The Effect of Upper Limb Prosthesis Type on Functional Outcomes and Satisfaction

by

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Dedication

For my family, whose support made this work possible.

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Preface

Chapters 2-6 have been written as separate manuscripts for publication. As such, there may be some repetition of content between chapters.

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Abstract

Despite the significant functional limitations imposed by upper limb loss, little research has focused on quantifying the functional success and satisfaction of prosthesis users. Most existing evidence comes from surveys, rather than experimental outcomes. Without a quantitative baseline, it is difficult to know where to focus attention for improvement in future prosthesis designs or to demonstrate whether new designs offer advantages over existing technologies. Thus, the goal of this dissertation was to quantify how functional outcomes and satisfaction relate to the type of prosthesis used.

The first aim was to determine if prosthesis type affects embodiment, or the extent to which prosthesis users perceived their prosthesis to be part of their body. We quantified embodiment for body-powered (BP) and myoelectric (MYO) prosthesis users based on a survey and two objective measurements of body schema and peripersonal space. Although BP users reported a stronger sense of agency over their prostheses in comparison to MYO users, other measures did not consistently differentiate experiences of embodiment based on prosthesis type. However, measurements of body schema varied depending on the cause of limb loss.

The second aim was to determine if prosthesis type impacts movement quality during activities of daily living. As an initial step for this aim, we quantified the reliability of movement quality metrics (three measures of smoothness and one measure of straightness) in healthy adults performing a variety of different tasks. Based on these findings, we then compared movement quality in BP and MYO prosthesis users during a subset of tasks (moving a can from a low shelf to a high shelf, placing a pill in a pillbox, and placing a pushpin in a bulletin board) using the

metrics that had the highest reliability. All movements were slower when performed with MYO prostheses, except for the reaching phase of the pill task. Object manipulation movements were consistently less smooth when performed with MYO prostheses. However, differences in curvature of the reaching movements between the prosthesis types varied across tasks.

The third aim was to determine if prosthesis type affects kinematic compensations during activities of daily living. We quantified lateral lean, axial rotation, and flexion of the trunk during the same three activities of daily living. The range of motion was greater in all directions for BP prostheses during each task—except axial rotation and flexion during the pin task, which were greater for MYO prostheses.

The fourth aim was to explore the factors associated with interest in noninvasive (myoelectric) and invasive (targeted muscle reinnervation, peripheral nerve interfaces, cortical interfaces) interfaces for prosthesis control. An online survey collected opinions from 232 individuals with upper limb loss on the interfaces. Relationships between interest in the interfaces and demographics, limb loss characteristics, and prosthesis use history were defined using bivariate analysis and logistic regression. There was increased interest in the invasive interfaces among individuals who were younger, had unilateral limb loss, or had acquired limb loss.

Taken together, these aims suggest that BP prostheses may promote embodiment and smooth movement, while MYO prostheses may minimize compensatory movement. Although emerging prosthesis technologies requiring surgical intervention may not be accepted by all individuals with upper limb loss, functional outcomes with these technologies should be compared to outcomes with existing BP and MYO prostheses to demonstrate the relative merits of each design.

CHAPTER 1. Introduction

Major upper limb loss is a debilitating condition that affects approximately 41,000 individuals in the U.S. (Ziegler-Graham et al. 2008). Given that the majority of these individuals are young and otherwise healthy (Esquenazi and Meier 1996), this population will require care for many years to ensure active and productive lives. Unfortunately, around 25% of individuals with upper limb loss ultimately choose not to use a prosthesis (Biddiss and Chau 2007b), which is partially due to the limited availability of different prosthesis choices.

Upper limb prostheses are broadly classified as passive or active devices. Passive prostheses appear cosmetically similar to the missing limb but do not offer functional movement. Active prostheses permit functional movement using control signals derived externally from the prosthesis user's body. Body-powered (BP) prostheses use a harness and Bowden cable to link body movements (typically glenohumeral flexion or scapular abduction) to movement of a terminal device. Myoelectric (MYO) prostheses rely on electromyographic signals recorded from the skin surface over muscles in the residual limb to actuate the terminal device. Myoelectric control commonly relies on a "direct" control scheme in which signals from an agonist/antagonist pair of muscles are used to control a single degree of freedom in the prosthesis (Wurth and Hargrove 2014).

BP prostheses are thought to have functional advantages over MYO prostheses in terms of durability, training time, maintenance, and sensory feedback, while MYO prostheses offer improved cosmesis and reduction in phantom limb pain. However, there is little empirical evidence to suggest that either device provides a significant advantage over the other (Carey et al. 2015, Carey et al. 2017). A recent review paper found that out of 31 articles offering insight on this topic, only 9 involved laboratory-based or clinical assessments (Carey et al. 2015). It is difficult to draw general conclusions from such a small body of literature, especially given the considerable diversity in the sample populations and experimental methods between studies (Biddiss and Chau 2007c, Carey et al. 2015).

Nonetheless, there may be functional differences between BP and MYO prostheses resulting from the way in which the prosthesis interfaces with the user. The human-prosthesis interface has an effect on the feedforward and feedback control systems involved in goal-directed reaching movements (Metzger et al. 2010). The feedforward control system predicts motor commands based on prior movements and depends on an internal model of the arm that accounts for its inertial properties. Reaching may be affected if an individual has not adequately updated their internal model to account for the new inertial properties of their prosthetic limb. The feedback control system corrects for errors during movement based on exteroceptive and proprioceptive input. The availability of this sensory information may differ between BP and MYO prostheses, leading to potential functional differences.

1.1 Sensory feedback in upper limb prostheses

The flow of sensory information in prosthesis users involves three different pathways (Childress 1973). The first pathway includes visual and auditory signals (Type-A), while the second pathway includes somatic sensory (e.g., vibration, temperature, touch) and proprioceptive information (Type-B). Type-C feedback is intrinsic to the prosthesis itself and does not require involvement from the user (e.g., automatic slip detection). The degree to which each type of feedback is available likely depends on prosthesis design.

Type-B feedback promotes extended physiologic proprioception (EPP), which means that the body's natural physiological sensors are used to understand the state of the prosthesis (Simpson 1974). Since Type-B feedback comes directly from the BP control interface (Bowden cable and harness system), a BP prosthesis user receives information about the prosthesis in the same channel through which the prosthesis is controlled (Weir 2003, Antfolk et al. 2013). Specifically, physiologic sensors of the controlling joint relay information about the prosthesis joint position (or terminal device aperture), velocity or forces. The modality of information used to operate the prosthesis is therefore similar to the modality of information used to control a natural limb (Doubler and Childress 1984). This availability of Type-B feedback in BP prostheses may offer a functional advantage through the provision of EPP. For example, healthy controls using BP prosthesis simulators have improved performance on compliance-discrimination tasks when cable-based force feedback is provided (Brown et al. 2016) and on tracking tasks when an EPP-based position control scheme is used (Doubler and Childress 1984, Doubler and Childress 1984b).

Conversely, MYO devices cannot directly offer EPP. The motors in MYO prostheses are velocity-controlled (output speed is directly proportional to input voltage), requiring the user to integrate velocity in order to control position (Weir 2003). This requires constant monitoring of the prosthesis using other feedback sources. Indeed, studies of visuomotor behavior in upper limb prosthesis users have shown that the gaze is fixed on the hand or the area of the object to be grasped for the majority of task completion time (Bouwsema et al. 2012, Sobuh et al. 2014). Auditory and incidental Type-B feedback (e.g., vibration from the motor, socket pressures) may also be available (Schofield et al. 2014), but this information must be consciously interpreted and does not promote EPP. The cognitive demand on the user is also elevated (Gonzalez et al. 2012, Antfolk et al. 2013), although this does not necessarily negate the utility of the feedback. For example, some MYO users develop sensitivities to vibrations from the motor and soft touches to the socket that act as a functional aid (Sörbye 1980, Wijk and Carlsson 2015). Nonetheless, these adaptive strategies are likely developed slowly over a long period of time, and may never be fully developed by some users.

1.2 Embodiment of upper limb prostheses

Increasingly, researchers are exploring the potential clinical benefits of promoting a sense of embodiment over assistive technologies (Pazzaglia and Molinari 2016), including upper limb prostheses. Experiencing a prosthesis as an integrated part of one's own body, or embodiment, is driven by the interaction of tactile, visual, and proprioceptive signals (Giummarra et al. 2008) that are spatially and temporally congruent (Botvinick and Cohen 1998). Restoring sensory feedback to individuals with upper limb loss has important implications for this reason (Di Pino et al. 2014). Indeed, sensory restoration has long been a priority for the research community (Schofield et al. 2014), as well as prosthesis users themselves (Biddiss et al. 2007, Kyberd and Hill 2011).

To this end, considerable effort has focused on trying to induce the experience of embodiment in prosthesis users through the application of appropriate sensory stimulation (Ehrsson et al. 2008, Marasco et al. 2011, Marini et al. 2014, Rognini et al. 2018). However, prosthesis users are capable of embodying their prostheses in the absence of such interventions. This may be due the fact that some degree of sensory input is still available through the prosthetic interface, including proprioceptive input (via the body-powered harness) and vibrations or pressure transmitted through the socket (Wijk and Carlsson 2015). Additionally, visual feedback that is synchronous with one's voluntary actions to control an external object can lead to an "agency-driven" embodiment of that object (Kalckert and Ehrsson 2012), suggesting that a sense of agency over the prosthesis may contribute to embodiment as well.

Embodiment in upper limb prosthesis users in the absence of intervention has been documented using several different methodologies, including interviews (Murray 2004, Murray 2008, Wijk and Carlsson 2015), questionnaires (Imaizumi et al. 2016) and behavioral measures (McDonnell et al. 1989, Canzoneri et al. 2013a, Gouzien et al. 2017). However, it is currently unclear whether the experience of embodiment differs depending on whether a BP or MYO prosthesis is used. Since embodiment depends on the integration of multisensory inputs and the availability of sensory input depends on prosthesis design, embodiment may also depend on prosthesis design. Specifically, people may embody BP prostheses more strongly than MYO prostheses due to the availability of both proprioceptive and visual feedback. However, this has yet to be tested experimentally.

1.3 Assessing upper limb prosthesis use

Ideally, an upper limb prosthesis will become incorporated into the user's daily life and promote independence in activities of daily living. The prosthesis should help the user achieve natural motor patterns in order to reduce biomechanical stress on the intact limb, energy expenditure, cognitive load, and possible overreliance on adaptive equipment (Smurr et al. 2008). This means that movements of the prosthesis should be smooth, coordinated, and accurate while evoking minimal compensatory motion (Smurr et al. 2008).

Evaluating a prosthesis user's progress towards achieving these objectives is an important component of both rehabilitation and research, so the need to objectively measure performance is widely recognized (Resnik et al. 2013a). However, there is currently no standard approach for applying outcome measures, making it difficult to communicate and interpret results across the rehabilitation community (Hill et al. 2009). Experts in the upper limb prosthetics field have recommended an approach based on the World Health Organization's International Classification of Functioning, Disability and Health, which defines a framework for the definition and measurement of health and disability (World Health Organization 2002). In the context of upper limb prosthesis use, this framework indicates that outcome measures should address 1) the technical capabilities of the prosthesis itself, 2) the prosthesis user's ability to function in daily life, and 3) the subjective impact of a prosthesis on the user's life (Hill et al. 2009). It is difficult to assess all three components using a single outcome measure, so it is necessary to use several for a thorough evaluation (Metcalf et al. 2007).

Unfortunately, few outcome measures have been validated for use in adult upper limb prosthesis users. A review paper (Wright 2009) identified only four psychometrically sound outcome measures that are intended for application to prosthesis users: the Assessment for Capacity of Myoelectric Control (ACMC), Southampton Hand Assessment Procedure (SHAP), Trinity Assessment of Prosthesis Experience Survey (TAPES), and the Upper Extremity Functional Scale (UEFS) from the Orthosis and Prosthesis Users Survey. More recently, psychometric properties have been validated for additional tests, including the Box and Blocks Test (BBT) (Resnik and Borgia 2012), Jebsen-Taylor Test of Hand Function (JTHF) (Resnik and Borgia 2012), Activities Measure for Adults with Upper Limb Amputation (AM-ULA) (Resnik et al. 2013a), Brief Activities Measure for Adults with Upper Limb Amputation (BAM-ULA) (Resnik et al. 2018), and the University of New Brunswick Test of Prosthetic Function (UNB) (Resnik et al. 2013b). However, these assessments have significant limitations. Several measures are based on self-reported responses to questionnaires (TAPES, UEFS), making them vulnerable to bias. Other measures are scored on visual assessment of task performance by a trained observer (AM-ULA, BAM-ULA, ACMC, UNB), and may be affected by the observer's experience or bias (de los Reyes-Guzmán et al. 2014). The remaining measures use completion time as the outcome

(SHAP, JTHF). These are more objective, but are too narrow in scope to fully describe a prosthesis user's functionality.

1.3.1 Kinematic outcomes

In contrast, outcome measures that are derived from movement kinematics are able to offer detailed, unbiased information that reflects the underlying motor strategies associated with the movement (de los Reyes-Guzmán et al. 2014). These outcomes may be applied to a variety of movements, including targeted reaching movements (i.e., point to point movements with defined starting and ending positions) and functional activities of daily living. Although assessments often require subjects to complete the activities of daily living using real props, many assessments use simplified versions of the tasks, which allow subjects to merely simulate the movements. Simulated activities of daily living have been shown to involve different movement patterns than functional tasks, however (Taylor et al. 2018).

A variety of kinematic outcomes have been used to assess prosthesis use. For example, joint angle patterns have been evaluated in a number of studies (Carey et al. 2008, Carey et al. 2009, Hebert and Lewicke 2012, Metzger et al. 2012, Major et al. 2014, Hussaini and Kyberd 2017, Hussaini et al. 2017). Although these studies have consistently demonstrated that prosthesis users increase proximal joint movements to compensate for reduced range of motion in the distal prosthetic joints, only Carey (2009) and Hebert (2012) directly compared between BP and MYO prostheses.

Movement quality metrics are also useful for revealing performance deficits in prosthesis users. In healthy individuals, movements are smooth (Rohrer et al. 2002) and straight with symmetrical velocity profiles (Flash and Hogan 1985), and demonstrate tight temporal coupling of reach and grasp (Jeannerod 1984). In contrast, movement of prosthetic limbs tend to be less smooth (Fraser and Wing 1981, Doeringer and Hogan 1995, Bouwsema et al. 2010, Cowley et al. 2017) and more curved (Cowley et al. 2017), with asymmetric velocity profiles (Bouwsema et al. 2010) and decreased coupling between reach and grasp movements (Bouwsema et al. 2010). This decreased movement quality is likely a consequence of impairments in the feedforward and feedback control systems governing prosthetic limbs. The extent to which movement quality is affected by prosthesis type (BP or MYO) is currently unknown.

1.3.2 Reliability of kinematic outcomes

It is important to understand the test-retest reliability of these kinematic outcomes so that meaningful trends in the data can be distinguished from experimental errors or natural variability in subject performance (Schwartz et al. 2004). Reliability has already been quantified in patient and healthy populations for upper limb joint angles (Caimmi et al. 2008, Levanon et al. 2010, Aizawa et al. 2013, Caimmi et al. 2015, Engdahl and Gates 2018) and a variety of movement quality metrics (Caimmi et al. 2008, Wagner et al. 2008, Schneiberg et al. 2010, Osu et al. 2011, Patterson et al. 2011, Caimmi et al. 2015). However, these studies have included a limited number of tasks and varying definitions of what constitutes a reliable movement. Further investigation of measurement reliability is needed to improve interpretation of kinematic outcomes.

1.4 Novel upper limb prostheses

Since high prosthesis abandonment rates are commonly reported in the literature (Biddiss and Chau 2007b), it is clear that user needs are not being met with current prosthesis options. There is also a weak correlation between prosthesis wearing time and satisfaction with functional ability (Davidson 2002), suggesting that even individuals who frequently use a prosthesis can be dissatisfied with its performance. Even if there are comparative functional advantages between BP and MYO prostheses, it cannot be denied that both devices have considerable shortcomings in comparison to a natural limb. Most current commercially available prostheses permit a single degree of freedom (open/close) (Graimann and Dietl 2013, Resnik et al. 2014a), making performance of daily tasks unintuitive and cumbersome. Consequently, individuals with upper limb loss report a desire for prostheses with improved dexterity, including independent control of the joints, increased range of motion, and wider variety of grasp patterns (Biddiss et al. 2007, Kyberd and Hill 2011).

Although fully-articulated prosthetic hands exist, neither BP nor direct MYO control schemes are sufficient to activate all of the available degrees of freedom. Alternative control schemes, particularly those that interface directly with the nervous system, are needed. A variety of myoelectric control strategies have been proposed to avoid direct control (Farina et al. 2014), including pattern recognition algorithms in which specific features are extracted from the recorded muscle activity and used to control different degrees of freedom in the prosthesis (Ajiboye and Weir 2005, Mattioli et al. 2011, Scheme et al. 2014). Mode-switching (e.g., through co-contraction of an agonist/antagonist muscle pair) is another way to increase the number of degrees of freedom that can be controlled from the same recording sites (Wurth and Hargrove 2014).

Targeted muscle reinnervation may also be used in conjunction with a MYO prosthesis. This procedure involves surgical relocation of peripheral nerves to residual muscles (such as the pectoralis major) in order to create additional surface recording sites for myoelectric control (Miller et al. 2008, Kuiken et al. 2009). The number of new recording sites that can be created is limited, however, because the entire nerve is used to reinnervate a muscle.

This limitation can be addressed by interfacing more directly with the nervous system. For example, neural signals can be recorded directly from the peripheral nervous system following implantation of recording electrodes in the residual limb. These electrodes can be placed around (Sahin and Durand 1998) or within (Clark et al. 2011) the nerve. Similarly, electrodes can placed on (Chestek et al. 2013) or within (Hochberg et al. 2012) the motor cortex to record from the central nervous system. Because these approaches record from the nervous system rather than from the muscle, they may offer a higher degree of specificity (Kung et al. 2013) and can be used to collect a high volume of independent control signals.

Many of these approaches are surgically invasive and carry some degree of medical risk. In a survey to explore interest in both noninvasive and invasive technologies among individuals with upper limb loss, most individuals (83%) expressed interest in non-invasive MYO control and comparatively fewer individuals (≥ 39%) were interested in invasive approaches (Engdahl et al. 2015). However, it is unclear what factors influenced whether an individual was interested in invasive technology. Understanding these factors can help guide the development of prostheses to specifically benefit those who are most likely to accept the technology. This is true for prostheses that use both novel and conventional (BP and MYO) control methods.

1.5 Summary of dissertation

Although abandonment of upper limb prostheses is common, there is little quantitative evidence on how prosthesis designs need to be improved. Most knowledge regarding the functional shortcomings of upper limb prostheses is self-reported by prosthesis users, or is extrapolated based on outcomes that lack specific quantitative details. It is important to quantify how prosthesis design choices relate to function and acceptance, so that future prosthesis designs are well-informed. The goal of this dissertation was to relate functional outcomes and satisfaction with a prosthesis to the type of prosthesis that is used.

The first aim was to determine if prosthesis type affects embodiment. We used one self-reported and two behavioral measures of embodiment to quantify the extent to which BP and MYO prosthesis users perceived their prosthesis to be part of their body. This work tested the hypothesis that BP prostheses are embodied more strongly than MYO prostheses due to greater availability of sensory feedback.

The second aim was to determine if prosthesis type impacts movement quality during activities of daily living. First, we quantified the reliability of movement quality metrics in healthy adults during activities of daily living. These findings suggested that unconstrained tasks can reliably be used to assess movement quality, providing justification for using these methods with prosthesis users. Consequently, we compared movement quality in BP and MYO prosthesis users during activities of daily living using measures of duration, smoothness, and straightness. This work tested the hypothesis that movement quality is diminished in MYO prosthesis users compared to BP prosthesis users.

The third aim was to determine if prosthesis type affects kinematic compensations during activities of daily living. Kinematic patterns were quantified as the range of motion required by the trunk to complete a task. This work offers insight on the comparative advantages of each prosthesis type in terms of compensatory movement.

The fourth aim was to explore the factors associated with interest in noninvasive (myoelectric) and invasive (targeted muscle reinnervation, peripheral nerve interfaces, cortical interfaces) interfaces for prosthesis control. An online survey collected opinions on four types of interfaces from 232 individuals with upper limb loss. Relationships between interest in the interfaces and demographics, limb loss characteristics, and prosthesis use history were defined using bivariate analysis and logistic regression. Outcomes from this work offer insight on which individuals might accept novel interfaces and which individuals might prefer to continue using BP or MYO prostheses.

CHAPTER 2. Differential experiences of embodiment between body-powered and myoelectric prosthesis users

2.1 Abstract

Prosthesis embodiment, or the perception of a prosthesis as one's own body, may be an important component of the rehabilitation process. The purpose of this study was to determine embodiment depends on the type of prosthesis that is used. Embodiment was quantified three ways in a group of six transradial body-powered (BP) and myoelectric (MYO) prosthesis users. First, we assessed ownership and agency over the prosthesis using a survey. Second, we assessed body schema using a limb length estimation procedure. Third, we assessed peripersonal space through a tactile extinction paradigm. BP users reported a stronger sense of agency over their prostheses in comparison to MYO users, but the other measures did not consistently differentiate experiences of embodiment based on prosthesis type. Nonetheless, the experience of embodiment is diverse and differences between BP and MYO prostheses might be detectable using other methodologies. Further exploration of this question is warranted.

2.2 Introduction

An important component of rehabilitation for individuals with upper limb loss is provision of a prosthesis, with the goal of replacing the missing limb to the fullest extent possible. Those who are involved with the rehabilitation process believe that a prosthesis user may benefit from perceiving the prosthesis as part of their body, rather than an auxiliary tool (Scarry 1994). This "embodiment" of a prosthesis is a complex phenomenon that encompasses many different characteristics, which makes it difficult to define succinctly. One proposed definition is that an object is embodied only if some of its properties are processed in the same way as the properties of one's own body (de Vignemont 2011). This definition includes spatial (if the space surrounding the object is processed as body space), motor (if the object moves like a body part and is perceived to be under one's control), and affective (if the same affective reactions are shown towards the object as towards one's body) components. Although the presence of all three components would indicate "full embodiment", an object can be embodied even if only some of these components are present. Similarly, an object might be embodied in some situations but not in others (de Vignemont 2011).

Converging evidence in the literature suggests that individuals with upper limb loss can embody their prosthesis (e.g., (McDonnell et al. 1989, Canzoneri et al. 2013a, Imaizumi et al. 2016)) or can be induced to embody a prosthesis through provision of appropriate sensory stimulation (e.g., (Ehrsson et al. 2008, Marasco et al. 2011, Rognini et al. 2018)). Embodiment is dependent on the interaction between afferent and efferent signals (Pazzaglia and Molinari 2016). Congruence between tactile, visual, and proprioceptive signals (Giummarra et al. 2008) that are easily interpretable and concordant with a sense of agency is particularly important. The process of embodiment occurs as the brain extracts statistical correlations from multisensory inputs to create the perception that the information is arriving from a single plausible spatiotemporal source (i.e., the embodied object) (Armel and Ramachandran 2003). For example, experiments involving the rubber hand illusion have shown that when afferent signals are veridical and similar to normal physiological feedback, embodiment is more likely to occur (Di Pino et al. 2014).

Embodiment of a prosthesis also becomes more likely when a prosthesis user perceives that this sensory information comes from the interface between the residual limb, the prosthesis, and the surrounding environment (Mills 2013). Sensory feedback from a prosthesis can be delivered to the user through visual/auditory or proprioceptive/somatic pathways (Childress 1973). In body-powered (BP) prostheses, movement of a terminal device is activated by body movements (glenohumeral flexion or scapular abduction) via a harness and Bowden cable system. Because the state of the physiological limb is mechanically linked to the state of the prosthetic limb, the user receives information about the state of the prosthesis through the same physiological pathways that are used to activate the prosthesis (Weir 2003, Antfolk et al. 2013). The resulting sense of extended physiological proprioception in the prosthetic limb (Simpson 1974) may minimize the conscious attention needed to control the prosthesis.

In contrast, the terminal device of a myoelectric (MYO) prosthesis is actuated by a battery-driven motor controlled using surface electromyography signals recorded from the residual limb. The motors are typically velocity-controlled (output speed is directly proportional to input voltage), which requires the user to integrate velocity in order to control position (Weir 2003). Because of this, the user must constantly monitor the prosthesis visually. Indeed, studies of visuomotor behavior in upper limb prosthesis users have shown that the gaze is fixed on the hand or the area of the object to be grasped for the majority of task completion time (Bouwsema et al. 2012, Sobuh et al. 2014). However, visual feedback involves slower reaction times compared to tactile feedback (Nelson et al. 1990) and also must be consciously interpreted placing a higher cognitive demand on the user (Gonzalez et al. 2012). Auditory and incidental somatic feedback (e.g., vibration from the motor, socket pressures) from the prosthesis is also available, but may not be accessible to all MYO users (Sörbye 1980, Wijk and Carlsson 2015).

2.2.1 Methods for quantifying embodiment

Currently, there is no consensus on the most effective method to use for assessing the degree to which a prosthesis user embodies their device. Both empirical and phenomenological methodology can be found in the literature (Mills 2013). While empirical research uses objective outcomes to describe how prosthesis embodiment affects physiology and behavior, phenomenological research uses interviews and questionnaires to describe the subjective experience of embodiment. In particular, phenomenology explores the personal, social, and cultural implications of prosthesis use (Murray 2008). Phenomenology generates expansive and detailed descriptions of prosthesis embodiment, but does not provide the concrete outcome measures required for an empirical understanding of the experience (Longo et al. 2008). Accordingly, the two approaches are complimentary. Several examples of phenomenological and empirical methods are described below.

2.2.1.1 Phenomenology

Phenomenology has been used previously to identify common themes in the experience of prosthesis embodiment. Among others, these themes include decreased awareness of the prosthesis over time, perceptual integration of the phantom and prosthetic limbs, and viewing the prosthesis as a corporeal structure rather than a tool (Murray 2004). Importantly, these experiences occur within broader social and cultural contexts that moderate an individual's relationship with their prosthesis (Murray 2008).

Other studies have used phenomenology to supplement their empirical findings. For example, the level of integration of the prosthesis was assessed using a questionnaire and then correlated with estimates of the maximum reachable distance achievable with the prosthetic and intact limbs (Gouzien et al. 2017). Another study explored whether embodiment could be induced using cutaneous touch in individuals who had undergone targeted muscle reinnervation surgery (Marasco et al. 2011). Questionnaires and self-reported experiences were used in conjunction with more physiological measures of residual limb temperature and temporal order judgements. Similarly, experiences of embodiment have been compared between frequent and infrequent prosthesis users on the basis of both postural control metrics and questionnaires (Imaizumi et al. 2016).

2.2.1.2 Body schema

The body schema is an adaptable representation of the dimensions of one's body parts. Since the body schema is primarily maintained through proprioceptive and tactile feedback, it is continuously (and unconsciously) updated during movement in response to the changing sensory inputs. It is well established in the literature that tool-use changes the body schema such that it expands to include the tool (Martel et al. 2016). Given that a prosthesis is essentially a tool (i.e. it changes the dimensions of the body in an attempt to replace the missing limb), it stands to reason that prosthesis use would also cause changes to the body schema. Indeed, similarities in the kinematic profile of intact and prosthetic limbs (Fraser 1984) and overestimation of residual limb length when wearing a prosthesis (McDonnell et al. 1989) both indicate that prostheses can be incorporated into the body schema. Similarly, wearing a prosthesis increases the perceived length of the residual limb during a tactile distance perception task (Canzoneri et al. 2013a) and causes prosthesis users to overestimate how far they can reach with the prosthetic limb (Gouzien et al. 2017).

2.2.1.3 Peripersonal space

The peripersonal space is a region around the body that is characterized by extensive multisensory integration between visual, tactile, and auditory inputs from the body and the area immediately surrounding the body (Cardinali et al. 2009). Peripersonal space seems to be a consequence of bimodal neurons that respond to both tactile stimuli delivered within the tactile receptive field and to visual stimuli delivered near the same region. Although the visual receptive field is linked to the tactile receptive field in this way, the visual receptive field is also quite malleable. A seminal study by Iriki et al. demonstrated that the visual receptive fields related to the hand peripersonal space in monkeys could be elongated towards the tip of a tool following training (Iriki et al. 1996). The tool seemed to be included in the visual receptive field for several minutes following training, at which point the visual receptive field returned to its original size. Importantly, the tool had to be actively used (rather than passively held) for the visual receptive field to expand.

Humans demonstrate a similar malleability through a phenomenon known as extinction. Following brain damage, some individuals are unable to perceive contralesional tactile stimuli applied to the hand when visual stimuli are simultaneously applied to the ipsilesional hand (Di Pellegrino and De Renzi 1995), such that the tactile stimulus is "extinguished". If the visual stimulus is applied far from the ipsilesional hand (~30 cm), the extinction effect is greatly diminished. However, use of a handheld tool that extends the length of the arm strengthens the extinction effect even when the visual stimulus is applied far from the hand (Farnè and Làdavas 2000, Maravita et al. 2001). Thus, peripersonal space in humans seems to be spatially dynamic and can expand to include a tool.

Extinction is a consequence of the "winner-takes-all" function of the attentional systems within the parietal lobe (Bonato 2012). Contralesional and ipsilesional stimuli must compete for attentional resources when presented simultaneously, so the ipsilesional stimulus tends to dominate due to the presence of brain damage (Maravita et al. 2002). Although a strong extinction effect might not be expected in prosthesis users (in the absence of brain damage), it is possible that differences attentional demand between the intact and prosthetic limbs could contribute to a milder form. A lack of awareness of the prosthetic limb may allow a visual stimulus applied to the intact limb to dominate over a tactile stimulus applied to the prosthetic side. As with tool use in stroke patients, this effect may vary as a function of distance if the peripersonal space expands to include an embodied prosthesis.

2.2.2 Current approach

Since prosthesis embodiment originates from the integration of multisensory inputs, but the availability of sensory input differs based on prosthesis design, it is possible that the extent of embodiment differs with prosthesis design as well. In particular, BP prostheses may be embodied more strongly than MYO prostheses since they offer proprioceptive feedback. The purpose of this work was to compare the experiences of embodiment between BP and MYO prosthesis users using a phenomenological approach, as well as objective measurements of the body schema and peripersonal space. We hypothesized that all three methods would reveal stronger embodiment of BP prostheses than MYO prostheses.

2.3 Methods

2.3.1 Subjects

We recruited six adults with unilateral transradial limb loss were recruited through the University of Michigan Orthotics and Prosthetics Center (Table 2.1). Each participant was required to have at least 6 months of experience using a prosthesis. Age- and sex-matched controls without upper limb loss were recruited from an online database (https://umhealthresearch.org/). Exclusion criteria for both groups included history of other serious musculoskeletal, neurological, or visual impairments. All participants provided written informed consent to participant in this study approved by the University of Michigan.

2.3.2 Embodiment survey

Participants completed a subset of a survey about prosthesis embodiment developed (Imaizumi et al. 2016), including four questions about ownership of the prosthesis and three questions about the sense of agency over the prosthesis. All questions were scored on a Likert scale, where higher scores reflect an increased sense of ownership or agency. The ownership questions and the agency questions were averaged into separate composite scores. If applicable, participants completed the survey separately for their BP and MYO prostheses.

Table 2.1. Participant characteristics.

Controls				Prosthesis Users									
ID	Age	Sex	BMI (kg/m^2)	ID	Age	Sex	BMI (kg/m^2)	Cause of Limb Loss	Affected Limb	Prosthesis Type	Time of Prosthesis Ownership	Weekly Prosthesis Use	Daily Prosthesis Use
C01	34	M	22.7	P01	32	M	23.0	Trauma	Right	BP (voluntary open split hook)	10 months	6 days	6 hours
										MYO (iLimb)	5 months	5 days	5 hours
C02	48	F	36.2	P02	52	F	35.8	Trauma	Right	BP (voluntary open split hook)	7 months	6 days	8 hours
C03	57	F	22.2	P03	55	F	30.0	Congenital	Right	MYO (single DoF hand)	33 years	7 days	2 hours
C04	63	F	23.7	P04	66	F	24.4	Congenital	Right	MYO (bebionic)	6 months	6 days	3 hours
C05	23	F	20.4	P05	26	F	25.2	Congenital	Left	MYO (bebionic)	10 months	5 days	8 hours
C06	29	M	22.7	P06	29	M	23.2	Trauma	Right	BP (voluntary open split hook)	24 months	7 days	10 hours

2.3.3 Limb length estimation

Participants completed a limb length estimation task (McDonnell et al. 1989) in which one limb was placed inside an opaque tube (length: 91.4 cm, diameter: 13.5 cm) until it made contact with a fixed plate (Figure 2.1A). Using the opposite limb, participants moved a sliding indicator along the outside of the tube to the position where they perceived the end of their hidden limb. Control participants performed the task with their dominant and non-dominant limbs (Figure 2.1B), while prosthesis users completed it with their intact and prosthetic limbs. The prosthesis users also performed the procedure with the arm affected by limb loss. When the prosthesis was worn (Figure 2.1C), participants indicated where they perceived the end of the prosthesis (P-PT) and the end of the residual limb (P-RL). When the prosthesis was not worn (Figure 2.1D), participants indicated where they perceived the end of the residual limb (NP-RL) and where they imagined end of the prosthesis would be if they were wearing it (NP-PT). Participants were asked to keep their fingers fully extended and touch the plate with their fingertips. However, some of the prosthesis users had trouble with this position either because it was difficult to fit the hand inside the tube or because the fingers passively flexed when touching the plate. In these cases, the prosthesis users were asked to make a fist instead. For prosthetic hooks or hands without finger extension, participants instead touched the plate with the most distal part of the terminal device.

The plate was fixed at 10 randomly-chosen locations between 5 cm and the length of the participant's limb (Table 2.2). The limb length estimation error was the difference between the plate position and the indicated position.

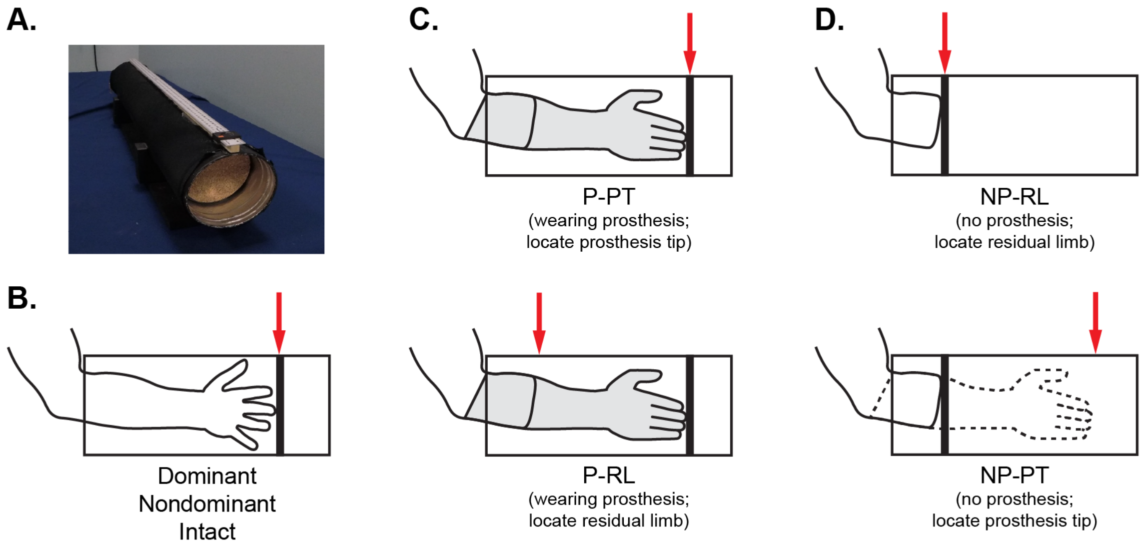

Figure 2.1. Conditions for the limb length estimation task.
Participants performed the limb length estimation task by placing their arm inside an opaque tube with a sliding indicator affixed to the exterior (A). They performed this task using their anatomical limbs (B), their prosthetic limb (C), and their residual limb (D). For each condition, they estimated (red arrows) where they perceived the end of their intact limb, prosthesis (P-PT), residual limb (P-RL, NP-RL), or where the prosthesis would be if they were wearing it (NP-PT).

Table 2.2. Limb lengths for prosthesis users.
Limb lengths for prosthesis users, measured from the lateral epicondyle of the humerus to the tip of the middle finger (if applicable) or the most distal point of the residual limb or terminal device.

ID	Intact Limb Length (cm)	Residual Limb Length (cm)	Limb Length Including BP Prosthesis (cm)	Limb Length Including MYO Prosthesis (cm)
P01	46.2	22.5	44.0	44.9
P02	33.6	15.9	35.4	n/a
P03	40.6	14.9	n/a	36.0
P04	43.5	11.0	n/a	33.4
P05	38.4	8.8	n/a	28.5
P06	48.0	12.0	38.0	n/a

2.3.4 Tactile extinction

2.3.4.1 Experimental set-up

Participants sat at a table with the arms spaced shoulder width apart, fingers fully extended, and palms facing down (Figure 2.2). Three green 10 mm LEDs were placed on the side of the intact arm (dominant arm for controls) at a distance of 40 cm beyond the fingertips (Far LED), at the fingertips (Mid LED), and at the elbow (Near LED). Gaze fixation points were located along the midline of the body at a distance of 20 cm beyond the fingertips (Far Fixation) and halfway along the length of the forearm (Near Fixation). The entire experimental set-up was contained within a three-sided box made of black foam to reduce visual distractions from the surrounding environment.

2.3.4.2 LED thresholding

In order to set the LED intensities to a level that was minimally detectable, each participant was presented with a series of ten 300 ms flashes and asked to respond when they saw a flash by pressing a foot pedal. The flashes were separated by a randomly-chosen interval (three to five seconds) of quiet time. If the participant detected fewer than eight of the flashes, the intensity was increased and testing was repeated until they detected eight or more flashes on two consecutive bouts of testing. During testing, participants were asked to focus on one of the fixation points. We repeated the thresholding procedure for four combinations of LEDs and fixation points (Near LED with Near Fixation, Mid LED with Near Fixation, Mid LED with Far Fixation, and Far LED with Far Fixation).

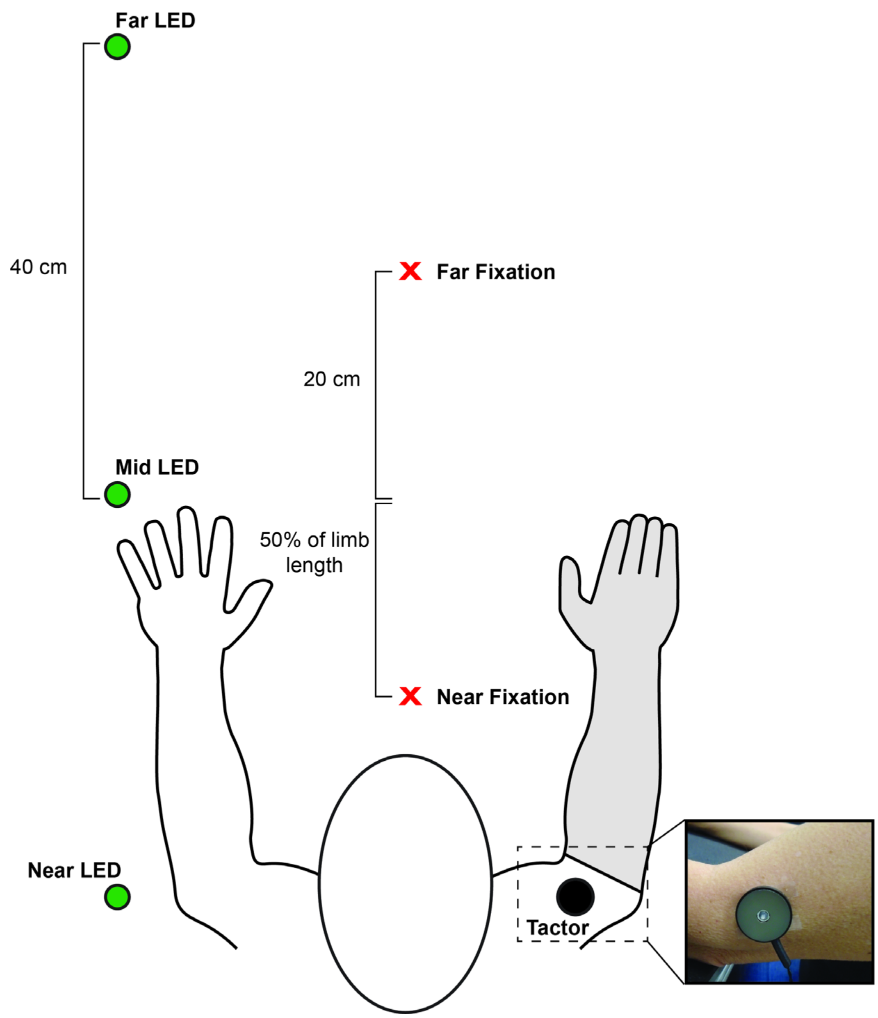

Figure 2.2. Set-up for the tactile extinction protocol.
A series of three green LEDs were place on the side of the intact arm (dominant arm for controls). A tactor was placed on the prosthetic side (nondominant arm for controls) approximately 0.5 inch proximal to the lateral epicondyle of the humerus (inset). This tactor consisted of a small actuator surrounded by passive housing that could be applied directly to the skin.

2.3.4.3 Tactile stimulation protocol

A commercially available tactor (C-2 tactor, Engineering Acoustics, Inc., Casselberry, FL) was placed on the arm affected by limb loss (non-dominant arm for controls). The tactor vibrated against the skin with a 250 Hz sinusoidal signal, resulting in a localized tactile stimulus. The center of the tactor was placed directly on the skin approximately 0.5 inch proximal to the lateral epicondyle of the humerus (Figure 2.2 inset). If the socket trim line or the triceps cuff of the body-powered harness obstructed this area, the tactor was placed more proximally and/or medially on the nearest unobstructed skin.

Participants completed the tactile extinction protocol a total of 40 times (Table 2.3). Each repetition was performed at a randomly chosen LED/fixation position with a randomly chosen tactor intensity (10 intensities x 4 positions = 40 conditions). The LED intensities were always set to a constant value (150% of the final values derived from the thresholding procedure). The tactor intensity varied between 5% and 95% of system output in increments of 10%. Each condition involved 16 trials presented in random order. On eight trials, a 300 ms flash from an LED was presented in conjunction with a 225 ms pulse from the tactor (visual/tactile trials). The start of the tactor pulse was delayed by 50 ms from the start of the LED flash. There were also four trials with only visual cues and four with only tactile cues. Each trial was separated by a random period of rest (three to five seconds). Thus, each participant was presented with 640 trials (16 trial x 40 conditions). However, P01, P02, C01, and C02 completed an earlier version of this protocol and were presented with only two visual trials and two tactile trials at each condition, for a total of 480 trials.

Table 2.3. Selection of conditions for the tactile extinction protocol.
The tactile extinction protocol involved 40 conditions. A condition is composed of one randomly selected LED/fixation point combination (out of 4 possibilities) and one randomly selected tactor intensities (out of 10 possibilities). A total of 16 trials were presented at each condition, including 8 visual/tactile trials (VT), 4 visual only trials (V), and 4 tactile only trials (T).

Condition (choose from each column)		**Trial (repeat all)**
LED/fixation point	**Tactor intensity**	
Near LED with Near Fixation	5%	Visual/tactile (8 trials)
Mid LED with Near Fixation	15%	Visual only (4 trials)
Mid LED with Far Fixation	25%	Tactile only (4 trials)
Far LED with Far Fixation	35%	
	45%	
	55%	
	65%	
	75%	
	85%	
	95%	

Participants pressed a foot pedal when they felt the tactor. They were informed that the LEDs would be flashing and instructed to ignore them while keeping their gaze on the designated fixation point. However, they were not told that the LEDs would be flashing in conjunction with the tactor pulses, or that the tactor intensity would be changed between conditions.

Response accuracy for each condition was the percent of tactor pulses detected by the participant out of the total number of pulses delivered. For each of the four LED/fixation point combinations, we calculated a moving average (window size=3) of the visual/tactile and tactile only response accuracy across the 10 tactor intensities. A cumulative normal function was fitted to each resulting curve as a function of tactor intensity (Table A.1 in Appendix A). We then determined the inflection point of the curve and calculated the difference between inflection points for the visual/tactile and tactile only curves for each of the four LED/fixation point combinations. A negative difference reflects an extinction effect (i.e., the visual/tactile curve is shifted to the right of the tactile only curve).

2.3.5 Statistical analysis

We assessed the magnitude of difference in outcomes between BP and MYO prosthesis users using Hedges' g as a measure of effect size:

$$g \coloneqq \frac{\overline{x}_1 - \overline{x}_2}{s^*}\left(1 - \frac{3}{4(n_1 + n_2) - 9}\right) \qquad \text{(Equation 2.1)}$$

$$s^* \coloneqq \sqrt{\frac{(n_1 - 1){s_1}^2 + (n_2 - 1){s_2}^2}{n_1 + n_2 - 2}}$$

where s* was the pooled standard deviation weighted for sample size. Effect sizes are considered small for $g \geq 0.2$, medium for $g \geq 0.5$, and large for $g \geq 0.8$ (Cohen 1988). Only comparisons with effect sizes ≥ 0.2 are reported.

For prosthesis users, we calculated Pearson correlations between error on the limb length estimation task and the composite ownership and agency scores from the survey, as well as daily length of prosthesis use.

2.4 Results

2.4.1 Survey

On average, BP users reported slightly higher ownership ($g = 0.28$) and agency ($g = 0.81$) scores compared to MYO users (Figure 2.3).

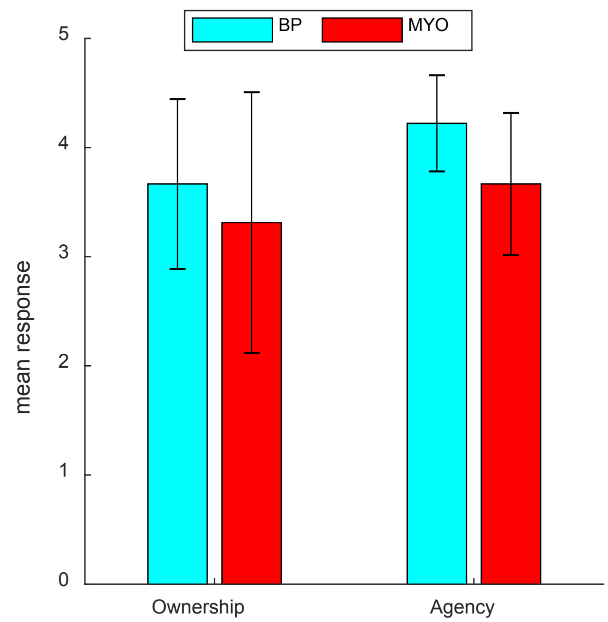

Figure 2.3. Self-reported embodiment.
Mean responses to the ownership and agency questions from the embodiment survey.

2.4.2 Limb length estimation

Estimation error was small (Figure 2.4A) for all conditions involving the anatomical limb (dominant: -1.4 ± 2.1 cm; nondominant: 0.6 ± 0.6 cm; intact: -2.0 ± 0.9 cm). There were few consistent differences in estimation error between BP and MYO prostheses. On average, MYO users underestimated their prosthesis length while wearing it (P-PT) but BP users overestimated (MYO mean: -2.3 ± 4.9 cm; BP mean: 2.5 ± 1.9 cm; $g = 1.01$). All prosthesis users overestimated their residual limb length while wearing their prosthesis (P-RL), but the error magnitude was higher for BP users than MYO users (MYO mean: 7.3 ± 7.1 cm; BP mean: 10.3 ± 2.0 cm; $g = 0.45$). Similarly, all prosthesis users overestimated their residual limb length when not wearing their prosthesis (NP-RL), but the error magnitude was higher for BP users than MYO users (MYO mean: 2.4 ± 3.1 cm; BP mean: 5.7 ± 1.9 cm; $g = 1.03$). Finally, all prosthesis users underestimated their prosthesis length while not wearing it (NP-RL), but the error magnitude was higher for MYO users than BP users (MYO mean: -4.8 ± 4.5 cm; BP mean: -2.1 ± 3.3 cm; $g = 0.58$).

However, these trends may be related to cause of limb loss for some conditions (Figure 2.4B). All of the BP users had acquired limb loss, while three of the four MYO users had congenital limb loss. Notably, two of the MYO users with congenital limb loss had worn their prostheses for less than one year (P04 and P05) and were quite accurate at estimating their residual limb length while wearing their prosthesis (mean: 1.55 ± 2.1 cm). In contrast, the other MYO user with congenital limb loss (P03) had used her prosthesis for 33 years and overestimated more significantly (mean: 10.1 ± 2.1 cm). However, all three participants with congenital limb loss accurately estimated their residual limb length when not wearing their prosthesis (mean: 0.9 ± 0.9 cm), while participants with acquired limb loss overestimated (mean: 6.0 ± 1.7 cm). Furthermore, all three MYO users with congenital limb loss underestimated their prosthesis length (mean: -4.4 ± 2.3 cm), while the participants with acquired limb loss overestimated their prosthesis length (mean: 3.0 ± 1.5 cm).

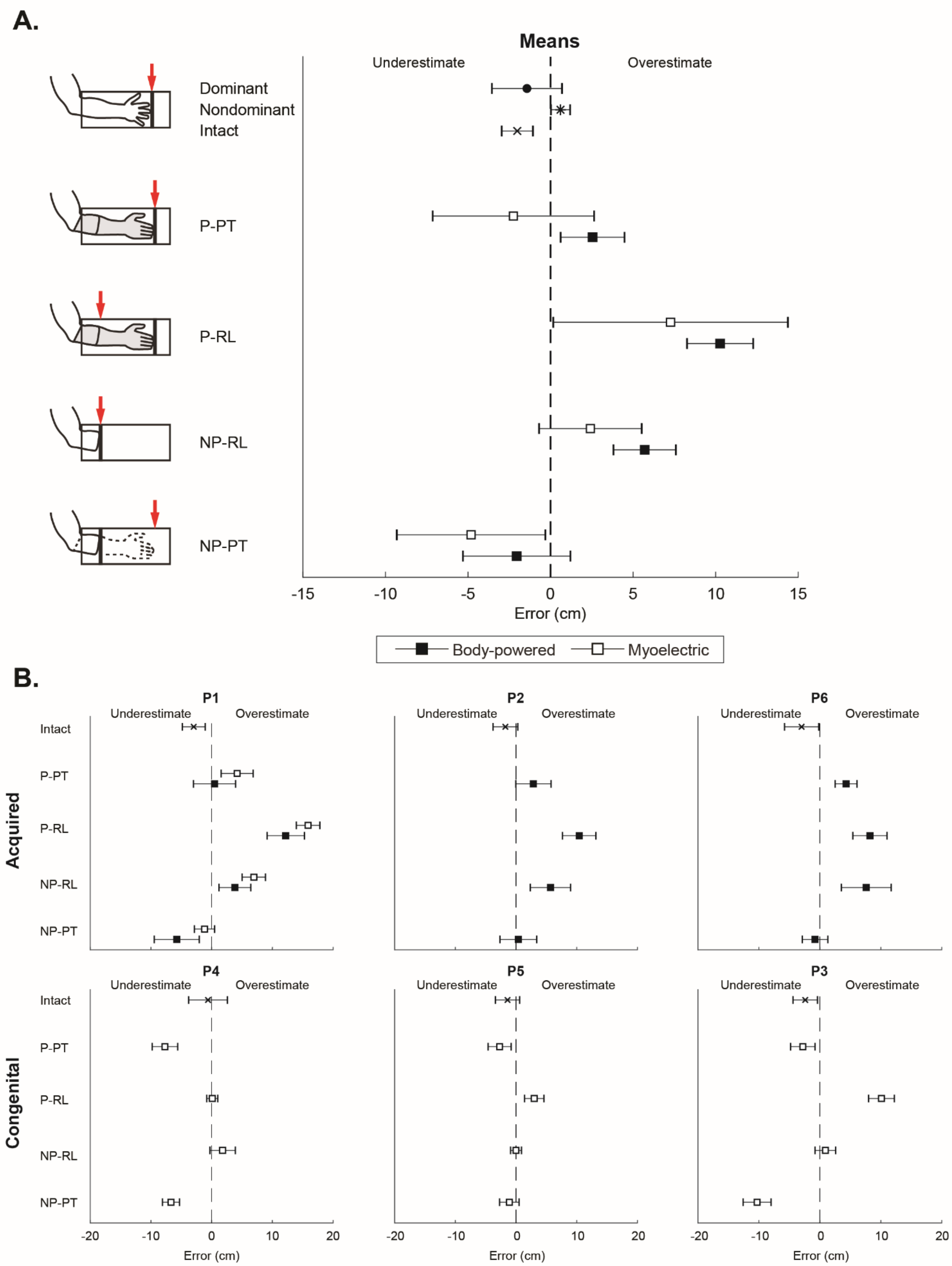

Figure 2.4. Limb length estimation error.
A) Average limb length estimation error for BP (solid squares) and MYO (open squares) prostheses, as well as the anatomical limbs for all prosthesis users and dominant and non-dominant limbs of controls. B) Average error across 10 trials for each participant.

There was a significant correlation (Figure 2.5A) between prosthesis length estimation error and agency as measured by the survey ($r = 0.851$, $p = 0.015$). There was also a significant correlation (Figure 2.5B) between prosthesis length estimation error while not wearing the prosthesis and hours per day spent wearing the prosthesis ($r = 0.885$, $p = 0.008$).

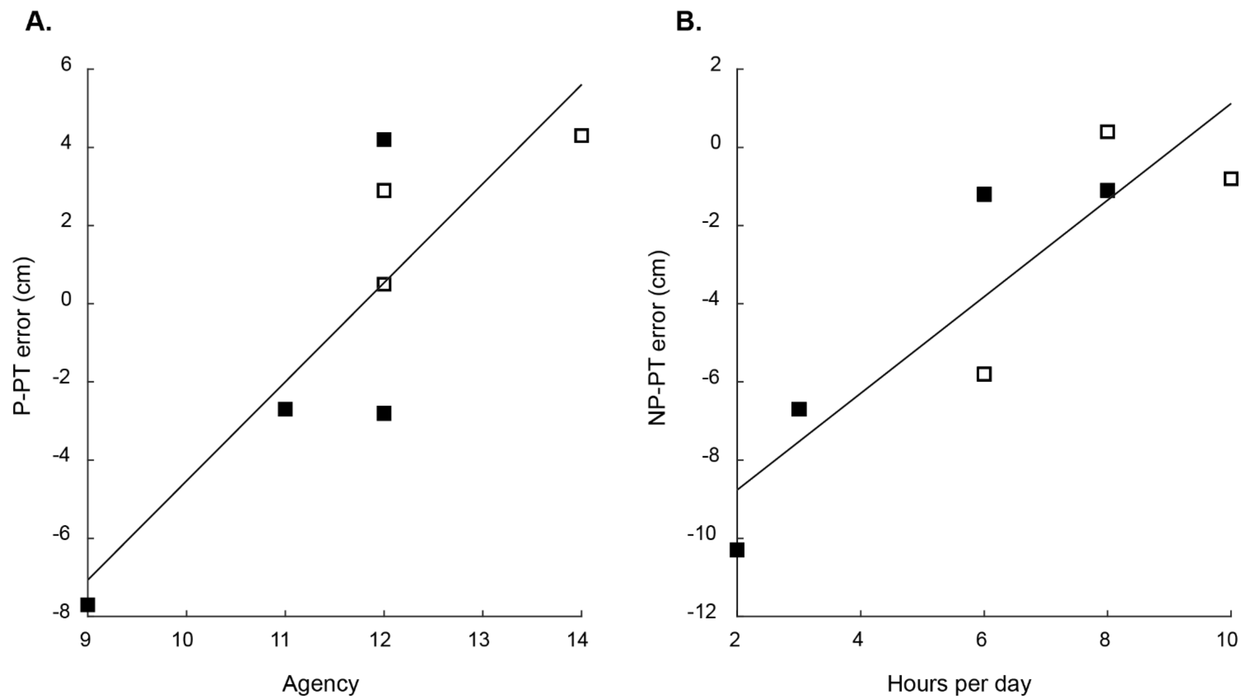

Figure 2.5. Correlations between limb length error and participant characteristics. Correlations between (A) mean P-PT error and agency, and (B) NP-PT error and hours per day wearing a prosthesis (solid squares = BP prosthesis users, open squares = MYO prosthesis users).

2.4.3 Tactile extinction

Out of the 1600 visual only trials presented across all participants, there were only 9 false positive responses in which the foot pedal was pressed in the absence of a pulse from the tactor.

Inflection points for the visual/tactile and tactile only curves should have been in the range of 0-100, which corresponded to the tactor intensities. However, the fit was very poor in some cases due to considerable variability in the data (Figure 2.6; also see Table A.1 in Appendix A). Data was excluded from further analysis if the inflection point was determined to be greater than 100 or if the adjusted R^2 was negative. A total of 7 visual/tactile curves and 3 tactile only curves were excluded for this reason. The adjusted R^2 exceeded 0.56 for all remaining curves and exceeded 0.8 for 95% of the remaining curves.

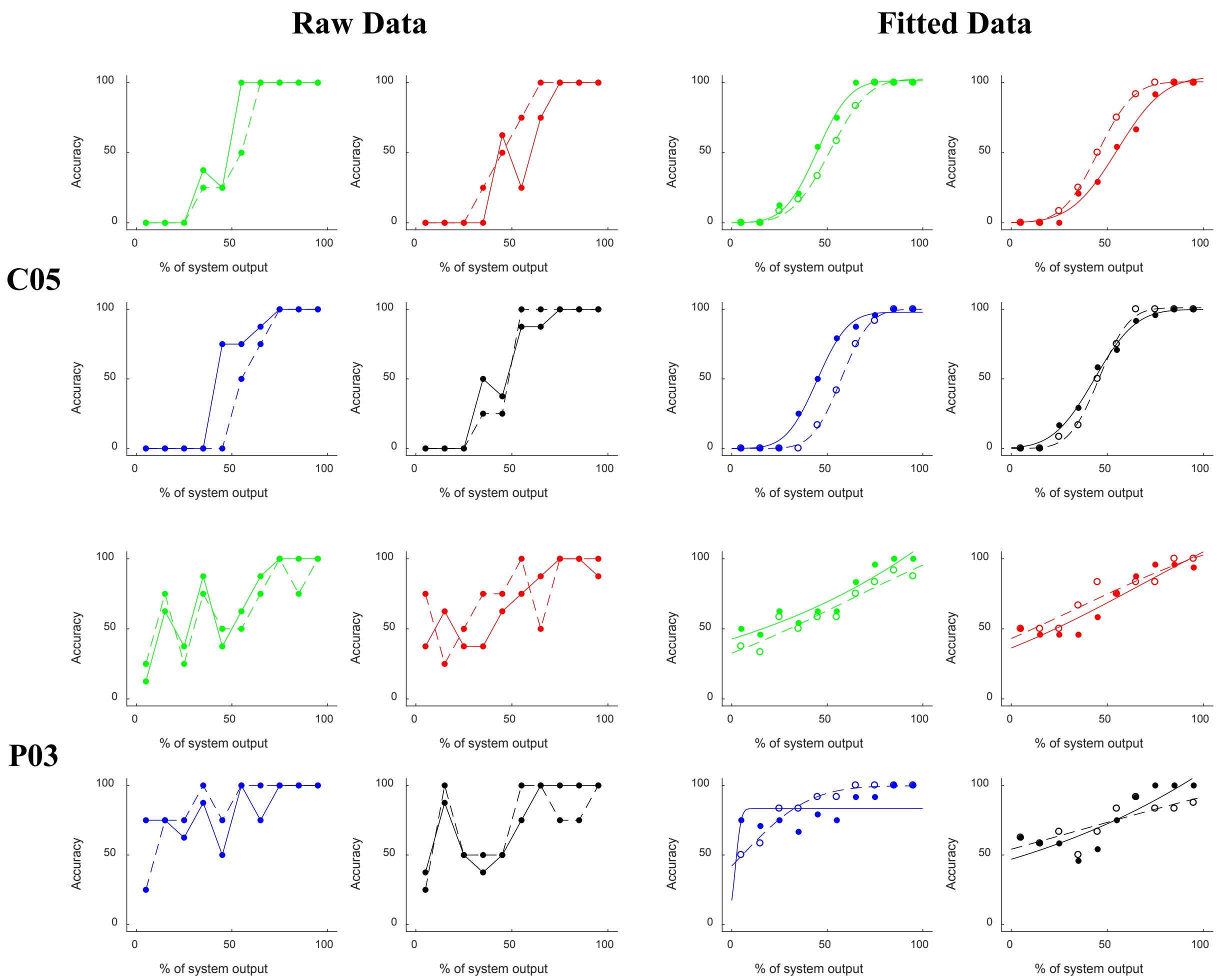

Figure 2.6. Representative visual/tactile and tactile curves.
Representative raw data (left) and data fitted with a cumulative normal function after calculation of a moving average (right). Solid lines correspond to visual/tactile trials and dashed lines correspond to tactile only trials. Curves for C05 (top) illustrate data that is modelled well by the cumulative normal function, while curves for P03 illustrate data that is poorly modelled by this function. (green = Far LED with Far Fixation; red = Mid LED with Far Fixation; blue = Mid LED with Near Fixation; black = Near LED with Near Fixation).

There were few consistent trends in inflection point differences based on prosthesis type or LED position (Figure 2.7). The average inflection point difference was -8.59 for BP prostheses on the Far LED with Far Fixation condition, and this difference gradually became more positive across the other conditions as the LED was closer to the participant. However, there was not a clear trend in inflection point differences based on LED position for MYO prostheses or controls. For most conditions, the inflection point difference was negative for BP prostheses and positive for MYO prostheses. Differences were small for the Far LED with Far Fixation condition (g = 0.54) and Mid LED with Far Fixation condition (g = 0.45), but large for the Mid LED with Near Fixation condition (g = 4.80). In contrast, the inflection point difference was positive for BP prostheses and negative for MYO prostheses on the Near LED with Near Fixation condition (g = 0.75).

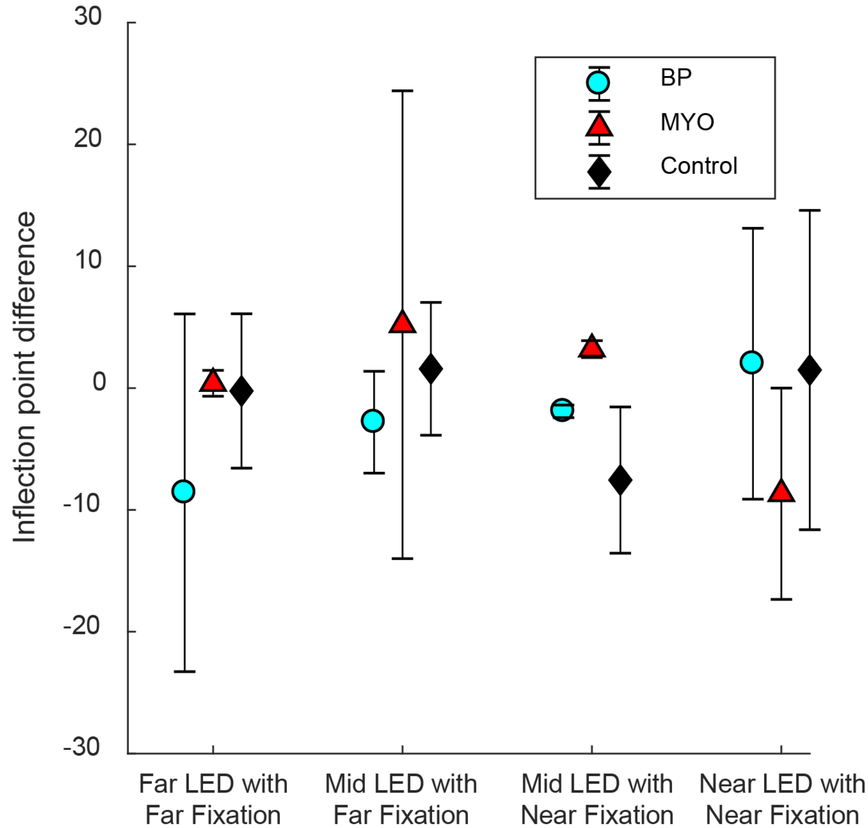

Figure 2.7. Inflection points.
Differences in inflection points between VT and T curves. A negative difference reflects extinction**.**

2.5 Discussion

The purpose of this work was to compare experiences of embodiment between BP and MYO prosthesis users using both phenomenological and behavioral methodology. We hypothesized that BP users would experience stronger embodiment than MYO prostheses due to the availability of proprioceptive feedback in BP prostheses. Our results demonstrate some preliminary evidence in support of this hypothesis, although findings differ based on the methodology.

2.5.1 Survey

Our survey results suggest that there are differences between the sense of prosthesis ownership and agency for BP and MYO users. Although the sense of ownership was only slightly stronger for BP users compared to MYO users, there was a clearer difference in the sense of agency. Ownership and agency are distinct cognitive processes (Kalckert and Ehrsson 2012), so it is reasonable that prosthesis users reported different experiences for each process. The sense of ownership is thought to depend on multisensory integration (Makin et al. 2008), such that matching between visual and somatic signals received from a limb allows for a sense of ownership to develop. Although we hypothesized that multisensory integration might be elevated for BP users (leading to stronger ownership), the lack of difference between ownership scores between BP and MYO prosthesis users does not support this idea. It is possible that the questions asked in this survey were simply insufficient to capture participant's sense of prosthesis ownership. Only four questions were included, and none of the questions have been validated. This could also explain why another study using the same survey showed that people who use their upper limb prosthesis at least 12 hours per day do not report stronger ownership of their prosthesis than people who use it fewer than 6 hours per day – even though the frequent users report stronger agency (Imaizumi et al. 2016). We elected to use this survey, despite its limitations, because there are currently no other published surveys available in the literature that ask directly about embodiment. Furthermore, this survey developed based on a semi-structured interview developed for prosthesis users (Wijk and Carlsson 2015), as well as a questionnaire which demonstrated psychometrically that embodiment of a rubber hand can be separated into ownership and agency components (Longo et al. 2008). While the survey is not validated, it is at least grounded in existing literature.

The mechanism underlying agency is different than for ownership. It is believed that agency depends on comparison between the efferent copy and sensory feedback (Blakemore and Frith 2003). When the efferent copy and feedback match, an individual will perceive that the movement has been performed as intended and will experience a sense of agency. The fact that MYO users reported a lower sense of agency may be evidence that they are not receiving adequate sensory feedback, causing a mismatch in the comparator. Additionally, it should be noted that MYO users experience a considerable degree of uncertainty in controlling their prosthesis. This uncertainty relates to unwanted activations that can occur when the electrode is not optimally interfaced with the residual limb due to socket fit or loading (Chadwell et al. 2016), as well as electromechanical delays associated with the hand itself. Indeed, (Saunders and Vijayakumar 2011) demonstrated that adding uncertainty to the controller of a closed-loop robotic hand resulted in significantly worse performance on a grasping task. In contrast, BP users may experience less uncertainty given the direct mechanical linkage between the harness and terminal device, which may result in a stronger sense of agency.

2.5.2 Limb length estimation

Results from the limb length estimation procedure were less consistent in demonstrating a difference between BP and MYO users. Another study which used this procedure found that children using MYO prostheses overestimated their residual limb length by an average of 7.9 cm (range: 0.6 -14.8 cm) when wearing their prosthesis (McDonnell et al. 1989). The average overestimation error for our participants was similar (mean: 8.6 cm, range: 0.1 – 15.9 cm).

Extension of the perceived residual limb length to more closely match the prosthesis length might be interpreted as evidence of spatial embodiment (de Vignemont 2011).

Although estimation error was slightly higher for BP users, the effect size was small. Instead, patterns in the results seem more dependent on cause of limb loss. We cannot compare this finding to (McDonnell et al. 1989), as their sample had both acquired and congenital cases but they did not differentiate error between the two groups. In our sample, the variability is largely driven by the two MYO users with congenital limb loss who had very low estimation error. This is presumably due to the fact that they had begun using their prostheses within the past year. They may have retained an accurate internal model of their residual limb length that was not altered by prosthesis use. In contrast, the participants with acquired limb loss and the participant with congenital limb loss who had 33 years of prosthesis experience showed clear alterations in their perception of residual limb length while wearing their prosthesis.

Similar malleability in perceived residual limb length with prosthesis use (Canzoneri et al. 2013a) and perceived arm length with tool use (Canzoneri et al. 2013b) has been documented during a tactile distance perception task. In the case of tool use, it has been proposed that this malleability is a consequence of sensory feedback resulting from actions performed with the tool. When the tool contacts an external object, sensory cues are delivered to the upper limb through the tool. This creates a perceptual expansion of the space in which body-related sensory information is located (Canzoneri et al. 2013b). Thus, the body representation expands to incorporate the tool or prosthesis so that the body is prepared to respond appropriately to stimuli that might interact with the body (Canzoneri et al. 2013a).

An additional explanation of the overestimation error could relate to the way the experiment was performed. A study of individuals with paraplegia found that they consistently overestimated their shoulder width as being closer to the width of their wheelchair (Arnhoff and Mehl 1963). Interviews following testing revealed that participants reported using the wheelchair as a reference point for estimation (i.e., visualizing the wheelchair width and subtracting to arrive at shoulder width). This strategy differs with that employed by healthy controls, who reported using their own body as a reference point. In our study, participants were always asked to estimate their prosthesis length immediately prior to estimating their residual limb length. Because they moved the slider from the position where they perceived their prosthesis inwards towards where they perceived their residual limb, it is possible that they were conditioned to use the prosthesis as a frame of reference. However, patterns in the prosthesis length estimation error might contradict this idea, as participants with congenital limb loss underestimated for this condition. It possible that participants with congenital limb loss were using their residual limb as a frame of reference for their estimate of prosthesis length, so that the prosthesis was perceived as being shorter and closer to the residual limb length. As we did not ask participants what strategy they used when estimating residual limb length, it is difficult to speculate further.

Accurate internal models of residual limb length in participants with congenital limb loss might also explain why they were highly accurate at estimating residual limb length while not wearing a prosthesis. In contrast, participants with acquired limb loss overestimated their residual limb length, perhaps reflecting a failure to update their internal model. It is worth noting that all participants with acquired limb loss were fairly new to prosthesis use (Table 2.1). Although no significant correlation was noted between duration of prosthesis use and residual limb length error, it would be interesting to explore whether this error decreases over time.

Prosthesis length estimation errors may have been related to how the prosthesis length was measured. The point on the prosthesis that participants were asked to locate might not have been the one that they generally used to interact with objects. Thus, the length that they were asked to estimate may not equal the length that they actually perceive. This is especially true for the MYO users who performed the task with a fist rather than extended fingers. It is interesting to note that MYO users in another study tended to overestimate how far they could reach with their prosthesis (Gouzien et al. 2017). Although the authors state that measured limb length did not differ between the intact and the prosthetic limbs, it is unclear what distal point on the prosthesis was used as a basis for measurement.

There were no clear patterns based on prosthesis type or cause of limb loss for prosthesis length estimation error when not wearing the prosthesis. However, it is interesting to note that the estimation error was quite small ($\leq$1.2 cm) for four participants. As there was a significant correlation between estimation error and hours per day spent wearing the prosthesis, these four participants also reported wearing their prosthesis the longest each day. In another study, prosthesis users with higher levels of prosthesis integration (which was assessed partially based on the number of hours the prosthesis was worn each day) had lower error when estimating how far they could reach with their prosthesis (Gouzien et al. 2017). However, low prosthesis length estimation error when not wearing the prosthesis did not necessarily correspond to low prosthesis length estimation error when wearing the prosthesis.

2.5.3 Tactile extinction

The extinction paradigm did not reveal any evidence supporting the hypothesis that BP users embodied their prostheses more strongly than MYO users. If prosthesis users had reduced awareness of their prosthetic limb, we expected that they would fail to detect subtle tactile stimuli applied to that side of their body in conjunction with a visual stimulus. Furthermore, we expected that embodying a prosthesis would extend this extinction effect such that prosthesis users would still fail to detect the tactile stimulus even when the visual stimulus was presented more distally. However, our results did not show these trends.

There are a number of possibilities that could explain these results. One possibility is that this paradigm is simply not sufficient to detect whether a prosthesis user embodies their device. Although there is some evidence from the surveys and limb length estimation task to suggest that our participants embodied their prostheses, it is important to note that embodiment is not an immutable or comprehensive construct. It has been argued that a prosthesis may be embodied only in some circumstances (de Vignemont 2011). Thus, a lack of correlation between different measures of embodiment does not mean the prosthesis is not embodied. While a prosthesis user's response to a specific circumstance may provide sufficient evidence to indicate that they embody their device, this evidence also may not be necessary to demonstrate embodiment (de Vignemont 2011). Interestingly, people with upper limb loss have been shown to favor their intact side when performing an adapted version of the landmark-position judgment task, suggesting that they have a mild visual neglect affecting their amputated side (Makin et al. 2010). While this might suggest that people with limb loss have attentional differences between their amputated and intact limbs, it does not necessarily mean that they will experience tactile extinction or that the degree of extinction would be modulated by prosthesis use. In fact, another study demonstrated that upper limb absence does not alter how people interact with stimuli presented within peripersonal space during a response selection task based on the Simon effect (Philip and Frey 2013).

Another possibility is that there were methodological flaws in the paradigm. Participants were given the opportunity to acclimate to the tactor prior to beginning the experiment, but many reported difficulty detecting it even when the intensity was close to maximum. The tactor was set to vibrate at 250 Hz, which is the frequency at which rapidly-adapting mechanoreceptors (Pacinian corpuscles) are maximally sensitive (Makous et al. 1995). However, ability to detect tactile stimuli is known to vary based on contact area (Verrillo 1963) and mounting pressure (Cohen et al. 2005). Contact area was constant between all participants since the same tactor was used, but mounting pressure likely varied. The tactor was initially mounted using only double-sided tape, but we wrapped it more securely using Coban tape if participants had difficulty detecting the vibrations. Furthermore, the visual stimuli may not have been sufficiently distracting to induce an extinction effect. The LED intensities were set to be only slightly higher than a detectable level for each participant, and participants were also instructed to focus their attention on the tactor rather than the LEDs. Altering the protocol to make the LEDs more distracting (i.e., brighter) and the tactor less noticeable (i.e., shorter stimuli) did not seem to help induce extinction in a pilot study involving healthy controls (Appendix A.2). It is possible that these changes could help induce extinction in prosthesis users, however. Finally, although we attempted to prevent fatigue in our participants by providing frequent breaks during the testing, it is possible that participants had difficulty maintaining focus. Failure to detect the tactile stimuli in some cases may be a result of their diminished focus, rather than a physiological reason.

Finally, it should be noted that this is a very small dataset with considerable within- and between-subject variability, possibly due to the methodological reasons described above. Given this variability, we had to first apply a moving average filter prior to fitting with a cumulative normal function. While this fit the data well in most cases, we did have to exclude some data due to a very poor model fit, further reducing an already small sample size. Further exploration with a larger sample size is needed.

2.5.4 Conclusion

This study demonstrated that BP prosthesis users reported a stronger sense of agency over their prosthesis in comparison to MYO users. This finding might be linked to difference in control methodology between the two devices, such that greater uncertainty in MYO control might lead to a reduced sense of agency. Although we did not demonstrate differences in embodiment between BP and MYO prostheses using the other methodology, continued exploration with a larger sample size is needed.

CHAPTER 3. Reliability of upper limb movement quality metrics during everyday tasks[1]

3.1 Abstract

Quantitative assessments of an individual's functional status commonly involve the use of movement quality metrics. The purpose of this work was to quantify the reliability of movement quality metrics in healthy adults during a variety of unconstrained activities of daily living (ADLs). Nineteen participants performed six ADLs (lifting a laundry basket, applying deodorant, turning a doorknob, placing a pill in a pillbox, placing a pushpin in a bulletin board, and drinking water from a glass) during two separate sessions. The ADLs were divided into reaching and object manipulation phases. Movement quality for each phase was assessed using three measures of smoothness (log dimensionless jerk, spectral arc length, and number of submovements) and one measure of straightness (index of curvature). Within- and between-session reliability was quantified using intraclass correlation coefficients (ICCs) and minimum detectable changes in measured units and as a percentage of their mean value (MDC%). Reliability was generally lower within-session than between-session and for object manipulation tasks compared to reaching tasks. The ICCs exceeded 0.75 for 5% of the within-session metrics and 73% of the between-session metrics. The average MDC% was 35% for the within-session metrics and 20% for the between-session metrics. Reliability was similar for most metrics when averaged across the tasks, but the number of submovements consistently indicated much lower reliability. Unconstrained ADLs can

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reliably be used to assess movement quality in functional settings that mimic real-world challenges. However, the specific movement quality metrics used in the assessment should be chosen carefully since some metrics perform dissimilarly when applied to the same data. In particular, it may be advisable to use the number of submovements in combination with other metrics.

3.2 Introduction

Movement smoothness is a fundamental property of skilled, well-coordinated motor behavior (Hogan and Sternad 2009). It likely reflects the nervous system's attempt to minimize movement error (Harris and Wolpert 1998) and energy costs (Nishii and Taniai 2009), as well as intrinsic mechanical filtering properties of muscle (Krylow and Rymer 1997). In healthy individuals, unconstrained point-to-point movements are approximately straight with symmetric velocity profiles (Flash and Hogan 1985) and highly blended submovements that overlap in time (Rohrer et al. 2004). These characteristics are altered or absent in movements made by individuals with upper limb impairments, such as those caused by stroke (Rohrer et al. 2002), Parkinsonism (Tresilian et al. 1997), cerebral palsy (Van Der Heide et al. 2005), and upper limb loss (Cowley et al. 2017). Consequently, smoothness is a useful tool for assessing performance deficits and improvement following therapeutic interventions.

Smoothness can be quantified using a variety of metrics that focus on different aspects of the movement. Some metrics relate to movement speed, such as the number of peaks in the speed profile, normalized mean speed, and the mean arrest period ratio (proportion of movement time during which the speed exceeds a given percentage of the peak speed) (Rohrer et al. 2002). Other metrics are based on jerk (rate of change of acceleration), including integrated squared jerk and root mean squared jerk (see (Hogan and Sternad 2009) for additional examples). Several metrics quantify the complexity of the Fourier magnitude spectrum (Balasubramanian et al. 2012, Balasubramanian et al. 2015) since smooth movements contain primarily low frequency components, while unsmooth movements contain additional high frequency components. Most of these metrics do not meet basic utility requirements—that is, they are not dimensionless, robust to signal noise, or monotonically responsive and sensitive to changes within a range covering the physiological spectrum of healthy and pathological movements (Balasubramanian et al. 2012, Balasubramanian et al. 2015). Recent comparative work suggests that the natural logarithm of dimensionless integrated squared jerk and arc length of the Fourier magnitude spectrum may be well-suited for meeting these constraints (Balasubramanian et al. 2012, Balasubramanian et al. 2015), although other metrics remain prevalent in the literature.

When using any of these metrics to assess movement quality, it is important to understand their test-retest reliability. This information facilitates appropriate interpretation of the data so that meaningful results can be distinguished from experimental errors or natural variability in subject performance (Schwartz et al. 2004). Reliability has already been quantified in patient and healthy populations for a variety of movement quality metrics, including index of curvature (Wagner et al. 2008, Schneiberg et al. 2010, Patterson et al. 2011), number of peaks in the speed profile (Wagner et al. 2008, Schneiberg et al. 2010), normalized jerk (Caimmi et al. 2008, Caimmi et al. 2015), and the median logarithm of instantaneous curvature (Osu et al. 2011). However, these studies have several drawbacks that limit their generalizability. Most studies focused on targeted reaching movements, which may not fully represent the complexity of movements performed in daily life. Two studies (Schneiberg et al. 2010, Osu et al. 2011) included simulated activities of daily living (ADLs), which have been shown to involve different movement patterns than functional ADLs (Taylor et al. 2018). It is important to understand how tasks are accomplished in unconstrained environments that mimic real-world settings, which may present additional challenges that are not involved in targeted reaching movements or simulated ADLs.

Additionally, all studies except (Patterson et al. 2011) and (Wagner et al. 2008) reported reliability in relative terms using Pearson correlations and intraclass correlation coefficients (ICCs). Neither of these metrics define the expected noise in the data in terms of the units in which the original measurement was made, as the standard error of measurement (SEM) and minimum detectable change (MDC) do. Given the expected magnitude of a movement quality metric for a normative population, deviations from normative movement can easily be identified using one of these absolute indicators of reliability. As such, the primary purposes of this study were to 1) establish the range of normative movement quality metrics for a variety functional ADLs, and 2) establish the test-retest reliability of these metrics in healthy adults. A secondary purpose was to explore the effects of filtering on the magnitude and reliability of these metrics.

3.3 Methods

3.3.1 SubjectsNineteen participants (9 male, age: 22 ± 4 years, height: 1.72 ± 0.10 meters, weight: 71.4 ± 14.2 kilograms) provided written informed consent for this institutionally approved study. Exclusion criteria included a history of serious musculoskeletal, cardiovascular, neurological, respiratory, or visual problems. Additional data from the same participants is reported in (Engdahl and Gates 2018).

3.3.2 Experimental protocol

Participants performed a series of six ADLs (Table 3.1; also described in (Engdahl and Gates 2018)) during two identical data collection sessions. The interval between sessions was at least one day (mean: 12 ± 10 days). During each session, the ADLs were repeated 10 times at a comfortable pace. Since we expected participants to be familiar with these common daily tasks, we did not provide practice time or instructions on how to complete the tasks. However, we did require that each ADL repetition began from a consistent initial posture. For standing tasks, the arms were relaxed at the sides. For seated tasks, the arms rested flat on the table at shoulder width. Object position was based on participant anthropometry (Table 3.1) and the table was aligned to the bottom of the rib cage while seated.

Reflective markers on the right wrist (radial and ulnar styloid processes) and hand (3rd and 5th metacarpal heads and base of the 3rd metacarpal) were tracked at 120 Hz using a 19-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA).

3.3.3 Data analysis

Marker position data were filtered in Visual 3D (C-Motion, Germantown, MA) using a fourth-order low-pass Butterworth filter. All analyses were performed on three separate data sets where data were filtered with either a 6, 10, or 20 Hz cutoff frequency. Analyses focused on the wrist joint center for all tasks except for turning the doorknob, which focused on the center of gravity of the right hand (calculated in Visual 3D). The wrist joint center was defined as the midpoint between the styloid markers. First, second, and third derivatives were calculated as the vector sum of derivatives of the wrist joint center or center of gravity position across all three planes of movement (Figure 3.1). Low-pass filters were applied after each differentiation (Salmond et al. 2017).

Table 3.1. Segmentation of reaching and manipulation tasks.
Each ADL was segmented into reaching and manipulation tasks based on position or velocity of the wrist.

ADL	Procedure	Task Name	Task Type	Segmentation Definition
BASKET	Participants stood facing the table with a 5 lb. laundry basket placed on the ground in front of it. They lifted the basket and placed it on the table.	Basket_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: WJC reaches global minimum in vertical direction
		Basket_Transport	Reaching	START: WJC reaches global minimum in vertical direction STOP: WJC reaches global maximum in anteroposterior direction
DEO	Participants stood 75% of arm's length from a stick of deodorant placed on the table along the midline of the body. They lifted the deodorant with the right hand, removed the cap with the left hand and simulated three swipes on the left axilla.	Deo_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: WJC reaches global maximum in anteroposterior direction
DOOR	Participants stood 75% of arm's length from a doorknob placed at waist height along the midline of the body. They turned the knob clockwise until its latch was fully retracted.	Door_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Mediolateral velocity of RSP reaches first local minimum
		Door_Turn	Manipulation	START: Mediolateral velocity of RSP reaches first local minimum STOP: Mediolateral velocity of RSP reaches last local minimum
PILL	Participants held a pill in the right hand while seated at the table. They placed the pill in a pillbox located at the midline of the body and in between the hands.	Pill_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Velocity of WJC reaches global minimum
PIN	Participants stood at 75% of arm's length from a corkboard with a 1 inch diameter circle drawn at eye level along the midline of the body. They placed a pushpin inside of the circle.	Pin_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Anteroposterior velocity of WJC returns below 5 cm/s
		Pin_Place	Manipulation	START: Anteroposterior velocity of WJC returns below 5 cm/s STOP: Anteroposterior velocity of WJC exceeds 5 cm/s
WATER	Participants sat at the table with a solid plastic cup containing approximately 150 mL of water placed at the midline of the body and in between the hands. They lifted the cup and took a sip.	Water_Reach	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Mediolateral velocity of WJC reaches first local minimum
		Water_Transport	Reaching	START: Vertical velocity of WJC reaches last minimum before peak vertical velocity STOP: Anteroposterior velocity of WJC returns below 5 cm/s
		Water_Drink	Manipulation	START: Anteroposterior velocity of WJC returns below 5 cm/s STOP: Anteroposterior velocity of WJC exceeds 5 cm/s

WJC = wrist joint center; RSP = radial styloid process

Each ADL was divided into independent movement phases that were classified as "reaching" or "object manipulation" tasks. A total of 11 reaching and manipulation tasks were included for analysis (Table 3.1). The outcome measures for each reaching task included smoothness and straightness, while outcome measures for each manipulation task included only smoothness (Figure 3.2). Smoothness was measured as log dimensionless jerk (LDJ), spectral arc length (SPARC) and number of submovements (nSUB). LDJ was the natural logarithm of the dimensionless integrated squared jerk:

$$LDJ := \ln\left(\frac{(t_2-t_1)^5}{A^2}\int_{t_1}^{t_2} \dddot{x}(t)^2 dt\right), \qquad \text{(Equation 3.1)}$$

where $x(t)$ was the position of the endpoint (wrist joint center or center of gravity), A was the arc length, t_1 was the movement start time, and t_2 was the movement end time. Because jerk is the rate of change of acceleration, increased LDJ magnitude reflects decreased smoothness (due to more rapid changes in acceleration). SPARC was the arc length of the Fourier magnitude spectrum of the speed profile (Balasubramanian et al. 2015):

$$SPARC := \int_0^{\omega_c} \sqrt{\left(\frac{1}{\omega_c}\right)^2 + \left(\frac{d\hat{V}(\omega)}{d\omega}\right)^2}\, d\omega; \qquad \text{(Equation 3.2)}$$

$$\hat{V}(\omega) := \frac{V(\omega)}{V(0)};$$

$$\omega_c := \min\left\{\omega_c^{max}, \min\left\{\omega, \hat{V}(r) < \overline{V}_{threshold}\ \forall\, r > \omega\right\}\right\},$$

where $\hat{V}(\omega)$ was the Fourier magnitude spectrum normalized to the DC magnitude ($V(0)$) and ω_c was an adaptively selected cutoff frequency. This cutoff frequency was the smaller of two values: 1) a fixed frequency (ω_c^{max}) chosen to cover the anticipated spectrum of the movement analyzed and 2) the frequency at which the magnitude for all greater frequencies is below a certain threshold ($\overline{V}_{threshold}$). According to recommendations from (Balasubramanian et al. 2015), $\overline{V}_{threshold} = 0.05$ and $\omega_c^{max} = 20\ Hz$. The Fourier magnitude spectrum for a smooth movement will be a smooth function of frequency (Balasubramanian et al. 2015). The shape of the frequency spectrum for an unsmooth movement will be more complex, such that the arc length of the spectrum is increased. Here, SPARC was defined as positive to reflect this relationship (i.e., an increase in magnitude corresponds to a decrease in smoothness). Submovements were defined as local maxima in the velocity profile, where $\frac{dv}{dt} = 0$ and $\frac{d^2v}{dt^2} < 0$. Straightness was defined using index of curvature (IOC), or the arc length of the position trajectory divided by the length of a straight line between the initial and final position. An IOC of one indicates a perfectly straight movement.

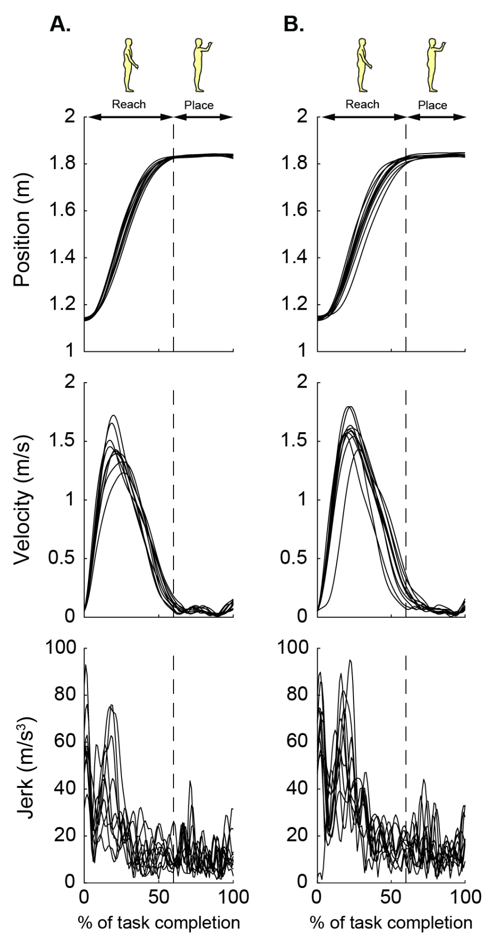


Figure 3.1. Representative position, velocity, and jerk trajectories.
Position, velocity, and jerk trajectories for ten trials are shown for one representative subject performing the Pin task during session 1 (**A**) and session 2 (**B**). The movement is divided into reaching and placing phases (shown by the vertical dashed line) following the conventions described in Table 1. The trials are normalized to 100% of task completion for visual clarity, although data was not normalized prior to calculating the movement quality metrics.

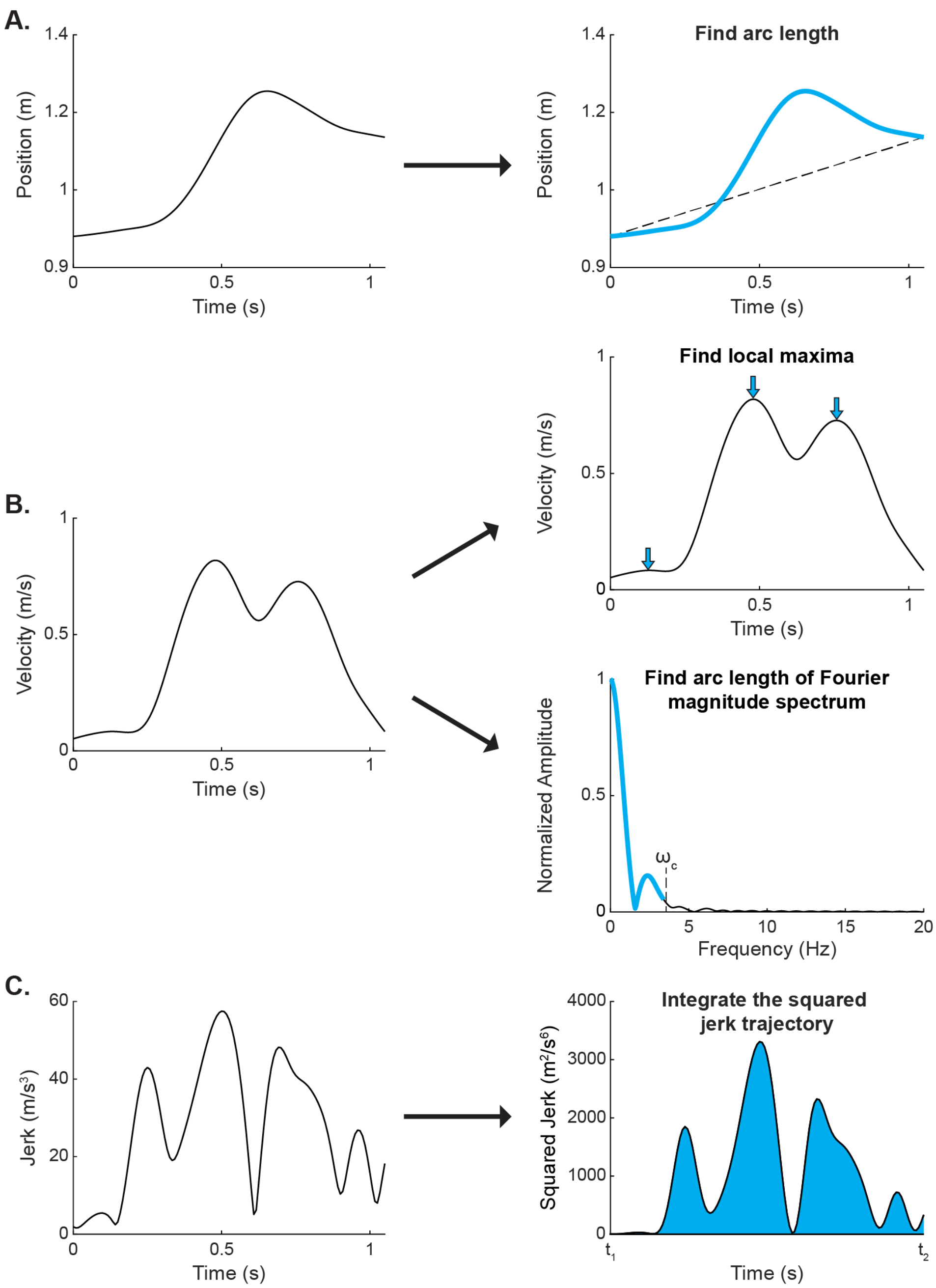


Figure 3.2. Definition of movement quality metrics.
Movement quality was quantified using four metrics of smoothness and straightness. Derivations of each metric are illustrated using a representative trial from the Deo_Reach task. **A)** The index of curvature (IOC) is a ratio of the arc length of the position trajectory (blue line) to the length of a straight line connecting the movement start and end points (dashed line). **B)** The number of submovements (nSUB) and the spectral arc length (SPARC) are determined using the speed profile. nSUB is the number of local maxima in the speed profile (blue arrows). SPARC is the arc length of the normalized Fourier magnitude spectrum (blue line) between 0 Hz and an adaptively selected cutoff frequency (ω_c). **C)** The log dimensionless jerk (LDJ) is based on the squared jerk trajectory, which is integrated (shaded area) over the duration of the movement (t_2 - t_1). This integral value is normalized based on the duration and arc length (blue line in A) of the movement.

3.3.4 Statistics

Within-session reliability metrics were calculated based on all 10 repetitions from the first session using data filtered with a cut-off frequency of 10 Hz. Between-session reliability metrics were calculated using the averages of all 10 repetitions from each session. Within- and between-session ICCs were calculated using (2,1) and (2,k) models for absolute agreement, respectively. We assessed heterogeneity of the data by checking the significance ($p < 0.05$) of the between-subjects variance from the two-way ANOVA. In order for the ICC to be valid, there must be significant between-subject variance (Portney and Watkins 2009). Eight invalid ICCs (4 within-session, 4 between-session) were excluded (Appendix B.1). Six of the invalid ICCs were calculated for nSUB. Following recommendations in (Portney and Watkins 2009), we considered ICCs > 0.90 to be clinically reliable, ICCs > 0.75 to indicate good reliability in other contexts, and ICCs < 0.75 to indicate poor to moderate reliability. Within- and between-session MDCs were calculated according to $\mathrm{MDC}_{95} = \mathrm{SEM} * 1.96 * \sqrt{2}$, where 1.96 corresponds to a 95% confidence interval and SEM is the square root of the mean square error term from the two-way ANOVA (Weir 2005). Because the movement quality metrics are all quantified on a different scale, MDCs are also presented as a percentage of mean values from the first session (MDC%) to facilitate comparison between metrics. All statistical analyses were performed using SPSS 24 (IBM, Armonk, NY).

3.4 Results

3.4.1 Normative movement quality metrics

In general, movement quality was higher for reaching tasks (LDJ: 7.65 ± 0.96; SPARC: 1.50 ± 0.07; nSUB: 1.38 ± 0.55) compared to manipulation tasks (LDJ: 9.65 ± 0.41; SPARC: 3.14

± 0.50; nSUB: 2.78 ± 0.34) (Table 3.2). Within the reaching tasks, the basket transport task had the poorest movement quality across all four metrics.

3.4.2 Within- and between-session reliability metrics

Reliability was generally poorer within-session than between-session (Figure 3.3 and Figure 3.4). Only 5% of all valid within-session ICCs were greater than 0.75 (mean: 0.53), compared to 73% of all valid between-session ICCs (mean: 0.80) (Figure 3.3). Similarly, the mean MDC% across all metrics (Figure 3.4) was higher within-session (35%) than between-session (20%).

These patterns are influenced by the fact that the reliability for nSUB was consistently worse than for the other three metrics. If nSUB is excluded, the discrepancy between the mean within- and between-session MDC% becomes much smaller (within-session: 16%; between-session: 11%). However, the discrepancy between the within- and between-session ICCs does not change (within-session mean: 0.55; between-session mean: 0.76). This is likely because only seven of the 11 within-session ICCs for nSUB were valid in the first place (Appendix B.1).

3.4.3 Reliability metrics across tasks

Reliability for LDJ, IOC, and SPARC was similar when averaged across tasks. The mean within-session ICC for these three metrics was 0.55 (LDJ: 0.55, IOC: 0.61, SPARC: 0.49) and the mean within-session MDC% was 16% (LDJ: 19%, IOC: 12%, SPARC: 18%). The mean between-session ICC was 0.82 (LDJ: 0.80, IOC: 0.83, SPARC: 0.81) and the mean between-session MDC% was 11% (LDJ: 13%, IOC: 8%, SPARC: 11%).

Table 3.2. Mean (SD) movement quality metrics for sessions 1 and 2.

		LDJ		IOC		SPARC		nSUB	
		Session 1	**Session 2**	**Session 1**	**Session 2**	**Session 1**	**Session 2**	**Session 1**	**Session 2**
Reaching	Basket_Reach	7.63 (0.45)	7.48 (0.36)	1.07 (0.04)	1.06 (0.03)	1.47 (0.04)	1.46 (0.05)	1.16 (0.18)	1.09 (0.11)
	Basket_Transport	10.02 (0.47)	9.84 (0.36)	1.56 (0.07)	1.54 (0.07)	1.65 (0.07)	1.65 (0.07)	2.85 (0.54)	2.64 (0.49)
	Deo_Reach	7.06 (0.48)	6.80 (0.34)	1.08 (0.07)	1.05 (0.02)	1.48 (0.05)	1.46 (0.01)	1.07 (0.25)	1.02 (0.04)
	Door_Reach	7.15 (0.45)	7.05 (0.38)	1.06 (0.03)	1.06 (0.02)	1.44 (0.02)	1.45 (0.02)	1.21 (0.26)	1.23 (0.25)
	Pill_Reach	6.86 (0.56)	6.76 (0.57)	1.09 (0.04)	1.10 (0.05)	1.46 (0.03)	1.47 (0.03)	1.04 (0.09)	1.03 (0.08)
	Pin_Reach	7.87 (0.45)	7.86 (0.33)	1.04 (0.02)	1.03 (0.01)	1.46 (0.01)	1.45 (0.02)	1.06 (0.10)	1.15 (0.18)
	Water_Reach	7.42 (0.65)	7.46 (0.58)	1.28 (0.20)	1.25 (0.14)	1.60 (0.12)	1.57 (0.10)	1.49 (0.44)	1.43 (0.33)
	Water_Transport	7.53 (0.46)	7.55 (0.49)	1.02 (0.01)	1.02 (0.01)	1.50 (0.06)	1.49 (0.04)	1.31 (0.19)	1.29 (0.26)
Manipulation	Door_Turn	10.20 (0.93)	9.94 (0.92)	n/a	n/a	2.89 (0.76)	2.84 (0.51)	3.27 (0.96)	3.08 (0.97)
	Pin_Place	9.69 (1.49)	8.99 (1.73)	n/a	n/a	2.82 (0.50)	2.74 (0.43)	2.84 (1.30)	2.48 (1.36)
	Water_Drink	9.62 (1.30)	9.47 (1.49)	n/a	n/a	3.83 (0.75)	3.73 (0.62)	2.57 (1.48)	2.45 (1.68)

LDJ = log dimensionless jerk; IOC = index of curvature; SPARC = spectral arc length; nSUB = number of submovements

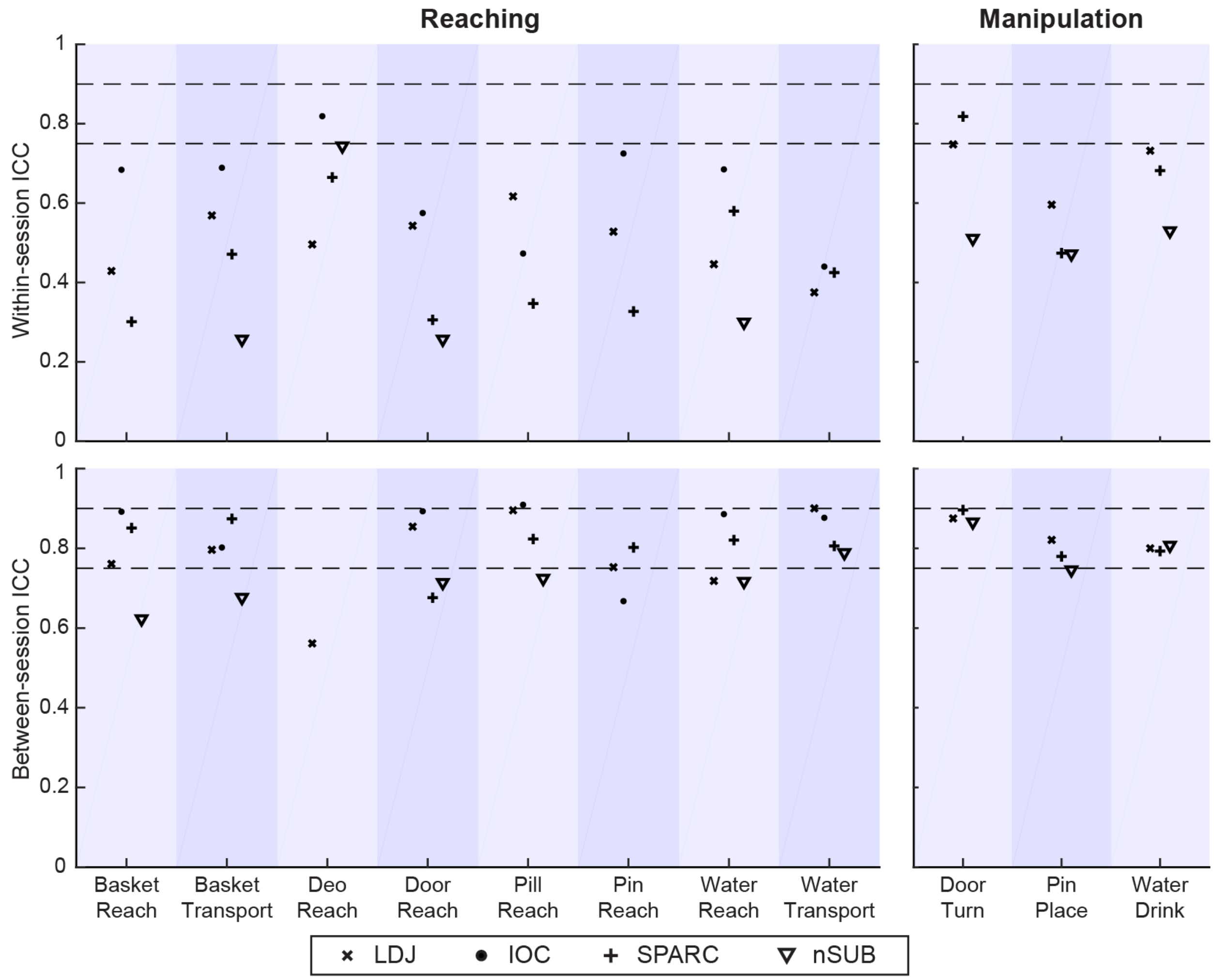


Figure 3.3. Within- and between-session intraclass correlation coefficients.
Within-session (top) and between-session (bottom) intraclass correlation coefficients (ICCs) are shown for each task. Invalid ICCs are not presented. Dashed lines indicate suggested thresholds for interpretation (ICCs > 0.90 indicate clinical reliability, ICCs > 0.75 indicate good reliability in other contexts, and ICCs < 0.75 indicate poor to moderate reliability).

The three object manipulation tasks (turning a doorknob, drinking water, and placing a pushpin) had the largest MDC% when averaged across LDJ, IOC and SPARC. This was true both within-session and between-session. Reaching to the doorknob and reaching with the pin had the smallest MDC% when averaged across LDJ, IOC and SPARC for both within-session and between-session. Interestingly, turning the doorknob also had the largest ICC across tasks when averaged across LDJ, IOC and SPARC for both within-session and between-session. There were no consistent patterns in the ICCs for the remaining 10 tasks. All ICCs, SEMs, and MDCs are included as supplementary material (Appendix B.1).

3.4.4 Effects of filtering

The choice of cutoff frequency affected the magnitude of the movement quality metrics (Appendix B.2). Higher cutoff frequencies resulted in higher magnitudes (less smooth movements), primarily for LDJ and nSUB. While MDC and MDC% also increased, ICCs were minimally affected by cutoff frequency.

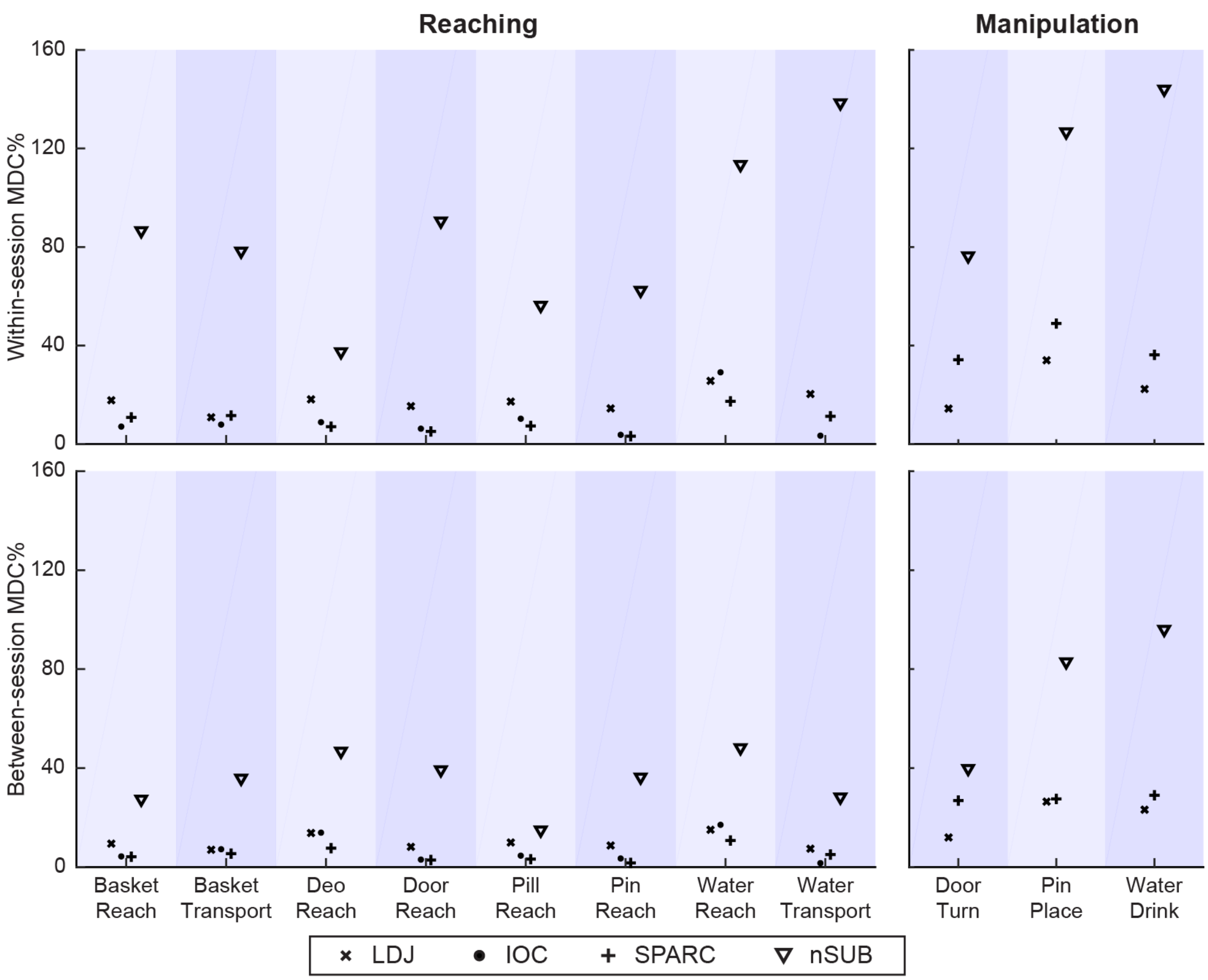


Figure 3.4. Within- and between-session minimum detectable change values.
Within-session (top) and between-session (bottom) minimum detectable change values for each task are presented as a percentage of the mean value from session 1 (MDC%).

3.5 Discussion

This work established normative movement quality metrics for unconstrained ADLs performed by healthy adults. The manipulation tasks had poorer movement quality than the reaching tasks, possibly reflecting the fact that multiple small corrections were needed to accomplish the task goal. Within the reaching tasks, the basket transport task had the poorest movement quality. This task was also the most physically demanding task, requiring to participants to lift a weighted laundry basket from the floor and place it on a table. The weight of the basket combined with the difficulty of navigating it around the table to avoid collision may have contributed to the poorer movement quality. The remaining reaching tasks either did not involve moving an object or involved a smaller, lighter object than was easier to transport (i.e., the plastic cup in the water drinking task). Despite the poorer movement quality, the basket task may still be valuable to use in a functional assessment given its unique task requirements.

Additionally, this work quantified the reliability of movement quality metrics in healthy adults performing unconstrained ADLs. Within-session reliability is often better than between-session reliability because differences in experimental set-up between sessions can introduce unintended variation in task performance. Although we attempted to keep our set-up consistent by standardizing the participants' initial posture and distance from the objects, this variability cannot be completely eliminated. In this study, within-session reliability was lower than between-session, suggesting that participants were not consistent in how they performed the tasks. However, this explanation contrasts with our previous work (Engdahl and Gates 2018), which explored the reliability of peak upper limb and trunk joint angles in healthy adults during ADLs. Several of the tasks included in (Engdahl and Gates 2018) were also included in this study (applying deodorant, turning a doorknob, and placing a pushpin). Since the within-session peak joint angle reliability for these tasks was generally high, it seems unlikely that there were major differences in participants' movement patterns between repetitions of the tasks. The low within-session reliability for the movement quality metrics might instead be attributed to how they were calculated. Since the peak joint angles were calculated directly from positional data at a single point in time, they characterized movement patterns primarily in a global sense. In contrast, the movement quality metrics were calculated using continuous positional data (IOC) or derivatives of continuous positional data (LDJ, SPARC, and nSUB). As such, they may be more reflective of small changes in performance than peak joint angles. They may also be more affected by signal noise and filtering characteristics, especially for the metrics that are based on derivatives. Indeed, increasing the filter cutoff frequency resulted in increased magnitude for the movement quality metrics (Appendix B.2). Of the three metrics calculated using differentiated data, filtering had the greatest impact on nSUB and LDJ. This is consistent with other work showing that SPARC is robust to changes in signal-to-noise ratio and filtering characteristics, especially in comparison to LDJ (Balasubramanian et al. 2015).

Nonetheless, within-session comparisons are less common than between-session comparisons. Most performance assessments involve making comparisons between different time points (e.g., before and after a therapeutic intervention), rather than within a single testing session. Between-session reliability is more relevant than within-session reliability in these situations. In fact, only one previous study of movement quality reliability even included within-session reliability (Osu et al. 2011). For most tasks, the between-session ICC exceeded the threshold for good reliability (ICC > 0.75; defined by (Portney and Watkins 2009)) and the MDC% was low. This suggests that the tasks would be an effective basis for identifying changes in movement quality across multiple testing sessions.

The reliability metrics presented here are comparable to those reported previously for different tasks. Between-session reliability for the IOC during reach to grasp and reach to target tasks has been reported for healthy adults (ICC range: 0.64-0.88; MDC range: 0.00-0.08) (Patterson et al. 2011), individuals with stroke (ICC ranges: 0.92-0.93 (Patterson et al. 2011) and 0.08-0.95 (Wagner et al. 2008); MDC range: 0.08-0.19 (Patterson et al. 2011); MDC% range: 7.4%-28.9% (Wagner et al. 2008)), and children with cerebral palsy (ICC range: 0.59-0.81) (Schneiberg et al. 2010). Between-session reliability for nSUB has also been reported for individuals with stroke (ICC range: 0.43-0.84; MDC% range: 24.4%-67.6%) (Wagner et al. 2008) and children with cerebral palsy (ICC range: 0.82-0.91) (Schneiberg et al. 2010). These similarities are encouraging, as they suggest that tasks do not need to be tightly constrained in order for individuals to perform them consistently. Unconstrained, functional ADLs can still be used in assessments of movement quality.

Reliability tended to be slightly worse for manipulation tasks than reaching tasks, as indicated by the comparatively larger MDC and MDC%. In contrast, the ICCs were similar across all tasks, which might be attributed to shortcomings in the definition of the ICC. ICC magnitude depends on between-subjects variability, such that high between-subject variability in a dataset can artificially increase the ICC even if trial-to-trial variability is high (Weir 2005, Vaz et al. 2013). Although the elevated MDC and MDC% values suggest the trial-to-trial variability was high (i.e., poor reliability), this may not be reflected in the ICCs if the between-subject variability was also high. Since the object manipulation tasks involved making multiple small adjustments to accomplish a goal, it is possible that larger movement variability was involved. Indeed, the within-subject standard deviations tended to be higher for the manipulation tasks (mean: 0.65) than the reaching tasks (mean: 0.18) (Appendix B.3). However, the comparatively lower reliability for manipulation tasks does not necessarily invalidate the use of those tasks in performance assessments. Manipulation tasks challenge different aspects of movement than reaching tasks, so they may be helpful in developing a more comprehensive assessment of functionality.

Of the four movement quality metrics, nSUB tended to show the worst reliability when applied to any task. This is likely due to several problems with how the metric is calculated. Despite the intuitive nature of this metric, it can be difficult to accurately identify peaks (Balasubramanian et al. 2012). Even when peaks are identified accurately, the number of peaks may not truly reflect the number of underlying submovements. The summation of multiple submovements can obscure peaks or create spurious peaks that do not correspond to any of the component submovements, and this issue may be further exacerbated by signal noise (Rohrer and Hogan 2003) and signal filtering characteristics. Aggressive filtering may conceal true peaks, while minimal filtering may not remove any spurious peaks (see Appendix B.2 for the effect of filter cutoff frequency on the magnitude of nSUB). Furthermore, this metric is insensitive to changes in the periods of arrest between submovements (Hogan and Sternad 2009, Balasubramanian et al. 2012), which is problematic since the temporal separation of submovements is strongly linked to overall movement smoothness. Additionally, the ordinal scale of this metric means that for movements with few submovements, a small change in the number of peaks reflects a disproportionately large change in smoothness (Balasubramanian et al. 2012). This issue is clearly seen in the elevated MDC% values for nSUB (range: 25%-156%), even though the range of MDC values is only 0.18-3.03 (Appendix B.1). Ultimately, metrics that rely on the entire continuum of data (such as LDJ and SPARC) are more likely to respond consistently and accurately to changes in movement smoothness than metrics like nSUB that focus on only isolated features of the data (Hogan and Sternad 2009, Balasubramanian et al. 2012).

This study established the reliability of movement quality metrics for healthy adults, expanding on prior work by including a wider variety of movement quality metrics and functional ADLs. Additionally, we provided normative data for smoothness metrics during ADLs, which can be used as a reference for future studies. These values are provided for three filter cut-off frequencies to facilitate comparison with other studies that used different methodology. Additional work is needed to quantify the reliability of these measures in specific patient populations and their sensitivity to detect changes over time.

CHAPTER 4. Movement quality characteristics of body-powered and myoelectric prostheses during activities of daily living

4.1 Abstract

Movement quality characteristics may differ between body-powered (BP) and myoelectric (MYO) prostheses due to differences in the availability of sensory feedback and the level of effort required to actuate the terminal device. The purpose of this work was to compare movement quality between BP and MYO prostheses during activities of daily living (ADLs). Six transradial BP and/or MYO prosthesis users performed three ADLs, including moving a can from a low to high shelf (CAN), placing a pill in a pillbox (PILL), and placing a pushpin in a bulletin board (PIN). Six age- and sex-matched controls also participated. Each ADL was divided into reaching and manipulation phases. Movement quality for each phase was assessed based on duration, smoothness (log dimensionless jerk and spectral arc length), and straightness (index of curvature) and compared between BP and MYO prostheses. All movement phases were slower when performed with MYO prostheses compared to BP prostheses, except the reaching phase for PILL. Reaches performed with MYO prostheses were more curved for PIN, but less curved for CAN. Manipulation phases were less smooth when performed with MYO prostheses. Movement quality was generally lower for MYO prostheses compared to BP prostheses. These differences may be related to the availability of sensory feedback and the method of terminal device actuation for each prosthesis type. Task requirements may also affect movement quality differently between BP and MYO prostheses, depending on whether grasping, releasing, or transporting an object is involved.

4.2 Introduction

Upper limb prostheses have many limitations that users would like to see addressed (Biddiss et al. 2007, Cordella et al. 2016), so it is unsurprising that approximately 25% of individuals with upper limb loss ultimately choose not to use a prosthesis (Biddiss and Chau 2007c). Most clinically prescribed prostheses are either body-powered (BP) or myoelectric (MYO). BP prostheses rely on a harness and cable system to link motion of other parts of the body (commonly the shoulder and trunk) to motion of the terminal device, while MYO prostheses use muscle activity recorded from the residual limb to control the terminal device. Prosthesis prescription depends on many factors including the prosthetist's experience, insurance coverage, and the patient's preference and functional goals (Carey et al. 2015). At present, there is little empirical evidence to suggest that either BP or MYO prostheses offer a significant functional advantage (Carey et al. 2015, Carey et al. 2017).

Few studies have characterized how prosthesis use affects the quality of movement. It is well-established that reaching movements made by anatomical limbs are approximately straight with symmetric velocity profiles (Flash and Hogan 1985), have highly blended submovements (Rohrer et al. 2004), and are strongly temporally coupled with grasping (Jeannerod 1984). Although movements made with upper limb prostheses have been studied less extensively, it is evident that movement quality tends to be lower for prosthetic limbs than anatomical limbs. For example, transhumeral BP prosthesis users required more discrete corrective movements with the prosthetic limb than with the intact limb to reach a target during a dynamic pointing task (Doeringer and Hogan 1995). In comparison to the intact limb, a transradial BP prosthesis user demonstrated slower movement times with the prosthetic limb and delayed closure of the terminal device during a grasping task (Fraser and Wing 1981). Similarly, transradial MYO and transhumeral hybrid prosthesis users had asymmetric velocity profiles during the reaching phase and decreased coupling between reach and grasp movements (Bouwsema et al. 2010). However, these studies did not require prosthesis users to perform tasks that were similar to those that might be encountered in daily life. Only one study explored movement quality during ADLs, demonstrating that transradial prosthesis users made movements that were less smooth and more curved with the prosthetic limb than the intact limb or than healthy controls (Cowley et al. 2017).

This decreased movement quality for prosthetic limbs may result from changes to the feedforward and feedback control systems with prosthesis use (Metzger et al. 2010). The feedforward control system predicts motor commands based on prior movements and depends on an internal model of the arm that accounts for its inertial properties. Reaching may be affected if an individual has not adequately updated their internal model to account for the new inertial properties of their prosthetic limb. The feedback control system corrects for errors during movement based on exteroceptive and proprioceptive input. The availability of this sensory information may differ between BP and MYO prostheses because of how each device is operated. In BP prostheses, somatic and proprioceptive feedback can be directly transmitted between the prosthesis and the body via the harness and cable system. Information about the prosthesis joint position (or terminal device aperture), velocity or forces is relayed through the body's natural physiologic sensors in a manner similar to the exchange of information in a natural limb (Doubler and Childress 1984). Conversely, vision is the primary source of sensory feedback in MYO prostheses and users tend to fixate their gaze on the prosthetic hand during task performance (Bouwsema et al. 2012, Sobuh et al. 2014).

The goal of this work was to compare movement quality between BP and MYO prostheses during ADLs. Differences in the availability of sensory feedback for BP and MYO prostheses might contribute to differences in movement quality between each device. Specifically, we expected MYO prosthesis users would have lower movement quality compared to BP users due to their greater reliance on visual feedback. Alternatively, movement quality might be influenced by the level of effort required to open and close the terminal device. Since the motors in MYO prostheses are powered by batteries, activating the terminal device requires little physical effort beyond what is needed to create the control signals (i.e., contracting muscles in the residual limb). In contrast, the power source for BP prostheses is the user's own musculature. This mechanical method of actuation is more effortful and could contribute to reduced movement quality on tasks that require significant movement of the terminal device.

4.3 Methods

4.3.1 Subjects

Six adults with unilateral transradial limb loss participated in this institutionally approved study (Table 2.1). Each participant had at least 6 months of experience using prostheses. Age- and sex-matched controls without upper limb loss were recruited from an online database (https://umhealthresearch.org/). Exclusion criteria for both groups included a history of other serious musculoskeletal, neurological, or visual impairments (other than upper limb loss).

4.3.2 Experimental protocol

Participants performed three ADLs at a comfortable pace (Table 4.1). Each ADL was repeated five times using the intact/dominant and prosthetic/nondominant limb. Prior to collection, prosthesis users were allowed to practice each ADL as long as they needed until they felt comfortable with it. Reflective markers on the wrists (radial and ulnar styloid processes, or approximate locations on the prosthesis socket) were tracked at 120 Hz using a 19-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA).

Table 4.1. Segmentation criteria for activities of daily living.
Each ADL was divided into reaching and manipulation phases based on position or velocity of the wrist joint center (WJC).

ADL	Procedure	Phase Type	Segmentation Definition
CAN	Participants moved a can* from a low (0.94 m) shelf to a shelf at shoulder height	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Anteroposterior velocity of WJC returns below 5 cm/s
		Manipulation	START: Anteroposterior velocity of WJC returns below 5 cm/s STOP: Anteroposterior velocity of WJC exceeds 5 cm/s
PILL	Participants held a pill while seated at a table. They placed the pill in a pillbox located at the midline of the body and in between the hands.	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Mediolateral velocity of WJC returns below 5 cm/s
		Manipulation	START: Mediolateral velocity of WJC returns below 5 cm/s STOP: Mediolateral velocity of WJC exceeds 5 cm/s
PIN	Participants placed a pushpin in a 1-inch diameter target drawn on a piece of corkboard placed at shoulder height along the midline of the body.	Reaching	START: Velocity of WJC exceeds 5 cm/s STOP: Anteroposterior velocity of WJC returns below 5 cm/s
		Manipulation	START: Anteroposterior velocity of WJC returns below 5 cm/s STOP: Anteroposterior velocity of WJC exceeds 5 cm/s

*different can sizes were used based on grasp aperture available with prosthesis

Prosthesis users estimated how many days per week and hours per day that they used their prosthesis (Table 2.1). The number of hours of prosthesis use per week was determined by multiplying their estimates of the number of days per week and hours per day that they use their prosthesis. They also completed the Trinity Amputation and Prosthesis Experience Scales – Revised (TAPES-R), which assesses adjustment to the psychosocial and physical demands of wearing a prosthesis (Gallagher et al. 2010). The TAPES-R contains three subscales that separately address psychosocial adjustment, activity restriction, and satisfaction with the prosthesis. The prosthesis satisfaction subscale is further divided into two parts that assess aesthetic (color, shape, appearance) and functional (weight, usefulness, reliability, fit, comfort) satisfaction using a three-point scale (not satisfied, satisfied, very satisfied). A separate question asks about general satisfaction with the prosthesis on a ten-point scale. Higher scores on all three scales reflect greater satisfaction. The TAPES-R was scored by summing responses to questions on the aesthetic satisfaction and functional satisfaction subscales.

4.3.3 Data analysis

Analysis focused on the wrist joint center (defined as the midpoint between the styloid markers), as this was the most distal anatomical reference point shared between the intact and prosthetic limbs. Marker data were filtered in Visual 3D (C-Motion, Germantown, MA) using a fourth-order low-pass Butterworth filter with a 10 Hz cutoff frequency. We calculated velocity of the wrist joint center by taking the derivative, low-pass filtering (Salmond et al. 2017), and calculating the vector sum of derivatives across all three planes of movement. These steps were repeated for second and third derivatives.

Each ADL was divided into reaching and manipulation phases, resulting in six movements for analysis (Table 4.1). The movement quality for reaching phases was assessed based on duration, straightness, and smoothness. For the manipulation phases, movement quality was assessed based on duration and smoothness.

Straightness was quantified using index of curvature (IOC), which is the total length of the path travelled by the wrist joint center divided by the length of a straight line between the initial and final position (Cowley et al. 2017). Smoothness was quantified using log dimensionless jerk (LDJ) (Equation 3.1) and spectral arc length (SPARC) (Equation 3.2). To reiterate, LDJ was the natural logarithm of the dimensionless integrated squared jerk:

$$LDJ := \ln\left(\frac{(t_2-t_1)^5}{A^2}\int_{t_1}^{t_2}\dddot{x}(t)^2 dt\right),$$

where $x(t)$ was the position of the endpoint (wrist joint center or center of gravity), A was the arc length, t_1 was the movement start time, and t_2 was the movement end time. SPARC was the arc length of the Fourier magnitude spectrum of the speed profile (Balasubramanian et al. 2015):

$$SPARC := \int_0^{\omega_c}\sqrt{\left(\frac{1}{\omega_c}\right)^2 + \left(\frac{d\hat{V}(\omega)}{d\omega}\right)^2}\, d\omega;$$

$$\hat{V}(\omega) := \frac{V(\omega)}{V(0)};$$

$$\omega_c := \min\{\omega_c^{max}, \min\{\omega, \hat{V}(r) < \overline{V}_{threshold}\ \forall\, r > \omega\}\},$$

where $\hat{V}(\omega)$ was the Fourier magnitude spectrum normalized to the DC magnitude ($V(0)$) and ω_c was an adaptively selected cutoff frequency.

4.3.4 Statistics

We compared the movement quality metrics for all movements using a one-way ANOVA to check for differences between the prosthesis users' intact limbs and the controls' dominant and nondominant limbs. There was only one significant difference (IOC for the reaching phase of the PILL task). Post-hoc tests revealed that the dominant and nondominant limbs were not significantly different, but the intact limb had significantly smaller IOC than the nondominant limb. As the primary purpose of this study was to compare movement quality for BP and MYO prostheses, we combined the data into a single category representing all anatomical limbs.

The magnitude of difference between movement quality metrics and self-reported metrics for BP and MYO prosthesis users were assessed using Hedges' g as a measure of effect size (Equation 2.1). Positive values for g indicate that the mean for MYO prostheses was higher than the mean for BP prostheses. We also calculated effect sizes to check for differences between anatomical limbs and prosthetic limbs (represented as the average of the BP and MYO prostheses). Positive values for g indicate that the mean for prosthetic limbs was higher than the mean for anatomical limbs. Only comparisons with effect sizes ≥ 0.2 are reported.

Given the small sample size for this study, we conducted a sensitivity power analysis to determine what level of effect could be expected at a larger sample size. For comparison, we also conducted a post-hoc power analysis ($\alpha = 0.05$, power = 0.8) to demonstrate the statistical power of the tests performed using the current sample size.

4.4 Results

P02 was unable to perform the CAN task due to insufficient grasp aperture. All other participants were able to perform all tasks.

4.4.1 Comparison of prosthetic and anatomical limbs

Movement quality was poorer for prosthetic limbs compared to anatomical limbs (i.e., movements made with prostheses were slower, more curved, and less smooth) across nearly all metrics and movements. The effect size was very large for the majority of these comparisons ($0.95 \leq g \leq 4.64$). There was also a small effect size ($g = 0.35$) and a medium effect size ($g = 0.77$) for SPARC for the reaching phases of the PILL and PIN tasks, respectively. Only one comparison, IOC for the reaching phase of the PILL task, had a negligible effect size ($g = 0.13$).

4.4.2 Comparison of BP and MYO prostheses

In general, movements made with MYO prostheses were had lower quality compared to those with BP prostheses (Table C.1 in Appendix C). This was true for both reaching (Figure 4.1A) and manipulation (Figure 4.1B) movements. Movements made with MYO prostheses were longer than those made with BP prostheses for both the reaching ($g = 0.48$) and manipulation ($g = 0.35$)

phases of the CAN task, as well as the reaching (g = 0.79) and manipulation (g = 1.82) phases of the PIN task. The manipulation phase of the PILL task was also longer for MYO prostheses (g = 0.83). Compared to reaching movements made with BP prostheses, those with MYO prostheses were more curved for the PIN task (g = 0.41) but less curved for the CAN task (g = -0.62).

Reaching movements performed with BP prostheses had higher LDJ for the CAN (g = -1.02) and PILL (g = -0.62) tasks, while manipulation movements performed with MYO prostheses had higher LDJ for the PILL (g = 0.55) and PIN (g = 1.25) tasks. Reaching movements performed with MYO prostheses had higher SPARC for the PILL (g = 0.36) and PIN (g = 0.43) tasks. Similarly, manipulation movements performed with MYO prostheses had higher SPARC for the the CAN (g = 0.33), PILL (g = 2.71), and PIN (g = 0.38) tasks.

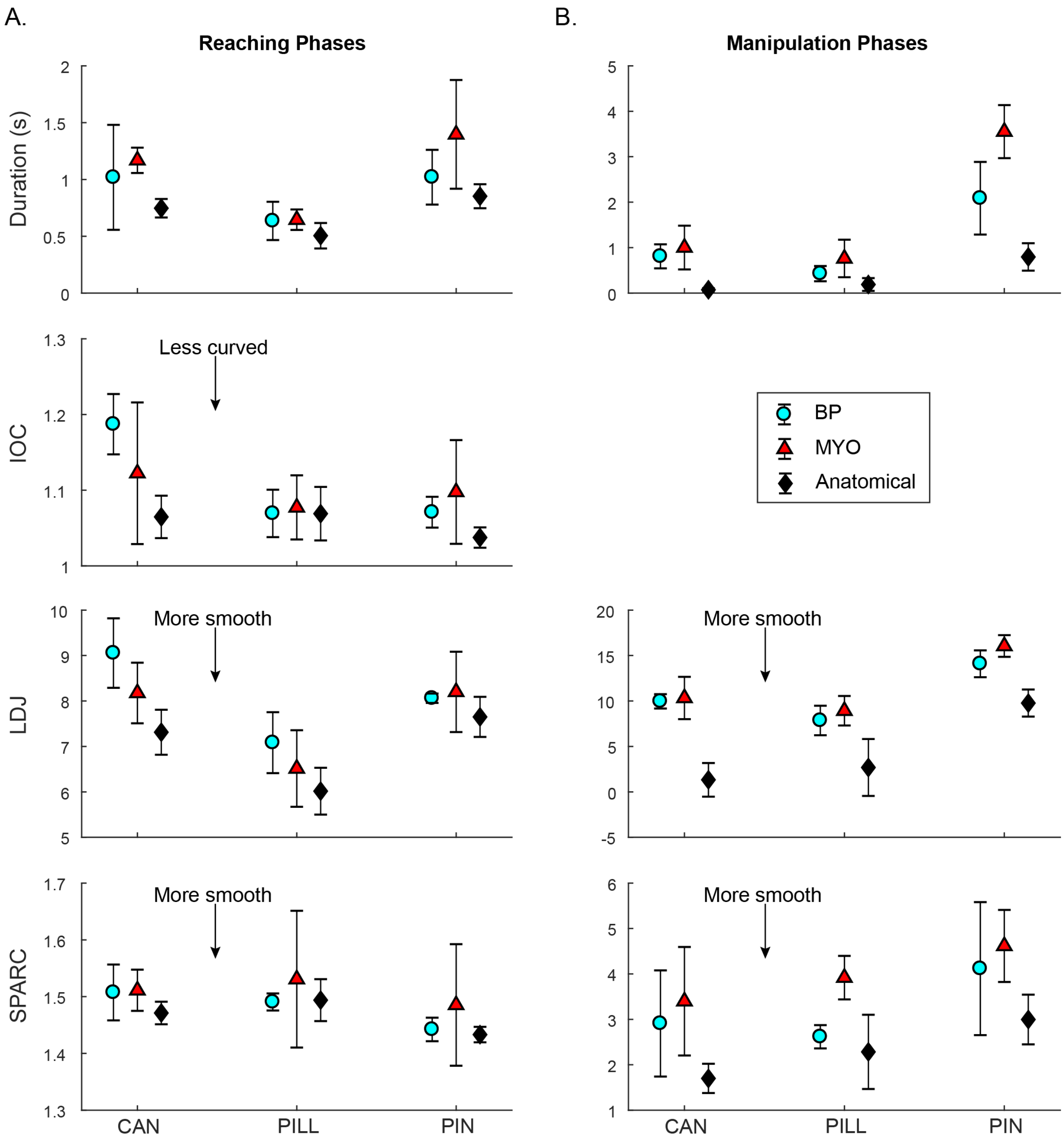


Figure 4.1. Differences in movement quality between intact limbs and body-powered and myoelectric prostheses.
Mean movement quality metrics for the prosthetic and anatomical limbs during the reaching (A) and manipulation (B) phases of the CAN, PILL, and PIN tasks. Error bars represent standard deviation.

4.4.3 Self-reported outcomes

In general, BP prosthesis users reported more frequent use of their prosthesis (51.3 ± 17.2 hours/week) than MYO prosthesis users (25.5 ± 11.8 hours/week). The effect size for this difference was large (g = 1.81). Although BP and MYO prosthesis users reported similar satisfaction (Figure 4.2) according to the general (g = 0.19) and aesthetic TAPES-R subscales (g = 0.00), BP users expressed greater satisfaction with their functional abilities compared to MYO users (g = 1.48).

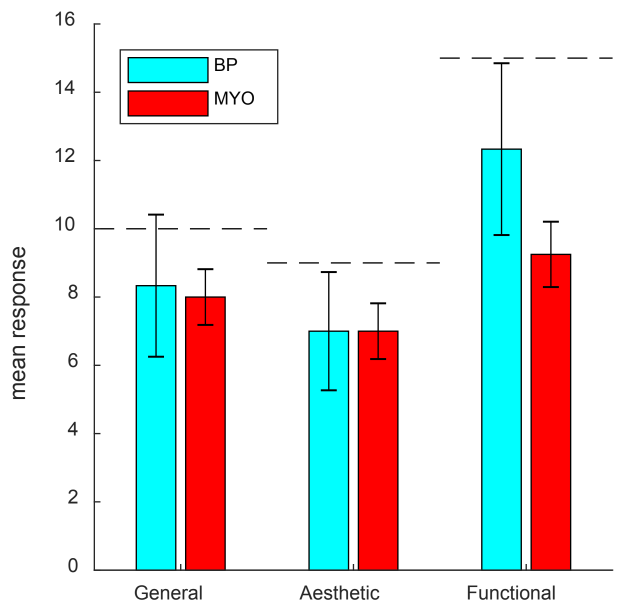


Figure 4.2. User satisfaction (TAPES-R) with body-powered and myoelectric prostheses. Mean scores for the aesthetic and functional satisfaction subscales and the single question about general prosthesis satisfaction from the TAPES-R. Dashed lines indicate the maximum possible score for each subscale.

4.4.4 Sensitivity power analysis

The sensitivity power analysis revealed that with the current sample size of 3 BP users and 4 MYO, the power is 0.05 to detect a small effect, 0.08 to detect a medium effect, and 0.14 to detect a large effect. The majority of the effect sizes calculated for this study were small or medium. For the two largest effect sizes ($g = 1.82$ and $g = 2.71$), the power was 0.60 and 0.93, respectively.

4.5 Discussion

The purpose of this study was to compare movement quality between ADLs performed with BP and MYO prostheses. Specifically, we hypothesized that MYO prosthesis users would make movements that were slower, less smooth, and more curved compared to BP prosthesis users. Although the effect sizes vary for each comparison, our results generally support this hypothesis. This finding is consistent with prior studies, which have demonstrated that prosthesis users make movements that are slower (Fraser and Wing 1981, Stein and Walley 1983, Cowley et al. 2017), less smooth (Doeringer and Hogan 1995, Bouwsema et al. 2010, Cowley et al. 2017), and more curved (Cowley et al. 2017) than those made by anatomical limbs.

Our finding that movement quality was worse for MYO users compared to BP users is concordant with participants' self-reported assessments of their prosthesis satisfaction and frequency of prosthesis use. In particular, MYO users reported less frequent prosthesis use and lower functional satisfaction in comparison to BP users. Satisfaction measured by the TAPES-R has been reported elsewhere for men with transradial limb loss using BP and MYO prostheses (Hafshejani et al. 2012a, Hafshejani et al. 2012b), but these studies found higher satisfaction among MYO users. Further study with a larger sample size is needed to explain these discrepancies.

Although our hypotheses was generally supported, we found a few unexpected trends. For example, reaching movements were more curved for the PIN task when performed with a MYO prosthesis but were more curved for the CAN task when performed with a BP prosthesis. This finding may be related to differences in task requirements for the reaching phase of the CAN task as opposed to the reaching phase for the PILL and PIN tasks. Participants had to actively prepare for grasping an object during the reaching phase of the CAN task, but they were already holding objects during the reaching phase for the PILL and PIN tasks (i.e., they were transporting the objects to a final location). Opening the terminal device on a BP prosthesis requires significant physical effort since movement of the terminal device is mechanically linked with a cable to movement of the shoulder and trunk. Pulling this cable may induce deviation from a straight-line trajectory during the reach, which was visually apparent in video of the BP users performing this task. In contrast, the terminal device on a MYO prosthesis can be opened simply by contracting muscles in the residual limb, which should not generate additional movement. It is reasonable to expect that the IOC would reflect these differences, as additional movement from opening the terminal device would increase path length travelled by the wrist joint center (thus increasing the IOC).

Another unexpected finding was that the reaching phases for the CAN and PILL tasks had higher LDJ values when performed with a BP prosthesis, indicating that the movements were less smooth. Conversely, the manipulation phases for the PILL and PIN tasks had higher LDJ values when performed with a MYO prosthesis and all movements had higher SPARC values when performed with MYO prostheses. The fact that these differences were not apparent from both smoothness metrics (LDJ and SPARC) relates to how the metrics are calculated. Since LDJ is based on the rate of change of acceleration in a movement (i.e., jerk), changes in acceleration associated with activating the terminal device should be reflected in the metric. It should also be reflected in the SPARC metric, which is based on the Fourier magnitude spectrum of the speed profile, presumably as increased magnitude of a higher frequency component. However, SPARC calculation inherently involves low-pass filtering due to the selection of an adaptive cutoff frequency (ω_c) that limits high frequency content if the normalized magnitude is below a certain threshold ($\overline{V}_{threshold} = 0.05$). It is possible that any additional high frequency content related to activating the terminal device was filtered out.

In addition to the differences in how the terminal devices are activated, differences in the availability of sensory feedback for BP and MYO prostheses might contribute to the observed differences in movement quality. Although BP prostheses offer proprioceptive feedback to the user as a result of the mechanical control system, MYO users are forced to rely primarily on visual feedback. Since visual feedback is slower than proprioception and requires additional cognitive processing (Gonzalez et al. 2012), it makes sense that MYO users would make slower movements compared to BP users. Indeed, lower speed of task completion for MYO users compared to BP users has been reported elsewhere (Stein and Walley 1983). Reduced movement speed could also relate to electromechanical delays associated with operating a motorized hand (Chadwell et al. 2016), possibly affecting the manipulation phases, (i.e., grasping in the CAN task and releasing in the PILL and PIN tasks). In particular, we observed that the MYO users tended to have trouble holding the pin securely, which may have made them slower and more cautious when pushing it into the corkboard. Furthermore, reliance on visual feedback could make it more difficult for MYO users to make small corrections for error during movement. This could relate to why MYO users had consistently had poorer movement quality for the manipulation phases of the PILL and PIN tasks. These tasks had high accuracy requirements since the objects had to be placed within a small target (0.9 in^2 slot for PILL; 1 inch diameter circle for PIN).

In general, our findings are likely influenced by the small sample size. There was considerable variability between subjects, especially between the MYO users. Including more participants may help reduce the between-subject variability. With the current sample size, only five of the effect sizes were large ($g \geq 0.8$) and the statistical power was very low. Participant recruitment is ongoing, so future work will focus on exploring these trends in the context of a larger sample size.

This study has demonstrated that movement quality during activities of daily living tends to be lower for MYO prostheses than BP prostheses. In general, movements performed with MYO prostheses were slower, less smooth, and more curved. These differences are likely related to the availability of sensory feedback and the method of terminal device activation for each prosthesis type. Task requirements may also affect movement quality differently between BP and MYO prostheses.

CHAPTER 5. Compensatory movements in body-powered and myoelectric prosthesis users during activities of daily living

5.1 Abstract

Upper limb prosthesis use involves compensatory strategies, but there is limited understanding of how prosthesis type affects these kinematic patterns. The purpose of this work was to compare compensatory trunk movements between body-powered (BP) and myoelectric (MYO) prostheses during activities of daily living (ADLs). Six transradial BP and/or MYO prosthesis users performed three ADLs, including moving a can from a low to high shelf, placing a pill in a pillbox, and placing a pushpin in a bulletin board. Six age- and sex-matched controls also participated. Three-dimensional peak-to-peak range of motion (ROM) for the trunk was calculated for each task. Compensatory strategies varied based on task requirements. Trunk ROM in all three planes was generally larger for BP prostheses compared to MYO prostheses, although axial rotation and flexion were greater for MYO prostheses during the pin task. However, there was considerable variability between participants. Increased ROM for BP prostheses may be related to additional compensatory movements induced by actuating the terminal device with a cable. Compensatory movements may also be affected by external factors such as training, socket fit, and choice of terminal device.

5.2 Introduction

Currently, the two most common types of upper limb prostheses are body-powered (BP) or myoelectric (MYO). BP prostheses are biomechanically powered, such that movement of proximal joints (typically the shoulder and/or trunk) is translated to movement of a terminal device via a harness and cable system. MYO prostheses are externally powered and residual limb muscle activity is used to control the terminal device. Both BP and MYO prostheses have significant technical shortcomings. Although some prosthetic components enable active wrist movement and use of multiple grasp patterns, most have limited degrees of freedom in the distal joints. The inability to control these joints, especially the wrist, requires prosthesis users to employ different kinematic strategies with the proximal joints or rely on the intact limb to accomplish activities of daily living (ADLs). Overreliance on the intact arm and compensatory movement of proximal joints likely contributes to the increased prevalence of overuse injuries among individuals with upper limb loss (Datta et al. 2004, Østlie et al. 2011).

Previous studies have demonstrated that compensatory strategies generally involve the trunk and shoulder. For example, transradial MYO prosthesis users had increased trunk movement compared to controls during ADLs that required forearm and wrist rotation (opening a door and lifting a box) (Carey et al. 2008). Similarly, transradial MYO prosthesis users required greater shoulder abduction and trunk motion in all three planes compared to controls to perform a series of goal-oriented tasks from the Southampton Hand Assessment Procedure (Major et al. 2014). During functional reaching tasks in the anteroposterior, mediolateral and vertical planes of motion, transradial and transhumeral BP and MYO prosthesis users had increased shoulder path distance and trunk movement in all three planes compared to controls (Metzger et al. 2012). A case study of a transradial single degree-of-freedom MYO prosthesis user also showed greater range of motion for the trunk and head compared to controls during performance of a clothespin relocation task (Hussaini and Kyberd 2017).

Given the differences in control strategies between BP and MYO prostheses, it seems likely that compensatory strategies will also vary based on which device is used. BP prostheses require movement of the shoulder and trunk to activate the terminal device, which may induce additional compensatory movement beyond what is required to make up for the limited distal degrees of freedom. However, only two case studies have directly compared compensatory movements between BP and MYO prosthesis users. One individual with transradial limb loss was found to use more elbow flexion when drinking from a cup and opening a door using a BP prosthesis than a MYO prosthesis (Carey et al. 2009). This individual also used more shoulder flexion when opening the door with the BP prosthesis, but more shoulder flexion when drinking with the MYO prosthesis. A separate study quantified compensatory movements of an individual with transhumeral limb loss performing a modified Box and Blocks task with different prostheses (Hebert and Lewicke 2012). The individual displayed less trunk movement when using a MYO prosthesis than when using a BP prosthesis.

While these studies represent an important step towards understanding how compensatory strategies vary between BP and MYO prostheses, they included a very small range of tasks. It is difficult to generalize these findings to other tasks that may have different functional requirements, such as reaching in different planes or manipulating objects of different sizes and shapes. Therefore, the purpose of this study was to characterize the compensatory trunk movements involved with transradial BP and MYO prosthesis use in a set of common ADLs.

5.3 Methods

5.3.1 Subjects

Six adults with unilateral transradial limb loss participated in this institutionally approved study (Table 2.1). Each participant had at least 6 months of experience using prostheses. Age- and sex-matched controls without upper limb loss were recruited from an online database (https://umhealthresearch.org/). Exclusion criteria for both groups included a history of other serious musculoskeletal, neurological, or visual impairments (other than upper limb loss).

5.3.2 Experimental protocol

Participants performed a series of ADLs using the intact/dominant and prosthetic/nondominant limbs. These ADLs included moving a can from a low shelf to a high shelf, placing a pill in a pillbox, and placing a pushpin in a corkboard (Table 4.1). Each ADL was repeated five times at a comfortable pace. Prior to collection, prosthesis users were allowed to practice each ADL as long as they needed until they felt comfortable with it.

The motions of seven body segments were tracked at 120 Hz using a 19 camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) and 24 reflective markers. Anatomical markers were placed on the acromion processes, medial and lateral humeral epicondyles, and radial and ulnar styloids for a static trial. Upper arm and forearm motion was subsequently tracked using clusters of four and three markers, respectively. Four markers were placed on the trunk (7th cervical vertebra, 8th thoracic vertebra, sternal notch, and xiphoid process) and three on the hands (3rd and 5th metacarpal heads and base of the 3rd metacarpal). For prosthesis users, the approximate locations of the epicondyles and styloids were estimated on the prosthesis socket. Hand markers were excluded on the prosthetic limb for BP users and placed in approximate locations for MYO users.

5.3.3 Data analysis

Marker position data were filtered in Visual 3D (C-Motion, Germantown, MA) using a fourth-order low-pass Butterworth filter with a 6 Hz cutoff frequency. A model containing hand (if applicable), forearm, upper arm, and trunk segments was created using the joint centers and local coordinate systems defined in (Gates et al. 2016). Trunk-room angles (lateral lean, axial rotation, flexion) were calculated using Euler angle decomposition according to rotation sequences recommended by the International Society of Biomechanics (Wu et al. 2005). Axes for lateral lean and axial rotation were flipped for movements performed with the left arm in order to match movements performed with the right arm. Under this convention, positive lateral lean indicates bending to the prosthetic/dominant side, positive axial rotation indicates turning towards the intact/nondominant side, and positive flexion angles indicate trunk extension.

A 5 cm/s velocity threshold for the wrist joint center (defined as the midpoint between the styloid markers) was used to define the beginning and end times of each repetition, which were also verified visually. Joint angle waveforms were then time-normalized to 100% of task completion. Peak-to-peak range of motion (ROM) was determined for the trunk in three dimensions: lateral lean, axial rotation, and flexion.

5.3.4 Statistics

We compared the trunk ROM for movements with prosthesis users' intact limbs and the controls' dominant and nondominant limbs using a one-way ANOVA. No significant differences were found, so we combined the data into a single category representing all anatomical limbs.

The magnitude of difference between trunk ROM for BP and MYO prosthesis users were assessed using Hedges' g as a measure of effect size (Equation 2.1). Positive values for g indicate that the mean for MYO prostheses was higher than the mean for BP prostheses. We also calculated effect sizes to check for differences between anatomical limbs and prosthetic limbs (represented as the average of the BP and MYO prostheses). Positive values for g indicate that the mean for prosthetic limbs was higher than the mean for anatomical limbs. Only comparisons with effect sizes ≥ 0.2 are reported.

5.4 Results

P02 was unable to perform the CAN task due to insufficient grasp aperture. All other participants were able to perform all tasks.

5.4.1 Comparison of prosthetic and anatomical limbs

Trunk ROM was greater in all three dimensions for each task when performed with prosthetic limbs compared to anatomical limbs. The effect size was very large for all comparisons ($1.15 \leq g \leq 3.37$).

5.4.2 Comparison of BP and MYO prostheses

In general, trunk ROM was greater for BP prostheses compared to MYO prostheses (Figure 5.1, Figure 5.2, Figure 5.3, Table D.1in Appendix D). ROM for lateral lean (g = -1.52), axial rotation (g = -1.45), and flexion (g = -0.90) were all greater for BP prostheses during the CAN task (Figure 5.1). ROM for axial rotation (g = -1.42) and flexion (g = -0.44) were all greater for BP prostheses during the PILL task (Figure 5.2). ROM for lateral lean was greater for BP prostheses during the PIN task (g = -1.45). However, ROM for axial rotation (g = 0.52) and flexion (g = 1.08) were greater for MYO prostheses (Figure 5.3).

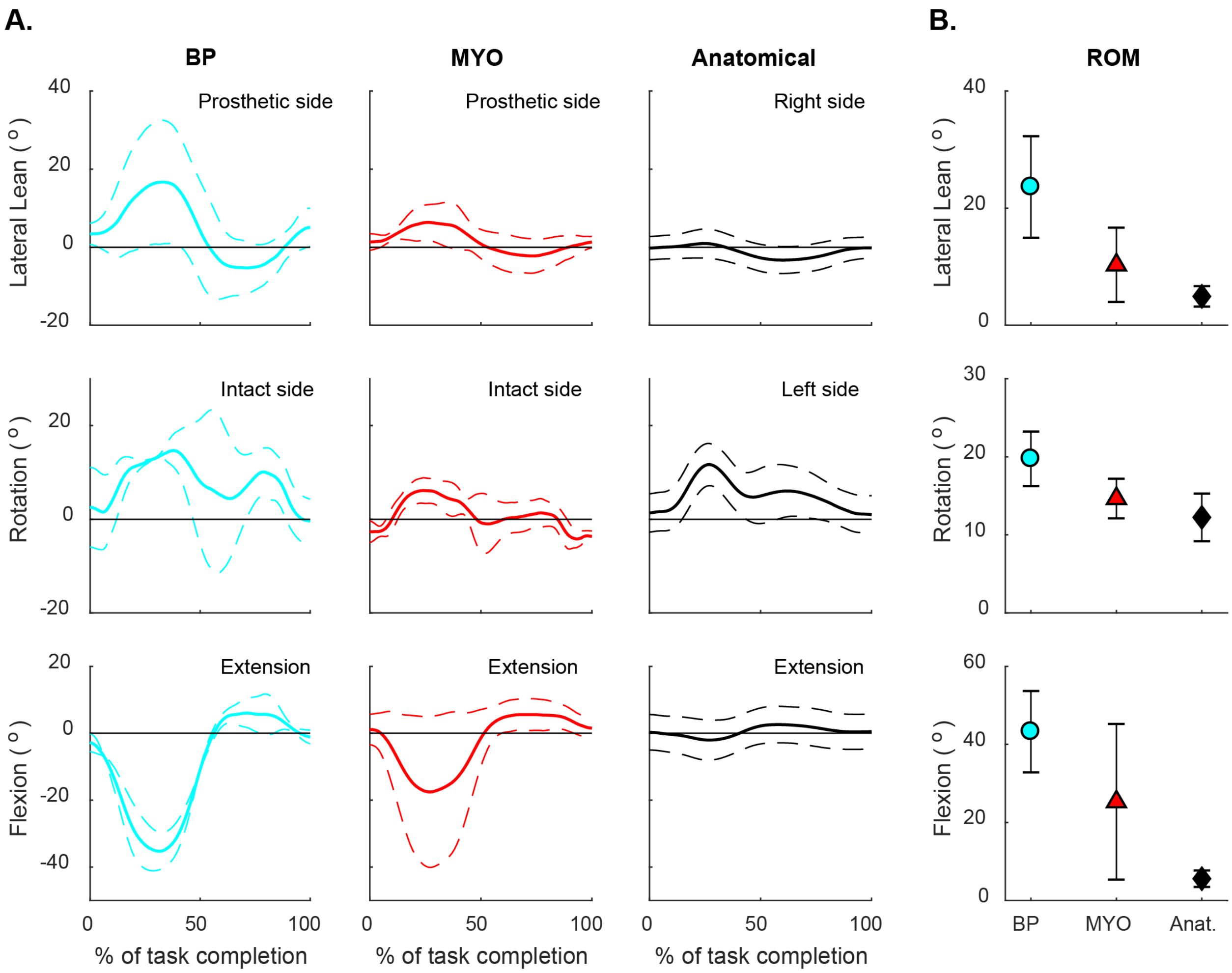


Figure 5.1. Kinematic patterns and ROM for the CAN task.
(A) The average (solid line) and standard deviation (dashed line) of the trunk angles for the CAN task. Note that for movements performed with the left arm, the lateral lean and axial rotation axes have been flipped to match movements performed with the right arm. (B) The average peak-to-peak ROM for the CAN task. Error bars represent standard deviation.

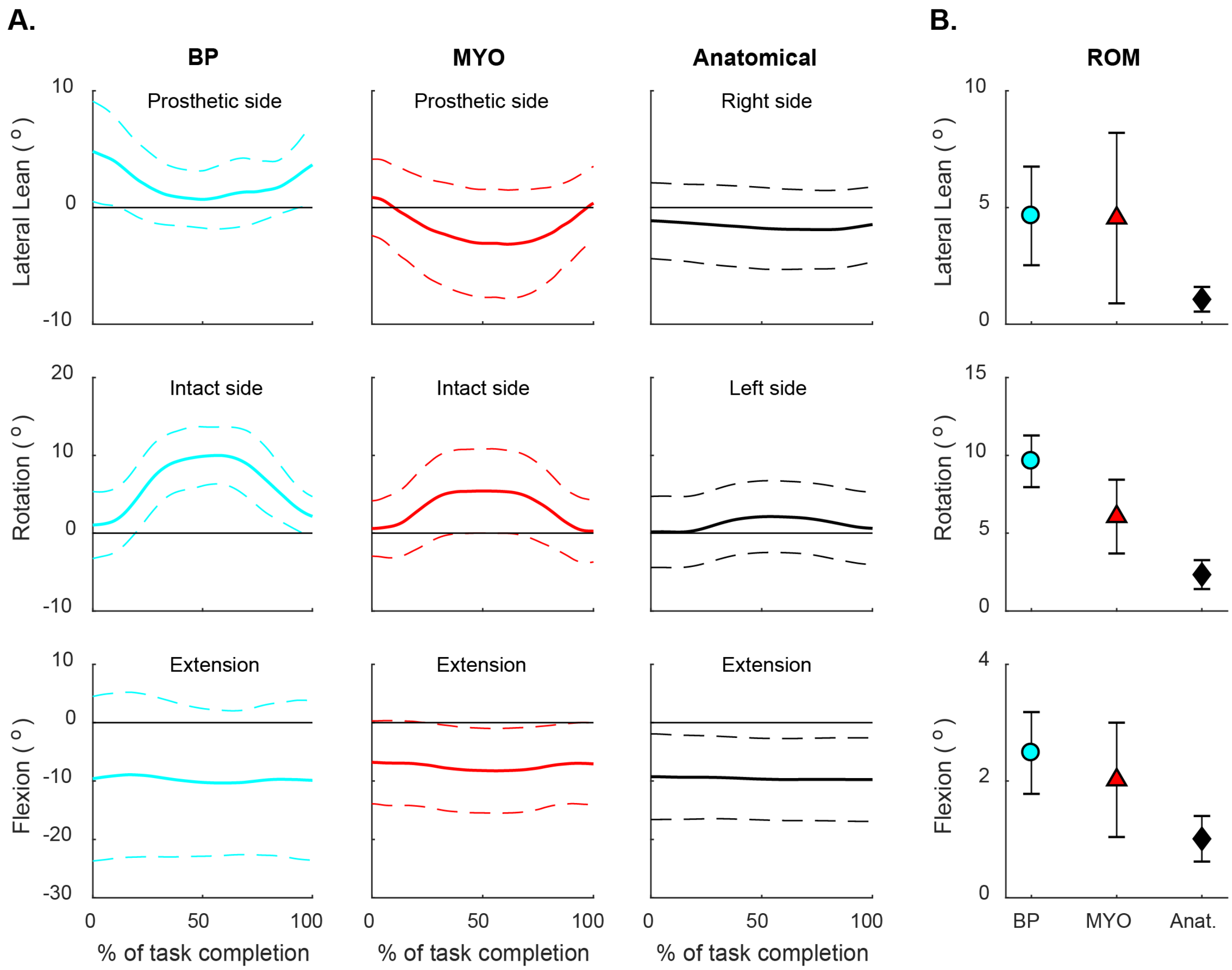


Figure 5.2. Kinematic patterns and ROM for the PILL task.
(A) The average (solid line) and standard deviation (dashed line) of the trunk angles for the PILL task. Note that for movements performed with the left arm, the lateral lean and axial rotation axes have been flipped to match movements performed with the right arm. (B) The average peak-to-peak ROM for the PILL task. Error bars represent standard deviation.

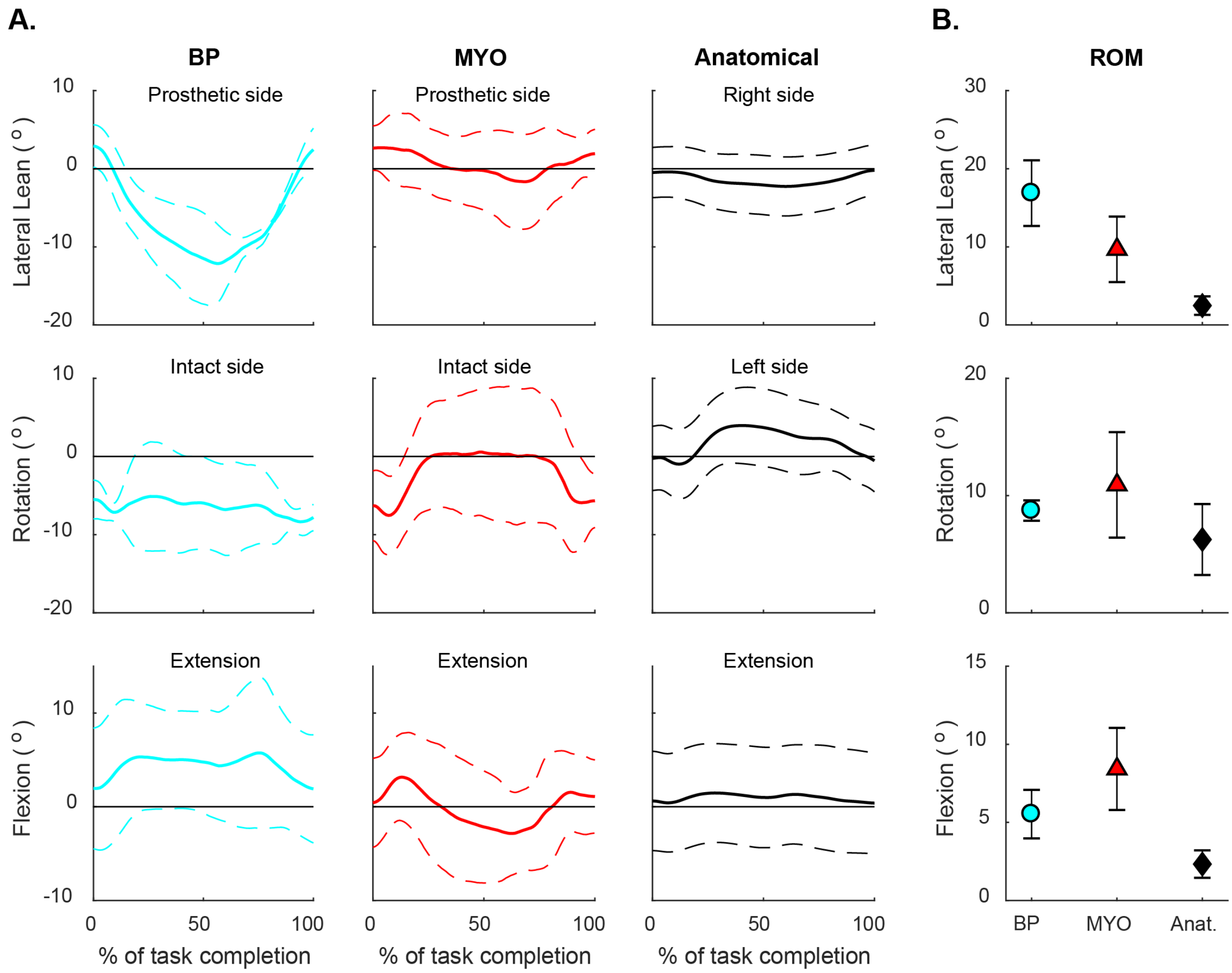


Figure 5.3. Kinematic patterns and ROM for the PIN task.
(A) The average (solid line) and standard deviation (dashed line) of the trunk angles for the PIN task. Note that for movements performed with the left arm, the lateral lean and axial rotation axes have been flipped to match movements performed with the right arm. (B) The average peak-to-peak ROM for the PIN task. Error bars represent standard deviation.

5.5 Discussion

This study quantified the compensatory trunk movements involved with BP and MYO prosthesis use in a set of common ADLs. Our results demonstrated that prosthesis users consistently required larger ROM of the trunk compared to controls. Furthermore, BP users required a larger ROM than MYO users for the majority of tasks.

There is considerable evidence in the literature supporting the fact that prosthesis users employ larger ROM of the trunk during ADLs (Carey et al. 2008, Metzger et al. 2012, Major et al.

2014, Hussaini and Kyberd 2017). Thus, our results showing increased trunk ROM for all prosthesis users corroborate previous findings. However, most prior work did not directly compare compensatory strategies between BP and MYO users. Only one of these studies included trunk angles in a case study of an individual with transhumeral limb loss (Hebert and Lewicke 2012), making it difficult to compare with our sample of individuals affected at the transradial level.

The larger ROM for BP prostheses compared to MYO prostheses is likely due to additional compensations induced by activating the cable system. Since BP prostheses are activated through shoulder flexion and/or scapular abduction, the act of opening the terminal device inherently affects its overall position. Additional compensatory movement is then required to maneuver the terminal device into the appropriate location to accomplish a given task. In contrast, MYO prostheses can be actuated simply by contracting muscles in the residual limb, which should not affect the overall position of the terminal device.

The CAN task required the largest ROM for trunk flexion, which was true for both BP and MYO prostheses. This flexion peak occurred during the phase of movement where participants were grasping the can. Flexion also coincided with a lateral lean towards the prosthetic side, as well as turning towards the intact side. These movements reflect the significant compensation needed to orient the terminal device appropriately given the lack of active forearm supination and wrist extension available in the prostheses.

The PIN task also involved significant trunk compensations, particularly for lateral lean. BP users tended to lean towards the intact side, while MYO users were able to maintain a more neutral angle. However, MYO users had greater ROM for axial rotation and flexion compared to BP users. It is visually apparent from video of the participants that BP uses employed less elbow flexion than MYO users. Keeping the arm more fully extended may have made it difficult to rotate or flex the trunk without colliding the arm with the shelf holding the corkboard, so the primary compensation would necessarily be lateral lean. In contrast, the MYO users flexed their elbows close to 90° when placing the pin in the corkboard. Rotating and flexing the trunk would therefore be more helpful in positioning the terminal device. As we have noted previously, additional exploration of the connection between elbow flexion and trunk movement is still needed.

The PILL task had a very small ROM overall, which is expected given that the task required minimal reaching to accomplish it. All prosthesis users rotated towards the intact side, likely to compensate for the lack of wrist extension in the terminal device. The BP users had a slightly larger axial rotation ROM in this direction (about 4°), however.

There are several important caveats when interpreting these results. First, there was considerable variability between participants in terms of the overall patterns of the joint angles, as well as the temporal location of peaks. This is a well-known problem for upper limb biomechanical analysis, as the kinematic redundancy of the upper limb makes it possible to accomplish any task using a variety of movement strategies (van Andel et al. 2008). Time-normalization of joint angle waveforms and averaging across trials and/or participants can therefore mask important features of the curves. For this reason, some have argued against the practice of time-normalization in favor of curve registration techniques such as dynamic time warping (Thies et al. 2017), which better preserves signal features and allows for separate quantification of temporal and amplitude variability. Our graphs of the overall kinematic patterns were certainly affected by this issue, which is evident from the large standard deviations seen in the trajectories. This creates some ambiguity for interpreting compensatory strategies employed by the prosthesis users. However, the peak-to-peak ROM was calculated based on individual curves before being averaged and is not affected by time-normalization. There was still considerable variability between participants, especially for

MYO users, which may be a consequence of the small sample size used in this study. Further study with a larger sample size is clearly needed.

A second limitation of this study is that differences in compensatory strategies cannot be linked to a single cause. The mechanisms of prosthesis actuation likely play a significant role, but other factors such as training, socket fit, and choice of terminal device are also relevant. Although the goal of training is commonly to minimize compensatory strategies as much as possible, it has also been suggested that training should instead focus on developing functional skills even if compensatory movement is involved (Bouwsema et al. 2012). Ultimately, training is highly individualized based on an individual's needs and abilities (Smurr et al. 2008) so long-term outcomes may vary considerably. Although we did not formally assess participants' with training and occupational therapy, it is worth noting that several participants commented that they had never performed these tasks with their prosthesis prior to the study. Participants were allowed to briefly practice the tasks, but it is possible that more extensive practice or training with the prosthesis could have changed their compensatory strategies. Indeed, a case-study demonstrated functional improvements following extensive training, even though the participant was considered to be an experienced prosthesis user (3 years of use) (Dromerick et al. 2008)

Characteristics of the prosthesis itself may also affect compensatory movements. For example, pain from an ill-fitting harness or socket may force users to adopt different kinematic strategies. One previous study on compensatory movement acknowledged this issue and assessed comfort among participants (Major et al. 2014), but found high levels of comfort in all participants and did not draw any further conclusions. Choice of terminal device is also relevant. There are significant mechanical differences between BP split hooks, single degree-of-freedom MYO hands, and multi-articulated MYO hands. Even within each category of device, there are differences across manufacturers and models. For example, activation forces and pinch forces vary across types of voluntary open split hooks and are also influenced by the number of bands that are used to hold the hooks closed (Smit and Plettenburg 2010, Smit et al. 2012). MYO hands vary as well, especially in terms of weight and grasp force (Belter et al. 2013). Terminal device variability across participants has been acknowledged as a limitation in previous studies of compensatory movement (Major et al. 2014). Although some papers have given details about which components their participants used, it is difficult to know how results may have been affected.

A final limitation of this study is that trunk angles alone do not fully represent a prosthesis user's compensatory strategies. The rest of the kinematic chain must also be considered, as trunk angles cannot be interpreted in isolation. For example, we have noted previously that trunk compensations are likely connected with elbow flexion during the PIN task. Similarly, other work has shown that elbow flexion can be more restricted by MYO prostheses than BP prostheses for certain tasks (Carey et al. 2009). Future work will include elbow and shoulder angles, which previous studies have shown to be affected by prosthesis use (Carey et al. 2008, Carey et al. 2009, Major et al. 2014).

This study has demonstrated that trunk ROM during activities of daily living tend to be larger for MYO prostheses than BP prostheses. This finding is likely related to differences in the way the terminal device is actuated since operating a BP prosthesis with a cable may induce additional compensatory movement. Task requirements may also affect the magnitude and direction of the compensatory movements differently between BP and MYO prostheses.

CHAPTER 6. Factors associated with interest in novel interfaces for upper limb prosthesis control[2]

6.1 Abstract

Surgically invasive interfaces for upper limb prosthesis control may allow users to operate advanced, multi-articulated devices. Given the potential medical risks of these invasive interfaces, it is important to understand what factors influence an individual's decision to try one. We conducted an anonymous online survey of individuals with upper limb loss. A total of 232 participants provided personal information (such as age, amputation level, etc.) and rated how likely they would be to try noninvasive (myoelectric) and invasive (targeted muscle reinnervation, peripheral nerve interfaces, cortical interfaces) interfaces for prosthesis control. Bivariate relationships between interest in each interface and 16 personal descriptors were examined. Significant variables from the bivariate analyses were then entered into multiple logistic regression models to predict interest in each interface. While many of the bivariate relationships were significant, only a few variables remained significant in the regression models. The regression models showed that participants were more likely to be interested in all interfaces if they had unilateral limb loss ($p \leq 0.001$, odds ratio ≥ 2.799). Participants were more likely to be interested in the three invasive interfaces if they were younger ($p < 0.001$, odds ratio ≤ 0.959) and had acquired limb loss ($p \leq 0.012$, odds ratio ≥ 3.287). Participants who used a myoelectric device were more likely to be interested in myoelectric control than those who did not ($p = 0.003$, odds

[2] A version of this chapter is published as Engdahl, S. M., et al. (2017). "Factors associated with interest in novel interfaces for upper limb prosthesis control." PLOS One **12**(8): e0182482.

ratio = 24.958). Novel prosthesis control interfaces may be accepted most readily by individuals who are young, have unilateral limb loss, and/or have acquired limb loss However, this analysis did not include all possible factors that may have influenced participant's opinions on the interfaces, so additional exploration is warranted.

6.2 Introduction

Despite the significant functional limitations that upper limb loss can impose, many individuals with upper limb loss choose not to use a prosthesis. The average prosthesis rejection rates reported in the literature are 26% for body-powered and 23% for myoelectric prostheses, although some estimates range upward of 50% (Biddiss and Chau 2007c). Among many other concerns, individuals with upper limb loss have reported a desire for prostheses with improved dexterity (including independent movement of the fingers and arm joints, increased range of motion, and wider variety of grasp patterns) (Biddiss et al. 2007, Kyberd and Hill 2011). The utility of such a prosthesis would be significant in comparison to most current commercially available prostheses, which permit only one degree of freedom (open/close) (Graimann and Dietl 2013, Resnik et al. 2014a) and can be cumbersome to use. Ultimately, this suggests that acceptance of a prosthesis may be improved if individuals with upper limb loss could be given multi-articulated prostheses that mimic the anatomic and physiologic complexity of the natural human arm. In fact, one survey reported that 68% of individuals who did not use a prosthesis were willing to reconsider using a prosthesis if improvements were made at a reasonable cost (Biddiss et al. 2007).

However, controlling a prosthesis with multiple degrees of freedom poses a significant technical challenge because it requires the collection of multiple independent control signals (Resnik et al. 2014b). The development of more advanced methods for prosthesis control is an active and rapidly-advancing area of research in which many options have been proposed. Here, we present an overview of the four primary categories of these methods: myoelectric control, targeted muscle reinnervation, peripheral nerve interfaces, and cortical interfaces. (A more detailed discussion may be found in (Ohnishi et al. 2007) or (Kung et al. 2013)). *Myoelectric control* refers to the use of electromyographic signals recorded from the skin surface over muscles in the residual limb. This method commonly relies on a "direct" control scheme in which signals from an agonist/antagonist pair of muscles are used to control a single degree of freedom in the prosthesis (Wurth and Hargrove 2014). It is generally possible to record only two independent signals from the residual limb (Ohnishi et al. 2007, Graimann and Dietl 2013) due to muscle cross-talk and co-activation, which limits the number of degrees of freedom that can be controlled. These sites may also be physiologically unrelated to the desired movement of the prosthesis (Wurth and Hargrove 2014), making the prosthesis unintuitive to use. Mode-switching (e.g., through co-contraction of the muscle pair) is one way to increase the number of degrees of freedom that can be controlled from the same recording sites (Wurth and Hargrove 2014). A variety of other myoelectric control strategies have been proposed to avoid direct control (Farina et al. 2014), including muscle pattern recognition algorithms in which specific signal features are extracted and used to control different degrees of freedom in the prosthesis (Ajiboye and Weir 2005, Mattioli et al. 2011, Scheme et al. 2014).

Some success has been documented with *targeted muscle reinnervation*, which involves surgical relocation of peripheral nerves to residual muscles (such as the pectoralis major) in order to create additional surface recording sites for myoelectric control (Miller et al. 2008, Kuiken et al. 2009). This ability to record a greater number of independent signals facilitates the use of more fully articulated prostheses than would be possible without surgical intervention. However, because the entire nerve is used to reinnervate a muscle, the number of new recording sites that can be created is limited. Furthermore, some of the original functions of the nerve may not be achievable with the reinnervated muscle (Stubblefield et al. 2009). As with traditional myoelectric control, pattern recognition algorithms may be used in conjunction with targeted muscle reinnervation to control prostheses with multiple degrees of freedom (Kuiken et al. 2009).

Some of the shortcomings of myoelectric control and targeted muscle reinnervation may be addressed by interfacing more directly with the nervous system. One approach involves the use of *peripheral nerve interfaces*, where electrodes are implanted in the residual limb to record neural signals from the peripheral nervous system. These electrodes can be placed around the nerve (Sahin and Durand 1998) or within the nerve (Clark et al. 2011). The other approach uses *cortical interfaces* for which electrodes are placed on (Chestek et al. 2013) or within (Hochberg et al. 2012) the motor cortex to record from the central nervous system. Because these approaches record from the nervous system rather than from the muscle, they may offer a higher degree of specificity (Kung et al. 2013) and can be used to collect a high volume of independent control signals.

Despite the purported advantages of targeted muscle reinnervation, peripheral nerve interfaces, and cortical interfaces, these three approaches have increased medical risk due to their surgically invasive nature. It is important to know whether individuals with upper limb loss feel that the potential advantages of having a more advanced prosthesis would outweigh the potential medical risks associated with the control interface. We recently conducted a survey of 104 individuals with upper limb loss to evaluate the interest of these individuals in noninvasive (myoelectric) and invasive (targeted muscle reinnervation, peripheral nerve interfaces, cortical interfaces) prosthesis control interfaces (Engdahl et al. 2015). Most participants (83%) expressed interest in non-invasive myoelectric control. Although the invasive interfaces were comparatively less popular, many participants (≥ 39%) still expressed interest in these technologies.

Each participant's views on the control interfaces was likely influenced by many factors. Previous literature on factors related to prosthesis acceptance lends support to this idea. The decision to use a prosthesis is thought to be motivated by a combination of predisposing characteristics, enabling resources, and established need (Biddiss and Chau 2007a, Biddiss and Chau 2007b). This includes a wide range of social (e.g., family support), clinical (e.g., time of fitting, training) and individual (e.g., gender, cause of limb loss) factors (Biddiss and Chau 2007c). Given the interrelated nature of these factors, it has been difficult to develop a substantive model to describe these relationships. In fact, a review of 89 articles on factors related to prosthesis use found that there was sufficient evidence to assume a relationship between only a few factors (level of limb loss, age, and lifestyle) and prosthesis acceptance (Biddiss and Chau 2007a).

The decision to use an invasive interface for prosthesis control may be similarly complex, if not more so given the additional considerations regarding medical risk. Some participants in our previous study used the free-form comment section to describe aspects of their decision-making process (provided as supplementary material in (Engdahl et al. 2015)), but meaningful conclusions cannot be drawn from these comments alone. Participants may have been influenced by more factors than they could succinctly describe, or there may have been factors that influenced them without their explicit awareness (such as gender). A more systematic exploration is needed to delineate these potential relationships.

Therefore, the purpose of this work was to explore the factors associated with an individual's interest in novel interfaces for prosthesis control. This information may help guide the development of future prostheses to specifically benefit those individuals who are most likely to accept the technology. Additionally, we investigated whether offering prosthesis functions customized to an individual's interests could increase their willingness to try a surgical procedure for prosthesis control.

6.3 Methods

6.3.1 Ethics statement

All subjects consented to participate in this study, which was granted exempt status and approved by the Institutional Review Board at the University of Michigan Medical School (HUM00077105).

6.3.2 Survey development

This study used an anonymous online survey (described in (Engdahl et al. 2015); full survey available in Appendix E.1) that was administered through Qualtrics (Provo, UT). The survey was initially developed based on the authors' previous experience in surveying individuals with paralysis regarding brain-machine interfaces (Blabe et al. 2015). All authors contributed to subsequent development of the survey. Descriptions of the prosthesis interfaces were written in collaboration with several other local clinicians and researchers. (It is important to note that the descriptions were simply intended to summarize the basic idea behind each interface because exact technical details continue to change as research progresses. As such, the descriptions also included a caveat about the availability of the technology.)

An initial draft of the survey was piloted on seven individuals during their appointments at the University of Michigan Orthotics and Prosthetics Center. Participants completed the survey at their own pace using a tablet computer and were allowed to provide verbal feedback on any question they did not understand while taking the survey. After completing the survey, they discussed their understanding of the questions with a researcher. Most feedback reflected confusion about medical or scientific terminology used in the questions, which prompted us to simplify the language as needed (e.g., changing "trauma" to "injury", "transhumeral" to "above elbow", etc.).

6.3.3 Survey distribution

All individuals over age 18 with upper limb loss above partial hand level were eligible to participate. The survey was advertised through various online forums and mailing lists, paper flyers at the University of Michigan Orthotics and Prosthetics Center, and the Amputee Coalition's *inMotion* magazine. Flyers were also given to clinicians (prosthetists, physical therapists, occupational therapists) for distribution in several institutions across the United States. Finally, the survey was administered via tablet computer to patients at the University of Michigan Orthotics and Prosthetics Center.

6.3.4 Survey design

The first part of the survey included questions about basic demographics, prosthesis usage, and satisfaction with functional abilities. After several early participants failed to answer all questions, the survey was updated to require a response to every presented question. However, some questions only appeared based on prior answers. For example, only participants with acquired limb loss were asked to provide their age at the time of amputation.

In the second part of the survey, participants were asked about their interest in myoelectric control (MYO), targeted muscle reinnervation (TMR), peripheral nerve interfaces (PNI), and cortical interfaces (CI). After reading a brief description of each interface, participants indicated the likelihood that they would try the interface if it offered each of six different functions. The functions were roughly ordered from basic to advanced, and the questions were phrased as: "With the procedures and risks in mind, how likely are you to have the device if it could let you <specific function>?" Responses were collected on a 5-point Likert scale from "very unlikely" to "very likely." The six functions included: 1) moving the hand slowly, 2) rotating the wrist, 3) performing a simple grasp with the arm in any position, 4) performing multiple types of grasp in which the force could be controlled, 5) performing tasks requiring fine motor control, and 6) having touch sensation.

Although myoelectric control does not require surgical intervention, it was included as a point of contrast for the three invasive interfaces. Current myoelectric technology does not offer all of the functions that were presented in the survey, so participants were forced to respond hypothetically regarding those functions. It is possible that participants who expressed interest in trying myoelectric control to achieve more advanced functionality would still be unwilling to try an invasive interface that offered the same features.

After publication of (Engdahl et al. 2015), we added a question asking whether there were any additional activities that participants wanted to perform with a prosthesis. Participants who responded "yes" were asked to list the activities and rate how likely they would be to try each of the four interfaces if they could perform those tasks. The questions were phrased as: "You wrote that you think it is important that your prosthetic allows you to do the following things: <activities listed by participant>. How likely would you be to try this device if it could let you do these things?" Responses were collected on a 5-point Likert scale from "very unlikely" to "very likely."

6.3.5 Data analysis

We selected 16 factors from the survey that may have affected participants' interest in trying the four prosthesis technologies, including 14 categorical variables (Table 6.1) and two continuous variables (age and time since amputation). While additional factors were available in the survey, several were excluded due to a lack of variability in the responses (i.e., ethnicity and race). The remaining factors were excluded because it was unclear how to code the responses in a way that would permit a meaningful statistical analysis. These questions allowed participants to select multiple answers (e.g., reasons for choosing not to use a prosthesis) or to provide free-form answers (e.g., current occupation), which led to considerable variability in the responses.

The outcome measure for each interface was a dichotomous variable indicating whether or not the participant expressed interest (i.e., responded "likely" or "very likely") to any of the six functions. Statistical analyses were performed using SPSS version 22 (IBM Corp., Armonk, NY, USA). Bivariate relationships between each factor and each outcome measure were explored using chi-squared tests (for nominal factors) and Mann-Whitney U tests (for continuous and ordinal factors). The false-discovery rate among the resulting 16 comparisons for each interface was controlled using the Benjamini-Hochberg procedure ($\alpha = 0.09$).

A series of logistic regression models were created to predict each outcome measure from the set of factors. Only factors that were statistically significant in the bivariate analyses were included in the logistic regressions. Time Since Amputation, Prosthesis Type, and Prosthesis Satisfaction were not included because they were only relevant for participants with acquired limb loss or participants who used a prosthesis. Separate models that included these factors were created for the appropriate subset of participants. Participants with missing data on any factor (n = 5) were excluded from the models. Given the lack of prior investigation in this area, we chose to enter all factors into each model simultaneously (forced entry). Interaction effects were not included in this exploratory analysis.

Table 6.1. Sample characteristics for all ordinal and nominal factors.

Factor	N (%)	Total N	Factor	N (%)	Total N
Gender			**Cause of Limb Loss**		
0. Male	139 (60%)	232	0. Acquired limb loss	186 (80%)	232
1. Female	93 (40%)		1. Congenital limb loss	46 (20%)	
Level of Limb Loss [a]			**Pain Frequency** [d]		
1. Partial hand	0 (0%)	232	1. Never	50 (22%)	231
2. Wrist disarticulation	20 (9%)		2. Less than once a month	36 (16%)	
3. Transradial	109 (47%)		3. Once per month	15 (7%)	
4. Elbow disarticulation	14 (6%)		4. 2-3 times per month	31 (13%)	
5. Transhumeral	60 (26%)		5. Once per week	10 (4%)	
6. Shoulder disarticulation	13 (5%)		6. 2-3 times per week	24 (10%)	
7. Forequarter	16 (7%)		7. Daily	65 (28%)	
Prosthesis Use			**Side of Limb Loss** [e]		
0. Yes	158 (68%)	232	0. Nondominant arm	76 (49%)	155
1. No	74 (32%)		1. Dominant arm	79 (51%)	
Myoelectric Use			**Unilateral/Bilateral**		
0. Yes	71 (31%)	232	0. Unilateral limb loss	197 (85%)	232
1. No	161 (69%)		1. Bilateral limb loss	35 (15%)	
Prosthesis Type [b, c]			**Functional Satisfaction** [f]		
1. Passive	21 (14%)	146	1. Very dissatisfied	14 (6%)	232
2. Body-powered	69 (47%)		2. Dissatisfied	39 (17%)	
3. Myoelectric	50 (34%)		3. Neutral	46 (20%)	
4. Adaptive or hybrid	6 (4%)		4. Satisfied	99 (43%)	
			5. Very satisfied	33 (14%)	
Prosthesis Satisfaction [c]			**Prosthesis Necessity**		
1. Very dissatisfied	6 (4%)	146	1. Very unnecessary	32 (14%)	232
2. Dissatisfied	14 (10%)		2. Unnecessary	37 (16%)	
3. Neutral	28 (19%)		3. Unsure	28 (12%)	
4. Satisfied	62 (42%)		4. Necessary	59 (25%)	
5. Very satisfied	36 (25%)		5. Very necessary	76 (33%)	
Education			**Lower Limb Loss**		
1. Some high school or high school degree	28 (12%)	232	0. Yes	35 (15%)	232
2. Some college or college degree	158 (68%)		1. No	197 (85%)	
3. Post-graduate or professional degree	46 (20%)				

List numbers indicate the coding for each factor.
[a] refers to highest level between arms for participants with bilateral limb loss; [b] not determined for 12 participants who used multiple prostheses with equal frequency; [c] refers to most frequently used prosthesis; [d] refers to pain in residual limb; [e] not determined for participants with bilateral or congenital limb loss; [f] refers to overall functional ability, regardless whether prosthesis is used

Table 6.2. Median values of continuous and ordinal factors.
Median values of continuous and ordinal factors for interested and uninterested participants.

Factor	**MYO**			**TMR**			**PNI**			**CI**			**Increased interest occurs with:**
	UI	**INT**	**p [b]**	**UI**	**INT**	**p [b]**	**UI**	**INT**	**p [b]**	**UI**	**INT**	**p [b]**	
Age	43.0	43.0	0.189	48.5	40.0	**< 0.001**	48.5	42.0	**.001**	48.0	37.0	**< 0.001**	Decreased age
Time Since Amputation [a]	14.5	7.5	**0.012**	12.0	7.0	**< 0.001**	14.0	7.0	**< 0.001**	11.0	6.0	**0.008**	Decreased time since amputation
Level of Limb Loss	3	3	0.843	3	3	0.951	3	3	0.678	3	3	0.782	n/a
Pain Frequency	2	4	**0.025**	2	4	**0.001**	2	4.50	**< 0.001**	3	5	**< 0.001**	Increased frequency of pain
Prosthesis Necessity	2	4	**0.005**	4	4	0.202	4	4	0.731	4	4	**0.026**	Increased perceived prosthesis necessity
Prosthesis Satisfaction [a]	5	4	**< 0.001**	4	4	**0.008**	4	4	**0.034**	4	4	**0.037**	Decreased satisfaction
Functional Satisfaction	4	4	**0.005**	4	3	**0.006**	4	3	**0.001**	4	4	0.370	Decreased satisfaction
Education	2	2	0.419	2	2	**0.005**	2	2	0.201	2	2	**0.015**	Decreased educational attainment

UI, uninterested; INT, interested; MYO, myoelectric control; TMR, targeted muscle reinnervation; PNI, peripheral nerve interfaces; CI, cortical interfaces.
[a] not relevant for all participants; [b] p values were calculated using Mann-Whitney U tests

Receiver operating characteristic (ROC) curves were used to describe each model's ability to predict the outcome measure. ROC curves are created by plotting the true positive prediction rate against the false positive prediction rate for the model using a range of threshold parameters. The area under the ROC curve (AUC) represents the probability that the model will rank a randomly chosen positive case (i.e., a participant who expressed interest in the interface) higher than a randomly chosen negative case (i.e., a participant who did not express interest in the interface). An AUC of 0.5 indicates that the model is performing according to chance, while an AUC of 1 indicates that the model is performing perfectly.

6.4 Results

A total of 250 individuals participated in the survey after the publication of (Engdahl et al. 2015). Responses were discarded if the participant stated that they had already taken the survey (n=14), declined participation after reading the consent form (n=4), had only partial hand amputations (n=8), or submitted an incomplete response (n=75). The remaining 149 responses were combined with the 104 responses reported in (Engdahl et al. 2015), and all 253 responses were screened for similarities in demographic information. Twenty-one apparent duplicates were identified and removed, leaving a total sample of 232 responses.

6.4.1 Interest in interfaces

Participants were considered to be interested in an interface if they stated that they would be "likely" or "very likely" to try an interface with any of the six pre-selected functions. Using this criteria, a majority of participants were interested in MYO (86%), TMR (58%), and PNI (64%), while comparatively few were interested in CI (38%). Cochran's Q test indicated significant differences among these four percentages ($p < 0.001$). Post-hoc comparisons with Bonferroni corrections revealed that all pairwise combinations of percentages were significantly different except TMR and PNI ($p = 0.041$).

6.4.2 Participant characteristics

The majority of participants were middle-aged (45 ± 15 years, N = 228), male (60%) and were educated beyond high school level (some college or college degree = 68%, post-graduate or professional degree = 20%). Most participants had unilateral (85%) and acquired (80%) limb loss, which occurred primarily at the transradial and transhumeral levels. The average time since amputation was 13 ± 14 years (N = 184). A majority of participants used a prosthesis at the time of survey completion (68%). Additional descriptive information is given in Table 6.1. Histograms of response frequencies for each factor are also available as supplemental material (Appendix E.2).

6.4.3 Bivariate relationships

A summary of the bivariate relationships is given in Table 6.2 and Table 6.3. Younger ages, lower educational achievement, decreased time since amputation, lower functional satisfaction, greater prosthesis satisfaction, higher frequency of pain, and greater perceived prosthesis necessity were all generally associated with greater interest in the interfaces. Males, participants with unilateral limb loss, participants with acquired limb loss, participants who use a prosthesis, and participants who use a myoelectric prosthesis were also more interested in the interfaces. While not every relationship was statistically significant for each interface, the direction of the significant relationships were consistent across the four interfaces.

6.4.3.1 Effects of quadrilateral limb loss

Exploratory analysis showed that 19 (54%) of the participants with bilateral upper limb loss were actually affected quadrilaterally (i.e., bilateral upper and bilateral lower limb loss). The remaining 16 (46%) participants with bilateral upper limb loss did not have lower limb loss. However, chi-squared tests revealed that interest in the interfaces was not significantly different between these two groups ($p \geq 0.268$).

Table 6.3. Odds ratios for nominal factors.
Odds ratios describing the effect of nominal factors on interest in the interfaces.

		MYO		TMR		PNI		CI	
Factor	**Reference Category**	**Odds Ratio**	**p [b]**	**Odds Ratio**	**p [b]**	**Odds Ratio**	**p [b]**	**Odds Ratio**	**p [b]**
Gender	Female	1.60	0.202	1.88	**0.021**	2.45	**0.001**	2.25	**0.005**
Unilateral/Bilateral	Bilateral	2.38	**0.045**	2.44	**0.015**	3.17	**0.002**	2.31	**0.046**
Cause of Limb Loss	Congenital	2.63	**0.014**	3.84	**< 0.001**	4.31	**< 0.001**	8.64	**< 0.001**
Side of Limb Loss [a]	Nondominant	0.76	0.628	0.93	0.845	0.79	0.530	1.20	0.573
Prosthesis Use	No use	2.85	**0.004**	1.31	0.334	1.04	0.894	1.30	0.373
Prosthesis Type [a]	n/a	n/a	0.098	n/a	0.185	n/a	0.392	n/a	0.532
Myoelectric Use	No use	18.05	**< 0.001**	0.35	0.205	1.07	0.808	1.19	0.544
Lower Limb Loss	No loss	0.64	0.332	0.62	0.190	0.49	0.103	0.71	0.390

MYO, myoelectric control; TMR, targeted muscle reinnervation; PNI, peripheral nerve interfaces; CI, cortical interfaces.
[a] not relevant for all participants; [b] p values were calculated using chi-squared tests

6.4.4 Logistic regressions

Only a few factors proved to be significant predictors for each model (Table 6.4 - Table 6.7). The significant predictors for MYO were Unilateral/Bilateral, Myoelectric Use, and Functional Satisfaction. The significant predictors for TMR were Age, Unilateral/Bilateral, Cause of Limb Loss, and Education. The significant predictors for PNI were Age, Gender, Unilateral/Bilateral, Cause of Limb Loss, and Functional Satisfaction. The significant predictors for CI were Age, Unilateral/Bilateral, Cause of Limb Loss, and Prosthesis Necessity.

Table 6.4. Summary of logistic regression model predicting interest in MYO.

	B [a]	S.E.	p	Odds Ratio	95% CI for Odds Ratio	Reference Category
Unilateral/Bilateral	1.846	0.569	**0.001**	6.335	[2.08, 19.33]	Bilateral
Cause of Limb Loss	0.828	0.578	0.152	2.288	[0.74, 7.10]	Congenital
Pain Frequency	0.085	0.107	0.426	1.089	[0.88, 1.34]	n/a
Prosthesis Necessity	0.339	0.182	0.063	1.403	[0.98, 2.01]	n/a
Prosthesis Use	0.030	0.538	0.955	1.030	[0.36, 2.96]	No Use
Myoelectric Use	3.217	1.087	**0.003**	24.958	[2.96, 210.19]	No Use
Functional Satisfaction	-0.448	0.228	**0.049**	0.639	[0.41, 1.00]	n/a
(Constant)	-0.611	1.343	0.649	-	-	-

Model χ^2 (7) = 48.3, p < 0.001.
[a] unstandardized regression coefficient

Table 6.5. Summary of logistic regression model predicting interest in TMR.

	B [a]	S.E.	p	Odds Ratio	95% CI for Odds Ratio	Reference Category
Age	-0.042	0.011	**0.000**	0.959	[0.94, 0.98]	n/a
Gender	0.558	0.347	0.108	1.747	[0.88, 3.45]	Female
Unilateral/Bilateral	1.106	0.427	**0.010**	3.021	[1.31, 6.98]	Bilateral
Cause of Limb Loss	1.190	0.475	**0.012**	3.287	[1.30, 8.33]	Congenital
Pain Frequency	0.030	0.074	0.690	1.030	[0.89, 1.19]	n/a
Functional Satisfaction	-0.220	0.144	0.127	0.802	[0.60, 1.06]	n/a
Education	-0.627	0.288	**0.029**	0.534	[0.30, 0.94]	n/a
(Constant)	1.977	1.067	0.064	-	-	-

Model χ^2 (7) = 49.9, p < 0.001.
[a] unstandardized regression coefficient

Table 6.6. Summary of logistic regression model predicting interest in PNI.

	B [a]	S.E.	p	Odds Ratio	95% CI for Odds Ratio	Reference Category
Age	-0.045	0.012	**0.000**	0.956	[0.93, 0.98]	n/a
Gender	0.973	0.364	**0.008**	2.646	[1.30, 5.40]	Female
Unilateral/Bilateral	1.366	0.439	**0.002**	3.920	[1.66, 9.28]	Bilateral
Cause of Limb Loss	1.328	0.484	**0.006**	3.773	[1.46, 9.73]	Congenital
Pain Frequency	0.059	0.078	0.454	1.060	[0.91, 1.24]	n/a
Functional Satisfaction	-0.412	0.157	**0.008**	0.662	[0.49, 0.90]	n/a
(Constant)	1.008	0.936	0.281	-	-	-

Model χ^2 (6) = 62.7, $p < 0.001$.
[a] unstandardized regression coefficient

Table 6.7. Summary of logistic regression model predicting interest in CI.

	B [a]	S.E.	p	Odds Ratio	95% CI for Odds Ratio	Reference Category
Age	-0.057	0.012	**0.000**	0.945	[0.92, 0.97]	n/a
Gender	0.644	0.361	0.074	1.905	[0.94, 3.86]	Female
Unilateral/Bilateral	1.029	0.487	**0.035**	2.799	[1.08, 7.27]	Bilateral
Cause of Limb Loss	2.235	0.636	**0.000**	9.346	[2.69, 32.49]	Congenital
Pain Frequency	0.009	0.077	0.909	1.009	[0.87, 1.17]	n/a
Prosthesis Necessity	0.274	0.119	**0.021**	1.316	[1.04, 1.66]	n/a
Education	-0.554	0.299	0.064	0.575	[0.32, 1.03]	n/a
(Constant)	-1.103	1.161	0.342	-	-	-

Model χ^2 (7) = 68.2, $p < 0.001$.
[a] unstandardized regression coefficient

All four models had good discriminatory power, as indicated by the ROC curves (Figure 6.1A). The AUC was similar for each model and was significantly greater than 0.5 in all cases ($p < 0.001$; MYO = 0.838, TMR = 0.770, PNI = 0.805, CI = 0.809) (Figure 6.1B).

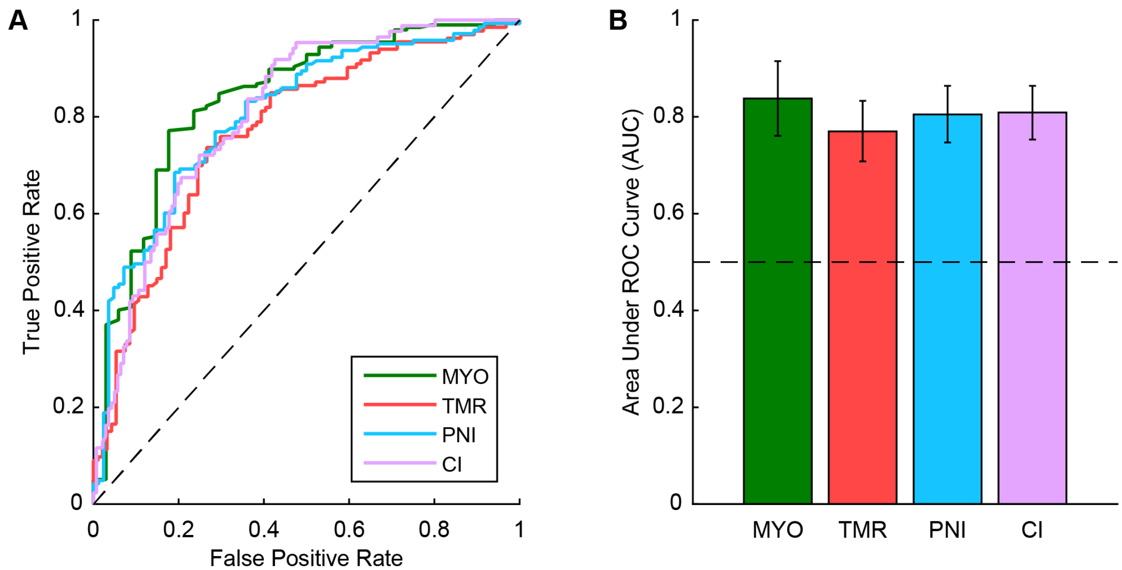


Figure 6.1. Discriminative power of the logistic regression models.
(A) ROC curves for each regression model. The diagonal reference line indicates performance according to chance. (B) Area under the ROC curve for each regression model. The horizontal reference line indicates performance according to chance. Error bars represent 95% confidence intervals. (MYO = myoelectric control, TMR = targeted muscle reinnervation, PNI = peripheral nerve interfaces, CI = cortical interfaces)

Separate logistic regressions for participants with acquired limb loss only and for prosthesis users only are presented in the supplementary material (Appendix E.3). While the combination of significant factors varied in comparison to the models presented here, the area under the ROC curve was significantly greater ($p < 0.001$) than 0.5 in all cases. For the models involving participants with acquired limb loss, Time Since Amputation was a significant predictor only for MYO, TMR, and PNI. For the models involving prosthesis users, Prosthesis Satisfaction was a significant predictor only for MYO.

6.4.5 Response to self-selected functions

Only 129 participants were asked whether there were additional activities that they wanted to perform with a prosthesis, and only 61 (47%) responded affirmatively. While there was considerable diversity in the functions that were mentioned, most functions could be classified into one of 12 different categories (Figure 6.2; see Appendix E.4 for a complete list). The most common functions related to sports and other recreational activities, followed by improved dexterity and grasping ability. Regardless of the interface, most participants (≥ 67%) were equally interested in the self-selected functions and pre-selected functions (representative example shown for PNI in Figure 6.3, see solid bubbles). Of these participants, 95% expressed interest in MYO, 47% in TMR, 63% in PNI, and 27% in CI. Few participants actually changed from uninterested to interested when the self-selected functions were added (green region in Figure 6.3), and some participants even changed from interested to uninterested (red region in Figure 6.3). However, paired t-tests revealed no significant differences in the most interested response among the pre-selected functions and responses to the self-selected functions ($p \geq 0.091$).

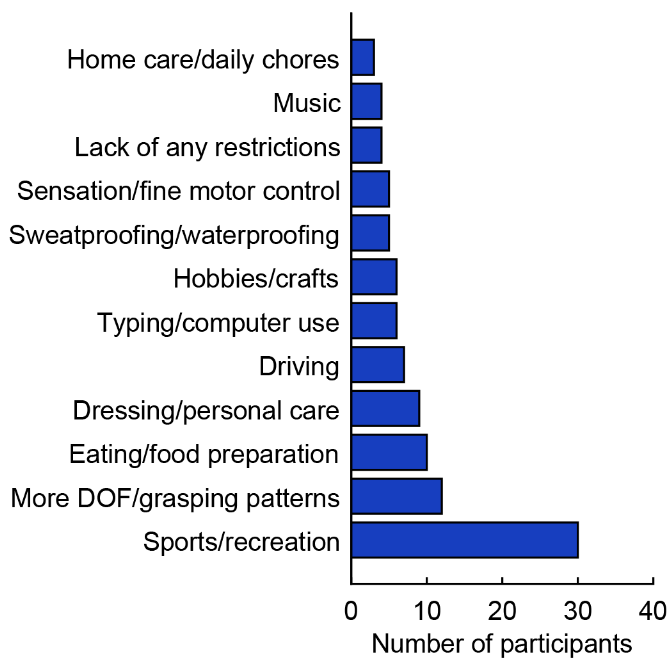


Figure 6.2. Additional categories of functions that were not already included in the survey.

Participants listed additional functions that they wanted to perform with a prosthesis that were not already included in the survey. In cases where a participant mentioned multiple functions that could be classified into a single category, the participant was counted only once for that category.

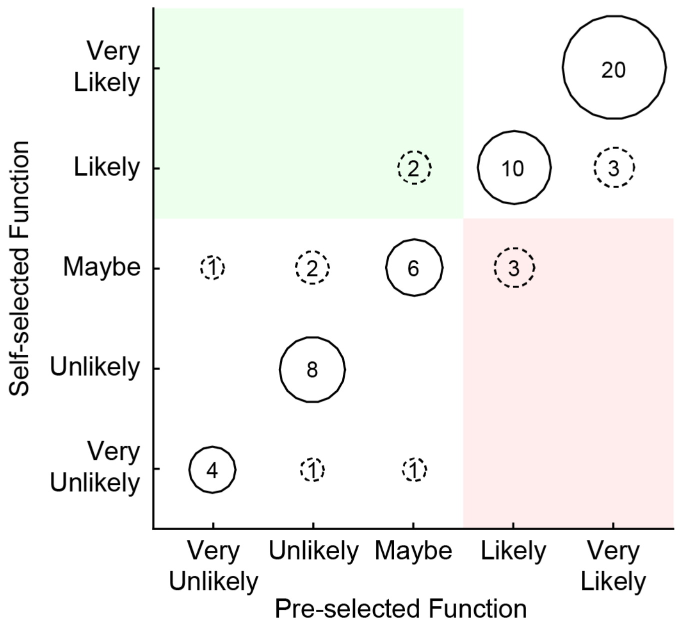


Figure 6.3. Distribution of interest in peripheral nerve interfaces depending on the availability of self-selected functions.
Participants indicated their interest in trying peripheral nerve interfaces if they could perform additional functions with a prosthesis that were not already included in the survey (vertical axis). These responses are presented in relation to the most interested response from the other six functions included in the survey (horizontal axis). The bubbles show the number of participants who gave each combination of responses. Dashed lines indicate a change in response. Green shading designates a change from an uninterested response ("very unlikely", "unlikely" or "maybe") to an interested response ("likely" or "very likely"). Red shading designates a change from an interested to uninterested response.

6.5 Discussion

The primary purpose of this work was to determine the factors associated with an individual's interest in novel interfaces for prosthesis control. Exploratory analyses revealed several common trends, although the degree of statistical significance varied between interfaces. In general, there was greater interest among males, participants with unilateral limb loss, participants with acquired limb loss, participants who use a prosthesis, and participants who use a myoelectric prosthesis. Greater interest was also associated with younger ages, lower educational achievement, decreased time since amputation, lower satisfaction with overall functional ability, greater satisfaction with a prosthesis (if a prosthesis was used), higher frequency of pain, and greater perceived prosthesis necessity.

When these factors were used to create regression models predicting interest in each interface, many were no longer significant. In fact, each of the four regression models identified a different subset of factors that were significant predictors. The only factor that was significant for all four interfaces was Unilateral/Bilateral, where individuals with bilateral limb loss were less interested in the interfaces than those with unilateral limb loss. It might be expected that individuals with bilateral limb loss would be comparatively *more* interested given the greater severity of their impairments. However, our findings suggest these individuals may believe the associated risks (especially for the invasive interfaces) outweigh any potential benefits. Without an intact limb to rely on during recovery from surgery, training, or in case of equipment malfunction, individuals with bilateral limb loss might be particularly concerned about any loss of function. This may also explain why quadrilateral limb loss was not associated with greater interest among participants with bilateral upper limb loss.

Age and cause of limb loss were also significant predictors for all three regression models involving surgically invasive interfaces. When given the opportunity to write free-form comments about the interfaces, several participants mentioned that they would have more seriously considered the invasive interfaces if they were younger (Appendix E.5). These participants generally did not feel that accepting the risks associated with these interfaces would be justified at their age. The fact that age and cause of limb loss were significant may also suggest that some individuals would have been more interested in the interfaces if the technology could be implemented at the time of amputation. Some participants expressed concern about needing an additional surgery to use the interfaces, as they had already been through numerous surgeries related to their initial limb loss (Appendix E.5). Although the survey did not specifically ask participants to consider when they would receive the interfaces in relation to their amputation, this may have been a confounding factor.

A separate analysis explored the effect of time since amputation among participants with acquired limb loss. Generally, participants who had experienced an amputation more recently expressed greater interest in the interfaces. As individuals become more accustomed to their condition over time, they may become less interested in alternative solutions beyond what is already clinically available. Interestingly, time since amputation was a significant predictor for myoelectric control, targeted muscle reinnervation and peripheral nerve interfaces, but not cortical interfaces. This trend may suggest that opinions on cortical interfaces are particularly static over time, regardless of whether individuals have become accustomed to their amputation.

Additionally, we explored whether offering prosthesis functions customized to each participant influenced their interest in each interface. The six functions that we chose to include in the survey may not necessarily encompass everything that is considered important by individuals with upper limb loss. We hypothesized that allowing participants to identify unique functions that they valued would prompt more positive responses to the interfaces. Our findings did not support this hypothesis, as a majority of participants did not change their responses when considering their self-selected functions in comparison to the six pre-selected functions. This trend could suggest that participants believed the six pre-selected functions were already comprehensive enough to facilitate their chosen activities, or that the surgical and/or training information was more influential to their decision than the functionality. Nonetheless, it is interesting to note the diversity in the types of functions that participants chose (Appendix E.4).

It is important to acknowledge several limitations that may have influenced these results. First, participants' responses to the myoelectric technology may have been biased in comparison to the other interfaces. Because myoelectric technology has been commercially available for decades, most participants were likely familiar with it already. This familiarity may have introduced additional variables into the decision-making process that were not relevant for the other, less familiar interfaces. Indeed, use of a myoelectric prosthesis was a highly significant predictor only for myoelectric control.

There were also limitations in the survey design, specifically in how the interfaces were described. We were constrained by the fact that some of the technologies do not exist outside of research labs, or do not currently exist in a form that offers all of the functions presented in this survey. It may be years before these interfaces are ready for widespread use, and the exact technical specifications are likely to change as development progresses. Consequently, we could not precisely define the training times, medical procedures, or medical risks. We also wanted the descriptions to be easily understood by individuals without a medical or scientific background, and omitted some details in order to maintain clarity. For these reasons, the descriptions were somewhat ambiguous and ultimately may have allowed participants to inaccurately infer potential benefits and risks. It would be informative to conduct another survey in the future when more technical details have been finalized so the descriptions reflect technologies that are truly available for clinical use. The results would likely differ from what we obtained here using more hypothetical descriptions.

The written comments suggest that some participants made their own inferences about the interfaces. For example, we instructed participants to assume that all the interfaces were waterproof to encourage them to evaluate each one on a broader, hypothetical level. However, waterproofing was mentioned numerous times as a desirable feature (Figure 6.2). Cost was also raised as a point of concern by several participants (Appendix E.5) even though they were instructed not to focus on cost when considering the interfaces (Appendix E.1). These findings suggest that some participants may have ignored or forgotten the instructions, or that their responses were affected by previous experiences with prosthetic technology. Cost may have been especially difficult to ignore, as many individuals with upper limb loss have difficulty obtaining adequate insurance coverage and experience a significant financial burden when acquiring, maintaining, and/or repairing their prosthesis (Resnik et al. 2012). Likewise, responses may have been influenced the participants' prior experiences with prosthesis sockets. Socket fit is extremely important in promoting functionality and comfort (Lake 2008), which many participants would have known from past prosthesis use. Although sockets were not emphasized in the survey, participants may have responded more negatively to the interfaces if they perceived that sockets were necessary and had negative opinions about traditional sockets. Concerns about sockets were in fact mentioned by several participants (Appendix E.5).

Similarly, the way that the functions were described may have introduced some variability in participants' responses. Although the first 35 participants viewed a slightly different wording version (as discussed in (Engdahl et al. 2015)), the vast majority of participants viewed the six functions in a cumulative manner where each successive function included the previous functions. We intentionally chose this wording in order to determine whether there was a "tipping point" where participants felt that the functionality of the prosthesis would outweigh any risks. It is possible that this wording prioritized functionality in a way that participants did not necessarily agree with. Using a dichotomous outcome measure that indicated whether participants were interested in at least one of the six pre-selected functions may have reduced the impact of this wording on our analysis. Additionally, most participants did not change their response to the interfaces when considering functions customized to their interest. This suggests the six pre-selected functions already included many of the features that prosthesis users value. Nonetheless, we acknowledge that presenting the functions cumulatively may have prevented some participants from choosing the specific functions they cared about.

Finally, the sample population may not accurately represent the larger population of individuals with upper limb loss because recruitment was primarily conducted online. We expected that online recruitment would be adequate because 84% of U.S. households owned computers in 2013 and 74% used Internet in the home (File and Ryan 2013). In an effort to include individuals who may not have computer or Internet access at home, we also made the survey available on tablet computer to patients visiting the University of Michigan Orthotics and Prosthetics Center. Our sample population matches the populations reported by other large-scale surveys in terms of several important demographic factors, including age, gender, prevalence of transradial limb loss, and prevalence of limb loss due to trauma (Atkins et al. 1996, Biddiss and Chau 2007b). (Note that the study by Atkins et al. (Atkins et al. 1996) was conducted entirely via mail). However, other characteristics of our sample population differ from what has been previously reported. Notably, the educational attainment of our participants exceeds what has been reported by Raichle et al. (Raichle et al. 2008), as well as national averages reported by the U.S. Census Bureau (Ryan and Bauman 2016). Census records from 2015 indicate that most adults (88%) had at least a high school degree, while 33% had at least a bachelor's degree (Ryan and Bauman 2016). In contrast, 99% of our participants had at least a high school degree and 53% had at least a bachelor's degree. This may indicate a sampling bias, as computer ownership and Internet use tends to be lower in households with lower educational attainment (File and Ryan 2013). Furthermore, computer ownership and home Internet use tends to be less common in Hispanic households compared to white, non-Hispanic households (File and Ryan 2013). Since our sample population was predominately white and non-Hispanic, this may be further evidence of a sampling bias. The results of this work should be generalized carefully given these limitations.

6.5.1 Conclusion

Our work has demonstrated that several factors are consistently associated with interest in novel interfaces for upper limb prosthesis control. Younger age, acquired limb loss, and unilateral limb loss were related to greater interest in surgically invasive interfaces. Interest in noninvasive myoelectric control was also associated with unilateral limb loss, as well as current use of a myoelectric prosthesis. Knowledge of these associations may be helpful to research efforts. For examples, researchers could try to involve individuals with these characteristics in testing and assessment of future devices. The information regarding specific benefits, medical risks, and training procedures that is gained as a result of this testing may eventually encourage those who are currently not interested to consider these interfaces.

Collectively, the work reported here and in our earlier paper (Engdahl et al. 2015) advances the literature in several important ways. Although it has been reported that individuals with upper limb loss are interested in novel interfaces for prosthesis control (e.g., (Biddiss et al. 2007)), our earlier paper was the first to actually quantify this interest. Our current analysis expands on those findings by identifying factors associated with the participants' interest. The propensity of prosthesis developers to pursue new technologies before the end users' needs have been clearly articulated is a detriment to individuals with limb loss (Biddiss et al. 2007, Carey et al. 2015), who may reject technologies which fail to meet their demands. Our work is valuable in this context, as it helps elucidate the perspectives of individuals with upper limb loss. However, we also recommend that additional studies are done to explore patient opinions in greater detail. It is clear from our work that a single survey is insufficient to understand every factor that motivates an individual's interest in new prosthesis technologies. A variety of other factors relating to the individual's medical history, lifestyle, and psychosocial state should be considered.

CHAPTER 7. Discussion

7.1 Summary and suggestions for future work

Existing literature offers limited quantitative insight on how upper limb prosthesis type affects functional outcomes and satisfaction. Without this information, it is difficult to know how future prosthesis designs should be improved. It is also difficult to demonstrate whether those designs offer advantages compared to existing technologies. This dissertation addresses several of these shortcomings by quantifying how embodiment, movement quality, and kinematic compensations are impacted through use of BP and MYO prostheses. It also explores what factors are associated with interest in emerging technologies for upper limb prosthesis control.

One major purpose of a prosthesis is to replace the missing limb to the fullest extent possible. Thus, embodiment of the prosthesis may be paramount to patient's success with the device. In order to delineate the relationship between embodiment and functional success, we first must understand how to characterize embodiment and how it develops. Based on the idea that prosthesis design may affect embodiment due to differences in the availability of sensory feedback, we sought to determine whether the experience of embodiment differs between BP and MYO users. Although BP users did report a stronger sense of agency over their prostheses in comparison to MYO users, our objective measurements of body schema and peripersonal space did not reveal clear differences between the two groups.

Nonetheless, it could be informative to continue exploring this question using additional methodology. Embodiment is not an "all or nothing" phenomenon. A prosthesis is likely embodied only in certain capacities (de Vignemont 2011), which may not be detectable by all methodologies.

For this reason, embodiment should be discussed in terms of specific quantifiable features. For example, overestimation of residual limb length while wearing a prosthesis can be interpreted as spatial embodiment (McDonnell et al. 1989), which is distinct from motoric or affective embodiment (de Vignemont 2011) and can be measured only with particular methodology. Even with the appropriate methodology, it will be difficult to establish clear benchmarks that indicate whether a prosthesis is embodied. For example, overestimation error may be simple to measure but it is not clear whether all error magnitudes are equally indicative of embodiment. There are many questions of this nature that must first be resolved if the broader functional implications of prosthesis embodiment are to be understood. *Future work should focus on developing a more cohesive definition of what it means to embody an upper limb prosthesis and what methodology can be used to reliably indicate if these conditions are met.*

If a prosthesis is to replace the missing limb to the fullest extent possible, it should also help the user achieve natural motor patterns. Specifically, movements made with the prosthesis should be smooth, coordinated, and accurate while evoking minimal compensatory motion (Smurr et al. 2008). Outcome measures based on movement kinematics are most appropriate for assessing these characteristics, but assessments often rely on self-reported responses, visual assessment of task performance, or task completion time instead (de los Reyes-Guzmán et al. 2014). While such assessments do have some basic utility for understanding how a prosthesis is used, they are insufficient to comprehensively assess the motor strategies involved with task performance. This is a major reason why there is currently a poor understanding of the relative functional advantages between BP and MYO prostheses (Carey et al. 2015, Carey et al. 2017).

To help address this issue, we quantified movement quality metrics and kinematic compensations in BP and MYO prosthesis users. MYO users consistently performed both reaching and object manipulation tasks more slowly than BP users, and were also less smooth when performing object manipulation tasks. However, BP users required a larger trunk range of motion to accomplish the same tasks and had greater movement curvature when reaching to grasp an object. This suggests a trade-off between the two devices in terms of functional advantages.

Collectively, these findings are likely related to differences in the control methods and the availability of sensory feedback between BP and MYO prostheses. Activating a BP prosthesis inherently requires excursion of the shoulder and/or trunk, whereas activating a MYO prosthesis requires only contracting muscles in the residual limb. This additional movement involved with BP use could contribute to the greater trunk compensations and movement curvature. However, BP users may be more equipped to make small corrections during movement since BP prostheses are thought to provide proprioceptive feedback through the mechanical control system. BP users might therefore have improved movement quality in comparison to MYO users, who rely primarily on visual feedback.

Future work should focus on identifying implications of reduced movement quality and compensatory strategies in terms of short-term burdens imposed on the prosthesis user and long-term prosthesis acceptance. As one example, several studies have speculated that poor movement quality and compensatory strategies might contribute to increased metabolic expenditure during prosthesis use (Metzger et al. 2012, Major et al. 2014, Cowley et al. 2017). It would be beneficial to demonstrate whether this relationship exists and whether it is affected by the type of prosthesis that is used, which may offer insight on factors contributing to acceptance or rejection of a prosthesis.

It is important to note that our findings were also influenced by task requirements. For instance, reaching movements were more curved with BP prostheses compared to MYO prostheses when reaching to grasp an object, but were less curved when reaching without the intent to grasp. *These contradictory findings highlight the need to assess prosthesis use during a wide range of tasks.* There is enormous diversity in the types of activities that a prosthesis user might choose to perform with their prosthesis, so it is problematic to generalize findings from a small subset of tasks. It should also be noted that the tightly-controlled laboratory environments often favored by researchers may not match the real-world settings in which prosthesis users actually perform tasks. *Future work that is focused on testing prosthesis users in everyday life is needed to further elucidate the short-term and long-term burdens imposed by prosthesis use.* That said, it is also crucial to understand how healthy individuals perform the same tasks in unconstrained settings and whether we can record reliable measurements in those settings. Our work demonstrates that unconstrained ADLs can indeed reliably be used to assess movement quality in functional settings that mimic real-world challenges.

Our work is not intended to make specific recommendations on whether there are absolute advantages to BP or MYO prostheses. In fact, it is unlikely that such a recommendation can be made. Despite the dearth of high-quality empirical evidence supporting these notions, the current consensus in the literature is that BP prostheses have advantages in areas such as durability, training time, and feedback, while MYO devices offer improved cosmesis (Carey et al. 2015, Carey et al. 2017). If each prosthesis type truly has relative advantages, prosthesis choice must be based on a patient's individual needs and preferences. By further clarifying some of these relative advantages, our work may eventually help inform the selection process. Furthermore, our work can also contribute towards developing a body of proof that can be presented to insurers, who require justification that prostheses actually help the user regain a "normal" level of function

(Resnik et al. 2012). *Collecting evidence that can be used to validate the utility of prosthesis provision should be a priority for future work.*

Nonetheless, we must acknowledge that prosthesis technology is rapidly expanding beyond the traditional BP and MYO designs. Patients may soon be faced with a greater array of options, forcing them to balance numerous considerations when selecting a prosthesis. It is important to understand how all of these technologies relate to each other in terms of functional performance and satisfaction. We have demonstrated that a sizeable portion of the population is not willing to consider surgically invasive prosthesis technologies (Engdahl et al. 2015) and that interest in these technologies tends to be higher among individuals who share certain specific characteristics (younger age, unilateral limb loss, and acquired limb loss) (Engdahl et al. 2017). This suggests a continued need to improve noninvasive technologies. *Understanding the advantages and limitations of existing technologies, based on suggestions for future work identified in this chapter, is an important step in this process.*

7.2 Long-term vision

I believe that continued development of upper limb prosthesis control techniques will have important implications relating to embodiment and functional outcomes for prosthesis users. Technologies that take advantage of existing communication pathways within the body by interfacing directly with the nervous system (such as targeted muscle reinnervation, peripheral nerve interfaces, and cortical interfaces) and/or decoding intent from noninvasively recorded control signals (such as pattern recognition with myoelectric control) show promise for promoting intuitiveness of prosthesis control. As a result, the prosthesis user may need to devote less attention to operation of their prosthesis, enhancing the naturalness of the experience. These same control techniques also offer expanded possibilities for delivering sensory feedback to the user. Sensory feedback can be delivered either via sensory substitution techniques (e.g., mechanically vibrating the skin of the residual limb in response to a stimulus) or modality matching techniques (e.g., eliciting proprioception and tactile sensations through direct neural stimulation with an invasive electrode). Having access to a diverse array of sensations that are easily interpretable may further improve the naturalness of operating a prosthesis. Taken together, I think these improvements in control and feedback will strengthen the user's experience of prosthesis embodiment.

I anticipate that embodiment of a prosthesis would have important functional consequences over multiple phases of the rehabilitation process. In the long term, I expect that experiencing a prosthesis like part of one's own body could facilitate achievement of more normative movement patterns. Specifically, embodying a prosthesis could help the user minimize energy expenditure, perform tasks without extraneous body movement, and avoid placing undue stress on the intact limb. I believe these advantages will ultimately be reflected as long-term acceptance of the prosthesis. Embodying a prosthesis might also improve training experiences early in the rehabilitative process. If the first prosthesis that a patient tries can be controlled in a manner similar to a natural limb and offers interpretable sensory feedback, it could reduce the amount of time needed to train with the prosthesis and acclimate to loss of the limb. In turn, I expect that this may reduce frustration with the prosthesis and make patients more likely to incorporate it into daily life.

On a related note, I think that embodiment, functional success, and ultimate acceptance of a prosthesis are dependent on other factors beyond the device design. It is likely that other moderating factors affect each individual's experience with a prosthesis. Even if two different patients are provided with the same type of prosthesis and undergo the same training procedures, I suspect that human factors like openness to new experiences, ability to problem solve, perception of the prosthesis itself, or frustration tolerance could interact to create different experiences with prosthesis use. Promoting success with a prosthesis may require considering factors beyond the device itself to understand individual patient perspectives.

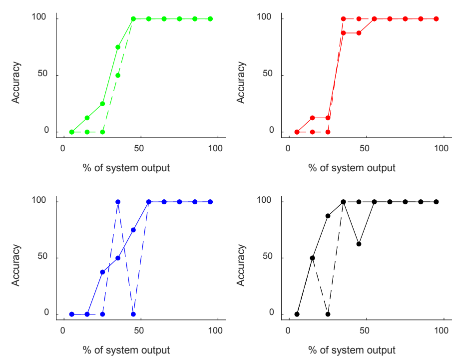
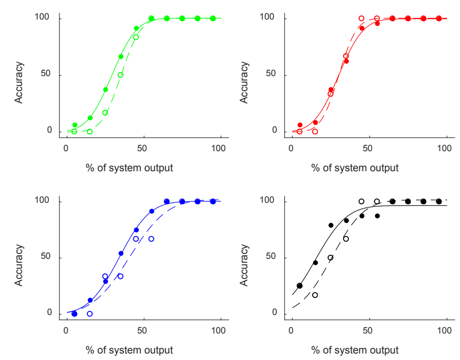
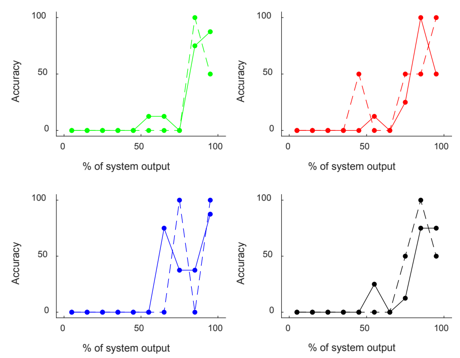
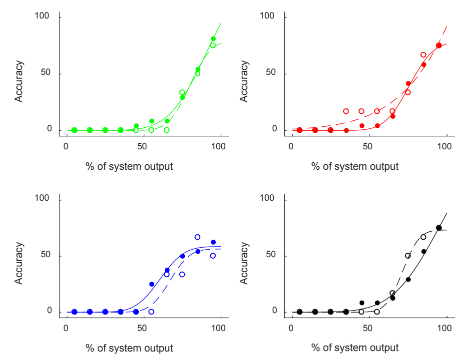
Appendices

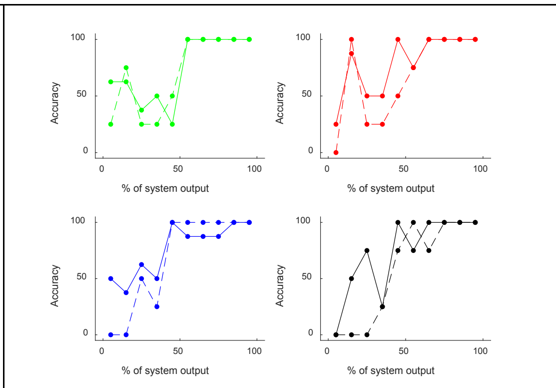
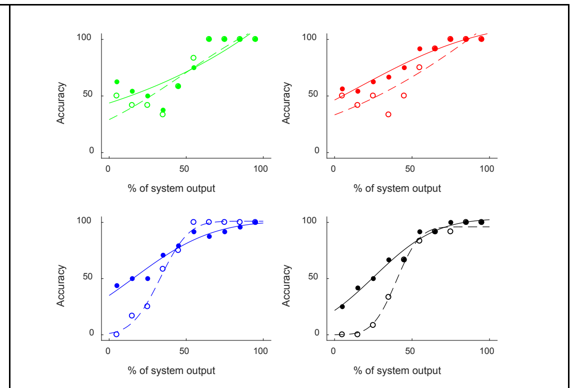
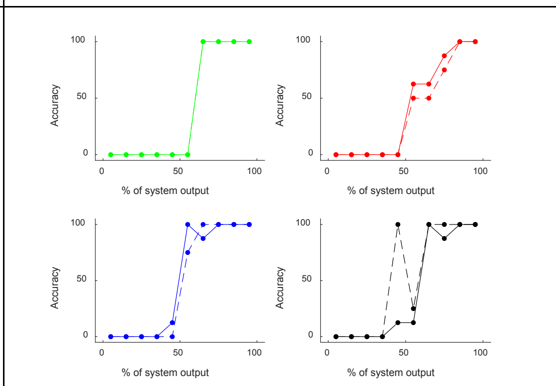
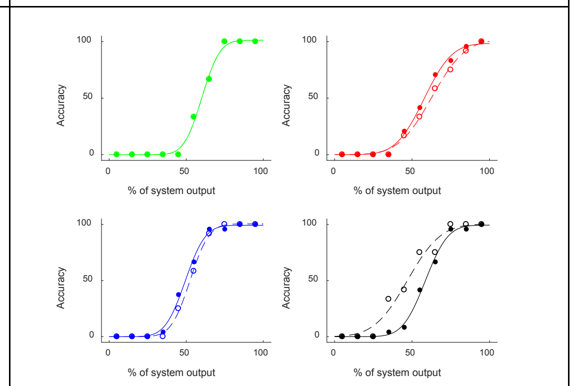
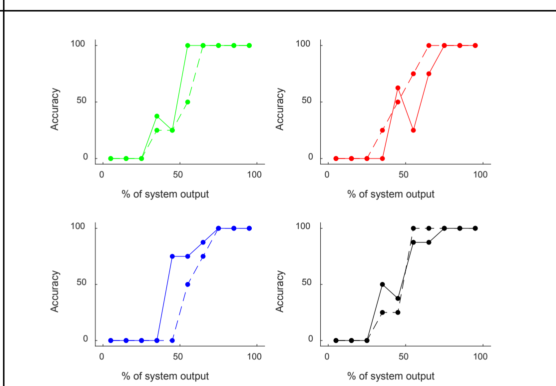
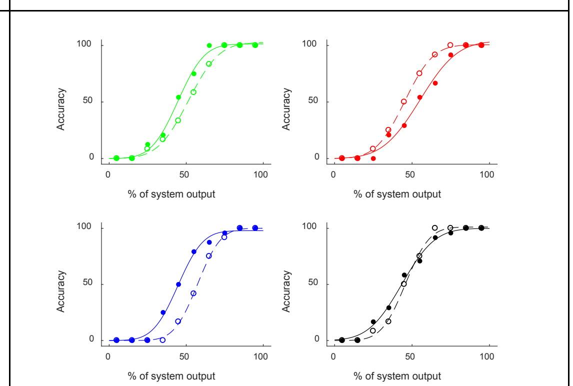
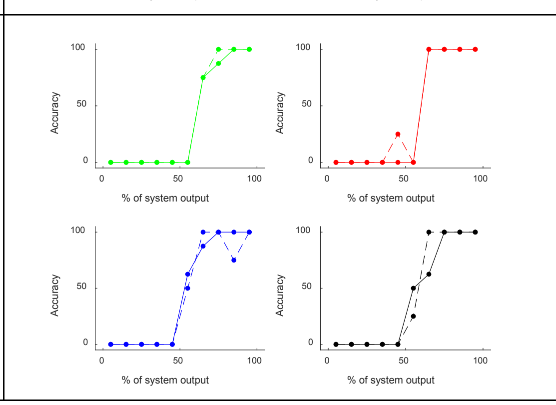
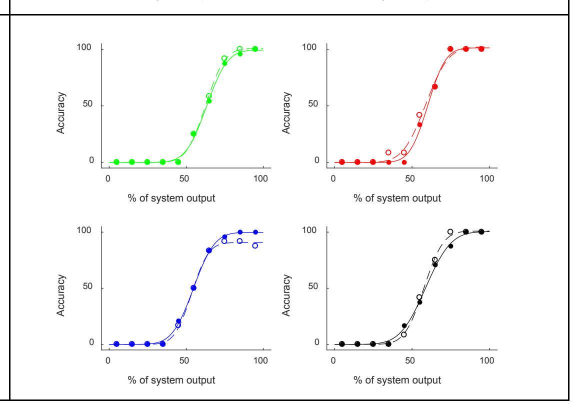
Appendix A. Supplementary material for Chapter 2

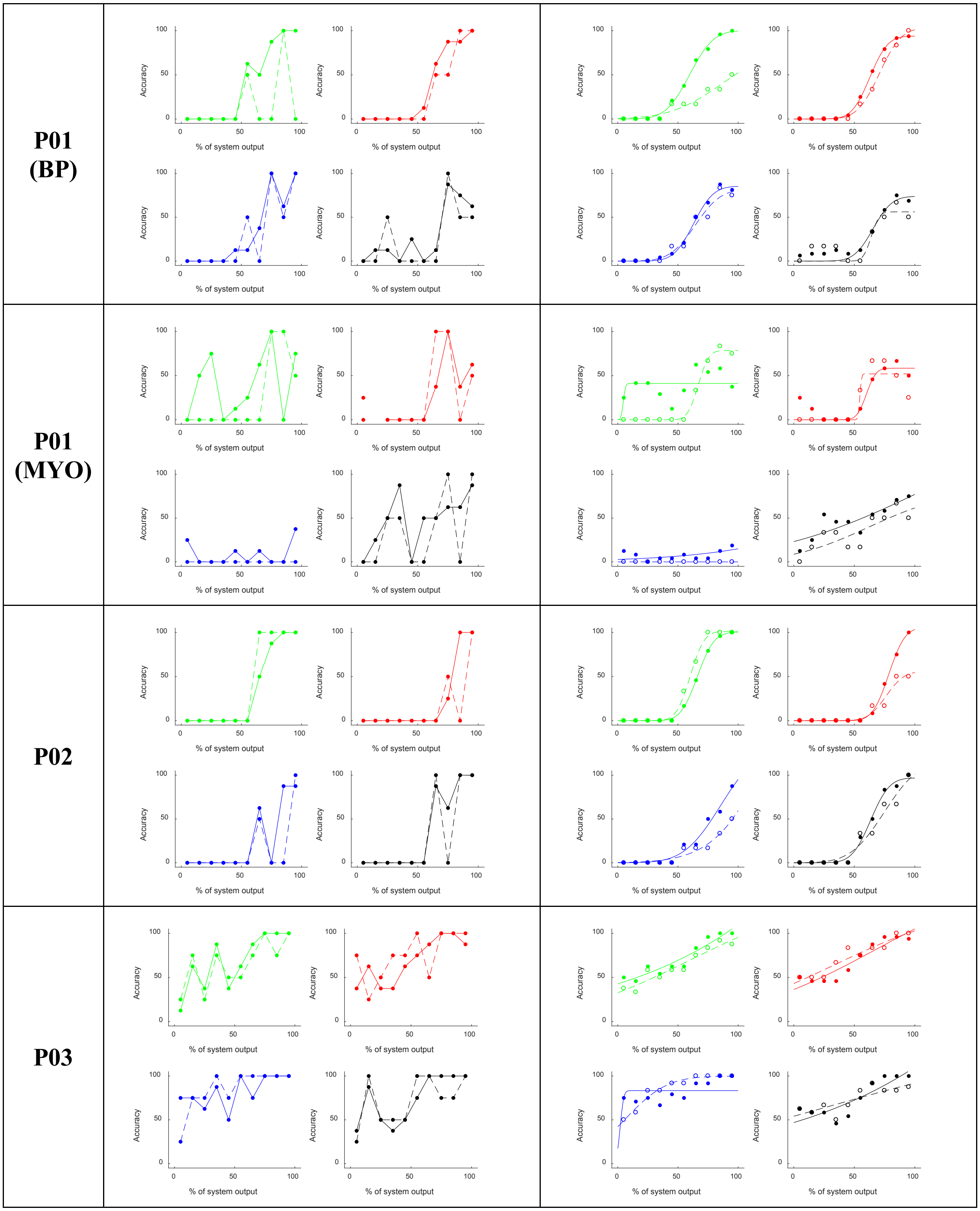
A.1 All results from the extinction paradigm.

Table A.1. Tactor detection accuracy.

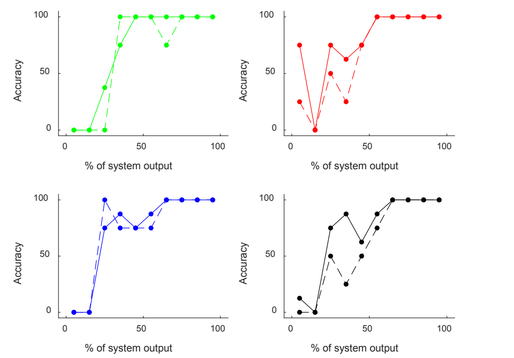
Raw data (left) and data fitted with a cumulative normal function after calculation of a moving average (right). Solid lines correspond to visual/tactile trials and dashed lines correspond to tactile trials. (green = Far LED with Far Fixation; red = Mid LED with Far Fixation; blue = Mid LED with Near Fixation; black = Near LED with Near Fixation).

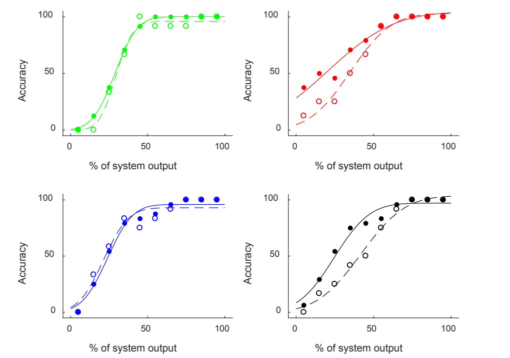
ID	Raw Data	Fitted Data
C01		
CO2		

C03		
C04		
C05		
C06		

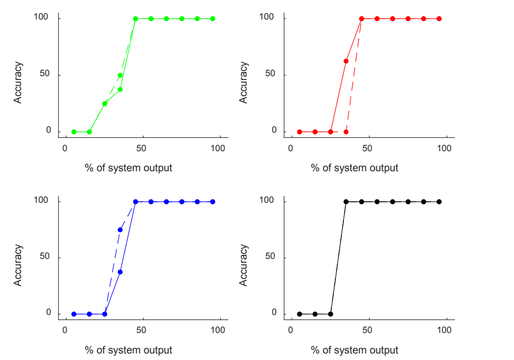

P01 (BP)
P01 (MYO)
P02
P03
Accuracy
% of system output
0
50
100

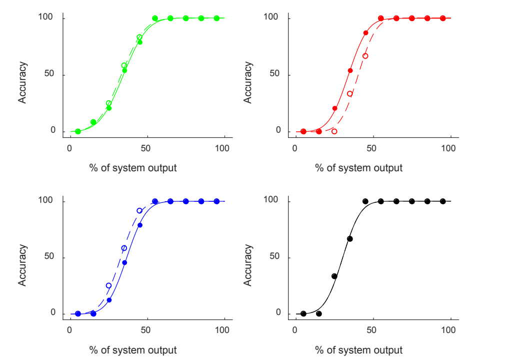
P04


Accuracy
% of system output

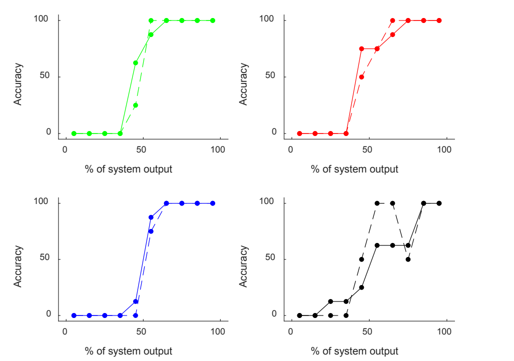

Accuracy
% of system output

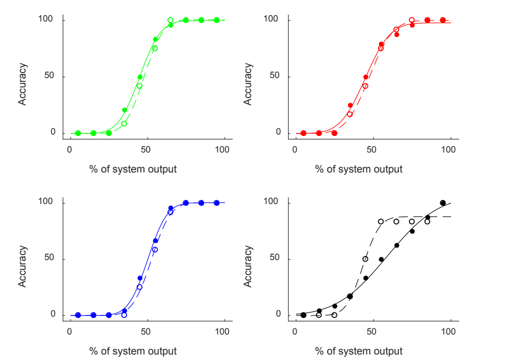
P05


Accuracy
% of system output


Accuracy
% of system output

P06

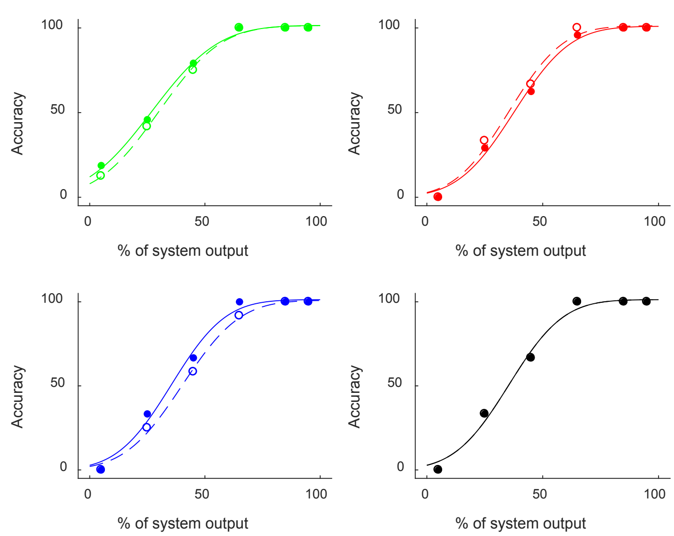

Accuracy
% of system output


Accuracy
% of system output

A.2 Pilot study with healthy controls.

In order to assess whether extinction could be induced by making the visual stimuli more distracting, we made several incremental changes to the protocol. First, we increased the LED intensity from being at a minimally detectable level up to the maximum intensity allowed by the system. The rest of the protocol remained the same. No consistent trends were seen for a single healthy control (Table A.2).

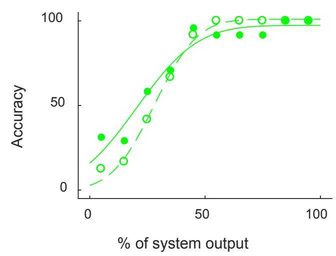
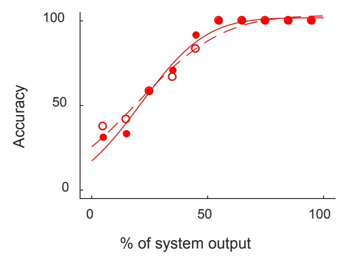
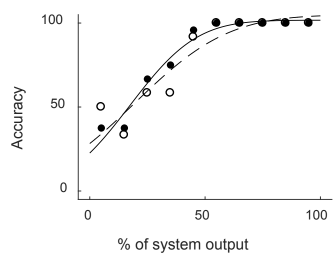
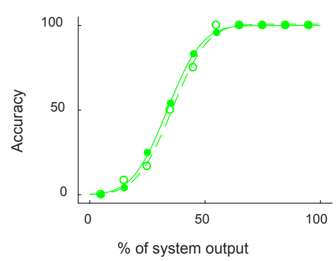
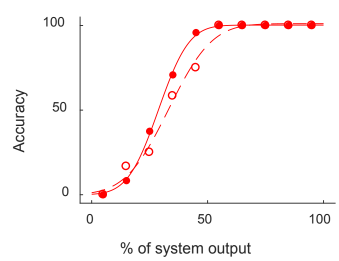
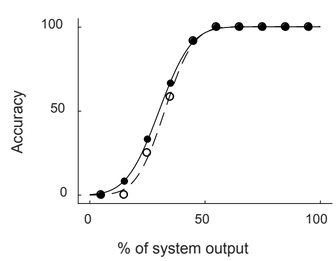
Table A.2. Tactor detection accuracy with maximally intense LEDs.
Solid lines correspond to visual/tactile trials and dashed lines correspond to tactile trials. (green = Far LED with Far Fixation; red = Mid LED with Far Fixation; blue = Mid LED with Near Fixation; black = Near LED with Near Fixation).

ID	Fitted Data
Pilot_01	

Next, the protocol was altered so that only three LED/fixation point combinations were included. Only the far fixation point was used, so the three conditions were Far LED with Far Fixation, Mid LED with Far Fixation, and Near LED with Far Fixation. The order of testing was also adjusted so that a single condition involved all thee LEDs for a given tactor intensity. Thus, only 10 conditions were included. Each condition had 48 trials (24 visual/tactile, 12 visual only, 12 tactile only), where the trials involving a visual stimulus were equally divided across the three

LEDs. The LED intensities were set to maximum intensity. Participants responded to the tactor by pressing the foot pedal, but also verbally indicated when they saw a visual stimulus. Again, no consistent effects were seen in two healthy controls (Table A.3).

Table A.3. Tactor detection accuracy for altered extinction paradigm.
Solid lines correspond to visual/tactile trials and dashed lines correspond to tactile trials. (green = Far LED with Far Fixation; red = Mid LED with Far Fixation; black = Near LED with Far Fixation).

ID	Fitted Data
Pilot_02	  
Pilot_03	  

Finally, we shortened the tactile stimulus to only 50 ms. All other aspects of the protocol were identical to what was described previously. Again, no consistent trends were seen across two healthy controls (Table A.4).

Table A.4. Tactor detection accuracy for shorted tactile stimuli.
Solid lines correspond to visual/tactile trials and dashed lines correspond to tactile trials. (green = Far LED with Far Fixation; red = Mid LED with Far Fixation; black = Near LED with Far Fixation).

<table>
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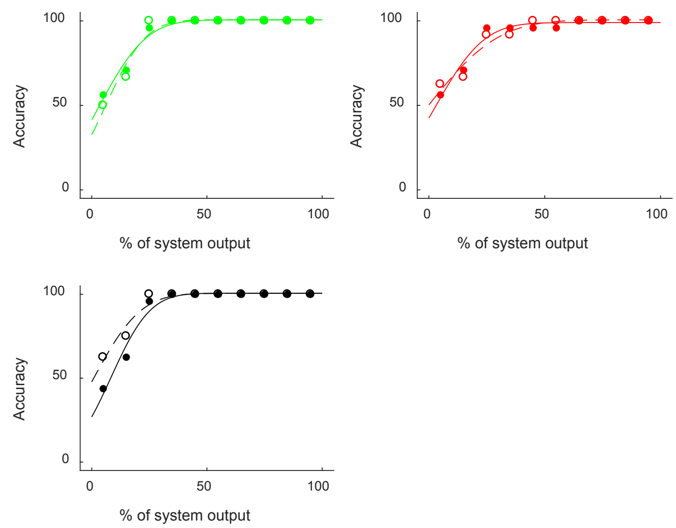
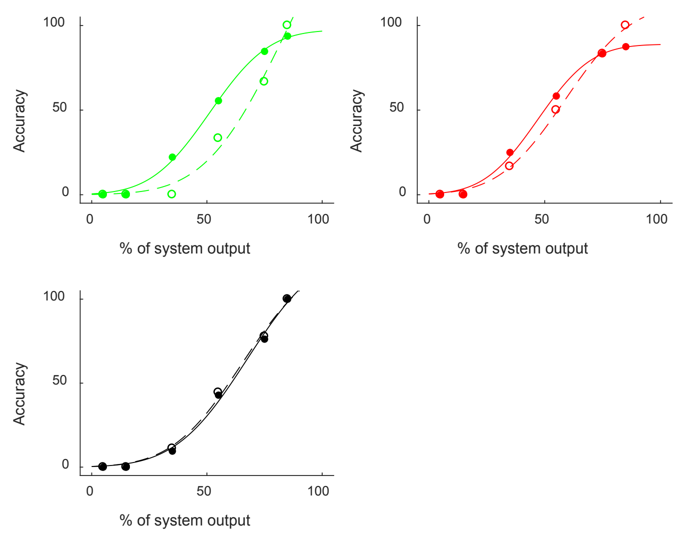
<tr><td>Pilot_04</td><td>
</td></tr>
<tr><td>Pilot_05</td><td>
</td></tr>
</table>

Appendix B. Supplementary material for Chapter 3

B.1 Complete within- and between-session reliability metrics for all tasks.

Available online at https://doi.org/10.1016/j.gaitpost.2019.04.023 under Appendix 1

B.2 Effect of filtering on movement quality metrics and reliability metrics.

Available online at https://doi.org/10.1016/j.gaitpost.2019.04.023 under Appendix 2

B.3 Mean (SD) within-subject standard deviations for the movement quality metrics.

Available online at https://doi.org/10.1016/j.gaitpost.2019.04.023 under Appendix 3

Appendix C. Supplementary material for Chapter 4

Table C.1. Complete movement quality metrics for all tasks.

			BP	MYO	Anatomical
Duration	**Reaching**	**CAN**	1.02 (0.46)	1.17 (0.11)	0.75 (0.08)
		PILL	0.64 (0.17)	0.65 (0.09)	0.50 (0.11)
		PIN	1.02 (0.24)	1.40 (0.48)	0.85 (0.11)
	Manipulation	**CAN**	0.81 (0.27)	1.00 (0.48)	0.07 (0.03)
		PILL	0.43 (0.17)	0.76 (0.41)	0.19 (0.14)
		PIN	2.09 (0.80)	3.55 (0.58)	0.79 (0.30)
IOC	**Reaching**	**CAN**	1.19 (0.04)	1.12 (0.09)	1.06 (0.03)
		PILL	1.07 (0.03)	1.08 (0.04)	1.07 (0.04)
		PIN	1.07 (0.02)	1.10 (0.07)	1.04 (0.01)
LDJ	**Reaching**	**CAN**	9.06 (0.76)	8.18 (0.67)	7.31 (0.49)
		PILL	7.08 (0.67)	6.51 (0.84)	6.02 (0.51)
		PIN	8.06 (0.10)	8.20 (0.88)	7.65 (0.44)
	Manipulation	**CAN**	9.97 (0.79)	10.33 (2.33)	1.33 (1.85)
		PILL	7.86 (1.63)	8.92 (1.62)	2.68 (3.13)
		PIN	14.10 (1.48)	16.05 (1.19)	9.77 (1.49)
SPARC	**Reaching**	**CAN**	1.51 (0.05)	1.51 (0.04)	1.47 (0.02)
		PILL	1.49 (0.01)	1.53 (0.12)	1.49 (0.04)
		PIN	1.44 (0.02)	1.49 (0.11)	1.43 (0.01)
	Manipulation	**CAN**	2.91 (1.17)	3.40 (1.19)	1.70 (0.32)
		PILL	2.62 (0.26)	3.92 (0.48)	2.28 (0.82)
		PIN	4.12 (1.46)	4.62 (0.79)	3.00 (0.55)

Appendix D. Supplementary material for Chapter 5

Table D.1. Mean (SD) trunk ROM for all tasks.

		BP	MYO	Anatomical
CAN	Lateral Lean (°)	23.62 (8.66)	10.32 (6.34)	4.94 (1.74)
	Axial Rotation (°)	19.74 (3.49)	14.65 (2.54)	12.24 (3.05)
	Flexion (°)	43.26 (10.42)	25.30 (19.95)	5.60 (2.12)
PILL	Lateral Lean (°)	4.67 (2.11)	4.55 (3.65)	1.07 (0.53)
	Axial Rotation (°)	9.62 (1.66)	6.07 (2.37)	2.34 (0.93)
	Flexion (°)	2.48 (0.70)	2.02 (0.98)	1.01 (0.39)
PIN	Lateral Lean (°)	16.88 (4.20)	9.68 (4.19)	2.48 (1.18)
	Axial Rotation (°)	8.72 (0.87)	10.91 (4.50)	6.25 (3.03)
	Flexion (°)	5.53 (1.55)	8.42 (2.63)	2.33 (0.88)

Appendix E. Supplementary material for Chapter 6

E.1 Full survey

Available online at https://doi.org/10.1371/journal.pone.0182482.s001

E.2 Distribution plots for the factors and outcome measures

Available online at https://doi.org/10.1371/journal.pone.0182482.s002

E.3 Additional logistic regression models

Available online at https://doi.org/10.1371/journal.pone.0182482.s003

E.4 All self-selected functions listed by 61 participants

Available online at https://doi.org/10.1371/journal.pone.0182482.s004

E.5 All written comments regarding the interfaces

Available online at https://doi.org/10.1371/journal.pone.0182482.s005

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