons, which interfere with the signal and adds noise. The distance between the source of the signal (neurons) and the implantable electrode (extraneural) varies and results in an unstable reading but can be resolved by more invasive electrodes (intraneural). The intraneural electrodes ensure good bio-compatibility, robustness, and stability for long-time sustainable performance. However, extraneural electrodes cause less damage to the neural tissue. Interferences that can hinder performance include the growth of fibrotic tissue around the electrode, electromechanical damage to the electrode, and changes in conductivity (inhomogeneity of the intrafascicular space).

Another invasive technique is osseointegration, a bone-anchored prosthesis that removes all discomfort which a socket brings. The prosthesis is attached to a titanium implant penetrating the skin, which in turn is attached to another titanium implant anchored in the bone (Fig. 3.2b) [93, 94]. With this technique, surface EMG (sEMG) sensors can be directly implanted within the residual limb (epimysial), which reduces muscular crosstalk and improves the signal-to-noise ratio [95]. However, osseointegration typically requires surgery prone to complications, though which most often tend to be minor, such as soft tissue infections [96]. Osseointegrated prostheses were primarily used for teeth implants and have since been used for facial prosthetics and anchoring hearing aids. In the 1990, the world's first surgery involving the implanting of a transfemoral osseointegrated prosthesis for a transfemoral amputee was performed by Dr. Per-Ingvar Brånemark and Dr. Björn Rydevik [97].

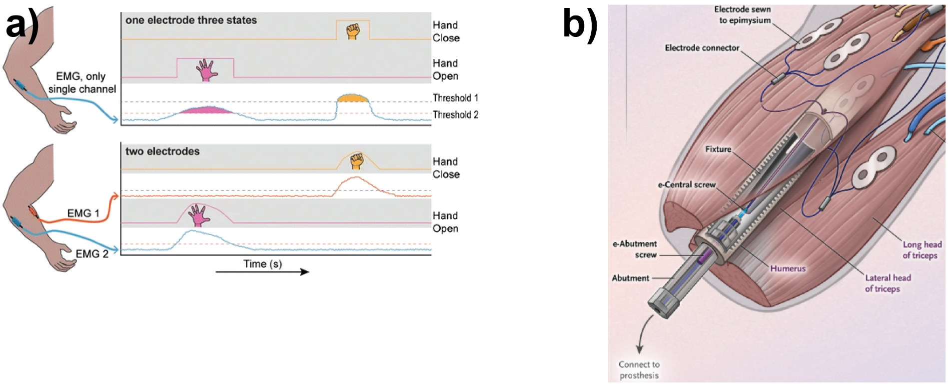

Figure 3.2: Different types of technology control systems. a) Myoelectric direct control (picture from Roche et al. [98], ©2014 Springer Nature), b) osseointegrated implantation where the prosthesis is attached to an abutment and a fixture for skeletal attachment. This system allows for a bidirectional communication to a prosthesis through an e-abutment screw and an e-central screw. Electrodes were implanted in the muscle. (Reproduced with permission from [94], Copyright Massachusetts Medical Society.)

3.3 Commercial myoelectric prosthetic hands

Open Bionics provides a 3D-printed myoelectric bionic arm with multi-grip functionality, called the Hero Arm, suitable for below-elbow amputees. The bionic arm was also the first 3D-printed arm that was clinically proven and is available in the United States of America, the United Kingdom, Europe, Australia, and New Zealand [99]. With the customised covers for the Hero Arm, the arm attracts, besides adults, younger patients, because the appearance can be customized to look like popular science fiction characters [90]. The Hero Arm comes with either three or four motors, where the former is smaller than the latter. The three-motor hand comes with a single tendon for each finger, while the four-motor hand has two tendons on the index and middle finger, which allows them to move independently. This allows the hand to have six different grip patterns, while the three-motor version only has four grip patterns. The grip patterns are grouped to make switching between them more manageable, which is done by pressing a button. The Hero Arm uses two EMG sensors to open the hand and close/perform the selected grip. The speed can be controlled by tensing the muscle gently or firmly to move the hand slowly or fast [100].

MyHand from Hy5 [101] is another myoelectric-controlled prosthesis that uses one motor to control three hydraulic cylinders. The hydraulic cylinders control two joints in the thumb, index, and middle finger, enabling a more adaptive grip. The ring and little finger move together with the middle finger. During the closing action, the hand will fold around any given object. MyHand provides five different grip patterns, where external manipulation is needed when switching to a specific grip pattern. For example, switching to finger point, the index finger needs to be held back while closing the hand. Hydraulic cylinders are more robust, more vigorous, and less noisy than motors, and as a pressure-based system, it provides more flexibility in the joints [85].

Michelangelo Hand by Otto Bock [80, 102] uses the Axon-Bus system that offers higher functionality and reduces complementary movements by the user. This is achieved by the active wrist and wrist rotation (supinate and pronate). Furthermore, the thumb, index finger, and middle finger are actively driven, and the ring and little finger passively follow the movements of the active digits. However, the mechanical coupling of all the digits prevents the index finger from flexing independently to form a pointing grasp pattern. This is possible with iLimb (by Touch Bionics Ltd., UK) [103] where all the digits independently perform flexion/extension, by containing more actuators [85]. VINCENTevolution 4 (by Vincent Systems GmbH, Germany) [104] is another advanced myoelectric prosthesis where the grip is conformed to the shape of the object that is grasped.

A startup in India, Robo Bionics [105], developed a prosthetic hand called Grippy, which uses mechanomyogram (MMG) sensors to read the mechanical signal instead of an electrical signal from EMG sensors. Using MMG sensors is suggested to be a better fit for patients who cannot achieve fine control of muscles and eliminates problems when using EMG sensors where the reading can be affected by humidity and heat. MMG sensors are, therefore, more suitable for the environment in India. Grippy provides an adaptive grip that conforms to an object's shape. The fingers close until they encounter some resistance. If a finger point is needed, the user must obstruct the index finger from closing when performing the closing command. However, it is not possible for the index finger to open and close in the finger point pattern, as can be useful in some situations, such as when handling a power drill. Grippy has three vibrators to provide the user with sensory feedback. Each of the vibrators is activated successively when closing and opening, though in a different order, and only one is activated when grasping an object. It also provides information of whether the object is soft or hard by making rapid and prolonged vibrations, respectively [90].

Coapt Engineering provides advanced myoelectric control technology using machine learning algorithms. The algorithms learn the patterns in the user's muscle movements, which are translated into intuitive and natural movements [106]. This control system is compatible with the majority of commercial upper limb prostheses [107]. Another pattern recognition system, Myo Plus by Otto Bock [80] is compatible with prostheses in the MyoBock family (BeBionic Hand, Myoelectric Speed Hands, and System Electric).

There are many control system solutions for providing dexterous and intuitive control of commercial prosthetics. However, sensory feedback systems are lacking. A few prosthetics provide simple feedback that uses vibration to alert users of certain simple actions such as contact, grip force, and gesture changes. To my knowledge, the commercial prostheses which provide vibrational feedback are Adam's Hand [108] (speed and power of the grip), MeHand [109] (grip/gesture changes, contact, and grip force), the LUKE Arm [110] (grip force), the Ability Hand [111] (contact and force feedback), Grippy [105] (vibration patterns for opening and closing grip, grasping, and softness), and Vincent Evolution [112] (contact and force feedback). Integrum's e-OPRA Implant System [97] is a solution for combining osseointegration with the possibility of an integrated neural interface (cuff electrode attached to a nerve) and embedded myoelectric sensors for better control and for providing the user with sensory feedback.

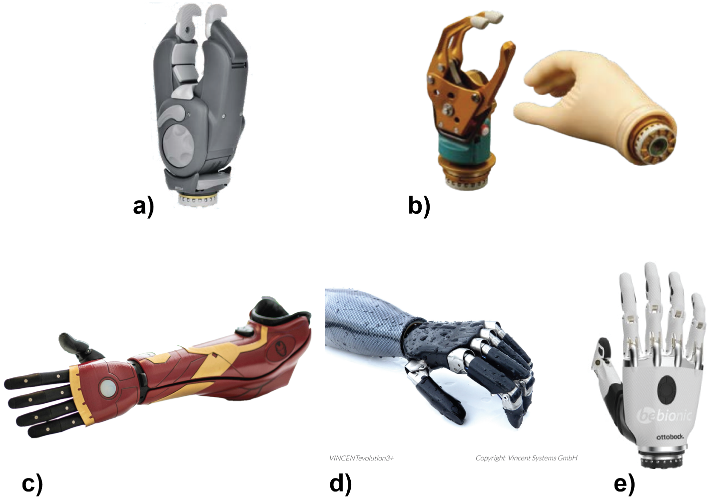

Figure 3.3: Examples of electrically powered prosthetic hands. The first row shows simple grippers. The second row shows more dexterous hands. a) DMC VariPlus System Electric Greifer (photo: Otto Bock [80]), b) MyoHand VariPlus Speed 2 (photo: Otto Bock [80]), c) Hero Arm (photo: Open Bionics) [99], d) Vincent Evolution 3 (photo: Vincent Systems [112]), e) BeBionic Hand (photo: Otto Bock [80]).

3.4 Prosthetic hands in research

Prosthetic hands have been explored in research for decades to find solutions to improve functionality (independently driven fingers, thumb abduction/adduction, and wrist flexion/extension) and control and to reduce the total weight.

CSIC in Spain developed an underactuated prosthetic hand, the MANUS HAND [113], using a crossed-tendon mechanism in the fingers. Compared to the traditional bar mechanism for inter-phalanx coupling, the crossed-tendon mechanism allows the lever arm to move independently of the finger angle, reducing the force required to generate torque. The MANUS HAND contains ten joints, where four (thumb, index, middle finger, and wrist) are independently driven. The ring and little finger are bendable and can be manually manipulated. A Geneva-based mechanism (continuous rotary movement translates into intermittent rotary motion) makes it possible to use one actuator to move the thumb in two planes. To control the grasp, sensors based on the Hall effect pick-ups were used to measure the angular position and grasp force, and are distributed on the hand, on the tips of the thumb, index, and middle finger.

SmartHand (developed by Cipriani et al. [114]) is a five-fingered underactuated transradial prosthesis with four brushed DC motors. The electrical brushed motors provide a good performance compromise, where this type of motor has the lowest power-to-weight ratio compared to other actuators. Another advantage of the SmartHand is the non-back-drivability transmission, which allows a small torque without consuming energy. In this way, the power can be turned off once the desired grasp/stability is reached. As for safety, the hand will not release the grasp of an object during electrical supply or battery failure [115].

The Defense Advanced Research Project Agency (DARPA) announced a new research project, "Revolutionizing Prosthetics", in 2005, due to the increased rate of upper limb loss, and charged DEKA Integrated Solutions Corp. (DEKA) to develop an upper limb prosthesis with high DoF [116]. They developed three prototypes (Gen 1, Gen 2, and Gen 3) with three configurations that fit different levels of amputation: transradial amputees, transhumeral amputees, and amputees with shoulder disarticulation. The DEKA arm had six grip patterns and ten powered joints for the shoulder configuration, with a control scheme with dual modes to switch between "hand mode" (control movements of hand and wrist) and "arm mode" (controlling elbow, shoulder, or a combination of both). The DEKA arm is now commercially available as the LUKE arm, which is manufactured by Mobius Bionics [110].

SSSA-MyHand (developed by Controzzi and colleagues) [117] uses three motors and a Geneva drive to actuate a five-fingered anthropomorphic hand. This construction allows the thumb to independently flex/extend and semi-independently abduct/adduct the thumb and flex/extend the index finger. As for the middle, ring, and little finger, flexion/extension is done simultaneously/synchronously. A single actuator is used for the semi-independent transmission, where a Geneva drive is used to abduct/adduct the thumb in parallel with a four-bar mechanism to flex/extend the index finger. The semi-independent movement of the thumb and index finger makes it possible for the SSSA-MyHand to perform most of the useful gestures that are used in ADL, comparable to commercial prostheses that use more actuators. This mechanism reduces the weight of the SSSA-MyHand, which is comparable to a single-DoF prosthetic hand.

Chapter 4

Artificial sensory feedback

Clinically available prosthetics restore the motor function of the hand to a certain degree, but most do not restore sensory feedback. Integrating both efferent motor output and afferent sensory feedback would provide a closed-loop control in a prosthetic device [118,119]. Without sensory feedback, it is difficult for the prosthesis user to control the exerted grasping force of the prosthetic hand and they thus need to rely on visual and auditory cues when interacting with objects [7]. Some commercial myoelectric prosthetic hands provide the user with vibrotactile feedback of simple actions, such as contact, grasp force, and when changing between gestures.

It has been recommended to include both implicit feedback (proprioception, prosthesis sound, and vision of the prosthesis) and explicit feedback (tactile feedback) in the design of a closed-loop sensory feedback system [120]. Implicit feedback is reliable enough for simple tasks, while additional explicit feedback is beneficial when grasping tasks become more complex and also has an advantage in ADL where the implicit (visual and auditory) feedback may be obscured [121].

Touch is crucial to convey information about surroundings, and it plays a critical role during emotional and social interaction. Implementing sensory feedback in prostheses could diminish the cognitive load and also increase task performance [122,123]. Additionally, sensory feedback is one of the desired features to enhance the user experience and increase the acceptance of wearing the prosthetic device [7,10,11].

Perceiving the prosthesis as a part of one's own body (embodiment) [124], where the prosthesis feels like an extension of the body rather than a tool, has been demonstrated to be crucial for prosthetics acceptance [125]. Embodiment is often conceptualised based on experimental phenomenology and can be divided into two subcomponents: *1) agency*: the percept that one is in control of the movement and *2) ownership*: the prosthesis feels like it is a part of one's own body [124]. Sensory feedback plays a crucial role in inducing ownership and improves the control of the prosthesis [7]. This enhances both ownership and agency of a prosthesis, hence contributing to prosthesis embodiment. Amputees who wear prostheses with a non-invasive sensory feedback system in ADL have reported an experience of completeness [126]. Furthermore, prosthesis ownership has shown to be significantly related to lower levels of PLP, and RLP [127].

Research on sensory feedback systems that can be used in prostheses has been undertaken for many years, where both invasive and non-invasive solutions have been presented [83, 119, 128–132]. Sensory feedback can be achieved by adding sensors to prostheses so as to receive a stimulus through touch or joint movements (proprioception). Through encoding algorithms, the stimulus is converted into electrical parameters to be sent to a stimulator (containing different types of actuators) in order to deliver sensations to the prosthesis user. The sensory feedback system would preferably deliver sensations which are intuitive in a way where the received sensation should closely match the expected sensation that is seen delivered to the prosthesis, in order to reduce the training and interpretation of the received sensation.

4.1 Sensors

There are different types of sensors that detect physical parameters. For a prosthetic hand, there should be sensors that can detect force, finger position (angle), slip, and temperature [133], but at the same time the sensors should be small (<100 mm^2 [133]), lightweight (<1 g [133]), and have low power consumption (<1 mW [133]) in order to fit into the prosthesis and to not contribute extensive weight. In terms of technical features, the sensors should provide a stable and repeatable output signal and be able to handle harsh environments and remain unaffected by an electric field or other physical properties, such as temperature, humidity, radiation, and chemical stresses. There should also be minimum tactile crosstalk between sensors or sensing elements [21]. Table 4.1 presents the most common commercial sensors that have been used to replicate the mechanoreceptors to some extent.

Different sensing technologies that can be used in dexterous robotic hands is discussed in a review by Kappassov et al. [128], where they also discuss how some design criteria might affect other factors. A design criterion for a high spatial resolution of a tactile sensor is dependent on the body site, where it can be 1 mm in size on the fingertip of the prosthesis or 5 mm on the palm [134]. In order to increase the spatial resolution, adding more sensing elements is required. This will increase the processing time and require more wires, which will increase the sensitivity to electromagnetic noise [128]. This could be resolved by shielding and smart wiring, though this will take up more space in the prosthetic hand. As the sensing element gets smaller, the signal-to-noise ratio also gets smaller and degrades the sensitivity

Table 4.1 Mechanoreceptors and corresponding artificial sensing

Sensing	**Receptors**	**Sensors**
Force	Merkel's discs (SA-I) Ruffini's corpuscles (SA-II)	FSR Hall effect Strain gauge Piezoelectric Capacitance
Position (angle)	Proprioceptors Ruffini's corpuscles (SA-II)	Hall effect Rotary potentiometer
Slip detection	Meissner's corpuscles (RA-I)	Acoustic Piezoelectric Accelerometer FSR matrix
Texture	Pacinian corpuscles (RA-II)	Accelerometer Acoustic
Softness/Hardness	Merkel's discs (SA-I) Pacinian corpuscles (RA-II)	Accelerometer Strain gauge
Temperature	Thermoreceptors	Thermistor

of the sensor, since the noise level will be closer to the sensor signal level [128]. A sensor's sensitivity is its ability to detect the smallest variation in pressure/force, which is favourable when manipulating fragile objects. However, a sensor with a high sensitivity could decrease the dynamic range which is the minimum to maximum detectable range. For the design of a dynamic sensor to detect vibrations during e.g., slip, discrimination of texture, or onset/offset of a contact, then the frequency response should be at least 400 Hz, with a sampling rate of at least 800 Hz (according to the Nyquist-Shannon sampling theorem). There are some tactile sensor designs that use soft material (elastic) to replicate the human finger and increase the surface friction, which eliminates the need to apply high normal forces to the handling object in order to keep it stable, as opposed to a material with low friction. However, the soft material limits the frequency response of the tactile sensor, creating a phase delay in the propagation of the waves from the mechanical vibrations during onset of contact and degrades the sensitivity. Furthermore, when pressing and releasing a sensor with soft material, the material will be compressed, and when released it might not expand immediately (hysteresis effect). Moreover, it might not regain its shape as it was prior to compression (memory effects). A tactile sensor should have low hysteresis in order to achieve a good dynamic response for vibration detection.

4.1.1 Tactile sensing technology

In order to perceive the correct haptic information when exploring an object, we interact with the object in different ways (*exploratory procedures* [135]). Lederman and Klatzy [135] videotaped blindfolded subjects while they were examining different objects. Depending on what stimulus property the subjects were judging, they used different exploratory procedures. When examining the texture, subjects executed a lateral motion across the object. When judging hardness, they pressed the object, they used static contact for temperature sensing, enclosure for volume and global shape, contour following for volume and exact shape, and unsupported holding for weight discrimination. The following introduces some of the common technologies that have been used to create sensors for prosthetic hands.

Piezoresistive sensors are effective in measuring static forces. Applying mechanical stress to the piezoresistive sensor, the electrical resistance changes in the semiconductive material. The simplest and cheapest type of piezoresistive sensor is force sensitive resistor (FSR). The sensor consists of a conductive film, a conductive print on a substrate, and a spacer in between. In an unloaded state (the circuit is open), the resistance is high, which drops when applying force. The larger area of the conductive film that comes in contact with the conductive print, the lower the resistance will be. Applying a force until the contact area reaches its maximum will lead to a saturation of the sensor. For prolonged pressure on a piezoresistive sensor, the resistance might drift, making the sensor more fit for qualitative measures than quantitative ones [136]. However, an FSR is a good and affordable solution in applications where accurate force measurement is not required. FSRs are commonly used to detect contact, detect force thresholds, and detect a relative change in force. An FSR has the advantage of being small enough to fit in an artificial hand. However, the FSR comes with some disadvantages, such as low repeatability, hysteresis, and temperature drift [133].

A capacitive tactile sensor consists of two conductive plates with a dielectric medium in between. The sensor can detect any changes of displacement of the conductive plates, which in turn changes the capacitance of the sensor. In this manner, the sensor can detect both touch and proximity events. The distance between the plates changes when applying normal force; the effective area between the plates changes while applying tangential force [19]. Capacitive sensors have a higher frequency response compared to piezoresistive sensors. However, capacitive sensors are susceptible to electromagnetic noise, have a non-linear response, are prone to hysteresis, and are sensitive to temperature [128].

A piezoelectric sensor converts mechanical energy into electrical energy, and is effective for measuring high frequency dynamic forces. The electric charge is generated by the piezo element (crystal or ceramic) when applying pressure on the sensor. The voltage can be measured, which is proportional to the pressure. It is best used for dynamic force, since the given static force will result in a drop in the output signal, and has a faster dynamic response than capacitive sensors. The sensor generates an output signal without any external power source. However, it requires an amplifier to make the signal output detectable. Some sensors come with an internal amplifier, which necessitates an external power source. A piezoelectric sensor is robust and therefore suitable for harsh environments. Furthermore, it is insensitive to electromagnetic interference. A piezoelectric accelerometer measures vibration and acceleration, and produces an electric signal when exposed to an external force that can be generated by vibration. A potential drawback with an accelerometer is its response to other vibrations that can occur in prosthetic hands, when only the vibrations from the skin during manipulation is desired [137]. This could be resolved by partially isolating the accelerometer using soft material.

A strain gauge changes its resistance with changes in strain, which is a deformation/displacement when stress is applied. A single strain gauge can be attached to an object, and when the object is stressed (bent or twisted), the resistance changes proportional to the deflection of the object. In order to measure the change in resistance, a Wheatstone bridge can be used. The resistance will decrease during compression and increase during stretching [136].

A thermal resistor – a thermistor – changes its electrical resistance with external temperature. There are two types of thermistors; negative temperature coefficient (NTC) and positive temperature coefficient (PTC) thermistors. For NTCs, the resistance drops with increasing temperature, making it common for temperature measurement. For PTCs, on the other hand, the resistance increases with temperature, making it a better fit for protecting circuits from overheating. Thermistors are non-linear but have good accuracy and repeatability [136].

A Hall effect sensor is a magnetic sensor that measures the changing voltage when the sensor is in an external magnetic field. The main disadvantage for usage in tactile sensing is its sensitivity to other external magnetic fields [133].

4.1.2 Sensor applications in artificial hands

Force/Pressure

When grasping an object, there are three-dimensional forces exerted on the skin. The forces form components at right angles and tangential to the skin, called grip and load forces. Increasing indentation in the skin increases the receptors' response. The afferents located superficially in the skin have small receptive fields, SA-I and RA-I, which detect skin deformations of relatively low frequency and high frequency, respectively [138]. The SA-I afferents detect sustained responses proportional to the pressure on the skin and fire more rapidly the heavier the object. SA-I detect normal and shear forces [139], while SA-II detects tangential forces [21]. RA-I afferents excite when breaking contact with an object and RA-II afferents excite when both initiating and breaking contact with an object [140].

An inexpensive capacitive sensor (SingleTact [141]) has been selected as a good candidate during an evaluation of commercial sensors to measure normal forces. SingleTact was selected in an evaluation to fit a specific case study, where SingleTact provided a higher accuracy sensory reading, and permanent modifications of the prosthetic digits could also be avoided [142]. SingleTact was compared to an FSR and a piezoresistive load cell – three cheap solutions. The case study concluded that an FSR was suitable for low-accuracy applications, whereas the load cell provided a higher accuracy. However, a modification of the prosthetic digit might be required.

Ge et al. [143] designed a capacitive sensor that can detect a grasping force in the range of 0–12 N and when an object approaches from a distance of approximately 100 mm (proximity detection) the capacitive sensor detects disturbances in the electric field. With collision control, a collision with an object can be avoided since the finger would stop steadily before making contact with the object. The sensors were integrated into the fingertips of a prosthetic hand.

Votta et al. [144] present a design for magnetic force sensors sensing both normal and shear forces. The finger contained a PCB with a 3-axis Hall effect sensor in the bottom part of the finger and a small magnet in the top part. When the fingertip touches an object, the magnet moves relative to the Hall effect sensor, and depending on the direction of the applied force, the sensor reads either a normal or shear force. This sensor was used for automatic grasping, where the normal force is measured, and when it reaches the normal threshold, the prosthetic hand enters a ”stable grasp” state. The hand enters the ”lifting” state when the shear force increases above a shear threshold. When the shear force goes below the shear threshold, after the ”lifting” state, the hand enters the ”release” state. The object is placed back on the table, and the hand automatically opens and releases the object during the ”release” state. Votta el al. compared the performance in a series of pick-and-place tasks with a normal EMG-controlled hand and found a slight improvement in releasing the object and as well as a decrease in the user’s mental burden.

Finger position

Proprioceptors respond to internal stimuli that senses when an organ stretches; in this way the receptors inform the brain about the body’s movement [16]. There is also evidence that the SA-II afferents can act as proprioceptors [138]. Using sensors that can read joint angles is common for determining finger position.

Hellman et al. [145] present a BairClaw index finger which uses the BioTac sensor. To measure four sets of joint angles, a Hall effect sensor was used, together with a diametrically magnetised ring magnet. The different sets of joint angles were flexion/extension in the MCP, PIP, and DID joints and adduction/abduction in the MCP joint. The sensors could measure the joint angles with a resolution of $\leqq 1°$.

Slip detection/texture discrimination

As previously mentioned, when grasping an object, there are both grip and load forces applied to the skin. When lifting an object, these forces have to be appropriate for taking up for the object's weight and the frictional conditions in order to prevent slipping [2]. The skin's ability to detect acceleration makes it possible to detect slip and identify textures. When an object is slipping, there is a response in SA-I, RA-I, and RA-II afferents. RA-II afferents detect micro-vibrations, which occur on smooth surfaces or textures with limited features [146]. To identify textures, the hand moves over the object, which creates tangential motion across the skin [2]. RA-II are responsible for detecting accelerations, and considering the receptor's sizable receptive field, the information regarding acceleration is non-spatial. Instead, it provides temporal, intensity, or modal information [147], which makes the RA-II afferents responsive to flutter, slip, and motion across the skin. Textures often contain features that are less than 1 mm in size. SA-I afferents provide spatial information about an object's form and texture [148]. The spatial variation in the SA-I response correlates with a surface's roughness [149].

Sensors sensitive to dynamic forces and movements are suitable for slip detection and discriminating textures. Such sensors could have piezoelectric characteristics or be accelerometers.

Cotton et al. [150] used thick-film printing techniques to prototype a fingertip. In order to detect vibration (associated with slip) they used a piezoelectric dynamic force sensor that can detect vibration. The slip signal provides information about the relative velocity between the finger and an object. In this study, the signal from the piezoelectric sensor was smaller than the background noise signal (50 Hz) for low velocities, but it was suggested to move the electronics closer to the sensor and use a battery as a power source.

Lowe et al. [151] used an accelerometer to detect slip in order to achieve stable grasping. The sensor unit was designed to be more sensitive to vibrations in prosthetic gloves than mechanical vibrations. The sensor could detect the onset of a slip in three planes and measure the slip distance of the object.

Romeo et al. [152] used a biomimetic fingertip with integrated piezoresistive MEMS sensors to detect slip. The sensors were provided with a 16 mono-axial signal for a novel slippage detection algorithm, which included digital filtering, rectification, and enveloping. The algorithm generates an ON/OFF slippage identification signal, which can detect relative movements between the sensor and the object smaller than 5 μm.

Mingrino et al. [153] developed a slip sensory system, which detects both normal as well as shear forces and local vibrations based on FSR technology and a dynamic sensor (piezoelectric polymer film). The system monitors the shear and normal force ratio before and during slipping. There was a significant increase in the ratio (shear/normal force) immediately before the object slipped from the grip. This ratio acted as a warning signal for an incident of slip. In contrast, the dynamic sensor generated a high-level signal when the object was slipping and provided helpful information when contact of the object was lost.

Microphones have a good ability to detect surface noise, such as when making contact with an object and during lateral motion. Furthermore, they have a simple implementation because of the wide variety of existing hardware and software for microphones. However, good care needs to be taken to remove the surrounding acoustic noise, which can be done by using a second microphone to detect the noise that can be filtered from the signal. Mayol-Cuevas et al. [154] developed a sensing pen using an electret piezoelectric microphone for tactile texture recognition. The sound input was segmented in order to obtain a fast Fourier transform (FFT), which could then be introduced to a supervised learning vector quantisation (LVQ) classifier system. The sensing pen detected 93% of eighteen textures, where the errors were similar to the intended texture.

Yi et al. designed a bioinspired tactile texture sensor to discriminate surface roughness [155], where polyvinylidene fluoride (PVDF) film sensors were used to replicate the RA-I in detecting the response and location of vibrations. Different features were extracted from the tactile texture signal to be evaluated together with two classification algorithms. The best classification accuracy for discriminating eight surfaces with various roughness values (RA, the absolute average relative to the base length, i.e. average difference between peaks and valleys) were 82.6$\pm$10.8% when using the standard deviation (SD) features as well as the k-nearest neighbours (kNN).

Multi-modal sensors in research

As is known, a human hand contains several receptors to interpret the information during touch. Therefore, multiple physical properties need to be transduced into an electronic signal. The tactile sensing includes the ability to detect temperature, texture, shape, force, friction, pain, and other related properties that are needed during manipulation of the surroundings [21]. Producing an artificial replica of the human hand has been a challenge, and the following text will mention a few studies that combine multiple sensors which try to replicate the natural hand with regard to touch.

Jin et al. [156] introduced a 5 x 5 stretchable multi-modal capacitive sensor with two vertically stacked capacitors (dual-capacitor). The positioning of the capacitors yields a unique combination of the signals in charge (positive or negative) and magnitude, making it possible to distinguish the sensing mode of the sensor. The sensor can detect curvature, pressure, strain, and touch. The sensor was able to absorb a pressure up to 75 kPa, with a pressure sensitivity of 1 MPa^{-1}. During strain, the capacitors increased linearly up to a strain of 25% and had a high touch sensitivity due to their relatively thin protective layer.

Cranny et al. [157] designed a fingertip that can measure grip force and temperature and detect the onset of slip. For the static force sensor, three thick-film resistors were connected in a pseudo-half-bridge circuit, making it independent of the position of the force put on the fingertip. The slip sensor had piezoelectric properties acting as a vibration sensor, and the temperature sensor was a PTC thermistor which could indicate whether the object was too hot or cold. The static force sensor showed high linearity up to forces of 50 N and with a maximum hysteresis of 0.7 N. The slip sensor showed a good response on the first impact with the object and in detecting the onset of slip. The temperature sensor was highly linear.

Ke et al. [158] proposed a design of a fingertip tactile sensor, where they used an FSR and PVDF sensor module. They used an FSR to measure static force, and by increasing the sensitivity and uniformity, they added a contact pad to match the sensitivity of the FSR. For the FSR module, the robustness was evaluated by applying different pressures to the fingertip and in different contact locations. It was concluded that there was high consistency in the pressing and release phases. Furthermore, the contact location had only a small influence during a light load (0–5 N). As mentioned in the study, shear force influences grasping, and this influence needs to be evaluated if it is to have any impact on this design of the fingertip tactile sensor.

Wettels et al. [159] developed a finger-shaped multi-modal tactile sensor, the BioTac®(Syntouch, LLC) [160, 161], which can sense force, vibration, heat flux, and temperature. The BioTac®has a bone-like core inflated by conductive liquid and is covered by an elastomeric skin. The skin and the liquid deform during force sensing, and electrodes read the electrical impedance. To detect texture or slip, microvibrations occur when an object is sliding on the surface of the finger, which propagates as sound waves in the conductive fluid. The sound waves are captured by a pressure transducer (hydrophone). The signal is further processed to improve sensitivity using a high-pass filter and amplification.

A passive pneumatic sensory system has been developed, where the sensor is passive and incorporated in the fingertips of a prosthetic silicone glove. When the sensor is pressed upon, air travels through a plastic tube connected to an actuator which will provide the user with sensory feedback [162]. A characterisation of this sensor it has been presented in Paper II [163], where its potential use in prosthetics is discussed. Importantly, compared to commercially available sensors, this custom-made sensor covers the entire digit, whereas multiple commercial sensors would be needed for this coverage.

Weiner et al. [164] presented a scalable embedded multi-modal sensor that can be used in prosthetic fingers using commercial sensors. Silicone rubber casting was applied directly to the finger with the system. The sensor system can measure normal and shear force, distance, acceleration, temperature, and joint angles. Accelerometers were used for slip detection, Hall effect sensors were used for joint angle measurement, barometer-based sensors (absolute pressure sensors) were used for measuring normal force, and a magnetic field sensor was used for normal and shear force measurements. The distance sensor was suggested for forming a pre-grasp when approaching an object. Their evaluation showed a hysteresis induced by the silicone rubber for the normal and shear force measurements. The system showed that multiple sensors could detect specific events during a grasping task, indicating high confidence in detecting events compared to a single sensor.

Segil et al. [165] developed a multi-modal sensor containing an infrared (IR) proximity sensor and a barometric sensor that were integrated into a prosthetic fingertip covered with an elastomer for a robust contact surface. The sensors measure proximity (0–10 mm), contact (0 N), and force (0–50 N). The IR sensor made it possible to detect contact forces close to 0 N. The multi-modal sensor was able to detect objects that applied forces on the fingertips in various spatial locations and angles of incidence. In their study, the force was applied on five different locations of the hand and three angles of incidence on the fingertips.

4.2 Sensory feedback methods

There are different methods to provide the prosthetic user with sensory feedback by applying sensations on a different part of the body, which would substitute the stimulation applied on prosthetic hands. The sensory feedback can be provided non-invasively (superficially on the skin) or invasively (targeted reinnervation, direct peripheral nervous system stimulation, or central nervous system stimulation).

The stimulation can be categorised as modality mismatched, modality matched, and somatotopically matched [166]. When a stimulation is somatotopically matched, the stimulus is perceived as being anatomically matched with regards to localization, i.e., if pressure is applied on the prosthetic thumb the prosthetic user would perceive the stimulation on their missing thumb. Modality-matched stimulation refers to a sensation that is congruent to the stimulation applied on the prosthetic hand, i.e., if force is sensed on a prosthetic finger, then force is also applied as a stimulus on eg. the residual limb. A modality-matched stimulus does not need to be somatotopically matched. However, is it suggested that the stimulation should be both modality- and somatotopically matched for intuitive sensory feedback, which could reduce the cognitive demand. Such methods could be neural stimulation, using referred sensations, and targeted reinnervation [129]. To apply a somatotopically matched non-invasive stimulation, the stimulation can be applied on amputees' PHM. Amputees with a PHM on the residual limb perform better in discriminating tactile feedback compared to amputees with very little to no referred sensations [64,162]. Studies have shown a better performance for amputees when provided with somatotopically matched compared to non-somatotopically matched feedback [65].

Invasive sensory feedback is prone to complications post-surgery, but can provide a somatotopically matched sensation which could remove additional training and cognitive effort. Chai et al. [167] have shown that the learning period to interpret the applied non-somatotopic sensory feedback is relatively short for amputees with a limited or no PHM. They proposed that a "3-day-effect" is needed to learn electrotactile sensory feedback on non-somatotopic sites, where the performance is comparable to somatotopic sites.

4.2.1 Non-invasive sensory feedback

Non-invasive sensory feedback conveys information to the user's skin without any surgical procedures, and has shown to provide enough information for good performance. Unlike a sensor, which converts mechanical energy or movement into electrical energy, an actuator does the opposite, e.g., it converts electrical energy into mechanical energy or movement [168].

Finding actuators for real-time non-invasive tactile feedback is a challenge, since the actuators need to be small and lightweight to fit in a prosthetic socket, and at the same time provide feedback that should be intuitive enough to require low cognitive load during use. Mechanotactile, vibrotactile, and electrotactile feedback are common methods to mediate the tactile stimulation applied on the prosthesis. Table 4.2 presents an overview of non-invasive sensory feedback techniques.

Table 4.2 Overview of non-invasive sensory feedback techniques.

Stimulation techniques	Parameters	Sensory information	Benefits	Limitations
Mechanotactile	amplitude (force)	contact force grasp force location of contact	close to "real" touch modality matched	bulky power consuming
Vibrotactile	frequency amplitude	grasp force grasp speed hand aperture	small-sized cheep low power	frequency and amplitude change dependently slow rise time
Electrotactile	frequency amplitude pulse width	grasp force grasp speed hand aperture surface discrimination	small-sized low power potential to produce rich sensory information	interference with EMG electrodes can produce unpleasant sensation
Hybrid (multimodal)	all of the above	all of the above	potential to produce rich sensory information	overload of sensations which can increase cognitive load

Modified from Svensson et al. [131].

Mechanotactile feedback

Mechanotactile stimulation often involves applying normal force or tangential stretch on the skin to provide feedback on touch and grasp. When grasping an object, pressure is applied on the prosthetic finger, which is fed back as pressure (mechanotactile) stimulation on the residual limb, providing a sensation that is modality matched. Therefore, mechanotactile feedback is often used to mediate force feedback.

Different actuators have been used to convey mechanotactile feedback, such as servo motors and pneumatics. Mechanotactile actuators are often larger and more power-consuming than other actuators, such as those use for vibrotactile or electrotactile stimulation. Mechanotactile actuators can vary the force that is exerted on the skin. However, actuators used for vibrotactile and electrotactile feedback provide more information, since more variables are changeable, such as frequency and amplitude for the former and frequency, amplitude, and pulse width for the latter. Lastly, a commonly mentioned distraction during subjective reports is the noise from the motors [142]. However, there are solutions to be found in research investigating other actuators that can be used for mechanotactile feedback that do not involve motors.

Mechanotactile feedback is a strong candidate for providing the user with sensory stimulation to enhance the feeling of ownership. The Rubber Hand Illusion (RHI) has commonly been used to invoke a sense of body ownership. Shehata et al. [169] show that there is a strong positive correlation between synchronous brushing (a brush strokes the hidden hand) and synchronous tapping (linear tactor). Therefore, mechanotactile feedback, which provides the subject with a tapping stimulation, could be a good technique in a sensory feedback system that gives the subject a sense of ownership and location.

Antfolk et al. [170] proposed a tactile display with five digital servos, providing mechanotactile feedback using a hinge mechanism where a button attached to a lever, presses on the skin. The motors were placed in a shape representing the fingertips of an open hand. They investigated how the tactile display influences on the EMG signals that are used to control a myoelectric prosthesis. The results showed that if the EMG sensors and the actuators are placed closely ($\sim$ 1 cm), the actuators would influence the EMG-signals. However, applying a finite impulse response (FIR) high-pass filter (with a cut-off frequency of 20 Hz) could remove the artifacts caused by the actuators.

A passive pneumatic sensory system has been developed, where the passive sensor consists of silicone bulbs shaped as fingertips on a silicone glove of a myoelectric prosthesis. When the passive sensor is acted upon, air will travel through a connected plastic tube and inflate actuating silicone pads placed on the residual limb, applying a small force on the residual limb which is modality matched [162]. This system was used in a longitudinal cohort study to evaluate the amputees' experience of wearing a prosthetic hand with an integrated sensory feedback system in daily life [126]. This system provides somatotopically matched sensory feedback to the amputees, since the actuating silicone pads were placed on the amputees' PHMs on the residual limb (Fig. 4.1a). The amputees experienced that the prosthesis felt like a part of their body and reported a reduction of PLP.

Schoepp et al. [142] designed two different actuators which were implemented in a prosthetic socket. Both tactors used a servo motor but were constructed differently, where one has a linear motion and the second is cable-driven (Fig. 4.1b). The cable-driven actuator has a lower vertical profile, which would take up less space in the prosthesis socket. The linear actuator is recommended if the space in the prosthesis socket is not of concern, since it applies higher force than the cable-driven actuator and consumes less current. Furthermore, the cable-driven actuator has a longer delay between the onset of contact on the sensor and the initial movement of the actuator than the linear actuator.

Borkowska et al. [171] developed a haptic sleeve to provide continuous mechanotactile feedback, with one small motor to pull a thread wrapped around the sleeve to compress and release compression on the user's forearm. The compressing was proportional to the pressure on the prosthetic fingertips as detected by FSRs. The dynamic range of the haptic sleeve was 0–5.1 N. The authors evaluated the impact the haptic sleeve has on adjusting grip force. They concluded that the haptic sleeve significantly increased the success rate in grasping objects. Furthermore, the energy expenditure was low when using the haptic sleeve, meaning there is less muscle fatigue during prosthetic use.

Proprioception is the body's ability to sense its position and movement, meaning body parts can be positioned without any visual feedback. The common suggestion for proprioceptive feedback in prosthetic hands is to provide the user with information about the hand's aperture.

Proprioceptive feedback has been proposed to aid in controlling a prosthetic hand while performing coarse single-DoF tasks in a virtual environment [172]. The results have shown an improvement in targeting accuracy when receiving proprioceptive feedback and when visual input was obscured compared to only visual feedback. For the easy tasks, visual feedback achieved the desired performance level. However, harder tasks combining both proprioceptive and visual feedback have shown a larger benefit in performance than independently.

Rossi et al. [173] developed a wearable haptic device (HapPro) to provide the user with proprioceptive feedback about the prosthetic hand aperture. The system was compared to a cart that rolls on the forearm with a pulley-based system. The position of the single DC motor that moves the fingers of the prosthetic hand was measured and mapped logarithmically to the position of HapPro. They showed that the system

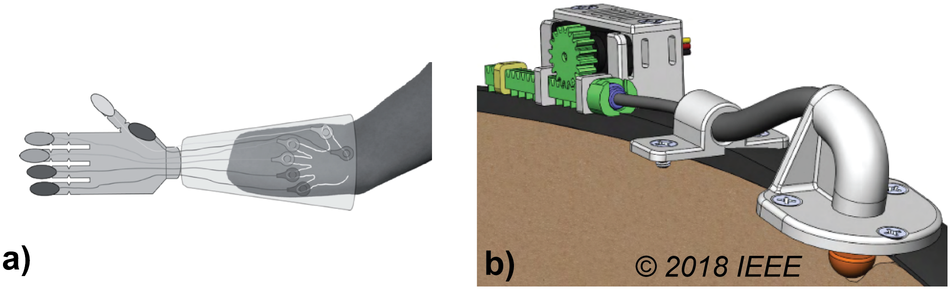

Figure 4.1: Examples of mechanotactile systems in research. a) Pneumatic sensory feedback system with sensing silicone bulbs on the prosthetic fingertips, which are connected to corresponding actuating silicone pads placed on the PHM (reused from Christian Antfolk [162], ©2012 Journal of Rehabilitation Medicine, CC-BY-NC 4.0 license), b) linear tactor where the motor and the head are connected with a Bowden cable (reused from [142], ©2018 IEEE).

should be used on the anterior side of the forearm, since the just-noticeable difference was more sensitive on the anterior side than on the posterior side. They concluded that the system is useful during passive and active exploration, placing an object in the prosthesis hand, and actively moving the prosthesis to grasp an object.

Battaglia et al. [174] developed a wearable single-DoF proprioceptive haptic device (Rice Haptic Rocker), which stretches the skin on the upper arm according to the motor position of the prosthetic hand's aperture. The haptic device stretches the skin with a maximum displacement of 10.5 mm to avoid slipping on the skin. The participants identified sizes of different spherical objects with a multi-DoF prosthetic hand, hence enabling them to estimate the prosthesis aperture. The results showed that the participants could identify the overall level of the aperture. However, it is challenging o convey proprioceptive feedback of a multi-DoF prosthesis using a single-DoF sensor's feedback.

Vibrotactile feedback

Vibrotactile systems provide a mechanical vibration to the skin, and by changing parameters such as amplitude and frequency, the feedback can convey a wide variety of information. However, it is reported to be annoying, cause discomfort, and desensitise the user after extensive use. Vibrotactile feedback is therefore recommended to be used as a short and alerting type of feedback [175].

Small commercial vibration motors are often used to generate vibrational tactile sensations on the skin. Depending on the vibration motor, it can generate vibrations perpendicular or tangential to the skin. Linear resonant actuators (LRA) provide vertical vibrations at a specific resonant frequency, while eccentric rotating mass (ERM) provides a rotational vibration. Both LRAs and ERMs are commonly used in sensory feedback systems due to their small size, low price, and low power consumption.

An ERM [176] contains a DC motor with a mass connected to a shaft (Fig. 4.2a). Spinning the offset mass at different speeds changes the vibration frequency and amplitude. Most commonly, the motors can be controlled by using a microcontroller to generate a PWM (pulse width modulation) signal. The input voltage determines the speed of the motor, which in turn changes vibration frequency and amplitude dependently. Due to the inertia of the mass, the ERM has a slow acceleration and deceleration. A vibrotactile device providing tangential force (shear force) to the skin has been made by combining three miniature vibration motors coaxially. It was thus possible to change both the vibration frequency and the vibration amplitude by the number of active vibration motors [177].

An LRA [178] is a voice coil motor similar to a mass-spring system, which is a resonance system driven by an alternative current (AC) voltage with a resonant frequency in the range of 175–235 Hz. The delivered current will activate the voice coil, which generates a magnetic field, exerting a proportional mechanical force that pushes on a mass containing a permanent magnet. The mass will move up and down in a linear motion and be enhanced by a spring (Fig. 4.2b). Driving the LRA in its narrow frequency range will optimise its power consumption and consume less power at its resonant frequency. The input voltage amplitude can vary the vibration amplitude with a fixed vibration frequency.

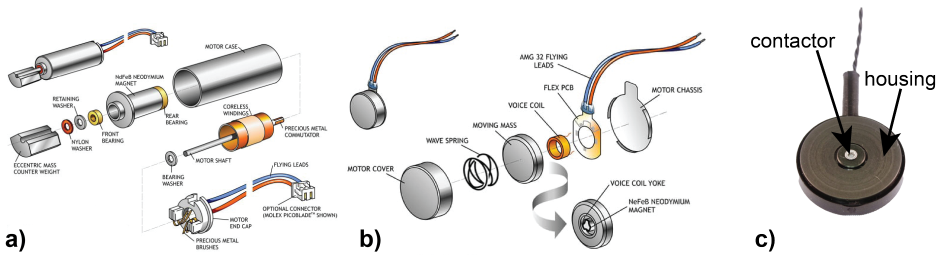

Figure 4.2: Three different vibration actuators. a) ERM (eccentric rotating mass) (photo: Precision Microdrives [176]), b) LRM (linear resonant actuator) (photo: Precision Microdrives [176]), c) C-2 tactor, with a contactor in the middle which oscillates perpendicular to the skin. The housing surrounding the contactor provides a strong point-like sensation (photo: the photo as being taken by Korey Mort, and photo copyright is owned by Engineering Acoustics, Inc. [179]).

Other linear actuators, such as the C-2 tactor and C-3 tactor (Engineering Acoustics, Inc., Casselberry, Florida, USA), are suitable for wearable applications [179]. The operating frequencies are in the 200–300 Hz (C-2 tactor) and 180–320 Hz (C-3 tactor) ranges, and a resonant frequency of 250 Hz and 240 Hz, respectively, which coincides with the skin's Pacinian corpuscles. The actuator contains a magnet attached to a spring, which moves between two switchable electromagnetic poles providing vertical vibrations. An indenter is attached to the magnet and protrudes from the casing, which encloses the rest of the elements [180]. The vibration amplitude and frequency can be somewhat be modulated independently, however with a strong resonant effect. The contactor is shielded by a housing that will deliver a strong and localised vibration (Fig. 4.2c).

Comparing the different vibration actuators, LRAs have several benefits over ERMs, such as faster acceleration (for their size), smaller size, higher efficiency, and a longer lifetime, since they do not contain any brushes which can wear off over time. However, LRAs are limited in frequency due to their narrow resonance peak. Furthermore, to achieve a good performance in acceleration, an LRA has to be driven close to the resonance frequency. However, this can be solved by implementing an auto-resonance algorithm to find the resonance frequency [178]. Huang et al. [181] compared the performance of an LRA and an ERM using a just-noticeable difference and two-point discrimination on the upper arm. They concluded that an LRA is better for binary applications and an ERM is better for encoding more information and conveying complex signals. However, vibrotactile feedback might not be recommended for two-point discrimination tasks because it introduces a sensation that the vibration is ”blurry”, making it difficult to distinguish between one and multiple vibrations. An LRA and C-2 tactor were evaluated when attached to different body sites and during loading [180]. Since all the elements are encased for an LRA, the LRA generates lower forces on the skin, which could affect situations where the user finds it hard to detect the vibration. However, the casing makes an LRA less susceptible to the load placed on the actuator, which provides a stable force that is exerted on the skin.

Vibration actuators placed on the skin introduce waves that propagate. Sofia and Jones [182] evaluated whether the wave propagation introduced by an ERM influences the ability to localise a point of stimulation. It was clear that the mechanical properties of the skin influenced the input from the motors. The vibration had a higher frequency on the palm (glabrous skin) but lower amplitude than on hairy skin. Furthermore, the location of the stimulation was easier identified on the palm. The structure of the skin is a possible explanation, where the glabrous skin has a thicker epidermis, making it stiffer than hairy skin [183]. The highest resonant frequency of 200 Hz has been reported to be on the fingertips, and the lowest resonant frequency of 100 Hz in the centre of the palm [184]. Both hairy and glabrous skin are sensitive at 200–300 Hz. Most vibrotactile actuators are designed to operate within this range [185]. Moreover, the LRA's resonant frequency varies depending on the placement on the skin, which is dependent on the stiffness and damping of the underlying skin [180].

Witteveen et al. [186] evaluated the grasping task performance by providing continuous sensory feedback about hand opening via an array of actuators placed on the forearm in a longitudinal or transversal orientation. An ERM was used for vibrotactile feedback, which was compared to electrotactile feedback. In some experimental conditions, an additional actuator was used for touch feedback, indicating that the hand opening was the same size as the object. There was no difference in grasping task performance between vibration and electrical stimulation without touch feedback, that is using only continuous feedback of hand opening. However, the time taken to complete the task was longer when receiving electrotactile feedback. Adding the touch feedback improved the performance only when receiving vibrotactile feedback. It was reported that it was difficult to distinguish the activation of an electrode in the array and the single electrode for touch feedback. It was also stated by some participants that vibrotactile feedback was more comfortable. The orientation of the actuators resulted in no significant difference in performance, which would make the tactile display flexible when designing a tactile display suited for an individual, depending on the length of the residual limb, where the vertical orientation would fit an amputee with a short residual limb.

Markovic et al. [9] evaluated an advanced vibrotactile feedback device containing four C-3 tactors attached to a rubber band and placed equidistantly on the participants' forearm. The C-3 actuators were activated in different patterns, with the vibrations delivered in different locations and amplitudes to convey information during grasping tasks. The frequency was fixed to 180 Hz due to noise produced during activation. This vibrotactile system provided the amputees with sensory feedback during different life-like experimental tasks to evaluate their ability to control a multi-functional prosthesis. The vibrotactile feedback aided in more complex tasks that required a more challenging prosthesis control than simple tasks, such as opening and closing control.

Alva et al. [187] created a system to provide proprioceptive feedback using vibrotactile actuators (ERM) in a wearable armband. The system provided the user with four vibrational patterns by varying the spatial properties and magnitude of the vibrotactile actuators. The system was tested during a reaching task undertaken in a virtual environment. Testing on eight able-bodied subjects, it was concluded that the system is a strong substitute for proprioceptive feedback. Each participated showed improved performance with different vibrating patterns, indicating that the vibration pattern should be customised for each user. This was specifically recommended for amputees, where the applied stimulation would be perceived differently due to potential muscle and nerve damage post-surgery.

Electrotactile feedback

Electrotactile feedback systems pass an electrical current through the skin. Contrary to vibrotactile technology, the stimulation parameters (amplitude, frequency, and pulse width) can be modulated independently.

Electrotactile feedback does not necessarily match a natural sensation, hence many users are not comfortable with receiving electrical stimulation [175, 188]. However, electrical stimulation elicits informative sensations such as buzzing, tingling, pressure, or vibration, but also less comfortable sensations such as pinching, itching, or sharp or burning pain [189]. Furthermore, the electrodes are lightweight, take up little space, and have an efficient power consumption.

Surface electrodes convey low-amplitude electrical pulses to the skin of the residual limb, which is known as TENS. Electrodes (anode and cathode) are attached to the skin, and the electrical stimulation travels from anode to cathode and stimulates the underlying nerves in the subdermal skin [190, 191]. The sensation can be perceived directly under the electrode, but the sensation can also be spread by placing the electrode near nerve bundles [189]. A stimulation device is used to modulate the parameters, such as the pulse width, amplitude, and frequency. Other parameters which can be altered to elicit different sensations are waveform and electrode size.

Spatial, temporal, and parametric properties are common electrotactile perceptual properties, where the latter includes detection threshold (DT), pain threshold (PT), just-noticeable difference (JND), and parameter-intensity properties (PIP). PIP reflect the sensitivity to electrical stimuli, which is the relationship between electrical parameters and perceived intensity [192]. Zhou et al. [192] proposed three methods to quantify the parametric properties; the modified staircase method, the bisection algorithm, and parametric-random algorithm. They concluded that linear mapping for physical parameters through electrical parameters is inaccurate since pulse width and amplitude are non-linearly mapped to perceived intensity. The perceived effects are highly individualised. It was also mentioned that bisection algorithms using a forced-choice approach to evaluate JND and DT were less susceptible to subjective bias than the modified staircase method. It has been shown that the stimulation frequency is important for control performance, when relaying information regarding grasping force and hand aperture. High frequency stimulation (at least >25 Hz) provides the best perception of electrotactile stimulation [193].

A main drawback with surface stimulation is the change of skin impedance during use. A suggested method, subdermal electrical stimulation, solves the common drawbacks of surface electrodes. The electrodes are placed under the skin through a minimally invasive procedure and have a more stable positioning, and thus also bypass skin impedance [194].

Franceschi et al. [195] developed a sensory feedback system that uses an electronic skin with 64 sensing elements which register mechanical stress and flexible matrix electrodes (32 stimulation electrodes) to provide electrotactile feedback. The participants identified mechanical patterns such as shape, direction, and trajectory. The participants successfully interpreted the stimulation that was delivered to their forearm. The electronic skin had a high sensitivity, making it able to detect slippage.

D'Anna et al. [196] applied electrical stimulation over the residual median and ulnar nerves to provide somatotopic sensory feedback. The electrical stimulation elicited sensations of paresthesia referred to the phantom limb. The sensation was clear enough for the amputees to be able to identify where on the prosthesis they were touched, which was either the whole hand, the ulnar region (little and ring finger), or in the median region (thumb, index, and middle finger). The amputees were also able to generate a desired grasping force.

Other

Some studies suggest auditory cues as a sensory substitution method. Audio-based sensory feedback would require training to learn to associate the sound to the received stimulation on the prosthesis. It has been show that when adding auditory feedback during myoelectric control, the mental effort was lower when controlling the myoelectric prosthesis than with only visual feedback [197].

Lundborg et al. [198] used microphones to pick up friction sound from four different textures. The signals from the microphones were transmitted to earphones and used hearing as a substitute for sensibility of the hand. The participants with non-sensate hands were able to identify the textures, suggesting the system is useful in substituting tactile information with acoustic information.

4.2.2 Invasive sensory feedback

It has been suggested that natural physiological sensory feedback can be restored by stimulating afferent peripheral nerves with electrical stimulation through invasive neural electrodes. Based on the placement of the electrodes, they are classified as extraneural, intraneural (interfascicular and intrafascicular), and regenerative [199] (Fig. 4.3).

The extraneural electrode is the least invasive electrode that does not penetrate the peripheral nerve. Cuff electrodes encircle the targeted peripheral nerve without restricting blood flow to the nerve. A recommendation for cuff design is that the cuff-to-nerve ratio should be 1.5 so as to compensate for nerve swelling [199]. Another extraneural electrode is the flat interface nerve electrode (FINE), which

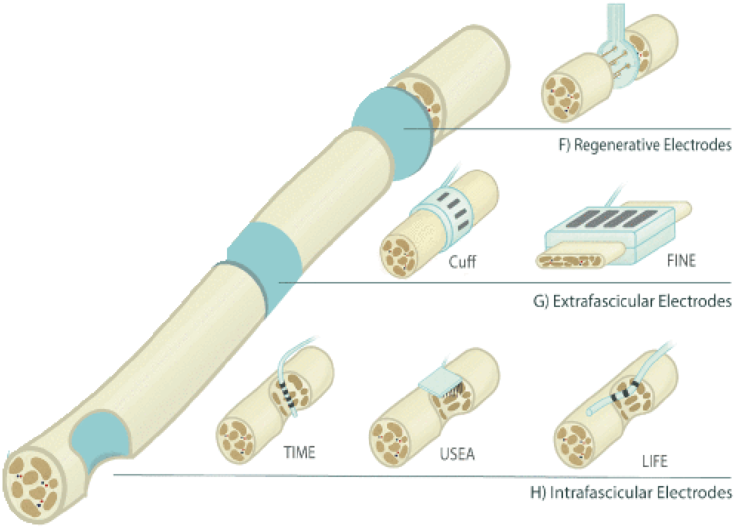

Figure 4.3: Neural stimulation with different implanted nerve interfaces (reused from Jabban et al. [200], ©2022 IEEE, CC-BY-NC 4.0 license.)

clamps the peripheral nerve and deforms it to a flatter cross-sectional area. FINE has the advantage that it provides a more selective stimulation compared to the cuff electrodes, which only stimulate the superficial fibres. With FINE, the fascicle flattens, and the more central fibres will be more accessible. Tan et al. [201] used FINE to provide two amputees with patterned electrical stimulation. The results showed that FINE is stable in the long term (more than a year) and one of the amputees experienced a relief in PLP. Neither of the mentioned electrodes will cause severe damage to the nerves if there is a considerable amount of strain, since they do not penetrate the epineurium of the nerve [202].

For a more selective stimulation of the nerves, intraneural electrodes can be implanted within nerve fascicles, which entails direct contact with the axons. The direct contact reduces the axons' stimulation threshold compared to extraneural electrodes. A longitudinal intra-fascicular electrode (LIFE) targets deeper nerve fibres, where the electrode penetrates the nerve longitudinally and is placed either between the nerve fibres or parallel to them. A transverse intra-fascicular multichannel electrode (TIME) penetrates the peripheral nerve transversally, which targets different fascicles within the nerve [203]. Boretius et al. implanted TIME in rats' sciatic nerves, which showed good results in activating selective fascicles. Both LIFE and TIME need a guided needle during the surgery to penetrate the nerve tissue and guide the electrode to the correct placement, which makes the surgery more complex and time-consuming [132]. The Utah slanted electrode array (USEA) contains an array of 100 electrodes [199]. Normann et al. [204] conducted a study with two transradial amputees, who underwent a USEA implantation into their median nerve and had the electrodes implanted for one month. The USEA could be stimulated from a single electrode or from a group simultaneously. By varying the parameters, the USEA could evoke more than 80 percepts.

Montero et al. [205] presented a myokinetic stimulation interface to restore proprioceptive sensations by remote vibration of magnets, which can be implanted in the residual muscles. Up to four magnets could vibrate when using twelve electromagnetic coils, and it was also possible to activate a single magnet. They demonstrated that the interface could generate directional and frequency-specific vibrations and proposed that magnets could be implanted in independent muscles to provide proprioceptive feedback for a single digit.

Targeted sensory reinnervation (TSR) is a surgical technique where regenerated afferent nerve fibres from the residual limb reinnervates in a skin area near the TMR site. Redirecting nerves to the skin provides a new pathway for cutaneous sensory feedback of the missing hand. Therefore, by touching the reinnervated skin, some amputees have reported feeling their missing hand on the skin [206–208]. This technique enables the patient to control the prosthesis with the residual limb and simultaneously sense touch and receive force feedback [208].

Another invasive approach is to use brain machine interface (BMI), which is a direct communication pathway between cortex and an external device. The signals from the cortex are recorded and translated into commands, which makes it possible for the amputees or patients with spinal cord injury to control an external device. However, the movements when controlling the external device are slow and relatively inaccurate when performing simple motor control tasks, as reported seven years ago [209]. This was suggested to be due to the lack of somatosensory feedback. A biomimetic approach is often mentioned together with intracortical microstimulation (ICMS), where the perceived sensation should be intuitive by reproducing naturalistic pattern of neuronal activity during stimulation. However, the effect ICMS has on cortical neurons are as of yet difficult to predict and ill-understood [209].

Raspopovic et al. [210] conducted a case study to restore natural sensory feedback in an amputee using a bidirectional prosthesis. The prosthesis was equipped with tension sensors to measure the force exerted by the index and the little fingers. The sensor inputs delivered afferent neural stimulation using two TIMEs, which penetrated the median and the ulnar nerve. The amputee performed different tasks, such as a fine control task, a grasping task, stiffness recognition, and object shape recognition. The amputee could effectively modulate the grasping force using the sensory feedback and could discriminate the different characteristics of an object (shape and stiffness).

Schiefer et al. [211] evaluated task performance on two amputees wearing a myoelectric prosthesis. One of the amputees is transhumerally amputated and has FINEs implanted on the median and ulnar nerves and a spiral nerve cuff on the radial nerve. The second amputee is transradially amputated and has a spiral nerve cuff on the radial, median, and ulnar nerves. A stimulator delivered stimulus to one or several electrodes, where the waveform of the stimulus was customised for each amputee by varying pulse width, pulse amplitude, and interpulse intervals. The stimulus provided a sensation of pressure, and one amputee received an additional information regarding the hand's aperture. Tests were done to evaluate the performance of functional tasks (standard ADL tasks), embodiment, and self-confidence when performing the tasks. The sensory feedback improved the amputees' ability to manipulate objects and decreased the number of failures during handling. Furthermore, subjective tests showed an increase in the sense that the prosthesis is a part of their body and that they felt more confident in handling objects.

Moxon [212] presented a ceramic-based, multi-site (CBMS) electrode that included four recording sites with a distance of 200 micrometres between the tip of the electrode. The CBMS was tested on rats to evaluate its feasibility for use in a neuroprosthetic device. Three tests were conducted, of which one was stimulating somatosensory neurons in the cortex. Rats use their whiskers to navigate an environment and discriminate objects, similar to how humans use their hands. During stimulation, action potentials in neurons that usually respond to a tactile stimulus of the whiskers were elicited. With the CBMS, the rats could successfully navigate through a maze. Approaching a split path, they received a stimulus that was passed to the left or right somatosensory cortex to make the rat turn in the desired direction.

Flesher et al. [213] evaluated the properties of the perception of the evoked sensation during intracortical microstimulation. A participant with tetraplegia, where the ability to voluntarily move the upper and lower parts of the body, participated in their study. An electrode array was implanted in S1, and during stimulation, sensations in the thumb, index finger, little finger, and palm were evoked. The participant reported spontaneous sensations (without the stimulation) in the form of tingling, which could be linked to the increased spontaneous firing of recorded neurons in S1. Microstimulation of S1 showed that the location of the sensation was matched somatotopically and provided a naturalistic sensation. It was also observed when manipulating the amplitude of the stimulus that the participant received a graded sensation which was proposed to be necessary during object manipulation when wearing a dexterous neuroprosthetic hand.

4.2.3 Multi-modal haptic sensory feedback in research

Using multi-modal tactile actuators would provide richer information than using only one type of actuator. This would broaden the possibilities to provide stimuli that could be modality matched to the stimulation applied on the prosthetic hand. However, great care needs to be taken to keep the sensory feedback system simple, in order to avoid increasing the cognitive load of interpreting the perceived sensation.

Jimenez et al. [214] used the BioTac®tactile sensor (SynTouch, LLC), fitted in a prosthetic hand, to detect force, vibration, and temperature and to provide direct sensory feedback using a tactile display. A loop with several pneumatic air chambers was used and fitted around the upper arm to convey force feedback. The signal from the sensor regulated the air pressure in the loop linearly and squeezed the upper arm as the pressure increased. A Peltier element was used to convey temperature changes and heat and cool the user's skin. Finally, a C-2 tactor conveyed vibration at a frequency of 250 Hz, and the amplitude was varied in proportion to the reading (vibration intensity) from the sensor. One unilateral amputee used the sensory feedback system to perform three tests to differentiate weight differences, detect contact forces, identify three temperatures (cold, room temperature, and hot), and differentiate roughness in textures. The amputee successfully performed the tasks with the tactile display and could differentiate weights with a weight difference of more than 20%. The amputee reported that the force feedback was constructive for assessing grasp quality and during contact with an object. However, the vibrotactile feedback was found to be somewhat distracting. It was mentioned that it was more effective to use the intact limb to perceive the other tactile properties such as temperature and texture.

Huang et al. [215] developed a multi-modal sensory feedback system containing a sensor array (five sensors) with a piezoelectric barometric sensor, an actuator array (five actuators) providing both vibrotactile and mechanotactile feedback, and Bluetooth low energy communication modules, which allowed a connection between the sensor array and the actuator array. The sensory feedback system was implemented in a prosthetic hand. Five sensors were covered with silicone and attached to the fingertip of a prosthetic hand. The actuator initially vibrated, followed by exerting pressure. Three amputees (of which two had PHMs) participated in the study. The actuators were customised, containing two DC servo motors and one cylindrical ERM, due to their effectiveness in delivering different intensity levels. The placement was adapted to the PHMs, and for the amputee without a PHM, the actuators were evenly distributed on the residual limb. The amputees received three types of sensory feedback – vibrotactile, mechanotactile, and a combination of both – and were asked to identify the stimulus location and intensity, which was applied in three levels. One amputee showed an improvement in the discrimination tasks when receiving multi-modal sensory feedback. All of the amputees mentioned that the vibration served as a warning signal which made them aware of the coming pressure sensation.

Abd et al. [216] recruited twelve participants who participated in their study, of which one had a congenital hand absence. They controlled a dexterous hand with two EMG surface electrodes, where one read muscle activity to control a tripod grip and the other to control the ring and little fingers. The prosthetic hand was equipped with three BioTac®tactile sensors (SynTouch, LLC) on the thumb, index finger, and little finger. A soft armband was designed to convey haptic feedback, containing three air chambers to convey information about the force proportionally to the three sensors. The armband also consisted of three ERMs, which were co-located with the air chambers. The vibrotactile actuators indicated whether an object was broken during the manipulation. The participants' tasks were to grasp, transport, and deliver two objects simultaneously into two separate bins without dropping or breaking them while their vision was obstructed, i.e., the readings from the two muscles were used to grasp two objects simultaneously; the tripod grip gripped a larger object while the ring and little finger grabbed a smaller object. The results indicate that the user could integrate multiple channels of bimodal haptic feedback to proportionally control the forces that were exerted on two objects that were grasped simultaneously. The delivery times of the objects were faster without the haptic feedback but there was a higher rate of grasp failures.

Chapter 5 Summary of included papers

This chapter briefly introduces the included papers. This thesis is composed of three published papers and one manuscript. Each deals with possibilities for non-invasive sensory feedback systems that could resemble the intended sensations obtained by the prosthesis. A passive pneumatic sensor (Paper II) that resembles a fingertip that can be implemented in a silicone glove belonging to a hand prosthesis demonstrated the advantages of producing consistent sensor readings for applications on curved surfaces (fingertip), when compared to commercial sensors (such as FSR and capacitive sensors). Electrical stimulation with mapped frequency created a partially matched sensation when laterally moving a microphone across textures to pick up texture-specific vibrations (Paper III). The received stimulus made it possible to differentiate the textures. Spatially matching the stimulation modalities with the intended sensation induces a more vivid RHI than stimulation modalities that cannot be mapped to resemble the intended sensation (Paper IV). Paper I shows that touch on predefined areas could be associated with specific fingers on participants with intact limbs after a structured training period. This opens up the possibility for amputees without referred sensations to learn the association, which can be sustained for at least two weeks and possibly reduce the cognitive load during prosthetic use.

Paper I: Touch on predefined areas on the forearm can be associated with specific fingers: towards a new principle for sensory feedback in hand prosthesis

This paper investigates whether a touch on predefined areas on the forearm can be associated with a specific finger over a learning period. Thirty-one able-bodied individuals participated in this study. A *tactile display* was developed and consisted of five servo motors which provided the individuals with mechanotactile stimulus, achieved by pressing on predefined points on the forearm, which was placed in such a manner as to resemble the position of fingertips. The individuals followed a computerised training programme for two weeks with follow-ups after one and two weeks. The stimulated areas on the forearm and the individual's response had high agreement, showed a distinct improvement until the third occasion, and showed stable progress for the rest of the period. It can be concluded from this study that it is possible to learn to associate stimulation on the forearm to specific fingers over after a short learning period.

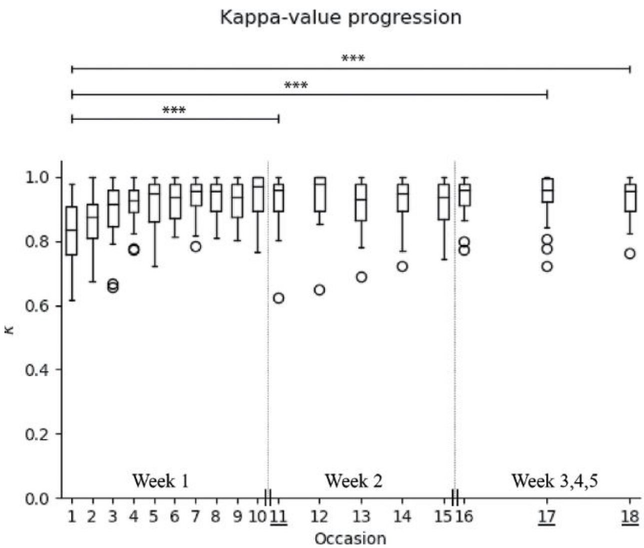

Figure 5.1: The box plot shows the progress in improvement (median kappa-values, 95% CI) over a learning period divided into 18 occasions. There is distinct improvement up until the third occasion, and the improvement between baseline and follow-ups are statistically significant.

Paper II: Characterization of pneumatic touch sensors for a prosthetic hand

An earlier study developed pneumatic touch sensors that can be integrated into a silicone glove for a prosthetic hand. This study performed different tests to look at typical sensor characteristics such as hysteresis, linearity, and impulse response for a single pneumatic sensor. The final test assessed the behaviour of the sensors integrated into the fingertips of a glove. For the tests, a stepper motor was used to apply a force gradually on the pneumatic sensor, whereupon the pressure was measured from the sensor during loading and unloading. The sensor was compressed with different sizes of indenters and at different angles. Air impulses were applied on a single sensor to assess the impulse response. For repeatability, the sensing glove with the implemented pneumatic sensor was placed on a myoelectric prosthesis which was used to repeatedly grasp a cylindric object. The sensor can respond consistently to pressure applied at different angles. It had a maximum hysteresis error of 2.39±0.17% and linearity with an error of 2.95±0.40%. Furthermore, the sensor provides a stronger sensation when handling sharper objects than blunt objects. The pneumatic sensor has a better potential for use in prosthetic hands compared to other commercial sensors, since it is stretchable, does not add any significant weight, and does not contain any electrical elements which could interfere with electrical noise.

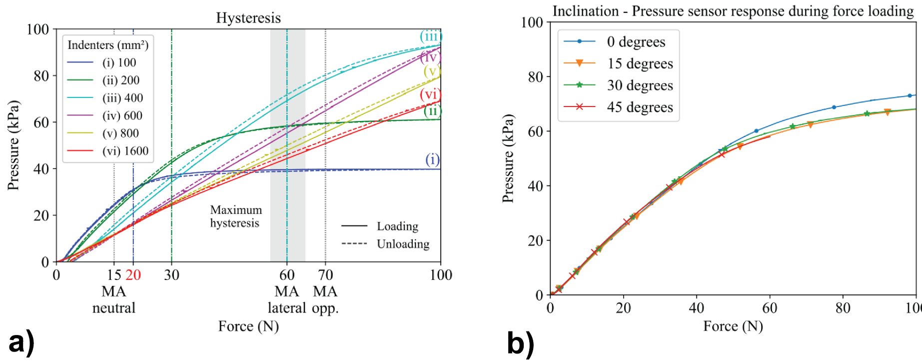

Figure 5.2: Selected results from the paper. a) Hysteresis of the pneumatic sensor when compression was made with different sizes of indenters. The grey area shows the occurrence of the maximum hysteresis (3.69 Pa). b) Consistent reading from the sensor when the indenter was applied at different angles. A difference in pressure level can be seen at different angles at higher forces, which indicates that the indenter hits the rigid bottom.

Paper III: Electrotactile feedback for the discrimination of different surface textures using a microphone

This study evaluates the possibility of using a microphone to pick up friction sound from different textures and process the signal into a characteristic electrical stimulation. The microphone signal was processed to provide a continuous transfer function between the microphone signal and the stimulation frequency. Median frequency was calculated on the transmitted signal, which provided good results in discriminating textures. Twelve able-bodied individuals participated in the study. The participants were asked to identify the stroked texture (felt, sponge, silicone rubber, and string mesh). The experimenter stroked the textures randomly (20 strokes/texture). The experiment was undertaken in three phases (training, with-feedback, and without-feedback). The participants could identify the textures with a median accuracy of 85% with the electrical stimulation, where the frequency was the only variable parameter.

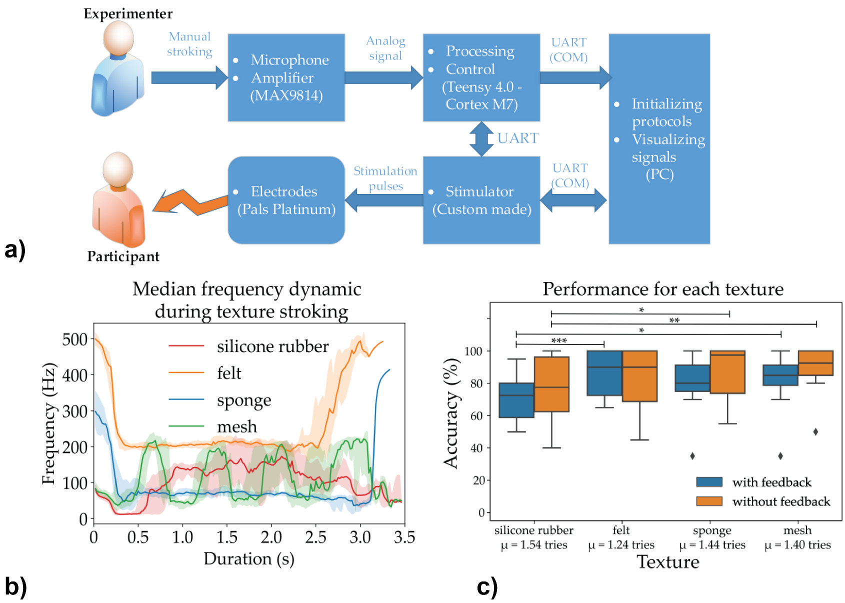

Figure 5.3: A summary of the study. a) An overview of the system. b) The median frequency of the audio signal. c) The box plot shows the performance of the twelve participants on identifying each texture.

Paper IV: The Rubber Hand Illusion evaluated using different modalities

This study used a modified RHI to evaluate common sensory feedback types: mechanotactile, electrotactile, and vibrotactile. This was done in order to investigate whewther modality-conflicting visuo-tactile stimulation could induce the RHI. Twenty-seven able-bodied individuals and three amputees were recruited for this study. The RHI experiment was divided into 10 experimental conditions to test asynchronous brush stroking, synchronous brush stroking, electrical stimulation, pressure, and vibration, all applied on hairy and glabrous skin. Each condition lasted for 100 seconds. The RHI was evaluated using two traditional tests: proprioceptive drift test and a subjective test (questionnaire). A third test used a visual analogue scale to evaluate the pleasantness of each stimulus applied on hairy and glabrous skin. The results showed that stimulus with matched modality elicits a more vivid RHI (brush-brush), but stimuli that was somewhat matched spatially (vibration and electrical stimulation) to the expected sensation (brush) induced a more vivid RHI than stimulus that was modality mismatched (pressure).

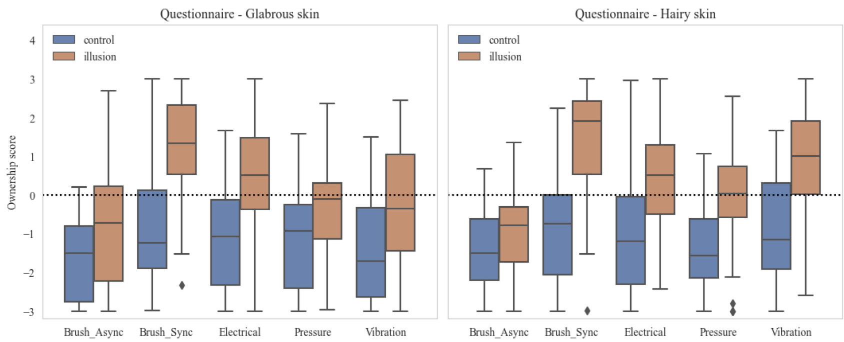

Figure 5.4: Questionnaire results showing ownership scores for each stimulus. The mean rating for the illusion and control statements is seen, where the rates for illusion statements are above the neutral rating (shown as a vertical line).

Chapter 6

Discussion and outlook

Sensory feedback systems in prostheses have long been of interest, and a number of studies have been published to find proper technical solutions to restore the sensory function of the hand. Providing a close to natural and physiological tactile sensation could enhance an intuitive use of the prosthesis. This could strengthen the sense of embodiment of the prosthesis, have a positive impact on quality of life, and less time would be required to learn how to use the prosthesis, which could increase the acceptance rate of using the prosthesis.

This chapter discusses the included papers, followed by a general discussion about sensory feedback systems and its future work.

Paper I showed that stimuli on a pre-defined area on the forearm could be associated with specific fingers after a short learning period. The results open up possibilities for amputees without referred sensation to learn a PHM which is viable for future studies. Other studies have shown an advantage of a present PHM for successfully eliciting sensations related to missing parts of the hand. Consistent placement of non-invasive sensory stimulation could give the subject an intuitive sensation after a short learning period. Limitations in this study were that some of the participants mentioned that they learned to discriminate the servo motors by their sound after a time. The speed and position were set to be the same for all servo motors. However, they generated different sounds/noises depending on how much the skin gave way to the pressure generated by the motor. Due to the positioning of the motors in an upside-down U-shape to resemble the finger's position, the motor placed in accordance with the middle finger would receive more resistance than those placed in accordance with the thumb and little finger, since the distal part of the forearm contains less fat and muscle than closer to the elbow. A solution for this could be to still use the same speed but control the position with force feedback so that each motor would stop when reaching the same contact force. Due to different sounds from the motors, the learning period might be faster than when not being able to use hearing as an aid, which is in accordance with the dual code theory, which holds that learning can be facilitated when involving more senses. However, since the outcome of this study was to evaluate whether touch on predefined areas can be associated with a specific finger and how long the learning period would be, with the benefit of hindsight, noise from the servomotors should have been minimised. On the other hand, there were only a few participants who reported using the motor noise as an auditory cue. Furthermore, for sensory feedback applications, mechanotactile feedback is usually controlled proportionally to the force received on the sensor. Therefore, controlling the servo motors by the measure force would have been more suitable. Lastly, it would be interesting to take this further and compare the learning period with other stimulation modalities to see whether this approach can be used for other non-invasive sensory feedback solutions.

Regarding some of the results when looking at the characteristic of the pneumatic sensor hand (Paper II), it would be interesting to see its continued improvement since it has some potential to be used as a ”sensing prosthetic hand”. The pneumatic sensor has some advantages as it is lightweight and can be incorporated into a prosthetic glove. Furthermore, it can sense pressure at different angles of incidence and is resistant to electrical interference since it does not contain any electrical elements. However, if the pneumatic sensor breaks (most likely because of a tear in the material), the whole prosthetic glove needs to be replaced. Spatial resolution is one of the features a tactile sensor should have, but since the pneumatic sensor is one unitary sensor, it does not possess the ability to discriminate spatial patterns. One solution for this would be to implement a microphone in the glove right outside the air bulb of the sensor. Thus a future suggestion for continuous work could be to implement a microphone to evaluate whether it can pick up sounds from a texture when surrounded with silicone. It would also be interesting to evaluate the linearity, hysteresis, and repeatability when applying pressure on the pneumatic sensor at different speeds to see whether the sensor's response is consistent. Three different speeds could be applied, representing the slowest, medium, and fastest grasping speed of a prosthetic hand. For this, other equipment would be needed since the stepper motor used in this study was a position-controlled motor and would thus not be able to stop immediately when reaching a particular force. Other future work could examine the potential of closing the feedback loop using the pneumatic sensor. The main idea of providing modality-matched sensory feedback would be to implement a mechanotactile feedback actuator, such as motors that provide pressure on the limb, which could be controlled proportionally to the force acting on the pneumatic sensor.

Most of the research into sensory substitution in prosthetics have used experiments in a controlled environment. Paper III considered this and was designed as a worst-case scenario, which means that implementing inconsistent manual stroking of different textures by the operator and not the user would yield an inconsistent feedback pattern. A suggestion would be to use the system in a combined passive and active learning in ADL. Another setup would be to use a glove with the implemented microphone where the subjects are stroking the textures themselves with both the visual and auditory cues obscured. For further development, noise-cancelling technique should be implemented to remove background noise picked up by the microphone. This could be done by adding a microphone to record background noise to be filtered out from the desired signal. However, this would add additional complexity to the design. Another suggestion could be when implementing a microphone in a multi-modal haptic sensor, another sensor used for detection of contact force and movement could activate the microphone only during texture discrimination. In other states, the microphone would be idle. During activation, a solution might be to alter the mechanical design of the microphone so it would amplify vibration from the finger and attenuate the surrounding sound, similar to the function of a stethoscope.

Manuscript IV shows that stimulation modalities with some features that matched the expected sensation (brush stroking) elicited a more vivid RHI than a different stimulus. In this case, the electrotactile and vibrotactile feedback were matched spatially to the classic RHI setup when the participant saw the rubber hand being stroked with a brush. Since the electrotactile stimulation contains a variety of adjustable parameters to generate richer a sensation than vibrotactile actuators, it could be interesting to undertake further experiments to evaluate the RHI. For example, the generated stimulus could be matched with the intended sensation by adjusting the parameters. A modified RHI experiment should be conducted, including actions done in ADL, such as contact force and grasping force feedback, including the different types of feedback: mechanotactile, vibrotactile, and electrotactile. Another observation, which has also been mentioned in other studies, is that electrical stimulation is not a natural sensation in ADL, which was perceived as uncomfortable for some individuals. The stimulation could have been adapted to each individual, so that they receive a sensation that is perceived as a moving sensation similar to the stroking path and yet is comfortable. In that case, the electrotactile stimulation could have contributed to an even higher ownership rate than when the stimulus was weak and only perceived on one spot. The RHI has been questioned regarding being a proper tool for measuring ownership of a fake hand and might be a measurement of the ability to generate experiences to meet expectations. A common limitation is the repeatability of the studies using the RHI since studies use different analysing techniques to conclude whether the RHI has been induced. Therefore, a guideline should be suggested for the RHI in order to increase the repeatability and reliability of the results.

An ideal prosthetic hand would restore both motor control and sensory feedback, which, according to literature, would also enhance the feeling of agency and ownership of the prosthesis. Agency and ownership are suggested to be two components necessary to induce the sense of embodiment of a prosthesis. Some surveys indicate that priority should focus on prosthesis control and comfort. With machine learning and pattern recognition, control of the prosthesis might improve and sensory feedback might be a redundant feature to be implemented in prosthetics. However, inducing the sense of ownership would require an implementation of sensory feedback, which should be as crucial as an intuitive motor control to induce embodiment. An interesting finding by Shehata et al. [169] shows that if the sensory feedback had a significant delay, this could more negatively affect the sense of agency than without the sensory feedback, hence this should also affect the sense of embodiment. It is therefore of utmost importance that the prosthetic user should receive the sensory feedback synchronously with their interaction with the prosthesis. It could be interesting for future studies when required solutions have been found regarding both motor control and sensory feedback to compare the performance of a prosthetic hand with intuitive sensory feedback and prostheses with good motor control to see whether there are any significant differences in the performance. In this way, it is possible to identify what sensory information is most important to aid the prosthetic's control and what sensory feedback is the most intuitive for the user.

In terms of PLP, there is still a lack of data and evidence that adding sensory feedback decreases PLP or whether this contributes to the feeling of embodiment. Furthermore, at the my first International Conference on PLP, held in 2021, it was agreed that PLP should be treated in an early stage of post-amputation and maybe even pre-amputation, since having pain might reduce the use of prosthetics. It was also mentioned that some pain should be treated and reduced before wearing a prosthesis. This could also be a reason why sensory feedback might be a redundant feature in case of diminished PLP. However, the question remains whether adding sensory feedback could prevent a comeback of PLP over longer periods of time.

The major challenge for upper limb prostheses is to fulfil technical and user requirements, such as being dextrous (anthropomorphic movements), being easy to operate, being lightweight, and restoring motor control and sensory feedback without increasing the complexity of doing so. Adding too much hardware to provide unnecessary sensory feedback would increase the system's complexity and increase the cognitive load for the user. Developing a prosthetic hand in the future should consider what needs ought to be met to improve the prosthesis's control and increase the feeling of embodiment (ownership and agency). This should be done without increasing the complexity and the weight of the system. Somatosensory feedback provides abundant information during manipulation of the surroundings and conveys various signals from multiple types of receptors in the human skin. For restoring this sensory function, a lot of sensory qualities need to be considered. Restoring touch non-invasively, multiple actuators need to be included to cover some of the qualities of touch. Nevertheless, actuators could be chosen according to how much sensory information they can convey per actuator. Suggestively, mechanotactile feedback should be one type of feedback, considering its matched modality to most basic ADL tasks. The stimulation modality could be combined with either vibrotactile or electrotactile feedback depending on how much sensory information is desired by the individual. Electrotactile feedback would be chosen for richer sensations due to its variety of changeable parameters, which could be encoded differently depending on which intended sensation the actuator should present. Since electrotactile feedback has shown to bring a variety of sensations, it could be interesting to ascertain whether it can resemble the sense of pressure enough to invoke a sense of ownership. While testing the sense of ownership with the RHI, mostly single modalities have been used, indicating that a single modality is enough to induce a sense of ownership as long as the stimulation modality is matched with the intended sensation. Using the RHI has shown that the sense of ownership is present even if the stimulus is not modality-matched; it is enough if the stimulus is matched spatially.

Implementing multi-modal haptic feedback could increase the system's complexity, which overwhelms the user with too much information, making the prosthesis more challenging to manage. To my knowledge, most research looks into continuous sensory feedback, since it more closely resembles the natural haptic sensations than discrete sensory feedback. Suggestively, the required sensations that should be presented to the user in order to feel that the prosthesis belongs to one's own body or for easier controlling of the prosthesis should be carefully evaluated. An evaluation of which sensations should be presented discretely or continuously to provide the user with enough sensory feedback during manipulation to reduce the risk of overwhelming the user with too much information should be undertaken. For simple tasks, it might be enough to provide discrete feedback, e.g., vibrotactile feedback could be used as a warning signal if gripping too firmly or loosely, whereas complex sensory feedback could be needed in more complex tasks. Previous studies have shown that it is possible to improve the task performance during complex manipulations, though have also increased the time needed to complete the tasks.

The sensory feedback used in research has mostly provided continuous feedback of the grasping force, where the user has to learn and interpret the feedback sensation to grasp an object without slippage or breakage. Grasping an object with a natural hand is an unconscious process where the grasping force regulates to avoid slipping or breaking an object without requiring attention or specific effort from the individual. Resembling natural hand grip adjustment, the feedback signal from sensors could send information back to the prosthesis for an automatic adjustment of the grip so as not to break or drop objects. In this way, the user does not have to put cognitive effort into learning how to interpret the signals to adjust the grip. Instead, the implemented sensory feedback could focus on delivering sensations during the exploration of objects, such as texture discrimination, receiving sensory feedback when making and breaking contact with an object, and receiving feedback during a maintained grip. However, it could also be interesting to evaluate whether this type of presented feedback would negatively affect the sense of ownership, making the prosthesis feel more like an alien hand than belonging to one's body. There could also be an option to the automatic grasp adjustment, e.g., the user could provide a specific muscle movement to switch automatic grasping to manual when reaching a firm grasp to increase or decrease grip strength further.

There are multiple sensory feedback systems in research; however, a majority have been used for short-term testing in labs and separately from the prosthesis. For a sensory feedback system to be wearable, the device should be small, lightweight, and comfortable for the prosthesis user. The next step could be to implement sensors and actuators in a prosthesis to provide multi-modal sensory feedback adapted to the prosthesis user's needs. The system could be tested for an extended period in order to evaluate and improve the system to make the system beneficial for the individual. Furthermore, to test if a complex system can provide with rich information (multi-modal stimulation) without confusing the user and increase the cognitive load. Adapting the prosthesis to the individual's everyday use and behaviour could increase the acceptance of wearing a prosthesis. However, technological limitations should be evaluated for such advanced systems, e.g. if existing processors have enough computational power to cover advanced motor control and hybrid sensory feedback system. Moreover, how long the prosthesis can be worn without charging. Even if an advanced prosthesis with complex motor control and sensory feedback could improve ADL, the prosthesis might still be rejected if the battery needs to recharge multiple times a day. Alternatives to harvest power for additional battery power could be considered, such as harvesting and converting biomechanical energy (body motion) to electrical energy to power the prosthesis with its embedded systems.

This thesis focuses on sensory feedback systems for use in prosthetic hands. However, such technologies are helpful in other applications, such as in controlling a robotic hand/arm operating in remote environments where there's a need for the human operator to receive sensory feedback; force, vibration, or alert/cue. Other applications could be when operating industrial robots which handle delicate objects, sensory feedback could mediate cues to ensure a proper grip of the object, or in surgical robots to aid the surgeon. Sensory feedback can also be helpful in inaccessible environments, such as controlling robots in space or the deep sea for exploration. Also, when operating robots in dangerous environments for proper handling of explosive/toxic objects. There's a broad field where human needs to be included in the system and needs sensory feedback to ensure proper handling.

Chapter 7

Populärvetenskaplig sammanfattning

Handens känsel spelar en viktig roll i hur vi hanterar omgivningen och i sociala interaktioner. Handen är ett komplicerat organ som kan hantera finmotorik, och med hjälp av känselkroppar i huden får vi information om beröring, tryck, temperatur (värme och kyla), vibration, och en känsel om kroppsposition. Amputation får stora konsekvenser, då man går miste om handens känsel och motorik. Utöver att mista handens funktionalitet kan psykologiska faktorer ha en påverkan på den amputerades kroppsuppfattning och sociala liv. Därutöver utvecklar vissa patienter kroniska smärtor efter en amputation, då man känner smärta i den amputerade/förlorade handen. Denna typ av smärta kallas för fantomsmärtor.

Stora forskningsframsteg har under senare tid gjorts inom styrning av proteser, vilket underlättat ersättningen av handens motorik till en viss grad. Det har dock inte gjorts lika många framsteg inom kommersiella proteser för att ersätta känseln. Det finns få kommersiella proteser med ett simpelt känselåterkopplingssystem där användaren får feedback om kontakt och greppstyrka, t ex i form av vibration på armen. Dagens proteser har känts otillräckliga, varför många enarms-amputerade avstår protes då de istället väljer att känna med amputationsstumpen. Att kunna känna med amputationsstumpen försvåras när man bär proteser. Dessutom behöver användaren förlita sig mycket på sin syn för att reglera greppstyrkan i de kommersiella proteserna, för att undvika att tappa eller ta sönder ett objekt genom att greppa det för löst eller för hårt. Detta kan bidra till en mental arbetsbelastning för användaren vilket gör att man väljer att inte använda sig av protesen. Det spekuleras därför i att om man kan implementera intuitiv känselåterkoppling i protesen så kan det tillföra en förbättrad kontroll av protesen och ge en känsla av helhet (känslan av att protesen tillhör den egna kroppen).

Denna avhandling fokuserar på att testa och jämföra olika typer av stimuleringar som kan ges till användaren och som matchar den mottagna stimuleringen. Det vill säga, kontakt med protesen (tryck) ska ges som tryck på den amputerade armen. Detta kallas för matchad modalitet. Stimulering kan även ges så att det matchar spatialt, dvs. positionen, formen och storleken på stimuleringen. Vilket innebär att spatialt matchad stimulering kan t ex både vara stationär (stimulering över större område) eller rörlig (en stimulering som förflyttas över ett större område), och den stimulering som sker på protesen ska förmedlas som densamma på den amputerade armen. Om stimuleringen som protesen utsätts för inte matchar stimuleringen som upplevs måste användaren lära sig att tolka kopplingen mellan stimuleringen som ges på protesen och på amputationsstumpen. Denna avhandling undersöker olika stimuleringar som kan liknas med den förväntade stimuleringen, både i termerna av var stimuleringen känns på och vilken typ av stimulering som upplevs. För att testa om stimuleringarna bidrar till känslan av helhet, jämfördes typerna av stimuleringarna i Rubber Hand Illusion experiment.

Flera amputerade upplever att den amputerade handen (fantomhanden) finns kvar. Genom att beröra amputationsstumpen upplever de en förnimmelse i fantomhanden. Denna typ av förnimmelse kan även kallas för fantomkänsla. Vissa amputerade har även en så kallad fantomkarta (phantom hand map, PHM), som också kan kallas för ”känselkarta”, där man genom beröring på amputationsstumpen kan hitta punkter som representerar delar av den amputerade handen (fingrarna och specifik sida av handen). Studier har gett en indikation på att detta fenomen kan ha uppkommit tack vare hjärnans plasticitet, där hjärnans kartbild över handen har tagits över av amputationsstumpen. På detta sätt har PHM en direkt koppling till hjärnan. PHM kan öppna upp möjligheter för att implementera ett känselåterkopplingssystem som kan stimulera de olika delarna på PHM för att ge användaren en intuitiv känsla och därmed minska den kognitiva belastningen. Denna avhandling har också kollat på möjligheterna att skapa en PHM hos amputerade med avsaknad av PHM.

Sammanfattningsvis, understryker denna avhandling att utveckling av ett intuitivt känselåterkopplingssystem som ger en matchande stimulering för framtida proteser kan lätta på den mentala bördan under hantering av en protes. För framtida proteser bör tekniska lösningar inom styrning och känselåterkoppling testas tillsammans för att ge en bra balans mellan den mekaniska kapaciteten och styrningen för att kunna vara till nytta för användaren. För avancerade lösningar kan bidra till för komplicerade proteser som bidrar till en högre mental börda. Det riskerar istället att användningen av proteser inte är värt mödan och leder till att man väljer att vara utan protes. En välbalanserad teknisk lösning för proteser, som inkluderar rörelser som efterliknar handens mekanik, intuitiv styrning och känselåterkoppling, kan leda till en känsla av att protesen tillhör den egna kroppen, och det är just vägen emot detta mål som detta arbete avhandlar.

Chapter 8

Acknowledgments

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Paper I

Touch on predefined areas on the forearm can be associated with specific fingers: towards a new principle for sensory feedback in hand prosthesis

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ORIGINAL REPORT

TOUCH ON PREDEFINED AREAS ON THE FOREARM CAN BE ASSOCIATED WITH SPECIFIC FINGERS: TOWARDS A NEW PRINCIPLE FOR SENSORY FEEDBACK IN HAND PROSTHESES

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Objective: **Currently available hand prostheses lack sensory feedback. A "phantom hand map", a referred sensation, on the skin of the residual arm is a possible target to provide amputees with non-invasive somatotopically matched sensory feedback. However, not all amputees experience a phantom hand map. The aim of this study was to explore whether touch on predefined areas on the forearm can be associated with specific fingers.**

Design: **A longitudinal cohort study.**

Subjects: **A total of 31 able-bodied individuals.**

Methods: **A "tactile display" was developed consisting of 5 servo motors, which provided the user with mechanotactile stimulus. Predefined pressure points on the volar aspect of the forearm were stimulated during a 2-week structured training period.**

Results: **Agreement between the stimulated areas and the subjects' ability to discriminate the stimulation was high, with a distinct improvement up to the third training occasion, after which the kappa score stabilized for the rest of the period.**

Conclusion: **It is possible to associate touch on intact skin on the forearm with specific fingers after a structured training period, and the effect persisted after 2 weeks. These results may be of importance for the development of non-invasive sensory feedback systems in hand prostheses.**

Key words: artificial limbs; amputation stumps; sensory feedback; upper extremity.

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LAY ABSTRACT

A drawback of currently available hand prostheses is the lack of sensory feedback. Some amputees experience a feeling of touch of the amputated hand when the residual limb is touched. This kind of referred sensation is called "phantom hand map". However, not all amputees experience "phantom hand map". Therefore, we examined whether touch on predefined areas on the forearm can be associated with specific fingers in individuals with an intact arm, using a tactile display during a 5-week training period. In conclusion, it is possible to learn to associate touch on predefined areas on the forearm with specific fingers after a structured training period, and the effect persisted after 2 weeks. These results may be of importance for the development of non-invasive sensory feedback systems in hand prostheses.

Losing a hand is devastating to the individual, with large physical and psychological consequences (1). The loss of sensibility and motor functions is a major problem for the affected individual. Advances in engineering have made it possible to build more advanced hand prostheses with improved grasping alternatives and range of motion (2, 3), but there is no hand prosthesis that is even close to replacing all of the lost functions (1). Control of motor functions in the hand is highly dependent on sensory feedback (4). One priority in prosthetic design that is desirable among arm amputees is how to provide the user with sensory feedback (2, 3, 5–9). It has been shown that sensory feedback improves grasping control and performance with myoelectric hand prostheses in inexperienced users (10, 11). Both invasive and non-invasive sensory feedback systems are under development (2, 9, 12–19).

Following an arm amputation, a phenomenon described as referred sensation may occur. It is described as an experience of touch of the phantom fingers when touching the skin of the forearm and is herein called a "phantom hand map" (PHM) (20, 21). The PHM is unique for each individual and can differ from 1 or 2 diffusely located areas on the residual forearm with referred sensations, to a very detailed map with several specific areas where touch is experienced as touching the lost hand (20, 21). Furthermore, when touching specific areas in the PHM there is cortical activation in the primary somatosensory cortex (S1), which very closely resembles activation seen after touching the different fingers in an able-bodied subject (22). A non-invasive method for sensory feedback in hand prostheses utilizing the PHM has been presented (23).

For non-invasive sensory feedback, a 3-fold process is required; firstly, a registration of the tactile stimuli by sensors is needed, secondly, actuators for transferring the stimuli from the sensors to the user, and thirdly, a process of relearning is necessary with adaptation in

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the central nervous system to interpret the new afferent signals (23). An important issue when designing sensory feedback systems in hand prostheses is how the feedback should be presented to the user in order to be easy to interpret. The most optimal way to present sensory feedback is a combination of modality as well as somatotopically matched solutions (19). Modality matched sensory feedback is when the feedback is analogous to the external stimulation of the prosthesis, and therefore logical in the interpretation for the user. For example, if the fingertips of the prosthesis receive pressure the user should experience the stimulation as pressure (19). Mechanotactile stimulation (pressure) has been proven to be easier to discriminate, compared with vibrotactile feedback (21). Ideally the feedback should also be somatotopically matched, meaning that the individual experiences the feedback as if it was applied to the corresponding location on the lost limb (19). To achieve somatotopically matched sensory feedback non-invasively, the PHM can be used as a target for the actuators of the sensory feedback (21, 24). Some amputees and all congenital amputees lack the PHM on the amputation stump and therefore also lack the possibility to use the PHM as an interface for transferring sensory feedback from a prosthesis (25). Thus, it is interesting to explore if it is possible to learn to associate stimulation on areas on the skin on the forearm with specific fingers of the hand, i.e. to induce an association of touching the fingers when the forearm is touched.

The ability in localizing stimuli in the PHM has been investigated using vibrotactile or mechanotactile (pressure) feedback and pressure stimulation surpassed vibrotactile stimulation in multi-site sensory feedback discrimination (26). A study of 7 amputees has reported that electrotactile feedback in somatotopically matched areas was better than non-somatotopically matched feedback for both accuracy and response time (26). In another study of 11 subjects (9 able-bodied and 2 amputees) electrotactile stimulation was used to compare somatotopically matched areas with non-somatotopically matched areas concerning correct identification rate and response time. Results indicate that areas on the skin without referred sensation (non-somatotopically matched area) of the phantom hand can be learned to be associated with predefined stimulation areas (27).

The aim of this study was to explore whether touch on predefined areas on the forearm can be associated with specific fingers, using mechanotactile stimuli. A further aim was to investigate if the associated sensory learning is influenced by age and sex.

METHODS

The study was conducted during 5 weeks for each participant, on 18 learning occasions, including follow-up at occasions 11, 17 and 18. Each occasion comprised 4 sessions (Table I). Each subject was provided with a silicone cuff to be placed on the forearm with 5 servo motors representing the 5 fingers, and constituting a tactile display that gave pressure stimuli in a pseudo random order during the learning sessions. The subject was seated in front of a laptop with the forearm resting on the table during the sessions (Fig. 1). While given stimulations on the forearm, the subject was provided with feedback on a screen with a photo of a hand with 5 fingers. The user application that was developed for the purpose of this study was used to control the tactile display (Fig. 1), provide the user with visual feedback, and log performance. The main menu of the user application can be seen in Fig. 2a.

Subjects

Able-bodied adults were included in the study and the exclusion criterion was regular medication with drugs that might inhibit concentration and learning. Thirty-five individuals enrolled in the study. The subjects were students recruited from the Faculty of Medicine, Lund University and staff at the Department of Hand Surgery, Skåne University Hospital, Malmö, Sweden.

The study was approved by the regional ethics review board in Lund (Dnr 2012/778) and all subjects gave their written informed consent. The study was conducted in accordance with the Declaration of Helsinki.

Learning protocol and follow-up

All subjects had a personal introduction to the programme and learning by one of the authors (UW), who instructed all subjects. The subjects were given a computer with a programme and the associated hardware, which they used at each learning occasion. The learning occasions were unsupervised during a 2-week period (Fig. 3) and the participant chose the location for training. Following the 2 weeks there were additional follow-up occasions. During the 2 weeks, there were 15 learning occasions in total. In the first week the training was completed twice a day during 5 days chosen by the subject (occasions 1–10). A mini-

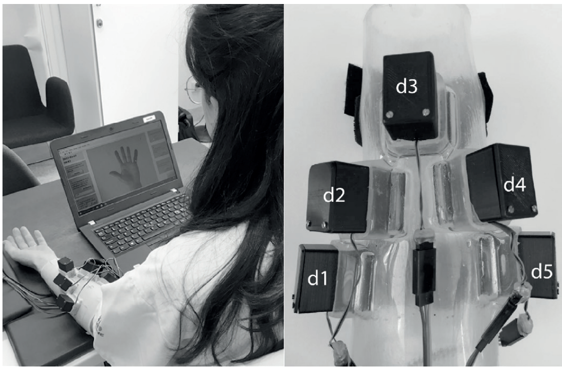


Fig. 1. Left: The training set-up. Right: The cuff that was used on the left forearm, with the servo motors in the black boxes. d1 – Thumb, d2 – Index, d3 – Middle, d4 – Ring and d5 - little.

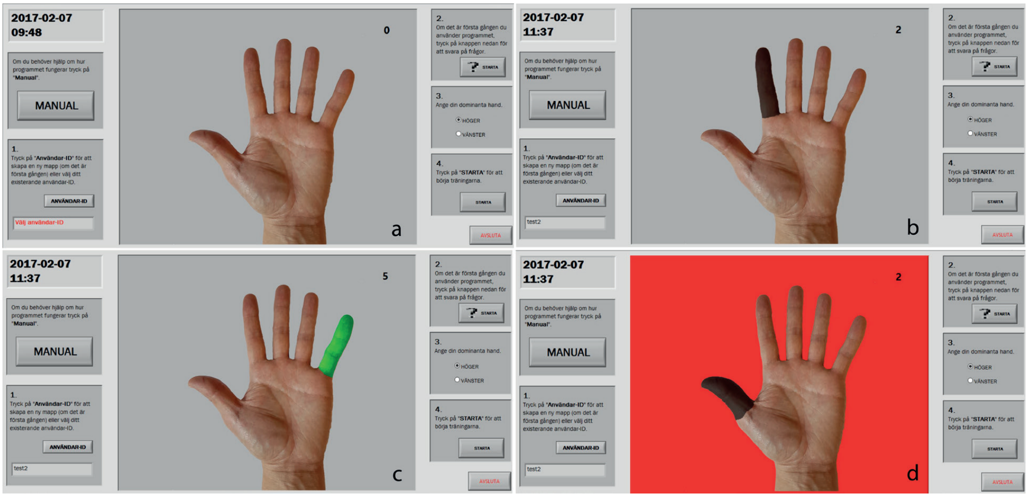


Fig. 2. Screenshots of the programme used during the training, showing the different training sessions. (a) Main menu. (b) Sessions 1, 2 and 4. (c) Session 3, correct answer. (d) Session 3, wrong answer (correct answer, thumb).

mum of 3 h should pass between each learning occasion. The following week the training was done once a day on 5 days chosen by the participant (occasions 11–15). Occasion 11 was the first follow-up and was completed in the same manner as the first training occasion. The second follow-up (occasion 17) took place one week after completion of the learning period, and the third follow-up (occasion 18) 1 week later (2 weeks after completion of the training period, i.e. 5 weeks in total to complete the training period).

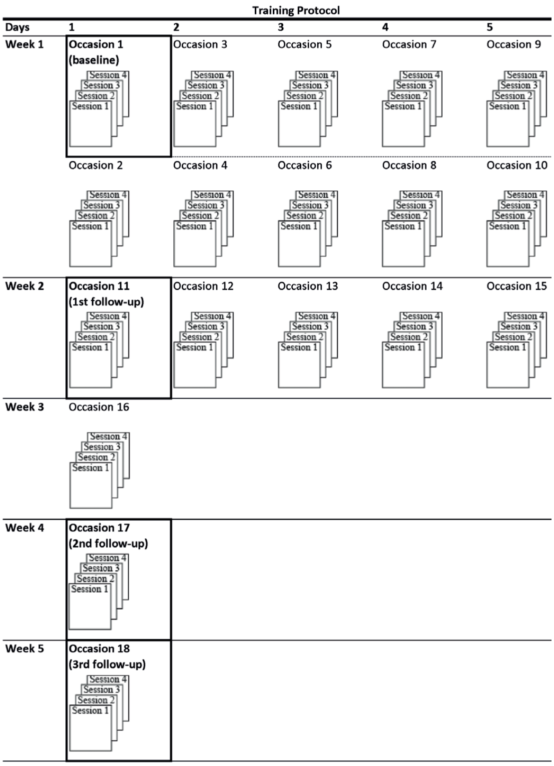
Learning

Each occasion contained 4 sessions (Fig. 3). Each session consisted of 30 stimulations (6 on each finger in a pseudo-random order). The 4 sessions were as follows:

- *Evaluation session;* stimuli were given on the skin of the forearm using the servo motors and the subject indicated the perceived finger location of the stimuli on the computer screen using a mouse (Fig. 2b).
- *Learning session;* stimuli were given on the skin of the forearm using servo motors and simultaneously the programme displayed which finger the stimuli should be associated with (Fig. 2b).
- *Learning with feedback session;* stimuli were given on the skin of the forearm using the servo motors and the subject indicated the perceived finger location of the stimuli and was given immediate feedback as to whether the response was correct (the marked finger turned green) (Fig. 2c) or incorrect (the background turned red and the correct finger turned black) (Fig. 2d).
- *Evaluation session;* Same as the first session (Fig. 2b).

The reason for starting with an evaluation session was to capture the acquired stimulation association from the previous occasion and, in addition, to be able to assess the learning curve between the occasions and not within the same occasion where the sessions are closely executed. Completing an occasion took less than 15 mins. The time lapse between the start of each single stimulation was 10 s.

Fig. 3. Learning protocol. Bold text with square borders denotes occasions for the analysis of the age groups and the learning progress. During the first week the training was done twice a day, the second week; once a day and the follow-up weeks (week 2, 4 and 5) only once. On each occasion, there are 4 sessions.



If the subject did not respond within 10 s, the next stimulation begun automatically and a non-answer was recorded.

Stimulation set-up

The study setup consisted of a *tactile display* (28) using 5 HS-40 Nano analogue servo motors (HI-TEC RCD, USA) incorporated in a silicone cuff with 3D-printed boxes. The boxes were placed in an upside-down U-shape, resembling the positions of the fingertips (Fig. 1), similar to previous work (28). During the sessions, the tactile display was placed on the left forearm. When positioning the tactile display for the first time, the boundaries were marked on the skin of the subject to ensure identical placement between each occasion.

A circular wheel horn was attached to the servo motor axis, which provided a rotational motion. A t-shaped rod was attached to the wheel horn and this mechanical combination converted the rotary motion of the motor to a linear motion of the rod (Fig. 4). The system provided a detectable indentation perpendicular to the skin (5 mm indentation, 17 mm^2 area) with a force that was sustained for 3 s. The distance between the stimulation points on the skin was 40 mm; the minimal distance to detect 2-point discrimination on the forearm (29). The servo motors were controlled by a microcontroller, Arduino Nano, which acquired data from a graphical user interface developed in LabVIEW (National Instruments, Austin, TX, USA) through a serial interface.

The graphical user interface guides the subjects through 4 training sessions, which are described in detail in previous section *Learning*. Prior to each session, the subjects got a descriptive pop-up window about the coming session. The programme was designed to be descriptive, to make sure that the subjects could use it unsupervised. During the sessions, a picture of a hand was shown. The subjects were instructed to select the finger, using a mouse, onto which they associated the perceived stimulation. Depending on which session was running, the subject was given visual feedback about their performance (Fig. 2). The software logged the subjects' information, such as age and sex, along with each subject's perceived stimulation value

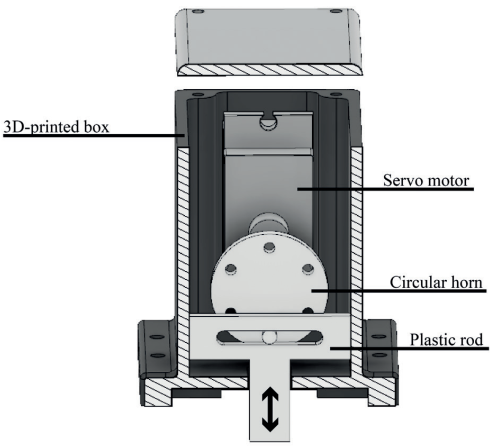


Fig. 4. Cross-section of the 3D-printed box, containing a servo motor with a circular horn, which provides with a linear motion together with the plastic rod, which in turn gives mechanotactile feedback on the skin.

and the actual stimulation value for each occasion and session. At the end of every occasion the subject had the opportunity to leave a comment about complications or other experiences during the learning sessions.

Data analysis

In order to evaluate the agreement between the actual stimulation and the response from the subjects, the linear weighted Cohen's kappa was calculated for each subject. By using a linear weighted model, a response that is more distant fingerwise to the actual stimulation were weighted more heavily than a response that is closer to the actual stimulation. The strength of the kappa value was assessed according to Brennan & Silman (30); values <0.20 are considered *poor,* values between 0.21–0.40 are considered *fair*, 0.41–0.60 *moderate,* 0.61–0.80 *good* and 0.81–1.00 *very good*. The kappa value was calculated for each individual that participated and the median kappa value and 95% confidence interval (95% CI) was calculated for each training occasion.

To determine if the changes in the learning curve were statistically significant the Wilcoxon signed-rank test was used. The kappa value was compared between paired observations; occasion 1 (baseline), 11 (1st day, week 2), 17 (2nd follow-up, week 4) and 18 (3rd follow-up, week 5). These occasions were chosen in order to analyse the learning progress between the first and second week, and also if 1 week without training would affect the new learned skill.

To assess if there were any differences between sexes the 2-tailed Mann–Whitney *U* test was used. This *unpaired* test can determine the differences between 2 groups and it is also useful in small groups (minimum 5).

To determine if there was a difference between age groups the 2-tailed Mann–Whitney *U* test was used. The age groups were divided into 4 different groups; 20–29 (1 male and 10 females), 30–39 (2 males and 7 females), 40–49 (2 males and 2 females) and >50 years (1 male and 6 females).

Both pre-processing of data and analysis were performed in Python, using packages such as Pandas (https://pandas.pydata.org/) SciPy (https://www.scipy.org/) and scikit-learn (http://scikit-learn.org/stable/).

For the analysis, the first evaluation session in every occasion was chosen to evaluate the progression of learning, which shows progression from the previous occasion rather than comparing the progress within a single training occasion.

RESULTS

Of the 35 subjects, 31 completed the study (25 women and 6 men). All but 2 were right-handed. The median age was 37 (range 22–66) years. The 4 individuals who did not complete the study dropped out at an early stage without having to state a reason.

The training protocol was structured, but, in some cases, there were minor discrepancies in the programme, as the subject performed the study unsupervised. A few days of delays in the programme were recorded (2–7 days) for some subjects. In total, 18 learning occasions were planned in the programme, and among the subjects, 16–19 learning occasions were recorded. The cause of this was either due to technical problems, some of the subjects repeated the occasion, or that the subjects missed some occasions.

By examining the learning progress during the training period, the results show that the weighted linear kappa value has a high median value throughout all occasions, and the baseline value was kappa =0.84 (>0.8; considered very good (30)). However, there is a distinct improvement up until the third training occasion (kappa = 0.92) (Fig. 5). The kappa value then stabilizes over the rest of the period; occasion 11 median kappa = 0.96, and continues to be high during the 2 last occasions; occasion 17 median kappa = 0.96 and occasion 18 median kappa =0.96. The improvements between baseline (occasion 1) and the chosen follow-ups (occasions 11, 17 and 18) were significant ($p<0.001$); baseline compared with the 1st day of week 2, 2nd follow-up week 4 and 3rd follow-up week 5. The agreement between actual stimulation and interpretation of stimulation (learning curve) peaked at occasion 12 (median kappa = 0.98). Outliers presented in Fig. 5 were unique individuals at every occasion of different age and sex, and did not follow a pattern that could be used for analysis.

The subjects' ability in distinguishing which finger was stimulated is shown in Fig. 6. It was easiest to distinguish the middle finger, where 95% of the answers were correct. Hardest to distinguish were the ring and little finger (84% correct answers), and most errors occurred when the stimulus was on the little finger and the response was the ring finger (16%).

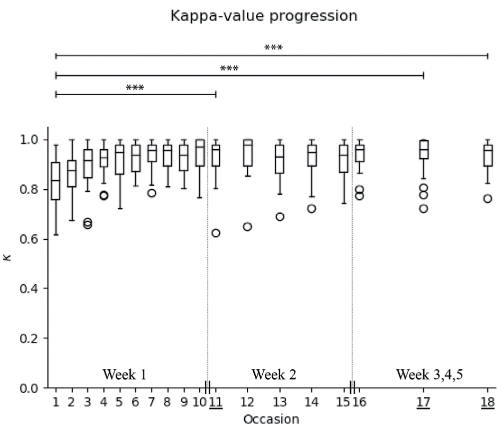


Fig. 5. The box plot shows improvement (in median kappa values) in learning during the 18 occasions, with 95% confidence interval (95% CI). The learning and evaluation was completed twice a day during occasion 1–10, once a day during occasion 11–15 was done once a day and once per week during occasion 16–18. The underlined occasions show the follow-ups. The improvement was statistically significant (***$p<0.001$) between the baseline and the 3 follow-ups.

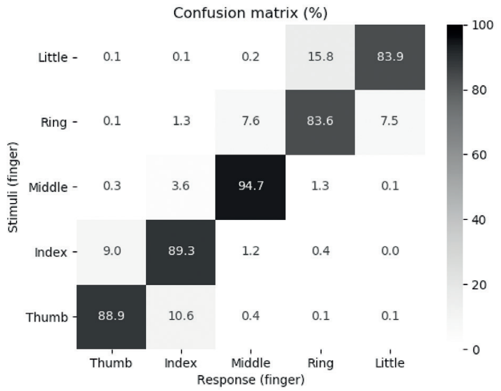


Fig. 6. The confusion matrix shows correct answers (in %). It was easiest to distinguish the area for the middle finger and it was as easy to interpret stimulation on a predefined area for the little finger as for the ring finger. The sum of the numbers shown in the matrix does not add up to exactly 100% due to rounding.

Comparing men and women, the 2-tailed Mann–Whitney *U* test showed there was no statistically significant difference in kappa value between the sexes. Of those participating in the study only 6 were men and 25 were women. The same statistical test also showed that age did not have any influence on learning, when comparing the kappa values of the different age groups.

Among the comments from the subjects a different sensation was described in the predefined area that the subject should associate with the middle finger. The sensation was reported to be perceived as tingling or as a stronger stimulation compared with stimulations of the other areas. Some subjects also reported minor differences in the sound from the servo motors during the stimulation.

DISCUSSION

This study shows that it is possible to induce an association between stimuli on the skin of the forearm with specific fingers following a structured training programme and that the association remain after 2 weeks. The results also show that it is easy to learn to interpret the stimuli on the skin of the forearm, and already after 3 training occasions the agreement between the actual stimuli and the response can be considered very good (30). The excellent agreement remains after 1 week of no training and still after 2 weeks after the end of the training programme. The fast learning that is shown in our group of 31 subjects is comparable with results presented by Chai et al. (27) who reported a "3-day-effect" in their study of 11 subjects during 7 consecutive days. The subjects in our study had a longer learning period and also 2 occasions per day during the first week, which indicates that the subjects learn even faster and within a day. The learning could even have a "3-occasion-effect" with at least 3 h in between occasions. In our study we used mechanotactile stimuli, which has been proven to be easy to interpret for sensory feedback (24). Mechanotactile is also a more common modality to receive as sensory feedback in daily use compared with vibrotactile or electrotactile stimuli, which was used by Chai et al. (27). The very good agreement between stimuli and responses in our study indicates that it is possible to learn predefined areas on the forearm skin that is comparable to referred sensation in capacity to localize the predefined areas. Chai et al. (27) show similar results, and non-somatotopically matched areas reached comparable levels to the somatotopically matched areas considering response time from the actual stimulation to the response of the perceived stimulation, during those 3 training days (27). Our result opens up for the possibility for amputees without referred sensations, as well as for congenital amputees, to learn the association and keep it prolonged for at least 2 weeks. Compared with our experimental learning set-up, prosthesis users would probably wear a prosthesis with sensory feedback more frequently and therefore get more confident with the sensory associations.

Learning as a concept is defined as an encoding of memory and is the process of "gradual changes in behavior as a function of training" (31). In the *dual code theory* there are separate "channels" to process information from different senses. Therefore, multiple senses should be used to facilitate learning, without exposing the working memory to fatigue (32). Three learning styles for adults are described; visual, auditory and kinesthetic, and the best learning is achieved when these 3 approaches are combined (33). In our study we apply visual and kinaesthetic (sensory) information at the same time, and in accordance with the dual code theory and the 3 learning styles this should ease the learning. A well-known concept in psychology and cognitive literature is the *spacing effect* (34). The spacing effect implies that practice is spread over a period of time and the opposite is when practice is massed at one or few close occasions. When the same amount of time is spent practicing, learning is most effective when spaced over time (34, 35). The memory tends to last longer, since spaced learning keeps new cells maintained (36). It has also been shown that the best learning occurs when the practice intervals were expanding over time (37). In the current study the *spacing* effect was applied and the occasions were spaced over a period of 5 weeks. In the first week the training occasions were made twice a day, the second week the training was made once a day and there was an interval of 1 week made respectively for the last 2 occasions. Another concept used in research for learning and memory is the *testing effect*. The effect in long-term memory is better when memory tests are made during the period of practice (38). In the present study every learning occasion included both a pure learning session and testing session where the subject received feedback on the responses. This may have been advantageous for learning.

No difference was seen in learning over time between the sexes. However, the group of men was small, only 6 men participated compared with 25 women, and the lack of statistical significant difference may be due to lack of statistical power. The results did not show any differences between the different age groups.

The U-shape of the tactile display imitates the order and positions of the fingers and may ease the intuitive interpretation of the stimuli of the predefined area with the specific finger. The middle finger was easiest to discriminate, whereas the little finger stimulation was most frequently mistaken, and instead associated with being the ring finger. A possible explanation for this is the U-shape. The stimulation for the middle finger was applied over the flexor tendons to the fingers and the median nerve, and some of the subjects reported a different sensation (tingling), or a stronger sensation of the stimulations of the area for the middle finger compared with stimulations of the other finger areas. The middle finger stimulation was applied in the centre and the most distally on the forearm and might have become a reference for the other stimulated areas which were either on the one or the other side of the middle finger. There was barely any misperception between the stimulations on different side of the middle finger (digit 1↔digit 4), (d1↔d5), (d2↔d4) and (d2↔d5), but it was more difficult to discriminate adjacent fingers (d1↔d2) and (d3↔d4). Nerve innervation is a possible explanation, the 3 radial sites (d1, d2 and d3) were applied to skin that is innervated by the median nerve, and the 2 ulnar sites (d4 and d5) were applied on skin innervated by the ulnar nerve.

Study limitations

Stimulation on the forearm comprised pressure from servo motors, and it is impossible to avoid mechanical noise. Since the speed of rotation of the servo motor was set to be the same, when applying pressure on the pre-defined area on the forearm, the 5 servo motors should sound similar. However, some subjects noticed that some servo motors could slightly differentiate in sound, which may have affected the performance in the progression of learning. According to dual code theory, the involvement of more senses can facilitate learning.

Therefore, for future clinical use, the noise could be an additional sense to enhance learning.

Clinical implication

The long-term aim of the study was to enable amputees who do not experience a PHM to use non-invasive methods of sensory feedback in hand prostheses, as reported previously (23). The present study was performed with able-bodied subjects who all had continuous afferent nerve signalling from the forearm and hand. This is in contrast to a forearm amputee who only has afferent signalling from the forearm. Furthermore, previous studies have shown that plasticity following a change in afferent patterns results in more nerve cells in S1 supplying the forearm area (39). The lack of (competing) afferent signals from the hand and an increased neuronal supply to the forearm leads us to suggest that a person with a forearm amputation would learn to associate touch on specific points of the residual forearm faster than able-bodied individuals.

The PHM is an ideal interface for transferring sensory information from receptors in the hand prosthesis to the amputee. However, some amputees lack a PHM, but the results of this study suggest that it is possible to learn to associate touch on predefined areas on the forearm with specific fingers. For clinical use it might not be necessary to receive stimulation from 5 sites. By applying only 3 predefined areas for stimulation in the U-shape (d1, d3 and d5), it might be even easier to discriminate the stimulations because of the increased distance between the stimulation points. D'Anna et al. (40) has also argued that trying to remember an increased number of received force levels is a cognitive burden, and that it is easier for the subjects to distinguish 3 different force levels than a larger number of levels. This argument could be applied to our study; that it might be easier for subjects to identify only 3 stimulation positions rather than identifying 5 stimulation sites. In a scenario with a myoelectric prosthesis only 3 stimulation actuators, instead of 5, in combination with the wider distance, could therefore make it easier to make adjustments of the areas for sensory stimulation when positioning the EMG electrodes for controlling the motor functions.

Future studies should assess the effects of the described training protocol and the possibility to learn to associate touch on predefined areas on the forearm with specific fingers in amputees without a PHM and in congenital amputees.

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Paper II

Characterization of pneumatic touch sensors for a prosthetic hand
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Characterization of Pneumatic Touch Sensors for a Prosthetic Hand

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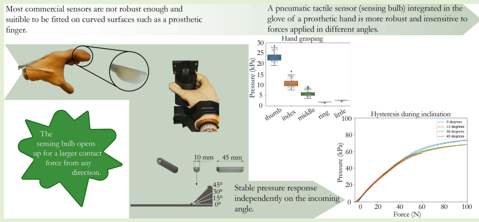
***Abstract*—This paper presents the results from the characterization of pneumatic touch sensors (sensing bulbs) designed to be integrated into myoelectric prostheses and body-powered prostheses. The sensing bulbs, made of silicone, were characterized individually (single sensing bulb) and as a set of five sensors integrated into a silicone glove. We looked into the sensing bulb response when applying pressure at different angles, and also studied characteristics such as repeatability, hysteresis, and frequency response. The results showed that the sensing bulbs have the advantage of responding consistently to pressure coming from different angles. Additionally, the output (pneumatic pressure) is dependent on the size of interacting object applied to the sensing bulb. This means that the sensing bulb will give higher sensation when picking up sharper objects than blunt objects. Furthermore, the sensing bulb has good repeatability, linearity with an error of 2.95±0.40%, and maximum hysteresis error of 2.39±0.17% on the sensing bulb. This well exceeds the required sensitivity range of a touch sensor. In summary, the sensing bulb shows potential for use in prosthetic hands.**



***Index Terms*—Pneumatic, touch sensor, non-invasive, sensory feedback, sensing glove.**

I. Introduction

THE tactile receptors in the human skin are exceptional sensors, the presence of which helps us to interact and explore our surroundings in activities such as manipulation and exploration [1]. There are several types of cutaneous receptors in the skin that detect vibration, force, shear, temperature, and pain. The receptors that respond to mechanical stimuli are called mechanoreceptors, which include Meissner's corpuscles, Pacinian corpuscles, Merkel cells, and Ruffini corpuscles [2]. The mentioned mechanoreceptors detect heavy pressure, vibration, light touch respectively skin stretching [3]. There is high density of mechanoreceptors in the human hand which makes it sensitive to delicate touch where the glabrous skin in the volar part of the hand is more sensitive than the hairy skin on the dorsal part, and the central whorl of the finger pulp being the most sensitive part as it contains the highest density of receptors [4]. In order to perceive different kinds of tactile sensations, all four mechanoreceptor types contribute to the flow of sensory information to the brain where the percepts are formed. Losing a hand entails the loss of thousands of mechanoreceptors but the receptors still remain in the residual limb. However, wearing a myoelectric hand prosthesis, which is a widely used choice among the commercially available prostheses [5], hinders the usage of the remaining receptors in the residual limb and takes away the ability to feel [6]. Thus, leaving the user with only visual input, the sound from the prosthesis, and sensations at the residual limb [7]. For upper limb prosthesis users, providing sensory feedback is highly desired. It has also been shown to improve the motor control of the prosthesis [8] and to reduce the need for visual input. Additionally, it helps the user to adapt to a new prosthesis and to learn how to use it more effectively [9].

The tactile sensing in an artificial hand should be capable of detecting temperature, texture, shape, and force [4]. However, the top priority is to provide feedback that enables grasping (e.g., of an object), touch, and proprioception [7]. Different kinds of sensors have been explored and developed to record sensory input, but few of them have been integrated into commercial prosthetic hands because this often leads to increased cost or to added difficulty during implementation [10].

In prosthetic hands, exteroceptive sensors are used to measure data during interaction with objects and environment. Depending on which sensing techniques are used, detection of normal or tangential forces, vibration, point contact, and temperature can be performed. To be fitted into the prostheses the sensors should meet criteria such as low hysteresis, robustness, and broad dynamic range [11]. The most common

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sensors used in prosthetic hands are force sensitive resistors (FSRs) [10], [12], [13], piezoelectric sensors [14], [15], and capacitive sensors [16]–[18]. Capacitive sensors have good frequency response, spatial resolution, and a wide dynamic range. Such a sensor can detect a normal force by a change in distance between the plates of the sensor or a tangential force by a change in the effective area (overlap of the plates) of the sensor. Nevertheless, it cannot distinguish between these two types of forces [2]. However, they are non-linear, have low accuracy, are more susceptible to environmental noise, and can be influenced by stray capacitance [19]. Resistive touch sensors are often simple, cheap, and have good sensitivity, although, they also have poor repeatability, high hysteresis, and poor frequency response. Furthermore, looking at FSRs, which have a limited active area, the actuator should look a certain way for the FSR to have good repeatability. Typically, the FSR sensors need a mechanical setup that positions the sensing force orthogonal to the FSR surface. The contact area to the FSR sensor should be a flat area that is slightly smaller than the FSR sensing area to provide evenly distributed pressure on the FSR [20]. These characteristics of the FSR make it less recommendable for use in prosthetics hands. This is because the FSR will be placed on a curved surface and the output signal from the FSR will be inconsistent when the force is applied at different angles. A piezoelectric tactile sensor is a suitable candidate to detect slip because it is able to measure vibration due to its very high-frequency response. However, such sensors have limitations. They are only able to measure dynamic force and tend to have poor spatial resolution [1], [2], [21]–[24]. Amongst the sensors mentioned above, FSRs and capacitive sensors are most commonly used to detect contact force. Frequently mentioned reasons are because they are cheap, simple, and lightweight.

Because most commercial sensors only detect a load applied in the center of a sensor, there is now a trend toward making sensors that are flexible and can be attached to a curved surface, such as a fingertip, and also to cover a larger area. Some studies have developed sensing skins [25]–[27], whereas others have developed sensors that use liquids [28]–[30] or air bladders [31], [32]. The latter sensors, integrated into a glove, are constructed to be as anthropomorphic as possible to imitate a real hand. Whether it is on a robotic hand that will interact with humans or on a prosthetic hand, such sensors make the prosthetic more comfortable during interaction and bring to the robotic hand the ability to handle fragile objects. A proposed soft tactile sensor uses a magnet which is immersed in a soft body structured finger, which also consists of a Hall-effect sensor to measure the intensity of the magnetic field generated by the magnet. This sensor detects normal forces [33]. However, such magnetic sensors are susceptible to other interfering magnetic sources and noise [34]. Because of interference of other magnetic objects, integration in robotics is of limited use [35]. The BioTac®tactile sensor (SynTouch, LLC) senses force, vibration and temperature [36] and has been crafted according to the human fingertip to be used in biomimetic systems. In prosthetic hands it has been used to provide closed-loop control of grasping force within the prosthetic hand.

Most commercial prostheses have a silicone glove and in the study by Antfolk *et al.* [31], silicone encapsulated bulbs were developed (hereafter called sensing bulbs) in the shape of fingertips, and were then integrated as a part of the glove. The sensing bulbs cover an area roughly equivalent to the proximal and intermediate phalanx of each finger. Plastic tubes connect the sensing bulbs to other silicone pads that act as a tactile display, which in turn, creates a non-invasive closed pneumatic sensory feedback system. The bulbs of this design spread to provide larger contact force in any direction with an anthropomorphic appearance. We suggest that this kind of pneumatic sensing bulb is robust and insensitive to direction. The design and the integration of the sensing bulbs into the glove is described in the aforementioned study. Consequently, in this study, the tactile properties of the sensing bulbs are characterized so that they can be used in conjunction with processing electronics to provide a finely adjusted transfer function and measure of the force and feedback provided to the user.

The paper is structured as follows. In Section II we briefly summarize some important features of the sensing bulbs that were mentioned in the previous study. Then, in Section III we describe the different ways used to characterize a single sensing bulb and also when the sensing bulbs are integrated into the glove. The results are presented in Section IV, with some further discussion in Section V where the characteristics of the sensing bulbs are compared, theoretically, to other state-of-the-art tactile sensors. A summary of the mentioned sensors can be seen in Table I. Finally, in Section VI, we draw our conclusions and make suggestions for future development.

II. Background: Pneumatic Touch Sensors

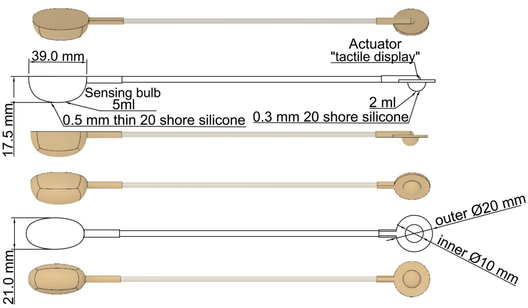
The design of this system was similar to that of an earlier study [31] with some minor changes: the sensing bulbs are of different size and are placed in different positions, only covering the distal phalanx of the rubber glove. It is mentioned in the aforementioned study that the sensing bulbs will be placed individually according to the type of prosthesis and also to how the prosthesis is handled by the user.

The sensing bulbs are made of high-temperature vulcanized (HTV) 20 shore silicone with the Young's modulus of 0.843 MPa. The thickness of the sensing bulb is 0.5 mm and the semi-rigid bottom-support is 1.75 mm with a 65 shore silicone. The semi-rigid bottom is to withstand the created pressure in the sensing bulb, resulting in bulging only on the sensing part of the sensing bulb. This eventuates a more accurately reading of the applied pressure. Some conclusions about the design was made in Antfolk et al. study during the development phase [31], where the thickness of the wall has to be compromised between durability and sensibility, therefore, the thickness should not be too thin nor too thick. The durability and the sensitivity were desirable in the design to make the sensing system durable to fit in a prosthetic glove, and meanwhile, provide with enough sensitivity to feed back the sensations. The force is sensed through the sensing bulbs and is mediated using air in a plastic tube that is connected to an actuator, providing the amputee with sensory feedback

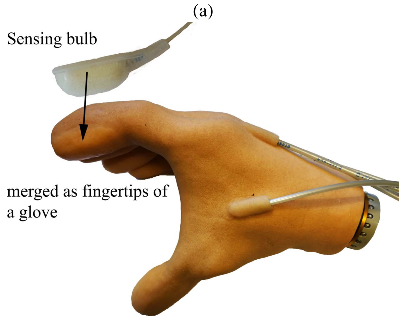
TABLE I
THEORETICAL SUMMARY OF THE PROPERTIES OF THE SENSING BULB AND THE COMPARED STATE-OF-THE-ART SENSORS

	Advantages	Disadvantages
Sensing bulb	✓flexible ✓low hysteresis ✓good precision ✓large sensitive area ✓consistent response to applied pressure at different angles	✗low spatial resolution[a] ✗output inverted dependent on size of object[b]
Capacitive sensor	✓spatial resolution ✓frequency response ✓wide dynamic range	✗very low precision[c] ✗non-consistent response to applied pressure at different angles
Resistive touch sensors	✓cheap ✓good sensitivity ✓large sensitive area ✓simple to use/connect	✗low precision ✗high hysteresis ✗output dependent on area size of object[d] ✗non-consistent response to applied pressure at different angles
Piezoelectric sensor	✓high frequency response	✗detects only dynamic force ✗output dependent on area size of object[d] ✗non-consistent response to applied pressure at different angles

[a] With the current design, spatial resolution is bad as the sensor is fairly large.
[b] Smaller area larger response, which actually could be an advantage, see discussion.
[c] Low precision due to; highly non-linear, susceptible to environmental noise. Often used as on/off sensor.
[d] Larger area larger response.



(a)



(b)

Fig. 1. (a) Pneumatic Sensing System: The sensing system is seen in different angles with dimensions. The actuator is seen in its active state, i.e. applied pressure on the sensing bulb makes the actuator bulge against the skin. When there is no pressure on the sensing bulb, the actuator is flat. (b) Sensing bulb: Top image, a single silicone bulb (not integrated into glove), Bottom image, glove with merged sensing bulbs in the fingertips for the MyoHand VariPlus.

on the residual limb when the silicone bulge against the skin (Fig. 1a). The sensing bulb is shaped to imitate a real fingertip (distal phalanx) (Fig. 1b) and having polyurethane foam inside to make the sensor stiffer and to facilitate a quick return to its initial form (e.g., after an object is released).

In this study the sensing bulbs were characterized by different experimental setups for an individual sensing bulb (top image in Fig. 1b) and for a prosthetic glove with five integrated sensing bulbs.

III. CHARACTERIZATION METHODS

The set-up of the experiments contained i) a VEXTA PK254-E2.04A stepper motor (Oriental Motor Co.,LTD., USA) having a step size of 1.8°/step ii), a force gauge (MARK-10, USA) with accuracy of ±0.1% full scale and a resolution of 0.02 N. This was mounted on iii) an miniature linear positioning system (Parker 402002, Parker Hannifin Corporation, Pennsylvania) that drove the force gauge on the z-axis to press on the sensing bulb to make it deflate and inflate (Fig. 2a). The stepper motor was driven using a NI MID-7604 stepper motor driver (National Instruments, USA). A setting of 10 microsteps per step was using on the motor driver. The linear positioning system uses a 1 mm lead screw, leading to a displacement of 500 nm/step of the system. The speed of the positioning system during the experiments were set at 5 μm/s. The speed is lower than the normal grasping speed, but could eliminate the rate dependencies.

Different sizes of objects with rectangular surfaces were attached to the shaft of the force gauge to act upon the sensing bulb. An integrated pressure sensor (MPXV5100G, NXP Semiconductors, The Netherlands) with accuracy of ±2.5% V_{FSS} and a sensing range of 0-100 kPa, was used to measure the pressure from the sensing bulb. The data was analyzed to evaluate the characteristics of the sensing bulb, such as, hysteresis, frequency response, and repeatability. Experiments, explained in section Sections III-A to III-C, were done on a single sensing bulb. While, for the experiment, in section III-D, tests were made on five different sensing bulbs that were integrated into a silicone glove [31] covering a MyoHand VariPlus Speed (Otto Bock, Germany).

LabVIEW, a visual programming language and environment, was used to control the measurement setup. Multiple tasks were done in LabVIEW: control of the stepper motor

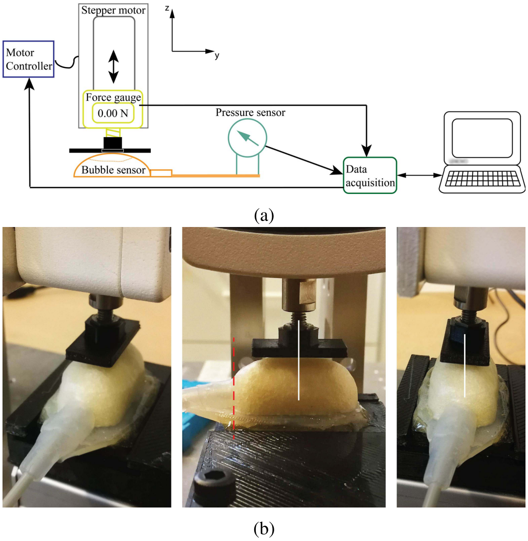


Fig. 2. (a) Experimental setup for Force-pressure characterization of the sensing bulb: Motor controller moves the gantry with the attached force gauge on the z-axis to press the sensing bulb and to measure the force applied to the sensor. The pressure sensor measures the pressure induced from the sensing bulb in the silicone tube. Further analysis was done after data acquisition. (b) Photograph of the positioning of the sensing bulb and the indenter. The center of the indenter was inlined with the center of the sensing bulb as the white line. However, the indenter should not exceed the dashed red line which is the attachment to the plastic tube.

in the z-axis, gathering of the data from the force gauge and from the pressure sensor using a NI USB-6341 DAQ device (National Instruments, USA). The DAQ device was also used to control the prosthetic hand in the second part of the measurement, which included a fully sensorized glove. The update rate of the experimental setup was set to 10 Hz.

Post-processing and visualization of data were conducted using Python with packages such as Pandas (https://pandas.pydata.org/). From the measured output voltage of the MPXV5100G sensor [37], the pressure was calculated according to equations supplied in the datasheet.

A. Setup: Compression With Indenters of Different Sizes

This experiment was done on a single sensing bulb, to evaluate how the sensing bulb behaved when it was compressed using objects of different sizes. The objects were rectangular plates, which acted as indenters in the experiments. The indenters were 3D-printed, using PLA filament (Young's modulus, 2.960 MPa), according to the sizes displayed in Table II. The size of indenter no. (v) was the same as that of the base of the sensing bulb, but when the sensor was compressed, the silicone of the sensing bulb widens, exceeding the size of the indenter. Therefore, an indenter twice the size (indenter no. (vi)) was chosen to cover the volume exceeded. The stepper motor was set to move in increments of one step during loading and unloading. The indenters were changed after each measurement, but the sensor remained

TABLE II
DIFFERENT INDENTERS

No.	Indenter surface area regardless of the area of the base of the sensing bulb	Base Area (mm^2)
(i)	12.5%	100
(ii)	25%	200
(iii)	50%	400
(iv)	75%	600
(v)	100%*	800
(vi)	200%	1600

* the same size as of the base of the sensing bulb.

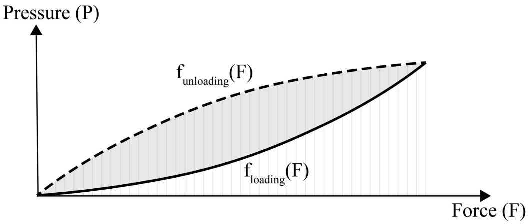


Fig. 3. Integral hysteresis definition: An approximated area is calculated between the two functions $f_{loading}$ and $f_{unloading}$ using a step size of 0.5 N. To get the error in percentage the approximated area is divided by the area for the function f_{loaded}.

at the same fixed position. The stepper motor slowly lowered the force gauge to apply pressure on the sensing bulb until the force reached 100 N (loading), and thereafter raised the force gauge until reaching 0 N (unloading). The slow pace of changes in pressure ensured isothermic conditions within the air compartment and the silicone membrane. It was chosen to apply pressure where the force gauge read up to 100 N, even though no more than 20 N is required for sensitivity range [4]. This was done to examine the characteristics of the sensing bulb under an extreme condition.

For the hysteresis evaluation, both loading and unloading phases were included. The hysteresis was only calculated for the three biggest indenters, (iv)-(vi), because of their linearity in the force range 0-100 N. Hysteresis was calculated for each indenter. The hysteresis was calculated according to the equation below:

$$hyst\% = \frac{\int_{x=0}^{100} f_{unloading}(x)dx - \int_{x=0}^{100} f_{loading}(x)dx}{\int_{x=0}^{100} f_{loading}(x)dx} * 100$$

The area between the function of loading and unloading state was calculated with the integral definition, which was divided by the loading area to get the hysteresis error in percentage (Fig. 3).

In the further analysis of the data, only the linear part was considered. Therefore, different end-forces were chosen for indenters of different sizes.

B. Setup: Compression From Different Angles

This measurement setup evaluated the angular dependency of the output of the sensing bulb when pressure was applied at different angles. This experiment was done on a single sensing bulb. Furthermore, four different mounting plates were 3D-printed, and then used to press against a tilted sensing

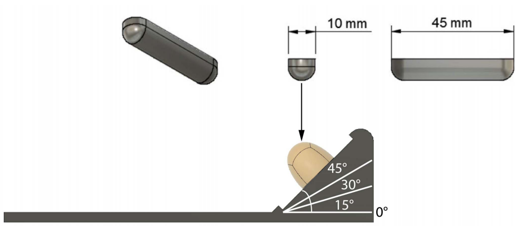


Fig. 4. Cylindrical indenter used in compression from different angles: The rounded bottom of the indenter was applied to the sensing bulb. Top image, on the left, seen from the bottom; In the middle, seen from the short-side; On the right, seen from the long-side. Bottom image shows the attachment used to position the sensing bulb in four different tilted positions: 0, 15, 30, and 45°.

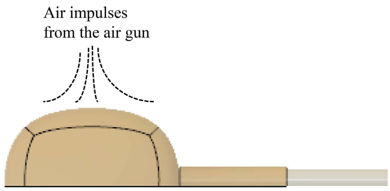


Fig. 5. A soft air gun was used to apply six successive air pulses to a single sensing bulb.

bulb at different angles (0°, 15°, 30°, and 45°). The indenters were pressed against the sensing bulb until the force gauge reached 100 N, however, at larger angles the pressure saturated before 100 N. Therefore, with the largest angle the loading stopped when the force gauge reached 60 N. A cylindrical indenter was 3D-printed to avoid uneven deformation of the sensor at the edges of the indenter (Fig. 4). The cylindrical indenter had a diameter of 10 mm and a potential contact area of 706 mm. The cylindrical indenter was aligned to the sensing bulb. Between the measurements with different pressure angles, the sensor position was adjusted to remain centered (although tilted) with respect to the centerline of the mounting indenter.

C. Setup: Impulse Response

The evaluation of the cutoff frequency was done by looking at the impulse response. A soft air gun was used to apply impulses to a single sensing bulbs (Fig. 5). The soft air gun used a 12 g CO_2 capsule. When the trigger was pulled, a pulse of air hit the sensing bulb. After each firing the CO_2 decreased and thereby, the amplitude of the impulses decreased. The soft air gun was mounted just above the sensing bulb, and six impulses were generated manually by pulling the trigger. A Fast Fourier transform was applied to the extracted sensor responses sampled at 100 kHz.

D. Setup: Glove With Sensing Bulbs

To assess the behavior of the sensing bulb in a realistic scenario, a measurement was conducted on five sensing bulbs that had been integrated into the glove as fingertips. The myo-electric prosthetic, with the sensing glove, was mounted and

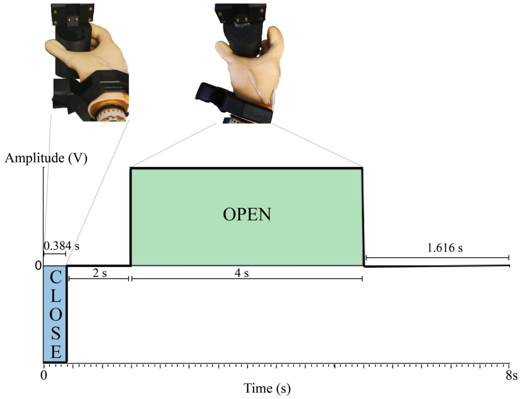


Fig. 6. (a) Control signal for the prosthetic hand. Amplitude: 0.5 (150 mm/s); Closing duration: 384 ms; Opening duration: 4 s; Time frame: 8 s. Duration between closing and opening was 2 s. Top figures illustrate the sensing bulb hand during the the closing and opening state.

remained in a fixed position during all the measurements. The MyoHand VariPlus speed was controlled by modulating a square wave signal in LabVIEW. The modulated signal was set by the amplitude and duty cycle, which defined the closing and opening speed of the hand, respectively, and how much the hand was opened or closed. The amplitude was set to 0.5, which is 50% of the hand's speed (300 mm/s), in other words, during the experiment the hand was moving at the speed of 150 mm/s. The sensor output was measured, and collected in LabVIEW, for all five sensing bulbs while the hand was repeatedly grasping and releasing an attached 3D printed cylindrical object (ϕ 60 mm) resembling the shape of a 0.5 L bottle. To investigate the sensor response repeatability, the hand grasped the object 100 times. The grasp was maintained for 2 s before release. The closing duration lasted for 384 ms and the opening duration lasted for 4 s to ensure that the hand opened completely. The hand remained open the last 1.616 s before repeating the grasping sequence. The change from close to open state happened within a time frame of 8 s. The experimental setup can be seen in Fig. 6.

IV. Results

A. Compression With Indenters of Different Sizes: Hysteresis

Driving the stepper motor until the force gauge reached 100 N showed a monotonically increasing non-linear curve with the indenter no. (i)-(iii). This naturally indicates that the indenter compressed the sensing bulb against the rigid bottom of the sensor earlier with smaller indenters, than with the larger indenters. As can be seen in Fig. 7, the curve becomes saturated with the smaller indenters (no. (i)-(iii)) because the indenter reached the sensor base at around 20, 30, and 60 N respectively, while for the larger indenters the curves are linear until 100 N is reached. Looking at the three biggest indenters with linear curves (indenter no. (iv)-(vi)), the maximum hysteresis occurs at 60.2±4.59 N (marked as the gray area in Fig. 7) with an error of 2.39±0.17% at the full-scale range

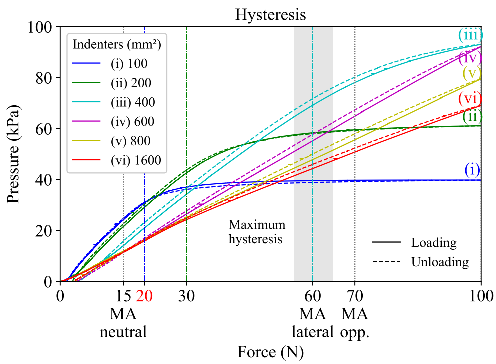
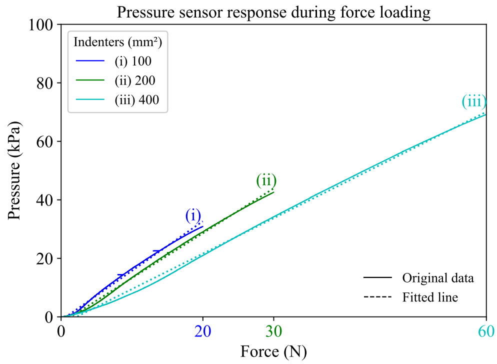


Fig. 7. Hysteresis of the sensing bulb: The solid lines represent pressure added (loading) to the sensing bulb and the dashed lines represent when pressure relieves (unloading). The vertical lines represent where the linearity occurs for each indenter and also the MyoHand VariPlus Speed, which has proportional gripping force of 0-100 N [39]. The force marked in red, indicates the maximum detected force required in the prosthetic hands. The gray area shows the occurrence of the maximum hysteresis for the (iv)-(vi).

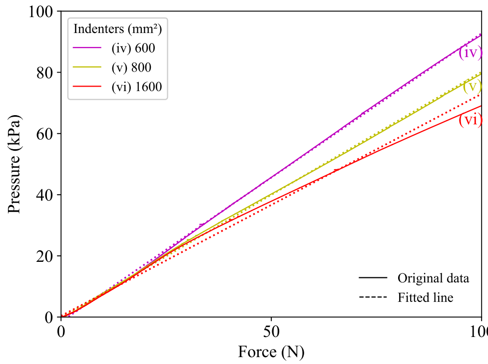
of the measured force (0-100 N). The error is less than the maximum error ($< 5\%$) suggested for sensors used in an artificial hand [38]. The reason for the hysteresis might be induced due to the characteristics of the silicone glove, such as the visco-elasticity of the silicone. Moreover, the figure shows that the sensor output is higher during unloading than during loading. This indicates that the sensing bulb was more compressed during loading. The maximal pneumatic pressure difference between loading and unloading was 3.69 Pa. The figure also compares the pressure applied on the sensing bulb with the maximum gripping force generated by the Otto Bock prosthetic hand, MyoHand VariPlus Speed [39].

B. Compression With Indenters of Different Sizes: Linear Regression

The response of the pneumatic pressure, when the sensing bulb was compressed by indenters of six different sizes, is shown in Fig. 8. It can be noted that the sensory output was approximately linear when forces below 20, 30, 60 and 100 N were applied with indenters (i), (ii), (iii) and (iv, v, vi), respectively. pressure was applied on the sensing bulb. For the three smallest indenters, (i)-(iii), the sensing bulb output was approximately linear until the force gauge reached 20, 30, and 60 N, respectively, when applying pressure on the sensing bulb Fig. 8a. For the remaining three indenters, (iv)-(vi), the sensor readings were approximately linear within the full range (0-100 N) (Fig. 8b). However, with a non-linearity for the biggest indenter (vi). Using linear regression, the coefficient of determination, R^2, was 0.9957, 0.9971, 0.9969, 0.9995, 0.9996, and 0.9960 with respect to increasing size of the indenters. The root-mean-square-error (RMSE) was 0.6025, 0.6806, 1.1101, 0.5762, 0.4592, and 1.1794 kPa. This implies that over 99% of the change in pneumatic pressure, for all the indenters, was related to the pressure applied. While $< 1\%$ depended on other variables, such as variable mechanical



(a)



(b)

Fig. 8. Force-pressure curve: Pneumatic pressure measured when the sensing bulb is compressed by indenters of different sizes: relatively small (a) indenter no. (i)-(iii) or larger (b) indenter no. (iv)-(vi). The curves show linearity at different load forces, where the end-force for each indenter was chosen as the point where the indenters reach the sensing bulb base. The dotted line shows the fitted line obtained from a linear regression.

properties of the sensing bulb because of its soft nature. It can be speculated that if the wall of the sensing bulb would have been thicker, the sensing bulb could have managed greater force. However, this would occur at the expense of the minimum force sensible. The deviation could also depend on the pressure sensor utilized: MPXV5100G [37], which had an accuracy of $\pm 2.5\%$ V_{FSS}. It can also be concluded that the calculated slope constants decrease when the sensor is deflated, when applying pressure with larger indenters.

C. Compression From Different Angles

As shown in Fig. 9, the characteristics of the sensing bulb remains the same when the pressure is applied at different angles. There is a difference in the pressure level depending on when indenter reaches the sensor base, resulting in pressure saturation for the sensing bulb. The reason for this is that, when the angle is larger, there is less area to counteract the pressure

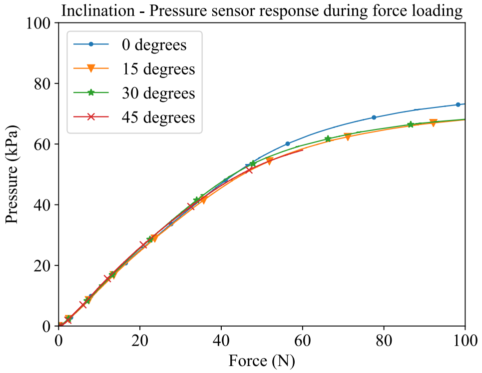


Fig. 9. Loading and unloading on the sensing bulb at different angles.

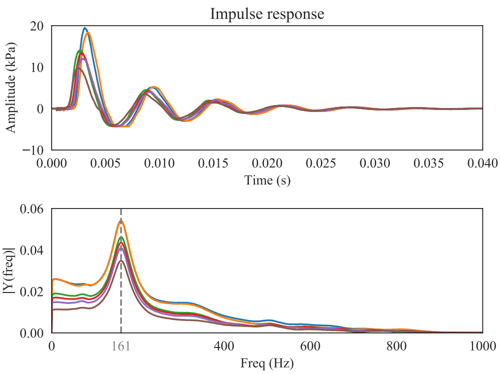


Fig. 10. Six manually generated impulse responses: top image, Response from the gun shot, bottom image. FFT analysis showing there is a resonant frequency at 161 Hz.

from the indenter. The consequence is that the indenter is not able to press with a great force. The contact behavior between the cylindrical indenter and the sensing bulb is different depending on the angles and during the loading on the sensing bulb. The contact area increases during loading. The two materials, silicone (sensing bulb) and plastic (indenter) create friction that stretches the sensing bulb and initiates greater tension on one side, whereas on the other side the mass gathers under the indenter when the angle increases. With no angle on the sensing bulb the mass encloses the cylindrical indenter.

D. Impulse Response

The response of the sensing bulb when generating an air impulse response is shown in Fig. 10. The amplitude of the impulse amplitude was between 10-20 kPa. The sensing bulb, with its resonant frequency at 161 Hz, covers the detectable vibrotactile frequency range of a Meissner corpuscle (3-40 Hz), a Merkel disk (0.4-3 Hz), and part of a Pacinian corpuscle (40-500 Hz) and Ruffini endings (100-500 Hz) [3], [19]. The sensing bulb had a rise time of 0.57 ms, which is within the range specified in the guidelines for a tactile sensing system in which the response time was limited to 1 ms [21].

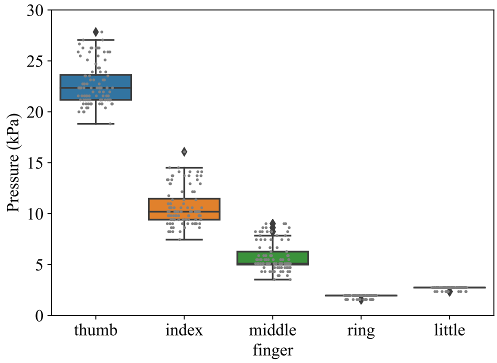


Fig. 11. Repeated measurement when the sensing bulb grasps an object 100 times, looking at the maximum pressure in each grasp.

TABLE III
PARAMETERS FOR THE SENSING BULB

Parameter	Value
Main advantage	Indifferent to angles
Hysteresis	2.95±0.40%
Coefficient of determination	>0.99
Sensitivity	0.82 kPa/N
Resonant frequency	161 Hz
Rise time	0.57 ms
Range	0-9.020 kPa (for the thumb)

The values in the table are based on the results when using the indenters (iv)-(vi), where the response did not saturate.

E. Glove With Sensing Bulbs

The data from the repeated measurements can be seen in Fig. 11. Because the MyoHand VariPlus Speed is a single-degree-of-freedom tripod prehensor, it forms a palmar grasp with thumb, index, and middle finger. This explains the results for the ring and little finger in the figure, where not enough pressure was put on the sensing bulbs to give a noticeable response. The prosthetic hand was placed to give a maximum response at the thumb and index finger and less at the middle finger. It could be noted that the median force measured by the thumb sensor was higher than on the other sensors. This is expected because the force is distributed among the opposing fingers. The repeatability of the sensing bulb hand shows a maximum discrepancy of 9.020 kPa for the thumb and a minimum discrepancy of 0.392 kPa for the ring and little fingers.

V. DISCUSSION

In this paper we characterized a pneumatic sensing bulb designed as an add-on for myoelectric and body powered prosthetic hands. The results can be seen in Table III. This was done by measuring the pneumatic pressure from the sensing bulb when applying pressure with various sizes of indenters and with different incoming angles. Additionally, the sensing bulbs were evaluated in a use-like scenario, while integrated within glove and fitted on a commercial powered prosthetic hand.

A. Comparative Analysis

The evaluation of the sensing bulb not only showed low hysteresis and good repeatability, but also showed stable pneumatic pressure response when pressure was applied at different angles. This is an advantage in prostheses if incoming pressures are not perpendicular to the hinge joint, because parts of the force vectors will be absorbed by the joint. Therefore, when using commercial sensors, such as FSR and capacitive sensors, the response will be inconsistent. However, use of a sensing bulb, which is not dependent on the incoming angle, will give consistent readings. Moreover, to get correct readings from the FSR, the pressure has to be applied evenly over the surface area. During grasping, the objects can have different shapes or cover only some part of the FSR active area. This makes the sensing bulb a better candidate for measuring grasping force. It was mentioned that FSR, instead of being a sensor for absolute force measurements, would be a better fit for detecting motion and position [23]. Furthermore, during manipulation of objects, the maximum contact force is encountered at the distal phalanx [40]. This requires the sensors to be flexible so that they can be placed on the fingertips of the prosthetic hands and still provide stable gripping of objects. Applying FSR on prosthetic fingers, which are soft to resemble the human body, gives an inaccurate sensor response. As for the sensing bulb, it already has the shape of a fingertip and it gives stable values wherever the pressure is applied. Despite the well-known characteristics of FSRs (simple and cheap), this kind of sensing technology is rarely used in commercial devices because of its poor performance [41]. A more suitable option is the capacitive force sensor, SingleTact, which has better accuracy than do other thin-film sensors [42]. Comparing a load cell and FSR, a calibrated SingleTact was recommended for use in prosthetic hands because of its adaptability to the shape of an artificial fingertip and its higher accuracy [16]. On the contrary, it has been shown that the error increases with indenter curvature [16]. The sensing bulb, itself, is a sensor with the shape of a fingertip which eliminates error caused by integrating a sensor on a curved surface.

B. Hysteresis

The sensing bulb showed only small hysteresis during the experiment. Because the pressure was applied slowly, see section A, small hysteresis was measured (Fig. 7). In this study, the hysteresis may be explained by deformation of the silicone or by a very small air leakage, because the sensing bulb did not regain its shape during unloading compared to the corresponding position during loading. Another explanation for the hysteresis could be that the viscoelastic HTV silicone might have induced properties such as creep and stress relaxation. It is desirable to make sure that the connection between the sensing bulb and the plastic tube is sealed properly to make it airtight. Even though it is often required that the tactile sensors, used in prosthesis, should have low hysteresis [11], human skin itself has high hysteresis [4].

C. Linearity

There is a higher sensitivity when smaller indenters are applied on the sensing bulb compared with larger indenters, which indicates that the sensing bulb hand has greater sensitivity when grabbing smaller objects. This is related to the physical relationship between force, pressure, and contact area for the sensor. However, what is interesting is that the sensor bulb response, stronger for smaller object, correlates to the natural perceived sensation using your hand. A smaller object of equal weight to a bigger will give a stronger perceived sensory [43]. As a consequence, having the sensing bulb fitted on a MyoHand VariPlus Speed, which has a grasp force of 100 N, the sensing bulb will give a non-linear response when grabbing small objects, although, such high force might not be necessary when handling small objects. Furthermore, compared to FSR [44] with its nonlinearity, where it has a higher sensitivity at low forces and wider variations [38], the sensing bulb is an arguably better candidate.

A non-linear characteristic could also be seen when applying an indenter larger than the sensing bulb itself. A possible explanation could be same as for the smaller indenters where pressure saturation is obtained early. The MyoHand Variplus performs a tripod prehension. The grasp will not adapt to the object since it has one degree-of-freedom. For this kind of prosthesis, the average contact force is detected at the thumb and the distal phalanges of the index and middle finger. Compared to human hand and prostheses with adaptive grasp, the contact area is smaller resulting in higher force holding an object. However, considering the force being distributed on a smaller contact area the maximum force was shown to only reach 24.9 N at the fingertips [40]. For the sensing bulb the force prediction discrepancy of nearly 10 N would not be applicable for such low forces as the sensing bulb has a linear characteristic up to 30, 60 and 100 N for the indenters (ii), (iii) and (iv, v, vi), respectively. With an exception of indenter (i) for which the sensor is linear up to $\tilde{2}$0 N which is a little shy of the maximum force of 24.9 N.

The sensing bulb is linear within the range of 0-100 Hz, and has a resonant frequency of 161 Hz. The required frequency response for force and position sensors should be the basis for such requirements ($>$100 Hz) [38]. The rationale is that it can detect deep pressure and high frequency vibration (e.g., can detect smooth surface objects). Looking at the skin's time response, which is $\sim$15 ms [45], the sensing bulb is relatively faster at 6 ms.

D. Intensity Resolution

When applying pressure on the sensing bulb, before it became flat, the force gauge showed a minimum value of 20 N for the smallest indenter (i). No more than 20 N is necessary to apply to the sensing bulb because the required sensitivity range of a tactile sensor to mimic a hand is 0.01-10 N [4]. However, doing a characterization in the full range (0-100 N) showed us the limits of the sensing bulb.

E. Spatial Resolution

When making an artificial sensor that aims to mimic a human fingertip, the spatial resolution should be 1-2 mm (>2 mm interpoint distance, from which the pattern discrimination should be more sensitive) [4]. However, because the sensing bulb is one unitary sensor it will respond to wherever the pressure is applied, thus it will not sense the position of the pressure, which provides neither interpoint discrimination nor pattern recognition. Moreover, the sensation will be fed back to the residual limb and compared to the two-point discrimination in the fingertips, which at the level of forearm varies between 30-45 mm [46].

VI. Conclusion

This study evaluates a pneumatic sensing bulb that could be used in body-powered and myoelectric prostheses. We validated that the sensing bulb has an advantage of sensing incoming pressure from different directions better than with current commercial sensors, such as most of the resistive sensors and piezoelectric sensors. The results showed an angle dependency during high forces, however, during use it has been shown that the contact force on the fingertips is maximum 24.9 N [40]. Furthermore, the sensing bulb is stretchable, which is highly desired when designing a sensor glove to be used on prostheses. Due to the material and construction of the sensing bulb, it will not add any significant weight when being implemented in the prosthesis. Furthermore, since the sensing bulb does not contain any electrical elements it is not susceptible to electrical interference while interacting with the surroundings. The sensing bulb has the potential for further development, such as integrating other sensing elements, adapting the glove to be fitted on a prosthesis with a higher degree-of-freedom hand and technological improvements to reduce the hysteresis (air leakage) further and also, to eliminate the non-linearity when interacting with bigger objects. As mentioned, one drawback is that, if the sensing bulb is not sealed properly there could be leakage and if it breaks, the whole glove has to be replaced.

Future work would be to integrate the sensing bulb into an active sensory feedback system with actuators to provide the user with sensory feedback with different modalities. Compared to the previous study [31], where the system was passive, the active system opens up the possibility to adapt the feedback to the user. It would also be possible to filter sensor data and transform the sensor data before feeding back the information to the user, thus avoiding redundant information. This would especially be informative if the number of sensors increases. The sensing bulb provides with solely one type of sensory feedback, pressure. Inasmuch as the hand contains different kinds of tactile sensing modalities, the tactile system should also contain different sensors (hybrid tactile sensors). Regarding the angle dependency effect during large compressions, the effect could be minimized by having the sensor bulb at a higher internal pressure. This would make the sensing bulb deform less. In conclusion, a comparison with different design configuration could be taken into consideration for future development. Moreover, focusing on adding more features to the sensing bulb glove, such as texture sensing. Other avenues that will be pursued is to subdivide the sensor bulb into smaller parts. Potentially, by having a 2x2 sensing bulb in the prosthetic finger new features such as force direction will be extracted.

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Paper III

Electrotactile feedback for the discrimination of different surface textures using a microphone
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Article

Electrotactile Feedback for the Discrimination of Different Surface Textures Using a Microphone

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Abstract: Most commercial prosthetic hands lack closed-loop feedback, thus, a lot of research has been focusing on implementing sensory feedback systems to provide the user with sensory information during activities of daily living. This study evaluates the possibilities of using a microphone and electrotactile feedback to identify different textures. A condenser microphone was used as a sensor to detect the friction sound generated from the contact between different textures and the microphone. The generated signal was processed to provide a characteristic electrical stimulation presented to the participants. The main goal of the processing was to derive a continuous and intuitive transfer function between the microphone signal and stimulation frequency. Twelve able-bodied volunteers participated in the study, in which they were asked to identify the stroked texture (among four used in this study: Felt, sponge, silicone rubber, and string mesh) using only electrotactile feedback. The experiments were done in three phases: (1) Training, (2) with-feedback, (3) without-feedback. Each texture was stroked 20 times each during all three phases. The results show that the participants were able to differentiate between different textures, with a median accuracy of 85%, by using only electrotactile feedback with the stimulation frequency being the only variable parameter.

Keywords: electrotactile feedback; texture sensor; non-invasive stimulation; friction sound; feature extraction

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1. Introduction

Every day the human hand is used to explore and interact with the surroundings. This is made possible by the delicate interaction between the sensory and motor systems in the peripheral and central nervous system. The human hand consists of 17,000 mechanoreceptors such as Meissner's corpuscles, Merkel disks, Ruffini organs, and Pacinian corpuscles, located at different depths in the skin and they react to different stimuli [1]. They are categorized depending on the size of their receptive fields, adaptation rate, and location in the dermis. Exploring a texture with the fingers elicits texture-specific vibrations in the skin, activating both Pacinian and Meissner's corpuscles which respond to high-frequency respectively to low-frequency vibrations [2].

Both the spatial pattern of the object manipulated and the temporal pattern with which the object is being manipulated play a role in texture perception. The different patterns are conveyed in afferent responses. The spatial, such as gratings and Braille dots (on the order of millimeters), evoke a response of slowly adapting type I (SAI) afferents. However, in discriminating natural textures the temporal pattern is more dominant. The temporal pattern is encoded in the responses of rapidly adapting (RA) afferents and Pacinian corpuscles [3].

The loss of a hand, through amputation, disconnects the afferent and efferent pathways from reaching their targets. In order to achieve the motor control necessary for object manipulation and object identification, the afferent pathways provide crucial information to close the loop between the hand and the brain and provide sensory feedback [4]. In addition, sensory feedback is essential for information about an object's physical properties, such as texture, softness/hardness, etc.

Several different commercial prosthetic hands restore some motor functions to the amputee. However, these prostheses do not provide any sensory feedback and sensory feedback has been highlighted, by prosthesis users, as one of the desired functions in prosthetic hands [5]. It has been suggested that sensory feedback in a hand prosthesis should be modality-matched, meaning that pressure on a finger should be experienced as pressure by the amputee. Furthermore, the feedback should be somatotopically matched, meaning that pressure applied on, for example, the prosthetic index finger should be experienced as sensory stimulation on the index finger by the amputee. Somatotopically matched and modality matched sensory feedback mimics normal physiology and thus may reduce the cognitive burden that sensory substitution imposes on the prosthetic user [6]. Furthermore, it has been shown that adding sensory feedback facilitates the control of the prosthesis [7].

One way to provide sensory feedback is to use transcutaneous electrical nerve stimulation (TENS), a technique that is based on high-voltage electrical pulses sent through a pair or a plurality of electrodes placed on the skin, to stimulate nerve fibers. It is commonly used to relieve pain [8] or provide electrotactile feedback [9]. In addition, TENS can play an important role in the control of manipulation tasks for prosthesis users [10–14] and assisting in the interpretation of objects [15]. TENS applied to the skin over the median or ulnar nerve in the amputation stump result in sensations experienced as originating from the median or ulnar nerve innervated fingers in the lost hand (somatotopic sensation) [16]. Using TENS could aid prostheses users to discriminate a surface's texture in a more intuitive manner and without sensory substitution, which is dominant in the case of other types of sensory feedback. Additionally, electrotactile feedback could potentially reduce phantom limb pain and stump pain in transtibial amputees [17] and also enhance the feeling of embodiment [18].

Technical solutions to provide sensory feedback of the force generated during a grasp are well explored, while feedback for texture perception for use in prosthetic hands is not. Interestingly, sensors used to detect texture-information are more common in self-organizing robots or robotics in applications such as health, eldercare, and manufacturing [19–23]. To provide an amputee with natural sensory feedback, implants that directly stimulate a peripheral nerve have been proposed. By using an artificial fingertip with a Micro Electro Mechanical System (MEMS) sensors using four transducing piezoresistors, the user could discriminate different textures based on the produced patterns of electrical pulses, which in turn, stimulate the nerves in the arm [24]. The small size and low power consumption in MEMS sensors are advantageous if used in a prosthesis. A proposed system for sensory substitution, to be used in prosthetic hands, used a force-torque sensor to obtain texture data from three different types of textures. By using a convolution neural network (CNN) algorithm, the different textures were classified and converted to vibrational stimulations [25]. Sensors based on Polyvinylidene Fluoride (PVDFs) films have been used for texture detection [26]. PVDFs, when stimulated by vibrations, display similar characteristics to fast adapting mechanoreceptors [22]. Yi et al. [27] developed a bioinspired tactile sensor based on piezoelectric materials, which was shown to mimic Meissner's corpuscles. In addition, multi-modal sensors have been used to identify different materials, by implementing multiple gauge sensors, to capture resistance changes, together with PVDFs to capture electric potential changes. As mentioned, PVDFs are equivalent to fast adapting mechanoreceptors while the gauge sensors represent the slow adapting mechanoreceptors that detect lateral stretch, hence, it detects the static properties of a stimulus [22,23].

It has been shown that participants who have lost sensibility in a hand can substitute it, to some extent, with auditory information [28]. The participants could differentiate between different textures by listening to the friction sound picked up by small microphones. Another study used a microphone attached to the forearm to show that vibrations occurring, when the fingertip was sliding over a rough surface, can propagate from the fingers to the forearm [29]. This suggests that a microphone is a good candidate to pick up a texture's acoustic characteristics. A classification analysis showed that the frequency composition in the texture-elicited vibrations consists of enough information to allow for the identification of different textures with high accuracy [30]. An early approach to texture recognition, a sensing pen was developed containing an electric microphone to classify different textures using neural networks [31]. A study showed promising results to use a capacitor microphone with an attached metal edge for texture sensing. When exploring different textures, the metal edge vibrates in different frequencies depending on the textural properties of the stroked material. The different textures could then be identified by using signal processing with the fast Fourier transformation (FFT), coupled with a supervised Learning Vector Quantization (LVQ) [32]. Another study implemented a node network of 10 microphones in robotic skin and classified different textures with a logistic regression model [33].

The current paper contributes a simple electrotactile feedback system with a computational method to convert recorded friction sounds, arising from stroking different textures, into somatotopic electrical stimulation in real-time. Electrical stimulation was chosen because of its easy application and the control of the stimulation parameters, such as the amplitude and frequency. The friction sound of a texture was recorded with a condenser microphone and median frequency was calculated of the audio spectrum. By analyzing the accuracy for discriminating different textures with the proposed system, conclusions can be made if the system is fit to be used as a texture sensing substitute. With the proposed system the participants had an overall median accuracy of 85% in discriminating different textures.

2. Materials and Methods

2.1. Participants and Ethics Approval

Twelve able-bodied participants, 10 males and 2 females (median age, 31 years; range, 24–44), with no known neurological disorders participated in the study. Two participants had previous experience of electrical stimulation, however, they were not familiar with the current study protocol. The rest of the participants had llittle knowledge of electrical stimulation in general. All participants provided informed consent, and the study was approved by the Swedish Ethical Review Authority (DNR 2020-03937).

2.2. Equipment

Four different textures [34] (see Figure 1) were used to evaluate the ability to discriminate between different surfaces based on electrotactile feedback.

Figure 1. (**a**) Close-up image of the omnidirectional electret condenser microphone with an amplifier module. An isolation cable was put on the circuit board for an easier grip of the microphone during the stroking and to reduce interference with the components on the printed circuit board (PCB). (**b**) The experimenter was stroking the different textures with a microphone in a proximal to distal direction. (**c**) Placement of the electrodes on the participant's forearm. The cathode was placed over the median nerve while the anode was applied on the upper part of the forearm.

An omnidirectional electret condenser microphone with an amplifier module (Adafruit MAX9814, Adafruit Industries, New York, NY, USA) (Figure 1a) was used as a sensor for the exploration of different textures, by picking up the friction sounds during stroking. The operating frequency range of the microphone was 20–20,000 Hz and the gain was set to 60 dBA. During the experiments, the experimenter was holding the microphone by hand while stroking the textures so that the enclosure was in direct contact with the surfaces (Figure 1b), thus, it was able to pick up the friction sound. The digitalization of the audio signal was done by a PJRC Teensy 4.0 microcontroller (32 bit 600 MHz ARM Cortex-M7 processor, using an NXP iMXRT1062 chip, PJRC.com, LLC, Sherwood, OR, USA). As the initial tests of the microphone-textures interaction showed that the friction-originated audio signal for the different textures was below 3 kHz, the microphone signal was sampled at 6 kHz with 16-bit resolution. The processing of the microphone signal and extraction of the signal features that depicted characteristic vibrational/audio responses during the tactile exploration was done in real-time by Teensy.

The extracted signal features were sent to a custom-made electrical stimulator capable of producing biphasic charge-balanced cathodic-first current-controlled pulses of amplitudes in the range from 0.1 mA to 10 mA (steps of 0.1 mA), and frequencies of 1 to 100 Hz. The DC/DC boost switching regulator was used to generate stimulation voltage which was maximally 38 V (depending on the skin impedance and the stimulation current). The stimulation control was done by an onboard PIC18F25K22 microcontroller (Microchip Technology, Chandler, AZ, USA). The microcontroller managed generation of the stimulation patterns and communicated with both, the Teensy microcontroller and PC, enabling alteration of stimulation parameters in real-time. The electrical stimulation was delivered to the participants through self-adhesive Pals electrodes (Axelgaard Manufacturing Co., Lystrup, Denmark), placed on the skin over the median nerve so the sensations following stimulation, were associated with the median nerve innervated fingers (thumb, index, and the middle finger) and palm area. An overview of the system can be seen in Figure 2. The PC included in the setup had a non-essential role as it was used just to initiate the protocol and visualize stimulation frequency in real-time during the experiments.

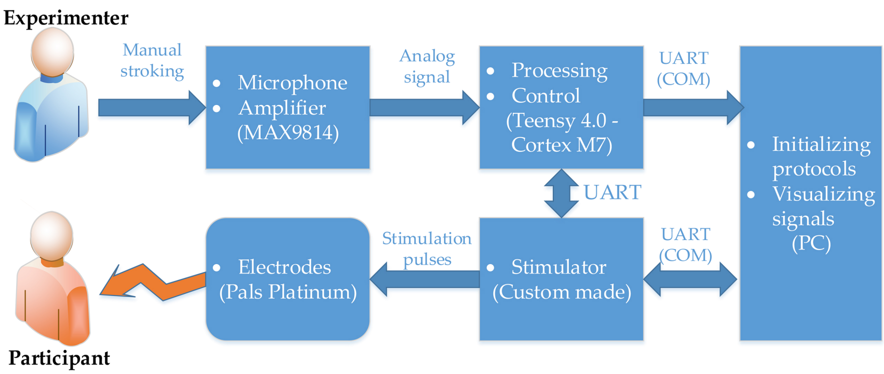

Figure 2. System overview. The manual stroking was performed by the experimenter with an omnidirectional electret microphone with the integrated amplifier. The audio signal was digitized by a Teensy 4.0 microcontroller for further signal processing and signal features extraction. The calculated median frequency was then sent to the custom-made electrical stimulator through a Universal Asynchronous Receiver/Transmitter (UART) connection. The electrical stimulation was then delivered through self-adhesive Pals electrodes attached to the participant's forearm skin over the median nerve. Both, Teensy and the stimulator were communicating with a PC using their UART connections.

2.3. Algorithm

The formulation of the algorithm presented in this paper was based on several empirical pre-tests (the algorithm evaluation stage) that were used to characterize friction-based interaction between the microphone and selected textures. These pre-tests identified the audio frequency range resulting from friction with selected textures and the behavior of several signal features, such as total signal power, peak frequency, mean frequency, and median frequency, during manual stroking.

The first step of the algorithm that was designed and implemented, as a part of this study, was the calculation of the frequency content of the microphone signal. The FFT calculation was performed on 2048 samples of the digitalized audio signal using a modified version of the Arduino library (http://github.com/kosme/arduinoFFT, accessed on 11 November 2020). The FFT was updated after every 128 samples, corresponding to a $\sim$50 Hz update rate with the sampling rate of 6 kHz. In the next step, the spectral components at 50 Hz and below 20 Hz were removed from the FFT spectrum.

The feature of the microphone signal that was heuristically chosen, was the median frequency of the audio spectrum. Besides this feature, several other well-known features, such as the mean frequency and the audio signal envelope, were tested in a small sample trial, but the median frequency showed the best results in discriminating different textures. In the next step, the median frequency was linearly translated into stimulation frequency. The rationale for devising the transfer function was to shift the median frequencies of the audio signal to the range of stimulation frequencies that are commonly used for sensory/neural stimulation [35,36]. The devised transfer function equation was:

$$stim = (medianf - lowerB)/scaling + 5, \tag{1}$$

where *stim* denotes the stimulation frequency (in Hz) that was sent to the stimulator, *medianf* denotes the median frequency signal feature, *lowerB* denotes the lowest median frequency that was empirically chosen as relevant, and *scaling* denotes the linear scaling factor used to constrain the possible median frequencies of the microphone signal into the range of stimulation frequencies produced by the stimulator. In the current study, the *lowerB* and *scaling* constants were empirically set to 50 and 10, respectively. The addition of 5 Hz was done to constrain the lower dynamics of the stimulation as extremely low frequencies would significantly reduce the information bandwidth delivered to the participant. In other words, sending a low stimulus frequency (e.g., below 1 Hz) would mean that the next stimulation pulse, and also the change of stimulation frequency, would have to wait for a long time (more than 1 s in the case of <1 Hz stimulation). This is the result of the electronic stimulator protocol which accepts updates only after producing a stimulation pulse specified by the last command. In addition, the stimulation frequency resulting from Equation (1) was constrained to 80 Hz, thus frequencies calculated as higher than 80 Hz were set to 80 Hz. The total processing time of the system is calculated to be 220 ms, which includes the communication delay (UART) of 20.6 ms, and due to the waiting until the last desired pulse is generated (max 200 ms in the case of 5 Hz stimulation). The processing time is considered to be fast enough to be considered "real-time" and should not affect the results as the reaction time to sensory stimuli is 50–300 ms [37].

Apart from the calculation of the stimulation frequency, the algorithm extracted the total signal energy from the microphone signal over the last 2048 samples, by summing the spectral components from 1 to sampling_frequency/2, which was used to enable/disable the stimulation. The stimulation was activated in the case of the magnitude exceeding the empirically chosen threshold, in this case, 300 mV. The threshold was set so that the stimulation was active only while the microphone was in contact with a surface.

The medians of the audio signal can be seen in Figure 3, with an error which is set to lower and upper quartiles of 25% and 75% respectively. As seen in the figure, there is some overlap between median frequency patterns resulting from stroking different textures. The silicone has a median frequency which gradually ramps up in frequency during the stroke, sponge and felt had a median frequency of approximately 200 Hz and 50 Hz, respectively while the mesh has a varying median frequency. The frequency responses presented in Figure 3 were obtained during the evaluation stages of the algorithm when the experimenter stroked the textures in an unobstructed (without any disturbances between stroking) and paced manner (the visual cue for executing strokes was given every 10 s). In the experiments with participants, the experimenter had to wait between consecutive strokes until the participant responded, which reduced repeatability leading to inconsistent stroke duration and dynamics. Another important difference between the algorithm evaluation stage and the study was that the median frequency update interval in the case of the evaluation stage was constant (∼50 Hz) while during the experiment the update was constrained by the latest stimulation frequency, thus ranging from 50 Hz to 5 Hz. This limitation was particularly noticeable when the median frequency of the audio signal was low, e.g., while the sponge or mesh was being stroked, and during the initial part of the stroke in the case of silicone rubber.

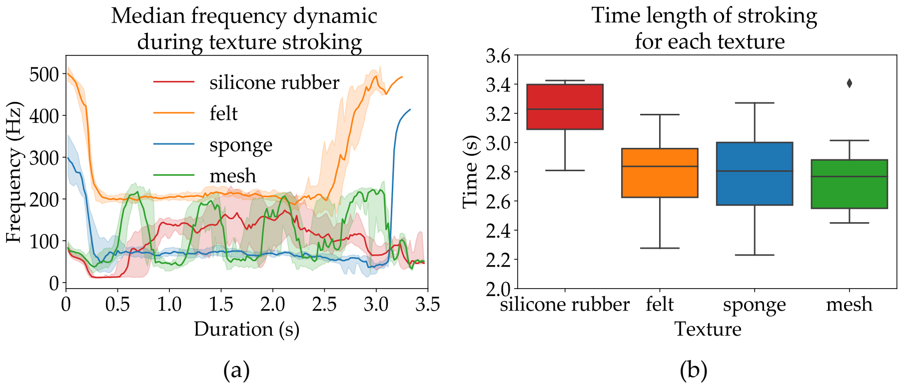

Figure 3. Consistency-test for the four textures. A total of 20 strokes were applied on each texture. (**a**) Median frequency error, set with lower and upper quartile (25% and 75%). There was a better consistency for felt and sponge, but the error is bigger for the silicone rubber and mesh. (**b**) The time consistency during the strokes.

2.4. Protocol

The experiment consisted of three phases: (1) Training phase, (2) feedback phase, and (3) without feedback phase. All phases were performed with participants sitting down in a quiet room. The initial step of the protocol was the placement of the stimulation electrodes. The anode (rectangular 7 × 10 cm Pals electrode) was placed on the ventral side of the forearm, while the cathode (round 2.5 cm in diameter Pals electrode) was placed over the median nerve, just proximal to the wrist as shown in Figure 1c. In this area, the median nerve is located between the tendon of the flexor carpi radialis longus muscle and the tendon of the palmaris longus muscle. Next, a stimulation at 50 Hz, with a duration of 2 s was applied to identify a sensory threshold, pain threshold, and the presence of the somatotopic sensation due to the nerve stimulation. After finding the sensory threshold with the resolution of 0.1 mA, the current intensity was increased in steps of 1 mA until reaching a painful level of stimulation. As the current output of the electronic stimulator was limited to 10 mA, the pain threshold was also capped at 10 mA. The amplitude of the current pulses that were used in the experiment was set to level = 3 (on a scale between 0 = no-sensation and 10 = maximum pain level), where the sensory threshold is level = 1 and the pain threshold level = 8. Level = 3 is considered to provide a distinct stimulation without evoking any pain. The stimulation amplitude for each participant can be seen in Table 1. At the current amplitude selected for the experiment, each participant was asked to inform the examiner about where he/she perceived the stimulation. If the stimulation was perceived in median nerve innervated skin areas, such as the thumb, index finger, middle finger, and part of the ring finger, but not at the electrode location, the stimulation was considered somatotopic. In the case of non-somatotopic sensation, the cathode was slightly relocated, and the previous steps were repeated until an electrode location leading to somatotopic sensations in median nerve innervated fingers were found.

Table 1. Participants in the study and the individually-set levels of stimulation amplitude (mA) where the stimulation frequency was 50 Hz and with a pulse duration on 250 µs. The perception threshold (Level 1) is the just-noticeable stimulation amplitude. The pain threshold (Level 8) is when the amplitude is too high for the participant to endure. The stimulation amplitude chosen for this study (Level 3) is the amplitude that was considered to provide the participant with a distinct stimulation that was also comfortable for prolonged exposure.

Participants	Perception Threshold (Level 1)	Stimulation (Level 3)	Pain Threshold (Level 8)
M1	3.2	5.1	10.0
M2	2.2	3.9	8.0
M3	2.8	3.4	5.0
F1	3.0	5.0	10.0
M4	3.1	5.1	10.0
F2	2.1	3.8	8.0
M5	1.3	2.1	4.0
M6	2.3	4.2	9.0
M7	2.3	3.4	6.0
M8	1.7	3.8	9.0
M9	2.0	4.0	9.0
M10	3.5	4.8	8.0

Upon establishing the stimulation amplitude, the training phase of the experiment was initiated. In this phase, the experimenter sequentially stroked each texture for 20 cycles (in total 80 strokes) with an estimated speed of 14–25 mm/s. During this phase, the participant got to watch the strokes and at the same time receive the electrotactile feedback.

After finishing 20 cycles, the participant was blindfolded and acoustically insulated using headphones, and the stimulation with the feedback phase was initiated. During this phase, the experimenter stroked the different textures in a proximal to the distal direction in a randomly predefined sequence, while the participant was instructed to verbally identify the texture. Up to two additional repetitions of the same texture were allowed, if requested by the participant. Upon receiving a response from the participant, the experimenter provided verbal feedback consisting of true/false statements and the information regarding the stroked texture (in the case of false response by the participant). This phase consisted of 20 repetitions of each texture (80 in total). Upon completing the phase, a short break was taken (approximately 5 min).

The final phase also comprised 20 repetitions of each texture in a new randomized order, but without feedback from the experimenter. As in the previous phase, two additional repetitions of the same texture were allowed for the participant.

It should be noted that all of the strokes were subjected to variability due to the manual execution by the experimenter. Specifically, with sticky or rough surfaces, such as silicone rubber, the vibrations resulting from the friction were unpredictable. In the case of smoother surfaces, the speed and consistency of the strokes were also directly mirrored in the frequency of the stimulation.

2.5. Data and Analysis

Two separation analysis was done in this study, one to analyze the consistency data and one on the following experimental data.

A custom-made LabVIEW program (Labview 2018, National Instruments, Austin, TX, USA) was used to record data during the evaluation stages of the algorithm development. These tests comprised stroking of each texture pseudorandomly by the experimenter and were used for further investigation of the consistency of the stroking. Two factors were explored; the consistency of the time of each stroke and the consistency of the frequency. The data analysis was performed in Python with several libraries, such as Scikit-learn (https://scikit-learn.org/, accessed on 11 November 2020) and SciPy (https://www.scipy.org/, accessed on 11 November 2020).

A Generalized Linear Mixed Effects model was fitted to the the data using jamovi [38–40]. The dependent variable was the accuracy of the responses of the subjects per phase and rod. A Poisson distribution using a log link function was used as this fits the type of data. Phase (with feedback and without feedback) and stimulus type (mesh, felt, sponge, and silicon rubber) were the factors and levels of the experiment. The participant was considered a random effect. Subsequent post-hoc comparisons for the different textures within each phase using Z-tests were corrected using Holm's sequential Bonferroni procedure.

3. Results

The accuracy of all 12 participants in identifying four different textures are shown in Figure 4a. It should be noted from this figure that the variance of accuracy for different participants is relatively large. Three of the participants had an accuracy higher than 90% in the experimental phase with verbal feedback, while this number increased to six participants during the last phase when no feedback was provided to participants. As the accuracy is not normally distributed, the total accuracy was calculated as the median of individual accuracy. The total median accuracy for all participants and textures was 85% (IQR 70–95%). The participants needed an average of 1.54, 1.24, 1.44, and 1.40 repetitions for each texture (silicone rubber, felt, sponge, and mesh) before responding to the stimulus generated by the different textures. The overall performance for each texture can be seen in Figure 4b. There was a statistically significant difference in terms of phase (with feedback, without feedback, $p = 0.034$) and texture ($p < 0.001$). The multiple comparisons test performed on textures per phase revealed there was a statistically significant difference between some of the textures, namely felt vs. silicone rubber in the with feedback condition ($p < 0.001$), mesh vs. silicone rubber in both feedback conditions ($p = 0.016$ (w/ FB), $p = 0.0086$ (w/o FB)), and sponge vs. silicone rubber in the no feedback condition ($p = 0.043$). For the other combinations there were no statistically significant differences.

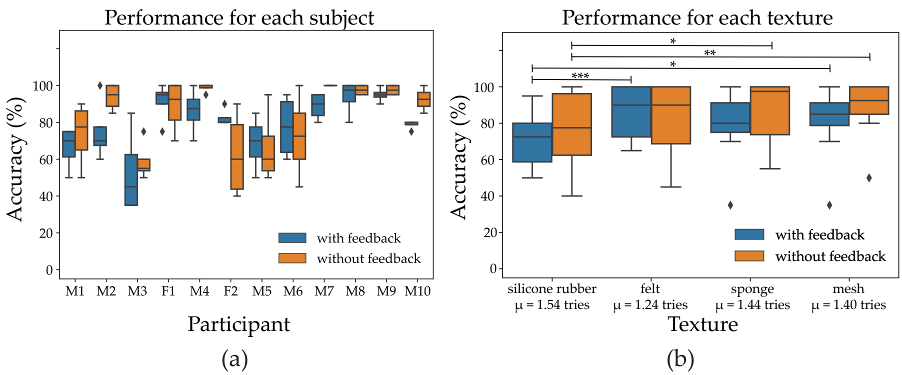

Figure 4. (**a**) The median accuracy for each participant where three participants were over 90% during the first phase (with feedback) and six participants during the second phase (without feedback). (**b**) The box plot shows the performance of the 12 participants on each texture. The x-axis also shows the average attempt the participants had for each texture.

The experimental results can be seen in the stimuli-response confusion matrix in Figure 5. During the final phase (without feedback) the median performance for each texture: Silicone rubber, felt, sponge, and the mesh was 77.5% (IQR 62.50–96.25), 90% (IQR 68.75–100.00), 97.5% (IQR 73.75–100.00), and 92.5% (IQR 85.00–100.00), respectively.

During the second phase (stimulation with feedback), felt was often mentioned, by the participants, as being the easiest to distinguish (85.4%) because of its higher median frequency. However, it was sometimes confused with silicone rubber (12.9%), since silicone rubber, on some occasions, gave inconsistent stimulation because of its sticky characteristics in the texture. This made the participants, on some occasions, hesitate if the stimulation stemmed from silicone rubber. Vice versa, silicone rubber was in 15% of strokes misidentified as felt. As mentioned, silicone rubber gave a low frequency at the beginning, which increases during the stroke, however, during manual stroking on few occasions the audio signal did not have this characteristic and instead was confused with the felt 15% of the time and 12.5% of the time it was misinterpreted as the sponge. In the final phase (stimulation without feedback), the performance improved for mesh, sponge, and silicone (+7.1%, +6.7%, and +5.9% respectively) while the ability to detect felt decreased slightly (−4.2%).

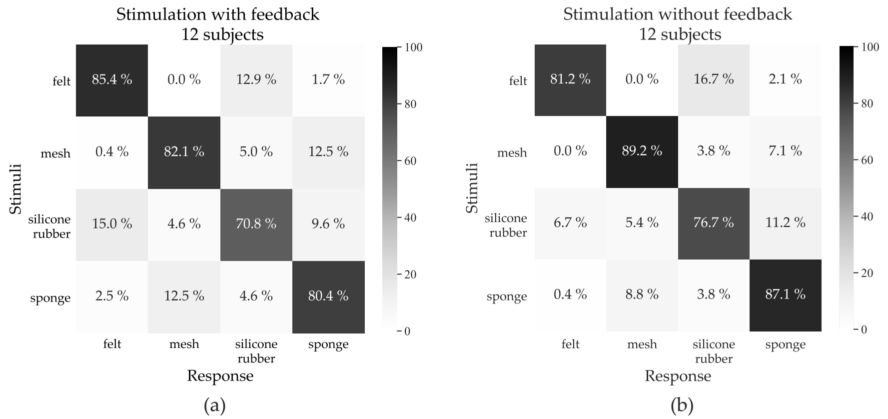

Figure 5. Confusion matrix for the identification of the different textures during the two phases. (**a**) Phase with feedback, where felt had the highest accuracy (85.4%). The lowest accuracy was when discriminating silicone rubber (70.8%), which was occasionally misinterpreted as felt (15.0%) and as sponge (9.6%). (**b**) Phase without feedback, where mesh had the highest accuracy (89.2%). In general the performance increased for all textures except felt in the phase without feedback. Mesh is easiest to discriminate (89.2%), sponge has also a high accuracy (87.1%). There was also a small improvement for the silicone texture (76.7% vs. 70.8%).

4. Discussion

The presented study was designed to assess the feasibility of developing a computational method for the direct conversion of the sound detected by a microphone, when stroking a texture, into an electrotactile stimulation pattern that could be used to distinguish between different textures. However, the microphone and the audio amplifier used in this study were regular, off-the-shelf components, not chosen for the specific purpose of measuring contact vibrations. It should also be noted that the experimental conditions in this study were designed to resemble conditions that would be expected in a real-world application of a feedback system in a hand prosthesis. Mainly, the sound was translated into the frequency of the electrical stimulation continuously, permitting variability of the feedback, in accordance with the natural variability in stroking velocity and pressure seen in a hand exploring an object.

The results of the study are encouraging since, after just a brief familiarization period comprising 20 strokes of each texture which lasted approximately 5 min in total, participants were able to achieve a relatively high overall median accuracy (85%). It should be pointed out that only two participants had any previous experience with electrical nerve stimulation which makes the familiarization of only 5 min even more rigorous in the case of naive participants. These participants had to get accustomed to both, the new sensation in general and the variation of stimulation frequency due to the interaction with different textures. It should be emphasized that the experiment was designed such that the participants had their vision and hearing occluded while in a real-life exploration of textures, the vision and tactile perception plays an equal role in identifying textures [41]. The same would apply to the auditory cues. When exploring a texture the sound of the stroking of textures can also help to perceive a texture's roughness [42]. It has also been shown that auditory cues could be more beneficial than visual and tactile cues to detect a material's stickiness [43].

Jamali and Sammut [22] used PVDFs to detect vibrations from materials for the classification of seven different surface textures based on three and five Fourier coefficients and with 50 learning samples per each. The stroking was done by a robot ensuring high repeatability of strokes. Furthermore, the results of Jamali et al. are intended for machine-based classification only, without involving human participants as recipients of the feedback information. The prediction accuracy for their algorithm, using a naive Bayes learner, was 78% when three Fourier coefficients were used and 83.5% when five coefficients were used. Compared to our study, the median accuracy of a human of 85% is a promising result for providing continuous feedback to participants. The same group [23] presented another method based on the learned classifier, resulting in a higher accuracy of 95% $\pm$ 4% on the unseen data. The setup consisted of an artificial robotic finger with implemented sensors that respond to stretch (strain gauges) and vibration (piezoelectric sensors). However, having a robotic finger with set pressure and velocity, the stroking of the material will be highly consistent, thus it could be debated if a learned classifier used in a controlled laboratory environment would perform as well in real-life manual stroking. Hughes and Corell [33] did consider the inconsistency in a human operator in their study, by stroking the textures by hand to include the variability in speed and pressure of the stroking. They implemented a network of sensor nodes, using omnidirectional microphones, embedded in silicone rubber for texture recognition showing that the skin prototype was able to identify 15 different textures with an accuracy of 71.7%. It should be noted that all of the aforementioned studies referenced within this paragraph present results on the ability of machine learning algorithms to discriminate between textures. It has not been shown if a human participant could match such performance in a real-time feedback setup (as presented in our study).

The present study could also be considered as the worst-case scenario as the feedback is directly proportional to the texture stroking dynamics which was completely governed by the experimenter, while the participant did not have any complementary information. Thus, we hypothesize that implementation of the concept of the proposed system, using a microphone as a sensor and electrical stimulation as a feedback mechanism, to provide information about a texture would significantly improve the accuracy of identifying a texture. The results support this hypothesis, where the accuracy is significantly improved between the second phase (with feedback) and final phase (without feedback) which was done in a short-term controlled experiment. Considering that the system will be used long-term and during activities of daily living, the user then would also use their natural feedback modalities, such as audio, visual, proprioceptive, or force, at his/her disposal. The user would then be able to incorporate his/her natural feedback modalities to further strengthen the internal models of interaction with different textures. Furthermore, with the continuous use of the proposed electrical feedback system, texture recognition would likely gradually improve because in our data there is a significant improvement in performance even between two consecutive tests. Hence, it could be argued that during long-term passive learning, the overall accuracy could be improved for the proposed system [44]. In addition, prolonged use accompanied with the spacing effect, where learning is spread over time [45], could also enhance overall accuracy.

One of the major sources of errors identified by the experimenters and participants was the inconsistency of the stimulation pattern dynamics for some of the textures. Manual stroking is prone to variations in applied force, stroking velocity, and velocity profile, all of which affect friction sound, and consequently electrical stimulation. However, all of these parameters are also subject to variation in the normal use of hands. The accuracy during the tests (which were between 70% and 89%) are the consequence of the high variability in median frequency of the audio signal resulting in the partial overlap between friction responses of other textures (see Figure 3a). In addition, the acoustic signals for the different materials might also deviate for each participant depending on where on the fabric/textures the strokes were applied. These deviations in signal were mainly noticeable for the non-smooth textures, such as mesh and silicone rubber. This can be seen in the results, showing a statistical significance in accuracy between silicone rubber and the other smoother textures (felt (w/ FB) and sponge (w/o FB)) supporting the discussion that a possible explanation for lower accuracy in discriminating silicone rubber due to its stroking inconsistency. There was also a significant difference in accuracy between silicone rubber and mesh indicating that despite the non-smooth texture mesh was easier discriminating compared to silicone rubber, whilst there were no significant difference between mesh and the other textures (felt and sponge), hence it can not be concluded that smoothness is crucial in being able to discriminate the textures. In the case of mesh texture, the distance between knots of the fabric was not constant, but upon rhythmic, consistent stroking this texture could resemble the sponge which was characterized by a low median frequency. In the case of silicone rubber, the sticky texture often resulted in the median frequency ramping-up (as shown in Figure 3a), but due to inconsistencies in applied force, this characteristic signal feature was sometimes missing, making it difficult to distinguish silicone rubber from felt or sponge. Both of these issues are the consequence of separation/partitioning between movement and sensation (the movement was performed by one person and the sensation is experienced by another person). This would be minimized in a real-world scenario where the same person does the movement and receives the sensory feedback, thus employing an internal forward model of hand movement and experiencing movement dynamics with complementary senses, such as proprioception, force, and vibration feedback (natural or externally generated). It can also be noted that the stimulation frequency increases with the stroking speed, and this applies also to the human hand where Manfredi et al. [30] recorded the skin vibrations during exploration of textures. The recordings showed an increase in frequency with increasing speed, and this applies both to non-periodic and periodic textures. Thus, having the same person performing texture exploring and perceiving the sensation, proprioceptive information will aid the central nervous system (CSN) to determine the velocity of the moving limb [46], in this case, the information of the stroking speed. Hence, the felt stimulation frequency could be associated with the stroking speed.

Due to the omnidirectional feature of the microphone, it picks up sound with equal gain in all directions, making it susceptible to background noises which is a limitation associated with the study. However, the experiment was performed in a quiet room which eliminated most of the background noises. If the currently described system was to be used for a prosthetic hand, background noise could be removed by using a noise-canceling sub-system where an additional microphone could detect only background noise.

Having in mind that other feedback modalities, such as force or hand aperture, would be prioritized in a prosthetic hand as they are coupled with the basic prosthetic hand functionality (grasping), the goal of this study was to evaluate the feasibility of using appropriate electrical stimulation for texture discrimination feedback using a microphone to pick up friction sounds of textures. As the feedback related to different sensors, such as force sensors, encoders, and microphones, would be combined within the same feedback interface (same electrical stimulator and electrodes), it was decided to dedicate only a single controllable stimulation parameter (frequency) out of many, such as, stimulation amplitude, frequency, pattern, and location, to texture exploration. Thus, frequency modulation of the electrical stimulation delivered to one cathode was chosen as a minimalistic setup, leaving other parameters available for other potential feedback modalities.

The presented system is designed to be portable. The hardware components comprising sensor and actuator sub-systems are based on two microcontrollers, ARM Cortex M7 and PIC18F25K22. The former processor handles audio signal sampling and execution of the algorithms presented in the paper, while the later processor is responsible only for executing electrical stimulation in a time-crucial manner. Additional circuitry related to the stimulator analog output stage, consisting of the step-up converter and discrete components, has a relatively small footprint (less than 4 cm^2 in the current version) and power consumption. Therefore, this hardware setup could be implemented within common hand prostheses by (1) integrating one or several miniature microphones on the prosthesis fingertips within a silicone glove, (2) placing at least one pair of electrodes over one of the major hand nerves, and (3) embedding all necessary electronics (including the prosthesis control part) on a single printed circuit board. As the presented system is self-contained, it could be integrated with existing and future powered hand prostheses or even in cosmetic prostheses.

5. Conclusions

This study presented an electrotactile feedback system with a microphone as a sensor, making it possible to pick up friction sounds from textures. In addition, a simple computational method to convert the signal transduced by the microphone into electrical stimulation was developed. The median frequency was calculated on the transmitted signal, since this feature had the best results in discriminating textures. The system provided the participant with somatotopic electrical stimulation from the processed microphone signal, which resulted in the participants being able to identify differences in textures. To the best of our knowledge, this concept is novel and there are no similar studies with the proposed system reported in the literature. The goal of this research was to devise an algorithm and self-contained hardware capable of supplying continuous feedback during texture exploration, and with future improvements, it would be interesting to investigate the performance during long-term use by a prosthesis user. The presented paradigm offers a unique feedback modality as there are no constraints regarding the number of detectable textures or their properties while the particular stimulation patterns resulting from stroking different textures could be learned by a user over time. In addition, the learning curve was steep, illustrated by the accuracy of 85% in participants (who had no prior knowledge of electrical stimulation) identifying different textures already after 20 repetitions.

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Institutional Review Board Statement: The study was approved by the Swedish Ethical Review Authority (DNR 2020-03937).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author. The data are not publicly available due to ethical restrictions.

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Conflicts of Interest: The authors declare no conflict of interest.

Abbreviations

The following abbreviations are used in this manuscript:

FFT	fast Fourier transformation
TENS	transcutaneous electrical nerve stimulation

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Paper IV

The Rubber Hand Illusion evaluated using different modalities

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Manuscript

The Rubber Hand Illusion evaluated using different modalities

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Abstract

Tactile feedback plays a vital role in inducing ownership and improving motor control of prosthetic hands. The prosthetic user needs to rely on visual input to adjust the grip without tactile feedback when grasping objects. However, commercially available prosthetic hands do not provide tactile feedback. The classical rubber hand illusion (RHI) has been used to induce ownership of a rubber hand, where a brush is stroking the rubber hand, and the user's hidden hand synchronously. This classical RHI provides modality-matched feedback, meaning that the stimulus on the real hand matches the stimulus on the rubber hand. The same RHI method has been used in previous studies with a prosthetic hand as the "rubber hand", suggesting that a prosthesis can be incorporated within the amputee's body scheme. Interestingly, applying stimulation with a mismatched modality where the rubber hand was brushed and vibration was felt on the hidden hand induced RHI. This study compares mechanotactile, vibrotactile, and electrotactile feedback to provide the user with tactile sensory feedback. Participants with intact hands took part in a modified RHI experiment. The rubber hand was stroked with a brush, and the participant's hidden hand received stimulation with either brush stroking, electricity, pressure, or vibration. The three latter stimulations are modality mismatched with regard to the brush stroke. Participants were tested for ten different combinations (stimulation blocks) where the stimulations were applied on the volar (glabrous skin), and dorsal (hairy skin) sides of the hand were performed. Using two standard tests (questionnaire and proprioceptive drift) showed that all types of stimulation induced RHI. Electrical and vibration stimulation induced a stronger RHI than pressure among the mismatched modalities. After completing more stimulation blocks, the proprioceptive drift test showed that the difference between pre- and post-test was reduced. This indicates that the illusion was drifting toward the rubber hand further into the session.

1 Introduction

After a hand amputation, both motor and sensory functions are lost. The functionality and appearance of the lost hand can be restored to some extent by prosthetic hands. However, some amputees choose not to wear prosthetics due to discomfort, dissatisfaction with the functionality, and lack of sensory feedback [1]. Sensory feedback plays a crucial role in inducing ownership, and several studies have shown that body ownership is associated with cutaneous touch [33, 28, 6]. Ownership of a prosthetic hand elicits a feeling that the prosthetic hand is a part of the body and could also contribute to improved control of the prosthesis [2]. The sense of ownership is a key element, together with the sense of agency, to create embodiment. With regards to prosthetic embodiment, the term is conceptualized in phenomenology where "the prosthesis is perceived as part of the body" [3], and has been demonstrated to be critical for prosthetics acceptance [4]. In contrast, the sense of agency is defined as a feeling that one controls the movement. Furthermore, high levels of prosthesis ownership have shown to be significantly related to lower levels of phantom limb pain (PLP) and residual limb pain (RLP) [5].

The Rubber Hand Illusion (RHI) is a well-established model for studying ownership of a rubber hand, where the illusion will induce a sensation that the rubber hand belongs to one's own body. The classical RHI was described by Botvinick and Cohen [6], where brushing both a rubber hand and the participant's own hand synchronously induced a sense of ownership of the rubber hand. Brushing both the rubber hand and the participant's hidden hand provides a visuotactile, modality-matched stimulation. The rubber hand is placed in front of the subject in an anatomically plausible position, and the participant's own hand (matching the side of the rubber hand) is hidden from view. Participants have reported to experience the RHI after brushing the hands synchronously for 10–110 seconds [7, 8]. On the other hand, brushing the participant's own hand asynchronously, where the delay is longer than 300 ms [9], related to the brushing on the rubber hand, will not induce the RHI [6]. Generally, the RHI is performed in two parts, the induction phase and the response phase, where the former exposes the participant to the stimulation and the latter involves tests to measure the strength of the illusion, such as objective measures (proprioceptive drifts) and subjective reports of ownership[10].

Ehrsson et al. [33] used functional magnetic resonance imaging (fMRI) to look at brain activity in healthy subjects, who were susceptible to RHI, while performing the classical RHI. They found that activity in the premotor and intraparietal cortex was correlated to the subjective measure of vividness, which defined how realistic and life-like the participants perceived the RHI.

Ehrsson et al. [11] performed the RHI experiment on eighteen unilateral transradial amputees. They showed that RHI was enhanced when brushing the index finger of the rubber hand and the referred phantom index finger on the amputation stump. Interestingly, Rosén et al. showed that the artificial rubber hand does not have to imitate the appearance of a biological hand to be perceived as belonging to the own body [12]. They performed RHI on five upper limb amputees who experienced referred sensations on the same phantom finger when brushing a finger of the robotic hand prosthesis and on the site of the stump. Furthermore, Rosén et al. suggested that the illusion could be maintained even during myoelectric control of the robotic hand prosthesis and tactile feedback. The classical RHI is based on visuotactile stimulation and has been compared to the moving RHI, where both passive and active movements were evaluated [13]. For the passive and active movements, the index finger of the rubber hand and the participant's own index finger were linked with a light stick. Hence, moving either of the fingers will create a synchronous movement. During passive movement, the experimenter moved the light stick, which moved the finger of the rubber hand and the participant's finger synchronously. The participant did the active movements by moving their finger, which will also move the finger of the rubber hand synchronously. Consequently, creating a feeling that the participant is in control of the movements (sense of agency of the rubber hand). The strength of RHI was equally strong in all three cases, suggesting that any combination of multisensory stimulation can induce RHI [13].

Human skin comprises hairy skin and hairless skin (glabrous skin, mainly found on the palms and sole of the feet). Glabrous skin is innervated by four types of mechanoreceptors which respond to different aspects of touch and transmits the information via fast-conducting Aβ myelinated fibres to the dorsal root ganglia and from there onwards to the central nervous system [14, 15]. A fifth mechanoreceptor has been found in the hairy skin, termed C-tactile (CT) afferents that respond to soft brush stroking with velocities at 1–10 cm/s [16] and code for pleasant touch [17]. Crucianelli et al. [18] conducted an RHI experiment on fifty-two females to study the impact pleasant touch has on body ownership by brushing at different velocities on the glabrous skin of the hand. A light touch with a slow stroking speed (3 cm/s) was deemed more pleasant and provided higher levels of subjective embodiment as evaluated by a questionnaire during RHI than a fast stroking speed (18 cm/s). Lloyd et al. [19] showed that a slow stroking speed positively influenced the subjective self-reports on ownership (questionnaire) and the pleasantness rating than during fast stroking speed (30 cm/s). Furthermore, stimulating the hairy skin of the hand, rather than the glabrous skin, showed a greater proprioceptive drift [19].

Vibrotactile, electrotactile, and mechanotactile feedback are commonly used research tools for investigating non-invasive sensory feedback in hand prosthetics. If these tools are used in solitary, when interactions occur with the surrounding, there can be a visuotactile mismatch between what is seen and what is felt. E.g., if a person sees a force being acted on a hand prosthesis but the stimulation felt is vibration, this can be seen as a visuotactile mismatch. D'Alonzo et al. [20] conducted a modified RHI experiment comparing the classical RHI with synchronous/asynchronous brush stroking and tapping, applying matched modality stimulations on both the rubber hand and the real hand. During mismatched modality stimulation, a vibration was applied on the hidden hand while the rubber hand was brushed or tapped. This was to investigate if the mismatched modality could promote the RHI. The stimulation was applied on the pulp of the distal phalanxes of the index and middle fingers. Synchronous tapping induced a vivid RHI but less than during synchronous brush stroking. Vibrotactile stimulation induced ownership of the rubber hand during mismatched modality stimulation, but the RHI was more vivid during brush-vibration than tapping-vibration. D'Alonzo et al. suggested that brush-vibration would induce a more vivid RHI due to the activation of the same mechanoreceptors, which is not the case during tapping-vibration. Shehata et al. [21] compared the classical RHI (brush-brush) to a modified RHI using mechanotactile feedback (tapping-tapping). The mechanotactile stimulation was also evaluated together with motor control of a simulated prosthesis. The responses on embodiment (including location, ownership, and agency) were evaluated. Brush stroking and mechanotactile stimulation (tapping) were compared for the sensory feedback part of the experiment. Tapping-tapping evoked similar embodiment responses as brush-brush (the difference was not significant) and had a strong positive correlation between the conditions. Furthermore, controlling a prosthesis (grasping objects) showed high agency responses, which were not influenced when adding mechanotactile sensory feedback. However, with asynchronous sensory feedback, while grasping objects with the prosthesis, the agency responses were lower than without sensory feedback.

In this study, we compared the commonly used stimulation modalities that can be used in prosthetic hands to evaluate if a type of stimulation modality provides a more vivid RHI, which could contribute to the body ownership of a rubber hand. The stimulation modalities were applied to the participant's hand while the rubber hand received brushstrokes. The different stimulation modalities were compared by conducting a modified RHI. The rubber hand received brush stroking, and the participant's own hand received brush stroking, electrotactile, mechanotactile, and vibrotactile stimulation.

2 Method

2.1 Participants

This study includes two cohorts of participants; twenty-seven able-bodied, right-handed individuals (18 males and 7 females; median age, 34 years; range 25–60 years) and three unilateral transradial amputees (3 males, in the age of 22, 42, and 47). In this study, the amputees are referred to as A1, A2, and A3. A1 was amputated on the right side one year ago. He displayed a phantom hand map (PHM) [22] on the residual limb. On this PHM, he experienced referred sensations from the volar side of the thumb, index and little fingers, as well as part of the palm on the distal part of the residual limb. The PHM was defined by using a pen as previously described ([12, 11, 22]). A2 was amputated on the left side 32 years ago. He experienced a shortened phantom limb, a phenomenon called telescoping, with the whole phantom hand intact and perceived that he could move the phantom fingers. However, he did not experience any referred sensations. A3 was amputated on the right side 18 years ago. He could perceive the movement of the phantom thumb and little finger when activating the residual forearm muscles but experienced no referred sensations. All participants used myoelectric prosthesis, and A3 had the prosthesis attached with osseointegration.

The study was approved by the Swedish Ethical Review Authority (DNR 2021-03630) and was conducted in accordance with the tenets of the Declaration of Helsinki. All participants were informed about the contents of the experiments, both verbally and in writing, and gave their informed and written consent.

2.2 Equipment

Two computers were used in this study, containing two different LabVIEW programs. One computer was available for the participant where a user ID was generated. The participant was asked to fill in demographic information and if they were new to the RHI experiment. After each experimental block, the participants were asked to perform different tests on the computer to assess the RHI. The experimenter had another computer from which the experiment parameters were controlled.

The setup for the RHI included a box (RHI box) with openable lids to obscure and reveal the rubber hand during the experiment. The RHI box was divided into two compartments, where a life-like rubber hand was placed in an anatomically/posturally congruent position on one side, and the participant's hand was placed on the other side (Fig. 1). The RHI box contained customized racks to adjust the position of sensors and actuators. One continuous rotation servo, FS90R (Feetech RC Model Co., Ltd., Shenzhen, China), was used to rotate a rod onto which brushes were attached. Two pairs of infrared (IR) sensors (Adafruit Industries LLC, NYC, USA) were used to detect the onset and offset of the brush stroking via a light transmitter and a photoelectric receiver. The IR sensors were connected to an Arduino MKR Zero. The IR sensors were used to keep count of the brushstrokes and sent an on and off signal to the actuators, which provided stimulation to the hidden biological hand. Three actuators were used to provide different tactile stimulations on the participant's hidden middle finger and the distal part of the residual limb. An HS-40 Nano analogue servo motor (HI-TEC RCD, USA) provided mechanotactile feedback on the hidden biological hand by converting rotary motion into linear motion (similar to what was done in Wijk et al. [23]). The motor was attached to a rod placed above the hidden biological hand, and the height was adjusted to provide a light touch on the skin. For vibrotactile feedback, an eccentric rotating mass motor (ERM) (Vibrating Mini Motor Disc ID 1201, 11000 RPM, Adafruit Industries LLC, NY, USA) was used and secured with tape to be kept in place during the experiment. For the electrotactile stimulation, an electrical stimulator was used to produce biphasic charge-balanced cathodic-first current-controlled pulses. The amplitudes ranged from 0.1 mA to 10 mA with a resolution of 0.1 mA, and frequencies of 100 Hz were used in the experiments. The stimulator communicated with both the Arduino microcontroller and PC. The electrical stimulation was delivered through self-adhesive Pals electrodes (Axelgaard Manufacturing Co., Lystrup, Denmark). The anodal electrode (rectangular 7 × 10 cm Pals electrode) was placed on the ventral side of the forearm. The cathode was placed on the proximal phalanx of the middle finger (circular 2.5 cm in diameter Pals electrode) and secured with tape to maintain the curved shape of the finger to maintain the electrodes in place.

The NI LabVIEW LINX toolkit [24] was used for an interaction with an Arduino Uno to control the actuators in this study.

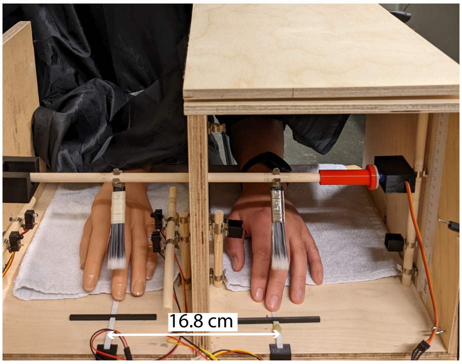


Figure 1: The experimental setup, where brush stroking was applied to the rubber hand and the participant's hand (classic Rubber Hand Illusion). The white rubber band was used as a guide for where to position the middle finger. The distance between the hands was 16.8 cm.

2.3 Experimental setup and protocol

In this study, we investigated four types of feedback (brushstrokes, pressure/force, vibrations, and electrical stimulation), which were tested on different parts of the hand (hairy skin and glabrous skin), with and without time delay (synchronous and asynchronous). To limit the number of combinations, asynchronous stimulation was only performed when the brush was stroking both the rubber hand and the hidden biological hand. Testing these conditions gave a total of 10 combination blocks. Throughout the paper, the stimulation blocks will be coded as [Asynchronous/Synchronous][rubber hand

Stimuli][hidden hand Stimuli]–[Hairy skin/Glabrous skin]. The description of the coding can be seen in table 1. Each participant took part in one session with ten stimulation blocks, where each block provided 30 stimulations, lasting for approximately 100 seconds. The ten stimulation blocks were randomized within the session and among the participants. The given stimulus was applied to the participant's hidden biological hand while a brush stroked the rubber hand. The participant had a 1–2 minute break between each block and was instructed to stand up and relax during the break. The experimental setup and protocol were the same for able-bodied participants and amputees, if not stated otherwise.

Asynchronous/Synchronous	**Stimuli**	**Hairy skin/Glabrous skin**
Asynchronous (**A**)	Brush (**B**)	Hairy skin (**H**)
Synchronous (**S**)	Pressing (**P**)	Glabrous skin (**G**)
	Vibration (**V**)	
	Electrical (**E**)	

Table 1: Abbreviation for the coding of the 10 stimulation blocks.

Identify stimulation amplitude

The initial step of the experimental protocol was to set the amplitude for the electrical stimulation for the participant. Different amplitudes were set for the glabrous and hairy skin. A two-second stimulation with a frequency of 100 Hz was given to define thresholds for; a) just perceiving stimulation, b) pain/uncomfortable stimulation, and c) the level where the participant felt the stimulation distinctly (stimulation level). After finding the sensory threshold, the current intensity was increased by 0.2 mA until a level was reached where the participant found the stimulation uncomfortable. The stimulation level used for the experiment was set to level = 2, on a scale where level = 0 referred to no-sensation and level = 10 maximum pain level. The sensory threshold was set to level = 1, and the uncomfortable threshold to level = 8. After finding the stimulus level, the participant was asked to inform the experimenter where they felt the stimulation. The stimulation should only be felt on one side of the middle finger (glabrous or hairy skin). Some participants perceived the stimulation solely right below the electrode, whereas others felt a stimulation travelling on the whole length of the finger. If the stimulation was felt on both sides, the current intensity was adjusted in steps of 0.1 mA until the stimulation was felt only on one side of the finger. Identifying the thresholds was done on glabrous and hairy skin separately. The mean stimulation level on glabrous and hairy skin for participants with intact hands was 1.67 mA and 1.98 mA.

The stimulation level for the amputees was set by starting at level = 1 and increasing the intensity until they felt a distinct stimulation.

Experimental session: Rubber Hand Illusion

The participant sat on a chair in front of a table facing the experimenter. The chair was set at a proper height, and the armrest was positioned, so the hidden biological hand attained a relaxed position in the RHI box. All jewellery and watches were removed to make the real hand closely match the rubber hand visually. The participant was wearing noise-cancelling headphones and listening to white noise during the experiments to remove any auditory cues that could be used. A sheet covered the participant's shoulder to obscure the arm for the hand in the box.

Before starting the session, a user identification number was generated in the LabVIEW program. The participant's hand was positioned at a fixed distance to the rubber hand. The distance between the rubber hand and the hidden hand was 16.8 cm. After completing the setup, the experimenter indicated that the session would start, and the following steps were performed: 1) the lid of the box was closed, and the participant was asked to perform the proprioceptive drift pre-test, 2) the experimenter opened the lid, and the participant was asked to fix their sight and focus on the rubber hand, 3) when the block was finished, the experimenter closed the lid, and the participant was asked to perform the proprioceptive drift post-test, 4) the participant continued with the questionnaire (9 questions) and finished with 5) rating the pleasantness of the stimulus. The different parts of the experiment can be seen in figure 2. After completing one stimulation block and the RHI tests, the participants took a break while the experimenter prepared for the next block. After finishing the entire experiment (session), the participant had the possibility to write free-text comments about the experience of the RHI.

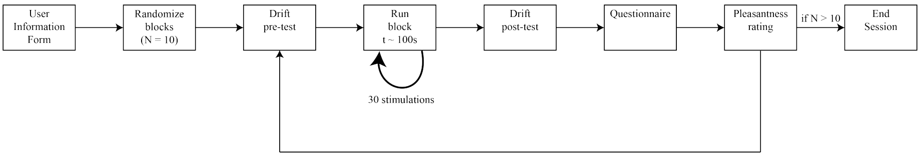


Figure 2: Experimental protocol for one session.

The SBH condition follows the original experiment done by Botvinick and Cohen [6]. The rubber hand was positioned in a congruent position to the hidden hand, and the hidden hand assumed the shape of the rubber hand where the fingers were slightly bent. The brush stroking started at the proximal phalanx of the participant's middle finger and the rubber hand and ended on the middle finger's metacarpophalangeal joint (MCP) joint. For the SGB condition, the fingers of both hands were fully extended to match the brush path, which started at the proximal phalanx to the distal phalanx. For the asynchronous condition (ABH and ABG), the brush stroked the hidden finger approximately 500 ms after the brush stroking on the rubber hand. During mechanotactile, vibrotactile, and electrotactile feedback, the actuators were placed on the proximal phalanx of the middle finger (Fig. 3). The position was chosen due to the onset of the brush on the rubber hand. The differences in brush stroking paths were due to technical reasons since the brush was fixed to follow a circular path with the setup where the accessibility of the middle finger was different when applying stimulus on the glabrous and hairy skin.

The stimulation was applied differently to the amputees than to the able-bodied participants. The amputees without a phantom map (A2 and A3) received stimulation on the distal stump at a central point for all stimulation blocks, where brushstrokes were given on the dorsal and volar sides of the rubber hand. Stimulation on the rubber hand's volar side was considered spatially (locally) mismatched. As for A1, the stimulation was applied on the phantom little finger during brushstrokes on the volar side of the rubber hand. The phantom little finger was chosen for practical reasons: distancing from scars where the sensitivity was reduced. The brush stroking and the mechanotactile stimulation were applied manually by the experimenter to the amputees' residual limbs.

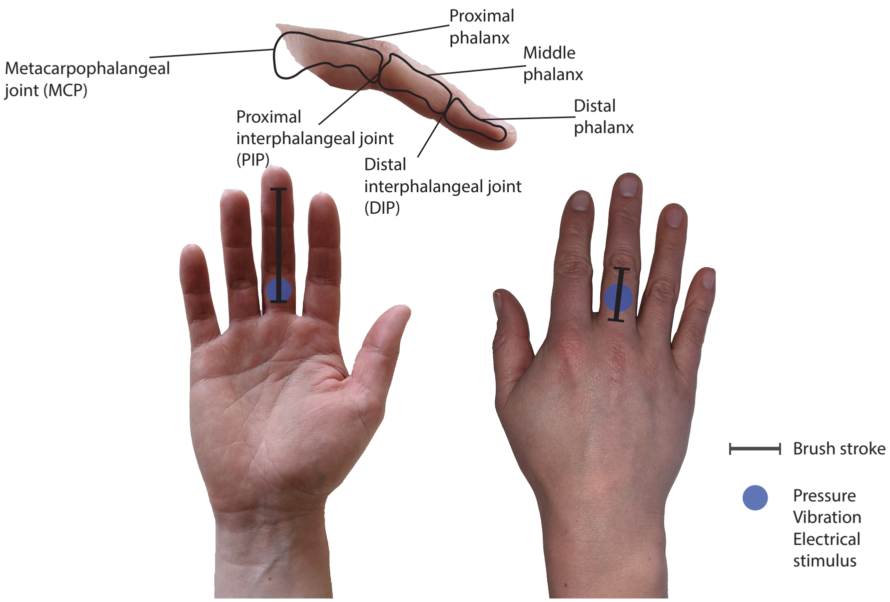


Figure 3: Bones and joints of the finger (upper picture). (Lower pictures) The lines on the middle finger show the brush stroke path, and the blue dot shows the position of the pressure, vibration, and electrical stimulus. On the glabrous skin (left), the brush stroking started on the proximal phalanx and ended on the distal phalanx. On the hairy skin (right), the brush stroking started on the proximal phalanx and ended at the MCP joint.

2.4 Outcome measures

Three tests were performed for each block to assess the RHI; proprioception drift pre- and post-test, questionnaire, and pleasantness test. After completing all stimulation blocks, the participant could by choice, convey comments about the RHI experience.

We hypothesize that stimuli that provide a sensation similar to brush stroking would provide a stronger RHI. Such sensations can be elicited by electrotactile and vibrotactile stimulation since the sensations could match the brush stroking spatially, meaning that the stimulus can be felt on a larger area on the finger. Furthermore, the vibrational and electrical stimulation would stimulate mechanoreceptors that are slowly adapting for a longer time than pressure, providing prolonged and richer information. Additionally, these stimulations could give a tingling sensation which might be similar to brushstrokes. The pressure stimulation would only apply stimulation on a smaller area, which should result in a weaker RHI than the other stimulation types.

Statistical analyses were performed to compare the different conditions and the two different locations, volar (glabrous skin) and dorsal (hairy skin). The analyses were performed in Python, using SciPy [25], Skikit-learn [26], and Statsmodels [27].

Proprioceptive drift test

The proprioceptive drift test is a pointing test which was done before (drift pre-test) and after (drift post-test) each block. A ruler was placed on the box (with the lids closed). The participant was asked to close their eyes and slide their finger on the ruler perpendicular to their hidden hand. They were asked to stop the sliding finger above where they felt their hidden middle finger. The proprioceptive drift was calculated as the difference between the pre- and post-test [28]. A positive drift indicated a drift towards the rubber hand, and a negative drift indicated a drift away from the rubber hand.

Normality was assessed visually using Q-Q plots and the Shapiro-Wilk test, indicating that parametric statistical tests could be used on the datasets. The proprioceptive drift was compared between asynchronous and synchronous brush stroking using paired t-test. This test was also used to compare the proprioceptive drift between hairy and glabrous skin. The synchronous brush stroking was compared with each stimulation type (electrotactile, mechanotactile, and vibrotactile stimulation) using one-way ANOVA.

Questionnaire

Directly after the proprioceptive drift post-test, the participant filled out a questionnaire containing nine questions adapted from Botvinick and Cohen [6] (see list below). The participant rated the statements using a seven-point visual-analogue-scale (VAS). This scale ranged from -3 (”absolutely certain that it did not apply”), 0 (”uncertain whether it applied or not”), and +3 (”absolutely certain that it did apply”). The rating was done by moving the indicator on the scale with a computer mouse using their contralateral hand. After each statement, the indicator was moved to the centre of the scale. Three statements assessed the illusion, which referred to if the sensation was felt on the rubber hand and if the rubber hand was felt as if it were one’s hand. Statement (S)1 and S2 assessed experience of referred touch, and S3 assessed the ownership of a rubber hand. The other six statements served as controls concerning hallucinations of own’s hand. The statements are randomized after each block and for each participant.

- S1 (illusion) *It seemed as if I were feeling the touch of the stimulation in the location where I saw the rubber hand touched.*
- S2 (illusion) *It seemed as though the touch I felt was caused by the brush touching the rubber hand.*
- S3 (illusion) *I felt as if the rubber hand were my hand.*
- S4 (control) *It felt as if my (real) hand were drifting towards the right (towards the rubber hand).*
- S5 (control) *It seemed as if I might have more than one left hand or arm.*
- S6 (control) *It seemed as if the touch I was feeling came from somewhere between my own hand and the rubber hand.*
- S7 (control) *It felt as if my (real) hand were turning rubbery.*
- S8 (control) *It appeared (visually) as if the rubber hand were drifting towards the left (towards my hand).*
- S9 (control) *The rubber hand began to resemble my own (real) hand, in terms of shape, skin tone, freckles or some other visual feature.*

In order to assess how many participants experienced the RHI, an ownership criterion was used [29]: Participants who scored higher than the neutral rating, 0, for the mean score of the illusion statements during synchronous brush stroking condition and had two points or more than the asynchronous brush stroking condition. The criterion was used on all data, including hairy and glabrous skin, but the criterion was also employed on the skin type separately.

The mean rating of illusion statements was compared between asynchronous and synchronous brush stroking to test whether the RHI was induced. Thereafter the mean rating of illusion statements was compared to the mean rating of control statements for all the stimulus types. The mean rating of illusion statements should have a higher rating than the control statements if the participants experienced an RHI [6, 12, 11]. In the BSB condition, the tests were performed on hairy and glabrous skin separately to assess differences in the mean rating of illusion and control statements between the hairy and glabrous skin. The results from the different stimuli were compared in order to assess if a specific type of stimuli induced a more vivid RHI.

Non-parametric statistics were used to assess ratings for the different statements. The Wilcoxon signed-rank test was used to analyse: 1) illusion statements between asynchronous and synchronous brush stroking and 2) mean rating for illusion statements and mean rating control statements for each type of stimuli. The analysis was done on the two different skin types to evaluate if the RHI has been induced when applying a certain stimulus type on a specific skin type. The analysis was done on each type of stimuli to evaluate whether the RHI was induced, including all skin types and separately. The mean rating of the illusion statement was compared between the stimulus types (brush stroking, mechanotactile, electrotactile, and vibrotactile stimulation) for each skin type using the Friedman test followed by a Dunn's post hoc test with Bonferroni correction.

There are a number of persons who are not susceptible to the RHI; it is still unknown why some are susceptible to RHI and others are not [10]. The classification of non-responders and responders is often based on subjective self-reports such as the RHI questionnaire and not according to objective measures. The low correlation between the two measures (proprioceptive drift and subjective self-reports) could result in different non-responders rates [10]. In this study, the number of non-responders was calculated based on an ownership criterion used in a previous study by Zbinden and Catalan [29]; the mean rating was calculated for the illusion statements (S1–S3) and was compared between the conditions, ABH and SBH. According to the criterion, the mean rating should be higher than the neutral rating, and the synchronous condition should rank at least one point more than the asynchronous condition, which was used for the present study.

Pleasantness rating

After the questionnaire, the participants were asked to rate the pleasantness of the stimulation on a VAS which is a subjective measure [30]. The scale ranged from -3 ("unpleasant), 0 ("indifferent") to +3 (pleasant). The rating was done by moving the indicator on the scale with a computer mouse using their contralateral hand.

The Mann-Whitney U-test was used to assess differences in pleasantness rating between hairy against glabrous skin for all conditions. Friedman test with Dunn's pairwise post hoc test (using the Bonferroni correction to adjust the p-value) was used to assess if there were any differences in pleasantness rating between the stimulus types grouped by skin types (hairy and glabrous skin).

3 Results

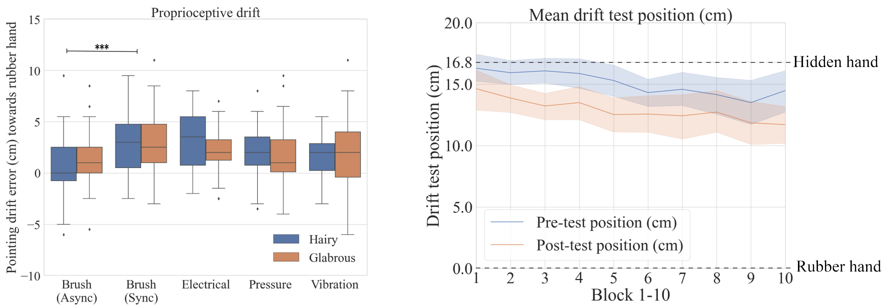
3.1 Proprioceptive drift

Able-bodied participants

There was a significantly greater drift towards the rubber hand following synchronous brush stroking (2.99±3.35 cm) compared to asynchronous brush stroking (1.17±3.16 cm) as shown by a paired sample t-test [t(106) = -4.01, p = .00019].

When comparing proprioceptive drift as a result of the different stimuli, the highest proprioceptive drift was seen with brush stroking, followed by electrical stimulation (2.65±2.68 cm), vibration (2.00±2.84 cm), and pressure (1.93±2.83 cm) (Fig. 4a). However, the differences between stimuli were not significant, determined by a one-way ANOVA [F(3, 51) = 1.64, p = .18]. Furthermore, the proprioceptive drift was not significantly different between hairy and glabrous skin using paired sample t-test.

As can be seen in figure 4b, the pointing position in the pre-test showed drifting towards the rubber hand after the completion of each of the stimulation blocks. The difference between the pre-test and post-test was slightly less after more completed stimulation blocks. A paired t-test was used to analyse the difference in pointing positions for the first and last stimulation blocks. This analyse showed that the pointing position was significantly closer to the rubber hand in the last block both in the pre-test [t(52) = 2.43, p = .02] and in the post-test [t(52) = 3.33, p = .003].



(a) (b)

Figure 4: a) Proprioceptive drift for each stimulus: Brush asynchronous stroking (BA), brush synchronous stroking (BS), pressure (P), vibration (V), and electrical stimulation (E). The proprioceptive drift towards the rubber hand was significantly greater with synchronous brush stroking than asynchronous brush stroking (p<.001), b) mean values from the drift test (cm) with a 95 % CI for all stimulus types. The figure shows that the pointing position already drifted towards the rubber hand before starting each block. 0 cm indicates the position of the rubber hand, and 16.8 cm is the position of the hidden hand.

Amputees

A1 had the highest drift towards the rubber hand when pressure was applied on the forearm than during brush stroking on the dorsal side of the rubber hand (Fig. 5). For amputees A2 and A3, the drift was following all types of stimulus as well as after asynchronous brush stroking.

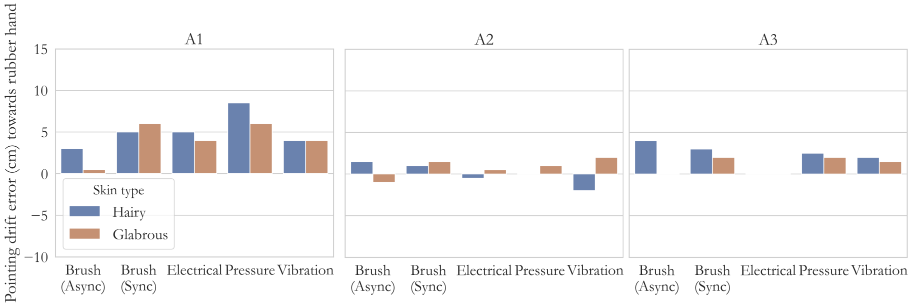


Figure 5: Proprioceptive drift for each amputee: The amputee who had shorter time elapse since amputation had greater drift towards the rubber hand than the other two amputees.

3.2 Questionnaire

Able-bodied participants

Figure 6 shows the pattern where the three illusion statements (S1–S3) tend to have more positive ratings than the control statements (S4–S9). Calculating the mean rating for the three illusion statements and the control statements, the illusion statements have a higher rating than the control statements (1.26 and -0.88) during synchronous brush stroking. In contrast, the rating is -0.91 (illusion statements) and -1.50 (control statements) during asynchronous brush stroking.

According to the Shapiro-Wilk test, the data was not normally distributed ($p<.05$).

Wilcoxon signed-rank test showed a difference when comparing the results from the illusion statements following synchronous and asynchronous brush stroking when the brush stroking was applied on hairy skin ($p<.001$) and glabrous skin ($p<.001$).

Furthermore, Wilcoxon signed-rank test showed a significant difference in rating comparing illusion statements and control statements following asynchronous brush stroking (glabrous: $p<.01$, hairy: $p<.05$), synchronous brush stroking (glabrous and hairy: $p<.001$), electrical stimulation (glabrous and hairy: $p<.001$), pressure (glabrous and hairy: $p<.001$), and vibration (glabrous and hairy: $p<.001$).

The Friedman test was used to analyse the rating of the illusion statements following each stimulation modalities. The test showed a significant effect of stimulation modalities on the ratings of the illusion statements (glabrous and hairy: $p<.001$). Dunn's post hoc analysis (using the Bonferroni correction to adjust p) showed a significant difference in the illusion statements between brush stroking and pressure (hair and glabrous: $p<.01$), where the mean rating is for brush stroking was 1.22 (glabrous) and 1.29 (hairy), and for pressure -0.26 (glabrous) and 0.14 (hairy).

A pairwise Wilcoxon signed-rank test was applied for each stimulus type comparing the results from hair and glabrous skin. The test showed no significant difference between hairy and glabrous skin for either stimulus type.

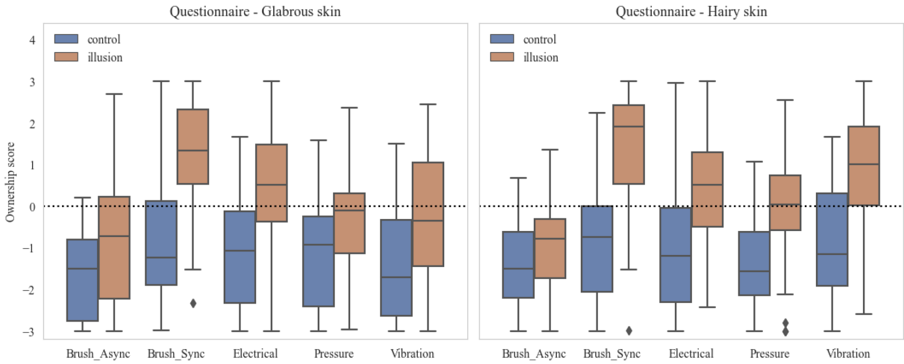


Figure 6: Results of questionnaires. The three illusion statements are marked with (*), and the rest (S4–S9) are control statements. The results show all stimulus types. The traditional RHI had a significantly higher rating than other stimuli. However, all types of stimuli induced the illusion.

Amputees

The amputees disagreed with the illusion statements (Fig. 7). A1 rated one of the illusion statement, *"It seemed as though the touch I felt was caused by the brush touching the rubber hand"* towards an agreement of 1.04, which had the same rate for one of the control statements, *"It seemed as if the touch I was feeling came from somewhere between my own hand and the rubber hand".*

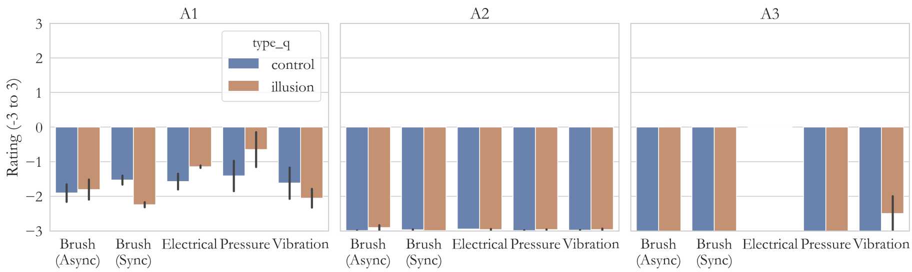


Figure 7: The bar plot shows the results of the questionnaire for each amputee. The two amputees who had lost their hand many years ago showed an apparent disagreement to the statements in the questionnaire.

3.3 Pleasantness rating

Able-bodied participants

Participants rated pleasantness higher with brush stroking (Fig. 8). They rated slightly higher when brush stroking was applied to hairy skin compared to glabrous skin for all stimulation modalities besides for electrical stimulation.

The Shapiro-Wilk test showed that data were non-normally distributed (p<.05). Thus, non-parametric tests were used to analyse data.

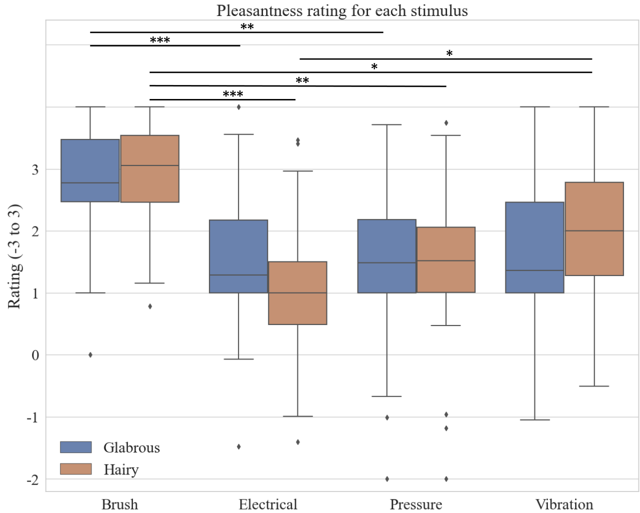


Figure 8: Pleasantness rating for each type of stimulus, where brush stroking was experienced as the most pleasant.

The pleasantness rating was compared between the two skin types, with all stimuli included, using the Mann-Whitney U-test. This test was used due to a different sample size (data loss), where the data was not saved after taking the test. There was no significant difference in pleasantness rating between the two skin types (glabrous and hairy skin). When results for each stimulus were analysed electrical stimulation on glabrous skin was close to significant ($p=0.06$).

The pleasantness rating on hairy skin was analysed between the different modalities using the Friedman test. It showed that the stimulation modalities had a significant effect on pleasantness rating ($p<.001$; Friedman test). Performing Dunn's post hoc test (using the Bonferroni correction to adjust the p-value), brush stroking was rated significantly higher than all the other stimulation modalities (electrical: $p<.001$, pressure: $p<.01$, and vibration $p<.05$). Vibration was rated significantly higher than electrical stimulation ($p<.05$). The Friedman test showed that stimulation modalities had a significant effect on the pleasantness rating when stimulating glabrous skin ($p<.001$). The Dunn's post hoc analysis (using the Bonferroni correction to adjust the p-value) showed that brush stroking was significantly more pleasant than electrical stimulation ($p<.001$) and pressure ($p<.01$).

Amputees

The pleasantness rating for each stimulus by the amputees can be seen in figure 9. A1 mentioned that pressure was most pleasant (rating: 1.03) and was rated higher than the brush (rating: 0.94) during the glabrous condition. A1 rated all the stimulation types similarly pleasant, 1.20–1.56. A3 experienced all the stimulation modalities as either pleasant or unpleasant (neutral pleasantness ranking).

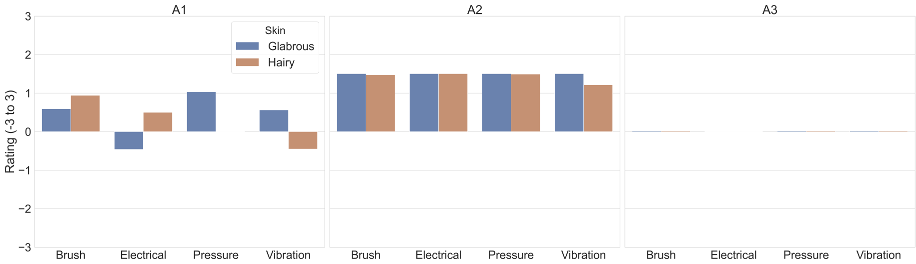


Figure 9: Pleasantness rating for each stimulus: Brush (B), pressure (P), vibration (V), and electrical stimulation (E).

4 Discussion

Previous studies have shown that if a hand amputee perceives their prosthetic hand as their own, they tend to use it more intuitive [31]. In order for this to happen, sensory feedback from the prosthesis is essential. In research, mechanotactile, vibrotactile, and electrotactile stimulation are common non-invasive methods to convey sensory feedback from a hand prosthesis to the amputee. This study assesses to what extent different types of stimulus, both mismatched and modality-matched, can induce the RHI. Furthermore, it also assesses if the possibility of inducing the RHI is different depending on if the stimulus is applied to the glabrous or hairy skin in the hand.

Electrical stimulation was perceived as uncomfortable due to the particular type of sensation it evokes, which is similar to when you hit your funny bone. Due to the uncomfortable nature of electrical stimulation, some participants had a lower uncomfortable threshold level hence a lower stimulation level which was experienced as a more local sensation directly below the electrode. Other participants who were more comfortable with the electrical stimulation could go higher in the uncomfortable threshold level; hence their stimulation level was perceived as a sensation on a larger area on the finger which better matched the brush stroking path. If the stimulation level could be set to match the sensation of the brush stroking, the sensation would probably be more intuitive and entail higher ratings in the tests. Participants who experienced a tingling sensation that travelled along the finger expressed a similar sensation to the brush. Some comments from those participants were: *"The electrical stimulation was most similar to the brush stroke."* and *"The electrical stimulation was surprisingly similar to the brush stroke, but pressure worked pretty good as well."*. This suggests that electrical stimulation can spatially match the brush stroking and, at the same time, provide a pleasant stimulation for an individual. For a more comfortable electrical stimulation some factors could be changed, e.g., decrease the frequency in order to increase the intensity (preferably 50–100 Hz to provide continuous sensation), use bigger electrodes, or prepare the skin with alcohol to remove dead skin to get an even distribution of the current.

A3 expressed that he did not feel that the rubber hand belonged to him. He mentioned that after 18 years without a hand, he became well aware of the loss and speculated that he probably could have been more susceptible to the RHI early post-amputation. This is supported by Mayer et al. [32], where amputees who are aware of their hand absence do not consider the prosthesis as their own even if they can see the prosthesis and sense their phantom hand. Ehrsson et al. [11] mentioned that as years go by without a hand, the amputee's perceptual system learns to accept the new body image without the hand and becomes less prone to experiencing the RHI.

4.1 Proprioceptive drift test

The proprioceptive drift was significantly greater with synchronous brush stroking than asynchronous, which is in accordance with the traditional RHI [6]. There was no significant difference in the proprioceptive drift toward the rubber hand between the stimulation modalities, indicating that stimuli with matched and mismatched modalities would contribute similarly to a proprioceptive drift towards a rubber hand.

For some participants, the illusion was strong even before starting the stimulation blocks which was later in the trial (e.g. block 2–10). This can be seen in the drift pre-test, where the participants almost pointed at the rubber hand's middle finger. This corroborates a prior study [28], where a drift towards the rubber hand was shown, even without stimulation. In addition, this drift increased gradually with sustained brush stroking [28]. The gradual drift in this study resulted in a small difference between pre- and post-test and would not show the actual drift towards the rubber hand. Botvinick and Cohen [6] demonstrated that the accuracy of pointing towards the rubber hand increased in proportion to the reported duration of the illusion. In this study, these patterns were not predicted since the breaks in between the stimulation blocks were added with the belief that the illusion would be broken between the stimulation blocks.

A1 had a phantom map, and only one year had elapsed since amputation compared to 32 years (A2) and 18 years (A3). According to Ehrsson et al. [11] findings, amputees with referred sensations reported higher scores on the illusion statements than amputees without, but the illusion scores were not significantly different. Furthermore, they showed no relation between the proprioceptive drift test and the time elapse since amputation.

4.2 Questionnaire

Previous studies where the RHI have been used have most often applied brush on hairy skin. Therefore, the ownership criterion for this study was employed on the data done on hairy skin, which showed that 6 of 27 (22.2%) were non-responders. Whereas there were 9 (33.3%) non-responders during conditions following stimulation of glabrous skin. However, the number of non-responders was not significantly different in hairy and glabrous skin. Therefore, if the ownership criterion was employed on stimulation on hairy and glabrous skin, 33.3% were non-responders. To replicate prior research, the statistical tests for evaluating if the RHI occurred were applied first on results of stimulation of hairy skin, then on results following stimulation of glabrous skin. Lastly, the difference between these two conditions was evaluated, which showed no significant difference in the RHI following stimulation on the hairy and glabrous skin. Therefore, the final tests were made on all data, including results from stimulation on hairy and glabrous skin. The literature has reported a high rate of non-responders (participants who are not susceptible to RHI), 23–28% [6, 13, 33], suggesting that the number of participants should be high to make any conclusion about the RHI.

The ownership criterion is arbitrary, but the commonly used is that the illusion statements should score at least one point above the neutral/indifferent rating [13, 33, 6, 34], and some only apply this for one of the illusion statement (S3) [35]. The interpretation of non-responders is assessed differently in previous studies, where some categorize non-responders as those who do not fulfil the ownership criterion, whereas some define responders and non-responders differently. E.g., where the responders had a positive rating (>1) for the ownership statements and non-responders had a negative rating ($\leq$0) [34] which would leave some participants in the rating scores 0–1, which is not included in either of the categories. If using the cut-off criteria where the mean rating for the illusion statements is $\leq$0 in this study (including data for both hairy and glabrous skin), there are 5 (18.5%) non-responders for the synchronous brush stroking, 9 (33.3%) for electrical stimulation, 12 (44.4%) for pressure and vibration. Trojan et al. [37] performed a pre-test to exclude non-responders (20–30%), where a similar criterion was used for the statement S3. Ehrsson et al. [33] also performed a pre-test to exclude non-responders (28%). This study showed a higher rate of non-responders compared to previous studies. One possible explanation could be that none of the participants has participated in an RHI study, which could contribute to a lower phenomenological control where the participants were unable to generate an experience that would meet the expectancies of the RHI [38]. It has been shown that trait phenomenological control can favour RHI ownership statements, suggesting that the questionnaire does not measure ownership but rather measures the ability to generate experiences to meet expectancies [38]. Some participants commented *"Once I felt an effect of the illusion, I think it increased a bit with time, so maybe the illusion would have been stronger with more time."* This was discussed by Riemer et al. [10] analysed and discussed the methodological differences in the RHI and suggested that the assessed RHI onset time (the time when the participant first perceived the feeling of ownership) varied between studies that included and excluded non-responders where the onset time was usually shorter in studies which excluded non-responders. In this study, all participants were included, which would cause skewness in the data compared to studies that exclude non-responders [10]. However, the methods in classifying non-responders are based on subjective reports (questionnaire) and not on objective measures (proprioceptive drift), and the literature shows an inconsistency in whether the proprioceptive drift correlates with the ownership ratings [10]. Hence, individuals that are classified as non-responders with one measure could classify as responders with another measure [10].

The illusion statement was ranked higher for brush stroking, providing a matched modality sensory feedback where the brush stroking was applied on both the rubber hand and the hidden hand. There was only significant difference between brush stroking and pressure. Electrical stimulation and vibration had a higher mean rating for the illusion statements than pressure, suggesting the two former modalities matched spatially with the brush stroking on the rubber hand. Moreover, both electrotactile and vibrotactile stimulation deliver dynamic stimuli, which hypothetically provide more activation (firing) of the hand's receptors during sensations across the receptive field. Some participants who had a higher stimulation level for the electrical stimulation could feel a tingling sensation extending along with the finger. This imitates the brush stroking travel path to some extent. For the ERM, the vibration elicits waves which probably propagate across the skin, leaving a sensation on a more extensive area of the finger. Stimulation using pressure induced a weaker RHI than the other stimuli, being the most spatially mismatched stimulus since the sensation is discrete and more defined than the other stimulation modalities, which gives a sensation in larger areas and is less defined. This would cause a more significant visuotactile conflict to the brush stroke.

4.3 Pleasantness

There was no significant difference in pleasantness rating for hairy skin versus glabrous skin. This seems reasonable according to Löken et al. [16] who showed that the pleasantness rating for brush stroking on the palm (glabrous skin) was not significantly different from the arm (hairy skin) if the brush stroking was alternating between the skin types, indicating that brush stroking the palm was equally pleasant as brush stroking on the arm. It was also suggested that the central processing is different for the hairy and glabrous skin since pleasantness in glabrous skin can be influenced by stimulation on the hairy skin; this effect does not apply vice versa. The brush stroking speed is ideally 3 cm/s for CT stimulation to show a significant difference in rating for glabrous and hairy skin [16]. This study used a brushstroke speed of 6 cm/s, which could slightly differ between the stimulation blocks. The slight variation in speed was due to the positioning of the brush since the friction between brush and rubber/biological hand varies depending on the height, and the servo motor was not strong enough to keep a stable speed. However, the variation in speed should not have any effect on pleasantness ratings on glabrous skin [30]. Due to the friction, the speed is assumed to be <6 cm/s, which would still be within the range of 1–10 cm/s where brush stroking was perceived as most pleasant [30]. For future studies, the pleasantness could be examined by applying stimulation on either glabrous or hairy skin for all types of stimuli, adding a short break, and finishing with the other. The order of the skin types could be randomized between the participants. This method could show significantly higher rating when stimulating on the hairy skin compared to glabrous skin since the alternation effect would not apply in this case [16].

Brush had a significantly higher pleasantness rating than the other stimuli on both hairy and glabrous skin, where stimulation on hairy skin was rated slightly higher than on glabrous skin. This could be explained by brushstrokes giving a light and slow-moving touch, which is effective for activating CT afferents [39]. Stimulation using vibration was ranked as the second most pleasant. Huisman et al. [40] showed that using multiple vibration motors created a haptic perceptual illusion to spatially match a brush stroke's path gave a similar subjective pleasantness rating as actual brushstrokes. Even though vibration does not activate CT afferents, Huisman and colleagues showed that the pleasantness rating followed a U-shape pattern similar to brushstrokes. This study only used one vibration motor. However, the perceived sensation of the vibration motor covered a large part of the finger, which can be seen as spatially matched to the brush stroke path. Today's electronic devices (phones, tablets, gaming consoles, etc.) use vibration as haptic feedback. This commonly used modality could positively reinforce the experience of pleasantness. Furthermore, Aβ afferents transmit information about discriminative touch, which is processed in the somatosensory cortex and represents learned touch based on previous remembered tactile experiences [41]. In contrast, CT afferents are processed in the limbic-related cortex (non-learned touch) and cannot be activated in isolation without activating Aβ afferents. A variety of perceived sensations was seen in the comments from the participants about vibration: *"Vibration was the most pleasant and felt like it induced more illusion, second to the standard brush."* while another participant commented: *"Vibration may have been a bit too strong to be realistic."* Electrical stimulation was the least pleasant, which could be explained by the fact that the majority of the participants were novices to this type of stimulus. It is not a familiar sensation that is encountered in daily life, and thus some experience it as an uncomfortable and unusual feeling which could have affected the pleasantness rating. Due to the parallel activation of CT afferents and Aβ afferents, CT afferents would act as a selector to distinguish velocities related to pleasantness during social touch to be further processed in the posterior insula, which is an affect-related cortex area [42]. It was also suggested that the discriminative processing in the SI and SII could influence the tactile processing in the posterior insula [43]. In the same way, the affective coding in the insular cortex could modulate responses in SI and SII. Posterior insula might also be activated during observational touch [44], seeing someone else being stroked by a brush. Based on this fact, seeing the rubber hand receive brushstrokes might affect the pleasantness rating on the other types of stimuli. With this in mind, future studies should investigate the pleasantness rating for the same stimulation modalities, excluding the visual input from the rubber hand to test the tactile feedback solely.

4.4 Limitations

A few participants' hidden fingers occasionally got tics, making the participant aware of their hand's position and disrupt the RHI, which could have interrupt the RHI since moving the finger will make one aware about the location of his/her own hand.

Two subjects could not remove their wedding rings, which might have affected the results. However, their data showed that they experienced the illusion.

Another limitation of the automatic setup compared to manual brush stroking would be while brush stroking the glabrous skin, the participant needed to position their hand the palm-side up while keeping the middle finger straight instead of slightly bent (during a relaxed position). This position might be difficult for some, which was also mentioned by one of the participants, *"It was difficult to keep arm with palm facing up. The arm wanted to naturally bend so I had to put in slight effort to keep it in place. This was not the case for when the palm was facing down".* The constant tension during the stimulation might affect the illusion in two ways: 1) the tension could make the participant more aware of their own hand's position and 2) the tension could hamper the participants ability to concentrate during the experiment. Tension was a problem during brush stroking since the brush was set to a specific position to provide light brush stroking starting on the proximal phalange. Furthermore, the servo motor was too weak to handle too much friction. To compensate for this, a soft brush and attaching the stick (that was holding the brush) only on one end. In this way, having one loose end would yield the brush on parts of the finger where there is more friction.

The experiment with amputees was limited in numbers. Due to the high rate of unresponsive to RHI in previous studies, the amputees in this study could be included in the group that is not susceptible to RHI. However, in a previous study, amputees have mentioned to experience ownership over their prosthetic when used in daily life even though they did not experience RHI [29].

5 Conclusion

In this study, we evaluated how different tactile stimuli (mechanotactile, electrotactile, and vibrotactile), that are commonly used in sensory feedback systems in prosthetic hands, can induce the RHI. We showed that all stimuli elicit body ownership of a rubber hand to some extent. The RHI becomes more vivid with tactile stimuli that imitate the received sensation on the rubber hand. In this case, electrotactile stimulation and vibrotactile could more closely imitate the brush stroking spatially compared to mechanotactile stimulation.

A slight drift towards the rubber hand could be seen after completing more stimulation blocks (time spent in one session), which gave unreliable results in the proprioceptive drift test and made it difficult to interpret whether the RHI occurred or not.

In contrast to previous studies, there was no difference in applying the stimulation on the hairy or the glabrous skin. However, future studies are needed to assess if the pleasantness rating depends on the order of the other stimulus types.

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