IMPACT OF PREHENSOR STIFFNESS ON QUALITY OF HAPTIC FEEDBACK IN BODY-POWERED UPPER LIMB PROSTHESES

by

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ABSTRACT

The design goal of a prosthesis is to replicate or replace the lost functionality of the missing limb. However, currently available upper limb prosthetic devices generally fall short of this goal, and as a consequence are often abandoned or rejected by the users. Potentially contributing to this shortfall is the lack of stiffness modulation in commercial prostheses — whereas intact individuals are able to adapt the stiffness properties of their limbs to the requirements of specific tasks, prosthesis users do not have this option.

One particularly troubled area of prosthesis functionality is the quality of haptic feedback provided by the prosthesis to the user. Body-powered prostheses, which account for the majority of all upper limb prostheses in use, provide a degree of haptic feedback by default, a consequence of the direct mechanical coupling between the actuating joint and the prehensor. The usually static prehensor stiffness is part of this mechanical linkage, and may be affecting the quality of the feedback transmitted to the user. Is it possible to improve the performance of the user in feedback-dependent tasks by modulating the prehensor stiffness of a body-powered prosthesis depending on the nature of the task?

This work, attempts to answer this question based on the results of three successive human subject studies, in which able-bodied volunteers used a prosthesis emulator system to complete feedback-dependent tasks at various prehensor stiffness settings. The main conclusions may be summarized as 1) prehensor stiffness has a quantifiable and significant impact on the quality of haptic feedback provided to a prosthesis user, 2) the optimal prehensor stiffness varies depending on the task, and 3) the users are aware of the impact of prehensor stiffness on their performance, and are able to make informed, task-dependent adjustments. These three conclusions serve as an endorsement for the inclusion of prehensor stiffness modulation as a control modality in the design of future prosthesis, which may result in increased user satisfaction and capability.

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LIST OF SYMBOLS

prehensor stiffness	K_D
test object stiffness	K_O
displacement magnitude	D
sensitivity index	d'

LIST OF ABBREVIATIONS

Body-powered	BP
Prosthesis Emulator	PE
Signal Detection Theory	SDT
Voluntary Closing	VC
Voluntary Opening	VO
Activities of Daily Living	ADL
Electromyography	EMG
Degree of Freedom	DOF
Two Alternative Forced Choice	2AFC
Four Alternative Forced Choice	4AFC

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For those that shall follow after.

CHAPTER 1

STIFFNESS MODULATION IN THE CONTEXT OF UPPER-LIMB PROSTHESES

A prosthesis aims to imitate the functionality of the limb it is intended to replace. In the case of prosthetic hands, both body-powered and robotic, much of the focus has been on position control and force control - the former is a virtual necessity for any active prosthesis, while the latter is a logical progression. However, limb stiffness modulation is a control modality often overlooked in prosthesis design, and this work aims to explore the potential impact of this shortfall on task performance of body-powered prosthesis users in tasks dependent on sensory feedback. The biomechanics of an intact arm enable individuals to modulate the directional endpoint stiffness of their limb through co-contraction of agonist/antagonist muscle pairs [1, 2] Often, limb stiffness is adapted to meet the requirements of a particular task [3–5] For example, stiffness modulation is useful when operating in unstable and unknown environments [3, 4, 6] or performing a constrained task [7]. In the area of prosthesis design, while the need for position and force control has long been recognized, the concept of prehensor stiffness modulation has been mostly ignored. Though the lack of control over the stiffness of a prosthesis was first pointed out as a limitation in 1982 [8], no current commercial upper-extremity prosthesis provides the capability of changing the stiffness of the prehensor. For body-powered prostheses, the prehensor stiffness is dependent on the stiffness of the mechanical spring installed within the prehensor assembly. Various manufacturers provide devices with a range of spring constants [9], but the spring stiffness cannot be easily changed during operation, thus preventing prehensor stiffness modulation. Thus, in contrast with an intact limb, a prosthesis user cannot modulate end point stiffness to match the requirements of specific tasks, potentially limiting performance. Introducing stiffness modulation as a control modality may improve user satisfaction with the functionality of the available prostheses, which currently remains low [10]. Among the many concerns cited, the quality of sensory feedback provided by the prostheses is often cited as an important factor in device abandonment [11]. Unlike robotic devices, body-powered devices inherently provide the user with haptic feedback while the device is in use. Due to the purely mechanical nature of the actuation and sensory feedback mechanisms in body-powered prostheses, prehensor stiffness may have a significant impact on the performance of the system. This work aims to expand on and supplement the existing body of literature by exploring the impact of stiffness modulation on the performance of a body-powered prosthesis system in various tasks. The central hypothesis of this work is that operating an upper limb body-powered prosthetic device at a single stiffness setting during different tasks limits or diminishes the quality of haptic feedback provided by the prosthesis. Thus, introducing stiffness modulation to a body-powered prosthesis may result in improved task performance, usability, and user acceptance.

1.1 Literature Review

The modulation of human joint impedance, and specifically joint stiffness, has been a topic of study for decades, generating a wide ranging body of literature. Numerous studies have analyzed the biomechanics involved, tested the stiffness response of individual joints and whole limbs during various tasks, and proposed different neural control strategies of various levels to explain the control exercised by the brain over the joint stiffness. Upper extremity prostheses continue to be an active research topic. The majority of this recent research has emphasized robotic prostheses, employing electromyography (EMG) control, despite the continued popularity of simpler body-powered devices. Though some groups have explored stiffness modulation as a potential control modality for prostheses, these efforts have also been almost exclusively aimed at robotic devices.

1.1.1 Joint Stiffness Modulation in Human Motion

Joint impedance, which encompasses joint stiffness, damping and inertial properties of the limb, is an intrinsic component of human motion, and has been the subject of numerous studies. Initial explorations of the subject [12], were limited to static estimates of impedance, and concluded that joint impedance was mostly a function of the passive elements within the joint. The prevailing assumption was that the spring-like properties of the joint arose from the spring-like properties of individual muscles and tendons. However, the 1980s saw two equally important developments that significantly enhanced our understanding of joint impedance. First, the concept of muscular co-contraction (also known as co-activation) was formalized [1, 8]. This provided a theoretical framework for joint impedance modulation - joint stiffness could be actively managed by the individual through the co-contraction of agonist/antagonist muscle pairs around a joint. Second, a rigorous experimental method for measuring the dynamic, directional impedance of human limbs was established [2]. This method relied on measuring the restoring forces following minute mechanical displacements of the limb from an equilibrium posture, and remains the primary technique for estimating joint impedance to this day.

Subsequent work [13] extended the mechanical perturbation method to include estimations of dynamic multi-joint impedance. Further work pointed to the feasibility of reducing the number of data collection trials necessary for accurate joint impedance estimation by using undisturbed velocity profiles as a basis for comparison [14]. Additionally, the use of stochastic disturbances, which have the advantage of eliminating voluntary response as well as evaluating all impedance components simultaneously, has been recommended [15]. As an alternative to mechanical perturbation, direct measurements of muscle co-contraction around the joint through EMG sensors have been used to estimate joint stiffness [6, 16, 17](While seemingly logical, this method runs into problems common with EMG measurements, such as dependence on static characterization of muscle behavior, limiting its precision, especially during dynamic movements. However, EMG measurements are capable of providing an on-line, quantitative estimate of joint impedance, something potentially useful in an active prosthesis context.

Of the three components of joint impedance – stiffness, damping and inertia — it is the first that has received the most attention. The stiffness of virtually all joints in the upper limb, ranging from the fingers [18, 19] and the hand [20], to the entire arm [21] has been characterized, using a variety of experimental tasks and apparatuses. It has been shown that while generating end point force contributes to the unidirectional stiffness of the limb along the axis of the applied force vector [22, 23], stiffness modulation can be independent of net joint torque [3, 16]. This torque-independent stiffness modulation is advantageous when adjusting for instability in the environment [3]. It has been demonstrated that humans have the ability to selectively alter the orientation of their endpoint stiffness to deal with directional instability in the environment through time varying co-activation of various antagonist muscles [4]. This ability holds across multiple memorized movement trajectories [5]. Increased levels of muscle co-contraction and joint stiffness have been observed during constrained pointing tasks [7, 17, 24]. Joint stiffness modulation has also been shown to be instrumental in limiting the effects of signal dependent noise within the neuromuscular system [7, 25]. Stiffness modulation is instrumental in learning novel tasks as well. Increased co-contraction improves stability during initial, uncertain movements, and as familiarity increases, co-contraction levels drop in order to reduce metabolic costs [6, 26].

While there is strong evidence of humans employing various joint stiffness modulation strategies to improve performance in diverse tasks, there is little indication that the other two components of limb impedance — joint damping and limb inertia — are similarly modulated. As may be expected, the contribution of limb inertia to joint and endpoint impedance is entirely dependent on the posture of the limb [21]. And despite numerous studies, no strong evidence of active joint damping modulation has been found [21, 24, 27].

In summary, joint stiffness, a component of overall limb impedance, is actively modulated by the central nervous system in order to improve performance in various tasks. This modulation is enabled through the co-contraction of agonist/antagonist muscle pairs around the joint. No strong evidence of task dependent joint damping modulation has been found to date.

1.1.2 Upper-Limb Prostheses

It is estimated that currently over 500,000 people are living with upper limb loss in the United States alone, and this number is expected to rise in the future [28]. Many activities of daily living (ADLs), actions essential to functional independence, are tied to hand and arm functionality [29]. In order to compensate for the loss of function, three broad classes of prosthetic devices are available on the market cosmetic, body-powered, and robotic (myoelectric).

Figure 1.1: The two types of active upper limb prostheses: (a) body-powered (Westcoast Brace and Limb), and (b) robotic (i-Limb Ultra)

Cosmetic devices are passive prostheses, intended to replace only the aesthetic appearance of the original limb, and thus present little interest from a task performance perspective. Body-powered and robotic prostheses are both considered active devices, as they are intended to restore varying levels of lost functionality to the user. An example of each type of active device is presented in Figure 1.1. Body-powered devices are usually cable actuated mechanical systems, relying on the movement of an intact joint (commonly the shoulder, contra-lateral to the site of the limb deficiency) to actuate a single degree of freedom (DOF) prehensor. Possible modes of operation include voluntary opening (VO) and voluntary closing (VC). In both modes the user actuates the prosthesis by applying tension to a cable. However, in

VO mode, the tension opens the prehensor aperture, while in VC mode the tension closes the aperture. Modern robotic prostheses are electromechanical systems, generally with more than one DOF actuated by electrical motors, with the control signal usually provided by integrated EMG sensors.

Variations in reporting standards, and the use of multiple types of prostheses by individuals make it difficult to arrive at a specific usage distribution between the three classes of devices. However, it should be noted that body-powered prostheses remain popular [10], with estimated usage rates either above [30, 31], or on par with those of robotic devices [32]. In addition to lower cost and weight, their primary appeal is the inherent presence of sensory force feedback, provided by the mechanical actuation action [33].

Recent advances in computing and sensor technology have contributed to something of a boom in prosthesis development [34], disproportionally targeting robotic devices [31]. Control of myoelectric devices has been improved through the development of implantable EMG sensors [32, 35] and direct neural interfaces provide the potential for further improvement [34]. Interfacing with the peripheral or central system also has the potential to provide natural somatosensory feedback to the user [36], a feature lacked by current myoelectric prostheses. Sensory feedback has long been an issue with robotic prostheses [37], and its presence has been shown to clearly improve the performance of the prosthesis user in functional tasks [38].

Despite these developments, abandonment rates for all types of upper limb prostheses remain high [29], reaching upwards of 40% [10, 29]. In addition to outright rejection, a significant portion of all device operators choose to use their prosthesis passively [10], effectively relegating it to cosmetic status. Individuals who reject prostheses are often dissatisfied with the level of functionality current technology provides, with 85% of rejecting users pointing to the low quality of haptic feedback provided as a significant factor in the decision to not wear the prostheses [11]. Additionally, prosthesis users with unilateral amputations report high rates of cumulative trauma disorder - additional injuries and health issues arising from overuse of the remaining healthy limb [29]. Despite these challenges, the opportunity to better address the needs of this population remains open, with 74% of rejecting users reporting a willingness to reconsider prosthesis use if improvements to device utility could be made at a reasonable cost [11].

In summary, despite recent advances in prosthetic technology, user satisfaction with upper-limb prostheses remains low. The many recent developments in sensor biointegration and computing, while promising, are still years away from widespread use, and are relevant only for those individuals that use a myoelectric prosthesis. Body-powered prosthesis users, a significant subset of the affected population, are stuck using a device that has not changed in any meaningful way in decades. A cost effective improvement in the quality of haptic feedback and comfort offered by body-powered prostheses has the potential to improve the lives of numerous persons with upper limb loss.

1.1.3 Stiffness Modulation in Prostheses

In 1982 the need for stiffness modulation in prosthetic devices was acknowledged [8] in a paper appropriately titled *Prostheses should have adaptively controllable impedance*. Over thirty years hence, prosthetic stiffness modulation remains relegated to the status of a lab experiment [31]. Current commercially available upper extremity prosthetic devices make virtually no provisions for modulating the stiffness of the end effector. In the case of body-powered prostheses, the end effector stiffness is a function of the stiffness of the spring in the prehensor, and the stiffness of the body joint driving the actuator. The stiffness of the prehensor spring varies between manufacturers [9], however, the spring stiffness cannot be modified on the fly, and the body joint can control stiffness in only one direction, a function of the cabled actuation system. In commercial robotic devices, joint stiffness is completely independent of user intention, and is secondary to velocity, and in some cases, force control.

On the research side, several recent studies have explored the utility of an upper-limb prosthetic device incorporating joint impedance and/or stiffness modulation. Sensinger and Weir designed and tested a single DOF series elastic actuated elbow prosthesis, with joint impedance controlled by EMG readings of the co-contraction of the users biceps and triceps

[39]. Though incorporating innovative mechanical solutions, the noisiness and unreliability of the surface EMG signal used as the control input presented serious challenges, with users having difficulty in maintaining a selected level of impedance.

Approaching the problem from a teleoperation perspective, Ajoudani et al. introduced the concept of tele-impedance [40, 41] , slaving the impedance properties of a robotic manipulator to the co-contraction levels in the operator's arms. This enhanced the controllability of the system, resulting in a reduction of contact forces during a peg-in-hole manipulation task, reinforcing the potential benefits of incorporating this control modality in a prosthetic device. Further work by the same group [42] as well as others [43] implemented the tele-impedance concepts in robotic prosthesis simulators.

The positive impact of impedance modulation on user performance in trajectory following and disturbance resistance tasks has been demonstrated in virtual [44] and idealized physical systems [45]. The capability of users to consciously self-determine the appropriate impedance characteristics of the system was also evaluated [44], with the results being somewhat discouraging — experiment participants were not able to independently identify the optimal impedance characteristics for maximizing task performance. It is possible that the use of continuously variable impedance characteristics, as well as the requirement of independently selecting stiffness and damping parameters may have overcomplicated the task.

Still, these studies have been subject to several limitations. In general, they either used idealized modes of control [44, 45], or were focused on EMG-driven systems, which while extremely promising, are less commonly used than the body powered prostheses [30, 31]. Additionally, all of them focused on the role stiffness modulation plays in enabling either precise motion, or stability against external disturbance, i.e. its impact on the *control* of the device, leaving aside its potential impact on the quality of sensory *feedback*.

In the realm of body-powered prostheses, Frey and Carlson [46] proposed a body-powered prosthesis with a variable mechanical advantage, but the main focus of the work was in raising the grip force of the device. Smit and Plettenburg performed a systematic characterization of commercially available voluntary closing (VC) body-powered prostheses [47], noting concerns about the mechanical efficiency of the devices. Subsequently, Smit et al., conducted a similar study of body-powered voluntary opening (VO) devices [48], and proposed a design of a mechanism to offset the excessive stiffness present in VC device joints [9]. However, this work stopped short of exploring stiffness modulation as a control modality. More recently, Brown et al. [33] conducted a study of the utility of force feedback in body-powered prostheses, confirming its importance in enabling users to complete an object stiffness identification task.

In short, despite the demonstrable utility of stiffness modulation as a control paradigm in an upper-limb prosthesis context, commercially available prostheses lack any stiffness modulation capability. Research in the area has overwhelmingly focused on robotic, myoelectrical devices, virtually ignoring the possible impact of stiffness modulation in bodypowered prostheses on task performance. As a consequence of the direct mechanical linkage used to actuate it, the introduction of stiffness modulation to a body-powered prosthesis has the potential to impact not only the controllability of the device, but also the feedback quality. Since feedback quality is an oft-cited reason for prosthesis abandonment, any improvements in this aspect of body-powered prostheses can have a lasting impact on many prosthesis users. However, no study to date has quantified the possible impact of stiffness modulation on the quality of haptic feedback provided to a prosthesis user.

1.1.4 Summary of Literature

In summary, joint stiffness modulation is an intrinsic control modality used by the central nervous system to adapt the impedance characteristics of the limbs to better suit the requirements of a particular task (environmental instability, etc.). Despite this, current prosthetic devices, which are called on to replace the functionality of a lost or absent limb, do not incorporate any provisions for stiffness modulation. This oversight is especially noticeable in light of the generally low levels of satisfaction among upper-limb prosthesis users, particularly when it comes to the quality of sensory feedback provided by the prosthesis.

An additional source of concern is the high incidence of comorbidities in the actuating joint among body-powered prosthesis users, likely caused by overuse.

The introduction of stiffness modulation to experimental prosthetic devices and simulators has been shown to result in increased performance in tasks ranging from peg-in-hole insertions to disturbance resistance. However, the vast majority of all recent prosthesis research has focused on robotic devices, despite the continued popularity of the body-powered designs. Among their many advantages, the direct mechanical linkages used to actuate the body-powered prehensors inherently provide the user with intuitive haptic feedback about the environment. Because of this mechanical connection, the employment of prehensor stiffness modulation in a body-powered device has the potential to positively impact the quality of sensory feedback received by the user in the course of various tasks. In addition, matching the stiffness of the prehensor with the requirements of the task has the potential to reduce unnecessary loading in the actuating joint, potentially reducing the occurrence of cumulative trauma disorder in prosthesis users.

1.2 Overall Goal and Work Structure

The central goal of the present work is to evaluate, in the context of body-powered prostheses, the potential of prehensor stiffness modulation to improve users' performance in feedback based tasks. If evidence of such potential is found, the introduction of prehensor stiffness modulation as a control modality in a future generation of upper-limb prostheses could lead to improvements in user satisfaction, capability and health. To evaluate this potential, this work attempts to answer the following three questions:

- **Does prosthesis prehensor stiffness have a quantifiable impact on the quality of haptic feedback received by the user?**

  If the quality of haptic feedback provided by the prosthesis to the user is independent of the stiffness of the prehensor, then introducing stiffness modulation to prostheses would be of little value (and this would be a very short thesis!). Throughout all experiments completed in the course of this work, I use subjects' performance in feedback-dependent tasks as a proxy indicator for the quality of haptic feedback they received – the higher the performance, the better the quality of feedback.

- **Is the effect of prehensor stiffness on haptic feedback quality task dependent?**

  It is possible that the quality of haptic feedback provided to the user of a body-powered prosthesis is optimized at a given prehensor stiffness, regardless of the task at hand. In this case, prehensor stiffness modulation is unnecessary, as simply optimizing the set prehensor stiffness would optimize the feedback. Alternatively, it is possible that the optimal prehensor stiffness varies depending on the nature of the feedback-dependent task (or on individual user preferences), in which case modulating the prehensor stiffness to match the requirements of the task is necessary.

- **Do users have the capacity to accurately asses the impact of prehensor stiffness on their performance?**

  The addition of any control modality to a user controlled system only makes sense if the users are willing and able to take advantage of the new capabilities. If the effect of prehensor stiffness on users' task performance is too minor or subtle for them to notice, it is unlikely that they would be willing to expend the necessary effort to adjust it.

In order to answer these questions, I conducted a series of human subject experiments, using a body-powered prosthesis emulator (PE) system to test the performance of volunteers in feedback-dependent tasks at various prehensor stiffness conditions. These experiments were organized into three separate studies, and a brief summary of their results is presented in Table 1.1, with the corresponding chapter providing extensive detail on the design and outcome of the experiments. All experiments made use of a repeated-measures design, and involved only able-bodied volunteers using a PE system described in detail in Section2.2. Along with answers to the three main questions listed above, each study provided additional insights into the influence of prehensor stiffness and its potential as a future prosthesis control modality. These insights are summarized in a brief Contributions section at the end of each relevant chapter.

Table 1.1: A Summary of the Three Studies Undertaken

Thesis Chapter & Study Title	Specific Aims	Main Contributions
Chapter 2 *The Effect of Prosthesis Prehensor Stiffness on Performance in an Object Discrimination Task*	Evaluate of the potential impact of prehensor stiffness on user performance in an object-identification task Determine the validity of conducting experiments using virtual objects as stand-ins for physical objects	Established that prehensor stiffness does have a quantifiable impact on user performance in an object-identification task, with a low prehensor stiffness resulting in improved performance Showed that the effect of prehensor stiffness on task performance may be observed using both virtual and physical objects
Chapter 3 *Task Dependent Effects of Prehensor Stiffness on Feedback in Body-powered Prostheses*	Investigate the effect of prehensor stiffness on user performance in a second feedback-dependent task: disturbance detection Confirm the findings of the previous study through additional data collection Determine if the impact of prehensor stiffness is present regardless of task difficulty	Showed that a high prehensor stiffness is advantageous in a disturbance detection task, and confirmed previous findings that a low prehensor stiffness is helpful during object identification tasks By demonstrating that the effect of prehensor stiffness on performance is task dependent, established a motivation for the introduction of prehensor stiffness modulation in body-powered prostheses Showed that the effect of prehensor stiffness is present regardless of the difficulty of the task
Chapter 4 *On the ability of users to independently adjust prehensor stiffness to enhance task performance*	Examine whether the effect of prehensor stiffness on user performance is linear Evaluate the ability of subjects to self-determine an appropriate prehensor stiffness level for a given task, based only on their own perception of their performance	Confirmed that prehensor stiffness has a linear effect on the subject performance in feedback-based tasks Established that subjects are aware of the impact of prehensor stiffness on their performance, and are able to identify an appropriate stiffness setting for a given task

CHAPTER 2

THE EFFECT OF PROSTHESIS PREHENSOR STIFFNESS ON PERFORMANCE IN AN OBJECT DISCRIMINATION TASK

In order for an upper limb prostheses to faithfully reproduce the behavior and function of the intact limb, the device must incorporate all control modalities of the organic appendage. One control modality that has been largely absent from prosthesis design is stiffness modulation. Able-bodied individuals have the ability to alter the directional endpoint stiffness of their arms through co-contraction of agonist/antagonist muscle pairs. In contrast, prosthesis users are largely forced to use a device of a set stiffness to perform all manipulations. In this chapter [1], I attempt to broaden the existing knowledge base by (1) using an experimental setup which emulates the control-feedback loop of a shoulder-driven body-powered prosthesis operated in voluntary closing (VC) mode, and (2) exploring the impact of prehensor stiffness modulation on the quality of haptic feedback the user receives about the environment. I conducted two human subject experiments, with subjects using the prothesis emulator to perform two feedback dependent tasks at different prehensor stiffness settings. Analysis of the results, based on signal detection theory, suggests that modulation of prehensor stiffness in a body-powered prosthesis can increase the performance of the user in object discrimination and classification tasks.

2.1 Introduction

A prosthesis aims to imitate the functionality of the limb it is intended to replace. In the case of prosthetic hands, both body-powered and robotic, much of the focus has been on position control and force control – the former is a virtual necessity for any active prosthesis, while the latter is a logical progression. However, limb stiffness modulation is a control

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[1] A preliminary version of the analysis discussed here has been presented at the 2015 World Haptics Conference, and an expanded version of the publication is currently under peer review in *IEEE Transactions on Haptics*

modality often overlooked in prosthesis design, and this paper aims to explore the potential impact of this shortfall on task performance of body-powered prosthesis users in feedback-dependent tasks.

The biomechanics of an intact arm enable individuals to modulate the directional end-point stiffness of their limb through co-contraction of agonist/antagonist muscle pairs [1, 2]. While generating end point force contributes to the unidirectional stiffness of the arm along the axis of the applied force vector, [22, 23], stiffness modulation can be independent of net joint torque [3, 16]. Often, limb stiffness is adapted to meet the requirements of a particular task [3–5]. For example, stiffness modulation, independent of applied force, is useful when operating in unstable and unknown environments [3, 4, 6] or performing a constrained task [7]. It should be noted that limb stiffness is only a component of overall limb end point impedance, with damping and inertial effects also providing contributions. However, evidence for damping modulation by humans is less conclusive [23, 24], and inertial effects are entirely posture dependent [21].

It is estimated that currently over 500,000 people are living with upper limb loss in the United States alone, and this number is expected to rise in the future [28]. Many activities of daily living (ADL), actions essential to functional independence, are tied to hand and arm functionality [29]. Abandonment rates for upper limb prostheses remain high [29], reaching upwards of 40% [10, 29]. In addition to outright rejection, a significant portion of device operators choose to use their prosthesis passively[10], effectively relegating it to cosmetic status. Individuals who reject prostheses are often dissatisfied with the level of functionality current technology provides, with 85% of rejecting users pointing to the low quality of haptic feedback provided as a significant factor in the decision to not wear the prostheses[11]. Additionally, prosthesis users with unilateral amputations report high rates of cumulative trauma disorder - additional injuries and health issues arising from overuse of the remaining healthy limb [29]. Despite these challenges, the opportunity to better address the needs of this population remains open, with 74% of rejecting users reporting a willingness to reconsider prosthesis use if improvements to device utility could be made at a reasonable cost [11]. Thus any changes to the design of the prosthetic device that result in increased functionality or have the potential to alleviate overuse injuries can have a significant impact on the prosthesis user community.

In the area of prosthesis design, while the need for position and force control has long been recognized, the concept of end effector stiffness modulation has been mostly ignored. Current commercially available upper-extremity prosthetic devices make virtually no provisions for modulating the stiffness of the end effector. In the case of body-powered prostheses, the end effector stiffness is a function of the stiffness of the spring in the prehensor, and the stiffness of the body joint driving the actuator. The stiffness of the prehensor spring varies between manufacturers [9], however, the spring stiffness cannot be modified on the fly, and the body joint can control stiffness in only one direction, a function of the cabled actuation system. In EMG-controlled, robotic devices, joint stiffness is completely independent of user intention, and is secondary to velocity, and in some cases, force control.

Recently several groups have begun to focus on this issue [40, 43, 45, 49]. Ajoudani et al. [40], as well as Hocaoglu and Patoglu [43] demonstrated tele-impedance control of a robotic system as a stand in for a prosthetic device. Blank et al. [45] demonstrated the utility of impedance modulation in two tasks (contact force minimization and trajectory tracking) using both virtual and robotic systems. However, these studies have been subject to several limitations. In general, they either used idealized modes of control [45], or were focused on EMG-driven systems, which while extremely promising, are less commonly used than the body powered prostheses [30]. Additionally, all of them focused on the role stiffness modulation plays in enabling either precise motion, or stability against external disturbance, i.e. its impact on the *control* of the device, leaving aside its potential impact on *feedback*. Smit and Plettenburg performed a systematic characterization of commercially available voluntary closing (VC) body-powered prostheses [47], noting concerns about the mechanical efficiency of the devices. Subsequently, Smit conducted a similar study of body-powered voluntary opening (VO) devices [48], and proposed a design of a mechanism to offset the excessive stiffness present in VC device joints [9]. However, this work stopped short of exploring stiffness modulation as a control modality. Frey and Carlson [46] proposed a body-powered prosthesis with a variable mechanical advantage, but the main focus of the work was in raising the grip force of the device.

To summarize, intact individuals possess the ability to modulate the stiffness of their limbs, and do so in order to improve performance in specific tasks. Individuals dependent on a prosthesis lack this control modality entirely, which can negatively impact their capability to complete various activities, including ADLs. In addition, when looking at body-powered devices specifically, the constant stiffness of the prehensor may contribute to the high incidence of secondary health issues in the contralateral shoulder. Current user satisfaction with prosthesis functionality is low, while rejection and passive use rates are high. Introducing stiffness modulation to a VC body-powered prosthesis may result in improved task performance, and potentially drive down the incidences of comorbidities in the contralateral actuating joint.

In this study I attempt to broaden the existing knowledge base by (1) using an experimental setup which emulates the control-feedback loop of a shoulder-driven body-powered prosthesis operated in voluntary closing (VC) mode, and (2) quantifying the impact of end effector stiffness modulation on the quality of haptic feedback the user receives about the environment.

The magnitude of the force feedback displayed by a body-powered prosthesis to a user during interaction with an object is proportional to the sum of two parameters: K_D, the device (prehensor) stiffness and K_O, the object stiffness, i.e.

$$F \propto K_D + K_O \tag{2.1}$$

Thus, if

$$K_D > K_O \tag{2.2}$$

the majority of the feedback force, F, is devoted to providing the user with feedback about the internal forces of the prehensor, instead of informing the user about the response of the object of interest. Therefore, I hypothesize that minimizing the value of the prehensor stiffness, K_D, will result in improved user performance during an object stiffness discrimination task.

This paper presents the results of two related and sequential human subject experiments. Twin aims were pursued in Experiment 1: (1) to explore the hypothesis that prehensor stiffness affects the ability of a body-powered prosthesis user to distinguish between two objects of different stiffnesses, and (2) to investigate whether this effect is observable when interacting with both real and virtual objects. The subjects were asked to use a prosthesis emulator (PE) device (shown in Figure 2.1, see [50] for details on an earlier version of the device) to manipulate an object and categorize it as either a low or a high stiffness spring. Each subject performed this task was with the PE system operating at both a low and a high stiffness setting, and with real and virtual objects. The results were evaluated through the application of signal detection theory (SDT) analytical framework, [51, 52] which has previously been shown to be viable in experiments involving physical and virtual object discrimination [53].

Experiment 2 was designed to explore the performance trends suggested by Experiment 1 further, by increasing the number of prehensor stiffness settings from two to three, and raising the difficulty of the task by expanding the object set from two to four, creating a classification experimental paradigm [51]. Only virtual objects were used for Experiment 2. Initial results from these experiments were presented in [54], where the analyses were limited to only descriptive statistics based on identification error rates, a metric which is demonstrably inferior to sensitivity indices [51]. The current paper substantially expands the analysis, results and discussion of these experiments through using the SDT framework, combined with inferential statistics.

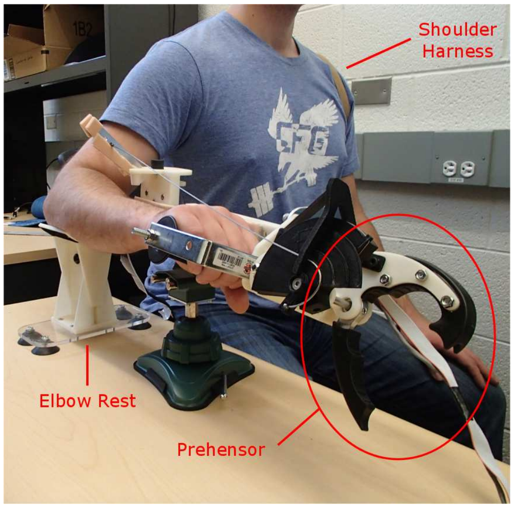

Figure 2.1: Prosthesis emulator experimental setup. Subjects applied tension on a Bowden cable via a shoulder harness which moved the prehensor for voluntary closing.

The remainder of this chapter is organized as follows: Section 2.3 is focused on a pilot study, an object discrimination experiment; the section provides a description of the methods and experimental procedure used in this pilot study, and presents and discusses the results obtained. Section 2.4 presents the same information for the subsequent object classification study. Section 2.5 contains a summary of the contributions this study made to addressing the overall questions targeted by this work.

2.2 Methods

2.2.1 Experimental Setup

The PE system consists of a single degree of freedom (DOF) prosthesis prehensor, attached to a 3D printed base which adapted it for use by able-bodied individuals (see Figure 2.1). The PE prehensor is mechanically linked by a cable to a harness worn by subject around the contra-lateral (left) shoulder, imitating the actuation mechanism of a body-powered upper limb prosthesis. Tension in the actuating cable, generated by a voluntary displacement of the subject's left shoulder, produces a torque around the prehensor shaft by acting on a lever arm rigidly attached to the shaft. The produced torque actuates the prehensor, closing the aperture. The output of an electrical motor (Maxon RE 30 DC 24V, 3.81A), is also connected to the prehensor shaft, through a cable drive with a gear ratio of 12:1 (Note: all torque measurements and constants are reported at the prehensor shaft). An image of these connections is presented in Figure 2.2. During the experiments, the torque of the electrical motor was employed to create a simulated, torsional spring with a constant of K_D. This linear torsional spring acted to keep the prehensor open, aping the behavior of a mechanical spring usually present in a VC prosthetic prehensor. Maximum range of motion for the prehensor is 100° , and the effective actuating lever arm may be set at five discrete values. The prehensor stiffness K_D, is continuously variable between it is practical minimum and maximum values - 3.7 N-mm/° and 9.3 N-mm/°. The lower limit is imposed by the frictional forces present within the system - a K_D below this threshold prevents the prehensor from opening once closed. The upper prehensor stiffness limit is a function of the maximum torque output of the motor connected to the prehensor shaft. This stiffness range is representative of currently available commercial body-powered prostheses [47].

Interaction of the prehensor with an object present inside of the prehensor aperture produces an additional torque on the prehensor shaft. This object interaction torque is displayed to the subject as a proportional rise in the force delivered through the cabled actuation system to the subject's shoulder. It should be noted that during this study the PE system operated in voluntary closing (VC) mode, meaning that the users' actuating motions resulted in the closing of the prehensor aperture. Many real-world body powered prostheses operate in voluntary opening (VO) mode, however only prostheses operating in VC mode provide the user with a much greater level of sensory feedback, and this mode of operation is commonly employed for demanding and intricate tasks.

During all data collection sessions, the subject remained seated, with the PE system positioned to the right of the subject, on a table approximately at waist height. The PE system was held immobile in an adjustable vise placed on the table (see Figure 2.1). The right elbow of the subject rested on a support, while the subject's right hand held the grip of the PE.

2.2.2 Analytical Framework

The data collected during both experiments was analyzed using the Signal Detection Theory (SDT) analytical framework. The techniques of SDT are specifically designed to evaluate the ability of human subjects to distinguish or discriminate between two or more similar stimuli [51] - in this case the varying levels of force feedback produced by the PE during interactions with objects of varying stiffness. The most directly applicable metric produced by SDT analysis is a sensitivity index, *d'*. In a simple two stimuli experiment, one of the objects is designated as the reference, the other as a signal. Based upon the subjects' responses, a subject specific sensitivity index, *d'* is calculated as follows:

$$d' = z\{H\} - z\{F\} \tag{2.3}$$

where H is the fraction of reference stimuli correctly identified, F is the fraction of false alarms (i.e. how often the subject mistook the signal object for the reference one), and the z{} operator denotes a z-transform, which transforms the rate statistics into a z-score. A sensitivity index of 0 indicates a subject making the discrimination decision by chance, and is consistent with results produced by random guessing. Higher values of *d'* indicate an increase in the subject's ability to discriminate between the two stimuli.

The use of *d'* as a performance measure in a discrimination experiment is preferable to other measures such as error rates. Unlike a simple error rate statistic, *d'* is explicitly independent of subject-specific biases, allowing for meaningful comparisons between subjects and conditions [51]. The hit rate and false alarm rates may be pooled between different subjects in a given condition to generate a single estimated group *d'*. This is most useful when the number of trials collected from each individual subject is relatively low, and this pooling approach is used in both experiments described here. In addition to *d'*, SDT recommends the use of decision time as a secondary performance measure. In the absence of a significant difference in sensitivity between different conditions, decision time is used as a "tie-breaking" indicator of performance [51].

In order to reveal statistically significant differences in the *d'* indices of various groups or conditions, confidence intervals for the measure must be established. The specific calculations involved are available in detail elsewhere [51]. Briefly, the variance, σ^2, is computed as follows:

$$\sigma^2 = \frac{H(1-H)}{2N_r\phi(H)^2} + \frac{F(1-F)}{2N_s\phi(F)^2} \tag{2.4}$$

where H is the fraction of reference stimuli correctly identified, F is the fraction of false alarms, N_r and N_s are the number of reference and signal trials respectively. The function $\phi(x)$ is defined as

$$\phi(x) = \frac{1}{\sqrt{2\pi}} e^{-.5z(x)^2} \tag{2.5}$$

Once the variance is calculated, confidence intervals may be established. All intervals reported here are at the 95% confidence level.

These analytical techniques are directly applicable to two alternative, forced choice experiments (2AFC). For experimental designs dealing with more than two possible responses, such as Experiment 2, modifications to the calculation of *d'* and the confidence intervals are required [51, 55], though they are not detailed here.

2.3 Experiment 1: Real and Virtual Objects

2.3.1 Prosthesis Emulator Configuration

During Experiment 1, the travel arc of the prehensor was limited to 90°. Tension in the actuating cable produced a torque around the prehensor shaft by acting on a 28.1 mm lever arm, provided by a pulley of the same radius to which the end of the cable was rigidly attached. During this experiment, in order to establish a low and high prehensor stiffness setting, the value of K_D switched between 3.7 N-mm/° and 9.3 N-mm/°, or K_D^- and K_D^+ respectively.

In order to provide two physical objects with varying spring constants, two Slo-Foam$^{\text{TM}}$ hand exercise foam blocks were used. Each foam block was fitted with a 3D printed interface, which enabled fast coupling and decoupling to and from the prehensor of the PE. The effective torsional spring constants of these foam blocks were experimentally determined, through repeated automated compression of the blocks in the PE prehensor at varying rates. Based on these experiments, the spring constant of the softer block, K_O^-, and the harder block, K_O^+, were estimated as $K_O^- \cong 10.8$ N-mm/° and $K_O^+ \cong 32.4$ N-mm/°. Here, the nonlinear and rate-dependent stiffness of the foam blocks was approximated by a simplified, linear model. Please see the Experiment 1 Discussion section for a more detailed examination of the impact of this approximation on the outcome of the experiment. In trials involving virtual objects, these torsional stiffness values were used to simulate the response of the foam blocks. Once the prehensor crossed a contact threshold, the prehensor motor simulated an additional linear torsional spring of the appropriate stiffness (K_O^+ or K_O^-), on top of the already present torsional spring acting to keep the prehensor open (K_D^+ or K_D^-). For the foam blocks, the contact threshold was a function of the physical dimensions of the block, and was set to 20°. For virtual objects the contact threshold was shifted to 30°, in an attempt to prevent saturation of the motor, which was called upon to simulate both the prehensor stiffness and the virtual object spring. A diagram of the PE system in this configuration is presented in Figure 2.3.

Therefore, the total force, F, displayed to the subject during any individual trial was

$$F = l(rK_D + (r - t)K_O) \tag{2.6}$$

where l is the lever arm of the cable interacting with the prehensor shaft, K_D is the device stiffness, r is the displacement of the prehensor from its resting position in degrees, t is the contact threshold, and

$$K_O = \begin{cases} K_O^-, K_O^+ \ if \ r \geq t \\ 0 \ if \ r < t \end{cases} \tag{2.7}$$

During all trials, to eliminate the potentially confounding effects of visual feedback, a partition was raised between the subject and the PE prehensor, blocking their view of the prehensor. To record the subject's responses, a keyboard was placed within reach of the

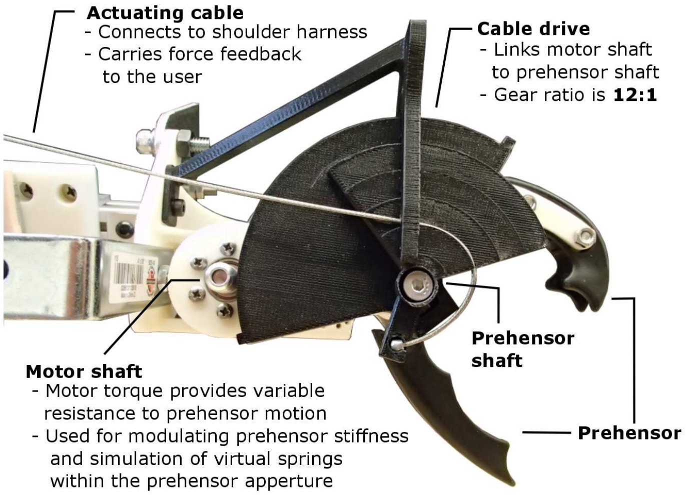

Figure 2.2: Detail of prosthesis emulator end effector linkages.

subject's left hand.

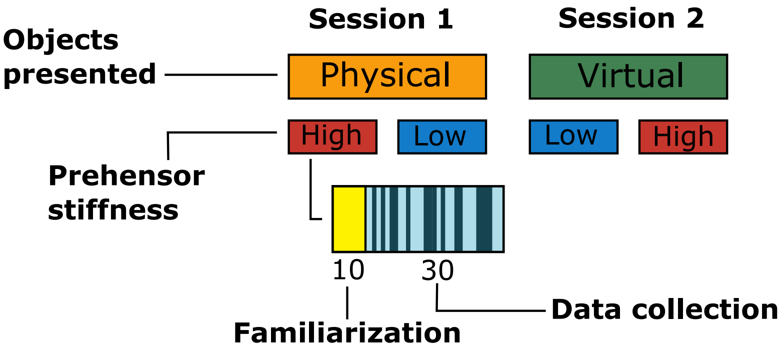

Figure 2.4: The order of presentation of various experimental conditions during in Experiment 1. The two data collection sessions were held on separate days. The order of presentation of experimental conditions (object type and prehensor stiffness) was randomized and counterbalanced between 8 subjects. In a given block, a total of 40 trials of the object stiffness discrimination task were completed. The first 10 trials of each block were considered to be part of the familiarization procedure. During each trial, a single low or high stiffness object was randomly selected and presented. Each object was presented a total of 20 times during each of the four blocks.

2.3.2 Procedure

Every subject completed two separate data collection sessions, held on different days. Each session was focused on a specific type of object, either real or virtual, meaning that no

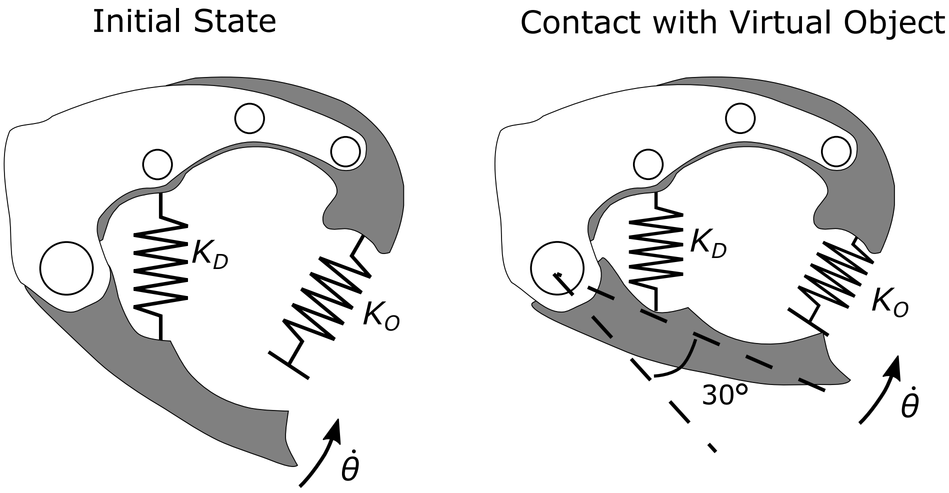

Figure 2.3: A diagram of the PE during the object identification task. Note that during the initial closing motion of the prehensor, it is opposed only by the prehensor spring , K_D. After the prehensor passes the contact threshold of 30 °, the effects of the virtual object begin contributing to the haptic feedback displayed to the operator.

subjects interacted with real and virtual objects on the same day. Every session was further subdivided into two blocks, each one dedicated to a single prehensor stiffness setting, K_D^- or K_D^+. The order of presentation of all conditions was randomized and counterbalanced across subjects, in an effort to reduce the possible impact of learning effects on the outcome of the trials. A visual example of the path of two subjects through their individual experimental condition sequence is presented in Figure 2.4. Each block contained 40 randomized trials, evenly split between trials dedicated to soft and hard objects (20 trials each). Unknown to the subjects, the initial 10 trials of each block were separated from the rest, balanced and randomized on their own. These initial 10 trials formed a dedicated training section within the block, and were excluded from all analyses. Finally, a familiarization session preceded every experimental block. During familiarization, the subjects were aware of which objects they were interacting with, and allowed to explore them at will. Between every block of 40 trials, the subjects took a break of 3-5 minutes.

In every trial, one object was simulated or physically placed within the aperture of the PE prehensor. The subjects were asked to interact with the object by actuating the prehensor through the cable actuation system, and categorize the object's stiffness as either high or low, based on the force feedback displayed to them through the shoulder harness. After making the determination, the subjects recorded their response by pressing the associated keyboard key. Crossing the contact threshold described above was used as an indicator of the start of the trial, and after the subjects pressed the keyboard key, the trial was considered over. Decision time was defined as the time between the two events. In addition to decision time, the subject's response was also recorded, to be compared to the actual object presented, establishing a measurement of the subject's accuracy and false alarm rates. During all trials, the subjects were asked to emphasize accuracy of object identification over a fast decision time. No ceiling was set for the duration of any trial, and the subjects were free to explore the object as desired during the trial.

While completing the physical object blocks, in between trials the subjects were asked to sit back and relax, while the foam block was detached from the prehensor. Given the randomized sequence of the two objects within all blocks, the foam block was always removed from the prehensor aperture, even if sequential trials featuring the same object were queued. Once the object was replaced in the prehensor, the subjects received a verbal indication to proceed with the next trial. During the blocks utilizing virtual objects, the next trial (and the next object) were automatically queued up once the subject ended the current trial and moved the prehensor to a neutral, open position.

2.3.3 Subjects

A total of 8 subjects participated in Experiment 1, (5 females and 3 males) and every subject was right hand dominant. The Colorado School of Mines Human Subject Research review process deemed the experimental protocol exempt from IRB review, and informed consent was given by all participants.

2.3.4 Experiment 1 Results

As discussed above, the initial 10 trials of every block were excluded from analysis. Due to a discrepancy in the experimental protocol, the complete data set of Subject 7 (S7) was also excluded from analysis. The sensitivity indices of the subjects revealed a second anomaly in the dataset - uniquely among all subjects and conditions, Subject 5 (S5) demonstrated a large negative *d'* value when interacting with real objects at the K_D^- device stiffness setting. A negative *d'* value indicates that a subject was more likely to identify the soft object as a hard one, and vice versa. When contacted for a follow-up interview, S5 reveled a previously undisclosed medical issue, which produced a reduction in sensitivity of the anterior part of the shoulder used to actuate and receive feedback from the PE.

The data of the remaining six subjects was pooled, and analyzed to produce the pooled sensitivity indices presented in Figure 2.5.

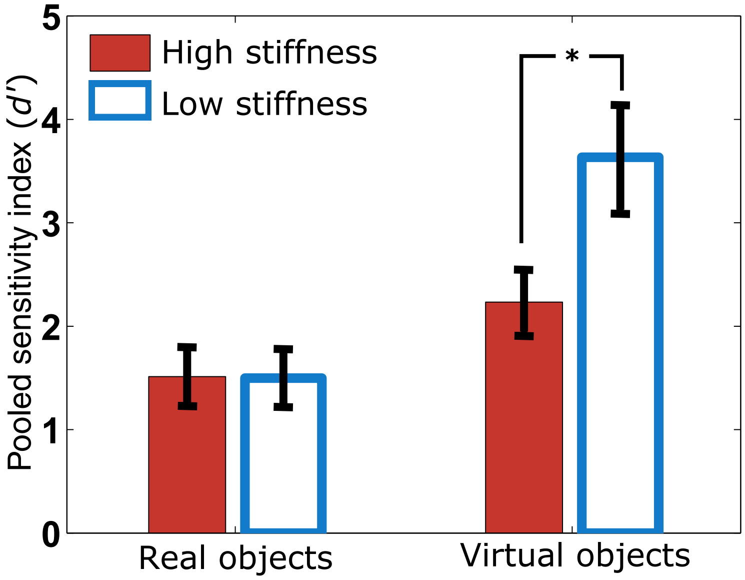

Figure 2.5: Pooled sensitivity indices (*d'*) for six subjects in Experiment 1, for the high and low prehensor stiffness condition, and for both real and virtual objects. Error bars indicate 95% confidence intervals.

Table 2.1 summarizes the average decision times for the six subjects in all conditions during the first experiment.

Table 2.1: Average and standard deviations of decision times in seconds for real and virtual objects, across six subjects

K	Real Obj.	Virtual Obj.
K_D^+	5.5 (±2.4)	4.6 (±2.7)
K_D^-	4.2 (±1.8)	3.2 (±1.9)

The decision time data were subjected to a two-way repeated measures analysis of variance, with the factors being the two device stiffness settings (K_D^- and K_D^+), and the two object types (real and virtual). The significance threshold for all statistical tests was set at $p < 0.05$. The main effect of device stiffness on decision time resulted in an F ratio of $F(1,6) = 8.39, p = 0.027$, indicating a statistically significant decrease in decision time when subjects were using a low stiffness device. The main effect of object type on decision times produced an F ratio of $F(1,6) = 5.33, p = 0.06$, indicating that the difference in decision times for real and virtual objects was not statistically significant. The interaction effect also failed to show statistical significance, $F(1,6) = 0.01, p = 0.91$. Including the data of S7 in the analysis results in a meaningful change only in the main effect of device stiffness, with the intact eight subject data set producing an F ratio of $F(1,7) = 4.34, p = 0.075$. As discussed above, objective considerations point to S7 as an anomaly in the data set. The inclusion or removal of S5 in the data set does not result in an alteration in the statistical significance of the decision time analysis.

The impact of device stiffness on decision time was confirmed as significant by one-way repeated measures analysis of variance applied to the reduced real and virtual object data sets individually, with $p = 0.027$ and $p = 0.031$ respectively. Similarly, when examined separately the subjects' decision time was not significantly affected by the type of object they were interacting with, real or virtual ($p = 0.19$ at the high stiffness setting, and $p = 0.058$ at the low device stiffness setting).

2.3.5 Experiment 1 Discussion

The subjects demonstrated improved performance in the object identification task when using a low-stiffness prehensor. This holds true for both the virtual objects and the physical sponges. In the case of virtual objects, the improvement in pooled sensitivity indices (*d'*) was statistically significant. While the real object pooled *d'* from the low and high stiffness conditions are virtually identical (see Figure 2.5), the decision time data (Table 2.1) shows that subjects were significantly faster in making the identifications when using a low stiffness prehensor. As mentioned above, if no significant difference is observed in the *d'* values between conditions, decision time becomes the secondary performance indicator [51]. Thus, both of the hypotheses for Experiment 1 have been confirmed - 1). a low stiffness prehensor does positively impact the performance of subjects in an object identification task and 2). this effect is observable with both real and virtual objects.

Besides evaluating performance, the *d'* measure can be used to draw additional conclusions about the tasks themselves. First, it is clear that the subjects had more difficulty identifying the real objects, with both pooled and individual sensitivities in this condition being consistently lower than for virtual objects. Most likely, this is due to the fact that the response of both the soft and the hard sponges becomes very non-linear as they are compressed close to the limit. Thus, while at the initial stages of the interaction the response of the two sponges is very distinct, as the subjects compressed them further, the force feedback in both scenarios increased rapidly as both sponges became essentially solid. On the other hand, discriminating between the linear response of the virtual objects appears to have been too easy for the subjects - at the low device stiffness condition, all subjects produced a perfect score in either the hit or the false alarm rate, indicating that the task was too simple [51]. In the virtual object condition, every subject demonstrated a higher sensitivity index when using the low stiffness prehensor. This was not entirely surprising, as the low stiffness of the device meant that a greater percentage of the force displayed back to the subject is contributed by the object. However, the high proportion of perfect scores is something that should be addressed through an increase in the difficulty of the task.

The discrepancy between the absolute *d'* values observed with real and virtual objects is not surprising. In fact, given the simplified nature of the virtual objects compared to the non-linear sponges, it would have been shocking if the subjects' response to both types of objects was the same. It must be emphasized that creating a perfect virtual replica of the physical sponges was never the aim of this work. As long as the impact of prosthesis stiffness on the overall performance of subjects is consistent in direction between the object types, then the simpler models can serve as useful tools in future experiments. Additionally, it is reasonable to expect that interaction with a different physical object, one with more linear spring characteristics and negligible damping, would produce results similar to the ones observed in the virtual object condition. Practical examples of such an interaction include distinguishing between plastic cups with different wall thicknesses, or squeezing a partially inflated bicycle tire to determine if it needs additional air.

At the individual subject level, the sensitivity results appeared to be highly varied. For example, when interacting with real objects, subjects 1, 2 and 4 performed noticeably better in the low stiffness condition, while subjects 3, 6 and 8 demonstrated better performance in the high stiffness condition. The order of presentation does not appear to have been a contributing factor to the split. The most likely explanation for this variability is the relatively small number of trials each subject completed - the subjects interacted with a given object only 15 times at each stiffness setting. At this number of trials, the individual *d'* values are very sensitive to random variation [51], and with more data collected, it is likely that most subjects would converge to a specific stiffness setting. In this work, the pooling approach was used to compensate for the limited data collected from individual subjects. While it is possible that individual preference plays some role in determining the "optimal" stiffness setting for this task, the current data set is insufficient to answer this question - future work will address this possibility.

Anecdotally, a large majority of the study participants expressed a preference for the lower prehensor stiffness condition, without being aware of their relative performance. This indicates that prosthesis users may be able to optimize the stiffness settings of their devices to match the requirements of a particular task, given the mechanical capability and an intuitive interface.

In summary, the first experiment provided support for the hypothesis that a lower prehensor stiffness results in improved performance when attempting to distinguish between objects based on their stiffness. However, the broader nature of this effect was not yet well understood. Is the effect of prehensor stiffness on a subject's sensitivity index monotonic? Is it linear? Will this effect hold as subjects are asked to distinguish between more objects?

Building off the initial results provided by Experiment 1, Experiment 2 was intended to further explore and detail the apparent trends in subject performance due to variations in prehensor stiffness.

2.4 Experiment 2: Four Virtual Objects

2.4.1 Prosthesis Emulator Configuration

While the experimental system from Experiment 1 remained largely intact, several minor adjustments were made to address issues that were highlighted by the pilot study. In spite of the preventative measures implemented, during Experiment 1 the motor used to simulate the prehensor spring and the virtual objects was occasionally saturated. This intermittent issue occurred only when simulating the hard virtual object at the high prehensor stiffness condition, producing an awkward and jarring sensation for the subjects. To address this problem, the maximum torques required were reduced, by restricting the travel arc of the prehensor to 50°, limiting the displacement of the virtual torsional spring at the prehensor shaft. In this configuration, the contact threshold for virtual objects was set at 20°.

As mentioned in the Experiment 1 Discussion section, the initial virtual object stiffness discrimination task appeared to be too easy for the subjects. For the follow up experiment, the task difficulty was raised by expanding the experimental objects set from two to four.

The four new virtual objects were loosely grouped into "soft" and "hard" categories, with two objects in each. The torsional stiffness coefficients of the four objects, in order of increasing stiffness, were as follows: O_1 = 4.3 N-mm/°, O_2 = 8.6 N-mm/°, O_3 = 17.2 N-mm/°, O_4 = 21.6 N-mm/°. These changes produced a classification experimental paradigm, a more difficult scenario compared to a discrimination task [51].

In order to compensate for the lower torques produced by the restricted displacement of the prehensor, and raise the levels of force feedback displayed to the users, the lever arm of the cable, l was shortened to 18.7 mm. The device stiffness set K_D was expanded to include three evenly spaced stiffness levels, $K_D = [K_D^1\ K_D^2\ K_D^3]$, with K_D^1 = 3.7 N-mm/°, K_D^2 = 5.6 N-mm/°, and K_D^3 = 7.5 N-mm/°. Thus, Eq. 3 above remained the same, while Eq. 4 became

$$K_O = \begin{cases} O_1, O_2, O_3, O_4 \ if \ r \geq t \\ 0 \ if \ r < t \end{cases} \tag{2.8}$$

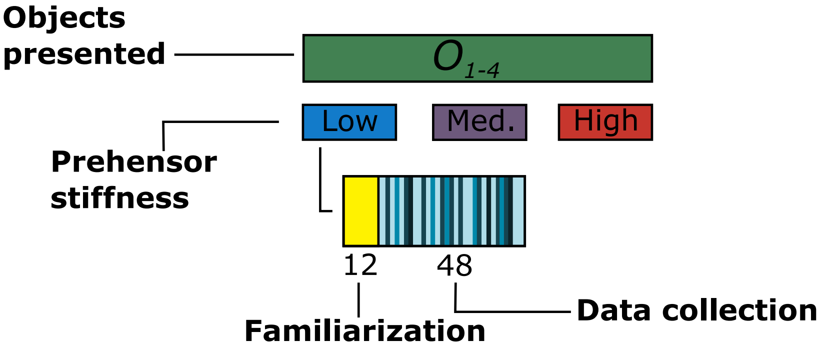

Figure 2.6: The order of presentation of various experimental conditions during Experiment 2. Each subject completed one session, composed of three 60 trial blocks. The order of presentation of the three prehensor stiffness conditions was randomized and counterbalanced between 12 subjects. The first 12 trials of each block formed part of the familiarization procedure. During each trial, a single object, out of a sample of four, was randomly selected and presented. Each object was presented a total of 15 times during each of the three blocks.

2.4.2 Procedures

During this experiment, each subject participated in one data collection session, composed of three trial blocks. Every block focused on a specific prehensor stiffness condition, low, medium, or high. The order of presentation of the three conditions was randomized and balanced between subjects, in order to minimize the effects of learning on the results. In each block, the corresponding value of K_D was set as the prehensor torsional spring stiffness. During each block, the subjects completed 60 randomized trials, evenly split between the four experimental objects. As in Experiment 1, unknown to the participants, the initial 12 trials of all blocks formed a hidden training section within the data, which was individually randomized and balanced (i.e. this training section included 3 trials for each experimental object). In addition, each trial block was preceded by a familiarization protocol. A diagram summarizing the procedure is presented in Figure 2.6. During familiarization, the subjects explored each one of the four experimental objects at will, cycling through the entire set at least twice. Experimentation commenced only when subjects indicated they were comfortable with the distinctions between the experimental objects.

The trial protocol was practically identical to the virtual object trials of Experiment 1. In each trial a single object was simulated in the prehensor aperture of the PE system. The subject closed the prehensor through a forward displacement of the contralateral shoulder. The simulated prehensor spring provided some initial resistance to the movement. Once the prehensor crossed the 20° contact threshold, the spring-like response of the virtual object began to increase the effective torque on the prehensor shaft, raising the magnitude of the force feedback displayed to the subject's shoulder. Once the subject classified the response as belonging to one of the established experimental objects by pressing the associated keyboard key, the trial ended. Similar to Experiment 1, the subject's response was recorded and compared to the actual object being simulated. Decision time was once again tracked, and defined as the period of time between the prehensor crossing the contact threshold and the end of the trial. To prevent physical and mental fatigue, subjects took a 3-5 minute break after every block.

2.4.3 Subjects

A total of 12 subjects participated in Experiment 2, 2 females and 10 males. All subjects, except for one, were right hand dominant. The Colorado School of Mines Human Subject Research review process deemed the experimental protocol exempt from IRB review, and informed consent was given by all participants.

2.4.4 Experiment 2 Results

Data from the first 12 (training) trials in every block for all subjects were excluded from further analysis, as discussed previously.

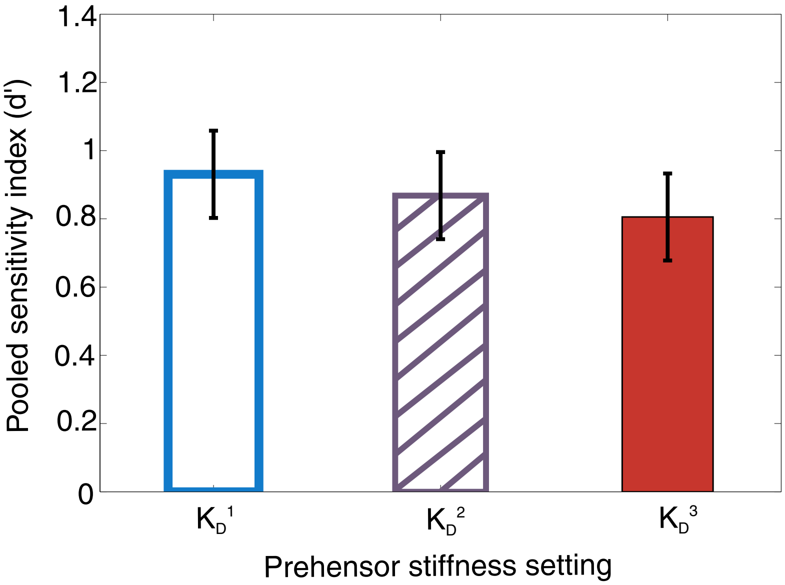

Figure 2.7: Pooled sensitivity indices (d') for Experiment 2 subjects at three prehensor stiffness conditions. Error bars indicate 95% confidence intervals.

The combined response matrices of all 12 subjects at each of the experimental conditions are presented in Table 2.2, with O_{1-4} representing the four objects in order of increasing stiffness, and R_{1-4} standing for the possible subject responses. Thus, the diagonals of each of the three matrices count the pooled number of correct responses given by all subjects at that condition. A total of 576 responses are recorded in each matrix.

Table 2.2: Pooled responses of 12 subjects in a 4-object classification task. O_1 to O_4: objects in order of increasing stiffness. R_1 to R_4: subject responses.

Low Prehensor Stiffness (K_D^1)				
	O_1	O_2	O_3	O_4
R_1	84	40	17	3
R_2	34	77	23	10
R_3	9	27	73	35
R_4	8	17	48	71

Medium Prehensor Stiffness (K_D^2)				
	O_1	O_2	O_3	O_4
R_1	72	59	12	1
R_2	29	76	32	7
R_3	6	27	72	39
R_4	10	14	48	72

High Prehensor Stiffness (K_D^3)				
	O_1	O_2	O_3	O_4
R_1	70	56	14	4
R_2	21	72	37	14
R_3	5	21	67	51
R_4	5	21	45	73

Table 2.3: Pair-wise and overall *d'* sensitivity indices of 12 subjects in a 4-object classification task

Prehensor Stiffness	d'(O_{1-2})	d'(O_{2-3})	d'(O_{3-4})
Low (K_D^1)	0.93	1.42	0.68
Medium (K_D^2)	0.84	1.35	0.61
High (K_D^3)	1.02	1.29	0.39

To obtain a single overall sensitivity index from the data sets for each experimental condition, techniques specific to M-alternative experiments were applied [55]. The results of the analysis are presented in Figure 2.7. Though no statistically significant differences in the subjects' sensitivity indices between conditions is evident, a slight linear trend toward better performance at lower prehensor stiffness is observed.

In addition, a classification experiment provides an opportunity to calculate individual sensitivity indices for each successive pair of objects. Based on the data in Table 2.2, it is possible to extract the pooled sensitivity of the subjects in differentiating between O_1 and O_2, between O_2 and O_3, and between O_2 and O_3. This is done by transforming the response counts into hit and false alarm rates, followed by applying Eq. 4.8 to successive pairs of rows. The results of this analysis are gathered in Table 2.3.

2.4.5 Experiment 2 Discussion

The data gathered during the four object classification task support the initial hypothesis: a lower prehensor stiffness is advantageous when attempting to discriminate between, or classify different objects. This is supported by the pooled *d'* indices reported in Figure 2.7. At the lowest prehensor stiffness setting, subjects demonstrated the highest sensitivity index, corresponding with better task performance. As prehensor stiffness increased, the *d'* index decreased linearly, with the highest stiffness setting producing the worst performance. The effect of stiffness on performance also appears to be linear.

However, the differences in sensitivity between conditions are not statistically significant. Compared with the virtual object condition results of Experiment 1, the magnitude of the effect of prehensor stiffness on sensitivity has dropped markedly. As an additional contrast to Experiment 1, no statistically significant differences in decision time was found between the three experimental conditions. The most likely explanation is that the transition to a four-alternative forced choice (4AFC) experiment design made the task too difficult for the participants, especially in the absence of correct-incorrect feedback. This forced the subjects to rely on their memory of the sensations caused by each stimuli. In the future, in order to modulate the difficulty of the task, the relative magnitudes of the stimuli should be modulated - i.e. brought closer together to increase task difficulty. Correct-incorrect feedback should also be provided, to help ensure that the participants are truly reaching their maximum potential sensitivity performance.

Turning to the pairwise sensitivity indices (Table 2.3), a few additional logical patterns emerge. Notice that at all prehensor stiffness settings, the highest pair-wise sensitivity was observed between O_2 and O_3. Recall that the four virtual objects were separated into two groups, with O_1 and O_2 making up the "soft" group, and O_3 and O_4 comprising the "hard" group. The stiffness gap between the groups was twice as large as the one between the group members, leading directly to the high between-group *d'* indices produced by the subjects.

For all three experimental conditions, the lowest pair-wise sensitivity was observed when subjects attempted to discriminate between the two "hard" objects. Recall that the stiffness offset between O_1 and O_2 was the same as O_3 and O_4. However, at higher object stiffnesses, the same change in absolute stiffness results in a progressively smaller percentage change. This effect likely accounts for the low sensitivities observed for these pairs, and appears to have the largest impact on the high stiffness prehensor system, with users demonstrating the lowest pair-wise *d'* of any condition when attempting to discriminate between the "hard" objects. The higher the overall stiffness of the prehensor-object system, the lower the percentage change in the total force feedback displayed to the user in response to a change in the stiffness of the object.

While most of the patterns observed in the pair-wise sensitivity indices support the initial hypothesis, the sensitivities displayed by the subjects for the O_{1-2} object pair bucks the trend. Here, the high prehensor system led to the best performance, followed by the low stiffness system, with the medium stiffness condition bringing up the rear. Given the consistent trend toward higher sensitivities at lower stiffness for the other object pairs, it is possible that this anomaly is simply the result of experimental noise.

A possible modification to the experimental protocol is providing correct answer feedback to the subjects after each trial. While this change in unlikely to meaningfully challenge the results obtained in the current study [51], such feedback will be incorporated in future studies. It is also possible that obtaining more data from each subject would affect the final outcomes. While more data is always preferable, practical limitations, such as user fatigue, prevent greatly extending the individual data collection sessions. In addition, this risk is somewhat mitigated through pooling the responses into a unified data set. In the current study, the subjects were limited to performing the given task at a prehensor stiffness setting set by the experimenter. Thus, at the current stage of experimentation, I can infer little about the subject's willingness to independently modulate the stiffness of the prosthesis emulator to maximize task performance if given that opportunity. To be sure, unless the user perceived value of the stiffness control modality exceeds the inconvenience and additional mental strain involved in modulating it, the users will be hesitant to integrate it into their prosthesis control scheme. Further studies are necessary to explore this issue in detail.

All but one of the subjects in this study were right handed. It may be possible that by asking right-handed subjects to use their non-dominant shoulder to interact with the PE, their performance was somewhat lower than it may have been otherwise. However, given that task performance did not depend strongly on the dexterity of the actuating limb such effects are likely to be negligible. Finally, the repeated-measures experimental design in combination with the counterbalanced order of presentation of the experimental conditions ensured that any advantages or disadvantages stemming from subject handedness were present in equal measure in all experimental conditions.

It is possible to look at the data collected in this study and conclude that all body-powered prosthesis prehensors should be as soft as possible, since that setting maximized performance in the experimental task, and lower stiffness would reduce the stress placed on the user's actuating joint during operation. However that inference would be misleading, as fixing the stiffness at a minimum point assumes that it is the optimal setting for all tasks, and robs the user of true stiffness modulation. Performance in other tasks may be maximized at different stiffness conditions, and the study presented in Chapter 3 explored this possibility. Specifically, the study to quantified the performance of subjects using a high stiffness prosthesis in a disturbance detection task.

2.5 Contributions

This study contributed to the understanding of the role of prehensor stiffness in PE user performance in the following ways:

- **Prehensor stiffness has a quantifiable and significant impact on the haptic feedback provided by a body-powered prosthesis**

  One of the overarching questions targeted by this work was whether the impact of prehensor stiffness on the performance of users during feedback dependent tasks was significant. Following the conclusion of this study, it is clear that prehensor stiffness does have a measurable and significant influence on the performance of users in an object identification task. While this is not sufficient to declare that prehensor stiffness modulation is necessary to optimize performance, it does encourage a deeper examination of this control modality.

- **The effect of prehensor stiffness on haptic feedback quality appears to be monotonic**

  Based on the results of Experiment 2 (section 2.4), it looks as though prehensor stiffness impacts user performance in the object identification task monotonically. However, given that no statistically significant difference between the three stiffness conditions was found, this requires additional investigation to confirm.

- **Virtual objects can serve as viable stand-ins for physical ones, for the purposes of evaluating the effect of prehensor stiffness**

  In the course of this study, the direction of the trends in user performance as a function of prehensor stiffness were consistent between real and virtual objects. Thus, the impact of prehensor stiffness on user performance in real-world scenarios may be reasonably inferred from the subject's performance during experiments using virtual stimuli.

CHAPTER 3

TASK DEPENDENT EFFECTS OF PREHENSOR STIFFNESS ON FEEDBACK IN BODY-POWERED PROSTHESES

The low quality of haptic feedback is a commonly mentioned deficiency in the current generation of prosthetic devices. Stiffness modulation, a control modality largely absent from prostheses, may have the potential to improve the quality of the haptic feedback provided to the user. To date, prosthesis users have been forced to use a single stiffness setting, regardless of the task they engaged in. In this chapter[2] , I report the results of two parallel experiments, in which able-bodied subjects used a body-powered prosthesis emulator to complete two feedback-dependent tasks. Experiment 1 focused on object identification based on apparent stiffness, while Experiment 2 dealt with detection of disturbances in the motion of the prehensor. In each experiment, the subjects completed tasks of varying difficulty with the prosthesis emulator operating in a high and a low prehensor stiffness condition. Analysis of the performance trends within the two subject groups suggests that a low stiffness prehensor improves subject performance in the object identification task, while a high stiffness prehensor is more suited to the disturbance detection task. This prompts the conclusion that introducing prehensor stiffness modulation to body-powered prosthesis may improve the quality of haptic feedback provided to the user and increase prosthesis user satisfaction.

3.1 Introduction

The central goal of this work is to quantify the potential impact of prehensor stiffness modulation to improve task performance of body-powered prosthesis users in feedback-dependent tasks. Data collected during a previous two part study described in Chapter

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[2] The results presented here have been submitted for publication to the journal *IEEE Transactions on Haptics* and are currently undergoing peer review

2 revealed that prehensor stiffness has a quantifiable effect on performance in object identification and classification tasks, with lower stiffness producing better performance.

However, this previous work was subject to several limitations. First and foremost, it relied on a limited pool of subjects. In addition, the use of a 4-alternative, forced choice classification paradigm proved challenging for the subjects, and produced several inconclusive results. Finally, the initial study focused on a single feedback-dependent task. Though it established that prehensor stiffness can impact task performance, this is not yet sufficient motivation for the implementation of prehensor stiffness modulation, as it is not yet clear if a high prehensor stiffness could be advantageous in a different task.

The current work directly improves on, and extends the previous study by:

1. Evaluating the influence of prehensor stiffness on user performance in two feedback dependent tasks - object identification and disturbance detection.
2. Using a parallel experimental design for both tasks, directly improving on the previous experimental protocol by providing the users with visual feedback regarding their performance.
3. Expanding the subject pool, with more data collected from individual subjects.
4. Directly exploring the interaction between task difficulty and the effect of prehensor stiffness on performance.

The potential impact of prehensor stiffness on performance in feedback-dependent tasks is mandated by the purely mechanical actuation system employed in body-powered prostheses. During interaction with an object, the force feedback that a body powered prosthesis displays to a user is proportional to the sum of K_D, the device (prehensor) stiffness and O, the object stiffness, i.e.

$$F \propto K_D + O \tag{3.1}$$

Thus, if

$$K_D > O \tag{3.2}$$

the majority of the feedback force, F, is dedicated to providing the user feedback about the internal workings of the prehensor rather than to informing the user about the response of the object.

Similarly, if the prosthesis prehensor encounters an unexpected displacement disturbance during closing, the change in the feedback force is broadly governed by the following relationship:

$$\Delta F \propto K_D * D \tag{3.3}$$

where ΔF is the change in feedback force, K_D is the prehensor stiffness, and D is the magnitude of the displacement disturbance in degrees.

I therefore hypothesize that a condition in which K_D is minimized will result in improved performance in an object stiffness discrimination task, and that this effect will be present regardless of the difficulty of the task. Additionally, I hypothesize that maximizing K_D will improve the performance of a prosthesis user in a disturbance detection task, and that this effect will be independent of the difficulty of the task.

This paper presents the results of two parallel human subject experiments, designed to test these hypotheses. The Experiment 1 focused on an object identification task, while the Experiment 2 asked the participants to complete a disturbance detection task. For both experiments, the subjects completed the tasks using a prosthesis emulator (PE) system (shown in Figure 2.1, see [50] for details on the original version of the device, which was developed as a teleoperation test bed). Both tasks were completed with both a low and a high prehensor stiffness.

The results were evaluated through the application of signal detection theory (SDT) analytical framework, [51, 52] which has previously been shown to be viable in experiments involving virtual object discrimination [53].

The remainder of this chapter is organized as follows: Section 3.2 details the experimental design, tasks, and procedures employed throughout this study. Section 3.3 presents all results obtained in the course of this work, with a discussion of the data provided in Section 3.4. Section 3.5 contains a brief summation of the contributions of this specific study to the overall goal of the work.

3.2 Methods

3.2.1 Setup

As described previously in Chapter 2, the PE system consists of a one degree of freedom (DOF) prosthesis prehensor, attached to a 3D printed frame. During operation, the PE is held in a horizontal orientation in a vise on a table. The subjects sit in a chair to the left of the PE, with their right hand grasping a rigid handle, and their right elbow resting on a nearby support (see Figure 2.1). The single DOF prehensor is actuated through a mechanical cable, which links the prehensor with the user's left shoulder. By moving their shoulder forward, the users generate tension within the cable, which in turn generates a torque around the prehensor shaft, closing the prehensor (see Figure 2.2). If the prehensor encounters an object or a position disturbance while closing, a change in the tension within the cable is produced. The users perceive this interaction as a change in the force feedback displayed to their shoulder through the harness. Different objects and disturbances produce varying levels of adjustment in the force feedback, allowing the user to differentiate between these stimuli based on their intensity.

This feedback/actuation mechanism mirrors that of a standard body-powered prosthesis operated in voluntary closing (VC) mode. This type of prosthesis is unique in providing the user with proportional force feedback about the interaction of the prosthesis and the environment, and is often used in challenging or delicate tasks.

In addition to the actuating cable, the prehensor shaft is also connected to a Maxon RE 30 DC (24V, 3.81A) motor through cable drive with a 12:1 gear ratio (see Figure 2.2). This motor is used to simulate the effect of a linear torsional spring acting around the prehensor shaft. The virtual spring acts to keep the prehensor open, and provides resistance to the closing motion initiated by the user. The stiffness of this virtual spring, K_D, can be freely adjusted within the physical limits imposed by frictional forces and motor characteristics. The experiments in this study were conducted at two discreet prehensor stiffness settings, low (K_D^-) and high (K_D^+). For all experiments, $K_D^- = 3.7$ N-mm/° and $K_D^+ = 7.5$ N-mm/° (note that all torque measurements and effective stiffness constants are reported at the prehensor shaft).

During all experiments conducted as part of this work, the lever arm between the prehensor shaft and the cable was set to 18.7 mm, and the range of motion for the prehensor was set to 50°. An encoder attached to the motor kept track of the angular displacement of the prehensor. Visual feedback of the prehensor displacement was blocked via a screen. Additionally, during all experiments, subjects wore noise-canceling headphones, which played pink noise. A keyboard was placed within the subject's reach, in order to record their responses during the experiment.

3.2.2 Subjects

Each experiment had a subject group of ten able-bodied individuals. The ten subjects in Experiment 1 (S1-S10, 4 female, 6 males) completed two experimental sessions focused on object identification. The subjects in Experiment 2 (S11-S20, 3 female, 7 males) completed two sessions of the disturbance detection experiment. There was no overlap between the two subject pools. Informed consent was given by all participants, and the procedure was deemed exempt from IRB review by the Human Subject Research group of the Colorado School of Mines.

3.2.3 Procedures

Both groups of study participants completed a pair of experimental sessions, held on two separate days. During one of the sessions, the participants completed their assigned task with the PE system set to the low prehensor stiffness setting (K_D^-). During the other session, the subjects completed the same task, but with the PE system operating at the high stiffness setting (K_D^+). This created a repeated-measures design, with prehensor stiffness acting as the independent variable.

The order of sessions was randomized and counterbalanced within the subject groups. The experimental design of the sessions was independent of the experimental task, both of which employed a two alternative, forced choice (2AFC) paradigm. Every session consisted of a 4 blocks of 50 randomized and balanced trials. During each block, the subjects were asked to use the PE to discriminate between a pair of stimuli – object stiffness for Experiment 1, disturbances for Experiment 2. Every block was preceded by a brief familiarization sessions, lasting 12 trials.

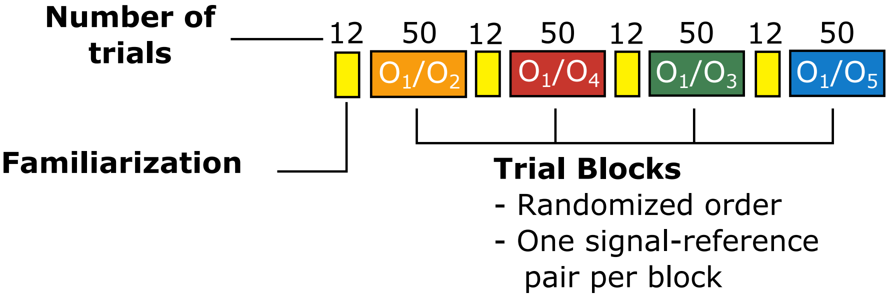


Figure 3.1: A diagram of the experimental procedure for one session of one subject, completed using either a low (K_D^-) or a high (K_D^+) prehensor stiffness. Each session contained four trial blocks, presented in randomized order, and preceded by brief familiarization sessions. The procedure was consistent between the object identification and disturbance detection tasks.

3.2.4 Experiment 1 - Object Identification

During the object identification sessions, participants were tasked with using the PE to interact with a pair of virtual objects, and correctly identify them based on their apparent stiffness. The objects were modeled as linear virtual springs, simulated by modulating the torque output of the prehensor motor. A total of five objects, with distinct stiffness coefficients, O_{1-5} were created. In increasing order, $O_1 = 6.5$ N-mm/°, $O_2 = 8.6$ N-mm/°, $O_3 = 10.8$ N-mm/°, $O_4 = 13.0$ N-mm/°, and $O_5 = 15.1$ N-mm/°. These five objects were organized into four pairs, by using O_1, the softest object, and coupling it with each of the remaining four objects (O_1 - O_2, O_1 - O_3, etc.). For all object pairs, O_1 served as the reference, while the higher stiffness object performed the role of the signal. One such pair was used as the reference-signal pair in each of the 4 blocks that made up every session. The order in which the object pairs were presented to the subjects was randomized.

As the subjects displaced the prehensor from its resting position, they initially felt only the force feedback resulting from the prehensor stiffness, K_D, acting against them. However, once the prehensor passed a predefined 20° contact threshold, the effects of the object spring (O_{1-5}) also started to contribute to the force in the actuating cable. See Figure 2.3 for a diagram of the PE system during the object identification task. As per Eq. 4.2, the force feedback displayed to the subjects was proportional to the sum of the prehensor and object stiffness coefficient. More specifically, the total force, F, displayed to the subject during any individual trial was

$$F = l(\theta K_D + (\theta - \theta_t)O) \tag{3.4}$$

where l is the lever arm of the cable, K_D is the device stiffness, θ is the displacement of the prehensor from its resting position in degrees, θ_t is the contact threshold, and

$$O = \begin{cases} O_{1-5} \ if \ \theta \geq \theta_t \\ 0 \ if \ \theta < \theta_t \end{cases} \tag{3.5}$$

Based on these relative differences in force feedback, during a specific trial the participants attempted to identify the virtual object they were interacting with as either the reference (O_1) or the signal (O_{2-5}). The participants were not limited in the number of times they could interact with the object during an individual trial, and many trials involved 5 or more discrete interactions. Note that as per the 2AFC paradigm, the subjects were always making a binary choice. Once they had made their decision, the subjects recorded their responses by pressing the keyboard key associated with either the reference or the signal object. In a change from the previous work, during all sessions, correct answer feedback was provided to the users immediately after their response was recorded. The correct answer feedback was displayed on a computer screen placed in front of the subject in numeric format, with a **1** indicating a correct identification, and a **-1** indicating an error.

A minor methodological error caused some of the experimental trial sets used during the object identification sessions to become slightly unbalanced. With 50 trials per block, a balanced trial set would include 25 reference trials and 25 signal trials, presented in randomized order. On average, the blocks featured only 1.0±1.1 unbalanced trials, and the impact of this error on the results was negligible.

3.2.5 Experiment 2 - Disturbance Detection

The procedures used during the disturbance detection experiments mirrored those employed during object identification trials, with the notable change of replacing the four virtual object pairs with four pairs of virtual displacement disturbances. Just as with the object reference-signal pairs, a total of five displacement disturbances of the prehensor were defined, labeled as D_{1-5} in order of increasing magnitude.

A diagram of the PE system during the disturbance detection task is presented in Figure 3.2. The disturbances were implemented as a displacement offset, added on top of the sensed prehensor position, θ that persisted for 2°. Specifically, as the subjects actuated the PE system, once the prehensor reached 25° of angular displacement, the value of the specified disturbance D_{1-5} was added to θ, and this change was maintained for 2°, after which it was removed. Thus,

$$F = l[(\theta + D)K_D] \tag{3.6}$$

where l is the lever arm of the cable, K_D is the device stiffness, θ is the angular displacement of the prehensor from its resting position, and

$$D = \begin{cases} D_{1-5} \ if \ 25^\circ \leq \theta \leq 27^\circ \\ 0 \ otherwise \end{cases} \tag{3.7}$$

is the disturbance applied as an offset in displacement, which is perceived by the subject as a pulse in the force feedback provided by the PE to their shoulder. This interaction resembles the scenario of an external impact/bump on a grasped object.

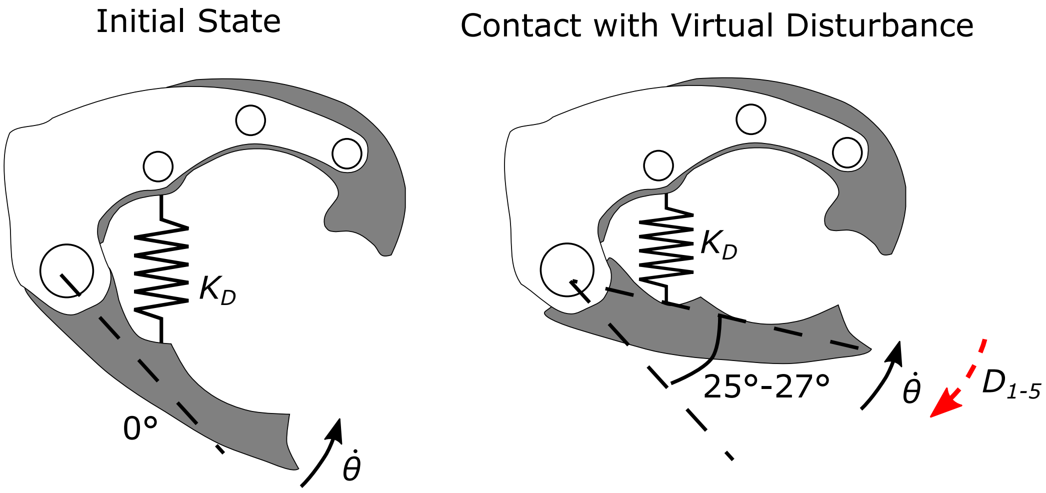


Figure 3.2: A diagram of the PE during the disturbance detection task. Once the prehensor crosses the disturbance threshold of 25°, a virtual disturbance D_{1-5}, is applied to its position, θ. The disturbance is implemented as a 2° offset added to the sensed prehensor displacement. The offset persists until the prehensor moves past 27° of displacement. This process provides a sensation of a brief but distinct disturbance in the motion of the prehensor.

Here, $D_1 = 70°$, $D_2 = 75°$, $D_3 = 80°$, $D_4 = 85°$, and $D_5 = 90°$. These magnitudes may seem unrealistic considering that the physical displacement of the prehensor was limited to only 50°. It must be emphasized that the virtual disturbance magnitudes represent only the commanded displacement offset of the prehensor. Due to the limited torque output of the PE motor, the actual displacement of the prehensor during the disturbance was far lower, as the actuating force provided by the subject's shoulder joint overpowered the additional torque generated by the brief disturbance. To clarify, as the motor attempted to displace the prehensor, it was constantly opposed by the forward motion of the subject's shoulder. Given the relatively high strength of the joint, the motor was unable to force the subject's shoulder back far enough to allow the prehensor to open by the commanded amount - and indeed applying forces strong enough to accomplish this to the subject would have unreasonably increased the risk of injury. The prehensor's position was affected by the disturbance, which produced a brief jump or stutter in the closing motion of the prehensor. While the prehensor was not physically displaced by the nominal magnitude of the simulated disturbance, the PE motor generated the same torque as if it had been. Thus, the pulse in the force feedback displayed to the participants was equivalent to one that would have been generated by a physical displacement of the prehensor (see Eq. 4.5).

The final magnitudes of the virtual disturbances were selected following pilot testing. The smallest disturbance, D_1, was used as the reference, and coupled with the remaining four disturbances (D_{2-5}) to form four reference-signal pairs. One pair was used during each of the 4 trial blocks that made up the two data collection sessions. The order of the blocks was randomized for each subject, but not counterbalanced between subjects.

Once again, the 2AFC protocol was followed. During each trial, one of two disturbances was displayed to the subjects, and they had to correctly identify them based on the relative differences in the force feedback provided to them. Correct answer feedback was provided.

The disturbances were displayed to the users only when the prehensor was in the process of closing. Dynamic stability in the transition phase was obtained by establishing a minimum $\dot{\theta}$, below which the disturbances were not displayed. None of the subjects had an issue with maintaining the required velocity of the prehensor. In fact, most of the participants did not notice this velocity "floor", as their preferred $\dot{\theta}$ was above the threshold.

3.2.6 Data Analysis

All of the data collected during the experiments was analyzed using the Signal Detection Theory framework (SDT). The application of SDT techniques to the responses produced by the particpants of the 2AFC experiments allows the calculation of a sensitivity index, *d'*. This index is calculated as follows:

$$d' = z\{H\} - z\{F\} \tag{3.8}$$

where H represents the fraction of correctly identified reference stimuli (i.e. O_1 or D_1), F is the fraction of false alarms (i.e. how often the subject mistook a signal stimulus for the reference), and the $z\{\}$ operator represents a z-transform, which transforms the ratios into a z-score. Note that unlike a simple error rate statistic, *d'* is a non-biased measure, allowing for meaningful comparisons between subjects and conditions. When the number of data points collected from individual subjects is relatively low, the sensitivity index is subject to significant random variation. In order to compensate for this, the responses of several subjects may be pooled to produce an estimate of the group sensitivity when identifying the stimuli [51]. The process of establishing confidence intervals on the pooled and individual sensitivity indices has been described in detail in Section 2.2.2.

3.3 Results

The averaged *d'* scores of all 10 subjects that completed the object identification experimental task are presented in Figure 3.3. The performance of the 10 subjects in the disturbance detection experiment are presented in Figure 3.3b. As a reminder, a repeated-measures paradigm was used for each of the experimental tasks. From the figure, it is clear that a low prehensor stiffness is advantageous for the object identification task, while high prehensor stiffness improves performance in the disturbance detection task. The effect is observed in all reference-signal pairs used in the experiment, and reaches statistical significance for two pairs in each task.

Table 3.1 presents the individual sensitivity indices of all subjects across all tasks, sessions and blocks. It is evident that while the aggregate data contains clear trends, at the individual subject level, significant variability is present. The most likely cause is the relatively low number of data points for each subject. However, given the random, but not balanced order in which the four blocks of reference-signal pairs were presented to the subjects, I considered the possibility that learning effects biased the data. In an effort to quantify the possible influence of learning effects on the subject's *d'* scores, the performance of each subject was analyzed relative to the group average, as a function of the block order. The results of this analysis are presented in Figure 3.4. In this figure, the values on the y axis indicate the difference between a single subject's d' in a specific block and the pooled d' for all subjects and blocks. A unique marker type represents a single subject. Therefore, a monotonic trend for a single subject across blocks would indicate existence of learning effects.

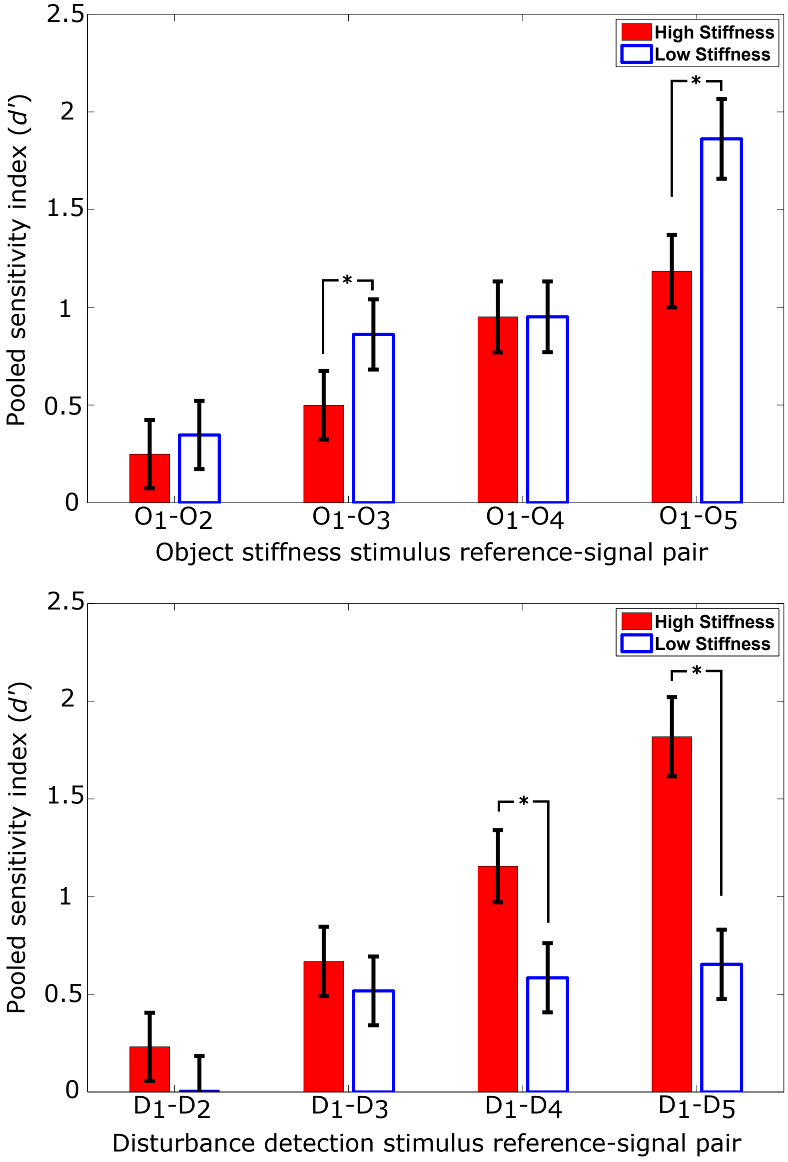


Figure 3.3: The pooled *d'* scores of subjects completing a) the object identification task and b) the disturbance detection task, as a function of the prehensor stiffness. Due to the repeated measures design, the performance of the same 10 subjects is evaluated in each paired bar graph. In both tasks, higher values of *d'* indicate a better ability to distinguish between the reference stimulus and the signal. Error bars indicate 95% confidence intervals, and statistically significant differences are marked with the * symbol

Table 3.1: Individual d' Scores of all subjects in Experiment 1 and Experiment 2

Experiment 1 - Object Identification

Prehensor Stiffness	**Stimuli Pair**	**S1**	**S2**	**S3**	**S4**	**S5**	**S6**	**S7**	**S8**	**S9**	**S10**
High	O_{1-2}	0.734	-0.464	0.224	1.147	0.194	-0.102	0.201	0.101	0.649	-0.097
	O_{1-3}	0.968	0.097	0.299	-0.008	0.721	0.415	0.304	0.415	0.869	1.046
	O_{1-4}	1.669	0.909	0.578	0.607	0.721	0.398	0.625	1.309	1.145	2.227
	O_{1-5}	1.548	1.771	1.594	1.411	1.470	0.180	0.811	1.533	0.710	1.653
Low	O_{1-2}	0.518	-0.008	0.821	0.050	-0.301	0.826	-0.619	0.533	0.869	1.006
	O_{1-3}	1.151	0.415	0.909	0.201	0.398	0.842	0.180	1.174	1.418	2.809
	O_{1-4}	1.546	0.385	1.070	0.495	0.000	0.398	1.177	0.721	2.457	2.773
	O_{1-5}	2.915	1.577	0.743	1.051	3.156	4.107	0.607	1.983	2.247	4.104

Experiment 2 - Disturbance Detection

Prehensor Stiffness	**Stimuli Pair**	**S11**	**S12**	**S13**	**S14**	**S15**	**S16**	**S17**	**S18**	**S19**	**S20**
High	D_{1-2}	0.207	0.102	0.000	0.000	0.000	0.201	0.409	0.533	0.109	0.826
	D_{1-3}	-0.101	0.101	0.756	1.065	1.095	0.409	0.941	0.941	1.095	0.612
	D_{1-4}	1.051	2.580	1.065	1.836	0.960	0.721	1.701	1.095	-0.105	1.424
	D_{1-5}	0.404	2.926	2.745	3.229	1.045	2.017	3.804	2.810	0.418	2.169
Low	D_{1-2}	-0.404	-0.317	-0.612	0.101	-0.612	0.101	0.404	0.304	0.404	0.791
	D_{1-3}	1.051	0.509	0.201	0.518	0.721	0.201	0.836	0.404	0.207	0.612
	D_{1-4}	0.201	0.518	0.656	0.826	0.612	0.555	0.409	0.721	0.418	1.095
	D_{1-5}	0.404	0.214	0.308	1.065	1.309	0.721	1.326	0.000	1.248	0.418

Overall, no clear trends are evident in Figure 3.4, indicating that learning effects did not substantially affect the individual sensitivity indices.

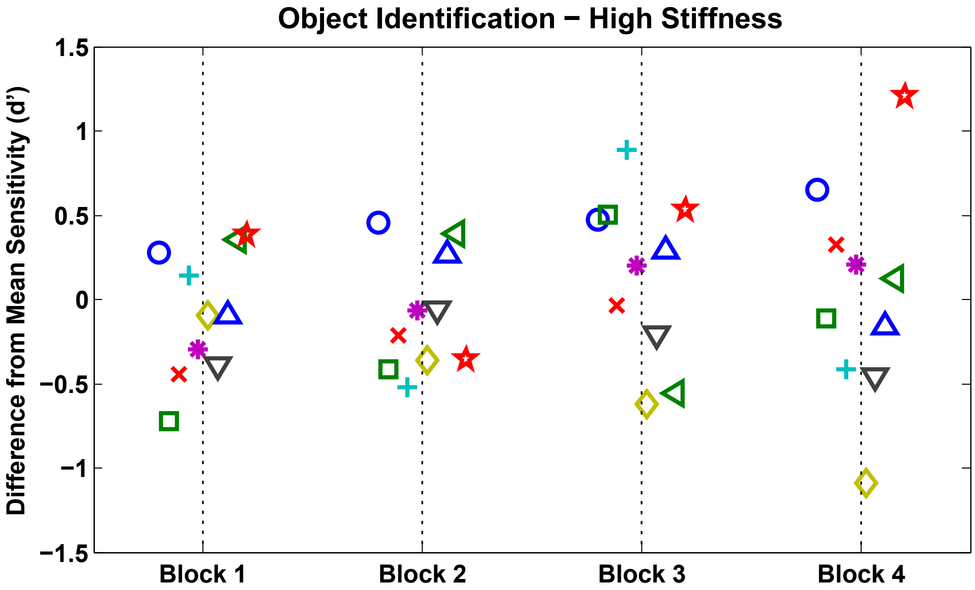


Figure 3.4: The *d'* sensitivity indices demonstrated by individual subjects, relative to the group average at the same condition, grouped by block number. Positive values indicate subjects outperforming the group average, while negative values are indicative of subjects under performing. A clear positive slope across the four experimental blocks would be indicative of a subject's performance improving as a result of learning.

3.4 Discussion

3.4.1 Object Identification

Prior to the study, I had hypothesized that subjects would demonstrate higher sensitivity in the object identification task when using a low stiffness prehensor. sensitivity indices presented in Figure 3.3a provide evidence for confirmation of this hypothesis. During the object identification task, subjects using the low stiffness prehensor demonstrated consistently higher *d'* indices relative to their own performance with the high stiffness prehensor. This effect is present across all four object pairs tested, and reaches statistical significance in two cases (O_1-O_3 and O_1-O_5). It is clear that a low stiffness prehensor enabled the subjects to improve their performance in the object identification tasks. However, since only two stiffness conditions were tested, there is not enough data to draw any conclusions about the consistency or linearity of the effect. The results presented in Chapter 2 suggests that the effect is monotonic and linear, but further testing is needed to confirm this.

The effect size of prehensor stiffness on object identification performance observed in this study was not uniform. When attempting to identify objects in the O_1-O_4 object pair, the subject's performance was virtually identical at the two stiffness settings. The data trends suggest that for this object pair, the subjects underperformed relative to their expected sensitivities in the low stiffness condition. As a reminder, the task difficulty objectively decreases as the difference in object stiffness grows (i.e moving from left to right in Figure 3.3a). Isolating the high stiffness condition, subjects demonstrate a steady increase in performance as the task becomes easier. However, in the low stiffness condition, subjects demonstrate virtually no gains in sensitivity between object pairs O_1-O_3 and O_1-O_4. This lack of improvement produces the marginal difference in performance between the two prehensor conditions at the O_1-O_4 object pair. Given the higher performance of the low stiffness prehensor across all other object pairs tested in this study as well as the previous study presented in Chapter 2, it is likely that the depressed sensitivity of subjects when interacting with this specific object pair is the result of high variability in data, both within and across subjects.

3.4.2 Disturbance Detection

My initial hypothesis held that during the disturbance detection task, subjects would demonstrate higher sensitivities when using a high stiffness prehensor. The pooled sensitivity indices of subjects completing this task at several difficulty levels are presented in Figure 3.3b. The data support this initial hypothesis - Experiment 2 subjects consistently demonstrated better performance in the disturbance detection task in the high stiffness condition. The effect is present across all reference-signal pairs, and reaches statistical significance for two of them (D_1-D_4 and D_1-D_5).

The observed effect size was non-uniform with respect to task difficulty. As the difference between the reference stimulus and the signal grew in magnitude, the performance advantage provided by the high stiffness prehensor became more and more pronounced. Focusing on each condition in turn, it may be observed that subjects using a high stiffness PE demonstrated a consistent and virtually linear improvement in performance as task difficulty decreased. However, in the low stiffness condition, the subject's performance stagnated. As may be observed in Figure 3.3b, despite growing differences in magnitude between the reference and stimulus, the subject's sensitivities changed very little for the latter three object pairs. It is possible that the low stiffness condition made the disturbance detection task too difficult for the subjects, and they hit the limit of their performance.

3.4.3 Prehensor Stiffness Modulation

The weight of the evidence collected during the two experiments strongly supports both of the initial hypotheses. Pooled sensitivity indices obtained from the aggregate data clearly point to low prehensor stiffness as being more suitable for an object identification task, while a high prehensor stiffness is associated with better performance in an disturbance detection task. Given this result, it is clear that limiting the prehensor of a body-powered prosthesis to a single stiffness setting artificially lowers the quality of haptic feedback received and stifles the performance of the user in feedback dependent tasks. No matter which stiffness setting is chosen, there will be a task or tasks in which the user's performance is suboptimal. In order to avoid this, the prehensor stiffness must be made variable, enabling the user to adapt the prosthesis to the requirements of a particular task.

Besides the technical challenges involved, the implementation of prosthesis stiffness modulation would depend on two factors:

1. the ability of the user to evaluate the impact of prehensor stiffness on task performance.
2. the willingness of the user to expend the effort necessary to adjust the prehensor stiffness.

It should be noted that the two factors are related - if the perceived benefits of stiffness modulation are high, then logically the users would be willing to work harder to achieve them. On a technological level, the latter of the two factors could be addressed through the design of an intuitive control interface that minimizes the mental and physical effort necessary to adjust the prehensor stiffness. Previous studies by Sensinger and Weir [39] and Blank et al. [44] have found that subjects may not be able to adequately control the stiffness setting of an artificial device if the control space is continuous. It is likely that allowing the subjects to select from a discrete set of stiffness settings would reduce the mental load involved in the task, and produce more reliable results.

However, the ability of users to independently assess the effect of prehensor stiffness on their performance in a feedback dependent task remains an open question. Given the relatively narrow range of viable prehensor stiffness settings, it is possible that users would be indifferent or insensate to the advantages provided by a specific stiffness setting in a given task. This problem is specifically addressed in Chapter 4.

The improvements in feedback functionality provided by prehensor stiffness modulation have the chance to encourage users that had abandoned their prostheses to resume regular wear [11]. Additionally, prehensor stiffness modulation has the potential to reduce the high levels of stress induced comorbidities in the actuating (i.e. shoulder) joints of body-powered prosthesis users.

3.4.4 Sensitivities of Individual Subjects

When looking at the performance of individual subjects (see Table 3.1), the clear, monotonic patterns present in the aggregate data do not emerge. While some subjects (S8, Experiment 1) do demonstrate a consistent improvement in performance when using the more advantageous prehensor stiffness, the sensitivity patterns of most subjects are far less clear, with many of them going against the group trend during one or more experimental blocks. Is it possible that the effect of prehensor stiffness on task performance is more complex than suggested by the group trends?

To answer this question, it is important to note that SDT analysis depends heavily on large numbers of trials, and aggregating data across groups is a valid approach when faced with a limited number of trials per subject [51]. Thus, it is likely that the variability in subject-level sensitivity indices would be reduced if more data points were collected. This conclusion is supported by the fact that even when operating the PE at a consistent stiffness, subjects' performance varies greatly as a function of the specific reference-signal pair. Despite the fact that the difficulty of both tasks objectively decreased as the difference in the intensity of the reference and signal stimuli increased, some subjects (S8, Experiment 2; S3, Experiment 1) demonstrated their lowest *d'* score during the "easiest" experimental block.

In addition to low trial numbers per subject, another possible source of this variability are learning effects. Since the order of presentation of the various stimuli pairs was not counterbalanced, it is possible that the subjects became biased over the course of the data collection sessions. The impact of presentation order on performance was quantified, and the results for the high stiffness condition of the object identification experiment are presented in Figure 3.4. These results are representative of the behavior of the subjects across all conditions. No consistent evidence of learning is present in the data. While a few subjects (see S1 in Figure 3.4) did demonstrate a relative increase in performance as they completed more blocks, the performance of others suffered with each completed block. No clear trend toward better task performance over the course of the experiment is evident. Therefore, it is likely that any learning effects, if present, were dominated by the differences in difficulty between blocks. This leaves the small number of trials collected at each condition as the most reasonable explanation between the variability seen in the performance of individual subjects.

3.4.5 Limitations

This study had several limitations. The first and most obvious is the use of only two prehensor stiffness settings as experimental conditions. This makes it impossible to discover if the effect of stiffness on task performance is linear. This experimental design choice was mandated by a need to limit the time commitment of individual subjects, and at the same time understand the breadth of the impact of prehensor stiffness on task performance. This limitation may be addressed by limiting the number of stimuli the subjects are interacting with, while increasing the number of stiffness conditions to five. Note that the results presented in Chapter 2 do provide evidence that the effect is monotonic, at least when dealing with object identification. In addition, the subjects had no control over the stiffness of the prehensor, performing all tasks at a preset stiffness condition. Future work will evaluate the capability of PE users to self–determine the appropriate level of stiffness for a specific feedback-dependent task.

The number of trials that each subject completed at a given condition was another limitation of the study, and its impact has already been discussed. This can be addressed by restricting the subjects to a single interaction with the stimuli during a given trial. In the current experimental design, it was common for subjects to explore a stimuli 5 or more times before making a decision. Thus, while the subjects were exposed to the stimuli multiple times, only a single data point was collected. Limiting the participants to a one interaction with the stimuli would allow for a dramatic increase in the number of trials, while keeping the time requirements virtually unchanged.

3.5 Contributions

This study contributed to the understanding of the effect of prehensor stiffness on haptic feedback quality in the following ways:

- **The impact of prehensor stiffness on the quality of haptic feedback provided by a body-powered prosthesis is task dependent**

  In the disturbance detection task, subjects' performance was maximized under a high-stiffness prehensor condition. Conversely, during object identification, a low prehensor stiffness resulted in better performance. This establishes solid motivation for the introduction of prehensor stiffness modulation as a control modality, since setting the prehensor to a single stiffness setting will inevitably result in a performance deficiency.

- **The impact of prehensor stiffness on the quality of haptic feedback is present independently of the difficulty of the task**

  During this study, the subjects in each group were asked to complete their assigned tasks at four difficulty levels, represented by the relative difference in the magnitude of the two stimuli they were asked to distinguish between. The effect of prehensor stiffness on performance described above (i.e. better performance with a low stiffness prehensor during object identification) was observed at all difficulty conditions, though the effect size did vary.

CHAPTER 4

ON THE ABILITY OF USERS TO INDEPENDENTLY ADJUST PREHENSOR STIFFNESS TO ENHANCE TASK PERFORMANCE

The studies described in the previous two chapters provide quantitative evidence for the conclusion that the stiffness of the prehensor of a body-powered prosthesis impacts the quality of haptic feedback the prosthesis provides. In addition, this effect is task-dependent—the prehensor stiffness that maximizes performance changes based on the task. Thus, it may be concluded that the introduction of prehensor stiffness modulation to BP prostheses would be beneficial. However, in order for stiffness modulation to be a useful control modality, the users must be able to make informed decisions about the impact of prehensor stiffness on their performance. Thus far, the subjects in the studies have operated the PE system at a given stiffness setting. If given control of the device, would they be able to make rational choices about the prehensor stiffness in order to improve their performance in a given task? In this chapter, I present the results of a human subject study intended to answer this question. The data collected suggest that after only a brief familiarization period, subjects are aware of the effect of prehensor stiffness on their performance in feedback-dependent tasks (object identification and disturbance detection). Given the opportunity, the subjects demonstrated a capability of selectively altering the prehensor stiffness in order to improve their performance in the experimental tasks.

4.1 Introduction

The work presented in this chapter serves as a capstone to a series of studies examining the impact the prehensor stiffness of a body-powered prosthesis emulator has on the performance of users in feedback dependent tasks. This section lays out the background and motivation for the series of studies, and places this work in the context of previous research.

4.1.1 Prosthesis users are unsatisfied with the quality of the devices

Over half a million people in the U.S. are living with some level of upper limb loss [28], and despite recent advances in technology, many prosthesis users are dissatisfied with the level of functionality provided by the devices. The rate of prosthesis disuse or abandonment remains high, reaching 40% in some populations [10, 29]. In addition, about a third of users choose to utilize their prosthesis in a purely cosmetic manner, essentially ignoring the functionality of the device [10]. This user dissatisfaction is present in both types of active hand prostheses - robotic and body-powered. There are many factors contributing to this discontent, but a commonly cited issue is the quality of haptic feedback provided by the prosthesis [11]. While currently available robotic devices offer virtually no such feedback, the venerable body-powered prostheses include this feature by default, courtesy of their mechanical linkage to an actuating joint. However, this feature comes with a cost, as the rates of cumulative trauma disorder in the contralateral limb are high [29]. Many of those who had abandoned their prosthesis have expressed a willingness to reconsider, provided that cost-effective improvements in device functionality could be made [11]. With body-powered devices currently being used by over 50% of the affected population [31], a design change which led to improvements in feedback quality would be very welcome.

4.1.2 Stiffness modulation is an under-utilized control modality in prosthesis design

One possible avenue for improving the feedback characteristics of body-powered prostheses is the introduction of prehensor stiffness modulation to the devices [31]. Limb and joint stiffness modulation is an innate characteristic of the biomechanics of human movement. Though many factors, such as muscle and tendon properties, contribute to joint-level stiffness characteristics [31], the main mode of control employed is the co-contraction of antagonist/agonist muscle pairs around the joint [1, 2]. Intact individuals have the ability to adapt the directional stiffness of their joints in order to better match the requirements of specific tasks. For example, stiffness modulation is used when performing a rapid, constrained motion [7], learning a novel task [6], or operating in an unstable environment [3–6]. Through co-contraction, joint stiffness characteristics may be regulated without affecting the joint torque [3, 16], though the application of force by a limb does increase its stiffness along the vector of the applied force [22, 23]. Note that while other factors, such as damping and inertia, contribute to the overall limb impedance, the evidence for active modulation of damping in joints is inconclusive [23, 24], and inertial effects are completely dependent on limb posture [21]. The need for prostheses to incorporate stiffness modulation as a control modality has been recognized for decades [8], however to date, it has not been implemented in any commercial prosthesis [31].

4.1.3 Recent research on stiffness modulation in a prosthesis context

In recent years, several research groups have investigated stiffness modulation in the context of prostheses and similar systems [39, 40, 43, 45, 49]. Both Ajoudani et al. [40], as well as Hocaoglu and Patoglu [43], explored the concept of tele-impedance by passing on the stiffness characteristics of an intact limb to a robotic system used as a stand-in for a prosthesis. Sensinger and Weir [39] created a 1 DoF elbow prosthesis with variable impedance, and tested the ability of users to control the device impedance through an electromyographic (EMG) interface. Blank et al. [45], used both virtual and robotic systems to evaluate the utility of stiffness modulation in improving user performance in two control tasks: trajectory tracking and contact force minimization. All of these studies however, had some common limitations. They either used robotic systems as stand-ins for EMG prostheses [39, 40, 43, 49], or employed idealized control interfaces [45]. This focus on advanced robotic systems in favor of the simple mechanical devices is a common feature of current research [31], which serves to reduce the applicability of these results to body-powered prostheses, which still make up more than 50% of the market. In addition, all of these studies examined the impact of stiffness on the control of the system, leaving the feedback part of the user experience unexplored.

While body-powered prostheses have not been totally ignored, only a few researchers have tangentially addressed the impact of prehensor stiffness on task performance. Smit and Plettenburg [47] conducted a systematic characterization of commercial voluntary closing (VC) body-powered prostheses, focusing on the mechanical efficiency of the systems. In subsequent studies, Smit [48] subjected voluntary opening (VO) devices to a similar analysis, and designed a mechanism intended to compensate for the excessive stiffness present in VC prehensors [9]. However, these studies did not examine the effects of actively adapting prehensor stiffness on task performance. Frey and Carlson [46] proposed a design of a body-powered prosthesis with a variable mechanical advantage, however the work was primarily concerned with increasing the grip force of the prosthesis.

4.1.4 Motivation

The current state of research and industry may be summarized as follows: upper-limb prosthesis users are generally dissatisfied with performance of the available devices, particularly when it comes to the quality of haptic feedback. Unlike intact limbs, prostheses generally lack stiffness modulation capabilities, which may negatively impact user performance and satisfaction. The control/feedback mechanisms of body-powered prostheses, which feature a direct mechanical coupling from the prehensor to an actuating joint, make it likely the stiffness of the prehensor could impact the quality of feedback received by the user. However, to date, body-powered prostheses have been largely passed over by the research community in favor of a focus on robotic devices. In my previous two studies, I have attempted to address this knowledge gap and quantify the potential impact of prehensor stiffness on the performance of subjects using a body-powered prosthesis emulator (PE) to complete feedback-dependent tasks.

During the first study [54, 56], described in detail in Chapter 2, I examined the influence of prehensor stiffness on the ability of subjects to distinguish between objects based on their apparent stiffness. For both real and virtual objects, the data collected suggested that a low prehensor stiffness resulted in improved performance in this task. This initial study was subject to several limitations however, such as a limited subject pool. In addition, while the utility of a low prehensor stiffness was supported by the data, this is not enough to determine whether true stiffness modulation has a place in BP prostheses. In particular, it was left undetermined whether a high prehensor stiffness could be beneficial during some tasks.

In the second study in the series, presented in Chapter 3, I attempted to improve and expand on the results of the first work. The main addition was the inclusion of a second feedback-dependent task - disturbance detection. A high prehensor stiffness condition resulted in improved subject performance in the disturbance detection task. In addition, the data collected reinforced the main conclusion of the initial study, namely that a low prehensor stiffness is beneficial during the object identification task.

Taken together, these two studies provided quantitative evidence of the potential improvement in prosthesis haptic feedback that could be realized through the introduction of stiffness modulation. However, several crucial questions remained unanswered:

1. *Is the effect of prehensor stiffness on feedback quality linear?*

   The use of binary experimental conditions (i.e. high or low prehensor stiffness) during the bulk of previous data collection sessions made answering this question impossible. While some of the data collected during the first study suggested that the effect was at least monotonic, the low volume of available data made this conclusion uncertain.

2. *Are the subjects capable of independently evaluating the impact of prehensor stiffness on their performance?*

   To date, subjects in my experiments completed all tasks at a prescribed prehensor stiffness. If given the ability to self-select the prehensor stiffness, would subjects be able to identify the stiffness setting most appropriate to the task?

4.1.5 Current work

The questions left unanswered by the previous works were used to define the two hypothesis tested during this study:

- **Hypothesis 1:** *the effect of prehensor stiffness on haptic feedback quality during both object identification and disturbance detection tasks is linear.*

- **Hypothesis 2:** *subjects are aware of the impact of prehensor stiffness on their performance, and given the chance will select a stiffness setting appropriate for the task.*

To test these hypotheses, I conducted a series of human subject experiments, in which subjects were asked to use a PE system to complete an object identification and a disturbance detection task at a five distinct prehensor stiffness conditions. In addition, subjects were asked to independently choose which stiffness setting they judged to be optimal for the task.

The results were evaluated through the application of signal detection theory (SDT) analytical framework, [51, 52] which has previously been shown to be viable in experiments involving virtual object discrimination[53].

The rest of this chapter is organized as follows: Section 4.2 describes the experimental methods and procedures employed, Section 4.3 presents the observed results, and Section 4.4 contains a discussion of the key findings. Finally, Section 4.5 presents the key contribution of this study to the overarching questions of this work.

4.2 Methods

4.2.1 Setup

The prosthesis emulator (PE) used during the previous two studies in the series (see Chapters 2 and 3) was not modified. Since it has been described in detail elsewhere, only a brief overview of the mechanical design is presented here.

The PE consists of 1 DoF prehensor, attached to a 3D printed frame, and actuated by a Bowden cable attached to a harness worn around the users contra-lateral joint (see Figure 2.1). Forward extension of the user's shoulder closes the prehensor aperture. If the prehensor encounters an object or a position disturbance while closing, a change in the tension within the cable is produced. This approximates the control/feedback interface of a body-powered (BP) prosthesis, operating in VC mode. Unlike a standard prosthesis, the tension which keeps the prehensor aperture open is provided not by a mechanical spring, but by an electrical motor (Maxon RE 30 DC). This motor is mechanically linked to the prehensor through a cable drive with a 12:1 ratio (see Figure 2.2). By adjusting the output torque of the motor in response to a change in the position of the prehensor, the PE system is capable of simulating a continuously variable prehensor stiffness (K_D). Its practical prehensor stiffness range is limited by internal friction on the low end, and max motor torque on the high end, and spans 3.7 mNm/° to 7.5 mNm/°. Previously, a maximum of three stiffness settings had been implemented. For this study, a total of five stiffness settings were defined, K_D^{1-5}, evenly spaced across the operational range of the PE. Thus,

$$K_D^{1-5} = [3.7, 4.7, 5.6, 6.6, 7.5] N - mm/^{\circ} \tag{4.1}$$

Part of the experiment required the participants to select the stiffness setting they felt was most appropriate for the task from the five available. A keyboard was used for this purpose, with a specific button assigned to each of the five stiffness settings.

During all experiments conducted as part of this work, the lever arm between the prehensor shaft and the cable was set to 18.7 mm, and the range of motion for the prehensor was set to 50°. An encoder attached to the motor kept track of the angular displacement of the prehensor. Visual feedback of the prehensor displacement was blocked via a screen. During all data collection sessions, subjects wore noise-canceling headphones, which played pink noise at a volume loud enough to mask the operational noise of the PE.

4.2.2 Subjects

A total of 10 able-bodied subjects were recruited for the study. The subjects (S1-S10, 4 female, 6 males) each completed two data collection sessions, each focused on either the object identification or the disturbance detection task. Informed consent was given by all participants, and the procedure was deemed exempt from IRB review by the Human Subject Research group of the Colorado School of Mines.

4.2.3 Procedures

The study participants completed both tasks over the course of a pair of experimental sessions, held on two separate days. The order of the task sessions was randomized and counterbalanced between the subjects. For both tasks, the overall experiment design was the same, with all sessions employing a 2-alternative forced choice (2AFC) paradigm.

4.2.4 Object Identification

During interaction with an object, the force feedback that a body-powered prosthesis displays to a user is proportional to the sum of K_D, the device (prehensor) stiffness and O, the object stiffness, i.e.

$$F \propto K_D + O \tag{4.2}$$

Thus, interaction with objects of different stiffness will produce variations in the force feedback displayed to the user, filtered through the stiffness of the prehensor itself.

During the object identification sessions, participants were tasked with using the PE to interact with a pair of virtual objects, and correctly identify them based on their response. The objects were modeled as linear virtual springs of different stiffness, simulated by modulating the torque output of the prehensor motor. The two objects used in the experiment were based on one of the object pairs employed in the previous study described in Chapter 3. The softer of the two objects, O^-, had a stiffness of 6 N-mm/°. The stiffer of the two objects, O^+, had a stiffness of 14 N-mm/°.

As the subjects displaced the prehensor from its resting position, they initially felt only the force feedback resulting from the prehensor stiffness, K_D, acting against them. However, once the prehensor passed a predefined 20° contact threshold, the effects of the object spring (O_{1-2}) also started to contribute to the force in the actuating cable. See Figure 2.3 for a diagram of the PE system during the object identification task. As per Eq. 4.2, the force feedback displayed to the subjects was proportional to the sum of the prehensor and object stiffness coefficient. More specifically, the total force, F, displayed to the subject during any individual trial was

$$F = l(\theta K_D + (\theta - \theta_t)O) \tag{4.3}$$

where l is the lever arm of the cable, K_D is the device stiffness, θ is the displacement of the prehensor from its resting position in degrees, θ_t is the contact threshold, and

$$O = \begin{cases} O_{1-2} \ if \ \theta \geq \theta_t \\ 0 \ if \ \theta < \theta_t \end{cases} \tag{4.4}$$

Based on these relative differences in force feedback, during a specific trial the participants attempted to identify the virtual object they were interacting with as either the soft object (O_1) or the hard object (O_2).

4.2.5 Disturbance Detection

If the prosthesis prehensor is unexpectedly displaced from its trajectory as it is closing, the change in the feedback force is defined by:

$$\Delta F \propto K_D * D \tag{4.5}$$

where ΔF is the change in feedback force, K_D is the prehensor stiffness, and D is the magnitude of the displacement disturbance in degrees.

Thus, two displacement disturbances of unequal magnitude will produce different changes in the force feedback experience by the user, which may be used to distinguish between the disturbances.

During the disturbance detection sessions, two disturbances D_{1-2}, varying in magnitude, were applied to the prehensor. A diagram of the PE system during the disturbance detection task is presented in Figure 3.2. The disturbances were implemented as a displacement offset, added on top of the sensed prehensor position, θ that persisted for 2°. Specifically, as the subjects actuated the PE system, once the prehensor reached 25° of angular displacement, the value of the specified disturbance D_{1-5} was added to θ, and this change was maintained for 2°, after which it was removed. Thus,

$$F = l[(\theta + D)K_D] \tag{4.6}$$

where l is the lever arm of the cable, K_D is the device stiffness, θ is the angular displacement of the prehensor from its resting position, and

$$D = \begin{cases} D_{1-2} \; if \;\; 25^\circ \leq \theta \leq 27^\circ \\ 0 \; otherwise \end{cases} \tag{4.7}$$

is the disturbance applied as an offset in displacement, which is perceived by the subject as a pulse in the force feedback provided by the PE to their shoulder. This interaction resembles the scenario of an external impact/bump on a grasped object.

The disturbance magnitudes matched those used in a previous study, described in detail in Chapter 3. Specifically, $D_1 = 70^\circ$ and $D_2 = 90^\circ$. These magnitudes may seem unrealistic considering that the physical displacement of the prehensor was limited to only 50°. It must be emphasized that the virtual disturbance magnitudes represent only the commanded displacement offset of the prehensor. Due to the limitations of the PE motor, the actual displacement of the prehensor during the disturbance was far lower, as the actuating force provided by the subject overpowered the additional torque generated by the brief disturbance. Note that while the prehensor was not physically displaced by the nominal magnitude of the simulated disturbance, the PE motor generated the same torque as if it had been. Thus, the pulse in the force feedback displayed to the participants was equivalent to one that would have been generated by a physical displacement of the prehensor (see Eq. 4.5).

The disturbances were displayed to the users only when the prehensor was in the process of closing. Dynamic stability in the transition phase was maintained by establishing a minimum $\dot{\theta}$, below which the disturbances were not displayed. This prevented the subjects from slowly "feeling out" the disturbance region. In general, the subjects preferred pace was well above the minimum $\dot{\theta}$ threshold.

4.2.6 Data Collection

The process of data collection was identical for both the object identification and the disturbance detection sessions. The only difference between the sessions was the type of stimuli presented to the subjects. In both cases, the lower intensity stimuli (soft object, small disturbance) was considered as the reference, and the higher intensity stimuli (hard object, large disturbance) was considered as the signal. During each individual trial, the subjects were presented with either the reference or the signal stimuli, and asked to identify which of the stimuli they had interacted with. In a departure from the procedure used in previous studies, the subjects were limited to a single interaction with the stimuli per trial. This change allowed for a 4-fold increase in the number of data points collected from each individual subject without increasing the time duration of the sessions. Volunteer pilot test subjects, familiar with the old protocol, commented that far from reducing accuracy, the introduction of an interaction limit helped by reducing the temptation to second-guess their initial reaction.

The progress of a subject through a single data collection session is summarized visually in Figure 4.1. The subjects started off with a brief familiarization session, which served to introduce them to the mechanics of the experiment and ensure that they were comfortable with the operation of the PE. The subjects completed four familiarization trials at each of the five stiffness settings, for a total of 20 familiarization trials. During this process they were provided with correct/incorrect answer feedback through the computer screen in front of them. Over the course of familiarization, the stimuli presented to the subjects alternated in a steady pattern. This allowed the subjects to learn the distinct haptic feedback queues of each stimulus in controlled environment. During actual data collection, the sequence of stimulus presentation was randomized.

Once they were comfortable with the experimental set up, the subjects were asked to choose which of the 5 prehensor stiffness settings they considered to be optimal for the specific task. In order to inform their choice, the subjects completed 30 randomized and

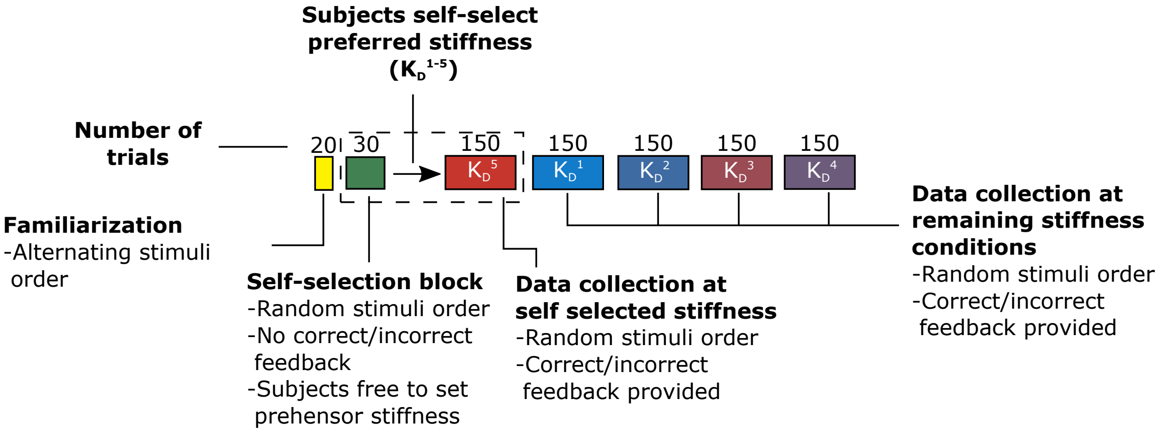


Figure 4.1: A diagram of the experimental procedure for one session of one subject. Subjects completed a total of 20 familiarization trials, 4 at each of the five available prehensor stiffness settings (K_D^{1-5}). This was followed by 30 randomized trials during which no correct/incorrect answer feedback was provided, and subjects were free to switch the PE system to any of the stiffness settings. Following this, the subjects selected a single stiffness setting from the five preset options as the most appropriate for the task. They completed 150 randomized trials at their self-selected stiffness setting. Following this, the participants completed and additional 4 trial blocks of 150 trials, with each block dedicated to one of the remaining stiffness conditions.

balanced trials of the session task. During this part of the experiment, the participants were able to freely select the stiffness setting of the prehensor. Crucially, no correct/incorrect feedback was provided to the subjects during this block, and thus their judgment of the suitability of a stiffness setting to a specific task was based only on their own perception of their performance. If the correct/incorrect feedback was still present, it is likely that the subjects would have relied on it almost exclusively in making their determination. Given the relatively small amount of trials in this block, a “lucky” run of a few correct answers at a disadvantageous stiffness setting could have greatly affected the subjects’ decisions. After completing the 30 trials, the subjects had to select one of the five available prehensor settings as their preferred one.

Once the subjects made their choice, they proceeded to complete 150 trials of the session task at their self-selected prehensor stiffness. The trials were randomized and balanced in 50 trial sub-blocks, allowing for the evaluation of learning effects. During this block,

Table 4.1: Self-selected prehensor stiffness of 10 subjects during two tasks

	Subject									
Task	**S1**	**S2**	**S3**	**S4**	**S5**	**S6**	**S7**	**S8**	**S9**	**S10**
Object Identification	4	3	3	4	4	2	5	3	3	2
Disturbance Detection	5	4	4	5	5	2	5	4	5	3

correct/incorrect feedback was once again provided.

Following the completion of the self-selected stiffness block, the subjects completed an additional 4 blocks of 150 trials. Each of the 4 blocks was dedicated to one of the remaining stiffness settings. The order of progression through the experimental conditions (i.e low-high stiffness, or high-low stiffness) was randomized and balanced between subjects. Correct/incorrect feedback was provided.

In summary, during each session, the subjects completed 5 blocks of 150 trials. The first of those blocks was completed at the stiffness setting judged by the subject to be most suitable to the particular session task. The rest of the stiffness blocks were completed in a prescribed order.

4.2.7 Data Analysis

All of the data collected during the experiments was analyzed using the Signal Detection Theory framework (SDT). The application of SDT techniques to the responses produced by the particpants of the 2AFC experiments allows the calculation of a sensitivity index, *d'*. This index is calculated as follows:

$$d' = z\{H\} - z\{F\} \tag{4.8}$$

where H represents the fraction of correctly identified reference stimuli (i.e. O_1 or D_1), F is the fraction of false alarms (i.e. how often the subject mistook a signal stimulus for the reference), and the $z\{\}$ operator represents a z-transform, which transforms the ratios into a z-score. Higher values of *d'* indicate a better ability to distinguish between two stimuli.

When the number of data points collected from individual subjects is relatively low, the sensitivity index is subject to significant random variation. In order to compensate for this, the responses of several subjects may be pooled to produce an estimate of the group sensitivity when identifying the stimuli [51]. The process of establishing confidence intervals on the pooled and individual sensitivity indices has been described in detail elsewhere [51, 56]. All confidence intervals were determined at the 95% confidence level.

The advantages of using a sensitivity index over a simple error rate comparison have been described in detail by other authors [51]. In summary, the use of the sensitivity index *d'* as a performance metric was motivated by the fact that it is not dependent on individual subject selection bias. Selection bias represents the individual tendency of each person to respond to every stimuli in the same manner - i.e some subjects will be more likely to identify everything as the high magnitude stimulus, and others will prefer the lower magnitude stimulus. Attempting to account for both types of bias using an error rate statistic can produce very misleading results [51]. The sensitivity index used throughout this work is directly derived from the actual error rates, but is not affected by the subject bias. As a point of reference, a sensitivity index of 0 may be considered equivalent to random selection between the two stimuli, or roughly a 50% success rate. Positive values of *d'* represent increased ability to accurately differentiate between the presented stimuli. For example, subjects who correctly identified 85% of the reference stimuli presented to them, while producing only 15% of false alarms would be assigned a sensitivity index of ~ 2.

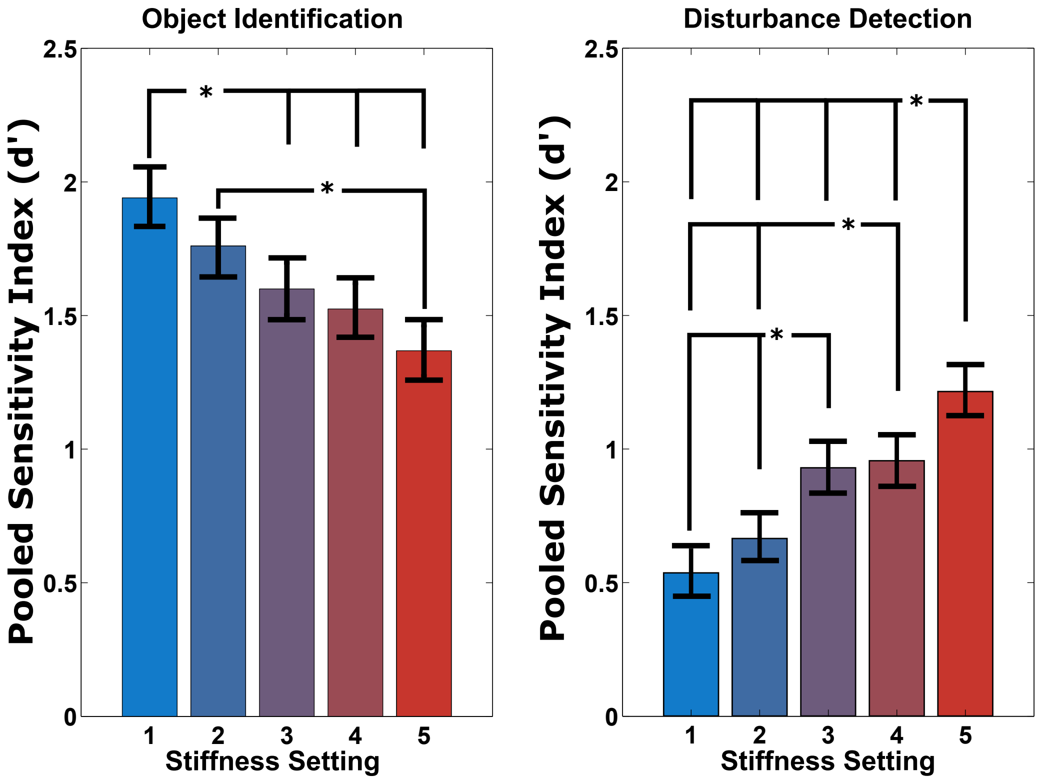


Figure 4.2: Pooled sensitivity indices for all 10 subjects across all tasks and experimental conditions. Statistically significant differences are reported at the 95% confidence level and are indicated with a *. Note the linear trend in performance relative to prehensor stiffness present in both graphs.

4.3 Results

The pooled sensitivity indices of all 10 subjects across all of the experimental conditions is presented in Figure 4.2. Statistically significant differences in senstivity index between experimental conditions are indicated with a * symbol. Note the clear impact of prehensor stiffness on performance in these two tasks. Lower prehensor stiffness resulted in better subject performance in the object identification task, and higher prehensor stiffness produced higher sensitivity indices during disturbance detection. In both tasks, the effect appears to be approximately linear.

The stiffness setting that each subject judged as most appropriate for a specific task is presented in Table 4.1. Note that out of 10 subjects, 8 preferred a higher stiffness for disturbance detection, and 2 subjects used the same stiffness for both tasks.

Histograms of the subject-preferred stiffness settings for each task are presented in Figure 4.3. Note that many subjects were willing to use very high stiffness settings during the disturbance detection task. In contrast, during the object identification task, subjects tended to stay to shy away from extremes, most often selecting one of the medium stiffness settings.

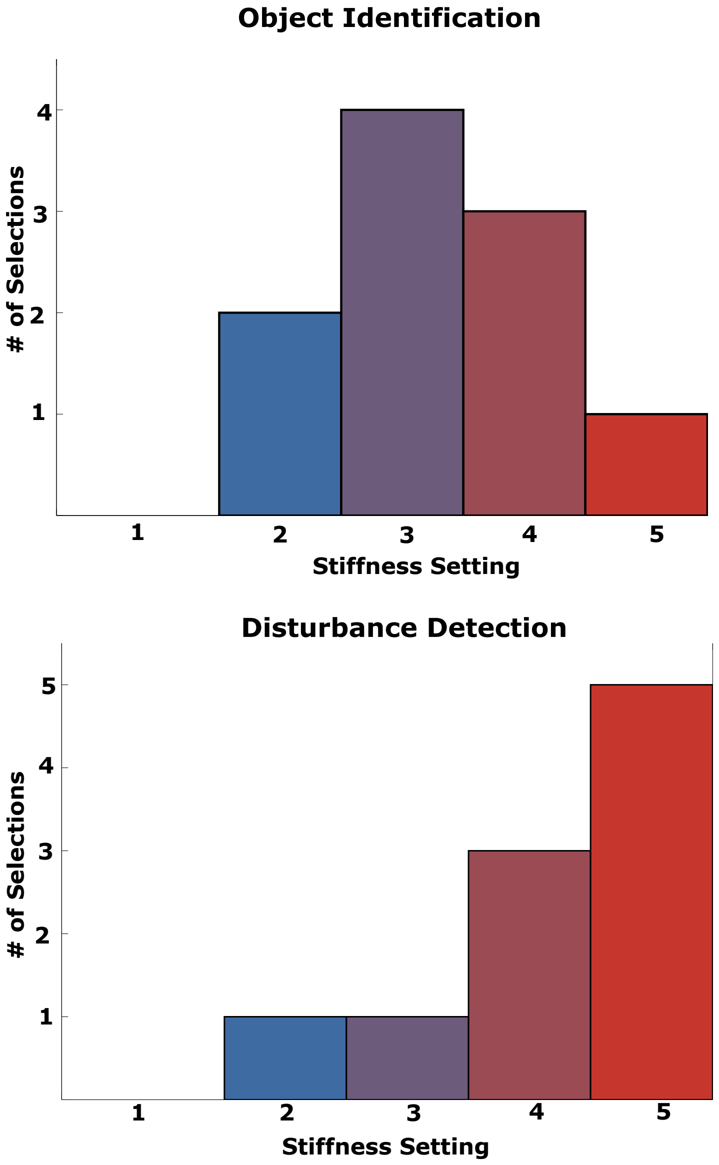


Figure 4.3: A histogram of the frequency with which subjects selected a specific stiffness setting during object identification (top) and disturbance detection (bottom)

The final two figures compare how well did each subject preform in their self-selected stiffness block, relative to their performance during the other 4 stiffness blocks. Figure 4.4 presents the data for object identification, and Figure 4.5 shows the results for disturbance detection. In both cases, the self-selected prehensor stiffness value of each is plotted on the horizontal axis, while the relative performance of the subject in that block is plotted on the vertical axis. Please note that because of its relative nature, the vertical axis is also discrete, with 5 possible rankings for the self-selected block performance, with 1 being the best, and 5 being the worst out of all the trial blocks a subject had completed. For example, a data point in the top right of the graph (high performance, high stiffness) indicates that the subject selected a high stiffness prehensor setting and performed relatively well during the task. The figures essentially add a second dimension to the histograms seen in Figure 4.3. Learning effects did not appear to influence the outcome of the overall experiment. However, the self-selected stiffness data collection block was always the first one completed, and it is possible that the subjects were still familiarizing themselves with the task. To account for this possibility, Figure 4.4 and Figure 4.5 also plot the same data but with the first 50 trials of the self-selected stiffness block omitted from analysis.

4.4 Discussion

4.4.1 Impact of Prehensor Stiffness on Performance

Based upon the data collected, **Hypothesis 1** may be confirmed - the effect of prehensor stiffness on the feedback quality during both tasks appears to be monotonic and approximately linear, though linearity is less pronounced in the disturbance detection task. Figure 4.2 shows that during the object identification task, the subject's sensitivity indices had a linear negative trend as the prehensor stiffness increased. Conversely, during the disturbance detection task, the trend was reversed, with the subject's performance steadily improving as the stiffness of the device increased. Though the linearity of the effect of stiffness on performance was not as clear during this task, given the underlying mechanics (4.6) it remains the most likely option. The observed performance trends fit with and complement

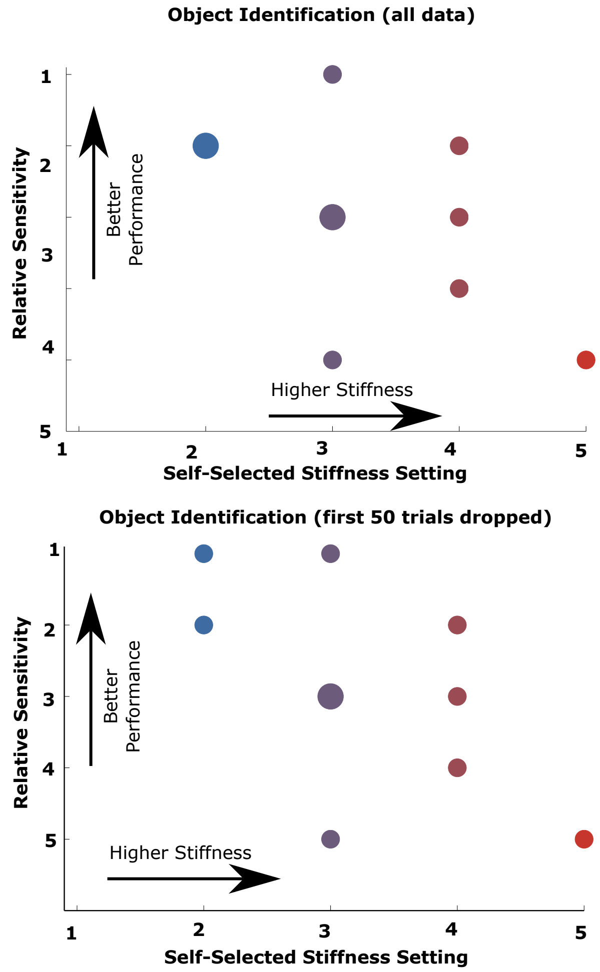
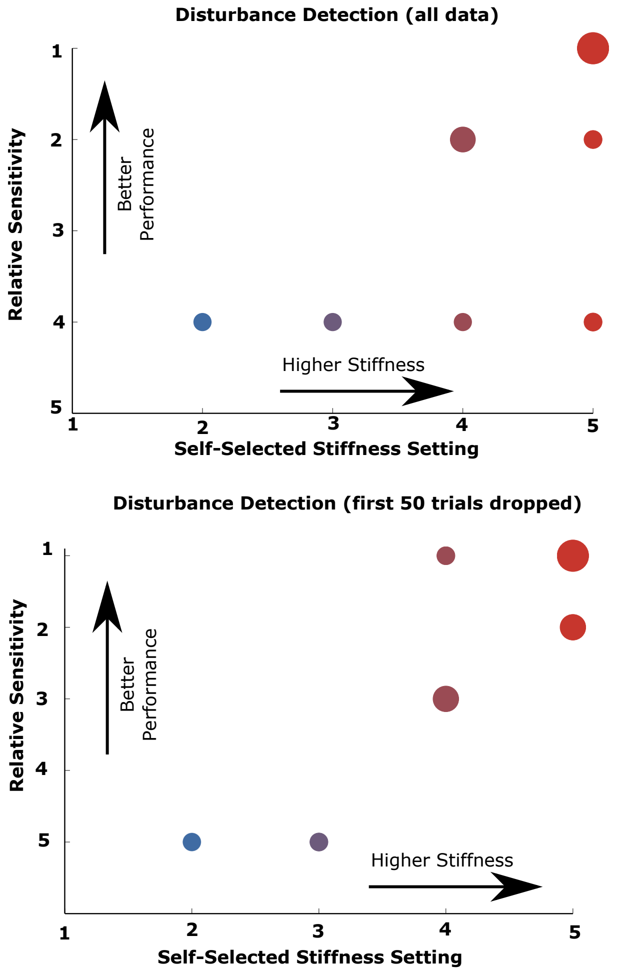


Figure 4.4: The relative performance of all subjects in the object identification task during their self-selected stiffness block, compared to their performance in the other blocks. The vertical axis is discrete, and ranks the subject's sensitivity in the self-selected stiffness condition relative to their sensitivities in the other conditions. A score of 1 means that in the self-selected stiffness condition, the subject demonstrated their highest performance. A score of 5 means that the self-selected stiffness block was the subject's worst. The horizontal axis is also discrete, and records the stiffness setting each subject selected. Given the discrete nature of the figure, overlap between the subjects occurs, and is reflected in the size of the markers. The top figure presents this analysis based on the entire data set, while the bottom figure omits the first 50 trials of each subject's data set.


Disturbance Detection (all data)

Disturbance Detection (first 50 trials dropped)

Figure 4.5: The relative performance of all subjects in the disturbance detection task during their self-selected stiffness block, compared to their performance in the other blocks. The vertical axis is discrete, and ranks the subject's sensitivity in the self-selected stiffness condition relative to their sensitivities in the other conditions. A score of 1 means that in the self-selected stiffness condition, the subject demonstrated their highest performance. A score of 5 means that the self-selected stiffness block was the subject's worst. The horizontal axis is also discrete, and records the stiffness setting each subject selected. Given the discrete nature of the figure, overlap between the subjects occurs, and is reflected in the size of the markers. The top figure presents this analysis based on the entire data set, while the bottom figure omits the first 50 trials of each subject's data set.

the results reported in my previous studies, which examined user performance at the two extremes of the current stiffness range (see Chapter 3).

It is apparent that the subjects found the disturbance detection task more challenging, across the entire set of prehensor stiffness settings. In fact, the highest sensitivity indices observed during disturbance detection barely reach the lowest recorded during object identification, with both occurring at stiffness setting 5. The most likely explanation is that disturbance detection involved a relatively discrete event, with the stimuli used being perceived as pulses by the subjects. On the other hand, during object identification, the participants were encouraged to close the prehensor almost all the way, their interaction with the simulated object lasting from 20°-50°. Coupled with no restrictions on the speed of the prehensor displacement, the subjects were able to evaluate the response of the object over the space of several seconds. Thus, this difference in sensitivities is not surprising given the nature of the tasks.

The drop-off in performance with respect to the prehensor stiffness appears to be steeper for the disturbance detection task. The decrease in sensitivity index between the best and worst stiffness setting was equal to 53% during disturbance detection, but only a 30% decrease was observed during object identification. In both tasks, the difference was statistically significant (see Figure 4.2). However, during disturbance detection, a small shift in prehensor stiffness was more likely to result in a statistically significant change in a subject's sensitivity index (see Figure 4.2). Thus it is reasonable to conclude that while performance in both types of tasks would be improved by the introduction of stiffness modulation to prosthesis, the impact of this control modality would be relatively greater in disturbance detection tasks.

4.4.2 Self-Selection of Stiffness

The second hypothesis tested during the study addressed the capacity of subjects to independently evaluate the impact of prehensor stiffness on their own performance in both tasks. Such awareness is inherently necessary in order for prosthesis users to take advantage of any stiffness modulation capabilities introduced to future prostheses. Based upon the data collected during this study, I can conclude that following only a brief familiarization period, and with no correct/incorrect feedback provided, subjects were able to selectively adapt their preferred stiffness to the specific task. For example, during disturbance detection, out of 10 subjects, 8 chose to use a higher stiffness prehensor setting than during object identification (see Table 4.1). The remaining two subjects chose to use the same stiffness setting for both tasks. It is reasonable to expect that with continued training, and visual feedback, the subjects' chosen stiffness settings would become even more optimized.

The histograms of the subject's selections, presented in Figure 4.3, provide additional details. While during disturbance detection, subjects were likely to choose one of the two highest stiffness settings - 4 and 5. Conversely, during object identification, subjects preferred to stay away from the lowest stiffness settings - no subjects selected setting 1, and only two subjects preferred setting 2. While the subjects selected relatively lower stiffness settings during object identification, they most often used the middle of the available stiffness range. This occurred despite the fact that the subject's actual performance in the task was maximized at the lowest stiffness setting. It is likely that additional considerations discouraged subjects from choosing the low stiffness condition. With a very soft prehensor, the PE system could feel "invisible" to the subjects - the low torques required to close the aperture provided little haptic feedback about the position of the prehensor. In the absence of visual feedback, this could be disorienting, and encourage the subjects to maintain their preferred stiffness in the middle of the available range. It is also interesting to note that higher prehensor stiffness settings require increased effort in operating the device. The final stiffness selection most likely depends on a trade-off between task performance and effort. At least for the range of stiffness values considered in this study, subjects were willing to exhibit more effort to improve their task performance. This can be taken as indication that BP prosthesis users would have sufficient motivation to make use of stiffness modulation.

Finally, it is possible that not all users' performance would be maximized at the extreme stiffness setting appropriate for each task. It is important to note that the relatively smaller number of data points available at the individual subject level makes quantitative comparisons much less reliable. Instead, performance trends at this level should be viewed as qualitative indicators. With that in mind, as shown in Figure 4.4 and Figure 4.5, the overall trend in performance during the self-selected stiffness matches the one observed in the entire data set (see Figure 4.2). When subjects selected a high stiffness setting to use during disturbance detection for example, they tended to preform well, relative to their own performance at other stiffness settings. However, this was not always the case. For example, several participants, who had elected to use a high prehensor stiffness setting, demonstrated surprisingly good performance in the object identification task. It is very likely that this is simply an artifact of the high variability present in a narrow data set, and that if the number of trials per subject were greatly increased, a more consistent pattern would emerge. However, even if the observed deviations from the overall trend accurately represent the real ability of the subjects, it reinforces the idea that introducing stiffness modulation to prostheses is a viable path toward improving performance.

4.4.3 Limitations

As the concluding study in a three-part series, the experimental design decisions made during this work represent an attempt at the systematic elimination of the limitations identified in my earlier work. The initial studies faced two consistent challenges - a use of binary prehensor stiffness conditions (low and high) and a low number of data trials for each subject.

Here, I have addressed the first limitation by introducing a range of stiffness conditions that spanned the viable operational range of the PE system. While 5 discrete stiffness settings fall short of the continuous stiffness modulation observed in intact limbs, it nevertheless represents a much more comprehensive search of the parameter space. Additionally, work by other authors has served to illustrate that the introduction of continuous stiffness modulation as a control modality can often lead to confusion for the study participants [39, 44].

However, this should not be taken as an indication that 5 stiffness levels will provide the ideal prehensor stiffness resolution for all tasks or all users. In practice, the optimal number of stiffness settings

Relative to my previous work, I have also increased the number of trials collected per subject and condition three-fold. This was achieved by limiting the subjects to a single interaction with the stimuli during a given trial. However, even with this increase, the uncertainty in the individual subject's sensitivity indices prevents drawing quantitative comparisons based on their performance.

The use of able-bodied subjects in this study may be considered an additional limitation. However,in the case of a unilateral amputation, the feedback and control mechanisms in the contra-lateral shoulder remain largely unaffected. Since the contra-lateral shoulder joint is used as the interaction point by both the PE system and body-powered prosthesis, the performance of able-bodied subjects can serve as a reasonable approximation. While its possible that given their greater experience in interpreting haptic feedback at the contra-lateral shoulder, subjects with an amputation would show greater sensitivities on average, this is not a given. In addition, regardless of their potentially higher sensitivities, subjects with an amputation would still benefit from the increased haptic transparency provided by appropriately modulated prehensor stiffness.

It must be acknowledged the practical ability and willingness of prosthesis users to take advantage of the increased performance offered by prehensor stiffness modulation would be strongly dependent on the nature of the provided control interface. While a purely mechanical solution for changing the effective prehensor stiffness is feasible, asking the user to expend the necessary mental and physical effort to make such a manual adjustment prior to each task may be too taxing. If available, EMG readings from the residual limb may be used an indications of the desired prehensor stiffness. This approach has the advantage of approximating the control modality present in the intact limb, but has the drawback of introducing an electrical solution to what is originally a purely mechanical system. Thus, if attempted, any control electronics introduced must be extremely low powered, such that the power supply and associated components do not increase the weight and cost of the device unduly. However, even if a purely mechanical device is the only feasible path toward introducing stiffness modulation, it would still be beneficial to the user. For example, if the user is unwilling to actively modulate the prehensor stiffness, they may still, potentially in consultation with their prosthetist, adjust the stiffness to a preferred static value - which may be easily changed over time in response to physiological changes, such as the development of CTD in the actuating joint.

The experimental tasks used here are intentionally abstracted away from specific ADLs. The stated goal of the work is to evaluate the impact of prehensor stiffness on the quality of haptic feedback - a frequent concern among prosthesis users [11] and an active research topic [33]. Results obtained here indicate that prehensor stiffness modulation can provide a viable path toward improving feedback-based task performance - and thus prosthesis functionality - which is another commonly cited deficiency of the current generation of prostheses [11].

4.5 Contributions

This study contributed to the understanding of the role of prehensor stiffness in PE user performance in the following ways:

- **Subjects are aware of the impact of prehensor stiffness on their performance** In order for prehensor stiffness modulation to serve as a viable control modality, prosthesis users must be able to judge for themselves the utility of a given prehensor stiffness in a specific task. Based on the response of the subjects during the self-selected stiffness blocks, this appears to be the case. During both tasks, the subjects demonstrated the ability to make rational decisions about the appropriate stiffness setting for a specific task following only a brief familiarization session.

- **The effect of prehensor stiffness on haptic feedback quality is monotonic and linear**

The performance trends demonstrated by the subjects during both experimental tasks as a function of prehensor stiffness confirm that prehensor stiffness has a monotonic and linear effect on the quality of haptic feedback provided to the participants.

CHAPTER 5

CONCLUSION

The central aim of this work was to evaluate the potential for improving the quality of haptic feedback provided by body-powered prosthesis through the introduction of prehensor stiffness modulation as a control modality. This overarching goal was split into three main questions, each of which was addressed over the course of one of the three successive studies conducted during this research:

- **Does prosthesis prehensor stiffness have a quantifiable impact on the quality of haptic feedback received by the user?**

  As demonstrated in the initial study, presented in detail in Chapter 2, prehensor stiffness does have a measurable effect on the performance of subjects in an object discrimination task. In the low prehensor stiffness condition, subjects demonstrated increased sensitivity when attempting to use the prosthesis emulator system to discriminate between two objects based on their stiffness.

- **Is the effect of prehensor stiffness on haptic feedback quality task dependent?**

  The data collected during a follow-up study, detailed in Chapter 3, showed that subject's performance in an disturbance detection task was maximized in the high stiffness condition. Additionally, a second object discrimination experiment was conducted as well, which reinforced the findings of the initial study. Thus, in two different feedback-dependent tasks the subjects' performance was maximized at two different prehensor stiffness settings. From this, it may be inferred that introducing prehensor stiffness modulation as a control modality does have the the potential to improve the quality of haptic feedback provided by a body-powered prosthesis.

- **Do users have the capacity to accurately asses the impact of prehensor stiffness on their performance?**

  During the final study in the series, discussed in Chapter 4, the subjects were asked to self-identify the prehensor stiffness that they felt was most appropriate for a given feedback-dependent tasks. Based on the results of the experiment, it is evident that the subjects were aware of the impact of prehensor stiffness on their performance. Thus, if prehensor stiffness modulation is introduced to body-powered prosthesis, it is likely that the users would be able to make informed and rational adjustments to the prehensor stiffness of their device in order to improve their performance.

While several previous works have focused on the utility of stiffness modulation in improving the controllability of prosthesis-like robotic systems [39, 40, 43, 45, 49], to my knowledge, this series of studies represents the first systematic investigation into stiffness modulation in the context of body-powered prostheses. On the basis of the data collected during this work and it is predecessors, I can confidently conclude that prehensor stiffness does impact the haptic feedback quality provided by a body-powered prosthesis operating in VC mode. Thus, the introduction of prehensor stiffness modulation into body-powered prostheses could improve the performance of such prostheses in various feedback-dependent tasks, and can reduce device abandonment rates that remain high due, at least in part, to low quality of feedback [11][10].

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