EMG Feedback for Enhanced Control of Myoelectric Hand Prostheses

Towards a More Natural Control Interface

Tchimino, Jack

DOI (link to publication from Publisher):
10.54337/aau539427829

Publication date:
2023

Document Version
Publisher's PDF, also known as Version of record

EMG FEEDBACK FOR ENHANCED CONTROL OF MYOELECTRIC HAND PROSTHESES

TOWARDS A MORE NATURAL CONTROL INTERFACE

BY
JACK TCHIMINO

DISSERTATION SUBMITTED 2023

AALBORG UNIVERSITY
DENMARK

EMG FEEDBACK FOR ENHANCED CONTROL OF MYOELECTRIC HAND PROSTHESES

TOWARDS A MORE NATURAL CONTROL INTERFACE

by

Jack Tchimino

AALBORG UNIVERSITY
DENMARK

Dissertation submitted

Dissertation submitted: 31/03/2023

PhD supervisor: Associate Prof. Strahinja Dosen,
Aalborg University

Assistant PhD supervisor: Associate Prof. Jakob Lund Dideriksen,
Aalborg University

PhD committee: Associate Professor Steffen Frahm, (chair)
Aalborg University, Denmark

Professor Max Ortiz Catalan
Chalmers University, Sweden

Associate Professor Christian Antfolk
Lund University, Sweden

PhD Series: Faculty of Medicine, Aalborg University

Department: Department of Health Science and Technology

ISSN (online): 2246-1302
ISBN (online): 978-87-7573-724-6

Published by:
Aalborg University Press
Kroghstræde 3
DK – 9220 Aalborg Ø
Phone: +45 99407140
aauf@forlag.aau.dk
forlag.aau.dk

Printed in Denmark by Stibo Complete, 2023

CV

Jack Tchimino was born in Thessaloniki, Greece in 1991. In 2015, he obtained a B.Sc. M.Sc. in Electrical and Computer Engineering from Aristotle University of Thessaloniki, Greece. He moved on to obtain a M.Sc. in Biomedical Engineering from Delft University of Technology, The Netherlands, with his thesis regarding biomedical signal analysis. Jack began his doctoral studies in the Department of Health Science and Technology of Aalborg University in 2019. His research focuses on novel artificial somatosensory feedback methods in upper limb prostheses to enhance their control and achieve a better user experience.

Un libro es un espejo y sólo podemos encontrar en él
lo que ya tenemos adentro

Carlos Ruiz Zafón – La Sombra del Viento

ENGLISH SUMMARY

The effects of upper limb loss are varied and far-reaching. The quality of life of amputees is severely impaired by their hinderance in executing basic tasks, their social participation and self-image are negatively affected, while many amputees face phantom limb pain. Modern myoelectric prostheses provide users with partial restoration of lost hand functionality; however, high prosthesis abandonment rates are still reported, despite the ever-growing sophistication of their design. An often-cited reason for users discarding their prosthetic hands is the lack of somatosensory feedback, in other words, the fact that modern prosthetic devices do not “feel”.

Many feedback generation and delivery methods have been put forward and studied in literature. This PhD thesis focuses on a relatively novel approach: that of EMG feedback. In a proportional control scheme, the strength of the muscle contraction that the user generates is proportional to the amplitude of the recorded myoelectric signal that drives the prosthesis and, by extension, proportional to the grasp force that the prosthesis applies. By communicating the amplitude of the user’s myoelectric signal back to them in real time, the user can use that information to precisely modulate their contraction up- or downwards to a desired level and apply the force they wish. An attractive feature of EMG feedback is that this can be achieved in a predictive manner before the hand closes around an object since the user receives feedback as soon as the hand starts moving.

The advantages of EMG feedback over more traditional methods, such as force feedback, have been demonstrated in the past. This work presents a comprehensive study of EMG feedback performance in various functional force-matching tasks and comprises of four parts. The first part regards the calibration of the myoelectric control loop in the presence of EMG feedback, aiming at performance optimization and enhanced force control. The advantage of EMG feedback over force feedback is further demonstrated in the second part, where it is shown that a prosthesis control loop integrating EMG feedback is less sensitive to control disturbances and the participants could perform a task correctly, irrespective of these disturbances. The third part establishes that EMG feedback enhances the control capabilities of the user, even when grasping compliant objects, a task which provides them with ample visual feedback, which one may argue could suffice for the correct application of force. Finally, the translatability of the system to other muscle groups and body areas, for use on higher-level amputees is demonstrated in a case study making up the fourth part of this work.

It is the author's hope that the evidence presented in this thesis will pave the way for the integration of EMG feedback in commercial prosthetic devices, with the singular goal of enriching their functionality and, thus, restoring the quality of life of their users.

DANSK RESUME

At miste en arm har en række vidtrækkende konsekvenser, herunder en væsentlig forringelse af livskvalitet grundet nedsat evne til at udføre simple opgaver og til at indgå i en social kontekst samt et forringet selv-billede og fantomsmerter. Moderne myoelektriske protester kan delvist genskabe funktionaliteten af amputerede hænder, men et stort antal brugere vælger ikke at anvende deres protese på trods af at deres design bliver mere og mere sofistikeret. En hyppigt angivet årsag til dette fravalg er manglende somatosensorisk feedback, eller med andre ord; det faktum at de ikke kan føle protesen.

Der har været foreslået mange metoder til at genskabe feedback i litteraturen. Denne PhD afhandling fokuserer på en relativ ny tilgang: EMG-feedback. Når protesens styres med proportionel kontrol, vil muskelkontraktionsniveauet som brugeren yder, estimeret ved EMG signalets amplitude, være direkte proportionel med protesens grebsstyrke. Ved at lade denne EMG-amplitude bestemme feedbacksignalet til brugeren i realtid, kan brugeren anvende denne information til med stor præcision justere deres kontraktionsniveau således protesen yder den ønskede kraft. En fordelagtig egenskab ved EMG-feedback er at det tillader brugeren justerer grebsstyrken allerede inden protesen griber objektet, idet brugeren får feedback allerede fra det tidspunkt hvor protesehånden begynder at lukke sig.

Tidligere studier har påvist visse fordele forbundet med EMG feedback i forhold til andre metoder til at genskabe somatosensorisk feedback. I denne afhandling præsenteres i fire dele en omfattende undersøgelse af effekten af EMG-feedback i forskellige realistiske protese-kontrol opgaver, der har det til fælles at en præcis kraft skal ydes. Den første del fokuserer på den optimale kalibrering af den myoelektriske kontrol løkke når den inkluderer EMG feedback, med henblik på at optimere præcisionen af protesens kraft. Fordelene ved EMG-feedback demonstreres yderligere i den anden del, hvor det vises at protesekontrol ved hjælp af EMG-feedback er mindre følsomt overfør forstyrrelser i kontrolsignalet, idet brugerene kunne udføre den ønskede kraft på trods af påvirkningen af disse forstyrelser. Den tredje del viser at EMG-feedback forbedrer brugerens kontrol over protesen, selv når den griber om et eftergivende objekt. I sådanne opgaver er den kraft protesen yder ellers i høj grad synlig for brugeren da den afspejles af objektets deformitet, hvilket normalt antages for at være tilstrækkelig information for at kunne yde præcis kontrol. I det fjerde og sidste studie påvises graden hvormed EMG-feedback kan anvendes på andre muskelgrupper og på andre

kropsområder med henblik på anvendelse til proteser for arm-amputationer ved skulderen.

Det er denne forfatters håb at de resultater der præsenteres i denne afhandling vil bane vejen for integration af EMG feedback i kommercielle proteser, med det endelige mål at forbedre deres funktionalitet og øge brugernes livskvalitet.

ACKNOWLEDGEMENTS

If internet memes are anything to go by, a PhD is a challenging experience even at the best of times. This one was, however, carried out during weird and uncertain times that affected all of us in ways that we likely still haven't fully grasped. Therefore, I consider myself extremely fortunate that, even when we were all stuck inside, disinfecting our hands, or following increasingly distressing news from around the world, I felt surrounded by people with whom I shared mutual support and understanding. To wit, there are many for whom I must express my appreciation, for their contribution, however big or small, during this journey.

First and foremost, this work could not have been possible without the support and guidance of my supervisor, Strahinja Došen, to whom I owe a great deal of gratitude for his encouragement and trust. Solon said, "I grow old, ever learning more and more" and my time at Aalborg University could not be a better exemplification of this ancient quote. Your contagious enthusiasm for our field inspired me to do my best to reach your standards and I feel I became a better scientist for that. Also, a very special thank you to my co-supervisor, Jakob Lund Diderisken for his help and valuable input along the way. Working with both of you has been an immense pleasure.

Pranav, Lisha, and Kaspar, where do I start? We redefined the concept of hygge. Explored exotic foods from the comfort of our homes. Traveled to the top of Jutland. Drank more GnTs that we should admit. Developed our green thumbs. Here's to more of that in the years to come!

Fabri, Odete, Jorge, Gaby, Juan, and Luisina, thanks for the movie nights, the cocktails, the barbacoas, the Duck Game sessions, the introduction to FromSoftware, even my first ever piñata and putting up with my rusty Spanish! Shoutout to Aruma, too!

And a huge thank you to all my friends and colleagues in Denmark for making the dark Danish days a little brighter, Santos, Shima, Felipe, Fernanda, Sebastian, Romain, Sabina, Luis, Taha, Daniel, Mauri, and Martin (I hope I'm not forgetting anyone!).

To my cerdas, Albert, Mauricio, and Andrés, thank you for the amazing trips to Greece and Spain (and a smaller one in France) and for listening to my rants and my countless ulcer-inducing dad jokes. Next stop is organizing a bachelor party and our next adventure (I'm driving). Lots of love and remember to wear your azabache.

María and Su, those grueling online Catan sessions were so enjoyable, even though I hardly ever won a game. From beer falling into places where it shouldn't (WINK) to the Ed Sheeran concert, and from watching Netflix remotely to the barrage of Snape stickers, thank you so much for being there 😊.

Panos, Natasha, Makis, Pantelis, Christine, and Eva, thanks for always keeping me up to speed on the latest GNTM drama and the latest gossip, of course. And even though I should be outraged that only one of you made the journey to the north to visit me, I can't stay mad at you. Thanks for tolerating another budding doctor in the group and thanks for all the good times whenever I come back south.

Kater, Anasta, Tony, Domna, thanks for all the laughs that are too many to list here, all the Pokémon talk and the endless Among Us intrigue!

Nick, Alex, Dimitris, Prodromos, and John, I hope blowing off steam during CS-GO was as entertaining for you guys as they were for me. Thanks for injecting your well-known dose of finesse in the proceedings when all reason was out the window.

Lastly, this dissertation is lovingly dedicated to my family, for their unconditional love and support throughout my academic journey.

And finally, dear reader, if you are kind enough to bother reading through this thesis and have not been mentioned by name, thank you too!

TABLE OF CONTENTS

Chapter 1. Introduction 19
1.1. Amputation and A Brief History of Prosthetics 19
1.1.1. Amputations...... 19
1.1.2. Prostheses...... 20
1.1.3. Modern Trends 20
1.2. Myoelectric Prostheses and User Perspectives 22
1.3. Feedback Control Loops...... 23
1.3.1. Anatomical Background of Somatosensory Feedback...... 23
1.3.2. Artificial Feedback in Prosthetics...... 25
1.3.3. EMG Feedback: State of the Art 26
1.4. Thesis Goals and Structure...... 30
Chapter 2. Methodology 35
2.1. Introduction...... 35
2.2. Prosthesis Control Scheme 35
2.2.1. EMG Recording and Prosthesis Motion 35
2.2.2. Prosthetic Hands...... 37
2.2.3. Discretization 38
2.2.4. Feedback Generation 39
2.2.5. Feedback Encoding...... 40
2.2.6. Vibrotactile Stimulation...... 41
Chapter 3. Summary of Studies......43
3.1. Study 1 – Calibration of a prosthesis control loop with EMG Feedback.. 43
3.1.1. Protocol...... 43
3.1.2. Results and Discussion...... 44
3.2. Study 2 – Disturbance Analysis and Comparison of EMG and Force Feedback...... 47
3.2.1. Protocol...... 47
3.2.2. Results and Discussion...... 49
3.3. Study 3 – Compliant Object Grasping...... 51

3.3.1. Protocol 51
3.3.2. Results and Discussion 54
3.4. Study 4 – Application and Evaluation on a Transhumeral Amputee 55
3.4.1. Protocol 55
3.4.2. Results and Discussion 58
Chapter 4. Conclusions 61
4.1. Addressing the Research Questions 61
4.2. Overall Conclusions 65
4.3. Future work 66
Chapter 5. Literature 67

TABLE OF FIGURES

Figure 1-1: The flow of control signals in a sound hand and across a myoelectric prosthesis control interface. 24
Figure 1-2: Three force control scenarios made possible with the provision of EMG feedback 28
Figure 2-1: The basic control loop for prosthesis control, including all types of feedback signals presented in this work 36
Figure 2-2: (A) Thalmic Labs Myo Armband, (B) OttoBock dry EMG electrode 37
Figure 2-3: Demonstration of (A) EMG and (B) force feedback resulting from the same muscle contraction. C: the different stimulation patterns corresponding to the discretized signals and the positioning of the vibrotactors around the upper arm. 38
Figure 2-4: The prostheses used in the experiments: (A) TouchBionics RoboLimb (B) OttoBock Michelangelo 39
Figure 2-5: A C2 vibrotactor 41
Figure 3-1: Demonstration of the parameter space evaluated in this experiment 44
Figure 3-2: Summary of the success rate results for each condition tested in this experiment 45
Figure 3-3: A myoelectric signal and its disturbed versions. The levels targeted are 2 and 4, specified in the plot 48
Figure 3-4: Main results for the performance in power grasps. The diamonds indicate the performance of the amputee participant. 50
Figure 3-5: The two conditions tested in this experiment. Condition A includes only incidental (visual) feedback, while condition B also integrated EMG feedback in the control loop. The three scenarios for the application of force are shown at the bottom part of the image 53
Figure 3-6: The main outcome measures of the third experiment: the success rate (A) and the trial completion time (B). 54
Figure 3-7: Schematic of the experimental setup elements placed on the transhumeral participant 57
Figure 3-8: The summary overall results of Study 4 58
Figure 4-1: The conclusions of the four studies that comprised this work 62

Chapter 1. INTRODUCTION

1.1. AMPUTATION AND A BRIEF HISTORY OF PROSTHETICS

1.1.1. AMPUTATIONS

An amputation is the removal of a body part due to trauma or disease, such as vascular diseases, tumors, and infections. As a medical procedure, amputation has a very long history, with reports as early as 2500 years ago [1]. A congenital amputation is a condition where fetal limbs have failed to develop properly *in utero*, with an infant being born without parts of its limbs.

Invariably, limb amputations restrict a person's ability to perform tasks of daily living. Losing a limb severely disrupts a person's quality of life and can lead to chronic psychological challenges [2], [3], a reduction in social participation [2], [4], and difficulties in returning to the workplace after the injury [5]. Painful sensations felt in the absent limb, known as phantom limb pain, is an additional issue faced by many amputees, which further complicates the rehabilitation process and negatively affects the amputees' quality of life [6].

In the United States alone, there are approximately 1.7 million people with amputations [7], of which 41,000 have upper-limb differences, the majority of whom are younger than 65 years of age [8], caused mainly by trauma or cancer [9]. Depending on the condition or the damage to the limb, there are several types of amputations.

A specific distinction is that between a disarticulation and an amputation, the former of which is the removal of a limb part not by bisecting a bone, but by separating the constituent bones at a joint. Hence, in the arm we can define 3 types: wrist, elbow, and shoulder disarticulation. Procedures that involve cutting through bones include partial hand amputations (transcarpal), amputations between the wrist and elbow (transradial), which are the most common traumatic upper limb amputations [9] and between the elbow and shoulder (transhumeral). Lastly, a forequarter amputation includes the removal of the entire arm, including the scapula and clavicle. This procedure is mainly performed as a last resort for the treatment of osteosarcomas in the proximal humerus or clavicle; however, by virtue of advances in the treatment of such tumors, it is rarely performed today and has given way to modern limb-sparing surgeries [10].

1.1.2. PROSTHESES

A missing limb can be replaced by a prosthesis, which mimics the original limb in form and function. Prosthetic limbs have been used throughout human history, with the earliest known prosthetic device being a wooden toe found buried with a 3000-year-old Egyptian mummy [11]. In his *Naturalis Historia*, Pliny the Elder mentions a Roman general from the 3rd century BCE who received a prosthetic arm that allowed him to return to battle, showcasing the importance of the functional restoration of missing limbs.

Our knowledge of prosthetic devices until recent years is rather fragmented since, for many centuries, amputations were characterized by a high mortality rate due to hemorrhaging or infection [12], while only affluent citizens could afford such devices. A historic example of a prosthetic arm is that of Götz von Berlichingen, a German knight who lost his arm circa 1500 CE. The 16th century French physician Ambroise Paré advanced the design of prosthetic arms by designing spring-loaded anthropomorphic prostheses. In both cases, the arms and digits were controlled by the amputee's intact hand but restored a fair amount of functionality [13].

The first body powered prosthesis was designed in the early 19th century by the German dentist Peter Baliff [14]and it was the first attempt to integrate active functionality in the prosthetic arm [12]. Through a network of harnesses and straps, the movement of a different body part was translated into movement in the prosthesis. In more recent years, these harnesses were replaced by Bowden cables, which are being used in body powered prostheses to this day [14].

1.1.3. MODERN TRENDS

In the 20th century, aided by significant technological advances, the interest has shifted to the development of powered prostheses, active motorized devices able to move independently, in response to the user's intent. These devices are characterized by a considerably higher design complexity than their predecessors and their motion is driven by biological signals generated by the user. It is reasonable to point out that the level of amputation is itself an important factor that informs the rehabilitation process as well as the design of the prosthesis interface. A higher level of amputation limits the prosthesis control capabilities available since there are fewer biological tissues to harvest signals from, to control a larger number of degrees of freedom [15].

Nowadays, myoelectric prostheses are arguably the most widely used type of powered prostheses, and the focus of this dissertation. Electromyographic (EMG) recordings recorded from the user's residual muscles are used to generate the control signals that drive the motors of the prosthesis. The first attempt at the development of such a device was in 1948 [12] but initially failed to garner enough attention for a clinical application. Nevertheless, since then, myoelectric prostheses have evolved and are now commonly used worldwide.

The electromyogram can be recorded in several ways. Most commonly, surface electrodes are used [16]–[25], which are placed onto the skin and detect the electrical activity of the muscles lying underneath. Percutaneous and intramuscular electrodes or other invasive interfaces achieve better signal quality than surface electrodes, since they are placed closer to the signal source [26], [27]. The use of invasive interfaces, however, is mostly limited to laboratory applications or experimental trials; the simplicity of surface electrodes makes them more suited for prolonged everyday use. The development of invasive interfaces must address issues of long-term biocompatibility [28] besides the implantation procedure itself, which a user may be unwilling to undergo in the first place [29].

Lastly, targeted muscle reinnervation (TMR) [30]–[32] is a medical procedure wherein the severed nerves of the amputated limb are rerouted onto a different muscle in the patient's body. That way, signals that would otherwise generate movements in the missing limb now elicit muscle contractions in a different muscle. The EMG generated in these contractions can, consequently, be detected by surface electrodes and drive a prosthetic limb.

It has been argued that muscles can be viewed as amplifiers of the electrical signals carried by the peripheral nerves that innervate them [16]. These neural signals are generated in the motor cortex of the brain and carried to the muscles to elicit contraction and generate movement. For this reason, the electrical activity of motor nerves can also be used as a control signal for a prosthesis [33], with many examples of electrodes for direct neural interfacing in the literature [34]. However, the necessity of surgery for the implantation of the electrodes, the complexity of the system, and the potential deterioration of the implants in situ [35], [36] have impeded the commercialization and wider use of such interfaces, although valuable research is still ongoing [37].

It must be noted that the experimental part of this PhD thesis utilized exclusively non-invasive means for EMG recording, for the participants to control myoelectric hands in all the experimental tasks.

1.2. MYOELECTRIC PROSTHESES AND USER PERSPECTIVES

Advances in fields like materials science and biomechatronics have paved the way for the development of a variety of myoelectric prosthetic hands by various manufacturers, both commercial and experimental. Prosthetic hands from major manufacturers include the Michelangelo and beBionic hands by OttoBock and the iLimb series by Össur. The VINCENTevolution hand by Vincent Systems is the only commercial hand that implements somatosensory force feedback, while the Hero Arm by Open Bionics is marketed as an affordable 3D-printed hand. In fact, 3D printing has garnered attention as an attractive option for the manufacturing of prostheses, specifically in developing countries that have limited access to high-end commercial prostheses [38], [39].

Research is, of course, still ongoing, with several laboratories designing and developing their own novel prosthetic hands. A prime example is that of the Soft Hand Pro [40], whose underactuated mechanical design is informed by the natural synergies of the human hand, to offer more natural grasping patterns. Other examples include the Hannes [41], and SmartHand [42] hands. The goal of all these devices is, of course, to restore functionality in amputees, but they differ in terms of several features.

When designing a prosthesis, one must not neglect the fact that they do so to enhance the quality of life of an amputee end user. A prosthetic hand system resulting from advanced biomechatronic design and development can very easily be abandoned by its user because they find its use impractical. In fact, prosthesis abandonment rates have remained high, despite advancements in their scientific field [43]. Studies have isolated several user requirements for prosthetic hands, non-compliance with which can lead to inadequate user experience and abandonment of the device. Lack of somatosensory feedback is very commonly stated as a major weakness of modern prosthetic hands [36], [44]–[50] and will be the focus of this work. Other requirements include dexterity [29], [36], [44], [47], [48], comfort [45], [47], [48], intuitive control [44], [45], [50], [51], short response time [7], [44], [45], reliability [47], [48], [50] and aesthetics [47], [48], [50].

The integration of feedback in prosthetic interfaces has been shown to greatly benefit the user experience in several ways. First and foremost, from a practical standpoint, feedback enhances the precision of the system by offering superior control and renders it less sensitive to noise and disturbances [52]. Prosthesis feedback replaces sensations arising from the lost limb with artificial ones, which promotes feelings of embodiment, in other words, the user is more likely to regard and accept the prosthesis as a part of their anatomy [36], while a reduction of phantom limb pain has been reported when the prosthetic hand is equipped with feedback [53], [54]. Nevertheless, despite mounting evidence towards the benefits of the integration of feedback in prosthetic hands, as well as the potential for an overall improvement of the users' quality of life, there is neither a consensus on an optimal feedback method or standardized system design, nor interest to integrate feedback systems in commercial prosthesis.

1.3. Feedback Control Loops

1.3.1. Anatomical Background of Somatosensory Feedback

Coordinated movement of a limb is the result of the continuous and bidirectional communication between the limb and the central nervous system. The brain's motor cortex generates commands that are relayed to the muscles through efferent nerve fibers, while afferent fibers transfer sensory information back to the brain. The afferent fibers inform the brain of e.g., limb position and orientation and muscle force, while the fusion of tactile information from different skin areas enables the identification of the stiffness and shape of a grasped object [55]. This communication is the reason why able-bodied individuals know e.g., the exact position of their limbs even if they are not looking at them, a sensory modality known as proprioception [56].

The human hand is a highly complex biological device. Besides its remarkable dexterity, it is an important means of social interaction and a body part with which we explore and perceive our environment [57]. The fine motor control of the hand, however, would not be possible if not for the dense network of sensory nerves embedded in its skin and muscles, relaying proprioceptive and exteroceptive information (regarding the position/orientation of the hand and its interaction with the environment, respectively) to the brain's sensory cortex [56], [58]. The hand has a disproportionately large representation in the motor and sensory cortices, with a very large number of neurons responsible for its closed loop control [56]. There are between 20 and 35 thousand nerve fibers reaching the hand through the wrist [59] and the fact that the vast majority of these are sensory afferents underlines the importance of sensory information for the control of the hand.

From a control systems perspective, this constitutes a closed loop system (see Figure 1-1A), wherein commands are generated by a controller (the brain) and their result is continuously monitored (via the sensory receptors) and corrected by the controller, thus generating a fluid, controlled, and highly precise motion. Proof that this somatosensory feedback is of paramount importance is what happens when it is absent, for example in patients suffering from deafferentating conditions, wherein their natural proprio- and exteroception are absent. The feedback information within the control loop in these cases is drastically reduced, rendering movements uncoordinated and imprecise. Input from other

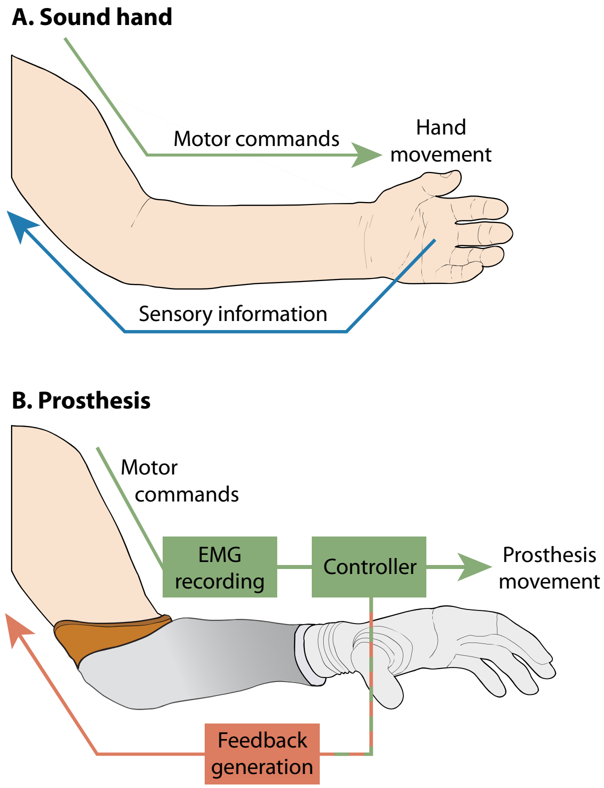

Figure 1-1: The flow of control signals in a sound hand and across a myoelectric prosthesis control interface.

senses must be used in lieu of normal somatosensory feedback to improve the precision of movements in cases like this.

The celebrated cellist Jacqueline du Pré experienced loss of tactile sensation in her fingertips due to multiple sclerosis, which affected her playing. She was initially forced to rely on visually observing and following the position of her fingers on the cello in order to perform, eventually abandoning her career altogether at the age of 28 [60]. Another characteristic and frequently mentioned example is that of the high-profile case of Ian Waterman, a patient who lost his sense of proprioception and, by extension, fine control of his limbs, who, nevertheless, managed to retrain himself to move his limbs over time, utilizing other senses, such as vision to close the control loop [61]. It can be argued that this is a strong indication of the ability of the brain to adapt and compensate for the loss of natural feedback.

The bidirectional path from the brain to the hand is severed after an amputation. As mentioned, a myoelectric prosthesis can offer restoration of hand functionality and can be controlled by contracting residual muscles. However, the path that the information follows is now unidirectional, with the brain sending commands to the muscles but not receiving any sensory information from the prosthesis. In the absence of precise feedback from the prosthesis, the information that the interface transmits to the user is reduced, forcing the latter to rely on other sources of feedback to achieve the desired prosthesis movement.

1.3.2. ARTIFICIAL FEEDBACK IN PROSTHETICS

The restoration of sensory flow to the brain and, therefore, the closing of the prosthesis control loop, can be achieved with the implementation of artificial somatosensory feedback. As a substitute for the afferent information that the user no longer receives naturally, variables regarding a system state are transmitted to the user by means of a form of stimulation, thus closing the control loop, as seen in Figure 1-1B. The wearer can now use this information to modulate and correct the generation of the control signals that drive the prosthesis.

There is a vast heterogeneity in the methods used to provide artificial somatosensory feedback as well as several ways to categorize them. Here we will use two major categorizations: invasiveness and the type of sensory information carried by the feedback.

Similar to EMG recording as mentioned before, feedback interfaces can be invasive or non-invasive. As the name suggests, invasive interfaces are embedded within the tissues under the skin. Examples can range from relatively simple subdermal electrodes [62], to peripheral nerve interfaces [63] and more complex surgical interventions, such as targeted sensory reinnervation [63] and even direct interfacing with the central nervous system [36]. Non-invasive means are markedly simpler and include electrotactile, mechano- and vibrotactile, auditory, visual, or temperature cues [64].

Schofield et al [65] provide three categories of feedback methods based on the type of sensory information that the feedback generates. Somatotopically-matched feedback is defined as the sensation perceived by the user as being anatomically matched to the part of the prosthesis that receives a stimulus. Modality-matched feedback is a stimulation scheme that is congruent with the type of stimulus that the prosthesis receives, e.g., a constriction band that can be used to communicate the prosthesis grasp force. Finally, substitution feedback transmits a system state through a stimulation paradigm that does not mimic the sensation that a natural limb would perceive in neither modality nor location; therefore, the user must be trained to interpret the stimulation provided by the feedback.

It should be noted that the method selected for feedback delivery in this work was that of vibrotactile stimulation, which constitutes non-invasive substitution feedback. The choice was motivated by the simplicity of the implementation; neither implantation nor extensive calibration (as e.g., in the case of electrotactile stimulation) was required. Of course, in order to increase the bandwidth of the feedback channel, in other words, to be able to transmit a sufficient amount of information to enhance the control of the prosthesis, the stimulation had to be cleverly designed, as will be made clear further on.

An important part of the feedback interface design process is, of course, the selection of the variable that will be communicated to the user, which also displays high heterogeneity across the literature. A widely used example is that of force feedback [22], [65]–[67], wherein the force applied by the prosthesis is encoded and transmitted to the user via the stimulation. Knowledge of the grasping force is useful for its more precise application, which is crucial for stable and safe grasping of objects. Other variables that have been considered include joint velocity [68], wrist rotation [69], [70], aperture [69] and discrete movement events [71].

An aspect of hand prosthesis function that can benefit greatly from the provision of feedback is the correct application of force. Despite its general benefits to the control of a prosthesis, we can identify an important drawback in force feedback. While the hand is in motion, the user can solely rely on their own muscle proprioception and potential incidental feedback arising from the prosthesis to modulate their contraction. The feedback is delivered to the user only after the prosthesis has applied force, thus only potentially offering *a posteriori* correctional ability. Therefore, if the user has overestimated the muscle contraction required to apply the desired force with the prosthesis, the generated grasp force will be correspondingly higher, potentially damaging a grasped object, rendering the feedback redundant.

1.3.3. EMG FEEDBACK: STATE OF THE ART

More recently, a novel method was put forward [23], wherein the user's myoelectric signal itself was transmitted to the user as feedback, named **EMG feedback.** In this approach, rather than measurements from the prosthesis or other extrinsic signals, the stimulation that the user receives depends on the amplitude of the myoelectric signal that they produce (which also drives the prosthesis). The use of EMG feedback "shortens" the feedback loop of the system (compared to e.g., force feedback) by informing the user of the signals they themselves generate.

In a proportional control scheme, used in most commercial and research applications due to its intuitive nature, the amplitude of the myoelectric signal is proportional to the force; therefore, EMG feedback can provide information about the grasp force of a prosthesis. Additionally, EMG feedback is activated as soon as the user initiates a muscle contraction and remains active for as long as that contraction is maintained. This enables the correction of a force that has already been applied but also the predictive control of this force, since with the help of EMG feedback the user can generate and maintain a muscle contraction of appropriate strength while the hand is still in motion. When the prosthesis closes around the object, it will apply the desired amount of force.

Examples of force control with EMG feedback can be seen in Figure 1-2. In the scenario in Figure 1-2A, the user generates and maintains a small contraction, the hand closes around the object and applies a grasp force. Using the information provided by EMG feedback, the user increases their contraction to the desired target level, thus reactivating the prosthesis, which increases the grasping force to the same level.

The predictive control feature of EMG feedback is seen in 1-2B and C. In 1-2B, the user generates a muscle contraction that activates the prosthesis, then, using the information provided by EMG feedback, proceeds to increase it to the desired level while the prosthesis is still in motion. In a properly calibrated system, the force applied when the prosthesis grasps the object will be the same as the level of the muscle contraction. The scenario in 1-2C shows that, in a similar fashion, EMG feedback can also allow the user to timely correct their contraction when the latter is overshot, thus avoiding the application of a larger-than-desired force and potential damage to the grasped object.

EMG feedback was introduced in [23], initially as complementary to force feedback. The participants received a combination of EMG and force feedback or solely force feedback on a computer screen while controlling a virtual prosthesis and performing routine grasping and force steering tasks. The study explored the potential benefits of the enhanced feedback loop on the control of the virtual prosthesis.

The results of this study indicated that when EMG feedback was added to a prosthesis control loop (already equipped with force feedback), the control performance in a force steering task was enhanced, while the participants were able to generate more consistent myoelectric commands compared to a system that only provided force feedback.

This approach allowed the participants to modulate their muscle contraction in real time. Additionally, the study introduced the notion of predictive force control made possible with EMG feedback, since the feedback was activated as soon

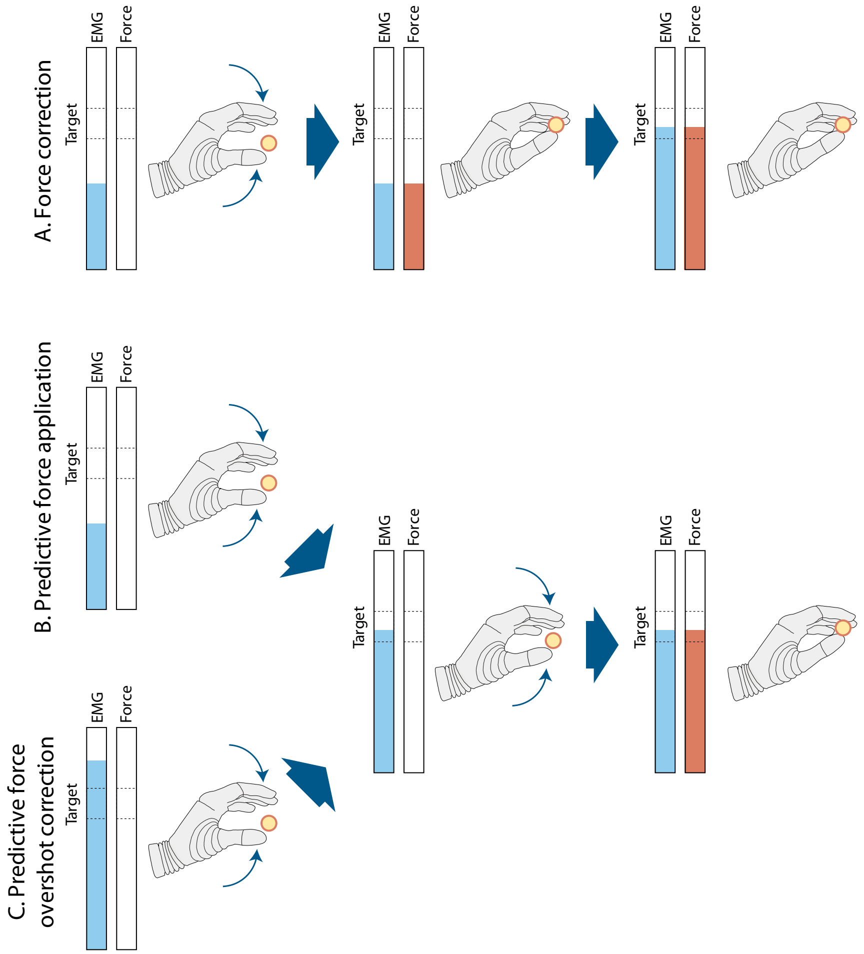

Figure 1-2: Three force control scenarios made possible with the provision of EMG feedback

as the participants initiated their muscle contraction, before a force was applied, as illustrated above. Therefore, they could apply the desired force by modulating to a suitable EMG contraction strength and maintaining that until the prosthesis closed.

This proof of concept was explored further in [22], where EMG feedback was directly compared to force feedback. The participants used a myoelectric prosthesis to perform a routine grasping task with a rigid object while receiving electrotactile EMG or force feedback. Notably, the myoelectric and force signals were discretized into 8 levels, each of which corresponded to a different feedback pattern delivered through electrotactile stimulation.

EMG feedback was shown to enhance the precision of both the myoelectric commands and the applied force. The performance of the able-bodied and the single transradial amputee that participated was significantly improved. Coupled with the nature of the experimental task (routine grasping), these results further establish EMG feedback as superior to force feedback, as well as emphasize the predictive control capabilities of EMG feedback, as mentioned above. Importantly, it was shown that EMG feedback can be used independently of force feedback, as opposed to [23], and still yield satisfactory results.

A similar comparison between EMG and force feedback was conducted under a different framework in [18], wherein the speed-accuracy tradeoff between the two approaches was evaluated. Besides an overall higher performance, EMG feedback was shown to offer the participants the ability to perform a grasping task faster, without negatively affecting the performance, while also promoting the repeatability of smoother myoelectric commands.

The team in [72] transmitted EMG feedback along with several other types through augmented reality (AR) and sound cues using a Google Glass device. The functional task in the experiment involved the relocation of sensorized clothespins, which the participants were required to grasp with a prosthesis, while restricting their applied grasp force with the help of the AR feedback they received. The task was executed with and without artificial feedback.

The results indicated that the provision of feedback significantly benefitted the control as the complexity of the task increased. The study was meant as a demonstration of the function of a complete prosthesis feedback interface, but it also provided further proof of the general benefits of feedback and the applicability of EMG feedback.

Additionally, the participants reported that they found focusing on the supplementary feedback more useful than their own intrinsic feedback sources, as the AR system provided them with more precise information. However, since they had to shift their focus from the prosthesis and grasped pin to the AR visual feedback and back, they reported that they preferred to rely on aperture or force rather than EMG feedback, as they were less dynamic and prone to rapid changes and, thus, easier to follow.

Since there is usually a direct mapping between the myoelectric signal amplitude and the velocity of the prosthesis motors, an approach wherein the velocity is selected as the feedback variable is also analogous to EMG feedback and explored in [73]. The feedback was delivered through audio cues and the goal of the study was to assess the robustness of the system when feedback was provided in response to perturbations compared to a condition without feedback in a virtual reaching task.

The outcome of a perturbation analysis of this kind is important, since it can offer valuable insight in the function of a prosthesis interface in a real-life application, given that experiments are usually conducted under nominal conditions, overlooking the effect of disturbance sources present in everyday prosthesis use [74]. The feedback interface tested in this study was shown to maintain the accuracy of the participants' performance despite the perturbations compared to the condition without feedback.

The approaches illustrated above fall under the broader category of biofeedback. By directly transmitting information about the muscle contraction strength in real time, EMG feedback effectively amplifies the information a user receives from their proprioceptive afferents during a muscle contraction. That way, the user can monitor and modulate their contraction in real time, using a more reliable sensory input (the feedback) as a guide.

1.4. Thesis Goals and Structure

Closing the prosthesis control loop with EMG feedback is a promising new technology, still being researched. A keen observer can identify some aspects of EMG feedback that can be explored further and knowledge that can be built upon. The goal of this PhD was to systematically evaluate and explore aspects of this novel somatosensory feedback method, in order to further establish it as a potential candidate for integration in commercial hand prosthesis interfaces. It is the author's belief that, based on the results of the experimental part of this work, as well as evidence presented in past literature, the implementation of EMG feedback can significantly enhance user experience.

The studies that have explored EMG feedback mentioned above used a proportional scheme for the control of the prosthesis. The first step in such a control scheme is the raw EMG processing and the creation of a suitable myoelectric signal. This pipeline most usually involves low-pass filtering (of a rectified raw EMG or a raw EMG envelope) and normalization to a percentage of the maximum voluntary contraction (MVC).

However, there is no consensus on the optimal selection of parameter values for the low-pass cutoff frequency or the calibration percentage and, due to the nature of the signal processing pipeline, the morphology of the resulting myoelectric signal, which is used to both actuate the prosthesis and generate the feedback, is highly dependent on these parameters. Therefore, it can be hypothesized that there is a set or range of parameter values that enhances the control scheme by offering better control capabilities and assisting the user in fully exploiting the benefits of EMG feedback.

The first study that comprises this work tackles this hypothesis by testing the performance of participants in a force-matching task with a hand prosthesis in nine conditions, each of which was characterized by a different parametrization of the EMG signal processing pipeline.

Research Question 1: *What signal processing parameter values optimize the control of a prosthesis control system equipped with EMG feedback?*

As mentioned above, a system equipped with EMG feedback has been shown to display superior robustness than one where no feedback is available. Moreover, despite being fundamentally different, both EMG and force feedback have been shown to offer notable benefits in prosthesis control, although EMG feedback has displayed a significant advantage over force feedback [22], [23].

A robust control interface will protect against unwanted behavior by the prosthesis, brought about by intrinsic or extrinsic disturbance sources, an important feature for appropriate use of the device in real-life scenarios. Due to their respective natures, and more specifically their difference in the timing of the stimulation, it can be assumed that users receiving EMG feedback will respond to disturbed control signals differently than those receiving force feedback. So far, however, their performance in disturbed conditions had never been directly compared; in fact, different feedback delivery methods are seldom compared under disturbed conditions [75].

Therefore, a robustness evaluation and comparison of EMG and force feedback in a functional force matching task was the next study to be conducted as part of this work. The control signals in the experiment were altered (amplified or attenuated) in random trials without the participants' knowledge, while the latter were performing a force-matching and relocation task, with EMG or force feedback, on a modified box-and-blocks task. Considering the properties of EMG feedback led to the hypothesis that it would offer superior performance and robustness to control disturbances than force feedback.

Research Question 2: *Can EMG feedback mitigate the effects of control disturbances and promote robustness in a prosthesis control loop more effectively than force feedback?*

Studies that explore the effects of artificial somatosensory feedback generally tend to employ functional tasks that involve grasping rigid objects and evaluate the system based on the performance in these particular tasks. Additionally, it is common for the experimenters to use some form of sensory deprivation on the participants (such as noise cancelling or visual obscuring of the prosthesis). This restricts the access of the participants to incidental cues arising from the prosthesis and forces them to rely exclusively on the artificial feedback that is being evaluated during a given experiment.

Such studies can offer valuable insight on the effects that a particular feedback system can have on a prosthesis control loop (in fact, noise cancellation was employed in Study 1 of this work); however, the restriction of incidental feedback does not reflect a real-world usage condition. There are, indeed, strong indications that in certain scenarios incidental cues are actually more valuable sources of information than artificial feedback [25], while supplementary feedback can be rendered redundant if it does not expand on the information available from incidental cues [76].

Incidental feedback is arguably more prevalent when grasping compliant objects, as the grasping force can be estimated based on the deformation of the grasped object, casting doubt over the usefulness of supplementary feedback. As illustrated above, EMG feedback offers an augmented version of intrinsic signals that cannot be made immediately obvious by visual observation of the grasped object; therefore, it would be reasonable to assume that it would, in fact, be useful for the control of a prosthesis, despite the presence of strong incidental cues.

Hence, the third study compared the performance of participants using a prosthesis with and without EMG feedback in a task involving grasping and reallocation of clothespins. Hypothetically, the performance in terms of force application would be similar in both cases, since the participants could use incidental cues to successfully complete it, but the properties of EMG feedback would offer faster completion time and more consistent commands.

Research Question 3: *Can EMG feedback improve prosthesis force control even in the presence of rich incidental feedback in the form of visual cues?*

Lastly, the cited works relevant to EMG feedback have tested their systems on below-elbow amputees or solely on able-bodied participants. There are no studies wherein the principles of EMG feedback are applied on higher-level amputees, therefore the efficacy of the system in this user population is unknown. Additionally, the system has to be drastically altered in order to accommodate the anatomy of a transhumeral amputee. For example, in the form in which it has been tested, the EMG signals are recorded from the forearm muscles, which are obviously absent in transhumeral amputees.

Therefore, for the final part of this work an exploratory case study was conducted, wherein the setup was altered and applied on a single participant with a transhumeral amputation, to evaluate the benefits of the control scheme with EMG feedback in a configuration that has never been tested before. Over four experimental sessions, the participant performed a force-matching task with a hand prosthesis under different configurations.

The pectoralis major muscle on the amputated side was selected for the recording of EMG signals, while the stimulation was delivered in the anterior and posterior shoulder area. The muscle selected displayed substantial atrophy due to disuse since the amputation, so it was interesting to evaluate the ability of the subject to control the interface with this particular muscle. Due to the condition of the muscle, we were expecting the participant to have difficulties integrating the control loop. Hopefully, EMG feedback would help the participant to overcome such obstacles.

Furthermore, the benefit of EMG feedback was evaluated by comparing its performance to a condition wherein no feedback was delivered. Lastly, the performance of the system with different EMG recording sites was explored, to assess the translatability of the EMG interface to different body areas

---

Research Question 4: *Does EMG feedback still benefit the control when the control interface configuration is altered for use in a transhumeral amputee?*

It is the author's belief that answers to these research questions will substantially enrich our knowledge and understanding of EMG feedback, further establish this approach as a viable solution for the provision of artificial somatosensory feedback in prosthesis control systems, and hopefully play a role in its eventual application in commercial devices. To that end, four experimental studies were conducted as part of this PhD, which will be presented in the following chapters.

The remainder of this thesis is structured as follows: Chapter 2 presents a detailed description of the functional and design principles of the experimental setups that were used in this work; Chapter 3 summarizes the methods and findings of the studies that comprise the work; Chapter 4 offers a discussion and conclusions based on the results, as well as future prospects in this promising field of supplementary somatosensory feedback in upper limb myoelectric prostheses.

Chapter 2. METHODOLOGY

2.1. INTRODUCTION

As stated before, this PhD thesis focuses on studying the integration of EMG feedback in the control loop of myoelectric hand prostheses. To that end, a series of four experimental studies was conducted, which included the execution of a variety of tasks using a prosthetic hand. The overall design of the setup was similar in all studies; thus, the elements used throughout them will be presented in this chapter.

2.2. PROSTHESIS CONTROL SCHEME

An overview of the basic form of the control scheme implemented in the experiments can be seen in Figure 2-1. In summary, the muscle contractions generate a surface EMG, which is recorded and processed. From the processing of the EMG, two signals are generated: one is translated into velocity commands for the prosthesis, and one is discretized and used to generate EMG feedback. If force feedback is used, then force measurements from the prosthesis are similarly discretized and used for feedback generation. In either case, each level of the discretized signal is translated into a different spatial pattern of vibrotactile stimulation, which is delivered to the user. The user also receives feedback through incidental cues from the prosthesis and their own muscle proprioception. The constituent parts of the block diagram are expanded upon in the following sections.

2.2.1. EMG RECORDING AND PROSTHESIS MOTION

Initiation of any voluntary movement starts in the motor cortex of the brain, which transmits neural impulses to muscles, which respond by contracting and producing movement.

The raw EMG resulting from the muscle contraction was recorded and underwent some processing steps to obtain a suitable myoelectric signal. The latter was produced by two methods. In Study 1, a Myo Armband (Thalmic Labs, USA)

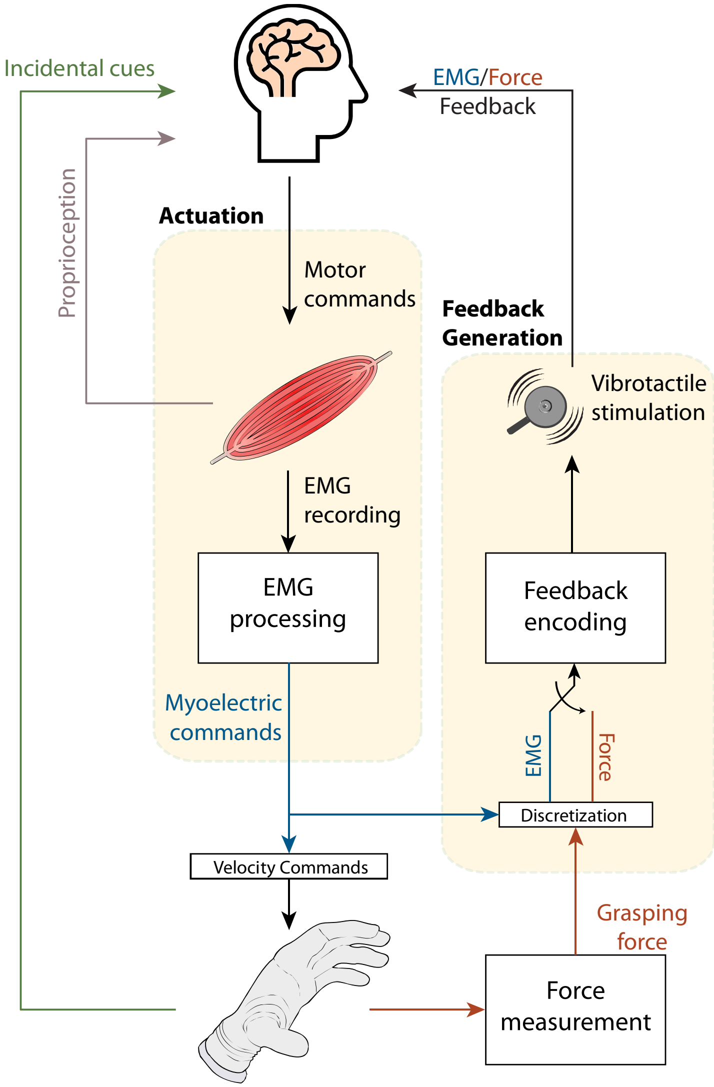

Figure 2-1: The basic control loop for prosthesis control, including all types of feedback signals presented in this work

was used to record the raw EMG, which was full-wave rectified and low-pass filtered, resulting in the signal envelope [77]–[79]. In Studies 2 to 4, a linear envelope of the surface EMG was obtained [67], [80], [81] using active dry electrodes (12E200, OttoBock, Duderstadt, Germany) and was also low-pass filtered. The equipment can be seen in Figure 2-2.

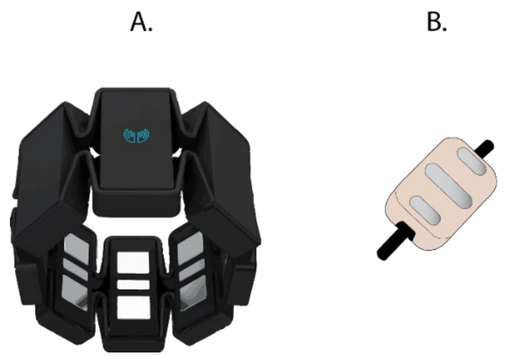

Figure 2-2: (A) Thalmic Labs Myo Armband, (B) OttoBock dry EMG electrode

This signal then underwent MVC normalization [22], [66], [82], [83]. The resulting myoelectric signal was used to generate the velocity commands that drove the prosthesis motors. The amplitude of the myoelectric signal was proportional to the closing velocity of the prosthesis, in other words a stronger muscle contraction would cause the hand to close faster and generate a larger force. The (normalized) myoelectric signal was mapped onto the closing velocity of the prosthesis, with 0 and 1 corresponding to no movement and the maximum closing velocity of the hand.

As seen in Figure 2-3, the EMG thresholds were selected so that the levels they defined were of increasing width. This was a conscious choice, in order to mitigate the effect of the higher signal variance that is exhibited in stronger muscle contractions, due to the multiplicative signal dependent noise that is intrinsic to the EMG [22], [23], [84].

Importantly, hand opening was implemented with simpler schemes since controlled opening of the prosthesis was not within the scope of this work. Henceforth, unless otherwise stated, the terms concerning EMG or myoelectric commands will be referring to signals generated for the closing of the prosthesis.

2.2.2. PROSTHETIC HANDS

Two myoelectric prosthetic hands were used, shown in Figure 2-4: a RoboLimb (TouchBionics, UK) and a Michelangelo (OttoBock, DE). The RoboLimb was used in Study 1 and offers velocity control over individual digits and thumb opposition (a total of six degrees of freedom). The Michelangelo was used in Studies 2 to 4 and offers velocity control over two degrees of freedom: opening and closing in two predefined grasp types (palmar and lateral) and wrist rotation.

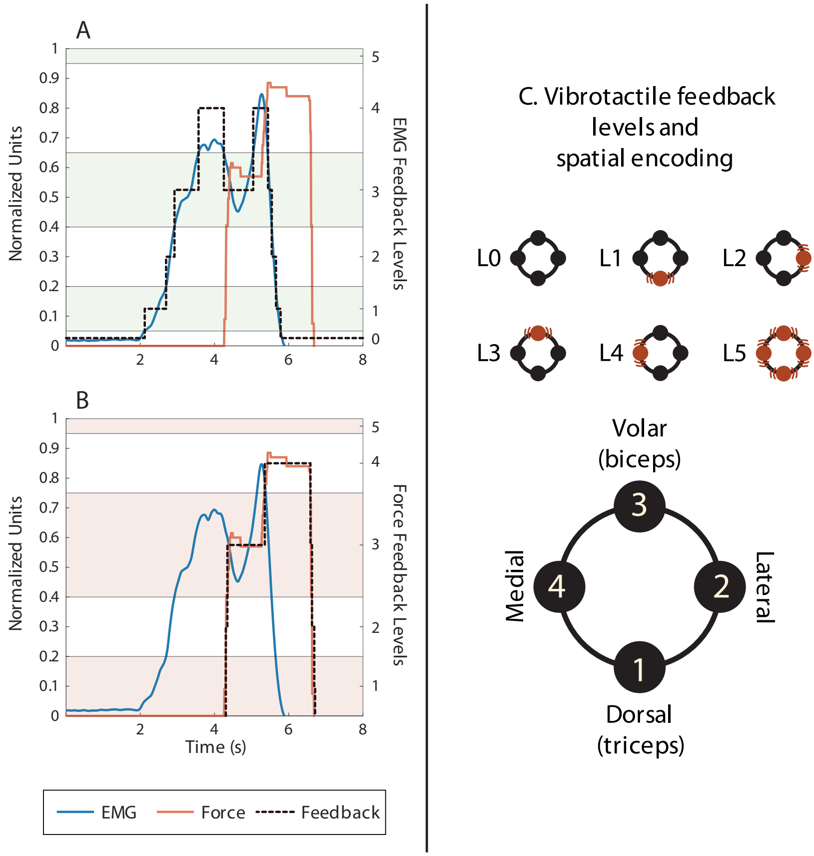

Figure 2-3: Demonstration of (A) EMG and (B) force feedback resulting from the same muscle contraction. C: the different stimulation patterns corresponding to the discretized signals and the positioning of the vibrotactors around the upper arm.

Both hands are non-backdrivable [25], meaning that the applied grasp force was maintained when the participants relaxed their muscles. An important difference between the two prostheses is that the Michelangelo has embedded force sensors that measured that output the grasp force that the hand applied, while the RoboLimb is not sensorized.

2.2.3. Discretization

The continuous myoelectric and force signals were divided in six and five levels, respectively, as shown in Figure 2-3A and B. One of the main functional principles of the interface was that the EMG and force levels corresponded to each

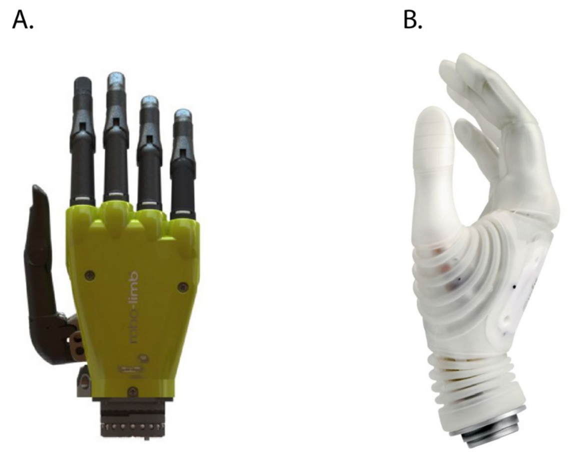

Figure 2-4: The prostheses used in the experiments: (A) TouchBionics RoboLimb (B) OttoBock Michelangelo

other, i.e., a maintained level *n* muscle contraction will result to the hand closing and applying a level *n* force. As mentioned, relaxation of the muscles did not cause a reduction of the force, due to the non-backdrivability property of the prosthetic hands used. However, if the hand was closed and applied a level *n* force, a contraction to a level higher than *n* would reactivate the motors and increase the force to that level. The lowest EMG level was deemed a dead zone and if the amplitude of the myoelectric signal lay within its range, the prosthesis was not activated and no feedback was delivered. As such, the EMG dead zone does not correspond to a force level.

2.2.4. Feedback Generation

The EMG and force levels that were communicated to the participants, implemented EMG and force feedback, respectively. The feedback was generated by using the discretized myoelectric signal that the participants generated or the discretized grasp force that the prosthesis applied, to implement EMG and force feedback, respectively, as will be elaborated further on. The effectiveness of discretized feedback has been illustrated in past research [22], [25], [67], [85], [86]. Importantly, in the experiments conducted, these two feedback types were never simultaneously activated (see switch in Figure 2-1).

Additionally, it is important to note that, in general, while in use, a prosthesis generates incidental cues that can also be regarded as feedback and used by the participants as a sensory substitute, that have been shown to benefit force control [25]. This incidental feedback includes auditory cues and vibrations from the prosthesis motors, as well as visual information. There are instances where the incidental information can be argued to be rich enough to render additional feedback redundant [25]. However, evidence suggests [23], [66], [82] that supplementary feedback greatly benefits the control capabilities of users. The participants' natural muscle proprioception is also a source of feedback and, thus, considered in the control scheme.

As seen in Figure 2-3A, EMG feedback followed the myoelectric signal and transmitted the level in which the latter lay at every moment. The participants started receiving a stimulation as soon as their myoelectric signal crossed the dead zone into level 1. As they modulated their contraction, they received a different stimulation, depending on the level of their myoelectric signal. When the participants relaxed their muscles and the amplitude of the myoelectric signal returned to the dead zone, the feedback was deactivated, but the force was maintained until the hand was opened.

Conversely, in the case of force feedback, the stimulation followed the same principles, but this time depended on the amplitude of the force signal measured by the prosthesis. In Figure 2-3B, the hand motion was initiated at ~2 s but the feedback was activated when the hand closed, and force was applied (~4 s). The participant relaxed their contraction but since the feedback now followed the force, it was maintained until the force was eliminated when the hand was opened.

Notably, this illustrates the major difference between the two feedback types described in the previous chapter, namely that EMG feedback provided real-time information to the user of a signal they had direct control over. The scheme could be used predictively [23], with the participant modulating to their desired level by following the EMG feedback and maintaining the contraction before the hand closed. The level of the force eventually applied should correspond to the level of EMG feedback that the participant received.

2.2.5. Feedback Encoding

Spatial encoding was selected for the delivery of feedback. The stimulations were delivered at four sites, positioned equidistantly and circumferentially around the upper arm of the participants, as seen in Figure 2-3C. In either feedback type, levels 1 to 4 corresponded to the delivery of stimulation at one of four sites, activated sequentially, while level 5 corresponded to the simultaneous stimulation of all four sites. Spatial encoding of feedback has been used in the past and has been shown to be effective and easy to perceive and interpret [67], [86].

The positioning of the stimulation sites (Figure 2-3C) was also an important design aspect. Their equidistant placement around the upper arm ensured that the participants could distinguish which tactors were active at any given moment and, thus, interpret the information provided by the feedback.

2.2.6. VIBROTACTILE STIMULATION

Figure 2-5: A C2 vibrotactor

Finally, feedback was delivered via vibrotactile stimulation using C2 and C3 vibrotactors (Engineering Acoustics Inc., Casselberry, FL, United States) at the stimulation sites around the upper arm (see Figure 2-3). A tactor can be seen in Figure 2-5. The vibration created by these tactors can be independently calibrated in gain and frequency. A moderate value for the gain was selected so that the vibrations were clear but non-intrusive. Their frequency was set at 230 Hz, which lies within the range of maximum sensitivity of the skin mechanoreceptors sensitive to vibration (Pacinian corpuscles) [87].

Chapter 3. Summary of Studies

3.1. Study 1 – Calibration of a prosthesis control loop with EMG Feedback

3.1.1. Protocol

As mentioned before, low pass filtering and MVC normalization in order to obtain the signal envelope is the first step in the processing of the recorded EMG signal. There is great heterogeneity in the scientific literature regarding the selection of the low pass filter cutoff frequency for the creation of the EMG envelope and the percentage of the MVC normalization. The hypothesis driving this study was that there was a value pair or range within the cutoff frequency – MVC normalization percentage parameter space for which the force control of the prosthesis when EMG feedback was provided was optimized.

Different values of these parameters result in a different morphology of the myoelectric signal that drives the prosthetic hand [22], [78], [79], [88]–[90]. It was hypothesized that this has direct implications for the ability of the participants to exploit EMG feedback and, by extension, the quality of the control scheme and the correct application of grasp force. The low pass cutoff frequency and the MVC normalization percentage affect different features of the signal and aspects of the system as a whole.

The cutoff frequency dictates the spectral content of the myoelectric signal, in other words, how smooth it is; a lower cutoff would result in a smoother, less noisy signal and vice versa. Therefore, selecting a lower cutoff frequency to eliminate higher frequencies that may degrade the quality of the signal seems to be the obvious choice. However, a reduction in the cutoff frequency brings about an increase in the system's response time, or its ability to respond to rapid changes in the input, namely the participant's muscle contractions, potentially introducing a latency that would negatively affect the control of the prosthesis.

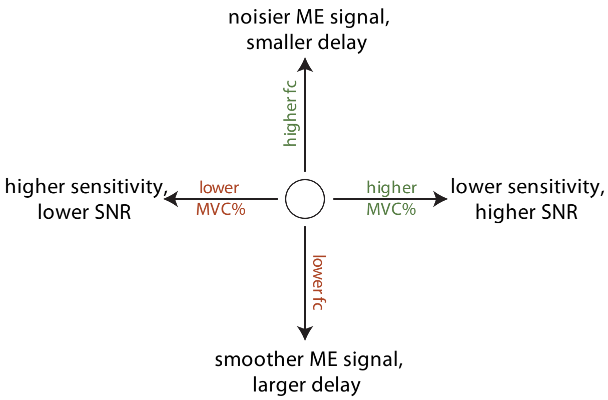

Figure 3-1: Demonstration of the parameter space evaluated in this experiment

The myoelectric signal is most often calibrated to a percentage of the MVC [23], [67], [91], [92]. Changing the MVC normalization alters the strength of the muscle contraction required to generate a myoelectric signal of a specific amplitude. Hence, a lower MVC calibration results in a more sensitive system, with smaller contractions necessary to saturate the myoelectric signal and reach the intermediate levels, and vice versa. A similar tradeoff exists in this case as well; specifically, weak muscle contractions generate EMG that is characterized by low signal-to-noise ratio (SNR) [93], which can also be detrimental for control. The interplay between the parameters is illustrated in Figure 3-1.

These tradeoffs render the relationship between the parameter selection and the control quality rather complex, with no value combination being obviously superior to the rest. The manner in which EMG feedback factored in the different configurations was also unclear. For this reason, this experiment involved the participants controlling a hand prosthesis with different system calibration conditions while receiving EMG feedback. Their performance would indicate if there was a parameter pair that offered better force control of the prosthesis.

Three values were tested for each of these two parameters, adding up to nine value combinations, making up a 3-by-3 parameter space. The values were 60, 40, and 20% for the MVC normalization and 0.5, 1, and 1.5 Hz for the cutoff frequency. The participants performed a force matching task with a prosthesis with an EMG-feedback-enabled control interface and their performance was evaluated based on their precision of EMG modulation.

3.1.2. Results and Discussion

The two parameters were tested independently to identify the best value for each one and gain insight on their overall effects on control. The results indicate

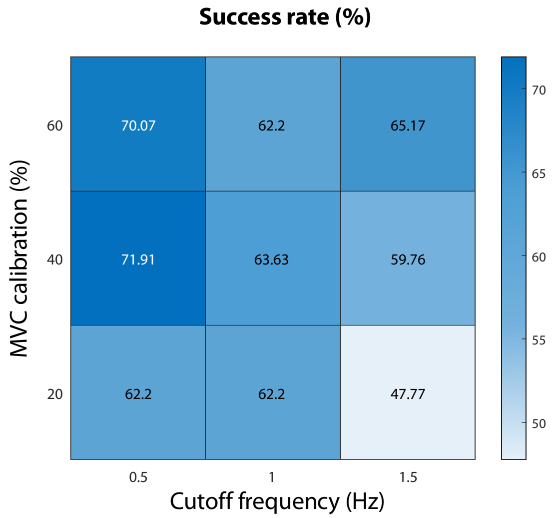

Figure 3-2: Summary of the success rate results for each condition tested in this experiment

that the performance was enhanced when the MVC was calibrated to a higher value. Normalizing to 20% of the MVC caused the participants' success rate to be significantly lower than when normalizing to 60 and 40%, both of which displayed an overall success rate of approximately 65%. A visual representation of this result is shown in Figure 3-2.

Evidently, when calibrated at 20%, the system became too sensitive for the participants to be able to properly control it. Judging by their success rate, the participants displayed a preference for a less sensitive system, at the expense of the requirement for larger muscle contractions to drive the prosthesis. At lower calibrations, the system became too sensitive for the participants to be able to control it properly, despite receiving real-time information about their myoelectric signal amplitude through EMG feedback, which, evidently, could not compensate for the high system sensitivity.

Different normalization values resulted in a different mapping of the participants' muscle contractions onto the normalized range of the myoelectric signal. Normalization to a lower percentage would require the participants to utilize a smaller contraction range to drive the prosthesis, while the myoelectric signal would be saturated with a smaller contraction compared to the case where a higher normalization percentage was used. The thresholds of the different EMG levels remain unchanged; however, in absolute terms, normalization to a lower percentage would require the participants to modulate their EMG to narrower ranges to achieve a desired force, which proved to be a significantly more demanding task.

Additionally, the higher SNR that characterizes the EMG arising from small muscle contractions could also contribute to the participants' inability to maintain their myoelectric signals within the narrow target levels, hampering their performance. Of course, stronger contractions are characterized by signals of larger variance, which may also be detrimental for prosthesis control. However, this will be addressed by the selection of a proper low pass cutoff frequency, as will be elaborated later on.

As far as fatigue was concerned, the value of 40% could be used as a compromise between sensitivity and avoidance of fatigue, while still offering good performance.

The analysis indicated that the performance with a cutoff frequency of 0.5 Hz was significantly higher than with 1.5 Hz (see Figure 3-2). A lower cutoff frequency would result in a smoother myoelectric signal, which seemed to be favored by the participants, compare to a signal that more closely followed the signal peaks that formed the signal envelope. Cutoff frequencies as low as the ones tested suppress rapid changes in a signal, potentially reduce the response time of a system, and introduce a lag between an actuation and its effect on the output. The system, however, did not seem to be affected by the action of a very low cutoff frequency low pass filter, rather the participants displayed higher performance in this case.

Moreover, as mentioned earlier, the preference for a higher normalization level would require the participants to execute stronger muscle contractions, which generate EMG signals of higher variance, due to the multiplicative signal dependent noise inherent to EMG [93]. The suppression of more rapid changes in the EMG that may have been caused by this variance rendered the myoelectric signal smoother and easier to control, eventually offering better performance of approximately 65% when the two lower cutoffs were used than the performance obtained with a cutoff of 1.5 Hz. Similar to the compromise regarding the selection of 40 rather than 60% MVC normalization, selecting 1 Hz as a cutoff for the filter is an acceptable compromise, since it both displayed similarly high performance with the lowest frequency tested and would limit any lag caused by the filtering, further reducing its effect.

The tradeoffs inherent in the parametrization of the EMG processing pipeline render the value selection rather complicated. This study offered a framework for the fine-tuning of a prosthesis control interface integrated with EMG feedback for superior force control, as well as an indication for which value ranges optimize force control. As it transpires, in an interface like the one presented in this work, low sensitivity and smoother control signals are crucial for optimal control.

Lastly, it was concluded that the division of the myoelectric signal space can be altered to increase the range of the levels and facilitate the task. This insight, along with the adoption of the 40% MVC – 1 Hz parameter pair for EMG processing were integrated in the subsequent experiments comprising this work.

3.2. Study 2 – Disturbance Analysis and Comparison of EMG and Force Feedback

3.2.1. Protocol

A vast majority of experiments in the field of prosthesis control are conducted in laboratories, under conditions free of noise or disturbance sources and it is under these conditions that control interfaces are tested and evaluated. However, these conditions do not correspond to the ones under which the prosthesis is meant to be used in daily life, where any number of extrinsic or intrinsic disturbance sources may contaminate the system and negatively affect the control, such as electrode shifts and changes in the electrode-skin interface [74], [94].

In the case of an open loop system, any disturbances would potentially not be perceived by the user, leading to unwanted behavior by the prosthesis, such as the application of an erroneous grasp force on an object. Such a scenario should be predicted during the design of the prosthesis control system and measures should be taken to mitigate the effects of control disturbances. By definition, adding a feedback loop to a control system can suppress these effects. Thus, the participants could become aware if the system behaved in a way it was not supposed to and modulate their muscle input accordingly.

Due to the nature of its signal processing steps, disturbances in the control loop would translate to the alteration of the myoelectric signal amplitude. Left unchecked, the errant control signal would result in the production of an undesired grasp force by the prosthesis.

Importantly, the perception of the system behavior and, by extension, its altered state differed between the two feedback conditions tested in this study: EMG and force feedback. In nominal conditions, EMG feedback provided real-time information on the participant's muscle contraction level (following the amplitude of the myoelectric signal), while force feedback communicated the force level and, thus, was only activated after the prosthesis had applied said force. Therefore, when force feedback was provided, there was a short time between the initiation of prosthesis movement and the application of force, wherein the participants received no stimulation.

Hence, when the control signal was disturbed, EMG feedback would make the participant aware of the altered system state more promptly than force feedback; therefore, they would theoretically be able to correct their muscle contractions before the hand could apply an erroneous force. This leads to the main hypothesis of this study, which was that when the control signal was disturbed, EMG would outdo force feedback in terms of participant performance.

To test this hypothesis, the participants (divided into two groups, each receiving one of the two feedback types) executed a modified box-and-blocks test to accommodate a force-matching task, wherein they were required to apply a specific force on a block, move it to a different compartment and release. An incorrect grasp force in a trial meant that they had to repeat it until they were successful. Two force levels were targeted (see Figure 3-3); namely, levels 2 and 4 as they are defined in section 2.2.3, corresponding to delicate and power grasping, respectively.

Some of the trials were characterized by an artificially altered myoelectric signal gain, of which the participants had not been informed beforehand [71]. The artificial disturbance was meant to emulate the effects of real-life disturbance sources. The two disturbance gain values that were used increased and decreased the myoelectric signal amplitude by 100% and 33%, respectively (see the example traces in Figure 3-3).

The performance of the participants was evaluated based on their force application precision, the number of attempts and time they required to apply the correct force in their successful attempt in each trial. A participant with transradial

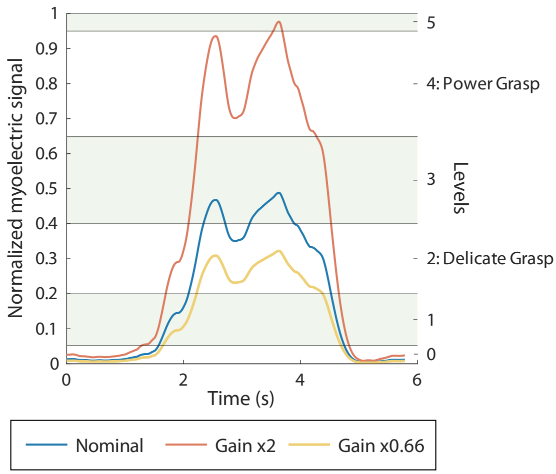

Figure 3-3: A myoelectric signal and its disturbed versions. The levels targeted are 2 and 4, specified in the plot

amputation was also recruited and performed the same task with a specially designed prosthetic socket. When the gain was altered, the participants receiving EMG feedback were expected to follow the feedback and correct their contraction as soon as they became aware of the disturbances. Hypothetically, this would lead to a higher success rate, fewer attempts per trial and faster completion time compared to force feedback.

It was assumed that the disturbances would affect the performance with both feedback types, but EMG feedback was expected to perform significantly better in both force levels. Despite the results not fully meeting this expectation, they indicate that EMG feedback does offer an advantage over force feedback on a number of key conditions.

It is important to note that the participants were given an extensive training session, wherein they were tasked with applying the desired forces under nominal, non-disturbed conditions. The goal of the training was twofold: first, the participants needed to get acquainted with the setup and familiarize themselves with the prosthesis force control and the task at hand and, second, it aimed at the development of the participants' internal models of the system [20], [95].

This way, the participants would internalize the mapping of their muscle contractions to the behavior of the prosthesis and the force they applied with it. A disturbance in the control signal would disrupt this balance, rendering the models temporarily invalid. In this case, the participants would have to follow the feedback to achieve the correct force, since it would be their only guide through the altered system dynamics.

As illustrated in 2.2.4, force feedback was only transmitted to the participants as soon as the prosthesis applied a force, as opposed to EMG feedback, which was activated and maintained when a contraction was initiated. Therefore, when force feedback is used, the participants would be aware of the altered dynamics of the system only after force feedback was activated. Due to this latency between the initiation of the muscle contractions and its delivery, it was assumed that force feedback was going to offer lower performance than EMG feedback.

3.2.2. Results and Discussion

When performing power grasps, the participants that received EMG feedback had a 13.3% higher success rate (73.3 against 60% with force feedback) with high disturbance gain and completed their successful attempts on average 1.16 s faster (2.48 s against 3.64 s) when the disturbance gain was low than the participants that received force feedback (Figure 3-4). Surprisingly, there were no differences in the performance metrics between the two feedback types in delicate grasps, while, on average, it took participants between 1 and 2 additional

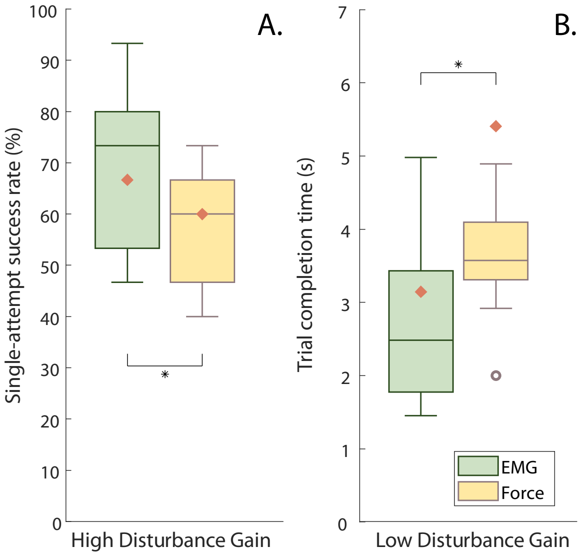

Figure 3-4: Main results for the performance in power grasps. The diamonds indicate the performance of the amputee participant.

attempts (after a failed initial one) to apply the desired force level across conditions.

When the gain was increased and the participants targeted level 4, they initially contracted their muscles like they were trained to (in other words followed their internal models of the system dynamics), since they had no *a priori* knowledge of the presence of the disturbance. However, this contraction would now cause an overshoot in the myoelectric signal due to its disturbed gain. This behavior was immediately communicated to the participant through EMG feedback; therefore, they could modulate it properly before an erroneous force was applied. Conversely, the participants who received force feedback became aware of the overshoot after the prosthesis had closed and applied a grasping force. Therefore, due to the delay between the initiation of the muscle contraction (and generation of an overshoot in the myoelectric command) and the delivery of force feedback, the disturbance could not be mitigated by the participants that received force feedback, resulting in their success rate to be significantly lower.

The case of the low-gain disturbed power grasp is also interesting. The success rate with both feedback types was similarly high, but the completion time with EMG feedback was significantly shorter, favoring this feedback type. The effect of the gain reduction was equivalent to the increase of the MVC normalization and, therefore, to the reduction of the system's sensitivity, which was shown to benefit control (see Study 1), which explains the high success rates. To explain the difference in the completion time, the behavior of the system and the participants during the trials must be examined in more detail.

EMG feedback made the participants immediately aware of the fact that the amplitude of the myoelectric signal was not high enough to achieve the power grasp target force, so the participants could readily strengthen their contraction to reach the desired level, thus increasing the closing velocity of the hand and reducing the completion time. On the other hand, this was not possible with force feedback, wherein the participants had to wait for the hand to close around a grasped object, only at which point they realized that the force was too low and increased their contraction following the force feedback to the appropriate level.

Interestingly, the number of repeated attempts was uniform across all conditions. After an initial erroneous attempt, the participants had to adapt to the new system dynamics and this adaptation was expected to require more attempts when they received force feedback. It can be assumed that the participants made use of incidental cues from the prosthesis to gauge the altered system state and adapt to it, since it has been shown that incidental feedback is also largely beneficial for the control of prosthetic devices [25].

Lastly, the amputee participant's performance closely followed the trends of the able-bodied participants, implying that there was little difference in the way that the amputee used the system. Moreover, EMG feedback offered markedly better performance in certain conditions, which indicates that it was also easily interpretable and intuitive by the amputee.

In summary, even though the performance differences between the two feedback types were not as pronounced as expected, EMG feedback is still further established as superior to force feedback. The participants could mitigate the effect of control disturbances and perform better with EMG feedback, similar to the behavior seen in nominal non-disturbed conditions [22]. The effect that incidental feedback has on the control of the prosthesis is further explored in the next study.

3.3. Study 3 – Compliant Object Grasping

3.3.1. Protocol

Studies 1 and 2 concerned the evaluation of EMG feedback when the participants grasped rigid objects; therefore, there was little indication of the force that the prosthesis was applying other than the artificial feedback provided to the participants. However, experiments like this address only a fraction of the activities that a prosthesis must restore for the user, since grasping and manipulating compliant objects is also very common in daily life.

Evidence suggests that incidental cues can benefit prosthesis control [25]. Therefore, it is possible for incidental cues to hamper the objective evaluation of a different feedback scheme during an experiment. For this reason, many experiments employ the occlusion of incidental cues from the participants (e.g., by having them wear noise-cancelling headphones to muffle out the sound of the prosthesis motors or blocking the prosthesis from the participant's field of vision) [96]. Nevertheless, such a configuration does not correspond to a real-life scenario since no user would actively seek to suppress incidental cues from the prosthesis; a user would utilize all available information to control the prosthesis.

Additional artificial feedback is redundant if the information it provides can already be gauged by observing incidental cues [68], [76]. The grasping of compliant objects is a prime example, since the user can receive rich incidental feedback concerning the grasp force simply by observing the movement of the prosthesis in relation to the deformation of the object. Arguably, the actual benefits of a feedback scheme can be evaluated more comprehensively when it is tested in the presence of incidental feedback.

Hence, the efficacy of EMG feedback when grasping compliant objects, a condition characterized by rich incidental feedback, was evaluated in this study. Both EMG and incidental feedback have been shown to offer good force control [22], [25] and they were expected to display similar performance in terms of force success rate. However, as mentioned, EMG feedback provided real-time information about the participants' muscle contractions while also being a more objective measure of the amplitude of the myoelectric commands. Hence, it was hypothesized that EMG feedback would assist the participants in achieving a shorter trial completion time. Additionally, EMG feedback was expected to benefit the participants by allowing them to generate more consistent commands and less variant myoelectric signals.

To explore this hypothesis, the participants were required to perform a modified clothespin task with three pins of increasing stiffness values (yellow, green, and black pins). The participants had to use the prosthesis to open the pins to their full aperture without applying more force than required to do so, i.e., until their handles just touched, as shown in Figure 3-5. The pins were selected so that a distinct force level had to be applied to each one to successfully complete the task, while the application of a larger force would imply the breakage of the hypothetical compliant object, deeming the trial as failed. A smaller force would not be sufficient to fully open the clothespins. The participants performed the experiment with and without EMG feedback.

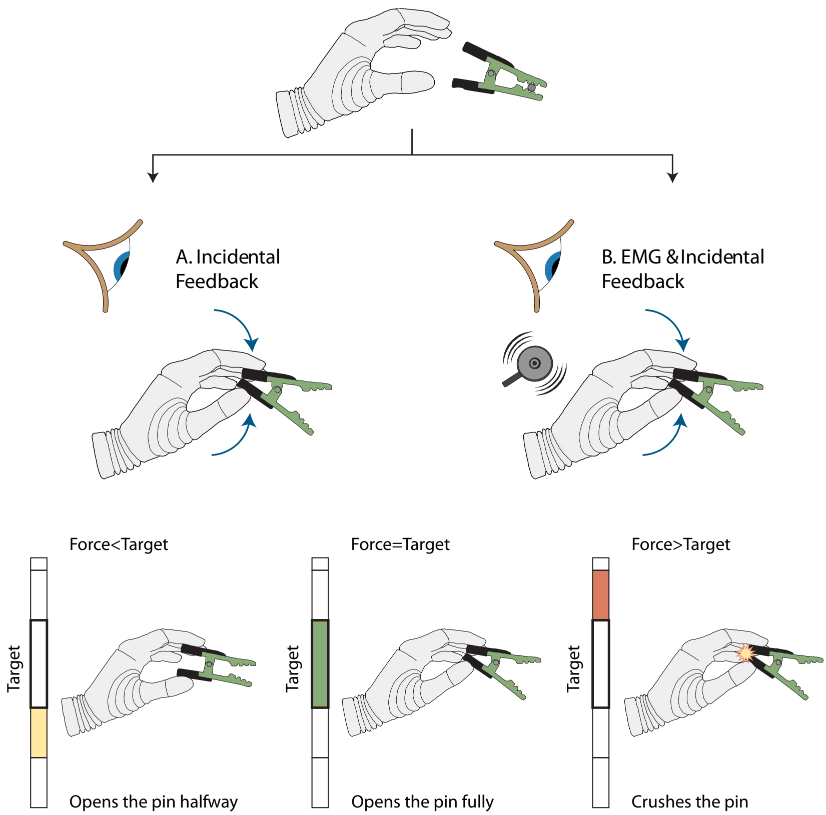

Figure 3-5: The two conditions tested in this experiment. Condition A includes only incidental (visual) feedback, while condition B also integrated EMG feedback in the control loop. The three scenarios for the application of force are shown at the bottom part of the image

In visual terms, the task was translated to bringing the handles of the pins together until they just touched. The instruction of the task, therefore, was to follow a visual cue that would signify the completion of a trial. This visual cue was the only indication for the completion of the task that the participants had when they did not receive EMG feedback. The way that a participant approaches and performs a task has been shown to be highly dependent on the feedback they receive [97].

The performance was evaluated based on the force success rate, the completion time of successful trials, the mean square jerk of the myoelectric signal [98], and the point-by-point variance [18] of the myoelectric signal in successful trials.

3.3.2. Results and Discussion

EMG feedback offered higher success rate (87.6%) overall than incidental feedback (72.2%) (Figure 3-6A). The myoelectric signals generated by the participants were smoother (lower mean square jerk) and more similar across trials (lower point-by-point variance) when EMG feedback was provided. Based on these results, the participants indeed seemed to largely prefer to focus on EMG feedback rather than incidental cues for the correct force application.

An unexpected result was the general lack of difference between the two feedback conditions regarding the trial completion time, which overall ranged approximately between 3 and 4 seconds (Figure 3-6B). Rather surprisingly, when receiving EMG feedback, the participants performed the task of grasping the yellow pin slightly but significantly slower than when they could only rely on incidental cues. The similarity of the completion time in the two conditions can be attributed to the choice of a similar strategy in both feedback conditions, however, it does not imply good quality control in the case of incidental feedback, as evidenced by its low success rate.

Additionally, a stepwise strategy was largely employed by the participants in both conditions when grasping larger pins, wherein they slowly closed the prosthesis onto the pin handles, applied a small force and then proceeded to gradually increase their contraction to open the pin. Notably, if the myoelectric signal

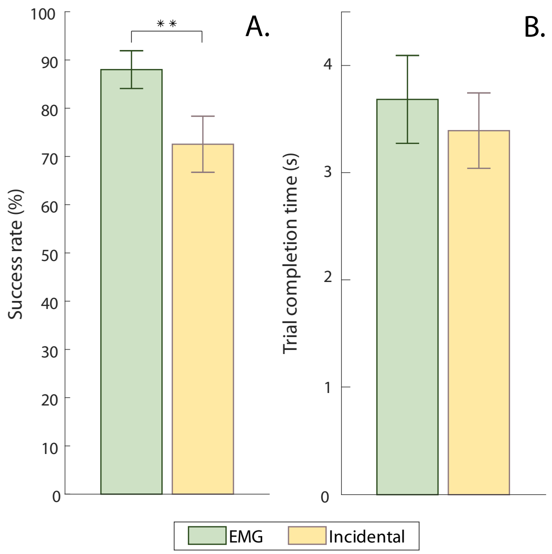

Figure 3-6: The main outcome measures of the third experiment: the success rate (A) and the trial completion time (B).

was not sufficiently high, the movement of the hand would halt with the pin only open halfway. A larger muscle contraction would be necessary to complete the trial in that case.

This choice of strategy indicated that the participants did not make use of the predictive properties of EMG feedback but may explain the difference in performance between the two conditions. When the participants received EMG feedback, they could modulate their muscle contraction based on the feedback, which was also an indicator of the applied force, thus generate a myoelectric signal of appropriate level, which was associated with the force required for each pin. Conversely, the generated force level was unknown to the participants when they only used incidental cues, in which case all they could do was generate myoelectric commands over whose precise amplitude they had limited control, a condition which, evidently, was prone to a larger number of errors.

The higher mean square jerk characterizing the myoelectric signals in the incidental feedback condition signifies the inability of the participants to generate stable and coordinated muscle contractions, relative to the EMG feedback condition. In the latter case, the smoothness of the signals can, again, be attributed to the real-time knowledge of the muscle contraction level, which allowed the participants to restrict their contractions to a desired range.

Furthermore, the enhanced coordination of the muscle contractions and the knowledge of their level through EMG feedback allowed the participants to train themselves to perform similar contractions across trials, thus achieving a lower point-by-point variance. This is a promising result for the quality of the control interface, since a lower variability across trials is indicative of motor skill learning [99].

If the information a feedback scheme communicates can be provided by observing incidental cues, it is rendered redundant with scarce benefits to the control loop [68], [76]. This is a reason to consider testing a supplementary feedback scheme in the presence of incidental cues and this is what this study explored. The results demonstrate the value of EMG feedback, which indeed enriched the information that the participants received from incidental feedback, manifesting in more precise and stable control of the prosthesis.

3.4. Study 4 – Application and Evaluation on a Transhumeral Amputee

3.4.1. Protocol

So far, the studies that comprise this work have dealt exclusively in the application of EMG feedback in configurations intended for transradial amputees, evidenced by the experimental setups, which have utilized EMG signals from the forearm muscles to drive a prosthetic hand and delivered vibrotactile feedback to the upper arm. In these studies, the EMG-feedback-enabled interface has been studied in terms of control optimization, sensitivity to control disturbances, and its ability to refine the grasping of compliant objects.

Artificial feedback has been shown to greatly benefit the control of a prosthesis, while it has been cited as an important part of the rehabilitation process [100]. However, it is clear that the setup must be heavily modified to be applied on users with transhumeral amputations. In their case, as the forearm is absent, there are no muscles directly relevant to hand control that can be used for EMG recording. In summary, the application of an interface similar to the one presented in the previous studies for the control of a hand prosthesis by a transhumeral amputee has to make use of muscles that are unrelated to natural hand control and deliver the vibrotactile feedback to a different body area.

This predicament leads to the research question of this case study, which regarded the ability of the myoelectric interface to be applied to different body areas to enable the control of the prosthesis by a very proximal transhumeral amputee. The study explored the performance of a transhumeral amputee participant using a modified version of the system in several conditions, to ascertain if force control of a myoelectric prosthesis is not just feasible, but also still intuitive and easy to interpret.

A single traumatic right transhumeral amputee was recruited for this case study. In the past, the participant had been using a prosthesis attached to a socket that covered his shoulder, with EMG electrodes embedded within. However, due to discomfort and signal degradation due to heat and perspiration under the socket, he rejected this interface in favor of an osseointegrated implant, on which a prosthesis would be attached (see Figure 3-7). At the time of the experiments, the participant was undergoing rehabilitation to ensure the integration of the implant with his residual humerus, ahead of the attachment of the prosthetic arm.

Four experiments were executed with this participant. Different recording sites were tested, to assess if the participant could effectively control the prosthesis using the muscles pictured in Figure 3-7 (left and right pectoralis major and left flexor carpi radialis). Tactors 1 and 2 were placed over the EMG recording site, as shown in Figure 3-7 (with 2 on top), while tactors 3 and 4 were placed over the participant's upper right back (with 3 on top), positioned similarly to tactors 1 and 2. Care was taken so that no tactors were placed over bones.

The positioning of the electrodes on the pectorals followed a trial-and-error test, wherein different points of the muscle in question were probed to identify one that offered the best signal quality. The flexor carpi radialis was identified with palpation, similar to the previous experiments. Importantly, and in contrast to the previous studies, the participant only had control over hand closing, while opening was operated manually by the experimenter. Finally, the tactors were distributed far enough from each other so that the participant could easily identify which tactor was active at any given moment, especially given the low spatial discrimination ability of the back area [101]. The prosthesis was secured on the table and positioned so that it could freely open and close.

The participant performed a force-matching task similar to those in Studies 1 and 2, wherein he was required to use his myoelectric signal to apply specific force target levels (levels 2, 3, and 4, as they were defined in 2.2.3). Each of the four experimental sessions tested different conditions under which the task was to be executed. The participant's performance was evaluated based on the force success rate, overall and for each target level, while the over- and undershoot rates offered further insight into the control.

Session 1 was an introduction to the concepts of EMG feedback and myoelectric control. The participant's performance was evaluated over a small number of trials, controlling the prosthesis with his right pectoralis major (on the amputated side). Good performance in this session would be promising for the next ones, since it would indicate that the participant could modulate his muscle contractions and correctly interpret the feedback.

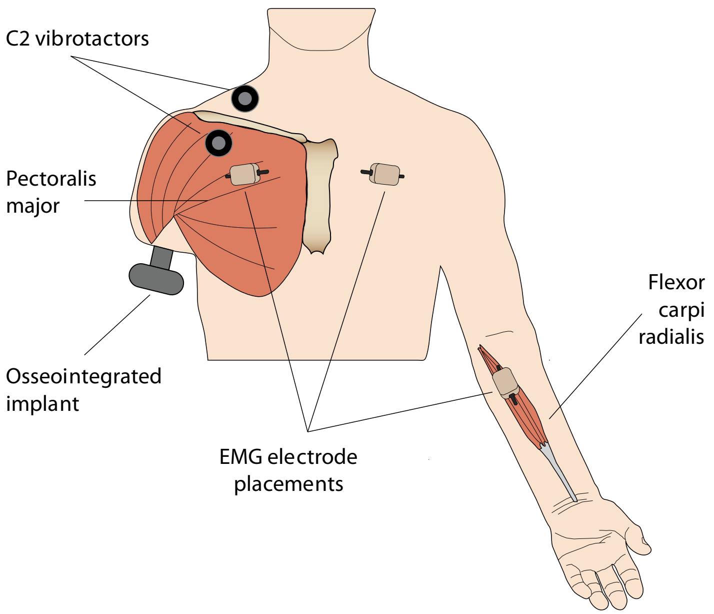

Figure 3-7: Schematic of the experimental setup elements placed on the transhumeral amputee participant

Session 2 aimed to ascertain the efficacy of EMG feedback in this new framework. This was achieved by the participant executing the experiment 3 times: without EMG feedback in the first and third run and with feedback in the second. The hypothesis in this session was that the feedback would benefit the control (which would result in higher success rate in the second run) and that it would have a learning effect on the participant. This learning effect would manifest itself in the third run (when the feedback is deactivated), wherein the success rate was expected to be higher than in the first run.

Sessions 3 and 4 tested the translatability of the myoelectric control to different muscles, while keeping the vibrotactile stimulation site unchanged. As seen in Figure 3-7, the right and left pectorals and the wrist flexor of the left hand were tested. They were tested in this order in Session 3 and reversed in Session 4. The performance of the participant would indicate the efficiency of the interface when applied to different body areas.

3.4.2. Results and Discussion

This case study involved the evaluation of the system seen in Studies 1, 2, and 3, translated to be applied on and controlled by a transhumeral amputee. Despite the recruitment of only one single participant, this study remains a source of valuable insight on the expectations one could have from a system like the one

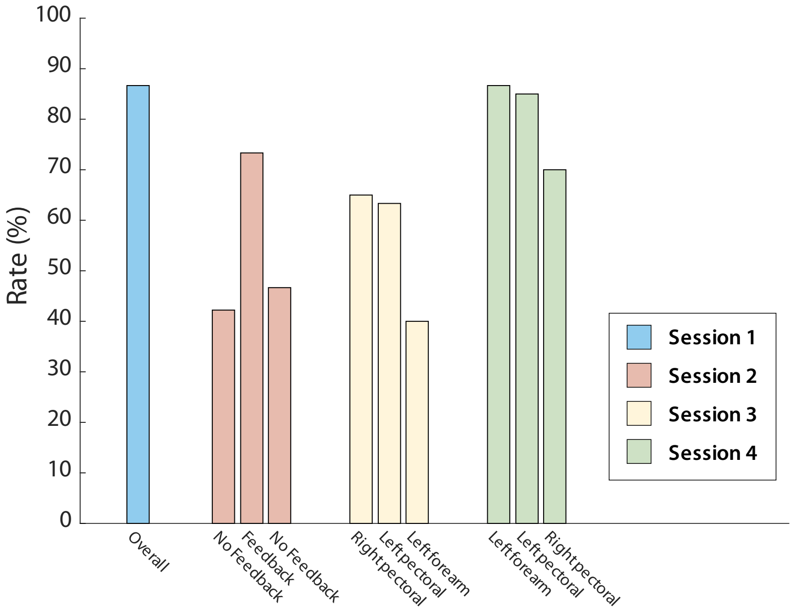

Figure 3-8: The summary overall results of Study 4

tested when applied under different conditions. The summary results are shown in Figure 3-8.

In the first session, the participant was introduced to the myoelectric control interface and the concept of vibrotactile EMG feedback. His performance was evaluated to ascertain the efficacy of the interface. Surprisingly, his success rate (over all target levels) was 86.7%. This exceeded expectations, considering this was the participant's first encounter with EMG feedback, while the system itself had never been used in its translated version, with EMG recorded from the pectoralis major and vibrotactile stimulation delivered around the shoulder area. The performance indicated that the participant could easily interpret EMG feedback and integrate it into his control loop, thus applying the target forces with precision.

Session 2 aimed at establishing the benefits of EMG feedback in the interface control, by having the participant perform three experimental runs, in only the second of which he received EMG feedback. As expected, the overall performance in the second run was 31% higher than in the first (73.3 against 42.2%), highlighting the effect that feedback has on control.

The feedback was expected to have an additional effect, namely that of learning, with the participant developing the ability to control the system in a feedforward manner when executing the task with feedback therefore maintaining a relatively high performance in the third run, where the feedback was deactivated. However, this expectation was not met, with the overall success rate being only 4.4% higher than in the first run. Possibly, his over-reliance on the feedback during the second run impeded any learning effect that the feedback could have provided [102].

Nevertheless, the participant performed a large number of undershoots when targeting level 4 in the third run, which may indicate that he was generating muscle contractions using a more conservative strategy, after receiving feedback in the second run. He likely resorted to this strategy to avoid overshoots; however, in the absence of feedback, this correction was imprecise, leading to 36.7% of level 4 trials failing due to undershoots.

The results of session 3 indicate that the ability of the participant to retain good control of the interface over the time between sessions (1.5 months) was overestimated. The overall success rate with both pectoral muscles was approximately 65% but dropped to 40% when the participant used his forearm muscles to drive the prosthesis. Additionally, trials targeting level 2 were characterized by a disproportionally large number of overshoots, with the participant overshooting all level 2 trials when using his forearm muscles.

A combination of insufficient training to refamiliarize the participant with the setup, the first-time use of two out of three muscles for interface control, as well as unwanted artifacts in the myoelectric signal generated by the participant could explain this striking drop in the participant's performance. These artifacts were mainly overshoots in the myoelectric signal often arising towards the end of the contractions, while the participant was attempting to relax. These artifacts were communicated to the participant via feedback; however, it is clear that he could not suppress them.

For this reason, in session 4, the artifacts detected in the previous session were pointed out to the participant, who was given instructions on how to suppress them. The overall success rate with the left forearm and pectoral was approximately 85% in session 4, while the right pectoral had a 70% success rate (similar to the success rate seen in the second run of session 2). This is an indication that with proper guidance the participant could reintegrate the control scheme and utilize the feedback to apply the desired forces, as well as that the performance of the system is robust to the positioning of the EMG electrodes. Additionally, the positioning of the vibrotactors did not seem to affect the interpretability of the feedback, since the participant utilized it effectively, as evidenced in the results of Session 2.

An interesting observation was that the participant could directly modulate his EMG to the desired interval and therefore directly apply the target force, without resorting to a stepwise strategy (an initial application of a low force and its subsequent gradual increase). This behavior suggested that the participant could utilize the predictive properties of EMG feedback, further evidence of the intuitive nature of the system and the participant's ability to integrate the feedback into his control loop.

In summary, the participant's performance exceeded expectations, particularly when controlling the interface with his right pectoral, considering that the muscle had atrophied due to disuse post amputation. The results suggest that the stimulation transmitting the EMG feedback was intuitive and interpretable for the participant to rapidly and effectively assimilate it into his control loop and enhance his performance.

The performance displayed by the participant was to par with that of able-bodied participants in the previous studies, despite the radical translation of the setup. Therefore, this interface shows promise for flexible application and implementation in myoelectric control systems.

Chapter 4. CONCLUSIONS

4.1. ADDRESSING THE RESEARCH QUESTIONS

The extraordinary complexity of the human hand and its control are really a natural wonder. Its replacement is of paramount importance to amputees, to restore lost functionality and their body image. However, fully restoring the functionality of such a sophisticated organ with our current methods and technology is still an elusive goal.

The restoration of sensory input from a prosthesis is considered to be a desirable feature in modern myoelectric prostheses. Therefore, extensive research has been dedicated to developing various different methods to develop a bidirectional control interface to implement intuitive and effective artificial somatosensory feedback.

The present dissertation contributes to this field by evaluating the function of a proportional myoelectric prosthesis control interface integrated with EMG feedback, the latter being a relatively novel approach of considerable promise as a method to close the control loop. Aspects of the control system were studied and assessed, answering the four main research questions presented in Chapter 1. The summary conclusions of the four studies are represented in Figure 4-1.

Research Question 1: *What signal processing parameter values optimize the control of a prosthesis control system equipped with EMG feedback?*

We identified an optimal parameter pair for the processing of the EMG signal recorded from the user, which allowed them to better utilize the EMG feedback

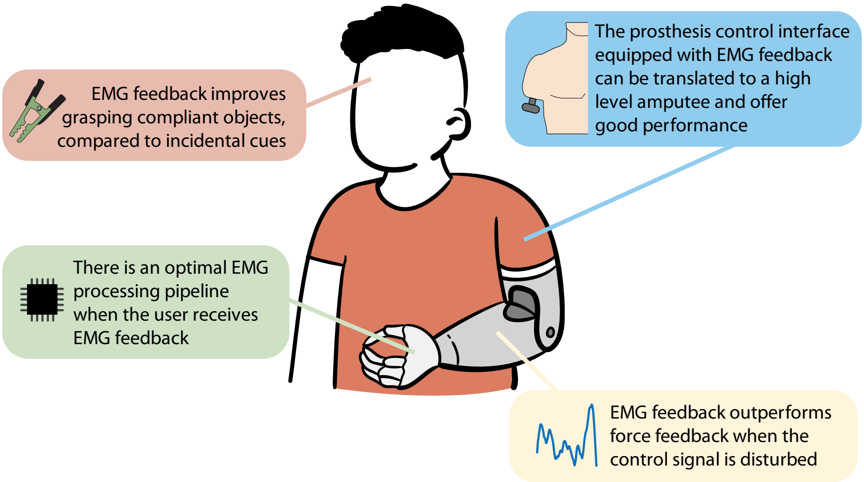

Figure 4-1: The conclusions of the four studies that comprised this work

and achieve specific force target levels with greater precision. This study offered a new set of guidelines for the parametrization of myoelectric control systems of this nature. These guidelines were used in the subsequent studies to select the EMG processing parameters to achieve optimal control.

As mentioned before, the scientific literature, even the studies that are directly related to the application and evaluation of EMG feedback, is characterized by considerable heterogeneity regarding the parametrization of the signal processing pipeline, with little consideration about its effects on the control loop. This study also indicated that a user's ability to exploit EMG feedback and control a prosthesis is highly dependent on the morphology of the myoelectric signal, which, in turn, depends on the signal processing pipeline.

The differences in the performance under the conditions tested in this study has provided valuable insight into the interplay between the parametrization of the control loop and the ability of the participants to effectively utilize EMG feedback and apply a desired grasp force with the prosthesis. Based on this, a more informed decision can be made regarding the selection of the EMG processing parameters.

The results are a step towards the standardization of the calibration of a prosthesis control interface equipped with EMG feedback. Knowledge of the optimal parameters can reduce the time required for pilot testing of a similar control loop in future studies, while, in a clinical setting, it can be used to offer better force control in daily activities, enhancing the user experience and quality of life.

Research Question 2: *Can EMG feedback mitigate the effects of control disturbances and promote robustness in a prosthesis control loop more effectively than force feedback?*

EMG feedback proved to be less sensitive to control disturbances than the more conventional and well-established force feedback, enabling the users to achieve their force targets more reliably, even in a condition of altered system dynamics. As mentioned, this study is the first one to directly compare EMG and force feedback in terms of their ability to mitigate the effect of control disturbances.

These findings further establish EMG feedback as a solid alternative to more conventional feedback approaches. A potential future application of artificial somatosensory feedback in commercial prostheses would benefit greatly from the integration of EMG feedback. The results of this study indicate the robustness of EMG feedback to control disturbances generated in the highly dynamic environment in which the device is expected to operate. Therefore, it can be assumed that disturbance sources such as changes in the skin-electrode interface (e.g., due to sweating) and electrode shifts that could normally hamper prosthesis control will no longer have a significant impact if the system is equipped with EMG feedback.

Additionally, the study establishes a reliable framework for the evaluation of artificial feedback under disturbed conditions. It is the author's hope that this approach will be adopted in future studies in the field.

Research Question 3: *Can EMG feedback improve prosthesis force control even in the presence of rich incidental feedback in the form of visual cues?*

This experiment was a more objective evaluation of EMG feedback, due to the similarity of the experimental conditions with a real-life scenario, thanks to the focus on the effect of incidental cues on the force control and their interplay with EMG feedback. Testing the system in a configuration resembling the daily life usage of the prosthesis offers important insight into its performance.

When presented with grasping a compliant object, a rich source of incidental feedback generated by the deformation of the object, participants showed a clear preference for the information provided by EMG feedback, rather than rely on incidental cues. This highlights the value of EMG feedback, since it implies that the information that it communicated to the participants enhanced and enriched the control loop, enabling precise and stable control.

Considering that incidental feedback has been shown to suffice for satisfactory force control [25], the results of this experiment are even more significant. They can be regarded as a verification of the benefits of EMG feedback in the prosthesis control loop.

Research Question 4: *Does EMG feedback still benefit the control when the control interface configuration is altered for use in a transhumeral amputee?*

So far, this work has shown that a properly calibrated myoelectric control system integrated with EMG feedback can offer good force control, which is maintained in disturbed conditions, while EMG feedback enriches the information obtained from incidental sources. Additionally, the performance of a transradial amputee participant in Studies 2 and 3 followed the trends set by the able-bodied participants, which was also a promising result, as was the fact that the amputee reported that the feedback was intuitive and easy to comprehend. This provided an indication about the use of the system by its target user group, namely transradial amputees, for whom it was initially conceptualized.

In Study 4 the translatability of the interface was demonstrated by fundamentally modifying its configuration and applying it on a very proximal transhumeral amputee participant, who displayed satisfactory performance, comparable to the one seen by participants throughout the previous studies. This participant also pointed out the intuitiveness of the interface and understood the functional principles within the first session without any previous experience or knowledge of similar setups.

Notably, the study yielded these results despite the recorded muscle having no functional relation to the movement that the prosthesis performed and the stimulation being delivered to a skin area farther from the end effector than in the configuration of the first three studies. These two outcomes were extremely promising and are further proof of the fact that EMG feedback was simple to use and can assist the participants in targeting precise force levels, regardless of the configuration of the recording and stimulation locations. Finally, this is an indication of the potential for the translatability of the setup in different body areas, while maintaining a satisfactory performance.

The results of the case study suggest that the features of the system make it suitable for use in different amputee groups, while opening up possibilities for further modifications based on the target application.

4.2. OVERALL CONCLUSIONS

There is now further proof for the value of EMG feedback as a candidate method for closing the proportional myoelectric prosthesis control loop, as it implements a simple, intuitive, and stable control interface, with easily interpretable stimulation. The system is adaptable and translatable to amputee participants, who can use it naturally and with ease, an indication of its potential for clinical applications.

As mentioned, feedback is an often-cited user requitement for prosthetic devices that has, so far, been overlooked in the design of commercial devices. The control of a prosthesis can be enhanced with the introduction of feedback, enhancing the user experience. It also arises from the results that EMG feedback promotes intuitive control and improves the reliability of a prosthesis interface, which, as stated in Section 1.2, are often-cited requirements that users have of their prostheses.

Besides offering good performance, the integration of EMG feedback in prosthesis interfaces also benefits the overall system design. The absence of additional sensors (for force, touch, etc.) that would be necessarily embedded in the prosthesis to generate the feedback signals would simplify the system considerably, decreasing the weight of the device and simplifying the software that controls it, while enhancing force control. This way, an interface equipped with EMG feedback would be more lightweight, simpler to manufacture and program, and potentially more cost effective. Additionally, a more lightweight prosthesis would be more comfortable to wear, thus fulfilling one more cited user requirement.

Therefore, in summary, an overall conclusion that can be drawn from this work is that EMG feedback can benefit many aspects of a prosthetic hand interface, from the system design to the user experience. This is an additional argument for the push for its introduction to commercial devices since the enhanced user experience may be instrumental in the reduction of prosthesis abandonment rates.

It is the author's sincere hope that the evidence presented herein will be instrumental in the advancement of the field of prosthetic hands. The integration of artificial somatosensory feedback in commercial devices is a crucial step in the evolution of such devices, which will offer improved user experience and enhanced quality of life to prosthesis users.

It is important to note that although these were overall the outcomes we expected, some aspects of the results were surprising and would merit further investigation. One major point was the control strategies that the participants largely selected, which deviated from the optimal one, namely the predictive control of force. Hypothetically, the selection of this predictive strategy may also have an effect on the trial completion time, which was somewhat of a confounding factor in Studies 2 and 3. Perhaps, a deeper understanding of the behavioral aspect of motor learning with this particular myoelectric interface will shed light on these questions.

4.3. FUTURE WORK

Regardless of their promise, the results of this study by no means suggest that the implementation of the prosthesis myoelectric closed loop control presented in this work is flawless. The exploration of other aspects of the complex interplay between the user and the control and feedback interface is still a work in progress.

An important step forward would be to compact the system, so that all its components and functions can fit in a prosthetic socket, which, of course, is the vision for the potential commercialization of this system. The relative simplicity of the hardware comprising the interface may facilitate this transition.

A fully mobile setup paves the way to the more comprehensive exploration of its performance. A longitudinal study where amputees use the system in their daily lives can be conducted to cement the long-term benefits of EMG feedback on user experience.

Moreover, a more comprehensive study of the learning effects of EMG feedback can be conducted, to ascertain if this feedback interface can be used for training the feedforward mechanisms of the users. Can it be a useful tool in establishing the user's internal models of the prosthesis interface in the long run?

The evaluation frameworks put forward in this work can be used and expanded upon to assess future iterations of the setup used in the experiments or entirely different configurations of prosthesis control loops.

And lastly, the control interface used in this work can be used for the development of different systems (other than prosthetic hands), such as e.g., myoelectric teleoperation of robotic arms. The integration of EMG feedback into such a system can potentially also benefit its control and performance.

Chapter 5. LITERATURE

Publications:

J. Tchimino, J. L. Dideriksen, and S. Dosen, "EMG feedback outperforms force feedback in the presence of prosthesis control disturbance," Front. Neurosci., vol. 16, no. September, pp. 1–13, Sep. 2022.

J. Tchimino, M. Markovic, J. L. Dideriksen, and S. Dosen, "The effect of calibration parameters on the control of a myoelectric hand prosthesis using EMG feedback," J. Neural Eng., vol. 18, no. 4, 2021.

J. Tchimino, J. L. Dideriksen, and S. Dosen, "EMG feedback improves grasping of compliant objects using a myoelectric prosthesis", 2023 – ***Submitted in the Journal of NeuroEngineering and Rehabilitation, under review***

J. Tchimino, R. L. Hansen, P. H. Jørgensen, J. L. Dideriksen, and S. Dosen, "Application of EMG feedback for hand prosthesis control in high-level amputation: A case study", 2023 – ***Submitted in the Journal of Neural Engineering, under review***

---

[1] N. Das, N. Nagpal, and S. S. Bankura, "A review on the advancements in the field of upper limb prosthesis," *J. Med. Eng. Technol.*, vol. 42, no. 7, pp. 532–545, Oct. 2018.

[2] H. Shahsavari *et al.*, "Upper limb amputation; Care needs for reintegration to life: An integrative review," *International Journal of Orthopaedic and Trauma Nursing*, vol. 38, no. December 2019. Elsevier, p. 100773, Aug-2020.

[3] G. Pomares, H. Coudane, F. Dap, and G. Dautel, "Psychological effects of

traumatic upper-limb amputations," *Orthop. Traumatol. Surg. Res.*, vol. 106, no. 2, pp. 297–300, Apr. 2020.

[4] F. Kristjansdottir, L. B. Dahlin, H.-E. Rosberg, and I. K. Carlsson, "Social participation in persons with upper limb amputation receiving an esthetic prosthesis," *J. Hand Ther.*, vol. 33, no. 4, pp. 520–527, Oct. 2020.

[5] J. S. Hebert and H. Burger, "Return to Work Following Major Limb Loss," in *Handbook of Return to Work*, Springer, Boston, MA, 2016, pp. 505–517.

[6] A. Esquenazi, "Pain management post amputation," in *Pain Management in Rehabilitation*, T. N. Monga and M. Grabois, Eds. New York, NY: Demos Medical Publishing, 2002, pp. 191–202.

[7] M. Espinosa and D. Nathan-Roberts, "Understanding Prosthetic Abandonment," *Proc. Hum. Factors Ergon. Soc. Annu. Meet.*, vol. 63, no. 1, pp. 1644–1648, 2019.

[8] M. González-Fernández, "Development of Upper Limb Prostheses: Current Progress and Areas for Growth," *Arch. Phys. Med. Rehabil.*, vol. 95, no. 6, pp. 1013–1014, 2014.

[9] D. W. Braza and J. N. Yacub Martin, *Upper Limb Amputations*, Fourth Edi. Elsevier Inc., 2018.

[10] M. Malawer and P. Sugarbaker, "Forequarter Amputation," in *Musculoskeletal Cancer Surgery*, Dordrecht: Kluwer Academic Publishers, 2004, pp. 289–298.

[11] J. L. Finch, G. H. Heath, A. R. David, and J. Kulkarni, "Biomechanical Assessment of Two Artificial Big Toe Restorations From Ancient Egypt and Their Significance to the History of Prosthetics," *JPO J. Prosthetics Orthot.*, vol. 24, no. 4, pp. 181–191, Oct. 2012.

[12] K. J. Zuo and J. L. Olson, "The evolution of functional hand replacement: From iron prostheses to hand transplantation," *Can. J. Plast. Surg.*, vol. 22, no. 1, pp. 44–51, 2014.

[13] A. J. Thurston, "Paré and prosthetics: The early history of artificial limbs," *ANZ J. Surg.*, vol. 77, no. 12, pp. 1114–1119, 2007.

[14] M. Hichert, D. A. Abbink, P. J. Kyberd, and D. H. Plettenburg, "High cable forces deteriorate pinch force control in voluntary-closing body-powered prostheses," *PLoS One*, vol. 12, no. 1, pp. 1–13, 2017.

[15] D. S. V. Bandara, J. Arata, and K. Kiguchi, "Towards control of a transhumeral prosthesis with EEG signals," *Bioengineering*, vol. 5, no. 2, pp. 10–16, 2018.

[16] D. Farina *et al.*, "The extraction of neural information from the surface EMG for the control of upper-limb prostheses: Emerging avenues and challenges," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 4, pp. 797–809, 2014.

[17] C. Castellini and P. Van Der Smagt, "Surface EMG in advanced hand prosthetics," *Biol. Cybern.*, vol. 100, no. 1, pp. 35–47, 2009.

[18] P. Mamidanna, J. L. Dideriksen, and S. Dosen, "Estimating speed-accuracy trade-offs to evaluate and understand closed-loop prosthesis interfaces," *J. Neural Eng.*, vol. 19, no. 5, 2022.

[19] P. Mamidanna, J. L. Dideriksen, and S. Dosen, "The impact of objective functions on control policies in closed-loop control of grasping force with a myoelectric prosthesis," *J. Neural Eng.*, vol. 18, no. 5, 2021.

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

[21] A. Sturma *et al.*, "A surface EMG test tool to measure proportional prosthetic control," *Biomed. Eng. / Biomed. Tech.*, vol. 60, no. 3, pp. 207–213, Jan. 2015.

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

[23] S. Dosen, M. Markovic, K. Somer, B. Graimann, and D. Farina, "EMG Biofeedback for online predictive control of grasping force in a myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, pp. 1–13, 2015.

[24] M. Tavakoli, C. Benussi, and J. L. Lourenco, "Single channel surface EMG control of advanced prosthetic hands: A simple, low cost and efficient approach," *Expert Syst. Appl.*, vol. 79, pp. 322–332, Aug. 2017.

[25] M. Markovic, M. A. Schweisfurth, L. F. Engels, D. Farina, and S. Dosen, "Myocontrol is closed-loop control: incidental feedback is sufficient for scaling the prosthesis force in routine grasping," *J. Neuroeng. Rehabil.*, vol.

15, no. 1, p. 81, Dec. 2018.

[26] J. M. Hahne, D. Farina, N. Jiang, and D. Liebetanz, “A novel percutaneous electrode implant for improving robustness in advanced myoelectric control,” *Front. Neurosci.*, vol. 10, no. MAR, pp. 1–10, 2016.

[27] H. A. Dewald *et al.*, “Stable, three degree-of-freedom myoelectric prosthetic control via chronic bipolar intramuscular electrodes: a case study,” *J. Neuroeng. Rehabil.*, vol. 16, no. 1, p. 147, Dec. 2019.

[28] M. Ortiz-Catalan, R. Brånemark, B. Håkansson, and J. Delbeke, “On the viability of implantable electrodes for the natural control of artificial limbs: Review and discussion,” *Biomed. Eng. Online*, vol. 11, no. 1, p. 33, 2012.

[29] A. W. Shehata, E. J. Scheme, and J. W. Sensinger, “Audible Feedback Improves Internal Model Strength and Performance of Myoelectric Prosthesis Control,” *Sci. Rep.*, vol. 8, no. 1, p. 8541, Dec. 2018.

[30] K. D. O’Shaughnessy, G. A. Dumanian, R. D. Lipschutz, L. A. Miller, K. Stubblefield, and T. A. Kuiken, “Targeted Reinnervation to Improve Prosthesis Control in Transhumeral Amputees,” *J. Bone Jt. Surg.*, vol. 90, no. 2, pp. 393–400, Feb. 2008.

[31] T. A. Kuiken *et al.*, “Targeted reinnervation for enhanced prosthetic arm function in a woman with a proximal amputation: a case study,” *Lancet*, vol. 369, no. 9559, pp. 371–380, 2007.

[32] T. A. Kuiken *et al.*, “Targeted muscle reinnervation for real-time myoelectric control of multifunction artificial arms,” *JAMA - J. Am. Med. Assoc.*, vol. 301, no. 6, pp. 619–628, 2009.

[33] B. Ahkami, E. Mastinu, E. J. Earley, and M. Ortiz-Catalan, “Extra-neural signals from severed nerves enable intrinsic hand movements in transhumeral amputations,” *Sci. Rep.*, vol. 12, no. 1, p. 10218, Dec. 2022.

[34] A. E. Schultz and T. A. Kuiken, “Neural Interfaces for Control of Upper Limb Prostheses: The State of the Art and Future Possibilities,” *PM&R*, vol. 3, no. 1, pp. 55–67, Jan. 2011.

[35] N. de la Oliva, X. Navarro, and J. del Valle, “Time course study of long-term biocompatibility and foreign body reaction to intraneural polyimide-based implants,” *J. Biomed. Mater. Res. - Part A*, vol. 106, no. 3, pp. 746–757, 2018.

[36] P. Svensson, U. Wijk, A. Björkman, and C. Antfolk, "A review of invasive and non-invasive sensory feedback in upper limb prostheses," *Expert Rev. Med. Devices*, vol. 14, no. 6, pp. 439–447, 2017.

[37] M. Ortiz-Catalan, B. Håkansson, and R. Brånemark, "An osseointegrated human-machine gateway for long-term sensory feedback and motor control of artificial limbs," *Sci. Transl. Med.*, vol. 6, no. 257, p. 257re6, Oct. 2014.

[38] H. E. M. A. Abbady *et al.*, "3D-printed prostheses in developing countries: A systematic review," *Prosthetics Orthot. Int.*, vol. 46, no. 1, pp. 19–30, 2022.

[39] C. Dally, D. Johnson, M. Canon, S. Ritter, and K. Mehta, "Characteristics of a 3D-printed prosthetic hand for use in developing countries," *Proc. 5th IEEE Glob. Humanit. Technol. Conf. GHTC 2015*, pp. 66–70, 2015.

[40] S. B. Godfrey *et al.*, "The SoftHand Pro: Functional evaluation of a novel, flexible, and robust myoelectric prosthesis," *PLoS One*, vol. 13, no. 10, p. e0205653, Oct. 2018.

[41] M. Laffranchi *et al.*, "The Hannes hand prosthesis replicates the key biological properties of the human hand," *Sci. Robot.*, vol. 5, no. 46, Sep. 2020.

[42] C. Cipriani, M. Controzzi, and M. C. Carrozza, "The SmartHand transradial prosthesis," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, p. 29, 2011.

[43] S. Salminger *et al.*, "Current rates of prosthetic usage in upper-limb amputees – have innovations had an impact on device acceptance?," *Disabil. Rehabil.*, vol. 0, no. 0, pp. 1–12, Dec. 2020.

[44] B. Peerdeman *et al.*, "Myoelectric forearm prostheses: State of the art from a user-centered perspective," *J. Rehabil. Res. Dev.*, vol. 48, no. 6, p. 719, 2011.

[45] L. C. Smail, C. Neal, C. Wilkins, and T. L. Packham, "Comfort and function remain key factors in upper limb prosthetic abandonment: findings of a scoping review," *Disabil. Rehabil. Assist. Technol.*, vol. 16, no. 8, pp. 821–830, 2021.

[46] E. A. Biddiss and T. T. Chau, "Upper limb prosthesis use and abandonment," *Prosthetics Orthot. Int.*, vol. 31, no. 3, pp. 236–257, Sep. 2007.

[47] E. Biddiss, D. Beaton, and T. Chau, “Consumer design priorities for upper limb prosthetics,” *Disabil. Rehabil. Assist. Technol.*, vol. 2, no. 6, pp. 346–357, 2007.

[48] D. Farina *et al.*, “Toward higher-performance bionic limbs for wider clinical use,” *Nat. Biomed. Eng.*, May 2021.

[49] C. Antfolk, M. D’Alonzo, B. Rosén, G. Lundborg, F. Sebelius, and C. Cipriani, “Sensory feedback in upper limb prosthetics,” *Expert Rev. Med. Devices*, vol. 10, no. 1, pp. 45–54, Jan. 2013.

[50] F. Cordella *et al.*, “Literature review on needs of upper limb prosthesis users,” *Front. Neurosci.*, vol. 10, no. MAY, pp. 1–14, 2016.

[51] K. Østlie, I. M. Lesjø, R. J. Franklin, B. Garfelt, O. H. Skjeldal, and P. Magnus, “Prosthesis rejection in acquired major upper-limb amputees: A population-based survey,” *Disabil. Rehabil. Assist. Technol.*, vol. 7, no. 4, pp. 294–303, 2012.

[52] J. M. Hahne, M. Markovic, and D. Farina, “User adaptation in Myoelectric Man-Machine Interfaces,” *Sci. Rep.*, vol. 7, no. 1, pp. 1–10, 2017.

[53] C. Dietrich *et al.*, “Sensory feedback prosthesis reduces phantom limb pain: Proof of a principle,” *Neurosci. Lett.*, vol. 507, no. 2, pp. 97–100, 2012.

[54] G. Chai, X. Sui, S. Li, L. He, and N. Lan, “Characterization of evoked tactile sensation in forearm amputees with transcutaneous electrical nerve stimulation,” *J. Neural Eng.*, vol. 12, no. 6, 2015.

[55] P. Haggard, “Sensory Neuroscience: From Skin to Object in the Somatosensory Cortex,” *Curr. Biol.*, vol. 16, no. 20, pp. R884–R886, Oct. 2006.

[56] E. N. Marieb and K. Hoehn, *Human Anatomy & Physiology*, 9th ed. Pearson Education Inc., 2013.

[57] M. Schiefer, D. Tan, S. M. Sidek, and D. J. Tyler, “Sensory feedback by peripheral nerve stimulation improves task performance in individuals with upper limb loss using a myoelectric prosthesis,” *J. Neural Eng.*, vol. 13, no. 1, p. 16001, 2015.

[58] A. Ninu, S. Dosen, S. Muceli, F. Rattay, H. Dietl, and D. Farina, “Closed-Loop Control of Grasping With a Myoelectric Hand Prosthesis: Which Are the Relevant Feedback Variables for Force Control?,” *IEEE Trans. Neural Syst.*

Rehabil. Eng., vol. 22, no. 5, pp. 1041–1052, Sep. 2014.

[59] H. P. Saal and S. J. Bensmaia, "Biomimetic approaches to bionic touch through a peripheral nerve interface," *Neuropsychologia*, vol. 79, pp. 344–353, 2015.

[60] L. O. Tierradentro-García, J. S. Botero-Meneses, and C. Talero-Gutiérrez, "The Sound of Jacqueline du Pré: Revisiting her Medical and Musical History," *Mult. Scler. J. - Exp. Transl. Clin.*, vol. 4, no. 2, p. 205521731877575, Apr. 2018.

[61] J. Cole, *Pride and a daily marathon*, 1st ed. Cambridge, MA: MIT Press, 1995.

[62] B. Geng, J. Dong, W. Jensen, S. Dosen, D. Farina, and E. N. Kamavuako, "Psychophysical evaluation of subdermal electrical stimulation in relation to prosthesis sensory feedback," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 3, pp. 709–715, 2018.

[63] S. Raspopovic, G. Valle, and F. M. Petrini, "Sensory feedback for limb prostheses in amputees," *Nat. Mater.*, vol. 20, no. 7, pp. 925–939, 2021.

[64] J. W. Sensinger and S. Dosen, "A Review of Sensory Feedback in Upper-Limb Prostheses From the Perspective of Human Motor Control," *Front. Neurosci.*, vol. 14, no. June, pp. 1–24, 2020.

[65] J. S. Schofield, K. R. Evans, J. P. Carey, and J. S. Hebert, "Applications of sensory feedback in motorized upper extremity prosthesis: a review," *Expert Rev. Med. Devices*, vol. 11, no. 5, pp. 499–511, Sep. 2014.

[66] G. Chai, H. Wang, G. Li, X. Sheng, and X. Zhu, "Electrotactile Feedback Improves Grip Force Control and Enables Object Stiffness Recognition While Using a Myoelectric Hand," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 30, pp. 1310–1320, 2022.

[67] A. M. De Nunzio *et al.*, "Tactile feedback is an effective instrument for the training of grasping with a prosthesis at low- and medium-force levels," *Exp. Brain Res.*, vol. 235, no. 8, pp. 2547–2559, Aug. 2017.

[68] E. J. Earley, K. J. Kaveny, R. E. Johnson, L. J. Hargrove, and J. W. Sensinger, "Joint-based velocity feedback to virtual limb dynamic perturbations," *IEEE Int. Conf. Rehabil. Robot.*, pp. 1313–1318, 2017.

[69] M. A. Garenfeld, C. K. Mortensen, M. Strbac, J. L. Dideriksen, and S. Dosen, "Amplitude versus spatially modulated electrotactile feedback for

myoelectric control of two degrees of freedom," *J. Neural Eng.*, vol. 17, no. 4, 2020.

[70] A. Erwin and F. C. Sup, "A Haptic Feedback Scheme to Accurately Position a Virtual Wrist Prosthesis Using a Three-Node Tactor Array," *PLoS One*, vol. 10, no. 8, p. e0134095, Aug. 2015.

[71] C. Cipriani, J. L. Segil, F. Clemente, R. F. Richard, and B. Edin, "Humans can integrate feedback of discrete events in their sensorimotor control of a robotic hand," *Exp. Brain Res.*, vol. 232, no. 11, pp. 3421–3429, 2014.

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

[73] E. J. Earley, R. E. Johnson, J. W. Sensinger, and L. J. Hargrove, "Joint speed feedback improves myoelectric prosthesis adaptation after perturbed reaches in non amputees," *Sci. Rep.*, vol. 11, no. 1, pp. 1–11, 2021.

[74] D. Yang, Y. Gu, N. V Thakor, and H. Liu, "Improving the functionality, robustness, and adaptability of myoelectric control for dexterous motion restoration," *Exp. Brain Res.*, vol. 237, no. 2, pp. 291–311, Feb. 2019.

[75] L. F. Engels, A. W. Shehata, E. J. Scheme, J. W. Sensinger, and C. Cipriani, "When less is more-discrete tactile feedback dominates continuous audio biofeedback in the integrated percept while controlling a myoelectric prosthetic hand," *Front. Neurosci.*, vol. 13, no. JUN, pp. 1–10, 2019.

[76] M. A. Gonzalez, C. Lee, J. Kang, R. B. Gillespie, and D. H. Gates, "Getting a Grip on the Impact of Incidental Feedback From Body-Powered and Myoelectric Prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 29, pp. 1905–1912, 2021.

[77] O. Barzilay and A. Wolf, "A fast implementation for EMG signal linear envelope computation," *J. Electromyogr. Kinesiol.*, vol. 21, no. 4, pp. 678–682, Aug. 2011.

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

[79] R. M. Rangayyan, *Biomedical Signal Analysis*, 2nd ed. New Jersey: John Wiley & Sons, 2015.

[80] T. Y. Fukuda *et al.*, "Root mean square value of the electromyographic

signal in the isometric torque of the quadriceps, hamstrings and brachial biceps muscles in female subjects," *J. Appl. Res.*, vol. 10, no. 1, pp. 32–39, 2010.

[81] K. T. Özgünen, U. Çelik, and S. S. Kurdak, "Determination of an optimal threshold value for muscle activity detection in EMG analysis," *J. Sport. Sci. Med.*, vol. 9, no. 4, pp. 620–628, 2010.

[82] L. Vargas, H. Huang, Y. Zhu, and X. Hu, "Evoked Tactile Feedback and Control Scheme on Functional Utility of Prosthetic Hand," *IEEE Robot. Autom. Lett.*, vol. 7, no. 2, pp. 1308–1315, 2022.

[83] M. Yoshida and M. Terao, "Suitable cutoff frequency of low-pass filter for estimating muscle by surface electromyogram," in *Proceedings of the 25th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (IEEE Cat. No.03CH37439)*, 2003, vol. 2, no. October 2003, pp. 1709–1711.

[84] K. E. Jones, A. F. d. C. Hamilton, and D. M. Wolpert, "Sources of signal-dependent noise during isometric force production," *J. Neurophysiol.*, vol. 88, no. 3, pp. 1533–1544, 2002.

[85] M. Markovic *et al.*, "The clinical relevance of advanced artificial feedback in the control of a multi-functional myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, p. 28, Dec. 2018.

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

[87] S. Gilman, "Joint position sense and vibration sense: anatomical organisation and assessment," *J. Neurol. Neurosurg. Psychiatry*, vol. 73, no. 5, pp. 473–477, Nov. 2002.

[88] T. D. Sanger, "Bayesian Filtering of Myoelectric Signals," *J. Neurophysiol.*, vol. 97, no. 2, pp. 1839–1845, 2007.

[89] M. Sartori, G. Durandau, S. Došen, and D. Farina, "Robust simultaneous myoelectric control of multiple degrees of freedom in wrist-hand prostheses by real-time neuromusculoskeletal modeling," *J. Neural Eng.*, vol. 15, no. 6, p. 066026, Dec. 2018.

[90] S. Muceli and D. Farina, "Simultaneous and proportional estimation of hand kinematics from EMG during mirrored movements at multiple

degrees-of-freedom," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 3, pp. 371–378, 2012.

[91] S. M. Radhakrishnan, S. N. Baker, and A. Jackson, "Learning a Novel Myoelectric-Controlled Interface Task," *J. Neurophysiol.*, vol. 100, no. 4, pp. 2397–2408, 2008.

[92] J. Barnes, M. Dyson, and K. Nazarpour, "Comparison of hand and forearm muscle pairs in controlling of a novel myoelectric interface," in *2016 IEEE International Conference on Systems, Man, and Cybernetics (SMC)*, 2016, pp. 002846–002849.

[93] L. Sörnmo and P. Laguna, *Bioelectrical Signal Processing in Cardiac and Neurological Applications*. Elsevier, 2005.

[94] C. Prahm *et al.*, "Counteracting Electrode Shifts in Upper-Limb Prosthesis Control via Transfer Learning," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 27, no. 5, pp. 956–962, 2019.

[95] S. Dosen *et al.*, "Building an internal model of a myoelectric prosthesis via closed-loop control for consistent and routine grasping," *Exp. Brain Res.*, vol. 233, no. 6, pp. 1855–1865, 2015.

[96] M. A. Wilke, C. Niethammer, B. Meyer, D. Farina, and S. Dosen, "Psychometric characterization of incidental feedback sources during grasping with a hand prosthesis," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, p. 155, Dec. 2019.

[97] H. Bouwsema, C. K. van der Sluis, and R. M. Bongers, "Effect of Feedback during Virtual Training of Grip Force Control with a Myoelectric Prosthesis," *PLoS One*, vol. 9, no. 5, p. e98301, May 2014.

[98] N. Hogan and D. Sternad, "Sensitivity of Smoothness Measures to Movement Duration, Amplitude, and Arrests," *J. Mot. Behav.*, vol. 41, no. 6, pp. 529–534, Nov. 2009.

[99] L. Shmuelof, J. W. Krakauer, and P. Mazzoni, "How is a motor skill learned? Change and invariance at the levels of task success and trajectory control," *J. Neurophysiol.*, vol. 108, no. 2, pp. 578–594, Jul. 2012.

[100] M. Guémann *et al.*, "Sensory substitution of elbow proprioception to improve myoelectric control of upper limb prosthesis: experiment on healthy subjects and amputees," *J. Neuroeng. Rehabil.*, vol. 19, no. 1, pp. 1–12, 2022.

[101] K. Myles and M. S. Binseel, "The Tactile Modality: A Review of Tactile Sensitivity and Human Tactile Interfaces," *Proc. - 12th Int. Symp. Haptic Interfaces Virtual Environ. Teleoperator Syst. HAPTICS*, 2007.

[102] R. Sigrist, G. Rauter, R. Riener, and P. Wolf, "Augmented visual, auditory, haptic, and multimodal feedback in motor learning: A review," *Psychon. Bull. Rev.*, vol. 20, no. 1, pp. 21–53, 2013.

ISSN (online): 2246-1302
ISBN (online): 978-87-7573-724-6

AALBORG UNIVERSITY PRESS