proCover – Sensory Augmentation of Prosthetic Limbs Using Smart Textile Covers

Joanne S.L. Leong

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Declaration

I hereby declare and confirm that this thesis is entirely the result of my own original work. Where other sources of information have been used, they have been indicated as such and properly acknowledged. I further declare that this or similar work has not been submitted for credit elsewhere.

Hagenberg, June 27, 2015

Joanne S.L. Leong

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Kurzfassung

Sämtliche heute erhältliche Prothesen verfügen über keinerlei Sensorik, die taktile Empfindungen erfassen und an ihre Träger übermitteln können. Während sich die Forschung vor allem auf High-End-Lösungen konzentriert, welche direkt in zukünftige Prothesen integriert werden sollen, präsentiert diese Arbeit eine alternative Möglichkeit, die schon bei heute gebräuchlichen Prothesen eingesetzt werden könnte. Eine neuartige textile Sensorsocke könnte eine nichtinvasive, leicht anwendbare und individualisierbare Möglichkeit darstellen, herkömmliche und auch kostengünstige zukünftige Beinprothesen mit Sensoren auszustatten. Im Dialog mit acht Beinprothesenträgern wurde ein Konzept für tragbare Sensoriken ausgearbeitet, die die speziellen Anforderungen von Prothesenträgern berücksichtigen. Die Diskussionen zeigten, dass vor allem die Individualisierbarkeit der Sensorlösung von großer Bedeutung ist, um den einzigartigen und auch wechselnden Anforderungen von Prothesenträgern gerecht zu werden. Basierend auf den so gewonnenen Erkenntnissen wurden neue Interaktionsmethoden zum dynamischen und individuellen Erstellen von Verbindungen zwischen Sensorregionen auf der Socke und tragbaren Aktuatoren in Form eines Vibrationsarmbands entwickelt. Der entwickelte Prototyp wurde in einer Pilotstudie von vier Beinprothesenträgern getestet, um die generelle Verwendbarkeit und Nützlichkeit in verschiedenen Szenarien zu evaluieren. Die so gewonnenen Erkenntnisse ermöglichen einen Ausblick wie textilen Sensoren in der Zukunft verwendet werden könnten, um Prothesen kostengünstig mit Sensoren auszustatten und amputierten Menschen trotz der Notwendigkeit einer Prothese ein reichhaltiges Erfassen ihrer Umwelt zu ermöglichen.

Abstract

Today's commercially available prosthetic limbs lack tactile sensation and feedback, while many recent advancements in this domain are focused on sensor technologies designed to be embedded into future prostheses. We present a novel concept and prototype of a prosthetic-sensing sock that offers a non-invasive, self-applicable and customizable approach for the sensory augmentation of a wide variety of lower-limb prosthetics, including present-day and future low to mid-end varieties. Through questionnaires and discussions with eight lower-limb amputees, we map the design space for sensing wearables for prosthetics and uncover the need for custom sensing solutions in this area to accommodate the unique and changing sensing needs of prosthesis-users. Based on these insights, we develop novel interaction methods for dynamic, user-driven creation and mapping of sensing regions on the foot to wearable haptic feedback actuators. We then perform an early assessment of our prototype in a pilot-study with four amputees, where we explore its utility in scenarios brought up by the amputees. We summarize our findings from the process and establish future directions for research into using smart textiles for the sensory enhancement of prosthetic limbs.

Chapter 1

Introduction

At present, many people rely on the use of a prosthesis to replace a limb or substitute a missing part of an arm or leg. Following the loss of a limb, prosthetics are critical devices that can help people regain the ability to perform and participate in a number of different physical activities. There are many reasons why a person may incur the loss of a limb. Some of the most common reasons include birth defects, cancer, physical trauma, and circulatory problems from diseases such as diabetes [29]. In 2005, it was estimated that approximately 1.6 million people were living with limb-loss in the United States alone. However, the number is expected to rise to approximately 3.6 million by the year 2050 [30], with the number of lower-limb amputees currently outnumbering the number of upper limb amputees at a ratio of four to one [37]. In light of this, many researchers are putting concentrated effort into improving these devices to better serve this important and growing group of people. Research into prosthetics is multidisciplinary, and spans across many different aspects ranging from health and comfort, to control and mobility.

The design and construction of prostheses that can emulate a natural sense of touch is of particular growing research interest. Over the last few decades, a number of solutions have been developed for the detection of pressure, slip, heat and texture [22]. Many of these are centered upon embedded sensor technologies, with the objective of restoring sensory capabilities for people who have lost a limb and must then rely on a prosthesis. However, many of the exciting innovations in this field will likely remain out of reach for many people, due to a multitude of factors pertaining to service accessibility, health status, personal attitudes towards elective surgery, and very prominently – monetary cost. Fortunately, sensory substitution devices in the form of wearables present a promising avenue for tackling some of these issues. Crafted carefully, such technology has the potential to be a cost-effective, self-applicable and non-intrusive means for prosthesis-users to augment their prostheses with sensory capabilities.

1.1 The Gap Between Low and High-End Prosthetics

The cost of a new prosthetic leg can be prohibitively expensive. Prices range from 5,000 to upwards of 50,000 USD [38], and are driven by the need for customization and high quality construction. A prosthetic limb must be uniquely crafted to fit the specific wearer's body, and at the same time must be built in a sturdy and fail-safe manner. A poor fit can lead to complications such as skin irritation and pressure ulcers, whereas technological failures can lead to serious physical injuries such as broken bones if the leg construction collapses beneath a user. Furthermore, prostheses are not onetime investments. Instead, they should be regularly replaced every couple of years which can dramatically increase the monetary burden carried by many amputees. Therefore, despite the present day availability of high-end prostheses with various advanced features (e.g. programmable, EMG-controlled, etc.), many people still rely on low-end prosthetics, such as those that operate on the basis of simple mechanisms like levers and straps.

The range of research directions being taken in the domain prosthetics is similarly very broad. On one hand, there is a low-end research stream, focused on making prosthetics cheaper and more broadly accessible. The demand for low-cost solutions has fueled the popularity for both 3D-printed and do-it-yourself (DIY) type prosthetics [9, 10]. On the other hand, there exists a high-end research stream in which researchers attempt to push the boundaries of what is possible with current prostheses. In recent years, advancements on this side have improved how humans can interface with prostheses both physically and mentally. From a physical perspective, changes in the mechanical and electrical design of prostheses have paved the way towards greater mobility. For instance, the development of powered prostheses have made it easier and less physically demanding to walk [1]. Mentally, research has demonstrated the possibility for brain-controlled prosthetics (prosthetics that can be controlled with thought), as well as prosthetics capable of triggering realistic tactile sensations in users [18, 23].

However, many of the high-end sensory technologies being developed are meant to be built directly into prosthetic limbs. As such, they cannot be applied retroactively. This presents a gap whereby present day prosthetics that lack sensory capabilities, as well as future low-to-mid range prosthetic solutions cannot benefit from these advancements. Therefore, we aim to introduce a low-cost sensing wearable that can be applied retroactively to prosthetics and can help close this gap.

1.2 *proCover*

This thesis presents *proCover*, a novel prosthetic-sensing wearable that offers a non-invasive, self-applicable, cost-effective, and dynamically customizable approach for the sensory augmentation of lower-limb prostheses. The wearable is constructed using a three-layer smart-textile composition that forms a pressure-sensing matrix. Sewn into a sock form-factor, the textiles are capable of enveloping and providing all-over sensing for a prosthetic foot. In the form of longer stockings or knee guards, the textiles can be worn over a prosthesis to extend sensing capabilities higher up the leg and even over the knee-joint. By driving motors in a vibration band worn around the arm with pressure measurements gathered from the textile garment, *proCover* restores a sense of feeling to prosthesis-wearers by substituting normal touch sensation with vibration feedback. The result is a specialized smart wearable concept for prosthetics, pictured in Figure 1.1, that enables users to adapt its functionality to fit their unique needs across a variety of different situations and activities.

Summarizing, the main contributions of this thesis are:

- A novel concept and prototype of a textile wearable that can be self-applied and retroactively used to augment a wide range of lower-limb prosthetics with customized sensing capabilities, and which offers coverage beyond the plantar region of a prosthetic foot.

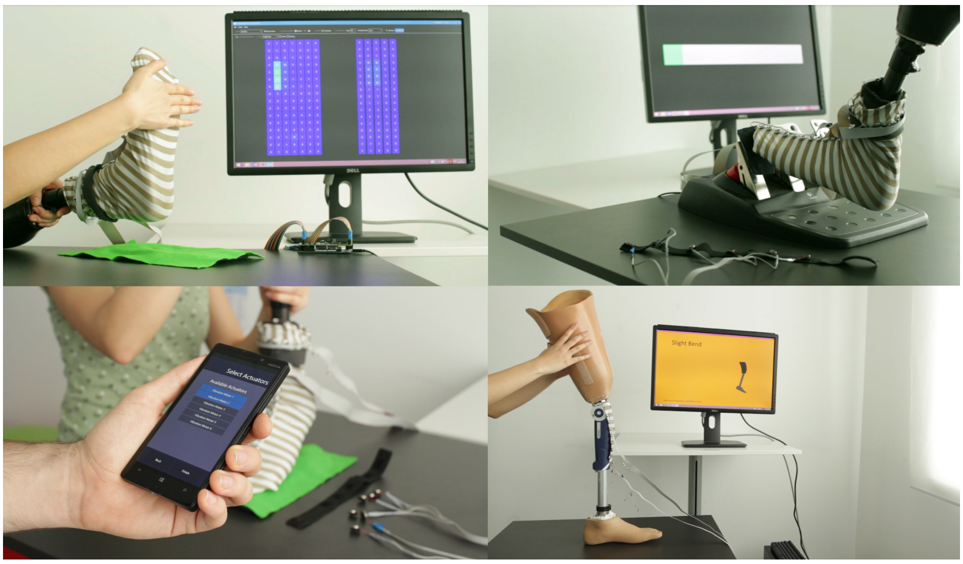


Figure 1.1: *proCover* allows for touch discrimination (top-left), pressure variability (top-right), dynamic user customization (bottom-left) and bend detection (bottom-right).

- Novel interaction techniques that allow for the customization of the sensing capabilities for prosthetic limbs. This includes the ability to both dynamically create distinct sensing regions from a high-resolution matrix of sensors and map them to feedback actuators.
- Outline of the design space for prosthetic sensing socks through extensive questionnaires and discussion with eight lower-limb amputees.
- An early assessment of the design of sensing textile wearables and their applicability to real users in different scenarios in a final pilot-study conducted with four lower-limb amputees.

1.3 Outline

This thesis covers all the details pertaining to the *proCover* project. We begin by providing an overview of related work in the domain of prosthetics and sensory substitution techniques (see Chapter 2). With this as a basis, we then outline major design considerations for creating sensory substitution wearables for prostheses (see Chapter 3). Afterwards, we create a first-prototype. The technical details of the implementation are explained thoroughly in Chapter 4. After creating a first prototype, we investigate its real-world potential by conducting a pre-study with eight lower-limb amputees. Based on the insights derived from information gathered from both questionnaires and face-to-face discussions (described in Chapters 5 and 6) we refine the prototype and incorporate additional features to address the concerns of the amputees. The technical implementation details for these developments are described in Chapter 7. Finally, we conduct an in-lab pilot-study (described in Chapter 8) with four of the previous eight study participants where they try on the refined wearables. The study is divided into three tasks in order to assess the utility of the wearables in the context of different scenarios our participants had brought up in the pre-study. We conclude by discussing the limitations of our work and presenting possible future directions for research in this area in Chapters 9 and 10.

Chapter 2

Related Work

2.1 'Feeling' in Biomechatronic Prosthetic Limbs

Recent advancements have made it possible to enable amputees to regain near-natural physical sensations through the use of artificial limbs that either directly or indirectly stimulate nerve endings. The use of electrodes, which encircle or pierce nerve bundles have facilitated real-time grasp perception as well as near-natural touch perception in prosthetic hands [18, 23]. Artificial fingertips enabling wearers to discriminate between different textures have also been made possible with the use of an electrode inserted into a nerve in the arm [15]. However, in our work we omit the use of invasive surgical procedures and implants, for which the process may be complex and for which the long-term effects are still being carefully studied [23] . Instead, we focus on wearable systems with haptic feedback mechanisms, which provide a less invasive and more cost-effective alternative for sensory feedback for prosthetic legs.

2.2 Non-Invasive Sensory Feedback for Prosthetics

Many systems were designed to improve balance and gait. Fan et al. [6] created a haptic feedback system comprising of four piezoresistive force sensors mounted on a leather insole and corresponding pneumatic balloon actuators mounted on a cuff worn on the middle thigh. Sabolich et al. [20] used pressure sensors adhered to the plantar surface of the prosthetic foot to relay pressure information via transcutaneous electrical stimulation. Crea et al. [3] as well as ORPYX® Medical Technology [5] have also explored the use of vibration feedback on the thigh and back respectively, driven by pressure-information from sensorized shoe insoles. Employing a similar technique with vibration motors embedded in the prosthetic socket and driven by discrete force sensitive resistors (FSRs) mounted on a shoe insole, Egger [39] discovered that even near-natural sensations could be elicited when the motors were applied to a patch of skin with regrown nerves on the patient's stump. However, these works have taken a generalized approach to introduce sensing into lower-limb prosthetics, since they have been designed to offer the same sensor configuration for each user. Additionally, they appear "sole-focused" – positioning discrete, hardware-based pressure sensors located exclusively along the sole (plantar side of the foot). In contrast, our work seeks to explore the utility of sensing applied to the whole surface of the foot, including the edges and dorsal side of the foot, and investigates the possibility for user-driven sensor configurations.

2.3 Electronic Skin and Smart Textiles

Tactile sensing technologies such as electronic skin (e-skin), artificial skin with human-like sensory capabilities [8], have applications in a breadth of disciplines ranging from medicine to aerospace [25]. While non-textile based approaches exist to creating electronic skin, many of which are promising in the field of prosthetics [13, 24, 27], we choose to focus on a textile-based approach. The reason for this is that non-textile based approaches require that they are embedded or adhered to prosthetic limbs. In contrast, textile-based sensors can be worn over prosthetics like ordinary clothing, allowing for a more accessible means for sensing that can be easily applied to a broad spectrum of prosthetic limbs.

Flexible, stretchable piezoresistive fabric is available for a wide variety of pressure-sensing applications, ranging from e-skin for robotic limbs [32] to smart casts capable of detecting a good fit [4]. Such fabric also has applications in more traditional wearables. While Büscher et al. created a dataglove [2], Sensoria Fitness [35] developed commercially available smart socks with three embedded textile-based pressure sensors in the sole of each sock to monitor running. Pressure-sensitive socks have also been developed by Perrier et al. [17] to help prevent pressure foot ulcers in diabetic patients, while embroidered sensing socks were developed by Alphafit GmbH to manufacture custom fit shoes for people with diabetic foot syndrome [33, 34, 26]. The broad applicability of piezoresistive fabrics was demonstrated in FlexTiles [16], where the authors showed its applications in automobiles and furniture in addition to wearables. Yet none of these works considered using fabric to augment prosthetics, which in itself is a challenging problem since prosthetics take on various shapes and sizes.

2.4 Customization in Prosthetics

Prosthetics need to be highly customized to ensure a good physical fit for the wearer. However, more precedent is now being given not only to custom fits, but custom functionality and style. Hofmann et al. [9] explored how a design process can engage users to create assistive technology that better meets their own unique needs, and Torres [36] created a prosthetic arm which enables children to construct an arm from LEGO®. In this work however, we will explore customization concerning sensing needs.

Chapter 3

Design Considerations

There are a variety of aspects that must be taken into consideration when prototyping a wearable for the sensory augmentation of prosthetic limbs. These are described below in the following sections.

3.1 Aspects of Sensory Substitution Systems

On a high-level, we observe the creation of touch-sensitive prosthetics as having two main sides: sensing and feedback. Sensing involves the detection and measurement of a multitude of different sensations such as pressure, slip, temperature, and proprioception [22], while feedback refers to the means in which the system interacts with the human body to relay information. As shown in Figure 3.1, a mapping between these two aspects is necessary to transform data collected from sensors into signals, which the user can then interpret. There are also many technical approaches for both sides.

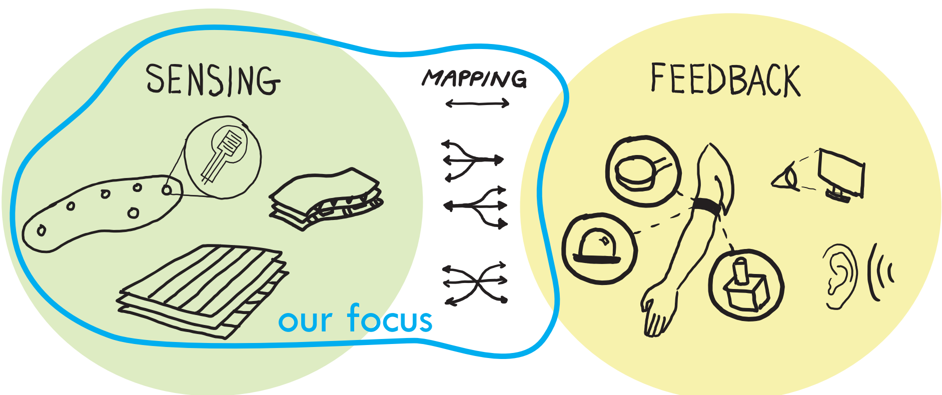


Figure 3.1: The problem domain can be viewed as having a sensor aspect mapped to a feedback aspect. In this work, our primary focus is on the sensing and mapping aspects.

While there are a multitude of feedback possibilities, in this work we focus on the sensing and mapping aspects of the problem. While it is recognized that users benefit from custom-fitting legs and ones that are programmable or specifically designed for different types of physical activities (e.g. walking, biking, running, climbing), we noticed that the approach taken to develop sensing solutions for prosthetic legs has been in contrast, inflexible. To our knowledge, no previous research has been conducted into using stretchy, high-resolution pressure-sensitive fabrics to create a wearable-sensing layer for prosthetics. However, we see potential for smart fabrics to provide novel, dynamic, customizable sensing solutions when combined with innovative mapping strategies. Using high-resolution pressure-sensitive fabrics would allow us to have enough pressure points at hand to change the mapping accordingly to meet the needs of the different users and their custom-fitted legs as well as the different physical activities they engage in.

3.2 Material Properties for Wearable Sensors

A number of factors should be taken into consideration when designing or choosing sensors to be worn on the body. In the situation where we want to augment prostheses with a sense of touch, such factors include the geometry of the human figure, bodily ranges of movement, and sensing input resolution (as depicted in Figure 3.2).

In wearable applications, it is ideal for sensors to be flexible and stretchable. In contrast to many popular touch-input devices that operate on a 2D plane (e.g. smartphones and tablets), body-worn sensors must be able to conform to more complex 3D geometries. Both hands and feet for instance vary in volume along their length and end in separate digits. By having stretchable rather than rigid hardware sensors, sensors can fit snugly over the surface of the appendages and more accurately conform to their phys-

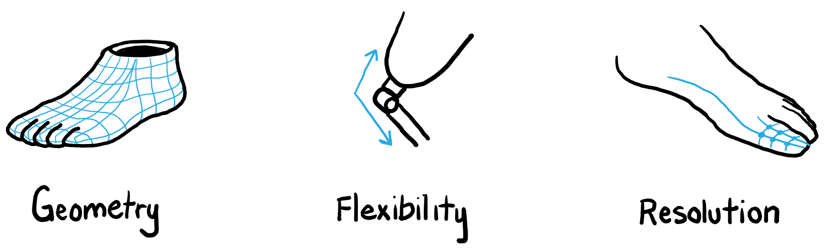


Figure 3.2: Many factors should be taken into consideration when making a wearable. The wearable should be constructed to fit the complex geometry of the human figure and should be flexible enough so as not to restrict movement. The sensing resolution should be sufficient to meet the needs of the wearer.

ical shapes. This would help to capture more relevant sensing information pertaining to the human figure. Furthermore, each part of the human body has its own unique range of movement which must also be considered when designing wearables. Materials should have a reasonable degree of stretch and flexibility so as not to restrict desired bodily movements. For a prosthetic leg, the sensing material should stretch under bending deformation such that it does not jam the hinge mechanism of the knee and impede the bending movement about the joint.

The issue of resolution is also significant when considering touch perception. At first glance, it would seem that the highest possible resolution is desirable for an input sensing layer and that approaching higher sensing resolutions would bring the sensing capabilities of a prosthesis closer to that of a real human limb. However, human limbs have a limited resolution for tactile perception (described further in section 3.3). Therefore, how sensory information captured by the system is processed and transformed into sensory feedback that can be understood by the user presents a bottleneck in the overall design of the system. So long as this bottleneck persists, the value of applying high-resolution sensors to prostheses remains diminished.

3.3 Tactile Feedback Resolution and Human Perception

While a high-resolution pressure map can be achieved, we note that one-to-one mappings between sensors and haptic actuators is unsuitable due to limitations in human tactile perception. Two-point discrimination thresholds (TPDT) are a measure of spatial tactile acuity, defined as the minimum spatial distance needed for a person to distinguish between two simultaneous stimuli from a single stimulus [12]. While it is influenced by a multitude of factors including bodily location and stimulus-type, previous research suggests that the TPDT for the fingertip and back for a static touch is 3mm and 39mm respectively [28]. Thus, there is clearly a limitation to the number of actuators that can be placed on a part of the body to represent sensor information. This is even more crucial for vibration-based stimuli, since vibrations are conducted readily through the body.

Chapter 4

proCover Prototype

As a first step in the design process, we decided to prototype a sensing sock system, which we named *proCover*. In this way, we could firstly explore the engineering challenges involved and assess the technical feasibility of the concept. Secondly, the resulting prototype would be a physical and interactive demonstration of the concept, which could be used to engage in critical discussion with prosthesis-users who have had no prior experience or familiarity with the concept of sensory substitution using a combination of smart textiles and haptic feedback.

The implementation of both hardware and software components was necessary to realize a working version of the *proCover* concept. In reference to Figure 3.1, hardware components are in the form of sensors and feedback actuators, while software is needed to map or transform the sensed signals into an output signal that could be used to drive the actuators. In this chapter, the resulting prototype is first described. Afterwards, both the process of creation and the technical details for each hardware and software component are explained in detail.

4.1 *proCover* System Description

The initial prototype of the proCover system pictured in Figure 4.1 features a pressure sensing sock with a three-layer fabric composition and measurement electronics (a custom shield mounted on an Arduino Due) for the sensing. For feedback, the system consists of a haptic feedback band with six vibration motors driven by an Arduino Micro. The software provides a foot-visualization that displays the location and intensity of pressure applied to different parts of the prosthetic foot. The software also determines vibration intensity for each motor based on both touch location and pressure.

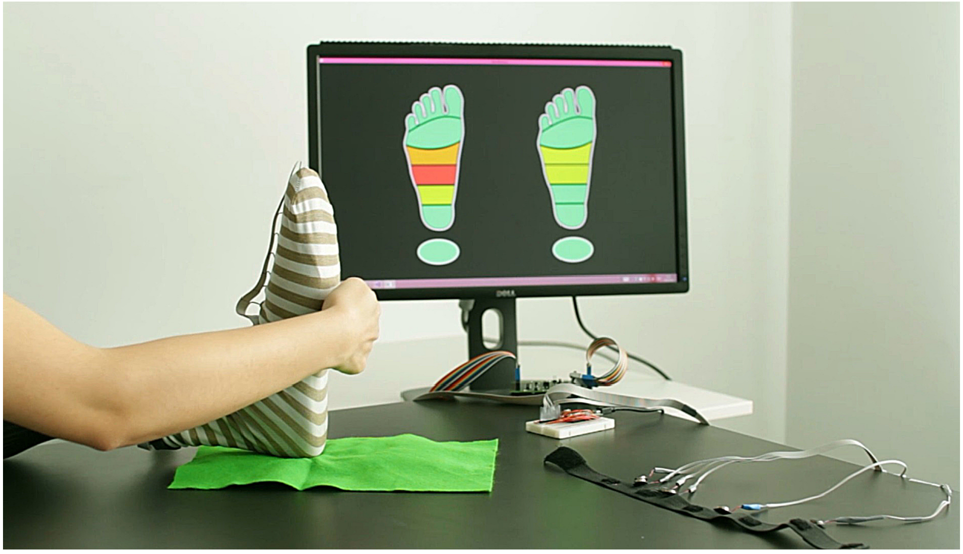


Figure 4.1: The sensing sock prototype consisted of a fabric-based sock and vibration armband driven by independent microcontrollers.

4.2 Hardware Implementation

proCover, pictured in Figure 4.1, consists of a textile-based sensor sock, electronics (wiring, and microcontrollers connected to a PC), and a vibrotactile band as hardware components.

4.2.1 Textile-Based Pressure Sensors

To create a textile-based pressure sensing matrix, we used the three layer fabric approach described in FlexTiles [16]. Two layers of *Narrow Stripe Zebra Fabric*[1], aligned orthogonally to one another and sandwiching a layer of *Eeonyx EeonTex^{TM} LG-SLPA fabric* were assembled to create the three layer composition. The zebra fabric has alternating strips of conductive and non-conductive fabric that are 8.125 mm and 9 mm wide respectively. The grey piezoresistive material has a specialized coating that allows it to behave as a variable resistor. As such, when mechanical force is applied to it, its resistivity drops. Both of these fabrics have four-way stretch (meaning they can stretch both crosswise and lengthwise). This stretchiness makes them suitable to create a wearable that can envelop the irregular 3D geometry of a prosthetic foot. A picture and diagram of this three-layer composition is provided in Figure 4.2.

A single sensing intersection tested from 25 to 1,000 g shows a high dynamic resistance change (6 kΩ to 0.42 Ω, $SD = 0.28$). However, we note

[1]This fabric is distributed by HITEK: https://www.hitek-ltd.co.uk/fabrics

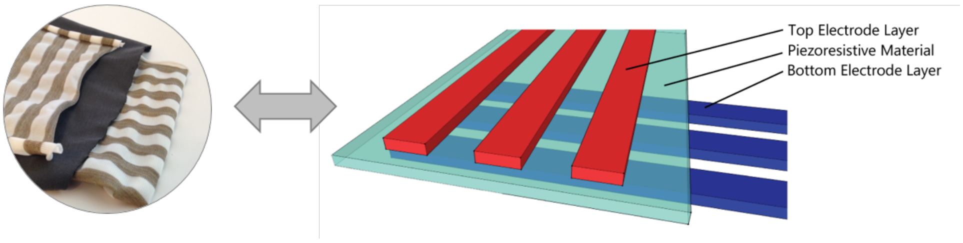


Figure 4.2: Pressure sensing matrix. Photo of three-layer fabric composition (left). Diagram of three-layer fabric composition creating a pressure sensing matrix with nine sensing intersections (right). Three rows (red) and three columns (blue) of conductive lines sandwich a layer of piezoresistive material (represented by the transparent blue layer).

that the pressure sensing matrix cannot be used as a scale. This is due to data loss that can occur between the single sensor cells. Instead, the matrix can be used to show a good relative force distribution.

4.2.2 Measurement Electronics and Electrical Connections

The measurement electronics used to read data from the pressure sensing matrix were the same as in FlexTiles [16]. A custom-built shield containing multiplexers (74HC4051) and shift registers (74HC595) was mounted on an Arduino Due microcontroller (SAM3X8E) with an internal analog digital converter (see Figure 4.3). Sensors in the matrix are measured sequentially at a rate of roughly 5 μs per sensor. Changes in resistivity are measured via the voltage change of a reference resistor connected in series.

Special steps had to be taken to connect the stretchy fabric layers to the measurement electronics. Electrical connections were established between the fabric and the shield using a combination of metal snap fasteners, ribbon cables, and jumper wires. Since the fabric was stretchy, Jersey snap fasteners needed to be used. The multiple blunt teeth of a Jersey snap fastener minimize the risk of tearing the fabric and help ensure that the fastener stays in place when the fabric stretches. The socket halves were attached to the fabric, whereas the stud halves were soldered to the stripped ends of the ribbon cables. Soldering was done with lead-free solder wire (RS SAC305, 0.71mm diameter). The soldering iron was heated fairly high (approximately 275 degrees Celcius) since the solder wire and the snap fasteners had a large surface area that tended to quickly dissipate heat. Twenty-pin flat grey ribbon cables were used alongside insulation-displacement contacts (IDCs). In this way, male to male jumper wires could be used such that the fabric could be plugged or unplugged from the measurement shield for easy transportation and debugging of the *proCover* prototype. A close-up of the electrical connectors are pictured in Figure 4.4.

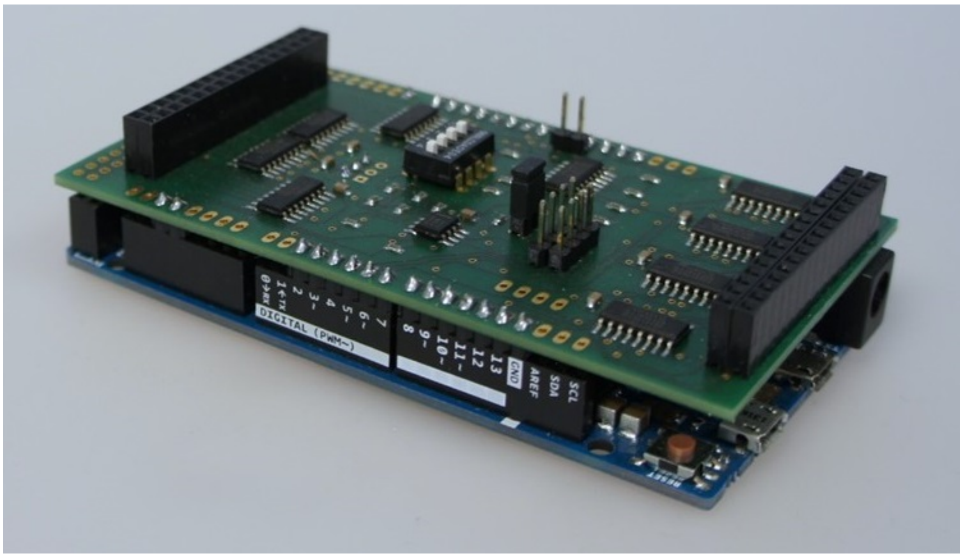


Figure 4.3: Measurement hardware for sensing sock prototype. A custom shield built in-house is mounted on an Arduino Due. Ribbon cable from the sock is plugged into the shield such that sensor readings can be taken.

While this approach decoupled the sock from the measurement hardware making them exchangeable modules that were easy to debug, it had the drawback that the soldered connections were vulnerable to breaking. It helped to ensure that studs were soldered in equal spacing with respect to each other, such that an equal distribution of tension could be maintained across them. It also helped to bury each lead in the groove of each snap fastener to distribute tension away from the soldered connection and minimize

Figure 4.4: Electrical connectors. Male snap fasteners, soldered to the ends of ribbon cable, are used to connect the fabric to the measurement hardware.

the risk of it twisting and breaking. Furthermore, it was helpful to strategically locate snap fasteners on the wearable in order to minimize abrasion against external objects which could weaken the soldered connections.

Other strategies could be employed to reduce strain on these connections and make them more robust. On the connections themselves, it could help to apply dabs of hot glue to provide more strain relief. It could also help to fixate the ribbon cable to the fabric garment to reduce the movement of the wires with respect to the metal snap fasteners.

4.2.3 Sewing a Sensing Sock

The sensing sock was designed using the same three-layer fabric composition described above. As such, zebra-fabric layers aligned orthogonally to one another and sandwiching the piezoresitive layer were used. However, instead of combining them to create a flat sheet, each layer was sewn into the shape of a sock. This was needed to create a deformable and stretchable pressure-sensing matrix which could be used to envelop the more complex 3D geometry of a prosthetic foot.

Creating the sock from this fabric required an understanding of sewing, fabric properties, as well as an understanding of the shape of the human foot. The first step in creating the sock was to investigate and select a template that could be used to sew a sock from the different types of fabric. While numerous templates exist, we had the special requirement that it should be compliant with the electrical properties of the aforementioned textiles such that the pressure sensing would be possible. Furthermore, the selected template should also allow for a snug fit around the foot. Lastly, the template should allow snap fasteners to be arranged such that the corresponding wires leading out from the sock would neither interfere with the interaction nor be at major risk of breakage.

Three different templates were researched and considered for sewing a sock from sheets of fabric. Typically, to ensure a good fit, templates account for the fact that the shape of the foot is thicker at the heel and tapers towards the toes. This is shown in Sock Template 1, 2 and 3 (pictured in Figure 4.5). These templates offer more material by the heel section to account for the fact that the heel section of the foot has the greatest circumference. However, as can be seen by the superimposed stripes of conductive fabric, they would require an unfavourable configuration of snap fasteners which includes the placement of fasteners on the heel of the foot. This would be unideal as it would interfere with the feeling of balance for the wearer when standing, and would also put the electrical connections at high risk of breaking. Therefore these three templates were not used for creating the sensing sock.

Due to the problems with snap fasteners pointed out in the previous three templates, we created a fourth template with a simplified geometry to address this issue. This template (pictured in Figure 4.5, bottom), offers

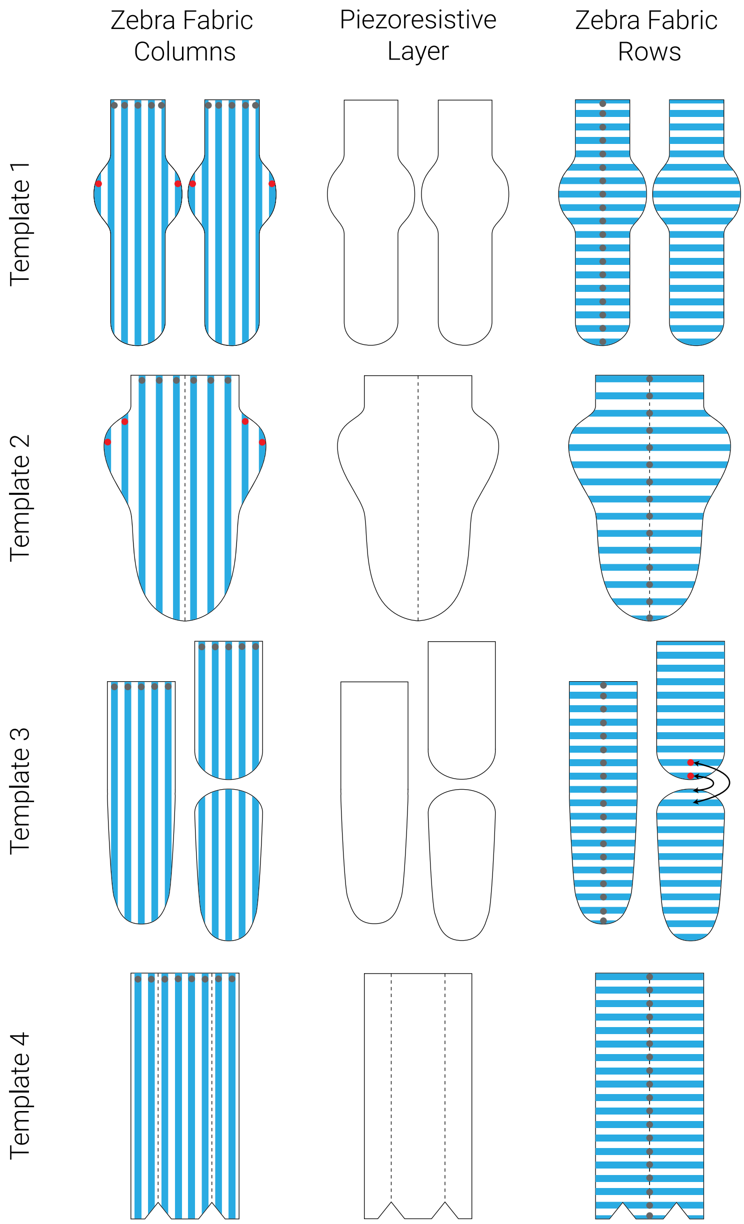


Figure 4.5: Sock Templates. Snap fasteners marked in red would need to be located on the heel of the foot for the conductive lines, making the corresponding templates undesirable. Template 4 was used to create the sensing sock. Although it did not offer more fabric to accommodate the thicker heel of the foot, the stretch in the fabric helped it fit. As it required one piece of fabric per sock, the template helped minimize the risk for error in aligning the conductive rows together.

Figure 4.6: Process shots from making the sock. A sheet of zebra fabric is being sewn into a sock using a sewing machine (top-left); a single sensing intersection is hooked up for testing its change in resistance under pressure (top-middle); female snap fasteners are attached to each conductive line on a sock (top-right); the three layers are ready (bottom-left); male-snap fasteners are soldered to the ends of each wire in a ribbon cable (bottom-middle); a column-sock with wires attached to the sock via snap fastener connections (bottom-right).

the advantage that the snap fasteners could then be positioned along the dorsal side of the foot for the rows and around the top ankle opening for the columns rather than being positioned along the sole of the foot. However, the drawback of this template is that it does not provide additional material for the heel of the foot. This would be particularly problematic if the fabrics used did not stretch, since the sock would then not be able to fit over the foot past the heel. However, since both the zebra fabric and piezoresistive fabric had considerable four-way stretch, the sock was still able to be donned. We noted however that the the sock then had a looser fit towards the toes of the foot, and a noticeably tighter fit around the heel of the foot. With regards to the pressure readings, the sensor matrix consequently had a higher pressure reading by the heel by default. However, this default pressure difference between the toes and heel of the foot could compensated in software by implementing an offset (described in the software section of this implementation chapter).

Once the template was selected, three fabric cutouts were made as illustrated under Template 4 in Figure 4.5. This included two cutouts of zebra fabric (each with a different conductive line alignment), and one cutout of the piezoresistive fabric. The dimensions of the cutouts were estimated by pinning the fabric sheets around a real foot to determine the necessary width and length of each cutout. For the first prototype, the number of columns and rows were chosen based on the fit of the sock around a real, 8.5 US (39 EU) shoe-sized left foot. The result was 16 rows long and 12 columns wide. The process of creating the sock is pictured in Figure 4.6.

Each layer was sewn using a sewing machine (Bernina Model B330), set up with a regular presser foot and a Stretch sewing needle. The Stretch needle, rather than a standard Universal needle, was needed to work with the stretchy zebra and piezoresistive fabric. Stretch needles have a rounded tip, whereas a Universal needle has a sharp tip. The rounded tip is better suited for use with stretchy fabric, as it moves around the threads in stretch fabric and preserves them. In contrast, a sharp or point-tipped Universal needle easily pierces through threads in stretch fabric, thereby causing them to snap, which destroys the integrity of the fabric. Since a regular sewing presser foot was used, the seams had a tendency to be wavy. For better results where seams remains flat, it is recommended that a walking foot be used instead. In this way, pressure is applied by both the foot and the feed dog to the top and bottom layers of fabric respectively such that the top and bottom pieces of fabric pass through the machine at the same pace and do not become stretched or misaligned during the sewing process.

The right sides (i.e. the side of the fabric superior in appearance) of each fabric cutout was sewn face-to-face, following standard sewing practices. This is to achieve the best aesthetic results, wherein the best side of the fabric is outward-facing, whereas the non-aesthetic side of the fabric as well as the seam faces inward for a garment. One can typically identify the fabric-type of a sheet of fabric by examining both sides of the sheet up close under a magnifying lens. Special characteristics depending on the type of fabric can be observed to determine which side is the right side of a fabric. For 'jersey' (also referred to as 'single-knit') fabrics such as the zebra fabric, the fabric curls *towards* the right side when a cut is made perpendicular to the grain. For the piezoresistive fabric, the glossy side rather than the matte side of the fabric is the right side of the fabric.

Regular, non-conductive cotton thread was used to sew the seams. Seams featured a 0.5 cm margin, which is important to give strength to the seams and prevent them from easily coming apart. The edges of the stretch fabric tended to curl, making it difficult to sew near the edges of each cut-out. To counter this, the fabrics were carefully pinned flat and sewn slowly to ensure proper alignment was maintained. Other possibilities to handle this would have been to use tear-away stabilizer or interfacing, or spray-on fabric stiffeners (which should later be removed so as not to interfere with the conductivity of the fabric). Non-conductive thread was used as it was important that no shorting would occur between rows of the sock. The cotton thread was sewn in using a zigzag stitch pattern to make the socks robust under stretch. Zigzag stitches, as opposed to a regular straight stitch would allow

Figure 4.7: Three textile layers (left) come together to form the *proCover* sensing sock prototype (right).

some give in the fabric both along the length and width of the sock making seams less likely to break under tension.

Along the toe seam of the sock with conductive columns, non-conductive fabric was sewn in between to separate the two layers. This would be needed to prevent lines that travel down the length of the foot along the dorsal side of the foot from connecting with the conductive lines travelling along the sole of the foot. This was necessary to prevent ambiguity in the measured pressure signals. If the lines were to electrically connect to one another, a signal created by a force applied to the top of the foot would not be distinguishable from a signal created from force applied to the sole of the foot, since they would share the same matrix row and column indexes.

As a final step, female snap fasteners were attached to each conductive row and column to facilitate connections to the microcontroller. The resulting sock, pictured in Figure 4.7, contained a total of 192 sensor intersections (16 rows $\times$ 12 columns), providing a resolution of 1.6 sensors/inch^2 to cover a foot with the approximate shoe size of 8.5 US (39 EU).

4.2.4 Haptic Feedback Band

While the sensing sock provides us with sensory input, we required a form of sensory feedback to complete the sensory substitution system. We chose to use robust, low-power, low-cost vibration motors as in [14, 21, 39], to affix to various parts of the body (see Figure 4.8) for sensory feedback. However, we note that many forms of feedback exist which can be incorporated into the design of sensory systems for prosthetics, such as nerve-interfacing electrodes and pneumatic actuators (as mentioned in Section 2.2). As the focus for this work was on sensing and sensor-feedback mapping aspects (see Figure 3.1), an active exploration into a wider range of actuator technologies was left for future work.

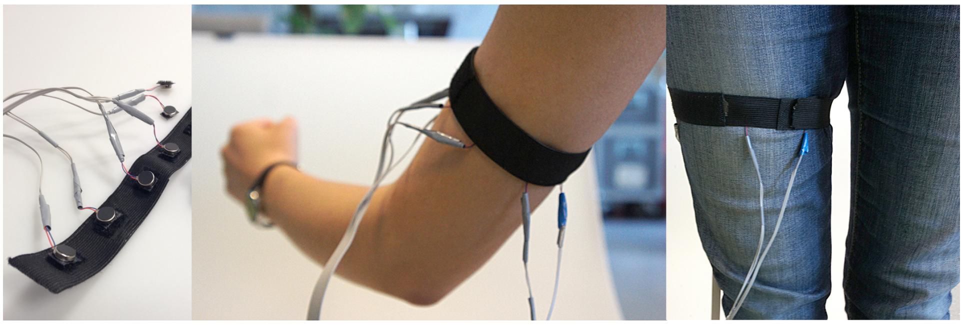


Figure 4.8: Array of vibration motors used for haptic feedback. The motors could be attached to different bands, and could be worn on different parts of the body such as the upper arm or thigh.

Two versions of the haptic feedback band were created. In the first version, the motors were mounted directly to an elastic band using adhesives. However, it was later realized that an impermanent mounting of the motors would be advantageous. In this way, the circuitry would be kept independent and interchangeable with different bands, increasing its versatility and making it easily applicable to different parts of the body using different lengths of bands. Thus, we created a second and final version of the haptic feedback system (pictured in Figure 4.8). This version was created using six vibration motors (Pico Vibe™ 10mm vibration motors) which were mounted with Velcro onto different lengths of stretchable elastic band and were controlled by an Arduino Micro board. These bands could then be worn around different parts of the body (e.g. arm, leg, torso, etc.). The motors were worn on the inside of the cuff directly against the skin in order to maximize sensation for users and make stimulation from each motor more easily distinguishable from one another. Sensor data from the sensing sock was used to drive these motors. The behaviour of the motors was governed by the software, described in the next section of this chapter.

4.3 Software Implementation

The software for this prototype handled the raw sensor data in order to drive a visualization for the forces applied to different parts of the foot. In addition, software was written to drive the vibration motors using the raw pressure data received from the sensing sock. The corresponding programs were written in WPF/C# and the Arduino programming language.

4.3.1 Foot Visualization Software

In the foot visualization pictured in Figure 4.9 (right), two images of a left foot are displayed on a black background. On the left, the pressure applied to the sole of the foot is shown. On the right, pressure applied to the top of the foot is displayed. The foot on each side is divided into different regions: *toes*, *ball*, *arch*, *heel*, and *ankle*. The arch is further divided into three segments (top, middle, bottom) to achieve more detailed visual feedback. This creates a total of seven segments per foot, including the oval segments that represent the front and back of the ankle. The segments can appear in three different colours. Green signifies no pressure, yellow signifies light pressure, orange signifies moderate pressure, and red signifies high pressure. The pressure readout for each segment was mapped to the peak pressure measured across all sensing intersections in the given sensing region.

For a sock with 192 sensors created from a 16 row by 12 column sensing matrix as pictured in Figure 4.9 (left), five columns were reserved for the top of the foot, whereas seven columns were reserved for the bottom of the foot. Two rows were used for the toes, three rows were used for the ball of the foot, five rows were used for the arch of the foot (with 2, 1 and 2 rows for the top, middle and bottom of this region respectively), three rows were used for the heel, and three rows were used for the ankle. In this prototype, the sensors included in each segment were hard-coded. However, it would be possible to enable the dynamic calibration of these segments in future iterations of the software.

Furthermore, the number of sensing regions created on each foot at this stage were chosen based off common terminology used to refer to different parts of the sole of the foot. This however sparked the question: “What

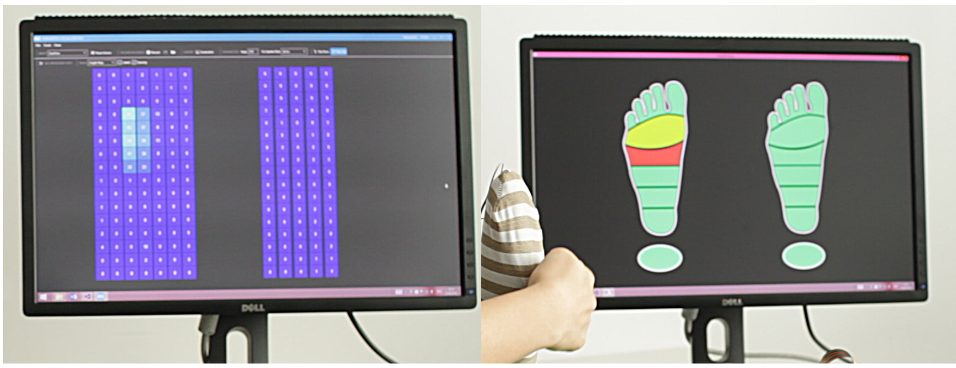


Figure 4.9: Software visualizations of pressure readings from the *proCover* sock. Individual sensing intersections from the sock are visualized using software developed in PyzoFlex [19] (left). Sensing intersections are grouped into distinct sensing regions along the bottom and top of the foot, as well as the ankle. Warmer colours indicate higher pressure readings (right).

sensing regions would make the most sense to have from the perspective of a lower-limb prosthesis user?" We decided to tackle this question in a pre-study with lower-limb prosthesis users (see Chapter 5).

4.3.2 Controlling Haptic Feedback

One-to-one mappings were created between segments and motors. One motor was assigned to each of the the six segments (*toes*, *ball*, *top-arch*, *mid-arch*, *bottom-arch*, and *heel*) along the sole of the foot. The intensity of the vibration for each motor was mapped to the peak pressure measured across all sensing intersections in the given sensing region. Therefore, the vibration feedback would correspond with the information displayed in the foot visualization. Since constant stimulation is undesirable in certain situations (for example, when the wearer stands still), the active vibration motors would time-out after 3 seconds of activation. This behaviour mirrors the behaviour of the motors in the sensing prosthetic leg created by Egger [39]. The time-out would reset however when the pressure level for the sensing region would drop below a moderate level of pressure.

4.3.3 Background Subtraction for Noise Removal

An offset function was implemented to ensure that tension in the sock when it is worn over the foot would not register as pressure readings in the system. As mentioned previously, the fabric template used to construct the sock meant that the sock would by default have greater tension around the heel of the foot than around the toes of the foot when worn. This is because no additional fabric was given to account for the greater thickness in this area. However, in the state where the foot is raised off the ground and no external objects are pressing against the foot, there should be no detected pressure on any region of the foot. To achieve this, the offset function was implemented. On a keypress of the spacebar, the current pressure measurements for each sensing intersection would be logged in an offset array. The offset array would then be subtracted from the subsequent pressure measurement arrays read in from the sensing sock. Doing this would cancel out the default pressure readings, such that no pressure is detected in the case where no object is in contact with the foot and sock.

Chapter 5

Pre-Study: Understanding Sensing Needs

In order to design a sensing wearable that would suit the needs of those with prosthetics, we decided to investigate more deeply what people's sensing needs would be, including possible associated factors such as their amputation-type, beliefs, and activities.

5.1 Method

In consultation with the eight lower-limb amputees, we investigated (a) the implications and potential of having pressure sensing on all surfaces of the foot, (b) the acceptability of a textile form factor for a sensing solution, (c) customization and personalization in the context of sensing for prosthetics, and (d) possible factors that influence users' sensing needs. We presented the participants with a demo of *proCover* and collected data in the form of a questionnaire. Overall, participants took twenty minutes to an hour to complete the questionnaire. We also collected insights through discussions with our participants.

5.2 Participant Demographics

Eight lower-leg amputees who use lower-limb prostheses, (3 female, 5 male) answered the questionnaire. Seven of them had one lower-limb amputation. Three participants had a transtibial (below-knee) amputation. Four participants had a transfemoral (above-knee) amputation. One participant had a double amputation (right: ankle disarticulation, left: below-knee). The participants ranged from 37 to 74 years of age ($M = 60.13$ years, $SD = 13.81$). While seven out of the eight participants were retired, examples of their professions were baker, bank teller, farmer, and hunter. The time for which they

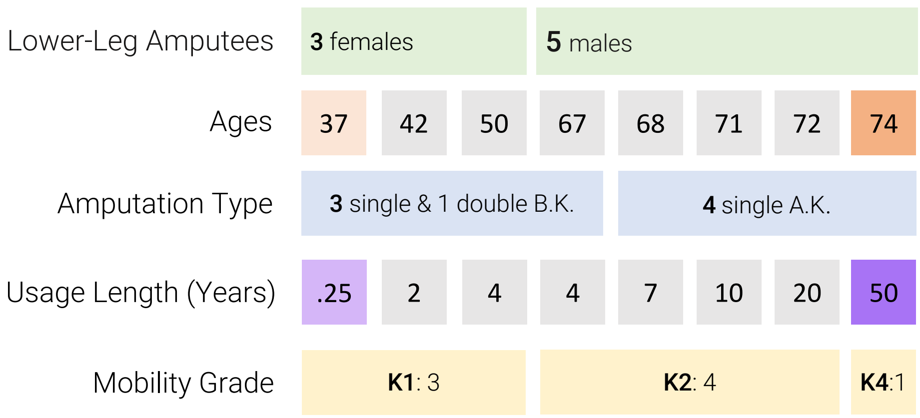


Figure 5.1: The pre-study participants were composed of people of different genders who varied greatly in age. They had different amputation types, degrees of experience using prosthetic legs, and levels of mobility.

used a prosthetic leg ranged from three months to fifty years ($M = 12.16$ years, $SD = 15.42$). An overview of the pre-study participant demographics is provided in Figure 5.1.

5.3 Pre-Study Results

The results from the questionnaire were collected and compiled. These results, explained in detail below, cover participant routines, activities, living circumstances and preferences regarding sensing on lower-limb prosthetics.

5.3.1 Participant Opinions on Sensory Feedback for the Foot

Having seen the prototypes of a sensing sock prior to completing the questionnaire, participants were asked to rate (1 = strongly disagree, 4 = strongly agree) the degree to which they believe that they would be able to do their activities more easily if their prosthetic foot (or feet) could detect when it is touching something. Five out of eight participants strongly agreed (62.5%), one agreed (12.5%), and two had no opinion (25%). The feedback as summarized in Figure 5.2 was very encouraging, as it showed that our potential users believe sensing technology, applied to their prostheses, could help them in better performing their activities.

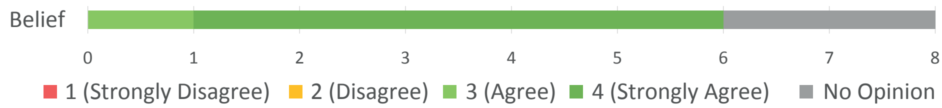


Figure 5.2: The degree to which the participants *believed* that having sensing on the foot would assist them in performing their physical activities.

5.3.2 Participant Socks and Footwear

Asked about their current use of socks, seven of the eight participants reported wearing socks over their prosthetic foot. Three reported changing their socks on a daily basis, two on a weekly basis, and one reported wearing socks only when needed (one sock-wearer did not answer this question).

Participants were also asked to indicate the types of shoes they wear. They responded with a spectrum of different shoe types. Running shoes were the most popular, followed by specialized shoes for prosthetics, sandals, hiking shoes, sneakers and dress shoes. Only one participant reported wearing strappy-sandals and crocs. No one selected options such as flip-flops, boots, ballerina flats/loafers, or high-heels. These results pictured in Figure 5.3 suggest that participants favour footwear that is flat, and can be fixed to the prosthetic foot securely. Six of the eight participants (with four up to fifty years of experience using their prosthetic limb) reported wearing three or more different types of shoes. The other two participants with the least amount of experience using a prosthetic limb (three months and two years of experience respectively) reported wearing only running shoes.

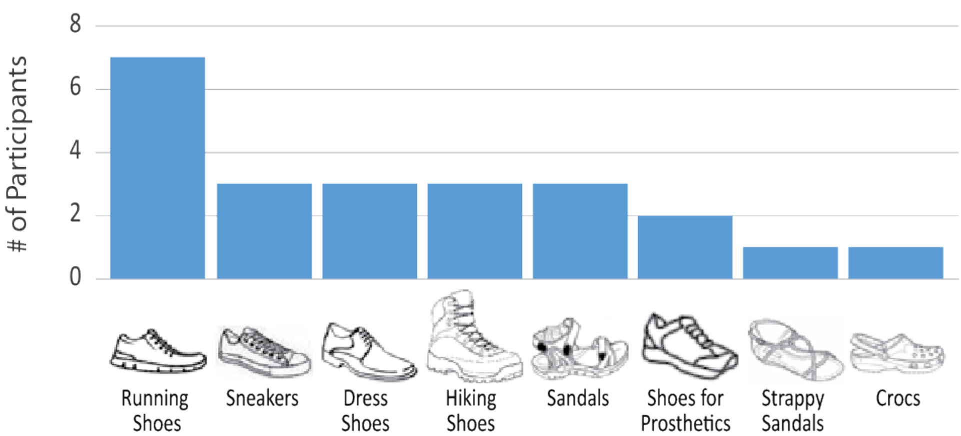


Figure 5.3: Shoes worn by the participants. Participants were asked to report in the questionnaire all the types of shoes that they wear.

The majority of the prosthesis users are accustomed to wearing socks on a regular basis, and primarily wear footwear that is compatible and designed to be worn with socks, meaning that a sensing layer in the form of a sock would be minimally disruptive to their normal routines.

5.3.3 Participant Activities

Lower-limb amputees are often assigned a mobility grade when they are fitted for a prosthetic leg. While slightly different systems exist, they generally contain grades ranging from 0-4. Grade 0 implies a patient does not have the ability to transfer or ambulate safely with or without assistance, and a prosthesis does not enhance their quality of life. Grade 1 implies the patient has the potential to use a prosthesis for transfers or ambulation on level surfaces. Grade 2 implies a patient has the potential to overcome small obstacles such as curbs. Grade 3 patients can move over wild terrain so long as not too much stress is put on the leg. Grade 4 patients would place high impact or stress on the leg, with distance and time capabilities similar to healthy individuals [31].

The mobility grades amongst the participants varied. Three participants identified as Grade 1, four identified as Grade 2, and one identified as Grade 4. This distribution is pictured in Figure 5.4. Participants self-reported partaking in a diverse range of physical activities. These included sports-related activities such as hiking, biking, wheelchair basketball, Qigong and Bavarian curling. Other self-reported activities included non-sports related examples such as walking, climbing up and down stairs, and shopping. Household chores including ironing, gardening, and even farm work (milking cows) were also mentioned by the participants.

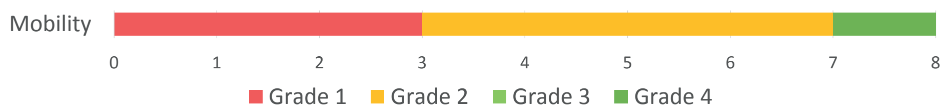


Figure 5.4: The mobility grades of the participants.

5.3.4 Participant Confidence Levels

Each person was also asked to report their confidence (1 = very insecure, 5 = very confident) in performing different activities (i.e. stair-climbing, ladder-climbing, car driving, bike riding). Six participants felt 'okay' or better with stair-climbing ($Mdn = 3$). Their answers are summarized in Figure 5.5. In contrast, seven participants used ladders ($Mdn = 2$), and four reported feeling insecure on ladders. Six of the participants could drive, all

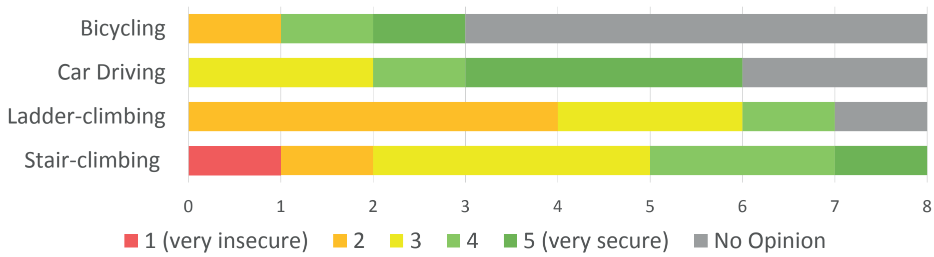


Figure 5.5: The degree to which participants feel secure while using bicycles and cars, and while climbing ladders and stairs.

of whom felt 'okay' or better ($Mdn = 4.5$). Two needed a left-foot throttle modification since they had right-leg amputations and all drivers owned automatic vehicles. We noted however that they occasionally drive the manual cars of their friends or family. Only three participants reported on bike-riding, each with a different level of confidence.

The participants were also asked to report their level of confidence (1 = very insecure, 5 = very confident) traversing different types of surfaces. Their answers are summarized in Figure 5.6. In general, participants felt 'confident' on firm, textured surfaces such as asphalt/concrete and carpet ($Mdn = 4$), 'okay' with gravel, grass, and hard flooring ($Mdn = 3$), 'insecure' on sand ($Mdn = 2$), and 'very insecure' on ice ($Mdn = 1$). One participant also reported feeling 'insecure' descending slopes. In general, they felt confident on firm textured surfaces such as asphalt/concrete and carpet, but insecure on sand and on ice.

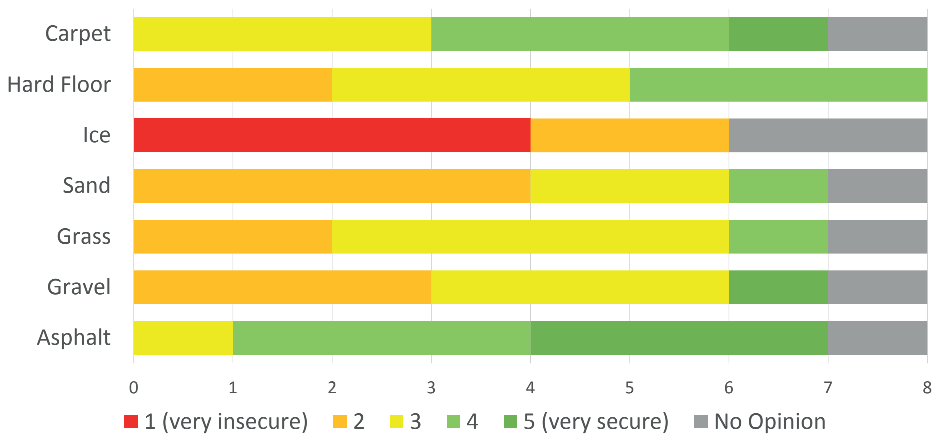


Figure 5.6: The degree to which participants feel secure traversing different surfaces. Participants felt most secure on asphalt.

5.3.5 Participant Importance of Sensing Regions

Participants were then asked to perform a colouring activity, where they shaded parts of the foot using different colours depending on how important it would be for them to sense in those regions. Green was used to signify 'important' whereas red was used to signify 'very important'. The shape of the foot was presented in three views (the sole, and two complementary three-quarter perspectives) for them to colour. Seven of the eight participants performed the activity. Figure 5.7 shows their individual responses (left) and a compilation of all the coloured responses (right). No two people provided the same response for the colouring activity; their responses illustrate that each participant had a different mental concept of what regions on the foot should have sensing. However, participants generally considered sensing on the sole of foot by the toes and by the heel as important.

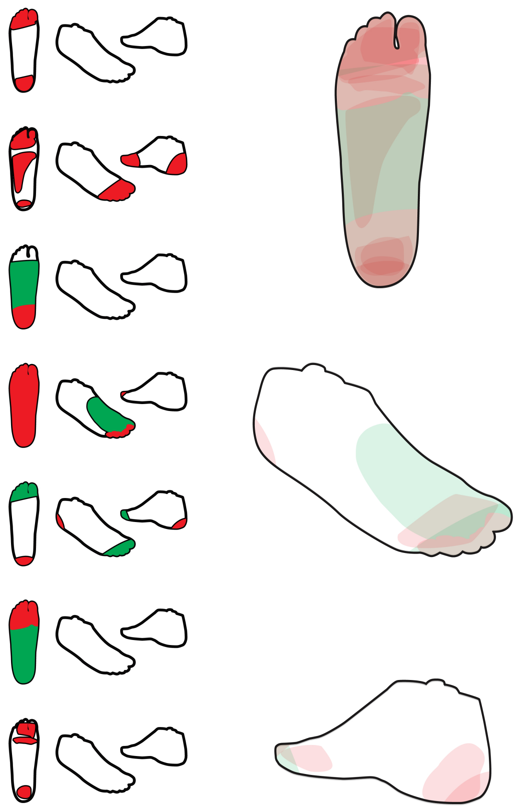


Figure 5.7: Seven colouring responses from participants expressing different sensing needs (left). Compiled responses (right) – left feet and right feet responses are superimposed over one another. Regions with high saturation were coloured by more participants. Green was used to signify 'important' whereas red was used to signify 'very important.'

Chapter 6

Discussion: Desired Sensing on the Foot

In this chapter, we analyze the data from the pre-study and present the resulting findings. Interestingly, answers received from the questionnaires and also in face-to-face discussions following the completion of the questionnaires illustrated that sensing needs can vary from person to person. Upon closer examination, their responses are shown to differ according to their amputation types and choice of physical activities.

6.1 Impact of Amputation Type on Sensing Needs

The range of responses differed between above and below knee amputees as seen in Figure 6.1. While above-knee amputees put precedence on the sole and heel, below-knee amputees had a wider range of desired sensing regions. Figure 6.2 summarizes which regions were marked as ‘very important’. The four above knee amputees mainly thought that the region below the toes (Sole-Toes) and under the heel (Sole-Heel) were very important. The three below-knee amputees had more varied responses; they also identified the whole bottom of the foot (Sole-*), the Front-Edge, and the top of the toes (Toes-Top) to be very important.

Our discussions with participants also revealed that the amputation type (either above-knee or below-knee) can highly influence a person’s sensing needs on the foot; a few participants in particular brought up a number of different, specific issues. One above-knee amputee with Grade 2 mobility stated “I want to know if I stand on my heel, and if the knee is locked securely.” He explained that his leg could only fully support his weight when fully extended. When bent, the knee would simply hinge under his weight, which could cause him serious injury if he accidentally puts pressure on the

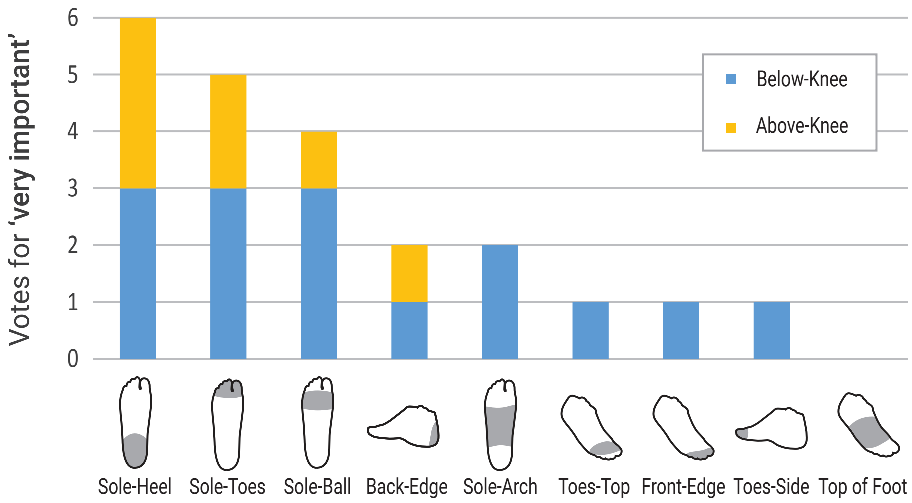


Figure 6.1: Sensing regions marked as 'very important' by study participants. Above-knee amputees desired a narrower range of sensory regions on the foot than below-knee amputees.

leg. At present, he regularly visually inspects his leg. However, the presented textile sensor on the heel could help him identify more easily whether his leg is fully extended and improve his sense of safety and security when ambulating. One woman with a below-knee amputation with Grade 2 mobility explained that she would like sensing along the front of the toes, stating "If I could feel if my forefoot [is caught on something], it would reduce the danger of tripping." As her current leg does not have any sensory capabilities, she cannot feel if her prosthetic foot catches on low-lying obstacles; feeling along the front of the toes could improve her safety by allowing her to preemptively correct the positioning of her foot in such situations.

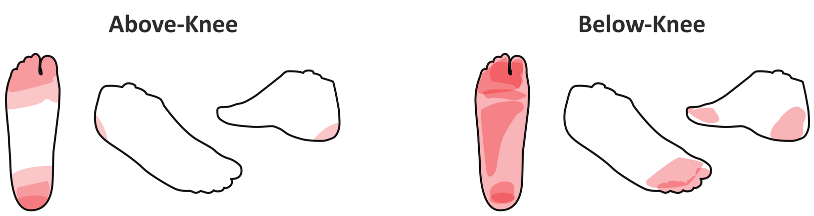


Figure 6.2: Compilation of sensing regions marked as 'very important' for above-knee and below-knee amputees.

6.2 Impact of Activities on Sensing Needs

Participants expressed desire for sensing on certain regions of the foot based on their different scenarios and activities. Figure 6.3 shows how the priority given to different sensing regions on the foot differs between activities.

Concerning walking, one participant recognized that her sensing needs were motivated by the types of walking surfaces she encounters. "Walking on hard-floor is very slippery. I think the area [in the middle] is of additional value for the sense of balance and better stability when walking on different surfaces such as a wet street, or when climbing stairs, etc," and added "When walking on gravel I could feel the pits better." The diversity in terrain due to their differences in texture and levelness contributed to participants' wishes to have sensing that was more widespread around the foot, that included Sole-Toes, Sole-Heel, Sole-Arch, Front-Edge, Toes-Side, and Back-Edge.

In contrast, participants expressed that while biking, the important sensing regions are more isolated to the Sole-Ball and Sole-Toes regions of the foot. This is easy to comprehend, as the ball ideally remains in firm, constant contact with the pedal while riding for maximal feeling of control.

Interestingly, the topic of crouching was also a scenario of concern for the participants. They would assume this position for instance, when gardening. As one participant explained, "When bending down, the stability would be better... when crouching, the toes are up in the air a bit and the point of gravity is on the heel." Assuming, maintaining, exiting this position requires shifts in one's center of gravity. As such, participants felt that the ability to feel the degree to which their weight is distributed towards the front versus towards the back would help them to maintain their balance, and gave emphasis to Sole-Toes and Sole-Heel.

Driving was considered by two participants (P1, P4) as an activity during which sensing would be very helpful. One above-knee amputee with Grade 3

Figure 6.3: Importance of different regions on the foot for sensing with regards to different physical activities. Left to right: walking, biking, crouching. Darker regions signify higher agreement. This graphic is derived from aggregating quotes gathered from different lower-limb amputees.

mobility explained that at times he is not aware if his prosthetic foot is in contact with pedals in the car. He explained that there was one incident where he did not realize his foot was against the gas pedal, pressed it down, and accelerated which resulted in a rear-end collision. The ability to better feel if his foot is against the pedal could help him to have better control over his car. Another lady with a below-knee amputation with Grade 2 mobility stated "When driving, I could react better with the clutch."

6.3 Summary: Sensing Socks for Lower-Limb Prosthetics

In summary, the results from the questionnaire provide the following take-aways regarding a pressure-sensing layer for lower-limb prosthetics:

- Users generally have a positive outlook on having sensory feedback for their prosthetic legs, and believe such technology can improve their performance in their activities.
- A sock form factor for the sensing layer is likely to be minimally disruptive to prosthetic-user routines, which can help with user acceptance and adoption. The majority of users wear socks regularly, and the majority of the footwear worn by users can be worn with socks, making this form factor more versatile than a sensing insole or shoe.
- Customization is valuable in sensing. It is beneficial for sock sensing regions to be variable in shape, size, location, and number to account for different user preferences.
- Activities have a large influence on which sensing regions on the foot are important. However, the sole of the foot, particularly by the heel and by the toes are generally important to prosthesis users. Most concerns relate to maintaining one's balance while standing and walking.

Chapter 7

Further Developments

While the original sock could fit a foot with shoe size 8.5 US (39 EU) and contained 192 sensor intersections (16 rows $\times$ 12 columns), we created another *proCover* sock to fit the foot sizes of all our participants. This sock could fit feet of shoe size 12 US (46 EU) or less, and contained 221 sensors (17 rows $\times$ 13 columns). Furthermore, in response to the insights gained from the pre-study, we developed and integrated new features into our sensing system to address the specific issues raised by the participants. We describe the details of our new developments below.

7.1 User-Configuration Tool: Mapping Sensing Regions to Haptic Feedback Stimuli

Adding to the original sensing sock and the vibration armband (described in Chapter 4), we developed a mobile app to allow for easy and dynamic creation and mapping of sensing regions on the foot to actuators. The system facilitates the dynamic creation of one-to-one, one-to-many, many-to-one, many-to-many type mappings.

This was in response to two observations from the pre-study. Firstly, we learned that participants have many varied opinions regarding desirable sensing locations on the foot and their relative degree of priority (see section 5.3.5). Secondly, we observed that participants typically engage in multiple different activities that demand different sensing capabilities (see section 6.2). This inspired us to develop a configuration tool. We considered that a single haptic sensory feedback system would unlikely equally meet the needs of each person, and rather, a dynamic system would be better able to handle these variations in needs and preferences amongst prosthesis users.

7.1.1 Interaction Design

Making configuration simple and quick to perform was a primary goal in the design of this tool. This is because many prosthesis-users may not have much prior knowledge or experience with wearable technology. Furthermore, the process of donning and doffing a prosthetic leg is already very complex and time consuming. Thus, if the sensing layer would greatly add to the degree of complexity in using a prosthetic leg, there would be a high risk that some users would find it too cumbersome to use in addition to their regular prosthetic leg and would choose to not use it at all.

The configuration tool allows users to create different mappings. We define a mapping as the relationship between one or more sensing intersections in the sock with one or more motors in the vibration armband. To create a mapping, a user follows a two-step process. First, while wearing the sock, the user can press 'Record' and either touch the foot or step on the region where they want a new sensing region to be located. Second, the user can assign which motors will vibrate when the sensing region they had selected is touched. For ease of use, users can also immediately test and fine-tune the selection as soon as a motor or multiple motors are selected by applying pressure to the selected region. This process is shown in Figure 7.1.

The system is also capable of storing multiple mappings, which can be enabled, disabled, or deleted. Multiple mappings in the system may be simultaneously enabled and active at any one time. This is to address the fact that prosthesis-users generally desire multiple sensing regions to be available at any one time. For example, a user may want to have the top of the toes, the ball of the foot, and the heel as three different sensing regions, mapped to three different motors on the armband before going on a hike on uneven

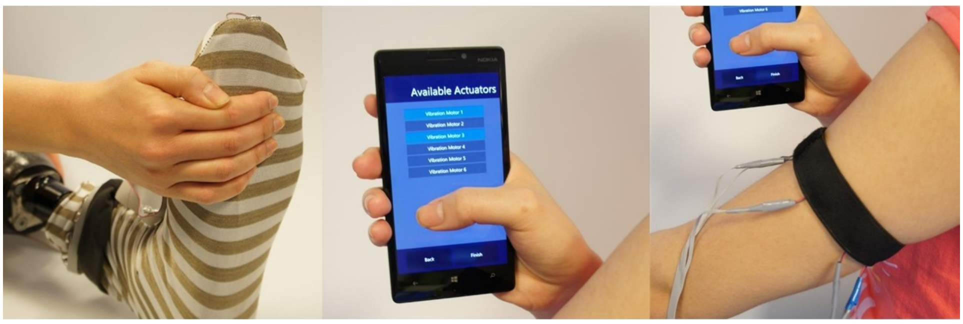


Figure 7.1: The configuration tool facilitates a two-step configuration process. First, users apply pressure to parts of the foot to select intersections for inclusion into a new sensing region. Second, users select which motors on the haptic feedback band will vibrate when pressure is applied to the newly created sensing region. Users can immediately test and readjust which motors are mapped to the region during the motor-assignment step before saving.

terrain. In future, it would also be beneficial to implement a feature for the system to store multiple sets of mappings. This way, users could then sort mappings into groups by activity-type and switch between sets of mappings in the system as *sensing modes*. For example, a user could then create and quickly switch between a walking, biking and driving-mode wherein their sensing capabilities can be made to match their needs during each activity.

7.1.2 System Infrastructure

The configuration tool prototype had a four-part infrastructure pictured in Figure 7.2. A mobile phone was used to collect and log user preferences. The original *proCover* sensing sock was used for both the selection of sensing intersections, and the detection of pressure within the created sensing regions. The original haptic feedback armband was used to communicate pressure information to the user. Information from both the phone and the sock were sent to a server. The phone sent information to the server over a wifi connection to trigger different modes in the server. The sock sent information to the server via serial connection, that was then used to drive the vibration intensity of the motors in the haptic feedback band.

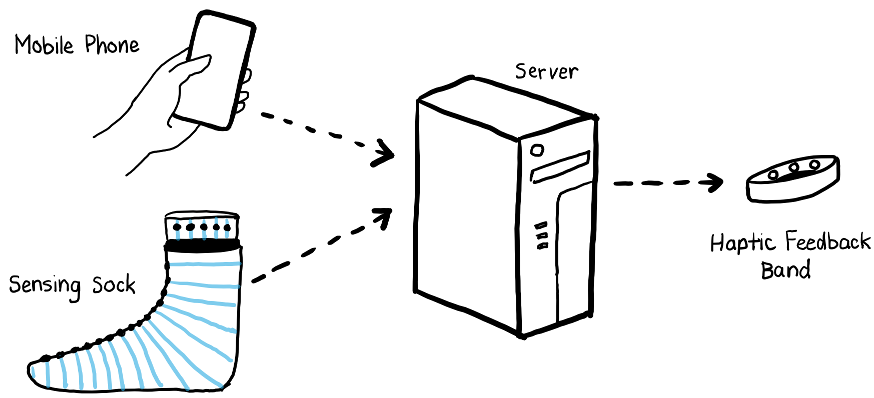


Figure 7.2: The system infrastructure for the customization tool includes a phone, server, the sensing sock, and the vibration armband. The phone communicates information to the server over a wifi connection; the sock and haptic feedback band communicates with the server over a serial connection.

7.1.3 Pressure Measurement Extraction Strategies

In addition to the main mapping functionality of the customization tool, an ability to adjust a sensor sensitivity threshold was also included. This threshold determined the amount of pressure that would need to be exerted on a sensor in order for it to be selected for inclusion into a sensing region. This addition was motivated by the results of initial tests done with the sock

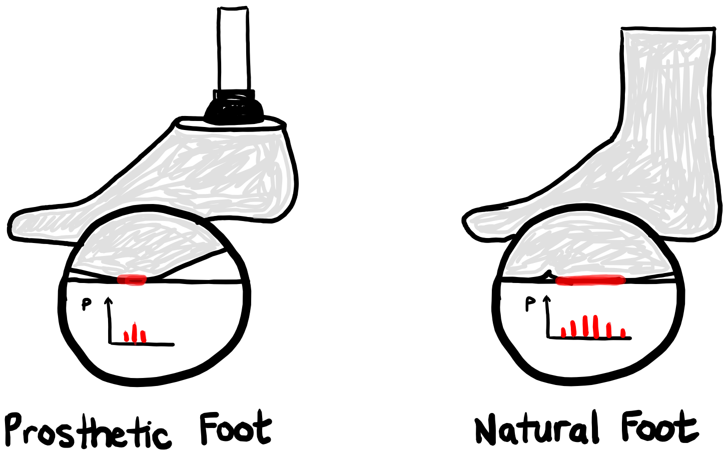


Figure 7.3: The greater stiffness of a prosthetic foot due to its material properties means that applied pressure is distributed over a smaller area than when a natural foot wears the sock. Therefore fewer sensing intersections would generally be triggered when the prosthetic foot would come in contact with an external object or surface.

over a prosthetic foot, which revealed that the pressure signatures from a prosthetic foot were different from those of a real foot. We observed that the greater stiffness of a prosthetic foot meant that pressure would distribute less over the contact surface of the foot. Figure 7.3 illustrates this phenomenon.

Three different strategies were explored for obtaining more accurate and representative pressure measurements for different sensing regions of the sock. The calculations and results of each strategy are discussed below.

Strategy 1: Maximum in Set

The first and simplest strategy was to use the maximum pressure value for all points in the set of selected sensing intersections as the overall pressure measurement for the sensing region. Let X be the set of all selected sensing intersections x containing a total of n intersections. Then the pressure measurement p for the sensing region is calculated as

$$p = \max\{x_1, x_2, ... x_n\}, \quad \text{where } x_i \in X. \tag{7.1}$$

This was the strategy used for the initial *proCover* sock prototype. While this method was effective when the sock was worn over a natural foot, it was not as effective when the sock was worn over a prosthetic foot. This is due to the fact that the peak amount of applied pressure could occur in the gaps between pressure sensing intersections. In the case where pressure would be applied in between sensing intersections, only small pressure signals would typically be detected around the point of contact for a prosthetic foot. Consequently the detected signals would not accurately represent the true amount of force that may be applied to the surface of the prosthesis. This strategy was also problematic in the cases where some sensors would malfunction or become shorted. In this case, the maximum pressure value would be read from the malfunctioning sensing intersection and would create a false reading reporting a very high amount of pressure on the corresponding sensing region of the sock.

Strategy 2: Average

The second strategy was to use the average pressure value for all points in the set of selected sensing intersections as the overall pressure measurement for the sensing region. This was calculated as

$$p = \frac{1}{n}\sum_{i=1}^{n} x_i. \tag{7.2}$$

This strategy has two benefits over Strategy 1. Firstly, it can help to mitigate the negative impact of noise or malfunctioning sensing intersections. Secondly, it can account for the fact that more than one pressure sensing intersection can be activated when the foot comes in contact with an external object or surface. This is particularly true in cases where the foot comes in contact with a surface, such as floor. However, this strategy is ineffective and was discarded since the average pressure values across all sensing intersections in a sensing region can be unrepresentative in the scenario where a point-pressure is applied (e.g. if a finger pushes against the sensing region).

Strategy 3: Upper Quartile Average

The third strategy was to use the average of all pressure values above the third Quartile (Q3), or the top 25% of the pressure values from the set of selected sensing intersections in the sensing region. Let M be the set of all pressure values m in the region sorted into ascending order such that $M = \{m_1, m_2, ...m_b\}$. Then b is the position index of the highest value, and a is the position index of the upper quartile. Then the pressure value can be calculated by averaging all pressure values above the upper quartile, i.e.,

$$p = \frac{1}{b-a}\sum_{i=a}^{b} m_i, \quad \text{where } a = \left\lfloor \frac{3(n+1)}{4} \right\rfloor. \tag{7.3}$$

This third strategy was selected as the best method for determining the pressure value for a custom sensing region of the sock. It is more robust to noise (generated from malfunctioning or shorted sensing intersections) than Strategy 1 since the reading is not dependent on a single sensing intersection. As such, a sudden peak in one intersection would not greatly impact the overall region's pressure value particularly when the foot is in contact with a large surface. At the same time, this method works better than the averaging strategy under point-pressures since it takes an average of only a quarter of the samples with the highest values. In this way, it minimizes the diminishing effect averaging can have, particularly when pressure is only applied to a small fraction of the whole sensing region.

7.2 Sensing Knee Guard: Detecting Bend States and Joint Position

We created sensing knee guards to determine the bend state of a leg using the same textile approach used for the *proCover* sock. We were inspired by the primary concern of one pre-study participant (described in Section 6.1), who expressed that he would not only want touch and pressure sensing, but also "position-sense" (often referred to as *proprioception*). We saw the potential in using the textile to not only detect pressure but also bending, and envisioned that either a knee guard or a longer stocking could be worn over the prosthetic joint to provide for this form of sensing. As such, we created several prototypes in order to realize this vision. We provide a description of each prototype below, and then explain the technical implementation for the corresponding hardware and software components.

7.2.1 Design Iterations

An iterative design approach was taken to create a sensing knee guard. Ultimately, three different physical prototypes, pictured in Figure 7.4, were constructed. Each prototype comprised a different set of features, and are described below. We note that for each prototype, pressure information was read-in from the garment (using the measurement electronics described in Section 4.2.2), and used to provide the user with visual feedback indicating the degree of bending in the corresponding knee-joint.

Prototype 1 – Full-Sleeve Design

The first sensing knee guard prototype that we created (see Figure 7.4, left) contains 100 sensing intersections (10 rows $\times$ 10 columns). It has a full-sleeve design, meaning it was sewn to encircle a limb fully. Its dimensions meant it could only fit around an elbow-joint. However, this physical prototype was sufficient to aid our early investigation into using input signals from pressure sensing textiles to extract the degree of bending in a joint.

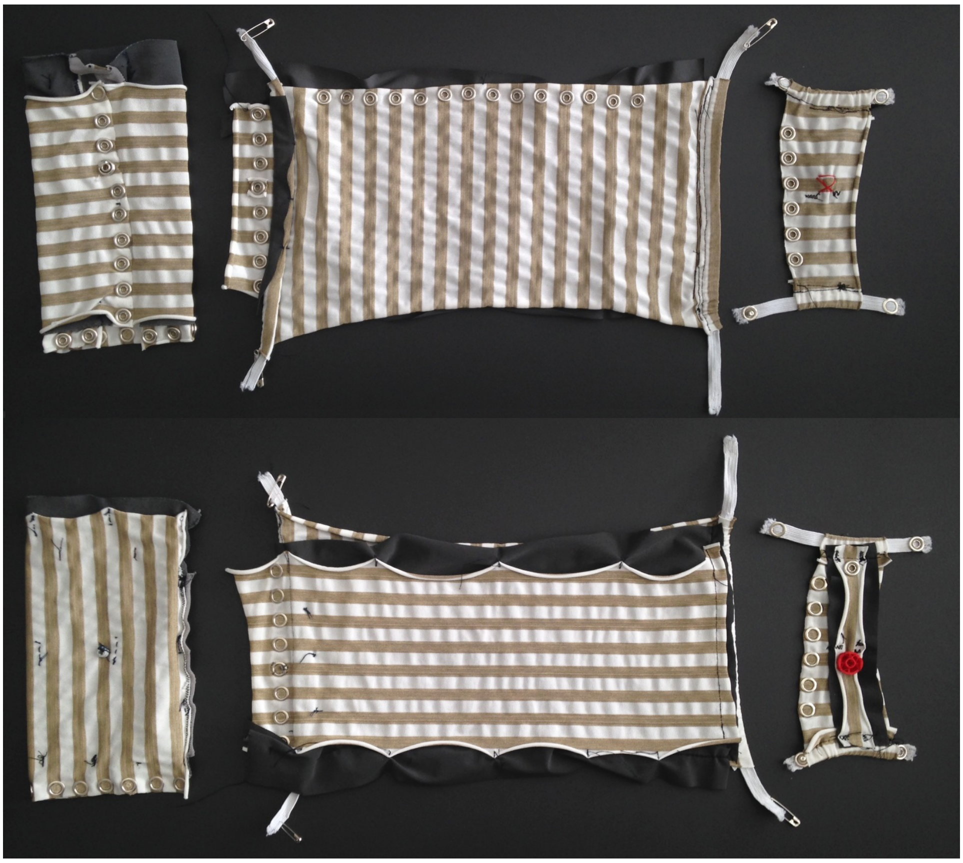


Figure 7.4: Sensing knee guard prototypes: Prototype 1 (left), Prototype 2 (middle) and Prototype 3 (right). Both sides of the prototypes are shown, with the outer sides exposed (top), and the inner sides exposed (bottom).

Prototype 2 – Broad Half-Sleeve Design

We expanded upon the first version of the sensing knee guard and built a second prototype (pictured in Figure 7.4, middle). In contrast to the first prototype, it has half-sleeve design, which allows it to be tied over a knee-joint. It is also slightly larger in its dimensions and contains 112 sensing intersections (14 rows × 8 columns). These design changes were driven by a desire to improve classification accuracy, to accommodate different sizes of prosthetic legs, and to make it easier to put on.

Prototype 3 – Mini Half-Sleeve Design

We created a third and final prototype of the sensing knee guard (pictured in Figure 7.4, right). Similar to the second prototype, it has a half-sleeve design. In contrast to the previous two designs however, it is much smaller in

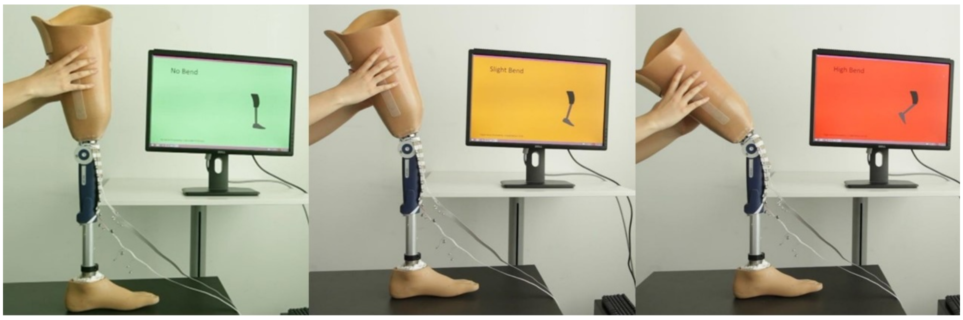


Figure 7.5: The final prototype of the sensing knee guard. Mounted on a prosthesis, it can provide an approximation of the amount of bending in the leg in degrees. A visual representation of the amount of bending registered by the system is displayed using an on-screen graphic of a prosthetic leg.

its dimensions and comprises six sensing intersections (6 rows $\times$ 1 column). Furthermore, it features the addition of an integrated plastic knee-cap and is designed to be mounted directly on a prosthesis instead of over pants (see Figure 7.5). These design changes were again motivated by the desire to extract bend information with higher precision.

7.2.2 Hardware Design Features

Different hardware features were developed in the prototyping process to improve the classification accuracy of the knee guard. We discuss the key hardware considerations below.

Half-Sleeve Versus Full-Sleeve Form Factor

The half-sleeve design used in the second and third knee guard prototype offered several benefits over the full-sleeve form-factor of the first prototype. Firstly, the half-sleeve design makes it easier to put on. Unlike the full-sleeve design that would need to be slipped on past the foot, the half-sleeve design can be strapped over the knee-joint directly. Secondly, the design makes it easily adjustable to accommodate different leg sizes; prosthetic legs often vary in diameter by the knee-joint. Thirdly, the half-sleeve design helps to ensure a tight fit over the knee-joint which is critical for classification accuracy. Using the straps, one can affix the knee-guard over the knee-joint such that it is held taut even in the no bend condition. Lastly, a full-sleeve design can cause lines to short on the concave side of the knee-joint, whereas a half-sleeve design eliminates this possibility, as seen in Figure 7.6.

Figure 7.6: The full-sleeve prototype in an unbent (left) and bent (right) state. This design presents a risk for conductive lines to come together and electrically short one another on the concave side of the joint, as can be seen in the bent state. A half-sleeve form factor eliminates this risk.

Utility of an Artificial Knee-Cap

Positioned at the apex of the convex side of a knee-joint, the integration of a plastic button helps to improve bend classification accuracy by amplifying the pressure signal at this point throughout the full range of motion. This hardware addition was integrated into the third prototype, simulating a knee-cap. Without this button, the range of pressure values that can be read from the knee guard is significantly smaller. A smaller range makes it more difficult to distinguish between the different bend-states. Note that the positioning of the knee-cap must be chosen based off the geometric profile of the leg to maximize its effectiveness. Examples of different knee-joint profiles can be seen in Figure 7.7.

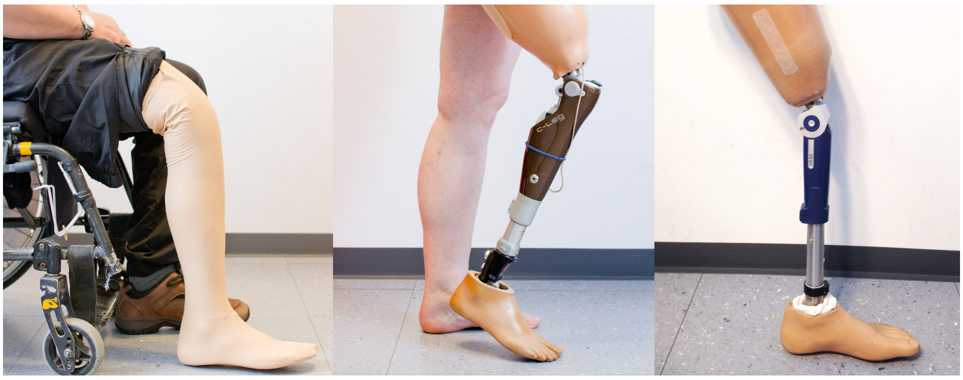


Figure 7.7: Careful consideration should be given to the placement of the artificial knee-cap in the sensing knee guard since prosthetic legs can have different geometric profiles.

Number of Sensing Intersections

Small dimensions rather than large dimensions for the knee guard also help to improve bend classification accuracy by minimizing sources for signal noise. As prosthetic knee-joints tend to be smaller than regular knees, we observed that the second prototype (the broad half-sleeve design) tended to have a loose fit, and that the loose-fitting sensors in the knee guard generated signals that did not correlate well with the bend state of the leg. Hence, their signals tended to interfere with rather than improve the classification accuracy. By keeping dimensions of the knee guard to a minimum in the third prototype, we reduced the amount of signal noise generated, and improved the knee guard's accuracy in determining bend-states.

7.2.3 Identifying Bend States

Three strategies were devised to determine the bending state of a leg using the sensing knee guards. The first two strategies were used to classify the leg as being in one of three states (i.e. no bend, slight bend, or high bend). The third strategy was developed to approximate the amount of bending in the leg in degrees. The implementation details for each approach are described in the following subsections.

Strategy 1: Protrusion Approach

The first strategy we devised was a protrusion-based approach, which we used in combination with the first prototype. This approach determines the degree of bending in the joint based off of the sensor subjected to the most amount of pressure when the joint is bent. This is located on the convex side of the leg at the apex of the knee-joint. A calibration phase is necessary to establish the thresholds needed to distinguish between the bend-states. To calibrate, the user must first capture the pressure readings both in a no bend state and in a maximum bend state. A low threshold is established based on the pressure reading in the no bend state, and a mid threshold is calculated as the midpoint between the two extremes. The system can then determine the bend-state by comparing the pressure value of the sensor to the thresholds.

It is critical for the knee-guard to be affixed securely in order to achieve a high degree of accuracy using this method. Physical shifting or translation of the knee guard over the joint changes the range of values that are detected at the key sensing intersection. Hence, in situations where this occurs, the system incorrectly classifies the bend-state (e.g. system identifies a slight bend instead of a high bend).

Strategy 2: Machine-Learning Approach

The second strategy we devised was a Support Vector Machine (SVM) approach, which we used in combination with the second prototype. This approach uses training data to create a classifier that can distinguish between different bend-states. For the prototype, we used the PyzoFlex FSR Matrix Visualization software [19] to capture unique SVM training data sets (i.e. 24 samples per condition) per person. We then input this data into Matlab and used the LIBSVM[1] library to find optimal parameters. In our bend detection application written in WPF/C#, we input these parameters to train a model for use within the application, which was leveraged for classifying new pressure patterns read-in from the knee guard while in use.

Ultimately, the SVM approach provided little benefit over the protrusion approach when the knee guard was applied to a prosthetic leg. In practice, we discovered that the geometric profile of a prosthetic leg creates a pressure signature that is characterized by only a few sensing intersections that show a significant change in pressure level. The remaining sensing intersections show little to no change in pressure readings and remain largely unchanged throughout the full range of motion. Thus, so long as the knee guard is affixed securely over the knee-joint, the bend information can be determined using simple thresholds rather than a machine learning approach.

Strategy 3: Improved Protrusion Approach

The third strategy we devised was an improved version of the first protrusion-based strategy. This was used in combination with the hardware setup of the third knee guard prototype (see section 7.2.1). When working with the first strategy, we saw that a single sensor with the addition of the knee-cap (described in section 7.2.2) had the potential to be used to give an approximation of the bend position. Therefore, in contrast to the previous two strategies that classified bend-states, this strategy attempts to extract an approximation of the amount of bending in a leg in degrees.

For this prototypical implementation, we assumed linear sensor behaviour (i.e. a positive linear relationship between the pressure signal read from the knee-cap sensor to the degree of bending in the leg) in order to approximate the degree of bending in the knee-joint. Let us define the range of bending in the leg to be between 0 and β degrees. Pressure measurements from the sensor range from 0 to 255. We define γ to be an offset in degrees (for background subtraction of default pressure readings) and define p as the pressure level that is read-in from the sensing intersection. Then the degree of bending in the leg, α, is calculated as

$$\alpha = \beta * \frac{p - \gamma}{255 - \gamma}. \tag{7.4}$$

[1] https://www.csie.ntu.edu.tw/c̆jlin/libsvm/

While this approach appeared to work fairly well in our brief tests with the prototype, it has a number of obvious shortcomings. Firstly, the true behaviour of the sensor is not perfectly linear. Hence, this strategy can only be used to achieve an approximation of the amount of bending in the leg. Secondly, the sensor offers a limited range of pressure readings and may become saturated prior to the maximum bend being reached in the prosthetic joint. This could result in critical information loss beyond a certain degree of bending. Therefore, caution should be taken before applying this strategy to realistic situations.

Chapter 8

Pilot Study

After one month, four participants (2 females and 2 males of ages 37, 42, 50, 74) from the first study were invited to the lab to test our revised textile-based sensing solutions. Two people were above-knee amputees (P3, P4), one person had a below-knee amputation (P2), and one person had double below-knee amputations (P1). They had used prosthetics legs for a different number of years (7, 10, 20 and 50). Each participant was using a different type of prosthetic leg (see Figure 8.1).

The pilot study was conducted with the overall goal of assessing the validity of the textile-based sensing concept for the sensory augmentation of real-world prosthetics. Our aims for this study were two-fold. Firstly, we wanted to test our technical implementations of the sensing sock and knee guard by applying them to a range of real prostheses. Secondly, we wanted to collect early impressions of our work from potential users. The pilot-study included three different tasks which were inspired by participant anecdotes from the pre-study. The study took approximately two hours to complete with all four participants.

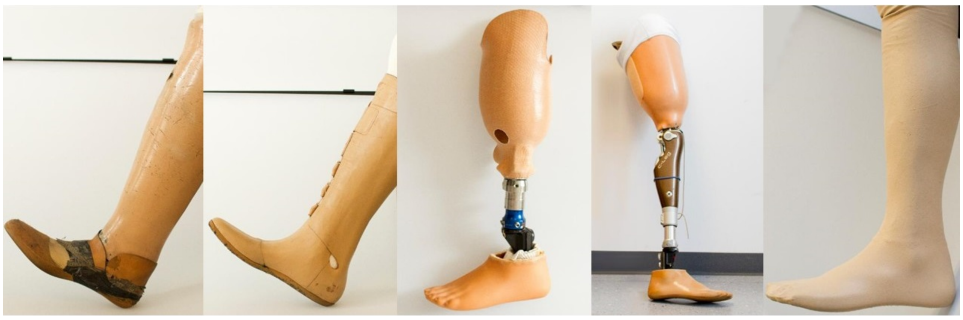


Figure 8.1: The participants' prosthetic legs.

8.1 Task 1 – Touch-Position Discrimination

The goal of this exercise was to assess the feasibility of dynamically creating distinct sensing regions on the foot, such that users can have the ability to determine which part of their prosthetic foot is being touched while wearing the sensing sock. This general feasibility assessment was necessary, considering the different form factors and materials of their prosthetic legs (see Figure 8.1). While geometric differences result in a different positioning of the sensor matrix on each of the prosthetic legs, different degrees of stiffness for the prosthetic materials results in different levels of pressure readings. Reliable dynamic mapping and adapative thresholds were needed respectively to handle these variations.

8.1.1 Task Procedure

We conducted a controlled experiment, where we pressed firmly with our hand either on the ball or the heel of their prosthetic foot, and asked the participants to report which part of their foot was being touched while having their eyes closed. This activity was performed under two conditions - with the sensing sock (as pictured in Figure 8.2), and without the sensing sock as we suspected that they might be able to feel something through their residual limb. In the condition with the sock, participants wore a vibration armband (either on the upper or lower arm), where the motor facing upwards was mapped to the ball of the foot, whereas the motor facing downwards was mapped to the heel of their foot. Each participant was assessed using the same sequence of touches, created by alternating randomly between the ball and heel of the foot. In total, each participant completed twelve trials (2 regions $\times$ 3 trials $\times$ 2 conditions). The number of correctly identified touches was logged and participants were asked to comment on the experience afterwards.

8.1.2 Results

All four participants performed this task. The dynamic mapping worked successfully, as all participants identified which region of the prosthetic foot was being pressed without errors when using the sensing sock. Conversely, on average, people produced an error rate of 75% ($SD = 0.083$) without the sensing sock. Without the sock, three participants (P2-P4) were observed to guess with minimal success which region was being touched. Despite the fact that P1 felt confident that he could correctly identify the touches based on the force he felt through his stump, he misidentified the touches five out of six times without the sensing sock, likely by misinterpreting the torque on his residual limb when his heel was not resting on firm ground. In addition to trying the system with her prosthetic leg raised, one participant (P2) also

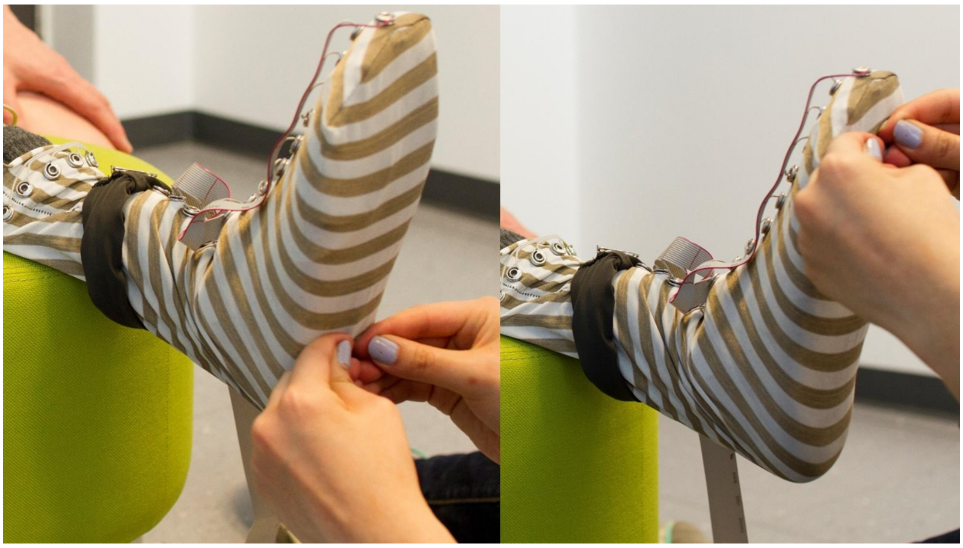


Figure 8.2: Each participant wore the sensing sock over his/her prosthetic leg, and each wore the vibration haptic feedback band snugly around his/her right arm. With their eyes closed, participants were asked to report whether the ball or heel of their prosthetic foot was being pressed. The number of correct answers were logged for each condition (with sock, without sock).

tested the set up in a standing position, where she shifted her weight from the ball of her foot to the heel of her foot. In this situation, she announced that she could now 'feel' how her foot contacted the ground through the vibration feedback on her arm.

All participants were asked to rate how challenging it was to use the sensing sock system (1 = Very Hard, 5 = Very Easy). All the participants rated the system as 5, or 'very easy' to use. Users were also asked to rate how easy it was to remember the mapping on the same scale. Unsurprisingly, all participants rated the mapping between sensor and actuators as 'very easy' to remember, as there were only two regions. However, P4 commented that he felt the task was more mentally demanding with the sock on, since he had to interpret the vibration feedback that corresponded with the pressing of different regions. When asked if they could imagine using this system in the future, we received encouraging responses. One below-knee amputee (P2) expressed she would like to use the system when walking (especially when walking on uneven terrain such as gravel), while an above-knee amputee (P4) stressed that he would like to use the system to feel his toes and heel while walking. P1 felt that his current legs gave him sufficient feedback through straps that led from his legs to a belt around his torso; however, he believed that people who are new to using a prosthetic limb would benefit from using this system.

8.1.3 Discussion

The results of Task 1 demonstrated the sensing sock in combination with the vibration armband provides a clear improvement over the sensory feedback that users otherwise rely on through their stump. Furthermore, the task confirmed that distinct sensing regions on the sock can be both created and mapped dynamically to haptic feedback actuators that users can quickly learn, memorize, and interpret. In fact, the mapping was so memorable that in the second task (described below), one participant (P2) exclaimed that she could feel her heel when she pressed the pedal, which she thought was a mistake. However, we simply mapped the region that was touching the pedal (in her case the ball of the foot) to a motor that happened to correspond with her heel in the first task.

8.2 Task 2 – Applying Pressure to a Car Pedal

The goal of this exercise was to assess the value of using the sensing sock to detect varying amounts of pressure created for instance when users apply force to pedals in a car. In turn, this information could be fed back to the user in the form of haptic feedback of variable intensity to provide an improved experience of operating the pedals of a car while driving.

8.2.1 Task Procedure

We conducted an experiment using Logitech G27 foot pedals. The pressure output of the sock was linearly mapped and scaled to the input voltage range of two vibration motors. Participants were instructed to depress the pedal to three different levels (shallow, medium and full) with their eyes closed under two conditions – with the sensing sock, and without the sensing sock. Left-leg amputees were asked to control the clutch pedal, while right-leg amputees were asked to control the gas pedal. The task was designed for participants to complete three trials per pressure level, for a total of nine trials per condition, and eighteen trials in total (3 pressure levels $\times$ 3 repetitions $\times$ 2 conditions). The number of errors (i.e. incorrectly performed presses) would be logged. At the conclusion of this exercise, participants were asked to comment on the experience under the two conditions, and whether they could imagine using the sensing system in the context of driving in the future.

8.2.2 Results

All four participants attempted this task. The results from this task were mixed. In general, all the participants found it easy to perceive the difference between no pressure and some pressure. However, they had more difficulty distinguishing between mid and high pressure levels. In general, the relative intensity of vibrotactile feedback was hard for them to assess.

Figure 8.3: Each participant wore the sensing sock over his/her prosthetic leg and wore the vibration haptic feedback band on his/her right arm. The haptic feedback band was adjusted for each participant to fit snugly around the arm. With their eyes closed, participants were instructed to depress a car pedal to one of three different levels (shallow, medium, full). The number of times they could correctly perform each press was logged.

Therefore, only small pressure changes on the sock were induced until the pedal was fully pressed. However, participants could clearly sense an increasing and decreasing stimuli when depressing and releasing the pedal. Furthermore, it is interesting to note that all the participants were observed to hover with their prosthetic foot rather than rest the heel of their foot on the ground while operating the pedals. This is likely because their ankles were inflexible such that they could not alter the angle of their foot. Participants could complete this task already very well without any sensory feedback, and had an average error rate of 11.1% ($SD = .079$). Therefore, the system led to a minimal improvement, lowering the average error rate to 8.3% ($SD = .048$).

In addition to testing the sensory feedback system with the pedals, we also invited participants to stand so they could experience feedback that would correspond to the pressure ranges reached when standing and shifting their weight. When asked about the concept of using pressure for the pedals, the responses were mixed. P2 expressed that she would want to feel the pressure so she could better operate the clutch. Two others expressed that they did not need such a system; P1 felt confident that he could apply the correct amount of pressure without the system, while P3 felt her good-leg was sufficient for the job. P4 expressed that he would see more value in being able to determine which pedal his leg was in contact with, rather than being able to feel the amount of pressure he was applying to a particular pedal.

8.2.3 Discussion

The results of Task 2 highlight that the range of applied force on the sensors generated from interacting with pedals is on a different level as the forces applied to the sensors when users stand. Therefore, we learned that the minimum and maximum pressure thresholds used by the system to determine the corresponding levels of feedback intensity should be adjustable. In this way, the system can provide feedback that corresponds well with the expected range of pressures that would be generated during different activities (e.g. driving versus walking).

Furthermore, based on our observations of their driving style, we learned that careful consideration is needed when creating sensing regions for a particular activity. We observed that people hovered their foot over the pedal, and displayed some inconsistency regarding which parts of their foot they used to depress the pedal. For example, they sometimes shifted their foot forward, pressing the pedal with the arch of their foot, while at other times they shifted their foot backward, pressing the pedal more with the ball of their foot. Therefore, it would be important that created sensing regions are made to account for such variations.

Lastly, the results for Task 2 are that some prosthesis-users are of the opinion that they do not need much additional pressure sensing support in the context of driving, while others would appreciate the additional feedback, especially when driving non-specially adapted cars. In addition, it would be interesting to explore other scenarios in which variable pressure and feedback would then be helpful.

8.3 Task 3 – Knee-Bend Detection

The goal of this task was to assess the feasibility of using the textile-based sensing solution to successfully detect the degree of bending in the prosthetic leg. Two participants with an artificial knee were asked to perform this task. For each participant, the bending sleeve (described in section 7.2.1) was affixed to their pants over the knee-joint.

8.3.1 Task Procedure

Training data was then captured for each participant for each of the three bending conditions (no bend, slight bend, and high bend). The training data was then used to train the system, after which we observed whether the system could successfully identify the state of their prosthetic leg for a total of nine trials (3 states $\times$ 3 trials). At the conclusion of this exercise, partici-

Figure 8.4: Two participants, each with an above-knee amputation, were asked to bend their prosthetic legs to three different degrees (no bend, slight bend, high bend) with the knee guard strapped over the knee-joint of their respective legs. The system was trained for each individual, and the number of times the trained system could correctly classify the true bend-state of the leg in test trials was logged.

pants were asked to comment on their experience as well as to comment on their desire for such a system in the context of their daily lives. Photos from this task are shown in Figure 8.4.

8.3.2 Results

The two above-knee amputees (P3, and P4) performed this task. While we envisioned a solution that would be fitted directly over the prosthetic leg, in the study, we refrained from asking participants to remove their clothing and instead fitted the sensing prototype over their clothing. In general, it was difficult to affix the sensor to their legs securely over their pants. As such, we observed that the sensor shifted while they moved, reducing the accuracy of the identification. Despite these conditions, the system classified six out of nine trials for the participant wearing very loose jogging pants (P4), and eight out of nine trials correctly for the participant wearing jeans (P3).

When asked about this sensing solution, participants stressed that bend detection is of most importance to improve safety from falling. For the participant with a simple hinged knee (P4), the detection of a slight bend in particular was of utmost importance, as he must ensure his leg is perfectly straight in order for his prosthetic to support his weight. Furthermore, both participants explained that preference would then be given to using this information to trigger automatic responses in the leg, rather than generating feedback that would be relayed to the wearer.

8.3.3 Discussion

Task 3 demonstrated the potential for our textile-based approach to be used to detect the bending of a knee joint. However, it is clear that it is crucial for the final sensing solution to fit very snugly around the prosthetic knee for optimal classification results. As we were not fully satisfied with the classification results for P4, we further improved upon the knee guard prototype by building an updated version that contains a slimmer sensor array and a button sewn into the fabric to simulate a kneecap (described in section 7.2.1). These changes stabilized the readings and allowed us to extract the angle of the joint with higher precision. Of course, while other types of bend sensors can be used, our results from this test demonstrate the potential for using our textile sensing approach to serve the dual purpose of augmenting the sensory capabilities of prosthetic feet as well as monitoring the bend-state of prosthetic legs.

Chapter 9

Limitations and Future Work

Working on *proCover* gave us a deeper understanding of the challenges pertaining to creating sensing wearables for prostheses. Our experiences from building multiple prototypes and also testing them with users showed both the promise and shortcomings of our solution. In this chapter, we outline the limitations of our technical implementations and study approach as areas for further improvement. We also point out possible future directions for research in this domain.

9.1 Technical Limitations

In the course of this thesis, we developed several prototypes of sensing socks and sensing knee guards. These prototypes were comprised of smart textile pressure sensors sewn into different form factors, electronic boards, and connectors used to forward information from the fabric to the measurement hardware. While our implementations served as successful proofs of concepts, there are a number of technical limitations to our approach that would need to be overcome before such sensing wearables can be used in practice by lower-limb amputees. In particular, we consider issues pertaining to the *robustness* and *portability* of the design.

Form-Factor In the pilot-study, we successfully applied our sensing sock to each of the different prostheses used by our participants. However, these tests showed that prosthetic ankles can present some difficulties in putting on the sock. While more modern prostheses tend to be thin around the ankles, one of our participants used a very old prosthesis that was very thick around the ankle. Consequently, particular effort was needed to pull the sock on fully past the ankle section of his leg. In addition, while some prostheses are designed with adjustable ankles, the ankles of our participants were immobile and incapable of plantar flexion (i.e. flexing the ankle to point the foot downwards). This can make it difficult to slide the sensing sock on, as it has to slide past the heel at a right angle. The considerable stretchiness of our sensing material helped make donning the sock easier; using sensing materials with an even greater degree of elasticity, similar to spandex, would help to improve future designs of the wearable.

Sensors In our short-term tests in the lab and also with participants in the pilot-study, no changes in the sensing behaviour of the textiles were observed. However, one factor we did not explore in this work is the possibility for the sensing behaviour of the textile to change as it is subjected to repeated or long-term physical stress. The material in the sensing knee guard prototype must operate under constant tension in order to produce clearly distinguishable pressure signals. Furthermore, our participants from the pilot-study informed us that their regular socks were prone to breaking down quickly when worn over their prosthetic feet. Hence, while we did not examine the durability of the sensors over the long-term, this would need to be done before these wearables could be relied upon on a regular basis.

Currently, we use a three-layer textile approach for pressure-sensing. Although we did not encounter problems with this in our tests, it presents a possibility for shearing to occur between layers. This possible shifting of the different layers could result in a misalignment of the sensing grid and misinterpretation of the pressure signals. Condensing the three-layers into a single sensing layer would eliminate this potential issue. Furthermore, it would help to simplify the fabrication process for the sensing garments.

Controllers In our prototypes, we relied on a server to process signals, and used separate printed circuit boards (PCBs) to capture input from the sensing garments and to drive feedback output. While our prototypical implementations were sufficient to conduct in-lab studies with participants, a number of changes would need to be done to make the system truly portable and wearable outside the lab. The server component should be removed such that all signal processing is done on a portable PCB, and the boards should be reduced in size and equipped to handle wireless transmissions in order to improve the portability of the system.

Electrical Connections In our work, we noticed that the soldered connections present a mechanical weak-point in our design. Under movement, these connections are put under tension which can cause them to snap. Strategies should be developed to either fortify or replace them before these wearables can be robust enough to withstand the physical stress of every-day use. One potential solution is to embed conductive lines directly into the fabric material which lead directly to measurement electronics mounted by the sock-opening. This way, shorter soft-hard connections can be used, which would experience less tension and be less prone to breaking.

9.2 Study Limitations

The pilot-study we conducted was brief in duration. Users only had a few minutes to adjust and familiarize themselves with the vibration feedback from the haptic armband. It is possible that users' task performance would have improved if they were given a longer period of time to become accustomed to the vibration feedback. It may also have been possible for users to attempt more sophisticated or complex tasks given a longer time frame. It would be interesting in future to observe whether users can learn and improve their ability to pin-point the location of vibration signals and better discern between different vibration intensities over time and in the context of different task scenarios.

9.3 Future Directions

There are a number of opportunities to expand upon the work done in this thesis. Future directions include exploring other feedback modalities and developing alternative interaction methods for the configuration of sensing regions as well as mappings. Beyond lower-limb prosthetics, we also see the potential to apply a similar approach for augmenting upper-limb prostheses with sensory capabilities.

9.3.1 Exploring Other Feedback Modalities

Sensory substitution systems represent a 'package-deal.' Their effectiveness in practice is determined by the quality of both the sensing and output; hence the importance of good feedback strategies cannot be understated. We worked with vibration motors as a first step towards exploring the potential of flexible mappings. However, there are still many possibilities for feedback modalities and mappings that amass to an even greater variety of options and flexibility for users. The use of more elaborate mapping functions as well as tactile phenomena such as sensory saltation [7], as was leveraged in Tactile Brush [11], may add even more value to such systems. Different feedback modalities (e.g. pressure feedback via pneumatic actuators, and auditory or visual cues) and possible combinations of them may have added benefits and should therefore be a subject of future work.

The possibility of triggering realistic sensations in the user using sensory wearables can also be further contemplated. A surgical procedure referred to as "targeted sensory reinnervation" can be done to remap the nerves that would originally lead to the missing limb to a patch of skin on another part of the body (e.g. the residual limb). Afterwards, touching this reinnervated patch of skin triggers the sensation of the original missing limb being touched for the patient. Thus, it is possible for a sensory wearable to trigger realistic sensations for the user if feedback actuators are positioned to interface with the reinnervated patch of skin. In fact, the tactic to leverage this surgical procedure in combination with a sensing prosthesis was devised and used by Egger [39]. This solution has the clear drawback however of requiring an invasive surgical procedure beforehand.

9.3.2 Improving the Interaction Design

In this work, we introduced a concept for the user-driven customization of sensory substitution wearables for prostheses. Our implementation of a configuration tool demonstrated one approach to user-customization of sensing regions and mappings to feedback actuators. In future work, other approaches can be taken in the interaction design of the customization tool. For prosthesis-users who are comfortable operating mobile devices, our application can offer a simple and efficient means for configuration. However, our experience with participants in the pilot-study revealed that many people are not familiar with using smartphones, and that it would also be important to create configuration methods that work without the use of smart devices in general. As an alternative, we considered incorporating simple button and LED interface modules on the sock and armband. By pressing a button on the sock, the user can toggle between a calibration mode and run-time mode for the sock. On the armband, buttons could also be integrated and used by people to select the motors that should vibrate in response to newly created sensing regions. Another possibility would be to implement the recognition of special grip gestures on the sock to trigger different modes rather than rely on hardware buttons.

9.3.3 Sensory Augmentation of Upper-Limb Prostheses

Moreover, we see potential in extending our approach to make sensing gloves for upper-limb prosthetics. The sense of touch is undoubtedly critical for controlling the movement of our hands. Hands perform a wide range of gestures and grips, and a sense of touch assists us in perceiving and understanding the amount of force and pressure we exert on different objects. A glove made using a similar textile approach could offer a possibility for upper-limb amputees to augment their prostheses with sensory capabilities. However, we suspect that the greater 3D complexity and possible range of movement for hands and finger digits will make the task more challenging. In particular, establishing robust soft-hard connections between the textile and the measurement electronics will be an obstacle towards making a portable and reliable sensory glove.

Chapter 10

Conclusion

In this work, we presented *proCover*, a novel prosthetic-sensing wearable that offers a non-invasive, self-applicable, cost-effective, and dynamically customizable approach for the sensory augmentation of lower-limb prostheses. We created several prototypes of sensing wearables, and investigated the design space for the concept in consultation with a diverse group of eight lower-limb amputees. Based on our insights from questionnaires and in-person discussions, we refined our original prototypes to incorporate new features. The introduction of novel customization capabilities made the system user-modifiable and capable of adapting to users' unique and dynamic sets of sensing needs, which change in accordance with their different physical activities. Techniques were also developed to leverage the use of the smart-textiles to detect bending in prosthetic joints. The validity of our wearable concept was confirmed in a pilot-study conducted with four lower-limb amputees. The sensing sock prototype was successfully used to dynamically create and map sensing regions on prosthetic feet to feedback actuators in an armband for each study participant. This enabled the prosthesis-users to distinguish between touches on different locations and to discern between different levels of pressure. Our sensing knee-guard prototype furthermore demonstrated the potential for our fabric approach to be used for bend-detection for prosthetic limbs in the same pilot study and in our own tests.

Ultimately, this work represents an early foray into the research space of prosthetic-wearables, and we hope that it will inspire similar research efforts moving forward. As the population of people needing a prosthesis continues to grow, it is important to consider how we can broaden access to these essential assistive technologies and enhance their user experience. Part of that can be accomplished through the creative innovation of technological alternatives, which can provide desired functionality to users at a reduced cost. We believe that smart-textile wearables can be part of this movement, and have worked to show one way they can be leveraged to improve the quality of life for those living with amputation.

Appendix A

Pre-Study Questionnaire

This six-page questionnaire was issued on March 10, 2016. Regular pens, as well as coloured pencils (red, green, and blue) were provided for participants to fill out the questionnaire. Eight lower-limb prosthesis-users completed the questionnaire over the course of approximately twenty minutes to an hour. A participant could choose to answer either question 16 or 17 depending on which side he or she wears a prosthesis. Double-amputees could choose to complete either question. The questionnaire is in German since the participants were native German-speakers.

Fragebogen: Beinprothese

Vielen Dank für Ihre Bereitschaft, unseren Fragebogen zu auszufüllen!
Wir würden gerne ein besseres Verständnis über Ihre Erfahrungen mit einer Beinprothese bekommen. Der Fragebogen dauert 10 bis 15 Minuten. Bei Fragen stehen wir natürlich gerne jederzeit zur Verfügung.

A. Allgemeine Informationen

1. Alter: ______________ 2. Beruf: ____

3. Geschlecht: ○ Männlich ○ Weiblich

4. Seit wie lange haben sie schon eine Beinprothese? ____

5. Welche Art von Amputation hatten Sie (z.B.: über/unter dem Knie, Teile des Fußes, ...)?

   ____

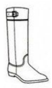
6. Schuhgröße (EU oder US)? ______________________

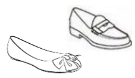
7. Tragen Sie Socken über dem Prothesenfuß? : ○ Ja ○ Nein

   Falls ja, wie oft wechseln Sie die Socken? _______________________________________

8. Welche Art Schuhe tragen Sie?

☐ 
Flip-Flops

☐ 
Flache Stiefel

☐ 
Ballerinas/Slippers

☐ 
Pumps

☐ 
Riemchen-Sandalen

☐ 
Stiefel mit Absatz

☐ 
Laufschuhe

☐ 
Riemchensandalen

☐ 
Sandalen

☐ 
Wanderschuhe

☐ 
Sneakers

☐ 
Schnürrer

☐ Andere: ____

B. Ihre Aktivitäten

9. Was ist Ihr Mobilitätsgrad?
Falls Sie sich nicht sicher finden Sie zusätzliche Informationen auf der letzten Seite.

K1 / Geringer Mobilitätsgrad	K2 / Mittlerer Mobilitätsgrad	K3 / Hoher Mobilitätsgrad	K4 / Besonders hoher Mobilitätsgrad
○	○	○	○

10. Wie sicher fühlen Sie sich beim ausüben folgender Tätigkeiten.
["nicht zutreffend" falls die Tätigkeit nicht möglich ist auszuführen]

Tätigkeiten	sehr unsicher 1	unsicher 2	okay 3	sicher 4	sehr sicher 5	nicht zutreffend
Stufen steigen	○	○	○	○	○	☐
Stehleiter benutzen	○	○	○	○	○	☐
Autofahren	○	○	○	○	○	☐
Fahrradfahren	○	○	○	○	○	☐

11. Falls Sie ein Fahrzeug fahren können (z.B. Auto, Motorrad, …):
   a. Welchen Typ von Fahrzeug fahren Sie: ____
   b. Benötigen Sie eine Änderungen am Fahrzeug, falls ja welche:

   ____

12. Wie sicher gehen Sie sich auf folgenden Oberflächen?
["keine Angabe" falls Sie keine Erfahrungen darüber haben]

Oberfläche	sehr unsicher 1	unsicher 2	okay 3	sicher 4	sehr sicher 5	keine Angabe
Asphalt / Beton	○	○	○	○	○	☐
Schotter / Kies	○	○	○	○	○	☐
Gras	○	○	○	○	○	☐
Sand	○	○	○	○	○	☐
Eis	○	○	○	○	○	☐
Parkett / Laminat / Fließen	○	○	○	○	○	☐
Teppich	○	○	○	○	○	☐
Sonstige						
1.	○	○	○	○	○	☐
2.	○	○	○	○	○	☐

13. Welche Art von körperlicher Tätigkeit üben Sie aus (z.B. mit dem Hund gehen, Einkaufen, Wandern, Laufen, Radfahren, Volleyball, ...) Bitte beschreiben Sie diese und geben Sie die Häufigkeit an.

Betätigung	<einmal im Monat	<einmal pro Woche	1-5 pro Woche	> 5 pro Woche
1.	○	○	○	○
2.	○	○	○	○
3.	○	○	○	○
4.	○	○	○	○
5.	○	○	○	○
6.	○	○	○	○
7.	○	○	○	○
8.	○	○	○	○

C. „fühlende" Beinprothese

14. Bitte Kreuzen Sie die Antwort an welche am besten zu Ihnen passt.

	trifft nicht zu	trifft eher nicht zu	trifft eher zu	trifft zu	keine Angabe
Ich könnte meine körperlichen Tätigkeiten einfacher erledigen wenn ich mit meiner Beinprothese Berührungen erkennen könnte.	○	○	○	○	○

15. Beschreiben Sie kurz Ihre Antwort in Frage 14.

16. **RECHTER FUß:** Skizzieren oder markieren Sie Abschnitte des Fußes, wo Ihrer Meinung ein Tastsinn hilfreich sein könnte. Verwenden Sie bitte die Farben aus der Legende rechts.

(rot)	**sehr wichtig**
(grün)	**wichtig**
(blau)	**nicht wichtig**

Fußsole

Seitenansicht

* Hinweis: Wenn Sie einen Fehler gemacht haben, notieren Sie bitte die Farbe, die Sie verwenden wollten und markieren Sie die Region mit einem Pfeil.

Haben Sie einen Bereich rot oder grün eingefärbt? Wenn ja, warum?

17. **LINKER FUß:** Skizzieren oder markieren Sie Abschnitte des Fußes, wo Ihrer Meinung ein Tastsinn hilfreich sein könnte. Verwenden Sie bitte die Farben aus der Legende rechts.

(rot)	**sehr wichtig**
(grün)	**wichtig**
(blau)	**nicht wichtig**

Fußsole

Seitenansicht

* Hinweis: Wenn Sie einen Fehler gemacht haben, notieren Sie bitte die Farbe, die Sie verwenden wollten und markieren Sie die Region mit einem Pfeil.

Haben Sie einen Bereich rot oder grün eingefärbt? Wenn ja, warum?

18. Haben Sie Erfahrungen mit Vibrationsfeedback?
(z.B. Vibration am Handy, Vibrationen an der Spielkonsole, ...) ○ Ja ○ Nein

Falls ja, an welchem Körperteil würden Sie sich eine Vibration/Feedback einer "fühlenden" Beinprothese erhoffen? Warum?

19. (Optional) Sonstige Anmerkungen:

Dies markiert das Ende des Fragebogens. Danke für Ihre Teilnahme!

Appendix B

Pilot Study Log Sheets

These forms were used to log information during the pilot-study conducted on April 6, 2016.

B.1 Task 1 – Log Sheets

Participant #: ________________

T1 : Configuration Tool

BASIC INFORMATION

Age: _________ Profession:____

Gender:	◯ Male	◯ Female
Have you been part of the first study :	◯ Yes	◯ No
Group:	◯ With Socks first	◯ No Socks first

How long have you been using a prosthetic leg(s)?

____

What type of amputation(s) did you have (e.g. above knee, below knee, partial foot, etc.)?

____

WITH SOCK/FEEDBACK

PUSH FOOT

Position	☺	☹
Heel		
Ball		
Ball		
Heel		
Ball		
Heel		

NO SOCK

PUSH FOOT

Position	☺	☹
Heel		
Ball		
Ball		
Heel		
Ball		
Heel		

Questions

How challenging was it for you to use the system?

	Not at all	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

Participant #: __________________

Mental Demand: How mentally demanding was the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Physical Demand: How physically demanding was the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Temporal Demand: How hurried or rushed was the pace of the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Performance: How successful were you in accomplishing what you were asked to do?

	Perfect									Failure
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Frustration: How insecure, discouraged, irritated, stressed, and annoyed were you?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

How can you imagine using this system in your daily life / as part of your routine?

Would you want different setups for different activities? YES/NO?

Participant #: __________________

How easy was it for you to map the region with the actuators?

	Not at all	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

How easy was it for you to remember the regions with the according actuators?

	Not at all	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

What regions would you initialize for walking?

What regions would you initialize for (another activity)?

Is it helpful to map sensor regions to motors of your choice?

What was problematic with the current setup? Further comments?

B.2 Task 2 – Log Sheets

Participant #: ________________

T2 : Driving Pedals

BASIC INFORMATION

Group: ○ With Socks first ○ No Socks first

WITH SOCK/FEEDBACK

Position	☺	☹
Light		
Middle		
Strong		
Strong		
Light		
Middle		
Middle		
Strong		
Light		

NO SOCK

Position	☺	☹
Light		
Strong		
Middle		
Middle		
Strong		
Light		
Middle		
Strong		
Light		

Questions

How challenging was it for you to use the system?

	Not at all	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

Mental Demand: How mentally demanding was the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Participant #: ________________

Physical Demand: How physically demanding was the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Temporal Demand: How hurried or rushed was the pace of the task?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Performance: How successful were you in accomplishing what you were asked to do?

	Perfect									Failure
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Frustration: How insecure, discouraged, irritated, stressed, and annoyed were you?

	Very low									Very high
With socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

How can you imagine using this system in your daily life / as part of your routine?

How easy was it for you to map the pressure level with the feedback?

	Very Difficult	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

How easy was it for you to remember the pressure level based on the feedback?

	Very Difficult	A little	Okay	Easy	Very Easy
With socks	☐	☐	☐	☐	☐
Using no socks	☐	☐	☐	☐	☐

Participant #: ________________

What was problematic with the current setup? Further comments?

B.3 Task 3 – Log Sheet

Participant #: _________________

T3 : Bending Test

BASIC INFORMATION

Position	☺	☹
No bend		
Middle bend		
High bend		
High bend		
No bend		
Middle bend		
Middle bend		
High bend		
No bend		

Questions

How challenging was it for you to use the system?

Not at all	A little	Okay	Easy	Very Easy
☐	☐	☐	☐	☐

Can you imagine using a knee-guard in your daily life to detect the bend of your knee?

Appendix C

CD Content

Format: CD-ROM, Single Layer, ISO9660-Format

C.1 PDF Data

Pfad: /

Leong_Joanne_2016.pdf Master's thesis as PDF file.

C.2 Video

Pfad: /

proCover-HD.mp4 . . . *proCover* Video

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