WEARABLE TACTILE DISPLAYS FOR MOTION FEEDBACK

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I certify that I have read this dissertation and that, in my opinion, it is fully adequate in scope and quality as a dissertation for the degree of Doctor of Philosophy.

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Abstract

Humans have a rich set of sensors in muscles, tendons, skin and joints that provide information about the position and motion of our limbs, even when we cannot see them. This sense is called proprioception. While the role of proprioceptive feedback in motor control is not completely understood, its value is evidenced by those rare cases where people lose proprioception. Simple motions become nearly impossible. Upper-limb amputees are often fit with a robotic arm that is controlled with sensors placed on residual muscles, making the control somewhat intuitive. However, they lack proprioceptive feedback from the prosthetic arm. They are forced to rely solely on vision to move the arm which can result in jerky, individuated motions and produces a large cognitive burden. Wearable tactile displays, placed on the skin, present a non-invasive method to provide feedback to amputees to restore lost sensations. In this work, we present user studies in which we evaluate the effect of using tactile displays to provide proprioceptive feedback to people controlling a virtual object (e.g. a prosthetic arm). Both vibration and skin stretch displays are tested with skin stretch being preferred as it can simultaneously provide a sense of position and motion. Benefits of the artificial proprioceptive feedback are shown for a targeted motion task. In one study, users control a virtual prosthetic elbow with electromyographic (EMG) sensors placed on the bicep and tricep muscles. When skin stretch feedback was provided, blind targeting accuracy improved and visual demand was reduced compared to no feedback.

In addition to replacing a lost sense of proprioception, wearable tactile displays can

be placed on the limbs for motion training. We also present the results of a study in which we provide real-time visual and vibration feedback to users about a potentially damaging knee joint force as they walk on a treadmill. The feedback allows them to adjust the way they walk to reduce the force much faster than with conventional motion training methods, where verbal or visual feedback is given intermittently.

Based on the results of these studies, it is possible to formulate some general guidelines about the use of wearable tactile displays. Due to the delays associated with feedback (natural or artificial), it is most appropriately used for slow or repetitive motions. Haptic feedback is most useful when vision is devoted elsewhere or is ambiguous. Skin stretch can be more effective than vibration when it is desirable to convey a sense of motion *and* direction rather than event cues.

Acknowledgments

I've always enjoyed reading colleagues' acknowledgment sections in their theses. It seems you can learn a bit about them beyond their technical work; what they value, what they spent their few free hours doing during graduate school etc. That makes writing my own section a bit terrifying for some reason. Nonetheless, I have a number of people who I owe a great debt of gratitude and who have made this work possible so here we go.

I have wanted to do a Ph.D. for many years. When I finished my Master's degree at MIT I was planning on staying for a few more years. However, funding in our lab was at a temporary low point and we had just had our first child so we decided that a real job might be best at that time. Fortunately, I was able to find a job that not only allowed me to do the things I liked to do (research) but offered a program for Ph.D. students which was better funded that most programs. Of course there was a catch: a three-year time limit. With a little effort, I was able to find a few professors that would agree to advise me with that constraint. I was torn between Stanford (with Mark Cutkosky) and Johns Hopkins (with Allison Okamura). Both seemed like great opportunities but I didn't think I could have convinced Cecy to move to the East Coast again so Stanford it was. We certainly did not regret that choice.

I'm indebted to Mark Cutkosky for having confidence in me to pull this off and also for his incredible mentorship over the past three years. He was always willing to let me do the things that I was interested in while providing insightful feedback whenever needed. It's a rare thing to have an advisor so talented yet so accessible

and approachable. I consider him a mentor and a friend. In particular I remember struggling with a study where users controlled a virtual object with a force sensor but the task was too easy and the feedback we were giving was not useful. I was trying to come up with a way to make the task harder but still realistic. Mark, (with only a couple seconds thought) suggested giving the object position dependent dynamics (like a real arm would have) and of course that was what we did and it worked well.

Scott Delp was also an inspirational teacher and mentor during my time at Stanford. One of the reasons I wanted to come to Stanford was to learn about biomechanics and my expectations in that regard were greatly exceeded. I was fortunate to be able to do some research with Scott for class projects that do not appear in this work. Thor Besier was also an incredibly great person to work with. He always had great ideas and insights into our research and was more than helpful whenever I needed something. He was an example to me of a solid researcher who also happened to be an extremely nice person. Scott and Thor provided the original suggestions for the study presented in Chapter 3 and Thor contributed all along the way. I also need to thank the other faculty who assisted with my research and/or served on the defense committee, many of them on short notice. Thanks to Professors Oussama Khatib, Gunter Niemeyer, Bernie Roth and Ken Salisbury.

I owe a special thanks to two colleagues in the Biomimetics and Dexterous Manipulation Lab (BDML). Karlin Bark created the skin stretch feedback used in most of the studies presented in this work. She is a brilliant designer and solid researcher. I'm amazed that the device that she developed just as I arrived turned out to be ideal for the applications I was interested in. I always enjoyed our discussions about everything from piezoelectric motors to psychophysics. Pete Shull was a major contributor to the work presented in Chapter 3. He was instrumental in getting the real-time Vicon system set up and running the study. He's already moving beyond that to new and interesting things and I wish him all the best in that and look forward to working together on papers in the future.

As my work was largely based on user studies, I'm indebted to everyone who

volunteered to participate in my experiments. If I could name you I would. Your time is greatly appreciated. Also, thanks to everyone (too numerous to name) in the BDML for creating a collaborative and fun environment. I always enjoyed asking/answering technical questions and sarcastically arguing over space with the biomimetics people. I hope someone feeds the fish now that I'm gone. Who will take the throne of the first one in each morning?

A few people from outside the university also contributed useful feedback. Prof. Brent Gillespie was an extremely valuable resource and a great person to have longer than expected discussions with. I lucked out that he happened to be on sabbatical here when I arrived. He provided suggestions about the visual demand study in Chapter 5. Prof. Neville Hogan provided typically brilliant insight into the motor control aspects of the work. Also, though the work did not end up making it into the thesis, I wish to thank Netta Gurari and Prof. Allison Okamura at Johns Hopkins University for the opportunity to collaborate on a stiffness discrimination study. Thanks for hosting me there for a week, I really enjoyed it. Also thanks to Curt Salisbury for always being a good friend and excellent source of technical knowledge.

Of course, none of this would be possible without my beautiful and kind wife Cecy. When she agreed to marry me, I don't think she bargained for seven years consisting of four moves across four states. I'm pretty sure she was a bit reluctant to leave our house three years ago to move back to campus and a small apartment but I don't think she complained (unless sarcastically). I am eternally grateful for her patience and support. I also need to thank our three small children, Anna, Adam and Haley (who was not born when we started this but decided to come early the week before quals). They have always made it fun to come home and I always looked forward to seeing them. I wish I had better words to express how I truly feel about Cecy and our children. Any other achievement or opportunity pales in comparison to the joy I receive from our family.

My parents also deserve much of the credit for the things I have been able to do in my life. Not only are they great examples and teachers but great friends that I

respect as much as anyone I have interacted with. I appreciate their constant love and support for my family and me. I hope to be able to live my life as a worthy tribute to them.

Sandia National Laboratories, through the doctoral studies program, provided the majority of the funding for this work. The program really is incredible and I'm thankful that I was able to participate in it. With three kids in the bay area, I don't think we would have survived on a research stipend, in fact I know we would not have. Tekes, a Finnish government organization and the King Abdullah University of Science and Technology provided supplementary funding for hardware and experiments.

Contents

List of Tables

List of Figures

Chapter 1

Introduction

Researchers in the field of haptics have sought to utilize the vast array of tactile sensors in the human skin to improve communication between man and machine. Haptic devices have the ability to convey a sense of interaction forces and textures with a remote or virtual environment. Potential applications for haptics are myriad and include remote robotic teleoperation and surgical simulation and training. While these technologically-demanding applications could have a huge impact, to date the most ubiquitous and successful applications of haptics involve some of the simplest technologies.

Nearly every cell phone or pager is equipped with a vibration device to alert the user that an event has occurred (e.g. an incoming call). Game controllers also utilize simple vibration motors to create illusions of force or impact. This type of feedback is portable in that it consumes little power and does not need to be grounded to a large structure. It can also be more easily conveyed to a single person (without the knowledge of those in proximity to them) than many audio or visual displays. However, the bandwidth of the feedback is quite low. This is due to the relatively poor ability to discriminate vibration parameters as well as a person's ability to remember what a particular vibration pattern means. This limits its use to simple "event-cue" feedback rather than providing a rich sense of information, such as could be conveyed

with verbal comments or on a visual display. Tactile sensation is not typically used for conveying symbolic information (though it can be done, for example Braille) but it is inherently multi-modal and distributed. This allows relatively low bandwidth information of various types (e.g. force, vibration, temperature) to be conveyed in a localized manner. It can also be used in parallel with vision and audition to improve the overall flow of information.

In this work, we will attempt to identify a few practical applications of wearable, portable tactile displays that take advantage of the strengths of haptic feedback while acknowledging its weaknesses. Of particular interest are applications where the feedback is related to the motion of a person's body. The use of skin stretch feedback, in addition to vibration, is a critical component of this work. Skin stretch has the ability to convey an analog sense of position and motion in a way that is more intuitive than vibration.

Two classes of motion-based feedback applications will be explored in this work. In the first class of applications, haptic feedback is used to convey some new channel of information to the user that they would not normally have. Within the constraints of this research, the feedback is related to desired motions of the person's body. This feedback could be based on some biomechanical parameter that the person has no natural sense of but is desirable to control (as in Chapter 3). Alternatively, the feedback could be based on some external desired motions (e.g. in learning a skilled movement).

The second broad class of applications we will explore in this work involves providing users a sense of feedback of their own movements or the movements of some tightly coupled robotic device. The feedback of motion- and position-based information we will refer to as proprioceptive or kinesthetic feedback, which can be either natural or artificial (when provided with wearable haptic devices). These terms will be defined more rigorously in Chapter 2. One of the clearest examples of an application of artificial proprioceptive feedback, and the one we will explore in the most depth, is the control of prosthetic limbs. Upper-limb amputees are often fit with

robotic limbs that can be controlled with electromyographic (EMG) sensors placed on residual muscles, making the control somewhat intuitive. However, they lack any proprioceptive feedback from the limb and are forced to use visual feedback or feed-forward control strategies to move the limb. The addition of haptic feedback could improve the ability of users to control the arm accurately and intuitively. This benefit can be directly due to the addition of feedback or from improved learning of the arm dynamics.

In both classes of applications described above, it is critical to understand the role of feedback (natural or artificial) in motor control of various tasks. In order to identify appropriate applications for the haptic devices, we must understand not only how sensory information is used in motions, but how proprioceptive and visual information are integrated. Unfortunately, there is much that is not understood on this topic. However, we can say something about what the central nervous system (CNS) seems to be doing. A more thorough discussion of some of these points can be found in Chapter 2 but some high level points will be presented here to provide a framework for the applications explored in later chapters.

Figure 1.1 shows a schematic representation of a person performing a movement task. Based on a knowledge of the task and a dynamic model of the body, the CNS creates a movement plan that it begins to execute by sending efferent signals to muscles. These muscles articulate joints and segments that produce motions. Feedback from sensors in muscles, joints, skin near the joints and the eyes is conveyed to the CNS via various neural pathways, each with respective delays. Some of this feedback directly affects the motion via low-level (e.g. spinal) feedback loops. Higher level feedback (e.g. vision), which is subject to longer delays, can alter the movement plan or provide precise positioning as needed. Higher level feedback, which is generally processed in the cortex, has the longest delays and imposes the highest cognitive burden. The large delays associated with all of the feedback (at least 50 ms and as much as 250 ms [68]) indicate that it is not likely being used for real-time motion control. Instead, feed-forward strategies (and perhaps some efferent copy of expected

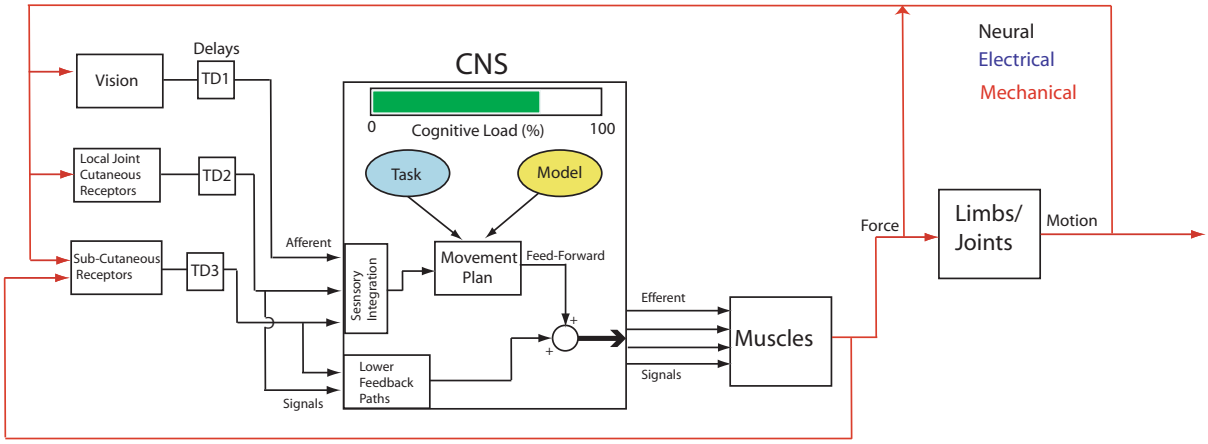

Figure 1.1: Schematic representation of a person performing a movement task based on a knowledge of the body's dynamics and the various sensory feedback elements present.

sensory feedback) are used to perform many movements at preferred speeds. The feedback may be used for slower corrections or adaptation of the dynamic models for future tasks. More will be said about this in Chapter 2.

When a person has a limb amputated and is fitted with an EMG-controlled limb, the efferent control path is somewhat intact but many of the feedback paths are eliminated or impaired (Figure 1.2). Electrical activity from residual muscles is detected to actuate movement of the robotic arm. Visual feedback is the only remaining sense of gross joint motion. However, we can place a haptic device on the skin near the amputated joint or on some other area that conveys a sense of the limb's motion to partially replace the lost sense. As with physiological senses, this feedback channel will have a delay associated with it, so its benefit will likely be for slower motions or for developing dynamic models of the limbs and joints. In myoelectric prostheses, motions are typically slower due to the lack of normal control mechanisms and the difficulty in building dynamic models of the controller and joints [103]. The addition of haptic proprioceptive feedback could reduce the visual demand associated with moving the arm, allowing vision to be devoted to other tasks. It could also improve

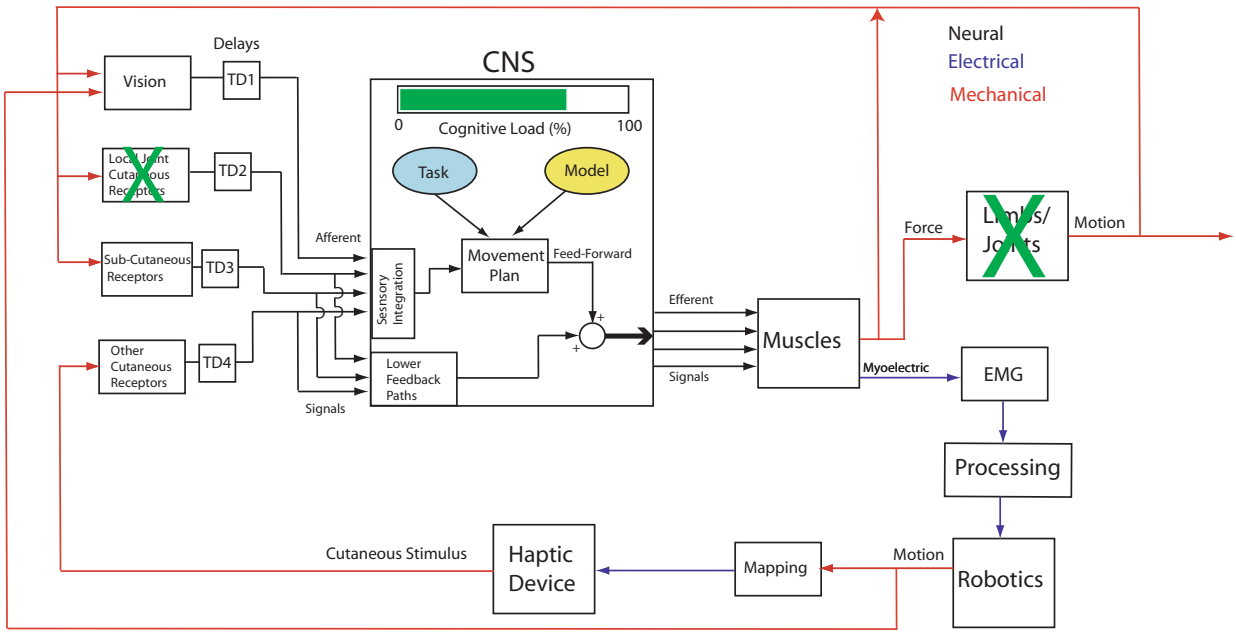

Figure 1.2: Schematic representation of a person performing a movement task with an EMG-controlled prosthetic limb. Many of the normal feedback paths are impaired or absent. Haptic feedback can be used to partially replace the lost sensation.

motion accuracy in the absence of vision. These specific hypotheses are tested in Chapter 5.

Based on this brief overview of the role of sensory feedback in movement, we can begin to formulate some guidelines for when haptic feedback might be useful for kinesthetic or proprioceptive feedback. First, when the feedback is related to desired motions of an unimpaired person, the feedback is likely to be beneficial for either slow movements, which may not be practically useful, or for learning a task in a repetitive fashion. In the latter case – which we think is most practical and has not been explored in depth to date – the feedback is not used in the current motion but in building feed-forward strategies for future movements. This feedback can be implicit (simply providing performance feedback to the person so they can iterate on

various strategies) or explicit (providing specific guidance on what motion corrections to make). In Chapter 3, we provide an example of implicit feedback for lowering a potentially harmful knee joint load during walking (a repetitive task).

The other application we will explore in this work involves the control of external devices (e.g. prostheses). In this case, the feedback is directly related to the control inputs from the users, which are limited to EMG or force in the studies we present. With both of these inputs, the user does not have natural proprioceptive feedback (at the joint level) as they would with a joystick or mouse. In the absence of proprioceptive feedback, users can potentially control the device with visual feedback or open-loop strategies based on models of the system's dynamics. We do not claim that the use of haptic feedback is superior to vision for most tasks, any more than natural proprioception is. However, we will often compare to completely feed-forward control strategies in the absence of vision for conditions when vision is not available. In these cases, the feedback is more beneficial if the dynamics or control properties of the system are not intuitive to learn. Additionally, we will evaluate if visual demand can be reduced with artificial proprioceptive feedback.

To summarize, we hypothesize that providing haptic feedback of motion-based information could be useful in at least the following situations:

- For motion guidance and correction of slow motions where visual feedback is ambiguous.
- For implicit or explicit motion guidance for repetitive tasks (Chapter 3).
- For control of systems with non-intuitive dynamics when vision is not available or is devoted to other tasks (Chapters 4 and 5).

Chapter 2 provides an overview of the major concepts of this research and presents the relevant prior work. In Chapter 3 we describe an experiment about motion training with visual and vibration feedback of a knee joint load that contributes to osteoarthritis (OA). Chapters 4 and 5 describe studies that use skin stretch (and

occasionally vibration) for artificial proprioceptive feedback in the control of various systems. The study presented in Chapter 4 involves controlling a cursor with non-intuitive dynamics with a force sensor. In Chapter 5, subjects control a virtual prosthetic arm with EMG sensors. Finally, Chapter 6 summarizes the main findings and provides discussion of the implications.

The primary contribution of this work is the demonstration of the utility of wearable tactile displays for two clinical applications. Vibration feedback is shown to be comparably effective to vision feedback in many respects in conveying biomechanical information for gait retraining. This has potential implications for osteoarthritis as well as other musculoskeletal or neurological diseases. Skin stretch is shown to be an effective method for providing artificial proprioceptive feedback. This has specific application to feedback in myoelectric prostheses. The feedback is particularly effective when vision is not available or for reducing the visual demand required to perform some motions. While this work is not expressly clinical in that we test healthy subjects rather than patients, it does establish the feasibility of two novel approaches at a practical level. The results presented here motivate further clinical studies of haptic technologies for these applications. At a higher level, we also establish some general guidelines for the use of wearable haptic devices for mobile use. For motion training, feedback is effective when provided immediately for repetitive tasks. Haptic feedback, which has a lower resolution than visual feedback but can be more easily used in portable systems, can still be effective for some tasks though learning may take longer. Vibration feedback is effective for low-bandwidth feedback, particularly when the feedback is localized or arrayed at multiple locations. Skin stretch is most effective for providing feedback of continuous motion and in closed-loop systems where the user has control of its motion.

Chapter 2

Background

This chapter provides a summary of prior work related to the topics of wearable haptic feedback and motor control. As there are a number of diverse topics that are relevant, some of the sections will provide only a cursory overview of the most relevant works. Background on specific applications is provided at the beginning of subsequent chapters as appropriate.

We first review the prior work in wearable haptic devices. The two primary modalities of feedback, vibration and skin stretch, are then discussed independently. We then discuss some of the relevant literature on proprioception and movement science. Finally, we discuss the prior studies of biofeedback for motion guidance.

2.1 Wearable Haptics

Most early haptic devices were designed to convey a sense of force to a person's hand. To do this effectively, they were "grounded" to a benchtop, allowing forces to be conveyed in the operational space of the endpoint. Wearable devices, which are not grounded, cannot convey forces in this way. They can produce net torques around joints however, and a number of such devices (exoskeletons) have been developed (e.g. [50, 69]). Of course, force is only one of many types of haptic information

that can be conveyed to a user. Mechanoreceptors in the skin can detect vibration, temperature, normal and tangential skin strains and pressure. An understanding of the types of receptors in the skin and their functional properties will help facilitate later discussions of the types of feedback devices that have been developed.

The human skin is generally categorized into either glabrous or non-glabrous (hairy) types. Glabrous skin is found primarily on the palms of the hands and bottoms of the feet. The vast majority of the skin surface area is non-glabrous. In this work we are primarily interested in feedback to this non-glabrous skin as the proposed applications involve placement of devices across the body segments. It is also beneficial to keep the hands free for other tasks and in some cases (i.e., amputees), the hand may be absent entirely. The following discussion will therefore focus primarily on the hairy skin. We will refer occasionally to devices used on glabrous skin as these seem to dominate the prior work.

Figure 2.1 shows the five main types of mechanoreceptors found in the human hairy skin [107]. The receptors are often categorized by their relative depth under the skin as well as the type of stimuli the respond to. Type I receptors are located superficially in the skin and have relatively small receptive fields. Type II receptors are found deeper in the skin and have large receptive fields. Slow adapting (SA) receptors respond to slowly changing stimuli while fast adapting (FA) receptors respond to more rapidly changing stimuli. As an example, the Pacinian corpuscle, which is known to be sensitive to vibratory stimuli, is classified as a FAII receptor because it responds to dynamic stimuli and is located deep in the skin. Table 2.1 summarizes the classifications and functions of the primary mechanoreceptors.

Rather than creating a sense of force at a point in task space or about a joint, there is clearly an opportunity to communicate information in a more compact package by stimulating a small number of mechanoreceptors [64]. This type of communication is not necessarily intuitive in the sense that it creates a realistic illusion of a tactile interaction but the nervous system can certainly learn to interpret the information with some training [113]. This idea of communicating information from external

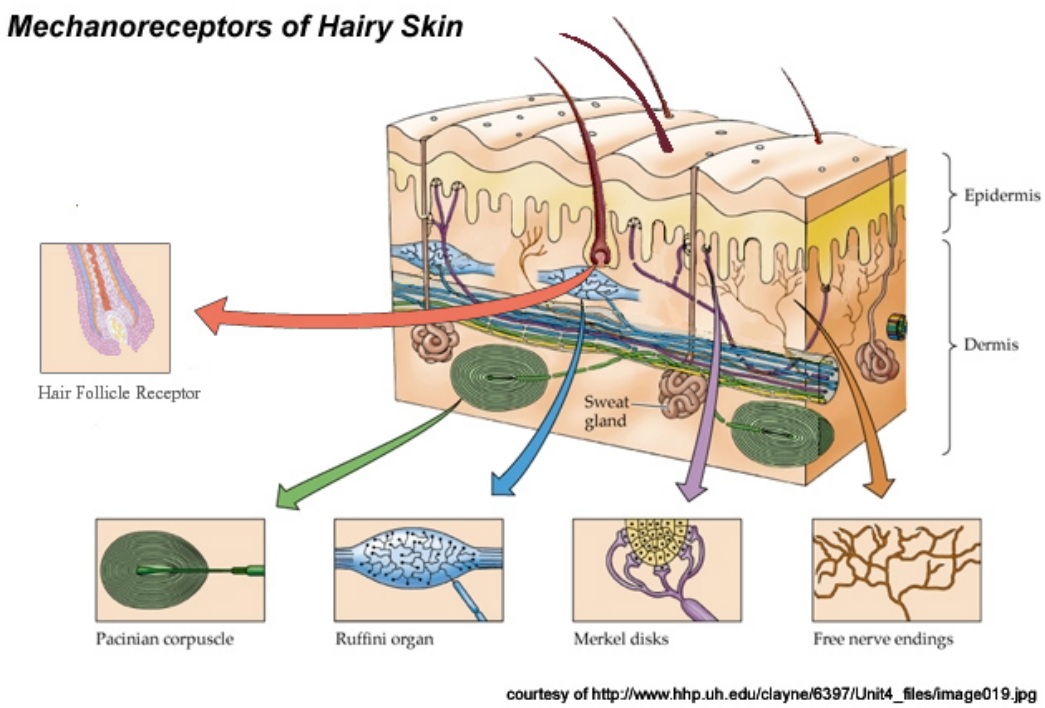

Figure 2.1: Mechanoreceptors found in hairy skin. Image taken from [72].

Receptor	Receptor Type	Median Field Size (mm^2)	Sensed Parameter
pacinian corpuscle	FAII	n/a	vibrations
hair receptor	FAI	113	hair displacement
ruffini ending	SAII	1.4	skin stretch
merkel disks	SAI	11	pressure and texture
field receptors (not pictured in Fig.)	FA	78	skin stretch and joint movement

Table 2.1: Characteristics of mechanoreceptors found in human forearm skin [107]. Sensed parameters suggested by Gilman 2002 [45].

sources (either to create a new sense or to partially replace a lost one) is often referred to as sensory substitution [73]. An effective sensory substitution device should have good information bandwidth and be as intuitive as possible.

A number of devices have been developed that stimulate cutaneous receptors in some way. The vast majority of them use either vibration (which is discussed in depth below) or electrical stimulation. Most mechanoreceptors (or adjacent afferent nerve fibers) can be artificially stimulated by injecting an electrical current into the skin. Electrotactile stimulation devices are designed to communicate information with electrodes that are placed on the skin. The current threshold required for a mechanoreceptor or nerve to respond (i.e., fire an action potential) depends on the depth under the skin and the diameter of the nerve fiber. Superficial and large diameter nerves have the lowest thresholds. In most embodiments, electrotactile displays stimulate a large variety of sensory responses including pressure, vibration, temperature and pain, creating a stimulus that is difficult to describe and is often uncomfortable [63, 64]. There have been some attempts to control the stimulus properties to selectively stimulate certain receptors [65]. One problem with cutaneous electrical stimulation is that the thresholds for stimulation change (due to skin conductivity changes) as the moisture content of the skin changes (e.g. due to sweating). This makes it difficult to control the stimulus for long periods. An electrotactile array called the Brainport has been created for use on the tongue due to the high receptor density and relatively constant moisture content [26]. While electrotactile displays have been effectively used in some experiments for sensory substitution [106, 113, 114, 116, 117], We do not consider them further here due to their subjective unpleasantness, relatively high power consumption [64], and inconsistency on hairy skin due to sweating as most of the expected applications will involve physical activity of the users. In the following sections we will discuss vibration, which is the most common form of tactile display for portable applications, and skin stretch which is a new alternative that has some unique advantages.

2.1.1 Vibration

A rapidly changing stimulus applied to the skin stimulates the fast adapting mechanoreceptors (Meisner and Pacinian corpuscles). Meisner corpuscles are sensitive to lower frequency stimuli while Pacinian corpuscles respond to higher frequencies (see Figure 2.2). The Pacinian corpuscles are most sensitive to frequencies around 250 Hz and are thought to be most important for detecting vibrotactile stimuli [64]. Unlike other mechanoreceptors, for which the firing rate increases with stimulus amplitude, Pacinian corpuscles fire at the frequency of stimulation [39, 95]. This implies that amplitude perception must be coded by the number of receptors that are active or with other receptors. Because the Pacinian corpuscles are type II receptors (deep), they have large receptive fields. This makes localization of vibratory stimuli somewhat difficult (i.e., the stimuli must be relatively far apart, depending on the skin area).

Extensive research has been completed to characterize vibration perception, studying the effects of stimulus waveform, contact area size, frequency, amplitude, and various other factors. Jones and Sarter provide a comprehensive summary of relevant findings [62], covering the last 60 years of research on vibrotactile stimuli to aid in designing tactile displays.

Vibrotactile stimuli are characterized in terms of the frequency and amplitude of stimulation. Variations in results are evident and are attributed to differences in experimental conditions and methods, choice of stimulus waveform, and size of contact area [43, 59, 77]. In general humans are most sensitive to frequencies in the 150-300 Hz range, though it varies slightly depending on the region of the body where the stimulus is applied [62]. For example, humans are most sensitive at the fingertips, less sensitive on the forearms, and least sensitive in the abdominal and waist area. This trend is even more pronounced in amplitude detection, where thresholds vary considerably more. The amplitude for detecting vibration at any frequency varies from 0.07 μm to 4 μm [111].

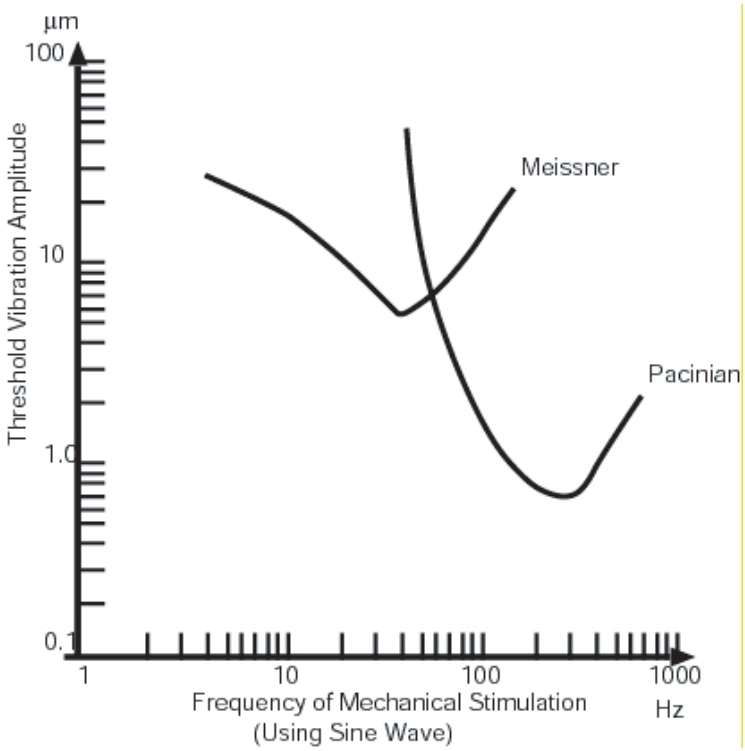

Figure 2.2: Frequency response of fast adapting receptors. Image taken from [58].

Vibrotactile difference thresholds have been studied considerably less than detection thresholds due to the difficulties in modulating changes in frequency and amplitude. As the amplitude of vibration increases, even though the frequency is constant, there is a perceived increase in frequency [81]. In addition, perceived frequency varies greatly across different regions of the body. Jones and Sarter produce a chart summarizing the findings of several studies to determine vibrotactile difference thresholds (see Figure 2.3). Here it can be noted that differences in measurement technique and environment may cause variation in threshold results, as there is a large discrepancy in difference thresholds at the finger. Frequency discrimination thresholds are typically presented as a normalized function of the reference frequency, $(\Delta F)/F_{ref}$, where (ΔF) is the frequency difference threshold in Hz, and F_{ref} is the reference frequency.

Frequency difference thresholds for the forearm range from approximately 0.2 to

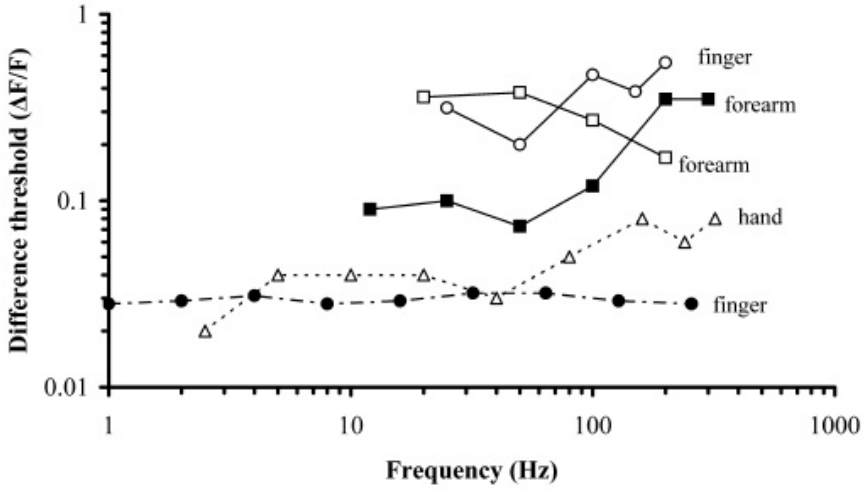

Figure 2.3: Vibrotactile difference thresholds as a function of frequency for pulses delivered to the forearm (filled squares, Rothenberg [94] ; open squares Mahns [77]), the finger (filled circles, Franzen & Nordmark [34]; open circles, Goff [46]), and the hand (triangles, Mowbray & Gebhard [82]). Figure taken from Jones [62].

0.4, where the thresholds are higher at lower frequencies [77]. Changes in amplitude can also be used to vary vibration intensity. Though there are variations in data, in general, amplitude based difference thresholds decrease with increasing stimulus intensity [59].

These perception studies typically used large actuators that have fast dynamics such that the amplitude (either force or position) and frequency of stimulation can be independently controlled. These actuators are obviously not practical for portable applications. For wearable devices, two types of actuators are commonly used to create vibrotactile stimuli. The most common type uses a simple DC rotational motor with an unbalanced inertia on the output shaft. In this case the frequency and amplitude of vibration are coupled (with amplitude approximately related to the square of frequency) and are determined by the voltage sent to the motor. The second class of motors used to create vibratory stimuli are simple linear actuators, including voice-coil and piezoelectric types. With these actuators, the waveform sent to the motor can be specified such that frequency and amplitude are theoretically

independent. However, most actuators designed for wearable use, particularly on the hairy skin, have mechanical dynamics that couple frequency and amplitude. In the studies presented in this work, we use a C2 actuator from EAI Inc (Figure 2.4). The frequency response of this actuator when placed on the skin is shown in Figure 2.5. These data were obtained by placing an accelerometer (Crossbow LP series) above the tactor which was strapped to the arm. A sinusoidal signal with a frequency that swept from 50 to 450 Hz over 5 seconds was sent to the tactor through a linear current amplifier (LM675) at a constant amplitude. The acceleration in the direction perpendicular to the skin was measured as the output at 2 kHz. The empirical transfer function estimate was then obtained by dividing the discrete Fourier transforms of the output and input. A clear resonance is evident around 250 Hz. Non-linear and non-monotonic magnitude coupling is found around this resonance. The device was tuned to the peak sensitivity of Pacinian corpuscles. In order to produce sufficiently large vibration amplitudes for wearable devices, most vibrotactile displays are tuned to a particular frequency. In the studies we present here that use vibration, we will therefore vary only the amplitude of the vibration while fixing the frequency at 250 Hz. The peak-to-peak acceleration produced by the C2 tactor was also measured for various amplitudes at 250 Hz and an approximately linear relation was found up to the current limit of the tactor. At this level, the tactor produced a peak-to-peak acceleration of 7.5 g when strapped to the skin.

Vibration on the fingertips has been used to provide a sense of force feedback [85]. On hairy skin, where receptor density is lower but there is more area to work with, arrays of vibration devices are often placed to create a sense of motion or direction [61,75,92,104]. The number of tactors needed to create a sense of continuous motion can be reduced due to a perceptual illusion called sensory saltation or the "cutaneous rabbit" [41, 42]. This illusion can be created by placing at least three tactors linearly on the body and presenting a few short stimuli at the first, then a few at the second etc. The person will often report that the stimuli were evenly distributed spatially [21, 104]. Interstimulus intervals should be between 20 and 300

Figure 2.4: C2 Tactor (EAI Inc).

ms (ideally about 50 ms). On the back, tactors should be located no more than 100 mm apart and three to six stimulus bursts per tactor are common [21]. Care must be taken with vibrotactile stimulation to avoid desensitization with prolonged use as fast adapting mechanoreceptors generally stop firing due to sensory adaptation after a relatively short period of continuous stimulation [10, 55, 88].

In summary, vibrotactile stimulation is a common choice for wearable tactile feedback due to its small size and ease of implementation. However the resolution is relatively low, particularly on hairy skin and is further limited by amplitude and frequency coupling in most devices as well as how many stimulus levels a person can remember. It is most commonly used to convey "event-cue" feedback (a single bit of information) or for some sense of localized feedback. For instance, if it was desired to instruct a subject to move their knee inward, a tactor on the outside of the knee could be used to provide this cue intuitively. The ability to localize is limited by the large receptive fields of the Pacinian corpuscles. Many users report is as annoying with prolonged use and it can lead to relatively fast desensitization. Vibration is most appropriately used when space is limited and/or many actuators with a low

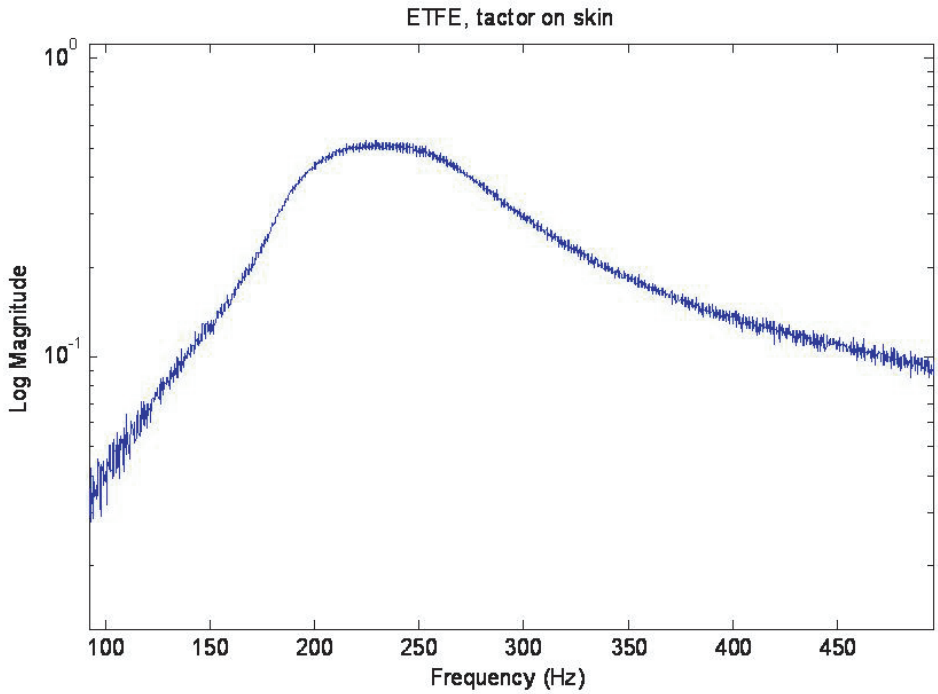

Figure 2.5: Empirical transfer function estimate (EFTE) for C2 tactor placed on skin. Transfer function is from input voltage to acceleration in direction perpendicular to skin surface.

resolution are needed.

2.1.2 Skin Stretch

In comparison to vibration, skin stretch can be used to activate slow-adapting as well as fast-adapting mechanoreceptors. Using skin stretch at low frequencies is attractive for wireless devices as it does not require much power; movements are small and velocities can be low. It has been shown in previous research that mechanoreceptors respond quickly and accurately to skin strain changes [29, 31, 89], and that humans are more sensitive to tangential forces than normal forces on the hairy skin of the forearms [11]. However, in comparison to vibration, few devices utilize skin stretch. Important exceptions include the work of Hayward and colleagues [52, 109]

and [11,31,37,71,74,83,89] who have developed fingertip (i.e., glabrous) displays that include skin stretch. Several investigators [31, 71, 74, 83] have also studied the mechanisms behind skin stretch. Makino [78] has developed a suction-based display that produces illusions of pressure on the skin, at least in part by producing localized skin stretch. However, non-glabrous skin stretch displays have been largely unexplored. Skin stretch is known to contribute to motion sensations at various joints [25, 29, 30] and elastic bandages placed on joints can enhance proprioception [90] due to increased cutaneous sensations.

In this section we will provide an overview of the devices used in these studies as well as some previous studies that established some of the perception and performance qualities of the stimuli. The majority of the design and performance testing was performed by Karlin Bark and much more detail can be found in her thesis [8].

Device Design

There are myriad ways of applying skin stretch. A single point can be moved linearly in one or two directions and can also rotate. Multiple points can also be used that move together or individually. A number of pilot studies were performed to determine the type of stimulus that produced the greatest dynamic range and resolution of feedback, with a desire to create a sense of motion as well as magnitude. It was found that two circular contact points (d=13 mm) spaced about 25 mm apart, that rotate about a central axis, produced a good sense of position and motion across a range of about ±45 degrees [8]. A benchtop device was created that produced the desired motions using a brushed DC motor with a capstan cable transmission (Figure 2.6). The device has a number of manually adjustable degrees of freedom to allow accurate placement on the skin and also includes a six-axis force sensor which allows monitoring of force and torque readings during motions. This device was used in the study presented in Chapter 4.

While the benchtop device established the effectiveness of the stimulus, a wearable

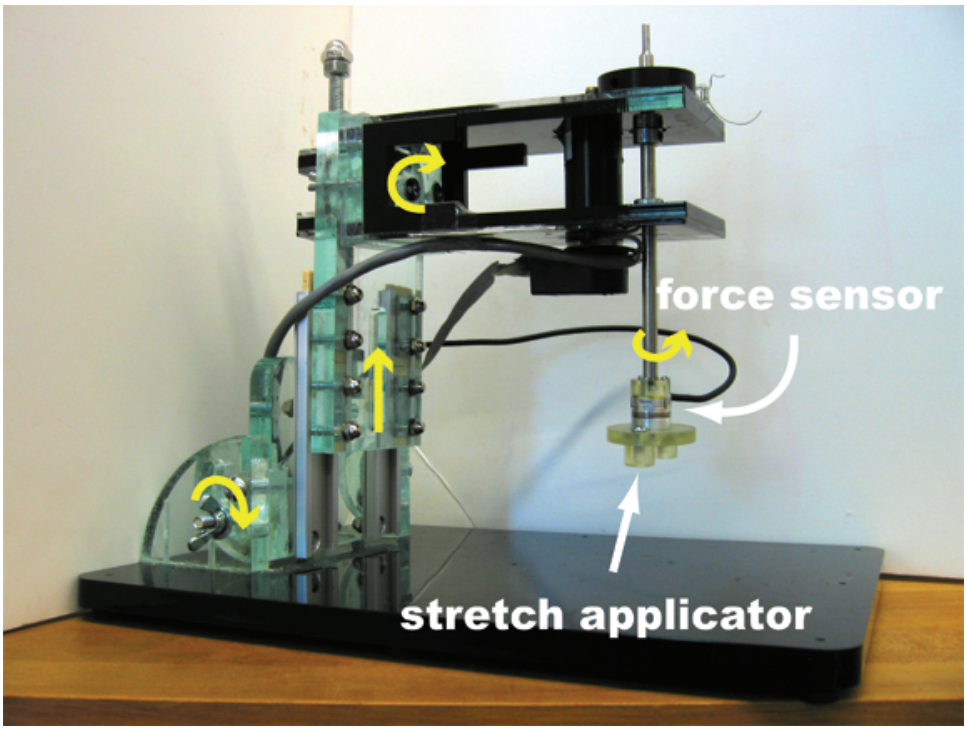

Figure 2.6: Benchtop skin stretch device with degrees of freedom labeled.

version was needed to meet the requirements of this work. In addition to being wearable, the device needed to be potentially portable, meaning that it should consume minimal power and have no components that would not transfer to portable use. It was also desired to create a motion that had very little vibration in the perceivable range so as to not confound the skin stretch stimulus for perception studies. Conventional DC motors were not ideal for this application as they needed to either be large enough to produce torques in the 0.5 N-m range (based on data from the benchtop device) or have a high transmission ratio. It was not desired to use a gearhead on the motor as these produce significant vibrations. A capstan pulley of appropriate size would have made the device prohibitively large. Finally, for portable use, a non-backdriveable actuator is preferable as it does not require a current to hold a position against an external torque.

A small piezoelectric rotational motor was identified that produces torques of up

to 0.1 N-m in a small package and is not backdriveable (Shinsei Motors USR-30). Additionally, it produces minimal perceivable vibration due to the high resonant frequency of the piezo-crystals. The specifications of the actuator are presented in Table 2.2.

	Motor Specifications
Diameter	30 mm
Thickness	9 mm
Maximum Torque	0.1 Nm
Holding Torque	0.1 Nm
Weight	20 g
Maximum Speed	150 rpm = 900 deg/s
Minimum Speed	15 rpm = 90 deg/s
Driving Frequency	50 kHz

Table 2.2: Ultrasonic motor specifications

An additional transmission ratio of approximately 5:1 was required to achieve the desired torque. This was accomplished with a capstan cable transmission. The actuators and transmission were packaged with an encoder for position feedback into the device shown in Figures 2.7 and 2.8. The specifications of the device are shown in Table 2.3. A newer, smaller version of the same device has also been designed with similar performance characteristics (Figure 2.9).

	Design Requirements	Device Specifications
Size	small	29 x 45 x 126 mm
Max Torque	0.2 Nm	0.6 Nm
Speed Range	$\leq$ 200 deg/s	15-150 deg/s
Weight	$\leq$ 200 g	115 g
Sensor	1 deg	1 deg (Hall Effect)
Resolution		0.05 deg (Encoder)

Table 2.3: Wearable skin stretch device characteristics

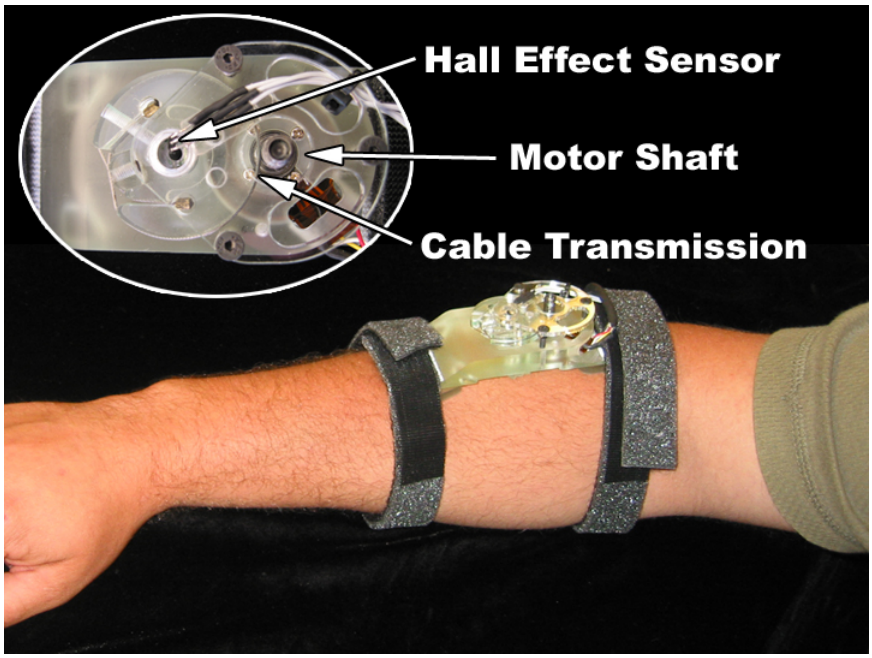

Figure 2.7: Wearable skin stretch device on arm.

Perception Tests

Two perception tests were performed with the skin stretch devices placed on the forearm. The first established difference thresholds for the device. In this study the benchtop device was used to present various stimuli in succession (three at a time) and the subject was asked to report the one that was different from the other two. The test stimulus was always larger than the reference stimuli. An adaptive three-up, one-down method was used to converge upon the just noticeable difference (JND). For details see [8]. The experiment was performed at two reference positions (10 and 30 degrees) and two speeds (fast and slow). The motion always started from zero degrees of rotation. The JND results for the twelve subjects tested are shown in Figure 2.10.

The normalized JND was significantly lower at the larger reference stimulus, which is intuitive as the torques and strains increase non-linearly with rotation. This non-linearity is accounted for by creating a non-linear mapping of skin stretch rotation to

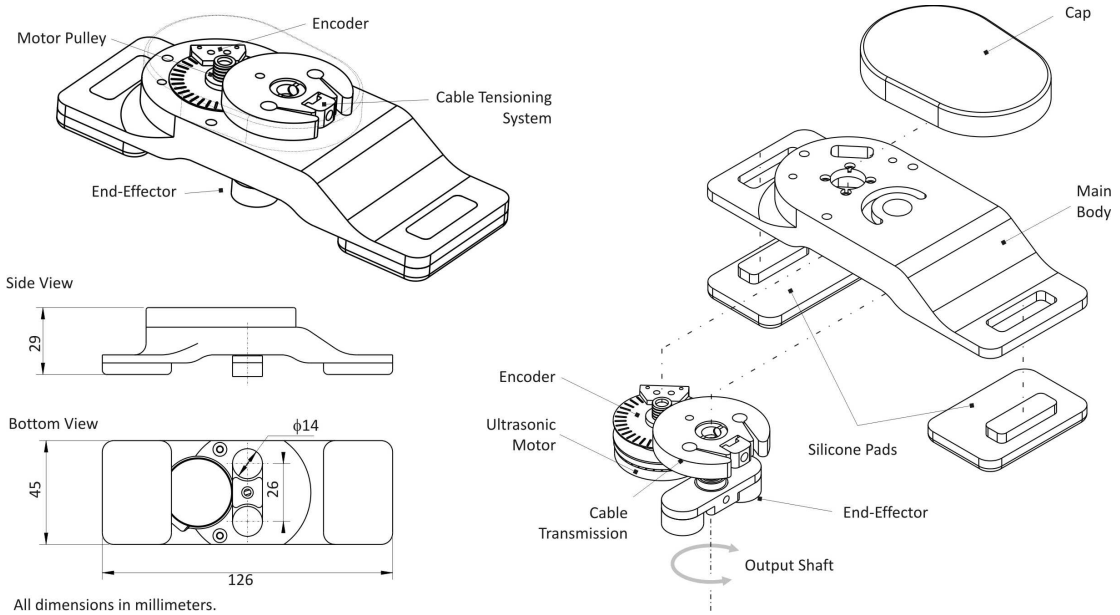

Figure 2.8: Wearable skin stretch device assembly layout and main dimensions.

virtual object position in some of the studies described in later chapters. The speed of the stimulus had no significant effect on the JND.

The second perception study tested how well subjects could linearly map the skin stretch feedback to the motion of a virtual object across the entire comfortable range of stimuli. The wearable skin stretch device was used in this study. Its rotation, along with the virtual object, was controlled by a rotational knob (see Figure 2.11). The virtual object was displayed on a monitor and was designed to look like the end-effector of the skin stretch device (Figure 2.12). The subject was given training in which the device was rotated together with the virtual object display for a few minutes. During this phase, they were also allowed to set the desired range of motion of the device.

In the experiment, two types of tasks were evaluated. In the first (called the active task) the desired position of the virtual object was displayed and the subject rotated the knob to move the skin stretch device until the perception matched the desired position. In the second task (passive) the skin stretch device was autonomously rotated

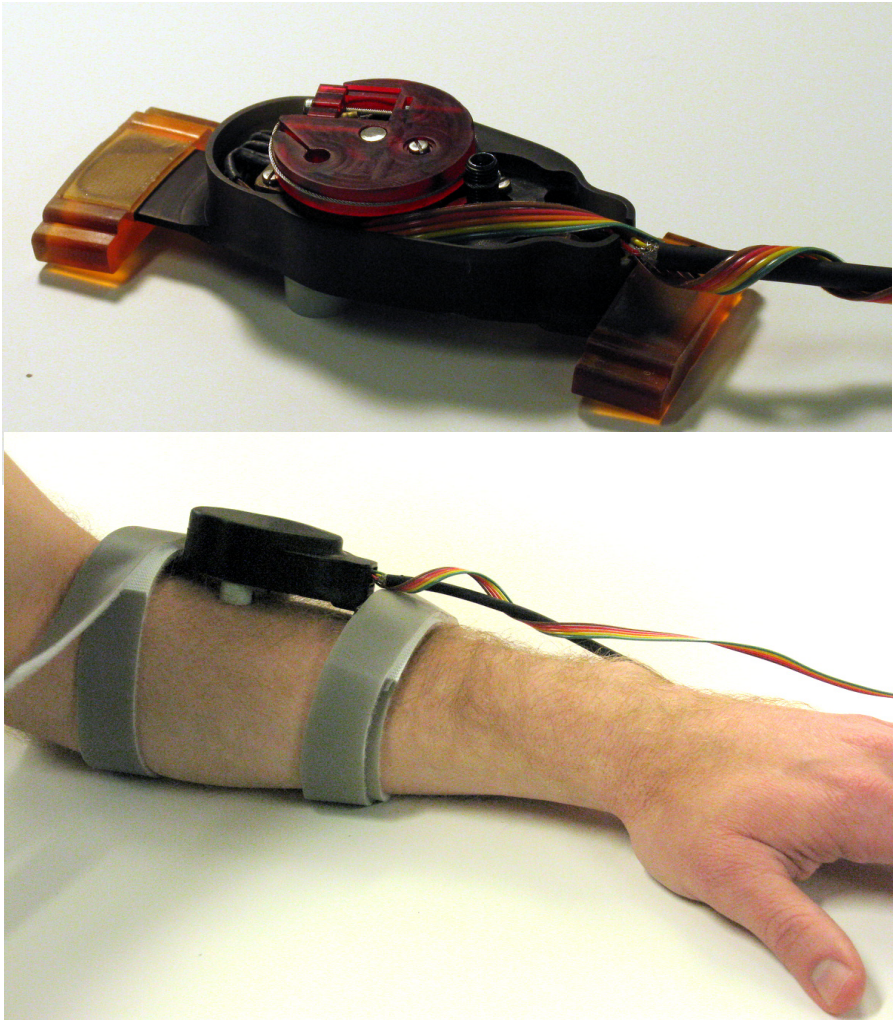

Figure 2.9: Most recent version of the wearable skin stretch device.

to a desired position and the subject moved the virtual object on the screen with the knob until it matched their tactile percept. In the active task, an afferent/efferent feedback loop was present that allowed the device to be dynamically positioned as needed. In the passive case, only a static sense of position was provided (after the motion was completed) and the feedback loop was visual. In both tasks, two end-effector configurations were tested. The "fixed" configuration refers to the case where the contact points were constrained to rotate with the end-effector. In the "free"

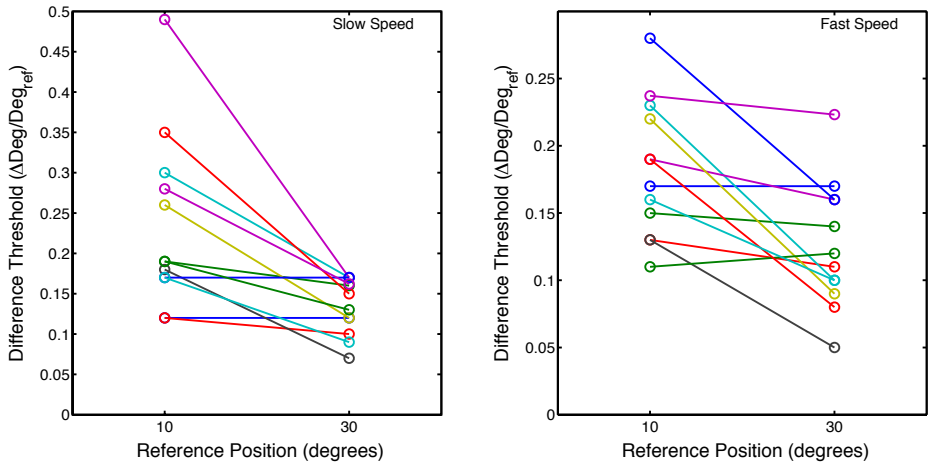

Figure 2.10: Results of difference threshold study for 12 subjects at slow (left) and fast (right) speeds. The difference threshold was lower at the larger reference position but speed had no significant effect (From [8]).

condition, the contact points could rotate about their own axes.

The results for ten subjects in the active and passive studies are shown in Figures 2.13 and 2.14 respectively. In the active case, subjects were able to effectively use the device to linearly map the orientation of the virtual object with an average residual of around 6 degrees. In the passive case however, the performance was much poorer. A number of direction errors were evident (points in quadrants 2 and 4) that are indicative of the subject having a sense of magnitude but not direction. These errors were reduced in subjects who had prior experience with the device, indicating that additional training could have potentially improved performance [8]. There was little difference in performance with the two end-effector configurations.

These results indicate that skin stretch feedback may be most effectively used when the user has either direct or indirect control over its movement. When the device is moved autonomously, there is some directional ambiguity. It also seems that the dynamic components of the stimulus (i.e., the sense of motion) are important for position perception. It is difficult to decouple the importance of static and dynamic

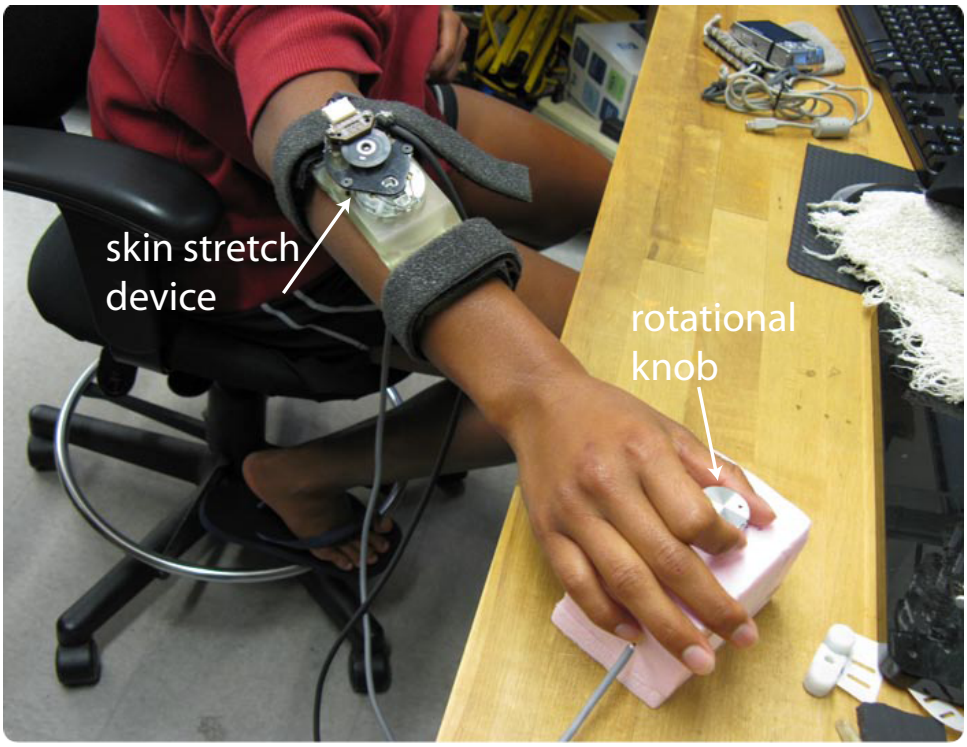

Figure 2.11: Setup for performance study with device and virtual object controlled by a rotational knob.

stimulus information from these results and it is similarly difficult to design a study that decouples them as it is not possible to have a step input of position (there is a dynamic *and* static component of the stimulus). However, we can infer from the improved performance in the active case that the ability to move and control the dynamic properties of the motion did improve performance. Subjects could use various strategies to reach the desired position. For example, they could move to a reliable point in the workspace (e.g. near the limits of the range) and the move back to an area that is perhaps less sensitive. Even though the gain of the knob rotation to device rotation was varied to avoid using finger proprioception to position the device, the subjects could potentially remap the workspace (relying solely on tactile feedback) by moving across the range in each trial if they desired. Additionally, direction errors are eliminated as the subject rotated the knob in the desired direction and assumed

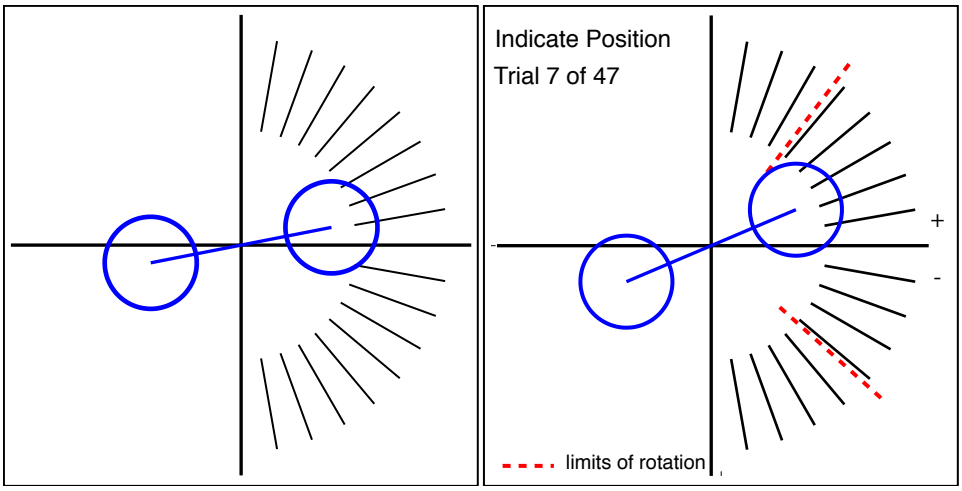

Figure 2.12: Display shown to subjects during trials. The left is the screen displayed during the training phase (showing the actual location) and on the right is the desired location as shown in the experiment trial. Dashed lines indicate the limits of rotation specified during the training phase.

that the device rotated in the correct direction.

2.1.3 Choice of Feedback Modality

Based on the discussion of vibration and skin stretch above, we can begin to make some conjectures about which type of feedback might be most appropriate for a certain application. Vibration is certainly simpler to implement and is effective for conveying exogenous event-cues or perhaps a small number of levels of information. It can be localized to some extent across body segments. Its prior use for vests and suits (e.g. see [61]) seems appropriate. Skin stretch on the other hand, has the ability to convey a sense of motion and position with a single device (which is larger than a typical vibration actuator but has a smaller footprint than would be required to array multiple actuators). It can also convey directional information, which vibration cannot do with a single actuator, but ambiguity in direction is possible if the stimulus is exogenous. Skin stretch can convey a fairly rich sense of motion in a small package but seems best suited for applications where the user has at least some control of the motion.

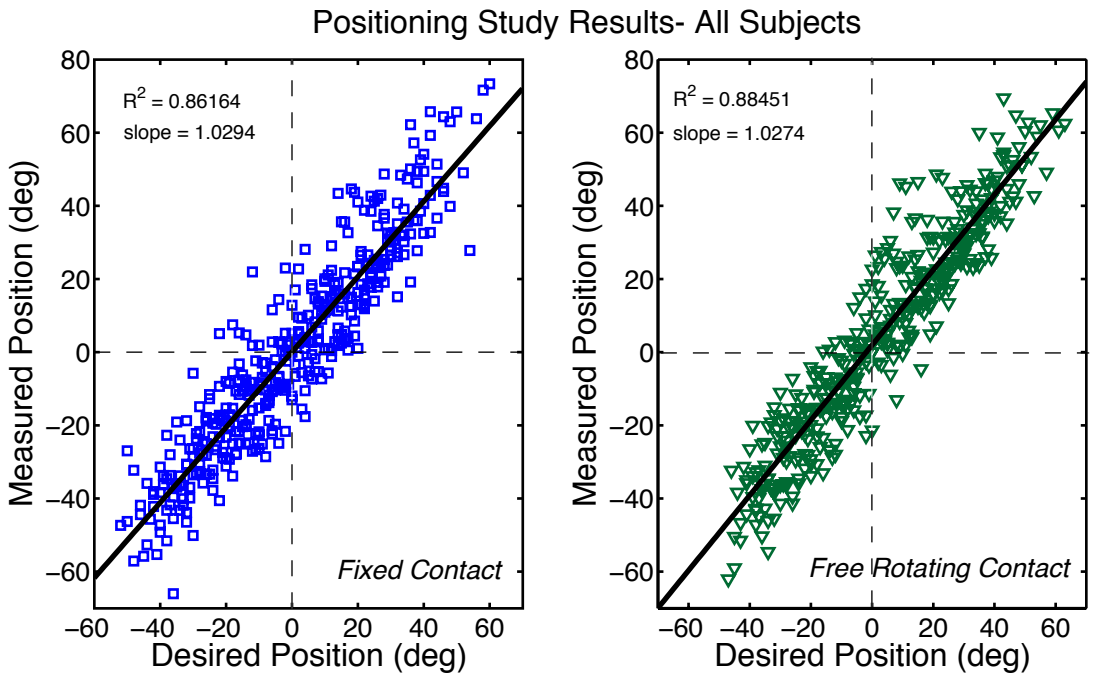

Figure 2.13: Data from all ten subjects in the active task. Two end-effectors (fixed and free) were tested and minimal differences were found in performance (From [8]).

This reduces the risk of direction errors and allows for compensatory strategies for positioning that reduce the insensitivity at small rotations. The perception results indicate that subjects can learn to uniformly map a workspace with minimal training with the device.

In the following chapters, we will use vibration for conveying coarse feedback of performance (Chapter 3) and skin stretch for creating a sense of position and motion of a virtual object. In Chapter 4, we will use both skin stretch and vibration to compare performance for positioning a virtual object.

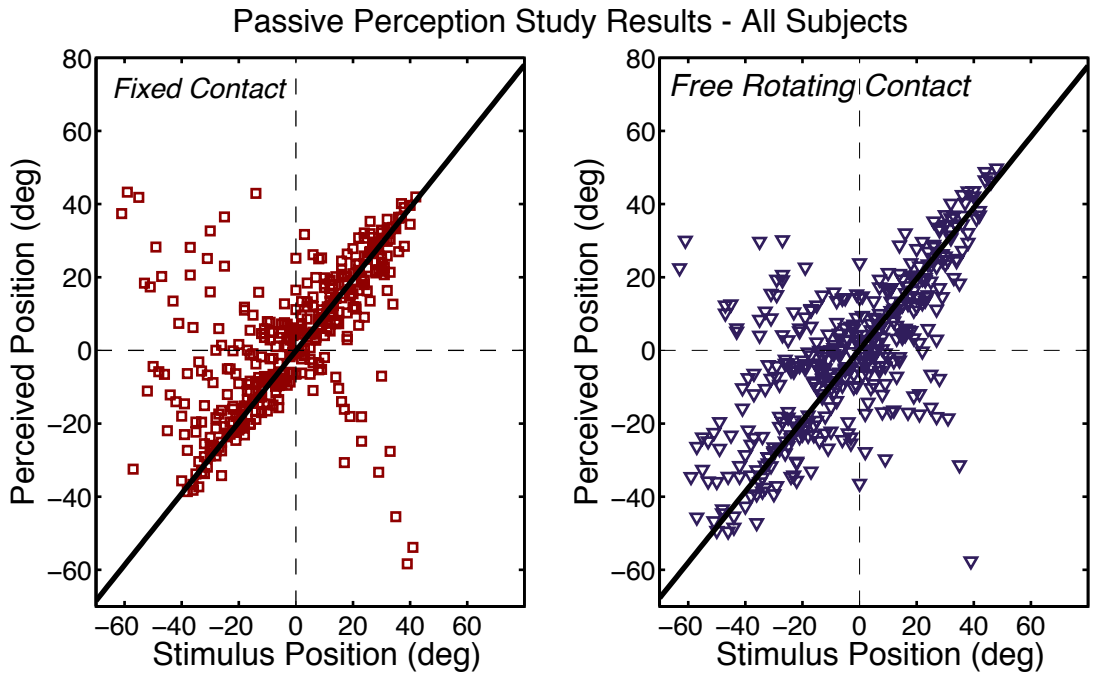

Figure 2.14: Data from all ten subjects in the passive study. Two end-effectors (fixed and free) were tested and minimal differences were found in performance (From [8]).

2.2 Proprioception and Kinesthesia

A standardized definition of the terms proprioception and kinesthesia is difficult to find in the literature. The use of these terms seems to be context dependent. Most commonly, proprioception is used to describe the sense of position and motion of body segments that is not due vision or the organs of equilibrium [15,39]. Kinesthesia has a similar meaning but is typically used to refer to motion rather than position. Kinesthesia also sometimes has a connotation of force. Proprioception and, to a lesser degree kinesthesia, are also sometimes used to refer to the low-level afferent feedback from sensors in joints and muscles (which in fact are often called proprioceptors). This can create come ambiguity in the use of these terms. We will use the term proprioception to refer to a gross (i.e., joint or endpoint level) sense of position and motion. We will use the term kinesthesia sparingly and only in reference to others' use

of the term in prior work. In all cases, it refers to a joint-level sense of motion. When referring to low-level afferent feedback, we will avoid use of the terms proprioception or kinesthesia to reduce ambiguity.

A number of sensory receptors contribute to the proprioceptive sense. The muscle spindles and Golgi tendon organs are sensitive to position, movement and force in muscles. Sensors in the joints give a sense of flexion and extension. Mechanoreceptors in the skin, including Ruffini endings and Merkel cells, also contribute to the sense of motion and position [39]. The brain integrates this afferent information to create a percept of the body segments' position and orientation.

The relative contribution of the various sensors and how the information is integrated by the CNS is not well understood. It is difficult to study proprioception as there are many contributors and they cannot easily be individually eliminated. Vibrotactile stimulation and skin stretch have been used to create illusory movements at various joints. These phenomena are often used to evaluate how different sensors contribute to kinesthesia [60]. Providing vibrotactile stimulation of about 75-100 Hz can create an illusion of tendon lengthening [47]. These illusions are created by muscle spindle activation and for many years it was thought that these dominated proprioceptive feedback based on this observation. However recently, skin stretch near the joints has also been found to create an illusory sense of motion [29, 30]. Collins et al. [25] found that skin stretch contributed to illusory movements at the index finger, elbow and knee and evaluated the relative magnitude of the perceived movements for various combinations of tendon vibration and skin stretch. Gandevia et al. showed that efferent signals contribute to proprioception [38].

Regardless of how the sense is created, the value of proprioception is unquestioned. Even when vision is present, proprioceptive feedback can improve qualities of targeted finger movements [70]. It has proven more effective than vision for stiffness discrimination [51]. The importance of proprioception is further evidenced by the tragic and thankfully rare cases of people who suffer from neurological conditions that eliminate this sense. Ian Waterman is one example of such a case [1]. He described his struggle

to do simple daily tasks using only visual feedback as a "daily marathon" [24].

Amputees who are fitted with robotic limbs no longer have proprioceptive feedback. They can use vision to position the limb, however. Many upper-limb jointed prostheses have been designed that are controlled by moving other joints (e.g. the shoulders). This allows amputees to have a sense of the position of the limb from receptors in the joints used to control the movement. The concept of creating a percept of an external object based on body motions is referred to as Extended Physiological Proprioception (EPP) [102]. Some more advanced upper-limb prostheses have been developed that are controlled with EMG sensors on residual muscles. This allows for less cumbersome and more intuitive control but no proprioceptive feedback is provided. This topic is revisited in more detail in Chapter 5.

2.3 Models of Feedback in Motor Control

How humans control movements of their limbs (and external objects) is the subject of much research and debate. A thorough review of this field is outside the scope of this document. However, we will provide a summary of some of the major theories while placing emphasis on the role of feedback (visual or proprioceptive) in motion control. It should be noted that the actual neural circuits that govern the motions are extremely complex and most of the work in movement science attempts to explain the gross behavior of this circuitry with simplified models taken from control theory.

As stated in the introduction, human motor control cannot be explained by a typical feedback control model due to the large delays associated with feedback. The fastest spinal reflex loops have delays of about 50 ms and visual feedback delays for arm movements can be as large as 250 ms [68]. The bandwidth of human motions is such that delays of this magnitude would cause instability if the feedback gains were not small. This has led researchers to propose other theories that explain how we control our movements. One such theory, called equilibrium-point control [14,32,33], theorizes that the brain simply sends a series of equilibrium points to the body to

create a trajectory. This was based on the observation that peripheral and low-level mechanisms such as reflex loops and the passive properties of the tissues tend to stabilize the joints around a point. In this theory, low-level feedback from muscle spindles etc. is used to maintain the actuators in certain regions where performance is stable and sufficient force generating capacity is available to compensate for external disturbances.

An alternative hypothesis that has gained prevalence in recent years theorizes that the brain (specifically the cerebellum) contains internal dynamic models of the body that allow efferent signals to be generated in a feed-forward fashion based on a desired motion [68, 80]. Many of the proposed control structures contain both an inverse model (for generating the motor command from a desired trajectory) and a forward model that is sent an "efference copy" and generates expected sensory feedback with a delay that is less than the actual feedback. The expected sensory feedback can be compared to the actual feedback (after a delay) to update the motor commands in a stable and robust way [80]. An example of the type of structure that is often proposed is shown in Figure 2.15.

Given this background, we can begin to hypothesize about the role of feedback in motor control. It seems that at a low level, there may be some degree of actuator management. However at a high level, which is the focus of this work, feedback can have a few roles. It almost certainly serves to update motor commands at slower rates as it is compared to expected sensory feedback from the forward model. It is also likely used to create the forward and inverse models so that future tasks can be executed as desired. There is some debate over whether these models are generalized to all tasks or are task specific [68].

Based on this discussion, the haptic feedback we provide should be focused on one of a few cases, as introduced in the previous chapter. It is useful when the motions are slow or the models are unreliable. It may also be useful for repetitive tasks or generalized learning. We explore repetitive learning in Chapter 3 but do not address generalized learning, though it could be an interesting study.

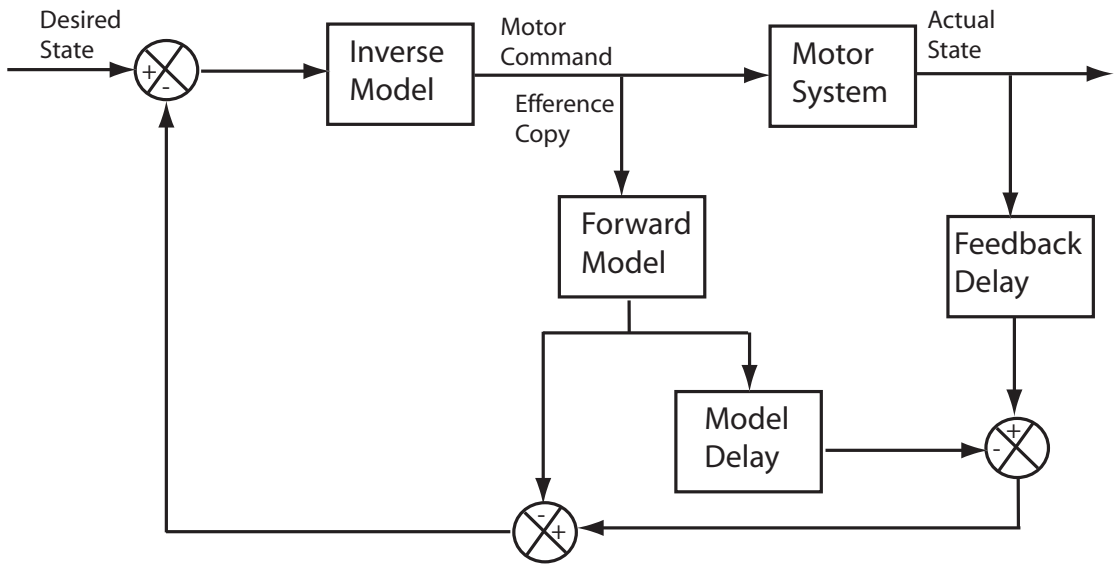

Figure 2.15: One model of motor control with forward and inverse models to compensate for feedback delays (Smith predictor [80]).

2.4 Feedback for Motion Guidance

We can finally look at the prior studies that have provided feedback to subjects to guide their motions. The importance of feedback in motor learning of a task is well established [12, 13, 17] with results suggesting the need for specificity and immediacy of the feedback [3]. Feedback can be given visually or verbally from a teacher but this introduces long delays and some information can be difficult to communicate with these means. Virtual reality displays can show desired motions and errors between actual and desired trajectories. However, 2D displays are not sufficient for displaying full 3D motions. Tactile feedback presents an intuitive way to guide motions due to the ability to localize feedback and minimize delays. This feedback can be given manually by a teacher but a person can only apply feedback at a small number of locations at a time. The use of tactile displays is a reasonable approach to provide localized, real-time feedback. However, due to difficulties in tracking motions in real-time, only two very recent papers have attempted this.

Bloomfield and Badler [16] placed vibrotactile devices on the upper and lower right arm and evaluated performance in a collision detection task compared to visual feedback. A motion capture system was used to track arm motions. More relevant to this work, they also tested the ability to train subjects to perform a new motion (martial arts poses) with the tactile feedback. In fact the training was successful for many motions (which were slow) but no comparisons were made to cases where no tactile feedback was given. The authors state that these studies were expected to be pilot studies but the results were included in the paper after examining the results.

Lieberman and Brazeal [75] also used motion capture and vibration on the arm. They evaluated the accuracy of arm motions (5 degrees of freedom (DOF) with the elbow on a table). Three of the DOF were hinge joints (elbow flexion/extension and wrist flexion/extension and ad/abduction) and the other two were rotary motions (upper and lower arm rotation). They found significantly lower errors and faster training with the tactile feedback than with 2D vision alone. Results were best in the hinge joints as the rotational vibration feedback may not have been easily interpreted. Again, the motions in this study were slow. The authors claim potential applications for a variety of situations but it is not clear how the training will translate based on the discussion above about feedback delays.

The work presented in the following chapters seeks to improve upon prior studies by using a new modality of feedback (skin stretch) to provide proprioceptive feedback and by using feedback for motion training of a very practical, repetitive task (walking).

Chapter 3

Feedback for Motion Guidance

In this chapter we describe an experiment in which we provide real-time biofeedback of a single biomechanical parameter to allow subjects to alter their gait pattern. As discussed in detail below, it is desirable to lower this parameter to prevent a common type of osteoarthritis. Feedback is provided with a visual display, which has the ability to convey the parameter with high resolution and can also display a time history from previous steps, to half of the subjects. The other half are given vibration feedback, which has a lower resolution and no time history but could potentially be integrated into a portable system for use outside the laboratory. This study was performed in collaboration with Pete Shull and Prof. Thor Besier.

3.1 Background and Motivation

Knee-joint osteoarthritis (OA) is a significant public health issue that affects millions [87]. The most common form is referred to as medial compartment OA and is distinguished by a reduction in cartilage thickness on the medial (inside) portion of the tibia-femoral joint. The disease is thought to be caused by non-uniform loading of the joint over time [6]. However, measuring these loads *in vivo* is extremely difficult. An external surrogate measure called the knee adduction moment (KAM) has been

identified that has been correlated to the onset, progression and severity of medial compartment OA [6, 97]. The KAM is the moment applied to the knee in the frontal plane and can be calculated during gait with knowledge of the ground reaction force (GRF) vector, the location of the center of pressure (COP) under the foot and the location of the knee joint center. These values can be measured in a gait analysis laboratory with force plates and a motion capture system. The KAM typically exhibits two peaks during the stance phase of walking which correspond to the peaks in the vertical GRF [56]. Of these, the first peak is the only one that has been shown to be higher in OA patients [57, 84, 108]. It has been proposed that the first peak is therefore the more important one to lower [36].

Severe cases of knee OA require total joint replacement which is a costly and invasive procedure. Interventions can be performed in an attempt to slow the progression of the disease before joint replacement becomes necessary. The success of these interventions is often evaluated by comparing the KAM pre- and post-treatment. The most common surgical intervention to lower the KAM is a high tibial osteotomy (HTO) [108]. This procedure involves the addition or removal of a small piece of bone on the proximal tibia in an attempt to redistribute the medial-lateral loading on the joint. A good outcome of HTO is considered to be a 30-50% reduction in the peak KAM. Two non-invasive interventions include orthoses, such as lateral heel wedges or knee braces, and gait retraining. Lateral heel wedges shift the COP medially but have only been found to produce modest reductions (less than 10%) in the peak KAM compared to HTO [66, 67]. Gait retraining strategies seek to alter the patient's gait such that the KAM is reduced. A few general strategies have been proposed to reduce the moment. These include walking with the toes pointed outward, increasing trunk sway, taking shorter strides, walking more slowly and what is called a "medial-thrust gait". Walking with the toes outward can reduce the second peak but does not impact the first peak significantly [5, 6, 49, 108]. Increased trunk sway can reduce both peaks but can be unnatural [27]. Walking more slowly may not be acceptable to some patients and as with all generalized modifications, may not be

effective for all patients. Therefore, an individualized gait modification is desirable.

Fregly et al. have used dynamic simulations to predict novel gaits that reduce the first peak of the KAM and are subject specific [35,36]. While this has been effective for a small number of subjects, it is very time consuming and potentially non-intuitive to train novice patients to walk with the new gait. In the following sections, we describe an experiment in which we calculate the knee adduction moment of the left leg while patients walk on a treadmill. On each step, we give them real-time biofeedback of the first peak of the KAM with visual or vibration feedback. The subjects then iterate on gait modifications until a satisfactory gait is identified that produces a lower peak KAM. This approach uses implicit feedback as we do not specifically tell subjects how to modify their gait (though we do provide suggestions before the trial). Subjects are simply given performance feedback. This approach has proven effective for motor learning in other complex, repetitive tasks [76, 91, 112]. Due to feedback delays, subjects do not necessarily use the feedback provided on the current step. However, due to the repetitive nature of the task, they are able to use the information provided on the current step in planning future motions. This study is an example of providing a person a sense of a physiological variable that it is desirable to change, but the person has no natural sense of.

3.2 Methods

3.2.1 Experimental Setup

The experiment was completed in the Human Performance Laboratory at Stanford University. Motion capture information was collected with a Vicon 8-camera marker-based system (OMG plc, Oxford, UK). Reflective markers were placed on various body segments to allow reconstruction of kinematic information. Three marker clusters were placed on each of the subjects' feet and upper and lower legs. Four markers were placed on both the pelvis and torso. Additional markers were placed on anatomical

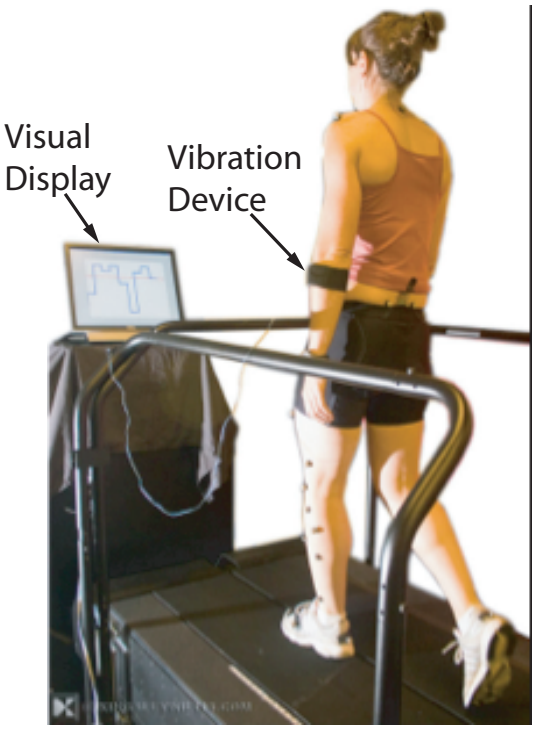

Figure 3.1: Experimental setup for knee adduction moment study. Reflective markers are used to capture motion data. Force plates in the treadmill gather force data. Feedback is provided with a vibration device or a visual display.

landmarks (i.e., femoral condyles and malleoli) but were removed after the static trial (with the exception of the left lateral femoral condyle marker). Two additional markers were also placed on the left lower leg to improve tracking in real-time.

Subjects walked on a split-belt, force-plate treadmill (Bertec Corp., Columbus, OH). The force plates in the treadmill provide GRF, moment and COP data. Force and marker data were collected and synchronized with Vicon Nexus software. The data were subsequently streamed via TCP/IP to Matlab (Mathworks, Natick, MA) where moment calculations were performed. Visual feedback was provided with a computer display in front of the treadmill. Vibration feedback was provided with a C2 Tactor (EAI Inc.) strapped on the left forearm (Figure 3.1). The visual feedback

was generated in Matlab and the vibration motor was controlled with the xPC real-time operating system (Mathworks, Natick, MA).

3.2.2 Calibration

After the markers were placed on the subject, they stood on the treadmill and a static trial was captured. Marker data from the static trial were used to build a model of the left tibia segment (consisting of the 5 tibia markers and the left femoral condyle marker as shown if Figure 3.2) so that it could be reliably tracked in real-time. The static trial data were also used in post-processing to locate joint centers and convert moments to the tibial coordinate frame. The left knee joint center was defined as the location of the left femoral condyle marker plus an offset in the negative y-direction (medial) in the lab coordinate frame that was equal to half of the knee width. The subject also reported their height and weight to normalize the joint moment ($Nm/(ht * wt)$).

After the model of the tibia segment was created, the subject completed a baseline walking trial. They were first asked to walk on the treadmill for 2-3 minutes at a self-selected speed to become comfortable with the device. We then recorded 15 steps of motion and force data and calculated the moment for each step. The moment about the knee joint, $\mathbf{M_k}$, was calculated as

$$\mathbf{M_k} = \mathbf{r^{k/COP}} \times \mathbf{GRF} \tag{3.1}$$

where $\mathbf{r^{k/COP}}$ is the position vector from the knee joint center to the center of pressure and **GRF** is the ground reaction force vector (see Figure 3.3). The knee adduction moment is the x-component of $\mathbf{M_k}$. All moment calculations during the experiment were performed in the lab coordinate frame, where the x-axis points towards the front of the treadmill. After post-processing the data, we converted the moment to the tibial coordinate frame to evaluate the adequacy of this approach. The average first peak of the KAM over the 15 baseline steps was defined as the

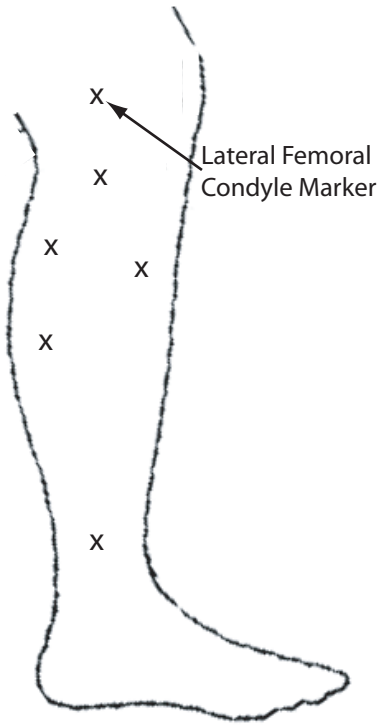

Figure 3.2: General location of tibia segment and knee markers tracked in real-time.

baseline KAM and the feedback we provided in the experimental trial was relative to this baseline. The first peak was defined by taking the maximum value of the KAM over the first 40% of stance phase. The stance time was determined from pre-trial walking data. Data from prior studies showed that the first peak always occurred before this point (e.g. see [56]) and this was validated by examining the data after the trial.

3.2.3 Experimental Trial

After the calibration and baseline trials, the details of the experiment were explained to the subject. The subject would walk at the same self-selected speed on the treadmill. They were told that they would be given feedback on each step (of the left leg) about a knee force that was desirable to minimize. Samples of the feedback (described

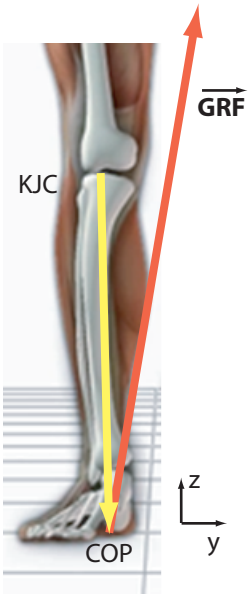

Figure 3.3: Vectors used for knee adduction moment (KAM) calculation. The KAM is the x-component of the cross product of the position vector from the knee joint center (KJC) to the center of pressure (COP) and the ground reaction force (GRF) vector.

below) were given prior to the experiment. The first peak of the KAM was calculated after 40% of stance and the feedback was provided as soon as possible after that (based on data collection and processing delays which were typically about 15% of the total stance time). The feedback was always presented before the left foot left the ground. Subjects were instructed to iterate on various gait modifications in an attempt to identify a new gait that produced a lower force but was still symmetric, sustainable, and acceptable. This meant that any modifications should be made on both legs, not result in unreasonably large increases in energy expenditure and not produce a gait that the subject would not consider using more permanently. We gave suggestions of some modifications to try based on prior studies. These recommendations were:

- walking with toes pointed in or out,
- increased medial-lateral trunk sway,

- taking longer or shorter strides (while adjusting stride frequency to maintain the same speed),
- loading the lateral or medial side of the foot,
- varying the distance between the feet (step width).

The subject was instructed to try various combinations of these (and any other) modifications until a satisfactory gait had been identified. We then allowed them to practice this gait for 1-2 minutes before capturing 20 steps of post-trial data. The first ten steps of this final trial were performed with the feedback device on, the final ten with the feedback off. The total time required to identify the new gait was recorded. The subject was also asked to report what strategies they chose for the final gait and to rate the awkwardness of the new gait compared to their normal gait on a 0-10 scale (with 0 corresponding to no different and 10 being extremely awkward).

Visual Feedback

Visual feedback was displayed on a screen in front of the treadmill. A stair-step plot of the current step's peak KAM as well as the peaks from the previous nine steps was shown. The baseline moment was displayed as a dotted line for reference. In addition to providing a time history of data, the visual feedback allowed data to be conveyed with high spatial resolution. These properties of the feedback allowed users to detect trends in the KAM based on the modifications they made.

Vibration Feedback

As discussed in Chapter 2, vibration feedback has a limited perceptual resolution. In pilot trials, subjects had difficulty distinguishing feedback amplitudes when a continuous scale (linear or logarithmic) was used. We therefore binned the feedback into three levels that were easily distinguishable (Figure 3.5). This put a fundamental limitation on the resolution of the feedback but reduced ambiguity in remembering

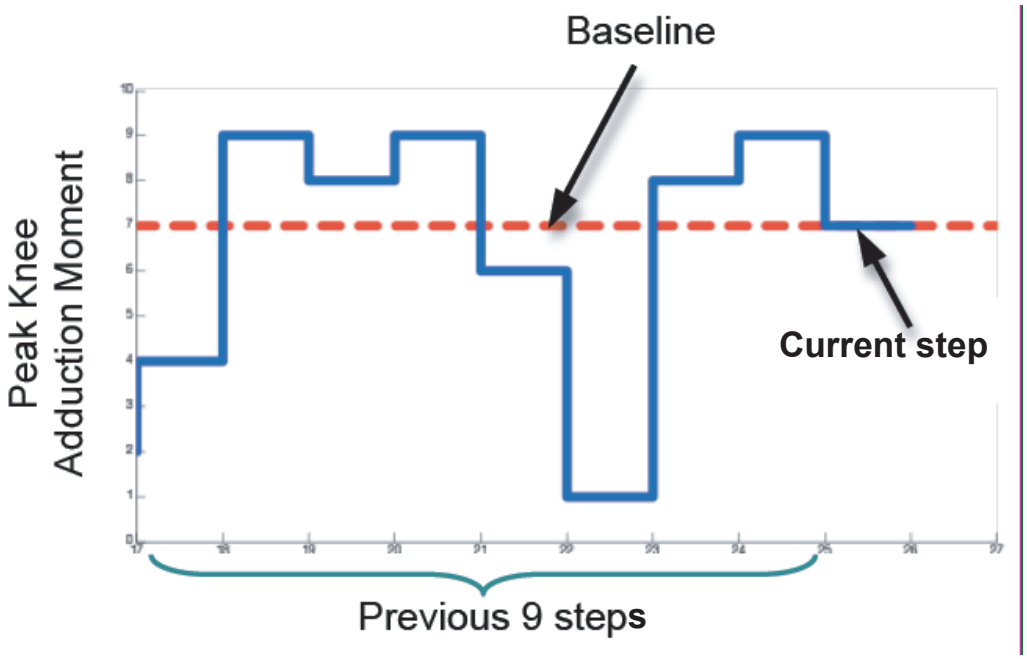

Figure 3.4: Sample of visual feedback display. The peak moment from the current step and the previous nine were displayed to the subject.

levels from previous steps. In all cases, a 0.5 second burst of vibration at 250 Hz was provided as soon as possible after the first peak in KAM. If the peak KAM on the current step was 80% of the baseline moment or larger, a large amplitude vibration was presented. If the peak KAM was 60-80% of the baseline, a low-amplitude vibration was provided. If the peak KAM was below 60% of baseline, no vibration was provided. The subjects' goal therefore was to minimize the amplitude of the feedback.

3.2.4 Participants

Sixteen healthy subjects participated in the study (11 male, 5 female, mean age 29, range 21-49). Eight of the subjects received vibration feedback and eight received visual feedback. Approval was received from Stanford's Institutional Review Board and all subjects gave informed consent prior to their participation.

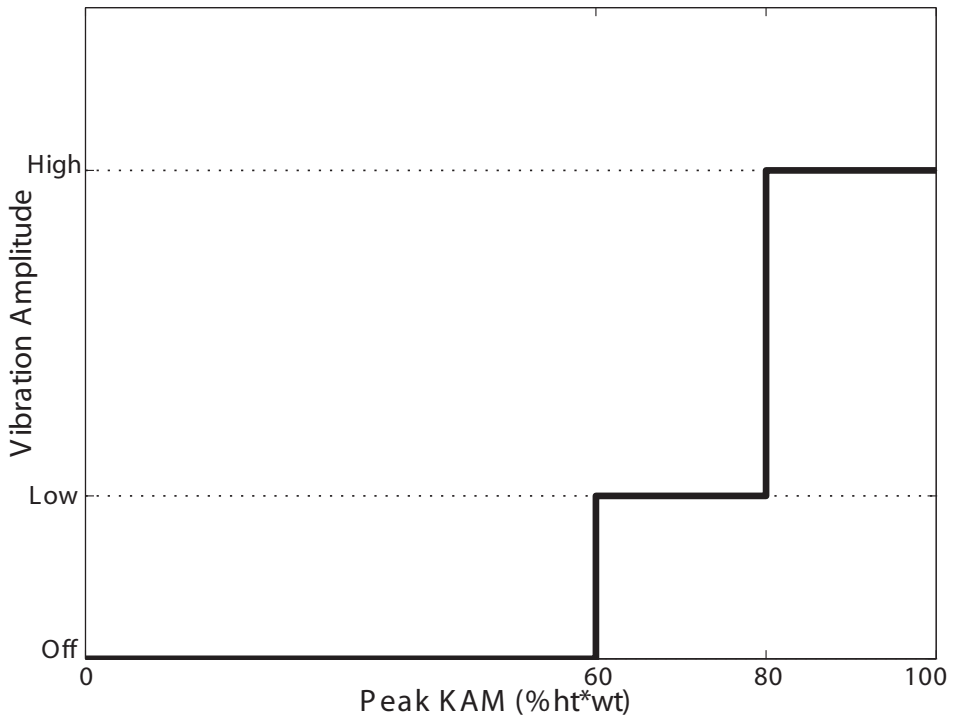

Figure 3.5: Mapping of peak knee adduction moment (KAM) to vibration amplitude binned into three easily distinguishable levels.

3.2.5 Data Analysis

The primary measures of interest from the study were the average first peak in the knee adduction moment (baseline and post-trial), the time required to complete the trial, and the awkwardness rating of the new gait. As discussed above, the knee adduction moment was calculated in the lab coordinate frame as this was a robust and simple way to perform calculations in real-time. However, the choice of coordinate frame for the knee moment can be important, particularly for non-standard gaits [18, 96]. We therefore used the captured data to calculate the moment in the tibia coordinate frame after the trial. We also used a more accurate method to determine the location of the knee joint center. This provided a check on the appropriateness of the approach used in real-time. Five of the sixteen subjects could not be post-processed due to poor static or dynamic marker data. The remaining 11 subjects'

data (six vibration, five vision) were analyzed as follows.

Using data from the static trial, we created a transformation matrix that related the coordinate frame created by the markers on the lower leg, to the coordinate frame created by the markers on anatomical landmarks. This matrix allowed reconstruction of the anatomical markers as "virtual markers" from the dynamic trial data. We also created a virtual marker at the knee joint center which we defined to be the midpoint of the line connecting the medial and lateral femoral condyle markers. As part of this analysis, we also filtered the force plate data (2nd order low-pass Butterworth with a cut-off of 30 Hz) and recalculated the COP. The knee joint moment was then calculated in the lab coordinate frame as above with this new data and was subsequently converted to the tibia frame with the coordinate transformation matrix from the static trial. The x-component of this moment was then the KAM in the tibia coordinate frame.

To evaluate statistical differences in the results of the experiment, various analyses of variance (ANOVAs) were performed. When comparing results within subjects (i.e., from baseline to final trials or in the lab and tibia coordinate frame) repeated measures ANOVAs were performed. When significant main effects were identified, paired t-test were used to compare cases. When evaluating aggregate results and comparing the two types of feedback a conventional ANOVA (between subjects) was performed with no pairing.

3.3 Results

The results for each subject are summarized in Table 3.1. For each subject, the type of feedback provided is given, along with the average first peak of the KAM for the baseline and final trials. No significant differences were found in the post-trial cases where the feedback was on and off so only the average of these 20 steps is presented for conciseness. The percent reduction in the KAM from baseline to final is also shown in Table 3.1 along with the time required to learn the new gait and the subjective

awkwardness rating. These results are based on the real-time calculations performed and are correlated to the feedback that was given to the subjects. All 16 subjects were able to reduce their peak KAM compared to the baseline case, though the amount of reduction varied from as little as 3% to more than 50%. Significant main effects due to both subject and trial were identified. The averge first peak of the KAM was significantly lower in the post-trial case ($p < 0.001$).

Sub #	Feedback	Baseline KAM (%h*w)	Post Trial 20 step ave KAM	Trial Duration (s)	Percent Reduction	Awkwardness Rating
1	Vision	5.80	4.15	292	28.45%	7
2	Vibro	2.95	1.53	867	48.14%	9
3	Vision	4.44	4.16	507	6.31%	4
4	Vibro	5.26	4.23	1139	19.58%	3
5	Vibro	3.02	2.92	968	3.31%	5
6	Vibro	4.84	3.89	538	19.63%	8
7	Vision	3.91	3.38	261	13.55%	7
8	Vision	4.38	4.14	598	5.48%	3
9	Vibro	3.80	3.67	505	3.42%	2
10	Vibro	3.57	3.26	431	8.68%	8
11	Vision	3.30	2.53	422	23.33%	8
12	Vision	3.66	3.31	381	9.56%	4
13	Vision	2.86	1.32	336	53.85%	3
14	Vision	4.21	3.31	350	21.38%	6
15	Vibro	4.84	3.27	384	32.44%	3
16	Vibro	2.88	1.91	455	33.68%	5

Table 3.1: Results for all 16 subjects. The Baseline KAM is the average first peak in the baseline trial. The Post Trial 20 Step Ave. KAM is the average first peak with the new gait. The percent reduction is a measure of the change in the average peak KAM from baseline to final. The trial duration is the time spent learning the new gait and the awkwardness rating is a subjective measure of the difference in the new gait compared to the subject's normal gait.

The percent reduction in the KAM from the baseline to final trials, along with the awkwardness ratings are presented in Figure 3.6. There is not a strong correlation

between the percent reduction and the awkwardness rating.

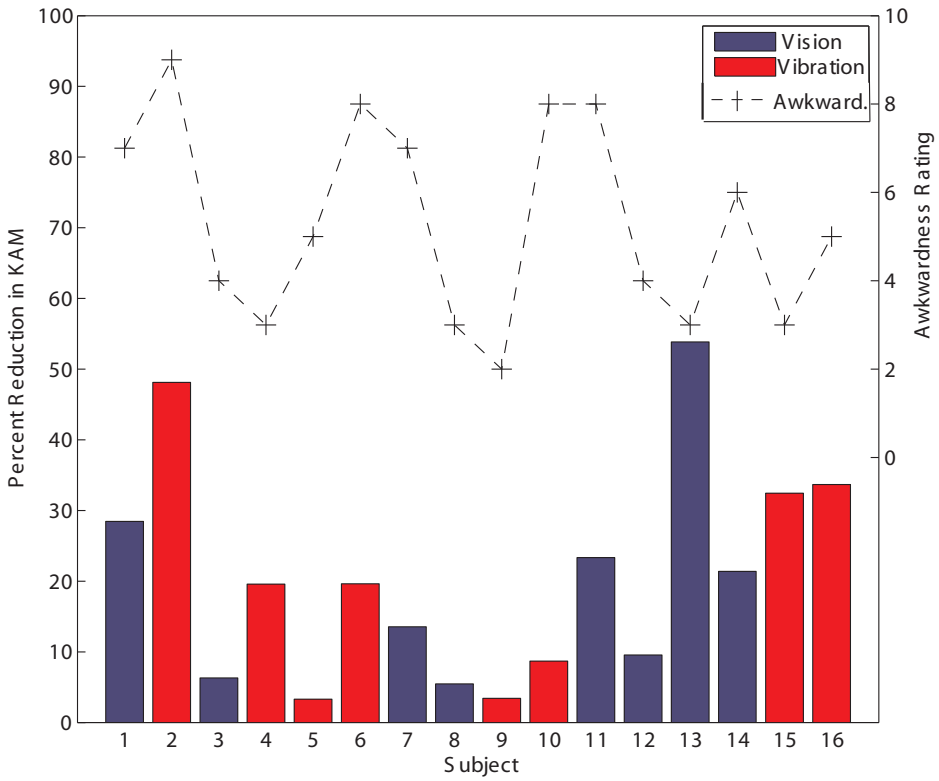

Figure 3.6: Percent reduction in the knee adduction moment from baseline to final trials along with the awkwardness rating for each subject.

Average results for all subjects and results grouped by the type of feedback provided are presented in Table 3.2. The percent reduction and awkwardness ratings for the visual and vibration feedback cases were not significantly different. However the duration of the trial was significantly shorter with visual feedback compared to vibration ($p < 0.01$).

The strategies that subjects reported using in their final chosen gait are shown in Table 3.3. The most common modifications chosen were toeing in (14 subjects), loading the medial side of the foot (6 subjects) and increasing the amount of trunk sway (4 subjects). Other modifications varied by subject and included some that were not included in the recommendations made before the trial.

	Baseline KAM	Final KAM	% Red.	Duration (s)	Awk. Rating
All Subs.	3.98	3.19	20.67%	527	5.31
Visual	4.07	3.29	20.24%	393	5.25
Vibration	3.90	3.09	21.11%	661	5.38

Table 3.2: Average results for all subjects and grouped by feedback type. Moments are in units of percent of height times weight.

Sub #	Strategies
1	Toe-in, load medial side of foot, slight hip thrust
2	Toe-in, load medial side of foot
3	Toe-in, feet more apart
4	Toe-in, medial knee thrust, longer stride, softer impact
5	Toe-in, feet closer together
6	Increased trunk sway, knee bent, medial heel strike, lateral push-off.
7	Toe-in, feet closer together, no arm swing
8	Toe-in, shorter stride
9	Toe-in, invert knees, shorter stride
10	Toe-in, load lateral side of foot
11	Toe-in, load medial side of foot, longer strides
12	Toe-in, leaning back, longer stride, less arm swing, knees flexed, widened stance, increased trunk sway
13	Toe-in, invert knees, increased trunk sway
14	Toe-in, load medial side of foot
15	Load medial side of foot, increased trunk sway
16	Toe-in, wider stance, load medial side of foot

Table 3.3: Gait modifications chosen by each subject.

Data for 11 of the 16 subjects were post-processed to calculate the KAM in the tibia coordinate frame. This analysis provided a check on the validity of performing moment calculations in the lab coordinate frame in real-time. In general, the knee adduction moments in the lab frame and the tibia frame were similar in the baseline trial (after correcting for the more accurate knee joint center). However, for the baseline trials where more abnormal motions were often present (particularly leg rotations in the transverse plane), the choice of frame had a larger affect on the moment calculation.

The average first peak of the knee adduction moment (for the baseline and final trials) calculated in real-time with the less-accurate knee joint center in the lab frame

and the post-processed peak KAM with the corrected knee center in the tibia frame are shown in Figure 3.7. While all subjects had a lower post-trial moment based on the real-time calculation, this was not the case with the more accurate analysis. In this case, two of the eleven subjects had a larger first peak with the new gait compared to baseline and one had a nearly identical peak moment in both trials. Those subjects whose average peak moment increased from baseline to final in the tibia frame were those who had a modest reduction based on the real-time calculations. The other eight subjects still had a smaller average peak moment in the final trial than the baseline. In some cases, the percent reduction was greater in the tibia frame than the lab frame. Overall, there was no statistically significant difference between the percent reduction values for the two frames.

While the first peak was the one we were most interested in reducing in this study, we were also interested in the effect of the new gait on the second peak of the knee adduction moment. The average second peak was calculated in the tibia coordinate frame for both trials for the 11 subjects with good marker data (Figure 3.8). The trend was not consistent as five subjects had a higher second peak with the modified gait and six had a lower second peak. The average peak was not significantly different from baseline to final trials.

3.4 Discussion

The fact that all of the subjects were able to reduce the average first peak of the KAM based on the feedback we provided was promising. This approach of providing real-time feedback for a repetitive task has the potential to greatly improve the speed of subject-specific gait retraining compared to prior studies [35, 36]. The fact that vibration produced a similar percent reduction in the KAM and awkwardness ratings was somewhat surprising. The resolution of the device is much lower than vision and there is no time history other than the subject's memory. This may have been in part due to the fact that the subject only had coarse control over the KAM and

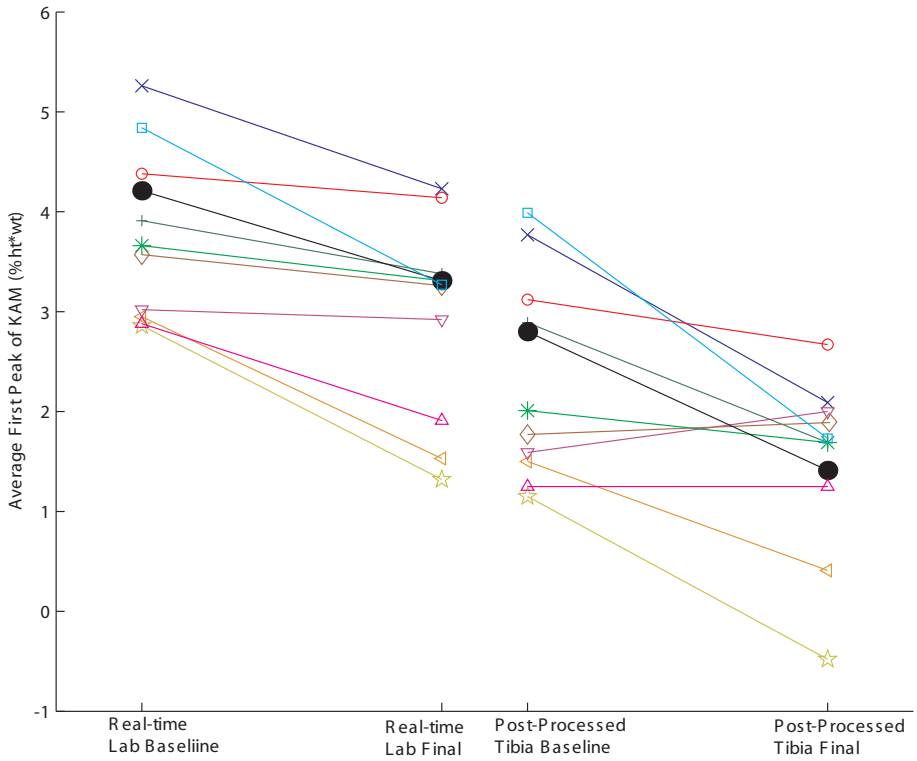

Figure 3.7: Average first peak of the KAM calculated in the lab frame in real-time (left) and in the tibial coordinate frame after post-processing (right).

so a relatively coarse feedback channel was still effective. However, the amount of time it took to converge on a new gait was much longer with vibration. In observing the study, this was most likely due to the fact that the subject would only receive a change in feedback from one step to the next if the peak KAM changed significantly (at least 20%). This made it more difficult to identify which strategies were effective and which were not. With visual feedback, trends of small changes could be detected and this information could be used to give feedback about various modifications more quickly. We feel that it may be most appropriate to first train subjects with visual feedback and then vibration could be used for maintenance of the adopted gait in a portable system. Of course, a portable system would also require sensors that could

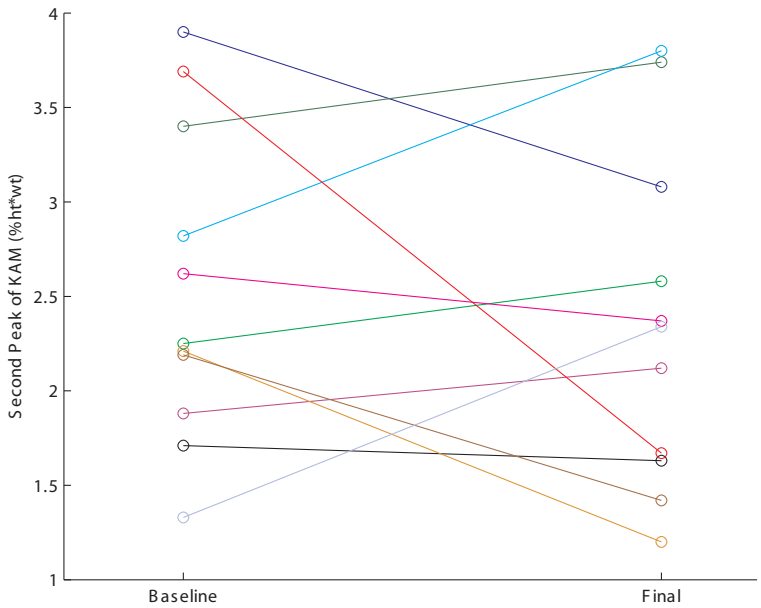

Figure 3.8: Second peak of the KAM in the tibial coordinate frame for baseline and final trials.

provide a reasonable approximation of what the motion capture system and force plates provided. The feasibility of this remains an open question.

There is some difficulty in assessing the aggregate results due to relatively large subject-to-subject variations. While subjects were instructed to adopt a gait that was sustainable and acceptable, there was significant subjectivity in how to interpret this. Some subjects adopted a more awkward gait than others. Despite this, there was not a strong correlation between the percent reduction in the moment and the subjective awkwardness rating. Instead it seems that some subjects' moment was very sensitive to small gait changes while others' was much less sensitive. This could be a function of anatomy (e.g. varus/valgus alignment) or the nominal gait pattern of the person. How this variability might affect future studies with OA patients is not yet clear.

In observing the gait modifications that subjects chose, there were some common trends. Toeing in was by far the most universal strategy used. This is interesting

as OA patients are often instructed to toe-out. This can reduce the second peak but not necessarily the first [5, 6, 49, 108]. Another common strategy was to load the medial side of the foot. To understand why these and other strategies might be effective, an understanding of how the moment is generated is helpful. The knee adduction moment has four contributing terms, all occurring in the frontal (yz) plane of the chosen coordinate system [56]. These terms are the y (medial/lateral) and z (vertical) components of the GRF and the y and z lever arms (see Figure 3.3). The z force and lever arm are primarily determined by the weight of the person and the length of the lower leg respectively and cannot be changed much with modest gait modifications. The y force and lever arm, however are more sensitive. The y force is small relative to the z-force but is multiplied by a larger lever arm. The y lever arm is small compared to the z lever arm but is multiplied by the much larger z GRF component. This makes the relative importance of these two terms similar (at least the same order of magnitude). The y component of the GRF can be reduced by increasing trunk sway such that the GRF vector points more laterally. Four subjects adopted this strategy. In this approach, the timing of the trunk sway is important in order to reduce the first peak. If the trunk sways at the wrong time it can increase the y GRF component. Assuming the direction of the GRF is fixed, the lever arm in the y direction can also be reduced by moving the knee joint center medially. Simply inverting the knee (using hip and ankle motions to move the knee medially) can have this effect. The knee can also move medially as a consequence of medially loading the foot or toeing in. Toeing in actually causes the foot to be loaded medially towards the beginning of stance and laterally toward the end of stance. This may account for the reduction in first peak but not necessarily the second in this study and the inverse in prior studies on toeing out. In this study we were interested in evaluating the efficacy of the feedback and therefore only provided broad suggestions of gait modifications. For clinical studies, faster and more effective training may be achieved with targeted coaching of previously identified strategies that are likely to be advantageous.

The analysis of the post-processed data showed that the choice of reference frame

was important, particularly for the modified gait. However, those who had a significant reduction in the KAM based on the real-time calculations generally had a greater percent reduction in the tibia frame. Those with modest reductions based on real-time data however, sometimes had a larger post-trial moment in the tibia frame. In either case, the subjects were able to reduce the parameter they were given feedback about, which seems to indicate that if the feedback was given in the tibia frame, subjects would similarly be able to reduce it. Based on these findings, for future studies, an attempt to convert to the tibia coordinate frame and produce a more accurate estimate of the knee joint center in real-time should be made. This is feasible but there are some potential issues. A larger number of markers must be tracked in real-time in order to create the coordinate system. In the real-time analysis in this study, only the left lateral femoral condyle marker needed to be detected at all times. To convert to the anatomical frame, a minimum of three are needed. If less than three are available at any time, the coordinate frame cannot be reconstructed. Perhaps more problematic, the system will often mislabel a marker after it has not been detected for a few frames. This causes all future calculations to be erroneous. The software must be robust to marker 'drop-out' to prevent these problems. Some additional computational burden will also result but this should not affect performance of the system. The creation of the transformation matrix requires a matrix inverse but this is only required once after the static trial. At each time step in the dynamic trials, a simple matrix/vector multiplication and a few vector additions (for the virtual markers) are all that are required.

This study showed the feasibility of providing real-time feedback for gait retraining. It also showed that simple haptic feedback could be effective, though learning was slower than with visual feedback. In addition to the computational improvements discussed above, future studies should test patients who have been diagnosed with OA as well as the long term retention of the adopted gait pattern. This approach provides significant benefits compared to other approaches as it is non-invasive, subject specific, and much faster than traditional gait retraining (a few minutes or hours

instead of weeks or months).

Chapter 4

Artificial Proprioception for Targeted Movements

In this chapter we describe an experiment in which we compared the ability of subjects to perform blind cursor movements without haptic feedback (in open-loop fashion) and with skin stretch feedback. To provide a comparison, simple vibration feedback was also tested. The task is roughly analogous to asking a person to move her hand specified distances, such as 10 or 20 cm to the left or right, without looking. In the present case, subjects apply forces to a single-axis load-cell held between the fingers. The force input controls a cursor that is attached to a virtual object that, like a human or robotic arm, has position-dependent dynamics.

While there are a number of ways in which skin stretch and vibration could be used to convey information, this study focused on evaluating the merits of using skin stretch feedback in a generic blind positioning task. The feedback is proprioceptive in that it is related to the position and motion of a virtual object. It is difficult to make a fair comparison between vibration and skin stretch feedback as neither method was optimized for this study. Instead, the general area of tactile stimulation was kept comparable. Single actuators of comparable footprint on the skin (a few cm^2) and capable of being used continuously, varying the signal to represent position were used.

More sophisticated approaches could be employed that utilize multiple actuators and add discrete event cues. However, in order to make as fair a comparison as possible between the modalities, we constrained the experiment to continuously varying signals with a single actuator in each case. This work was performed collaboratively with Karlin Bark.

4.1 Methods

In the following sections we describe two experiments used to assess the effectiveness of skin stretch feedback for proprioception. We will first describe the methods for the closed-loop targeting study. We will then describe a perception study that was used to validate the choice of feedback mappings.

Subjects were asked to perform cursor movements using a single-axis load cell with two strain gauges in a half-bridge configuration, held between the fingers and thumb. The use of a force (versus position) input minimizes the use of the subject's own proprioceptive sense as the load cell replaces actual hand motion. The virtual dynamics of the cursor are described in Section 4.1.1 below. A virtual workspace with a range of motion from 0-10 units was displayed on a computer monitor (Figures 4.1, 4.2). For each task, the cursor appeared at a starting position that was randomly chosen from (3, 5, 7) and the subject was instructed to move 2, 4 or 6 units to the right or left. Movement commands were constrained so that the desired end location was always in the range of 1-9. Subjects could not see the cursor during trials but they were given post-trial visual feedback throughout the experiment (final cursor position and desired position shown on the workspace). Subjects were instructed to attempt to move the desired number of units, stop the cursor, and press a button when they thought the virtual object was brought to a stop. Each of twelve possible combinations of starting positions and movements was repeated three times, for a total of 36 trials.

The experiment described above was performed under four feedback conditions

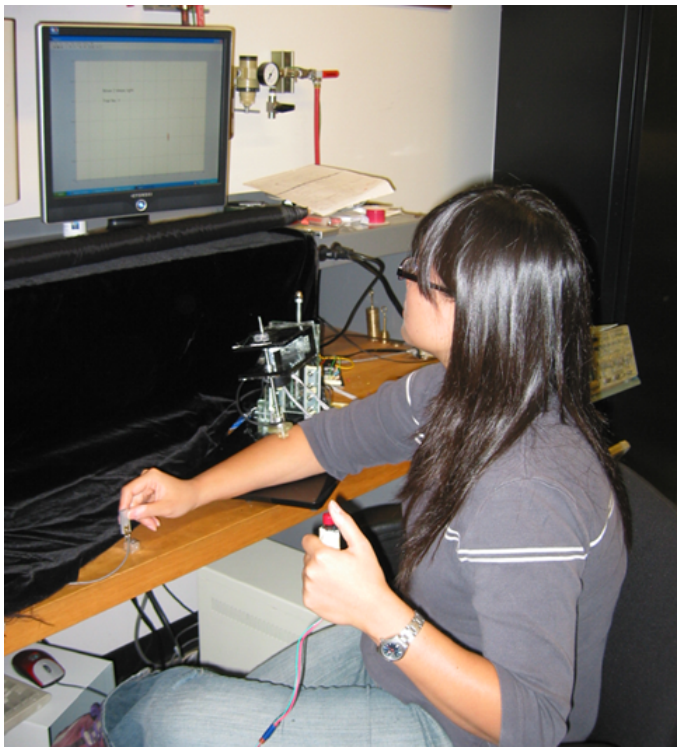

Figure 4.1: Subject completing the task with right arm controlling force sensor and left hand controlling button to end trial.

for each subject. First, subjects were tested with no haptic feedback. They were then tested with vibrotactile and skin stretch feedback. Then they repeated the no-feedback trial to evaluate training effects over the course of the experiment. Half of the subjects did the vibrotactile feedback trial before the skin stretch and the others did them in the opposite order. Before each trial, subjects were given training with visual feedback (cursor and workspace visible) for about one minute and then given ten practice trials with post-trial vision feedback, identical to the actual trials. For the no-feedback case, this practice allowed subjects to learn the position dependent cursor-dynamics and open-loop movement strategies. In the feedback cases, it allowed them to learn the haptic mapping of the cursor position.

Ten subjects were tested (three female, seven male). Four had little or no experience with haptic devices; the other six had at least moderate experience. The

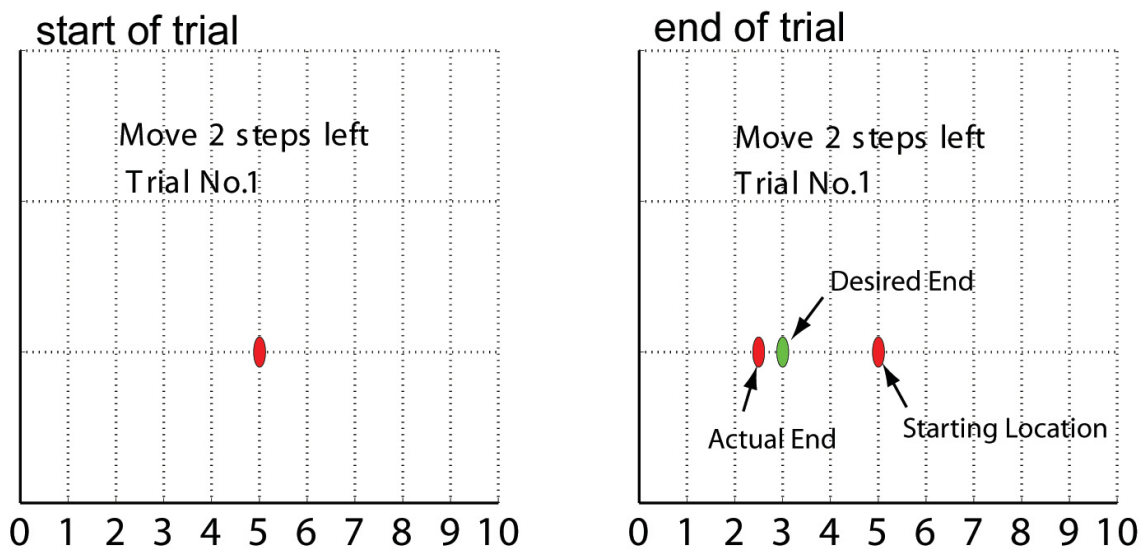

Figure 4.2: Screen shots of experiment task. The left graph shows the screen presented to the subject at the start of each trial. The right graph shows the screen presented to the subject at the end of each trial, showing desired cursor position and actual position to provide post-trial feedback.

experiment took about one hour to complete.

4.1.1 Cursor Dynamics

In pilot tests with four subjects we determined that if subjects were asked to move repeatedly to a single target, or one of a small number of targets, they often used feedback in a "pattern matching" mode. That is, they moved until the vibration or skin stretch felt like it did previously when they were over the target, and then they stopped. This way of using the feedback seemed closer to an event-cue (event = target reached) than proprioception. Consequently, we revised the experiment to utilize varied starting locations and amounts of movement in the left or right direction, as described above.

We also discovered that the virtual object attached to the cursor should have non-trivial dynamics. If the virtual object had a fixed mass and damping, subjects quickly learned open-loop strategies such as pulsing the force applied to the sensor a

certain number of times, or applying a steady force and counting "beats," to move the object a desired distance with accuracy. We hypothesized that analogous strategies do not work with human or prosthetic arms in part because the arm dynamics, and the mapping of muscle effort to movement, change continuously as a function of the arm configuration. Accordingly, we gave the cursor an inertia that varies somewhat like the endpoint inertia of a two-link robot arm whose end effector is constrained to move along a single direction in space. The endpoint inertia will be a polynomial involving sine and cosine functions of the position. A simplified approximation is a sinusoid, so that the cursor dynamics become:

$$m(x)\ddot{x} + b\dot{x} = F(t) \tag{4.1}$$

where b is the cursor damping, *F(t)* is the force applied to the force sensor which produces cursor motion, *x(t)*, and *m(x)* is the mass, which varies as

$$m(x) = 6 + 5sin(\frac{2\pi}{10}x) \tag{4.2}$$

The period of the mass variation matched the length of the visible workspace and the maximum and minimum mass were 11 and 1, respectively, with units such that a force magnitude of $F = 1$ and a mass of $m = 1$ resulted in an acceleration of 1 workspace unit/second^2. The damping was set to a constant value of $b = 10$. (The sinusoidal variation of the cursor mass can be seen superimposed on some of the subjects' data in Figure 4.15.)

A small deadband region was also added to reduce drift, such that the force applied to the cursor was related to the force from the sensor, F_j, by

$$F(s) = \begin{cases} 0 & |F_j| < 0.2 \\ F_j & |F_j| \geq 0.2 \end{cases} \tag{4.3}$$

Subjects removed their hands from the force sensor and the force was re-zeroed

before each trial (which lasted approximately 15 seconds) to ensure that there was no drift or bias force.

Subjects were told that the behavior of the cursor was position dependent but they were not told the actual mapping. As described above, they were given time to practice moving the cursor while it was visible before the experiment.

4.1.2 Vibrotactile Feedback

The vibrotactile feedback in this study was provided by a C2 Tactor, from EAI Inc. The tactor was placed on the arm, just below the elbow joint using a Velcro strap as shown in Figure 4.3. The actuator was controlled with a computer running the Mathwork's xPC Target real-time operating system through a current amplifier, controlling the actuation force.

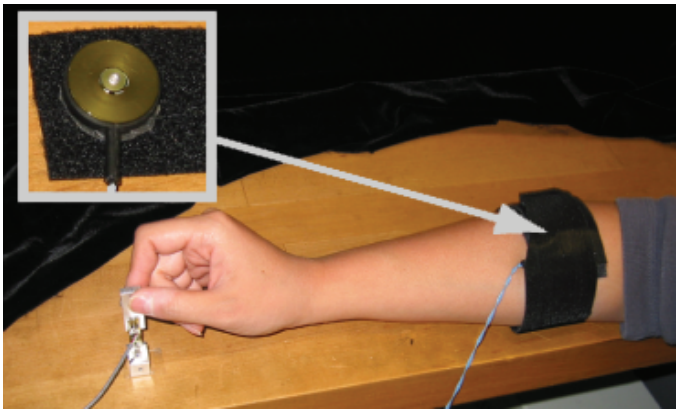

Figure 4.3: Vibrotactor strapped to test subject, placed on forearm below the elbow joint. The force sensor used to control cursor position is also pictured. Forces greater than 200 mN were needed to move the cursor.

Several position mappings were evaluated in pilot trials, including linear and non-linear mappings of frequency, amplitude or both. Sine waves were used in all cases. Because of the dynamics of the actuator attached to the skin, varying the frequency also results in magnitude variations, though not linearly or even monotonically. This made a variable frequency mapping difficult to use. In addition, varying the frequency

continuously results in Doppler effects when the cursor is moving rapidly, which cause the instantaneous frequency to be higher or lower depending on the direction of motion.

A pager motor (DC motor with unbalanced inertia) was also used in pilot trials. This motor exhibited an approximately linear relationship between input voltage and magnitude and an approximately quadratic relationship between frequency and voltage input. However, the ranges of frequencies (from about 50 Hz to 175 Hz) and magnitudes that the motor was capable of producing were smaller than with the C2 Tactor.

Three subjects were tested with both the pager motor (input voltage varied as a function of cursor position) and the C2 Tactor (forcing amplitude varied as a function of cursor position at 250 Hz) and all three did better with the tactor. Based on these studies, we determined that varying the amplitude of the sine wave sent to the C2 Tactor at a constant frequency of 250 Hz provided the most effective position mapping. Pilot trials also showed that a logarithmic amplitude mapping was more effective than a linear one. This result is consistent with other findings in the literature that amplitude perception follows a logarithmic pattern [85].

The final mapping chosen obeyed the following relation:

$$A(x) = 0.5 \times 10^{0.06x} \tag{4.4}$$

where *A(x)* is the amplitude of the stimulus and x is the cursor position. This results in a small but perceivable stimulus level at $x = 0$ and a stimulus near the current limit of the actuator at $x = 10$ corresponding to $A(x) = 2$, which produces a peak acceleration of approximately 7.5 G as measured by an accelerometer on the tactor, in contact with skin. When the cursor moved outside the $0 \leq x \leq 10$ units workspace, the stimulus saturated at the values for 0 and 10, respectively. While more rigorous studies would be required to determine an optimal mapping of vibrotactile stimulation, this was the vibration mapping that our pilot subjects found

most intuitive and performed best with.

4.1.3 Skin Stretch Feedback

The benchtop skin stretch device was used in these studies for controlled application of skin stretch. The skin stretch device was attached to the subject's arm, just below the elbow (Figure 4.4), in an area similar to where the vibration tactor was placed. The contact points were placed such that the line connecting them was perpendicular to the forearm.

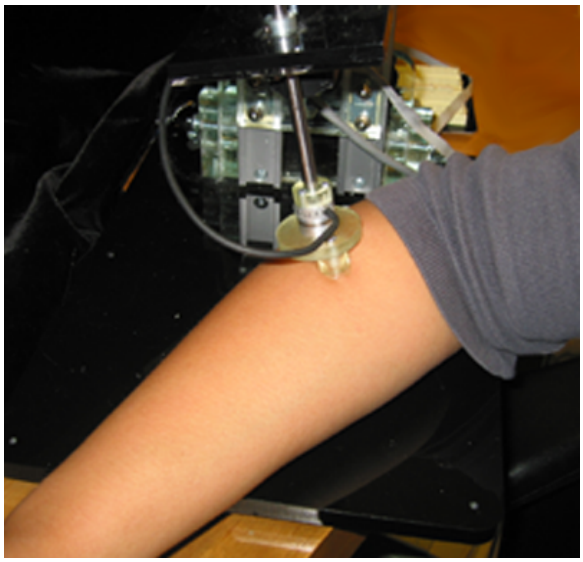

Figure 4.4: Skin stretch device attached to forearm below the elbow joint. Two contact points are attached using adhesive tape and rotate to apply skin stretch. Total contact area is approximately 2.5 cm^2.

For these experiments, a range of ± 45 degrees of rotation was mapped to the cursor position. This range of stretch was determined to produce sufficiently large, but comfortable, magnitudes of feedback in pilot studies.

One advantage of skin stretch feedback over vibration is the ability to convey direction. For these experiments, the rotation was set to 0 degrees when the cursor was in the middle of the workspace, at 5 units. The device subsequently rotated clockwise

or counterclockwise according to the direction of the cursor movement. Because there is relatively little information about perception of skin stretch applied to hairy skin, determining the optimal mapping of the cursor to magnitude of stretch is an open-ended problem. In pilot studies, a linear mapping of cursor position to degrees of skin stretch was first evaluated. Pilot subjects noted that there appeared to be a region of the cursor position surrounding $x = 5$ where it was difficult to detect rotation of the device. This region of uncertainty was also found in the perception studies described in Chapter 2. At higher levels of stretch, the skin stiffens nonlinearly [8] so that linear increases in displacement produce more than linear increases in stress. Thus, when skin stiffness is low, at low rotations, a greater change in rotation is required to elicit sensations; at higher angles, near the saturation limits, smaller changes in angle are detectable. To account for these effects we used a slightly nonlinear monotonic, fifth order polynomial to map cursor position to rotation. A plot of the polynomial mapping is shown in Figure 4.5. The slope of the polynomial is also shown, which indicates the rate of change of skin stretch with respect to changes in cursor position. The varying slope reflects the hypothesized variation in skin compliance at low and high rotations, respectively.

Following the approach taken with vibration feedback, when the cursor left the region $0 \leq x \leq 10$ the skin stretch rotation was saturated at that the minimum or maximum angles (± 45 degrees). Although further testing is required to determine optimal mappings of skin stretch stimulation, the subjects in pilot tests found the slightly nonlinear mapping easy to interpret and they performed better with it than with a linear mapping.

4.1.4 Perception Study

To evaluate the effectiveness of the mappings used in the present study, a perception study of both vibration and skin stretch feedback was performed in which subjects were presented stimuli corresponding to five cursor positions and they were asked to

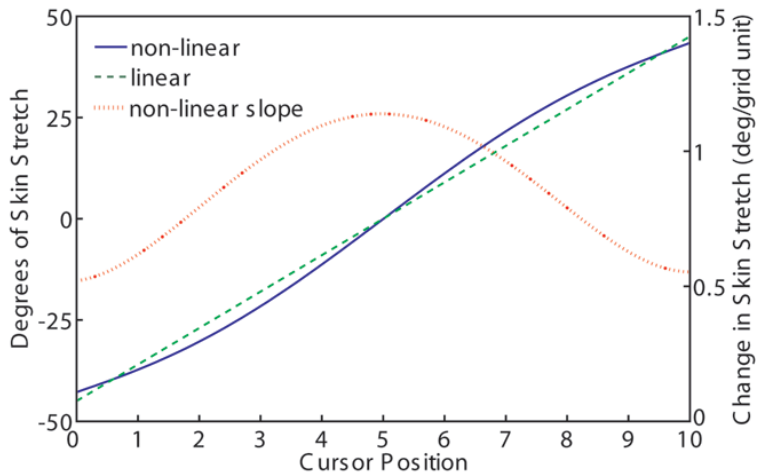

Figure 4.5: Skin stretch mapping relative to cursor position. Both the initial linear mapping as well as the non-linear method used are shown. Though the non-linear mapping is close to linear, at cursor locations near x=5 the increased change in stretch is great enough to overcome threshold limits to improve subject performance. The slope of the nonlinear mapping is also shown, reflecting the hypothesized variation in skin compliance.

report their perception of the stimulus. The five cursor positions corresponded to the final locations in the cursor workspace subjects were asked to move towards in the targeting study. If the subjects could distinguish the different stimuli either by perceived magnitude or position, the use of our mappings could be validated.

One of the unique features of the skin stretch feedback is that it is inherently continuous and therefore, cannot be instantaneously changed in magnitude as vibration can be. Frequency and amplitude characteristics of a vibratory stimulus can be changed in software, prior to presenting the stimulus to the user. In contrast, skin stretch rotation must always ramp to the desired position. In order to apply 30 degrees of stretch, the rotation must pass through 10 and 20 degrees. Subjects can use not only the static position of stretch, but also the dynamic motion as the stretch applicator moves to distinguish between different stimuli. This makes a classical perception study like those typically used for vibration not practical. A more intuitive method of reporting perceived skin stretch with directional components was

to think in terms of rotational displacement rather than magnitude. Hence, subjects were asked to report their perception of the orientation of the skin stretch device, not simply the strength of the stimulus. For vibration, because it lacks a clear directional component, subjects were simply asked to report a perceived magnitude.

We also observed that subjects could use information about the motion of the device to detect the amount of static skin stretch by integrating the velocity sensation. In our closed-loop task, as described in Section 4.1, the dynamics of the cursor were position dependent which made it more difficult for subjects to use this open loop strategy. To prevent subjects from simply timing the duration of the stimulus, we presented stimuli (Skin Stretch and Vibration) that started at one of three easily distinguishable levels corresponding to the extremes and middle of the workspace (0, 5 and 10 units) and ramped the stimulus at a constant speed to the desired stimulus level (1,3,5,7 or 9 units). This followed the protocol outlined for the closed loop experiments. In this method, subjects were able to use a sense of static or dynamic information or both to estimate position. Vibration stimuli were applied in a similar manner to keep the experiment protocol consistent between feedback methods.

Prior to the experiment, subjects were presented with the three possible starting stimulus levels (min, max and zero) and they were asked to provide a numerical value on an absolute scale including integers or fractions [118] of their choosing corresponding to the minimum and maximum. For vibration, these numbers were constrained to be positive numbers. For skin stretch, one was positive and the other negative (for the opposite direction of rotation) but they did not need to be equal in value. Subjects were not told that the rotations were in fact symmetric. The center point, where no rotation was applied was constrained to be zero. After this initial calibration phase, subjects were given the changing stimuli as described above and were asked to report their perception on the scale that they had chosen by entering a numerical value into a spreadsheet. For each subject, 75 stimuli were presented (15 at each stimulus ending location) in random order. Five subjects were tested with each of the haptic feedback methods. To prevent subjects from using audio cues to distinguish between

stimuli, headphones were worn throughout the trials and subjects were instructed to look away from the skin stretch device when in use. Subjects were not told that there were only 5 different levels of stimuli and the stimuli were ramped to the desired levels at constant rates of change for both vibration and skin stretch. Vibrotactile trials were completed in 10 minutes while skin stretch trials lasted approximately 15 minutes.

4.2 Results

4.2.1 Perception Study Results

The results of the perception studies for vibration and skin stretch are shown in Figures 4.6 and 4.7 respectively. For each case, the results were independently normalized by dividing each of the subjects' responses by the mean of all of their responses for each point and then multiplying by the mean of all subjects' responses at that point [85] to account for subject to subject variation. The resulting means and standard deviations were then calculated and plotted.

For vibration, the perceived magnitudes were roughly linear on the average as a function of workspace location (recall that the actual mapping was logarithmic as given by Equation 1). The standard deviations were smaller at low stimulus amplitudes. For data analysis, a one-way ANOVA comparison of means was conducted for each of the points and each of the adjacent points were statistically distinguishable ($p < 1 \cdot 10^{-5}$).

The skin stretch perception results were slightly less linear with more sensitivity near the extremes of the workspace. This indicates that a slightly more non-linear skin stretch mapping that the one chosen may have been more appropriate. Error bars were also quite large with skin stretch feedback, however upon further inspection of the data, we found that variances were generally low with the exception of a few outliers. A total of eleven outliers out of 375 total data points were identified and

two subjects had no outliers. For example, in one trial when the starting position was zero workspace units (about -45 degrees of skin stretch) and the stimulus ramped to 9 workspace units (about 40 degrees of skin stretch) the subject entered a negative number of appropriate magnitude but opposite sign based on their chosen scale. This could result from them making an error in detecting the direction of movement or from them simply making a sign error. As this was an open loop perception study similar to the passive study presented in Chapter 2, these errors were expected and are not found in a closed-loop task. The red curve in Figure 4.7 shows the results if these points are corrected (sign reversed). As expected, sensitivity is improved and standard deviations reduced in this case. Even without correction all four sets of adjacent points are statistically different ($p < 0.001$) and with corrections the largest p-value was less than $1 \cdot 10^{-8}$.

Overall, the perception results indicated that the mappings used were satisfactory as users were able to statistically distinguish the various points from one another.

4.2.2 Targeting Study Results

The data collected from the experiments were analyzed to determine the effectiveness of providing haptic feedback in blind movement tasks. The main parameters of interest were the absolute and relative error (absolute error divided by desired movement length) in final cursor position and the instantaneous velocity of the cursor at the end of each trial. A repeated measures analysis of variance (ANOVA) comparison of means was conducted to determine if the means in position error were significantly different across the various feedback methods: no feedback (NF1), vibration (V), skin stretch (SS), and the final no feedback case (NF2). The ANOVA method used was a post hoc analysis conducted using the Bonferroni criterion [40]. The effects of varying step sizes and the starting/ending positions of the cursor were also studied. The data were grouped into several subcategories to identify patterns and trends.

Error bars on the plots below indicate standard errors. In general, standard

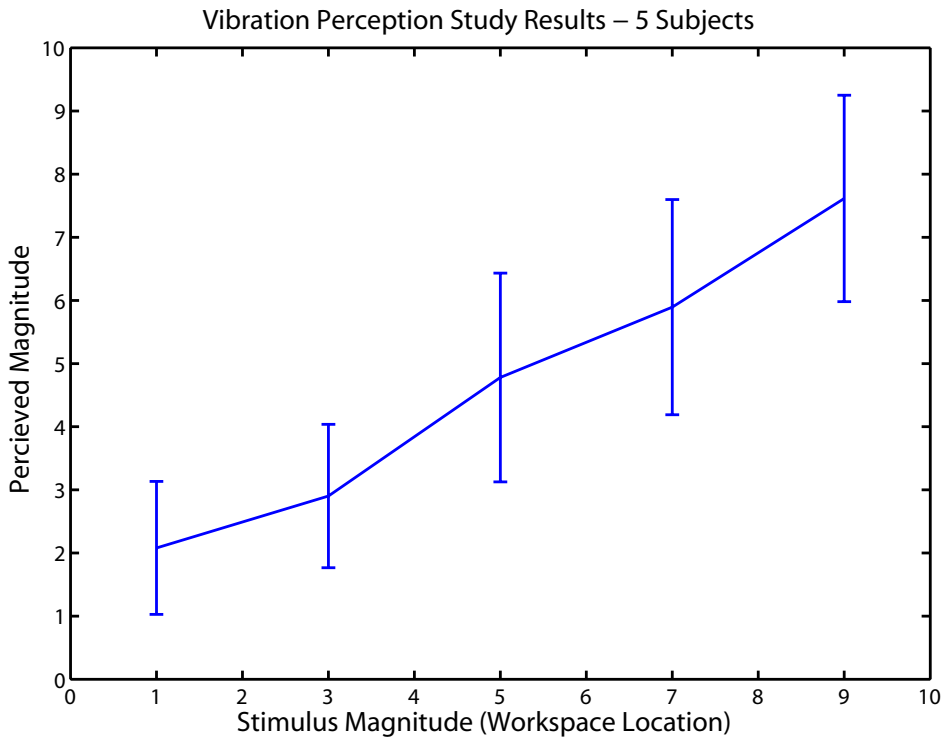

Figure 4.6: Perception curve for vibrotactile stimulation normalized over five subjects. The x-axis represents the workspace location that the stimulus corresponded to. The actual stimulus amplitude varied logarithmically with workspace location as given in equation 1. Error bars are plus and minus one standard deviation.

deviations were quite large relative to the means. This is largely due to the fact that the difficulty of the task caused standard deviations within subjects to be similar in magnitude to inter-subject standard deviations (such that normalization does not significantly decrease standard deviations). The task was designed to be comparably difficult to moving a real arm a specified distance without looking, which we expect would also result in relatively large variances. Quantitative results across all subjects as well as anecdotal observations of the most interesting cases are presented in the following sections.

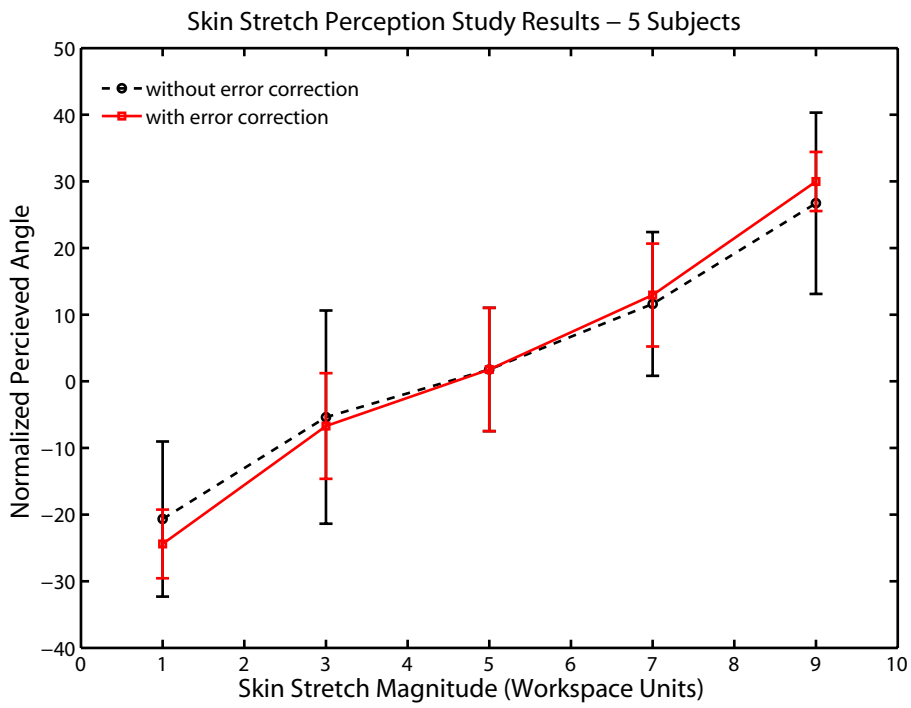

Figure 4.7: Perception curve for skin stretch normalized over five subjects with and without direction error correction. The x-axis represents the workspace location that the stimulus corresponded to. The actual amount of rotation applied is shown in Figure 4.5. Error bars are plus and minus one standard deviation.

4.2.3 Cursor Position

Overall Error

Repeated measures ANOVA applied to the absolute errors revealed significant main effects due to the feedback method, $F(3, 10) = 7.848$, $p = 0.001$. For relative errors, main effects due to feedback method were also found, $F(3,10) = 9.415$, $p = 0.0001$. Post hoc paired t-test analysis was then applied to the data using Least Squared Differences, where Table 4.1 presents the p-values comparing the various feedback methods.

As anticipated, the addition of skin stretch feedback improved movement accuracy.

	NF	SS	VL	NF2
NF	x	**0.002**	**0.035**	0.214
SS	**.002**	x	**0.057**	**0.014**
VL	**.035**	**.057**	x	**0.088**
NF2	.214	**.014**	**.088**	x

bold denotes significance at $\alpha = 0.10$ level

Table 4.1: P-values for post hoc paired t-tests comparing absolute errors. Skin stretch and vibration feedback were significantly different than no feedback at the $\alpha = 0.10$ level. Skin stretch was also significantly different from vibration.

	NF	SS	VL	NF2
NF	x	**.0002**	**.027**	.228
SS	**.0002**	x	**.052**	**.008**
VL	**.027**	**.052**	x	.103
NF2	.228	**.008**	.103	x

bold denotes significance at $\alpha = 0.10$ level

Table 4.2: P-values for post hoc paired t-tests comparing relative errors. Skin stretch and vibration feedback are significantly different than no feedback at the $\alpha = 0.10$ level. Skin stretch is also significantly different from vibration.

As seen in Figures 4.8 and 4.9, the relative and absolute position errors decreased with vibration and skin stretch feedback. Overall, in both absolute and relative error analysis, skin stretch produced statistically significant smaller error values (at the $\alpha = 0.10$ level) when compared to no feedback and vibrotactile feedback. The standard deviation of the errors was also lower with skin stretch than the other cases. Vibration feedback resulted in lower position errors as compared to receiving no feedback ($p < 0.10$). Of the 10 subjects tested, 9 performed best with skin stretch feedback (Figure 4.10).

Overall, subjects had significantly less error in the second no feedback case than

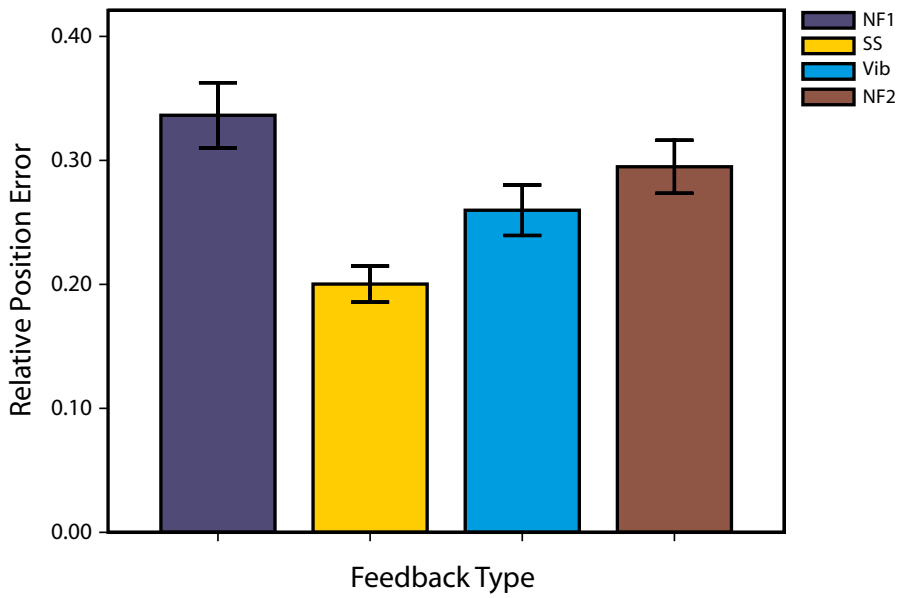

Figure 4.8: Average relative position errors for ten subjects. Error bars are plus one standard deviation. Skin stretch results in significantly smaller relative errors ($p < 0.05$) than no feedback.

the first, indicating that some improvement was taking place over the course of the experiment due to practice. However, this trend was not consistent across all subjects. Some did worse the second time and reported that they had become somewhat dependent on the feedback and had difficulty moving accurately when it was removed. We also performed a linear regression on the relative position errors across the 36 trials for each feedback case to see if significant improvement was occurring over the course of the trials. No significant trends were found in any of the feedback cases.

Error by Step Size

The position errors were sorted into various subcategories according to step size (the number of units the test subject was asked to move the cursor), and the desired ending position of the cursor. At first glance, when sorting the data by step size, it was clear that the addition of haptic feedback provided benefits over no feedback at

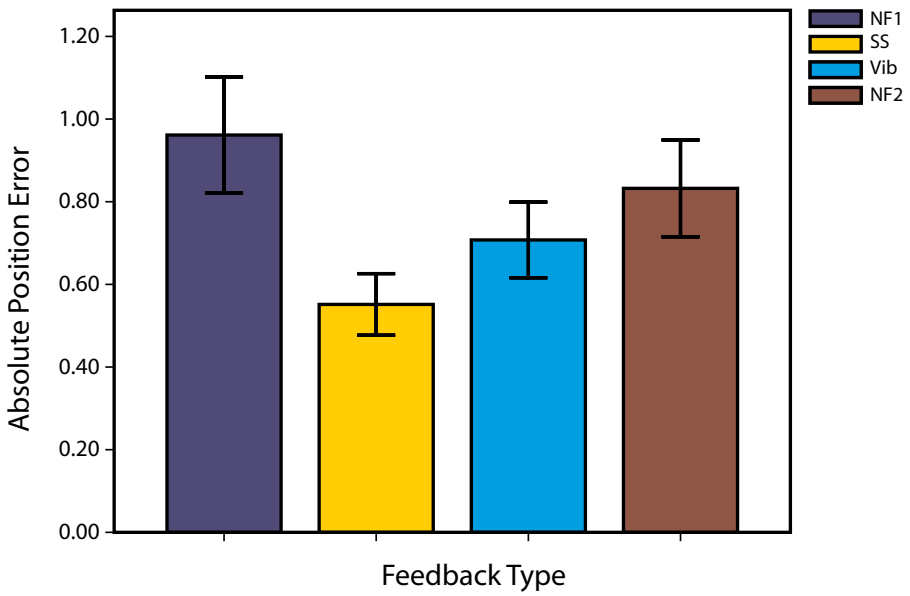

Figure 4.9: Average absolute position errors for ten subjects. Similar to relative error, skin stretch results in significantly smaller absolute errors ($p < 0.05$) than no feedback cases, though vibration does not.

each step magnitude (Figure 4.11). As expected, the absolute error increased as the desired step size increased, for all feedback modes. When no feedback was provided, the relative errors did not change significantly as a function of step size (Figure 4.12). However, for the two feedback modes, relative error decreased as step size increased from 2 to 6. This trend indicated that subjects seemed to be getting a sense of absolute position when haptic feedback was provided. If a true sense of position were provided, we would expect to see uniform absolute errors at all step sizes, such that relative errors decreased for larger step sizes.

Error by Target Location

The overall error as a function of target location is shown in Figure 4.13. Skin stretch again resulted in the lowest errors at all targets. There are at least two position

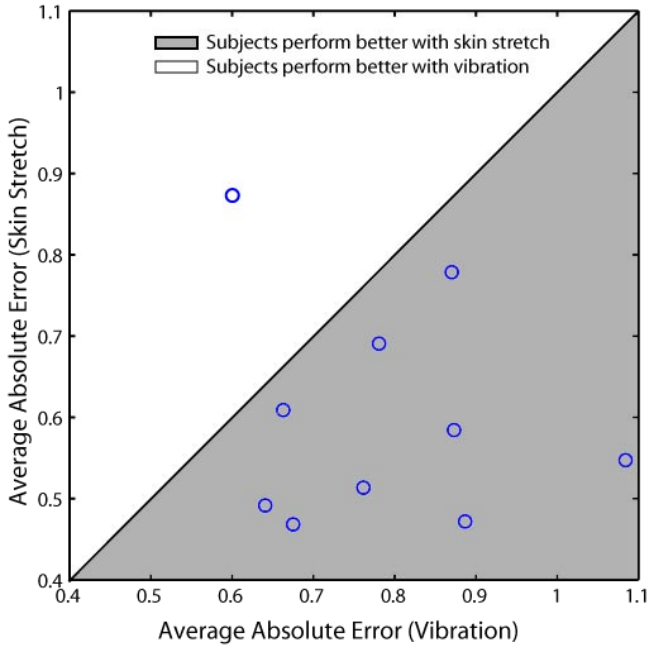

Figure 4.10: Average absolute position errors using vibration feedback and skin stretch feedback for each of the ten subjects. Each dot represents one subject. Dots in the shaded region indicates subjects who performed better with skin stretch feedback (9 of 10).

dependent factors that could influence these results. One is position dependent perception qualities of the haptic feedback, which are small (see perception study results above). The more dominant effect seems to be the cursor inertia. Errors were generally lowest near target location 7, which corresponds to a lower cursor inertia. The cursor was easier to stop in this location.

4.2.4 Final Velocity

Overall Velocity

The cursor velocity was calculated through differentiation (forward difference method) of the recorded cursor position throughout the experiment. No additional data filtering was needed since the cursor dynamics effectively act as a low-pass filter. The

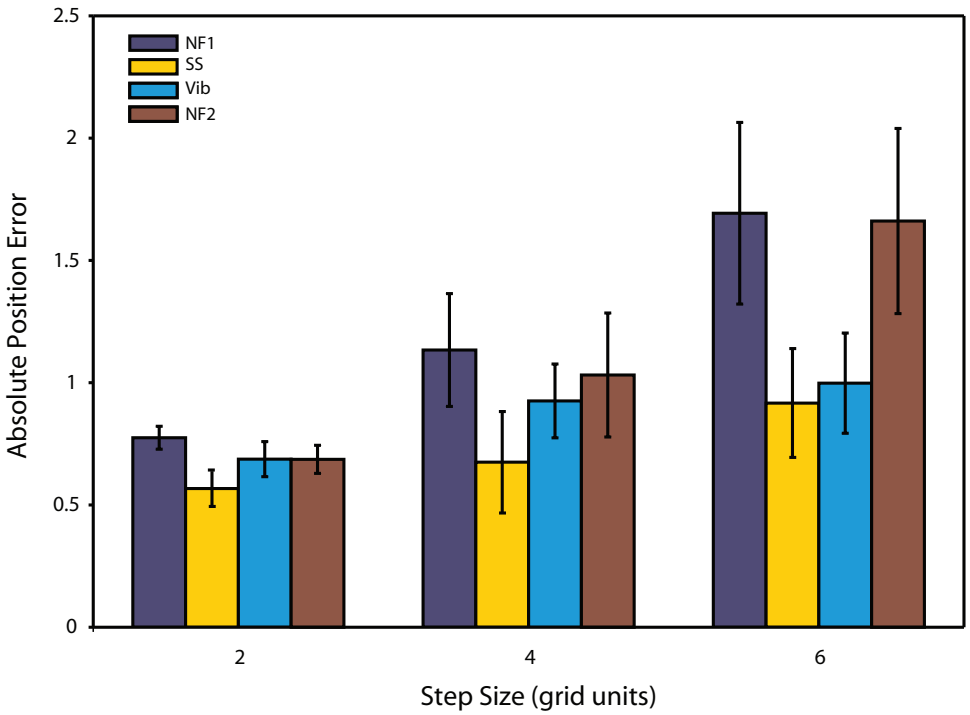

Figure 4.11: Average absolute position errors relative to step size. As expected, absolute errors tend to increase with increasing step sizes across all feedback modes. Both skin stretch and vibration feedback result in smaller errors at each step size, with skin stretch performing best, though not significantly.

instantaneous velocity of the cursor when the subject pressed the button (i.e., when the subject believed the cursor had been brought to a stop) was determined for each trial. For velocity analysis, it was discovered that 2 of the 10 subjects recorded velocities that were up to 4 times larger than the velocities measured during their first no feedback trial. These subjects were considered to be outliers as it was apparent they made no attempt to stop the cursor from moving before pushing the button in subsequent trials, though position errors were not large in comparison. The average cursor velocity for the remaining 8 subjects is seen in Figure 4.14.

Here, it is clear that skin stretch provided users with a sense of motion, allowing

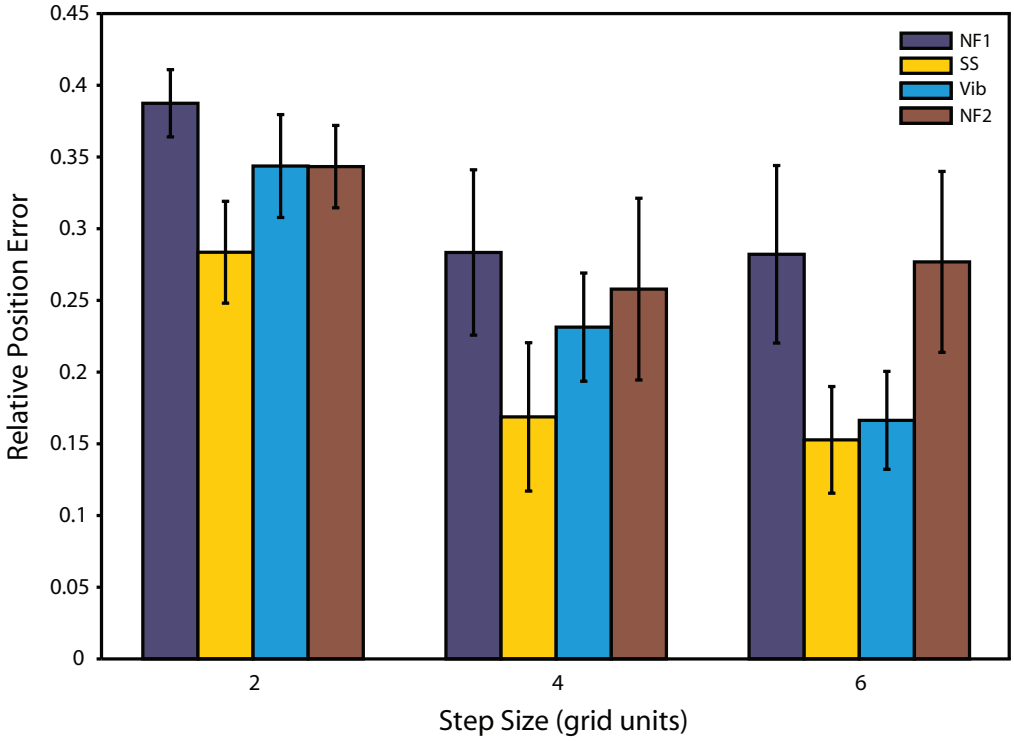

Figure 4.12: Average relative position errors by step size. The general trend is that relative errors decrease with increasing step sizes with feedback. At large step sizes (6), subjects perform better with feedback than without, though there is no significant difference between skin stretch and vibration. In addition, relative error decreases as step size increases from 2 to 6 when feedback is provided

them to more consistently stop the cursor. Velocities measured with skin stretch feedback are significantly lower than all other feedback modes with p-values less than 0.05.

It appears that the subjects were only able to easily detect velocity when skin stretch feedback was provided. However, we also note that the average velocities at which the subjects moved the cursor were lower with skin stretch than in the vibration or no feedback modes. It appears that because skin stretch provided a better qualitative sense of velocity, subjects moved more slowly in this case. In the

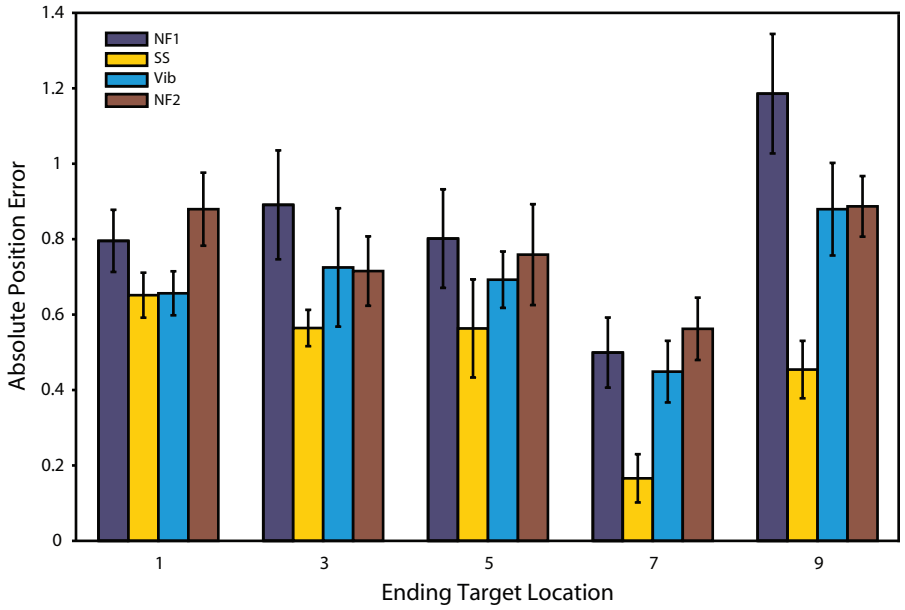

Figure 4.13: Average absolute position error versus desired target location. Skin stretch produced the lowest error at all targets. Lower errors are found where the cursor inertia is low (i.e. near point 7).

other cases, there was little sense of cursor velocity and no motivation or ability to limit speed.

Velocity by Cursor End Position

A point of interest when examining the final velocities is the correlation with cursor position and inertia. As the cursor inertia increases, while damping remains constant, more effort is required to bring the cursor to a stop. In all feedback methods, the average ending velocity of the cursor relative to the desired ending position closely matches the pattern of the inertial changes (Figure 4.15). A cursor position of 7 corresponds to a low cursor inertia, and correspondingly, the measured ending velocities of the cursor are lowest at that location. This suggests that subjects were indeed experiencing the challenges of controlling the varying cursor dynamics. However, the

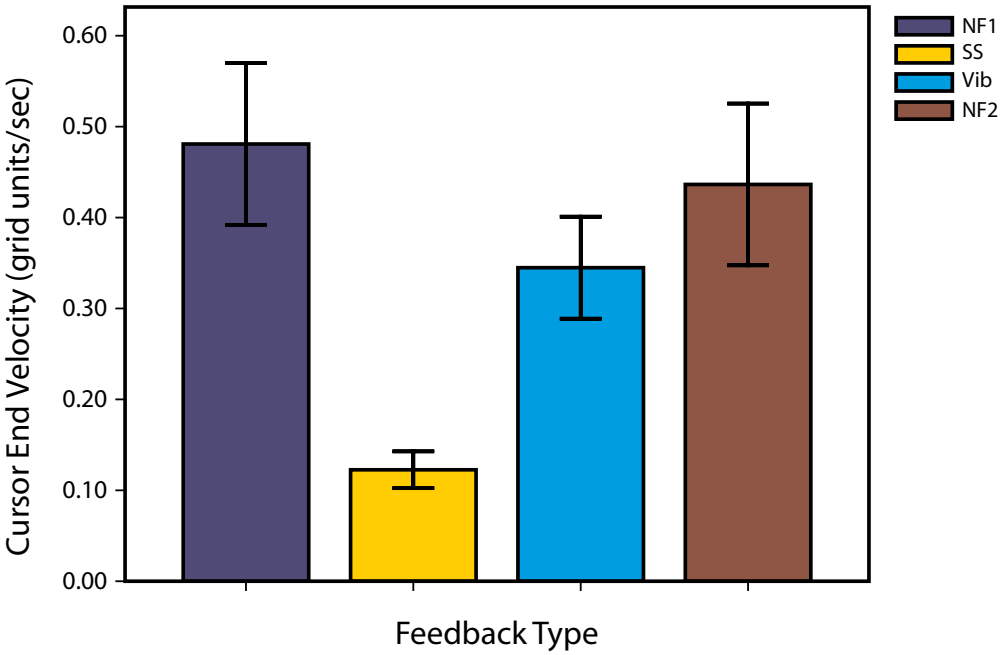

Figure 4.14: Overall average ending velocities. Skin stretch is far superior to all other feedback modes in maintaining low end velocities ($p < 0.05$).

errors are consistently lower with skin stretch and the variation with cursor position is generally lower than with other feedback modes.

4.3 Discussion

The results of the present study indicate that skin stretch may be an effective method for providing proprioceptive feedback in wearable displays. Skin stretch is an important part of the proprioceptive sense and provides a useful mapping for position information. In particular, skin stretch gives a realistic sense of dynamic motion in addition to static position using just a single stimulus, which could provide benefits for many applications. As observed in previous studies, this feedback is effective for closed loop control, providing a better sense of location and position when vision is

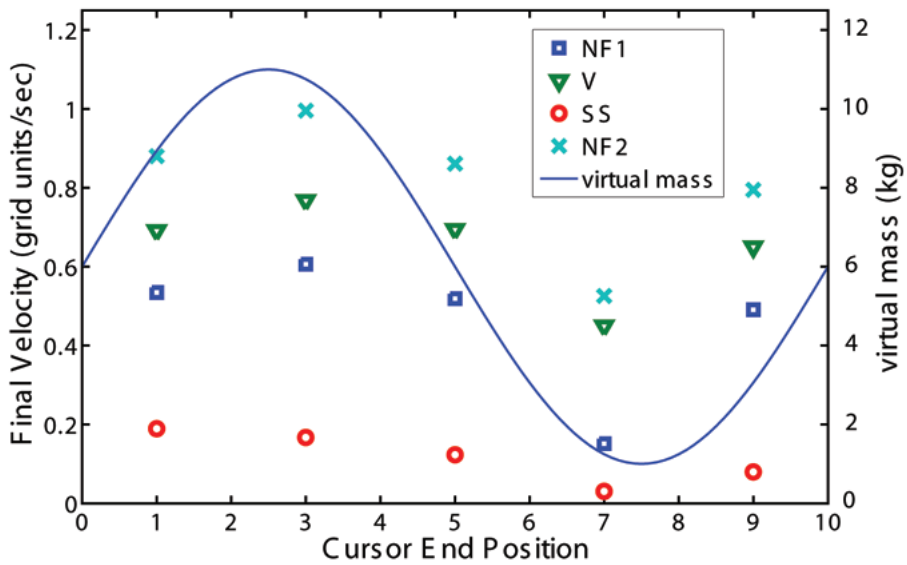

Figure 4.15: Average ending velocities with respect to desired ending cursor position. Average ending velocities from each feedback mode follow the sinusoidal characteristics of the varying inertial properties of the cursor with respect to cursor position.

not present. Though subjects were able to position the cursor with smaller errors using skin stretch feedback, the true value of skin stretch feedback lies in its ability to convey a sense of motion. Subjects could easily detect whether the cursor was moving or had stopped and in combination with the motor commands through the force sensor input, users had a clear idea of the cursor direction. This perception of motion becomes even more important when control channels or dynamics of a system are complex. In this study, the dynamics of the system were complex to make it more difficult for user to simply use open-loop strategies to position the cursor. In systems where the dynamics are simple, skin stretch feedback may not provide a good cost to benefit ratio. However, in systems such as a prosthetic arm, the dynamics can be difficult to predict and the control channels can be noisy. Myoelectrically controlled prostheses, which use electrical impulses produced from muscle contractions to control high-impedance prosthetic joint actuators, may contain noisy signals. Coupled with friction and the changing inertia and dynamics of the arm itself make it a difficult

system to control. It is this type of application in which skin stretch may provide to be a very effective method of feedback. This is explored in the following chapter.

Although vibrotactile feedback was provided in these experiments and found to be less effective in positioning accuracy than skin stretch, it must be noted that it still remains an attractive choice for wearable haptic applications due to its size and power characteristics. In the present study, the amplitude of the vibration was continuously varied according to cursor position as this was found to be the most effective continuous mapping. However, other strategies could be employed where discrete changes in amplitude or frequency occur at specific intervals. This would require the subject to "count" to some extent to perform movements and the inherent resolution of the feedback channel would be limited. In any case, when vibration is used, it is desirable only to turn on the stimulation when a movement is being made as many subjects reported that continuous vibration was annoying. In addition, neural adaptation causes desensitization over time [10, 55]. Multiple vibration stimulators could also convey a sense of motion if they are appropriately sequenced [104]. However the stimulators must be spaced relatively far apart due to the large receptive fields of the deep Pacinian corpuscles. Because skin stretch activates primarily superficial, slow acting receptors, it is less susceptible to these issues.

Although these initial studies indicate there are unexplored benefits to using skin stretch for tactile feedback, there are a few practical issues that must be addressed when attempting to implement skin stretch. Particularly if the skin contact area is relatively small, as with the device used in this study, care must be taken so that the device does not slip on the skin. How well the device works is highly dependent on the where it is placed on the skin and on the subject. The device tends to slip more if there is a lot of hair on the skin or in areas where skin curvature is high. Though the Red-E-Tape was found to adhere adequately to the skin, it is assumed that some amount of slip occurred throughout the skin stretch trials. The varying stiffness of skin on different parts of the body, as well as subject-to-subject variation of skin properties, present significant design and control challenges. Some individual calibration may be

required when the device is attached. Because skin stiffness properties also depend on the configuration of the body or limbs, the perceived magnitude of stretch also appears to change. In early pilot experiments, we attempted to place the skin stretch applicator near the elbow joint to provide something closer to an illusory sense of joint movement. However, we observed that as the trials progressed, subjects would move and bend their elbows slightly to re-adjust their seating configuration. Using the wearable device would be more comfortable for users and we can infer from these results and previous studies that the results from a study such as this would be similar with the wearable device.

The present study did not fully assess the effects of training on movement accuracy. Because subjects were given post-trial vision feedback throughout the experiment, some improvement over time was expected. In fact, most subjects had lower errors in the second no feedback trial than the first but no significant improvement was found across the 36 trials for a given feedback case. All subjects were given uniform training in this study. To fully assess learning effects, a future study should be performed where the training method is a controlled variable.

Overall, skin stretch feedback was found to provide effective feedback. It cannot be used to replace vision, yet it can be used to reduce the dependence on vision by providing a relative sense of position and motion. The following chapter builds on this idea by applying this type of feedback to a more realistic application and assessing the visual demand for another task with and without feedback.

Chapter 5

Artificial Proprioception for Myoelectric Prostheses

5.1 Introduction

Recent advances in engineering and medicine have provided more sophisticated upper-limb prostheses as well as improved means of controlling them. However, most upper-limb prostheses lack direct sensory feedback. While still not approaching the richness of human sensory systems, the current state of technology is such that prostheses could be equipped with a variety of sensors to measure position and motion, forces, temperature and vibration. A more challenging problem, however, is to relay this information back to the amputee. Promising research is being done in the field of neural prosthetics that may allow direct stimulation of neurons in the central or peripheral nervous system (e.g. [28,86]). However, these approaches are invasive, and more research and development will be needed before this becomes a viable option.

Wearable haptic displays can provide a non-invasive approach to communicating information from sensors in a prosthetic limb to the amputee. As early as the 1960's, a number of researchers evaluated tactile displays to provide users a sense of grip force in myoelectric hands. These included cutaneous [9,98,100] and implanted [4,23]

single electrode displays as well as vibrotactile displays [115]. Mann [79] implemented a vibrotactile display of position proprioception in a myoelectric elbow joint. The prevailing opinion at this time was that the feedback would be beneficial if not critical for the advancement of myoelectric prostheses. While these studies found benefits to the feedback, it was not widely adopted clinically. It is not clear why amputees and clinicians did not respond to this concept, but there are a number of potential contributing factors. First, the feedback modalities used can be subjectively unpleasant [64, 99]. There may also have been some aversion to adding complexity in the prosthesis without providing a more significant functional benefit. Most traditional myoelectric prostheses have high-impedance actuators in the joints which, coupled with the noise in the myoelectric signal (MES), may limit the user's ability to control force even if they have some afferent feedback. Amputees also have some limited sense of force in the prosthesis through vision (deformation) and reaction forces at the socket interface. Recently, as more sophisticated arms and hands are being created, haptic force feedback is again being evaluated, and may be better accepted clinically as amputees adopt more technologically advanced prostheses. Cipriani et al. implemented vibration feedback of grip force in an anthropomorphic hand that was myoelectrically controlled and found qualitative benefits in unimpaired subjects [22].

Less work has focused on haptic proprioceptive feedback for prostheses. In the absence of proprioception, amputees can rely on vision to position the arm. When visual feedback is not available, users can potentially rely on internal models of the arm and controller to perform some tasks with open-loop strategies. This has led some to question the value of proprioceptive feedback in prostheses (e.g. see [54]). As discussed in Chapters 1 and 2, the role of proprioceptive feedback in human motor control is not completely understood. The substantial delays associated with the feedback (50-250ms, depending on the modality [68]) indicate that it is likely not being used for real-time control for most motions at preferred speeds. Many motions seem to be performed in feed-forward fashion based on an internal model of the body [68]. How this idea translates to myoelectric prostheses, where arm dynamics

and control parameters are quite different (along with typical movement speeds) is also not well understood. However, there is evidence suggesting that feedback would be useful. Proprioceptive or visual feedback could be used to construct an internal model of the system, which could improve long-term performance. Visual servoing consumes a large portion of the user's visual capacity that could be devoted to other simultaneous tasks. High visual demand has been cited as a drawback of myoelectric arms by amputees [7]. Additionally, some benefits of proprioception have been shown even when vision is present [70]. The addition of proprioceptive feedback creates a closed-loop system that may make the prosthesis a more intuitive extension of the user. This is evidenced by the fact that body-controlled prostheses (including some with external power, e.g. [2, 19, 20, 53, 110]) have proven popular in part due to the fact that they take advantage of the principle of extended physiological proprioception (EPP) as first described by Simpson [102]. Myoelectric prostheses can provide more intuitive and less cumbersome control schemes but lack this element of feedback, which some amputees have cited as a reason for rejection of the prosthesis [101]. In the present studies, we aim to show quantitative benefits of proprioceptive haptic feedback in blind targeting accuracy and a reduction in visual demand for a continuous movement task. The visual demand measurement is an attempt to quantify how often a user must glance at the arm in order to complete the task.

In the study presented in the previous chapter, we found it necessary to give the virtual object non-intuitive dynamics in order for the feedback to be beneficial compared to feed-forward motions. For myoelectric prostheses, we hypothesize that the inconsistent and noisy characteristics of the MES, coupled with the nonlinear joint dynamics (due to static friction), will make open-loop strategies less effective and the haptic display more beneficial.

In the following sections, we evaluate the wearable skin stretch device with unimpaired subjects controlling a virtual prosthetic elbow joint with EMG electrodes on the bicep and tricep muscles. The experimental setup approximates a transhumeral

amputation as the elbow joint is effectively fixed by having participants hold a stationary handle with their hand. We use a blind targeting task to compare their accuracy with the haptic device to the cases where no feedback is provided (open-loop movements) and where proprioception is provided on the contralateral arm. These two comparison cases establish what we consider the worst (no feedback) and best (contralateral proprioception) afferent cases for blind targeting while maintaining the salient efferent characteristics of myoelectric control. In addition, we compare the visual demand required to roughly position the virtual elbow continuously in a visual occlusion study with skin stretch compared to no feedback.

5.2 Methods

In this section we describe the control of the virtual arm, including EMG data processing steps as well as the control of the feedback devices. Finally, we describe the calibration procedure and the experimental protocol for the two user studies. The wearable skin stretch device was used for these studies.

5.2.1 Control System

We used a simple controller that captures the salient characteristics of many myoelectric prostheses and was easy for the subjects to learn to use in the present work. Figure 5.1 shows a schematic representation of the primary control elements of the experimental setup. The control elements can be divided into three sub-sections; EMG data collection and processing, virtual prosthetic arm dynamics, and control of the skin stretch device. Each of these is discussed in the following subsections. Matlab's xPC real-time target was used to perform the data collection and closed loop control at 1 kHz.

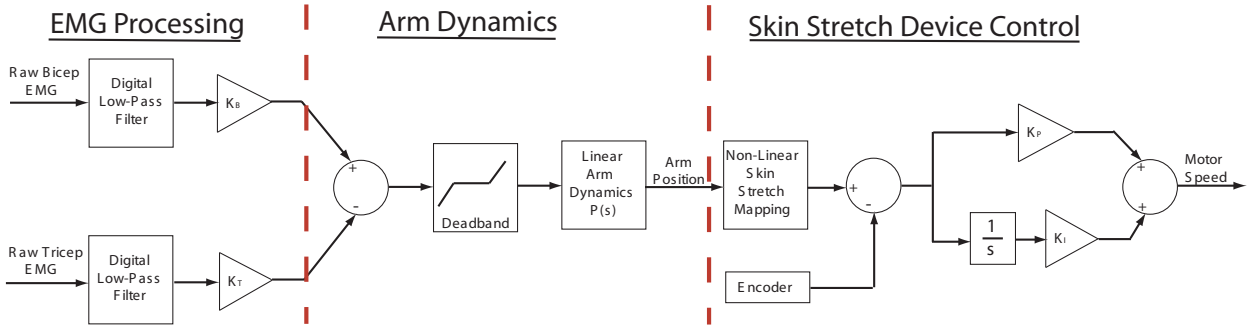

Figure 5.1: Schematic representation of the primary control elements in the system.

EMG Data Processing

Differential EMG electrodes with pre-amplification (Delsys 2.1) were used to collect the MES from the bicep and tricep muscles. A second stage handheld amplifier with gain of 1000 (Delsys Bagnoli 2) was used for further signal amplification. Each EMG channel was sampled at 1 kHz, the DC bias was subtracted (found during an initial calibration procedure described below), and a low-pass Butterworth filter with a cut-off frequency of 6 Hz was applied. Gains were then applied to each channel before sending the signal to the virtual arm. These gains varied between subjects and were determined as part of the calibration procedure.

Virtual Prosthetic Arm Dynamics

After the gains were applied to the filtered MES, the tricep signal was subtracted from the bicep signal to create a virtual torque that was sent to the arm. This virtual torque is analogous to the current sent to the motor in a prosthetic limb (multiplied by the torque constant of the motor). A deadband was used to simulate static friction in the arm which required a minimum torque of 0.5 units to be applied to the arm for it to move. The torque was saturated at 5 units. Units are defined such that one unit of torque produces an angular acceleration of one rad/sec^2 when applied to one unit of inertia (in the absence of friction).

After applying the non-linear elements, the torque was sent to the linear dynamics

of the virtual arm, which were modeled as a simple inertia and viscous damping such that the transfer function of arm position θ to torque T is:

$$\frac{\theta(s)}{T(s)} = \frac{1}{Js^2 + bs} \tag{5.1}$$

where J is the inertia of the arm and b is the viscous damping. In the present case we used values of $J = 0.5$ and $b = 2.5$. The free integrator in the denominator of the transfer function indicates that the system has no stiffness and the arm does not return to an equilibrium point when no torque is applied. On the virtual display, an arm angle of zero was represented by a horizontal line. The arm position was limited to $\pm$ 60 degrees, corresponding to a total of 120 degrees of elbow flexion.

Skin Stretch Device Control

For the experimental trials where skin stretch feedback was provided, our intent was to provide the user with an absolute sense of the virtual arm's position. When the arm's position was zero (horizontal), zero skin stretch was provided, and when it was at its limits (plus or minus 60 degrees), the skin stretch device was also rotated to its maximum value, which was limited to $\pm$ 40 degrees in this case. However, in prior studies we found that a linear mapping of arm position to skin stretch resulted in difficulty in detecting positions around zero. We therefore implemented a nonlinear mapping by fitting a fifth order polynomial that was constrained to pass through zero and the endpoints, had a slope of 1 near the origin and a slope of 1/2 near the limits (Figure 5.2). While more work remains to determine an optimal mapping, pilot subjects were better able to use the nonlinear mapping than a linear one in closed-loop targeting studies. Any future improvements in the device design and mapping should improve performance compared to the results presented here.

After the desired position of the skin stretch device was determined by evaluating the polynomial in Figure 4.5 at the current virtual arm position, it was compared to the encoder reading of the device to calculate the position error. A proportional

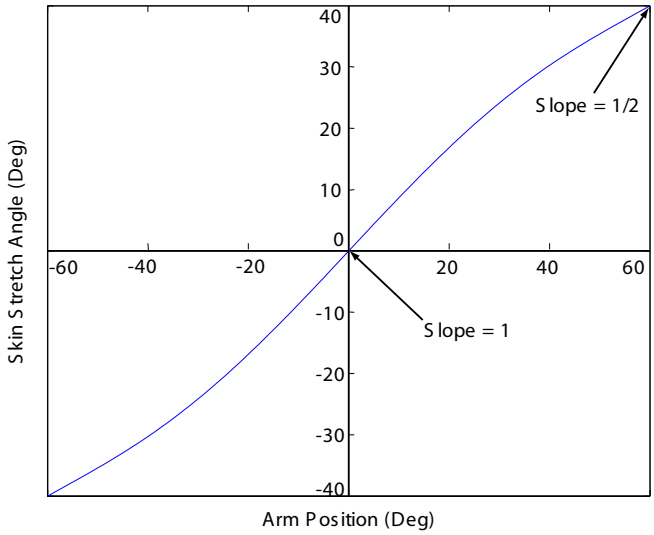

Figure 5.2: Non-linear mapping of virtual arm angle to skin stretch rotation. Mapping is a fifth-order polynomial with a slope of one near the origin and 1/2 near the limits.

plus integral (PI) controller was then used to set the speed of the motor based on this position error. The proportional gain (K_p) was 0.75 and the integral gain (K_i) was 0.1. A small driver board provided by the manufacturer (Shinsei Motors, D6030) was used to control the motor. In addition to the analog speed channel, which was specified as described above, two digital lines were used to control the direction of rotation. Due to these direction changes and the fact that the motor has a minimum speed at which it can rotate determined by the resonance of the piezo elements, some chatter around the desired position was sometimes encountered when the arm moved very slowly. We therefore implemented a deadband of 0.1 degrees on the position error. The position error due to the controller was always less than 0.8 degrees.

5.2.2 Experimental Setup

Setup Overview

The experimental setup is shown in Figure 5.3. The participant was seated in front of a computer display. EMG electrodes were placed on the right bicep and tricep

muscles near the center of the muscle body with the differential electrodes aligned approximately with the fiber direction. A reference electrode was placed on the back of the right hand. The wearable skin stretch device was placed on the back of the right upper arm, proximal to the elbow, near the tricep electrode. The participant gripped a handle with their right hand. A button was placed on the handle so it could be easily pushed with the thumb. The setup was designed to simulate a transhumeral amputation as the elbow joint was effectively fixed. Participants could produce the MES by pulling up or pushing down on the handle. Pulling up approximates the bicep acting in isolation, and pushing down approximates the tricep in isolation.

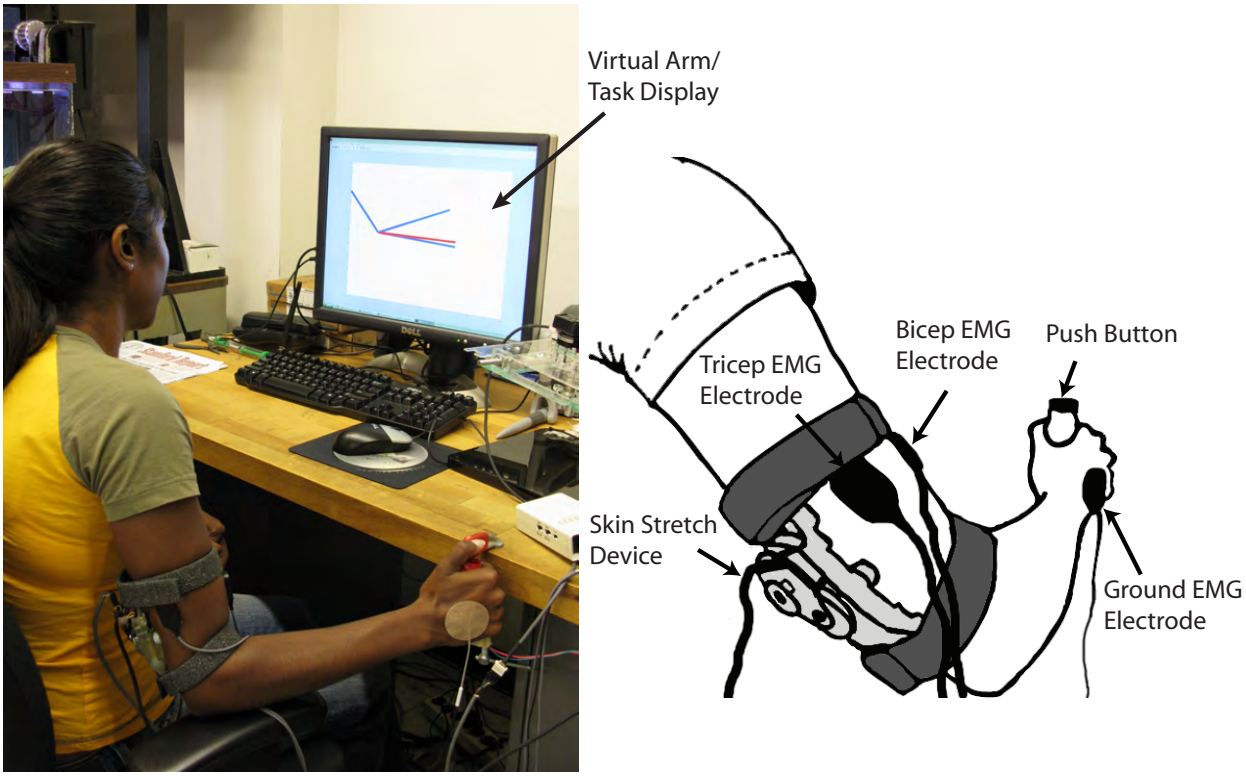

Figure 5.3: Experimental setup. Right pane shows schematic view of the primary input and feedback elements.

In the no feedback case, the skin stretch device was turned off. In trials where contralateral proprioception was provided, the left arm of the participant was placed

in a single degree of freedom manipulandum that could position the elbow joint at a desired angle in the horizontal plane. The elbow angle was set to match the angle of the virtual arm with zero degrees of elbow flexion corresponding to minus 60 degrees of rotation of the virtual arm, and 120 degrees of elbow flexion corresponding to 60 degrees of rotation of the virtual arm (Figure 5.4). The elbow manipulandum used a DC motor (Maxon RE025) with a 13.5:1 capstan cable drive transmission. A PID current controller on the desired elbow angle was implemented that produced dynamic tracking errors less than 2.0 degrees and static position errors of less than 0.21 degrees in all trials. A screen was placed above the manipulandum so the arm motion was not visible to the subject.

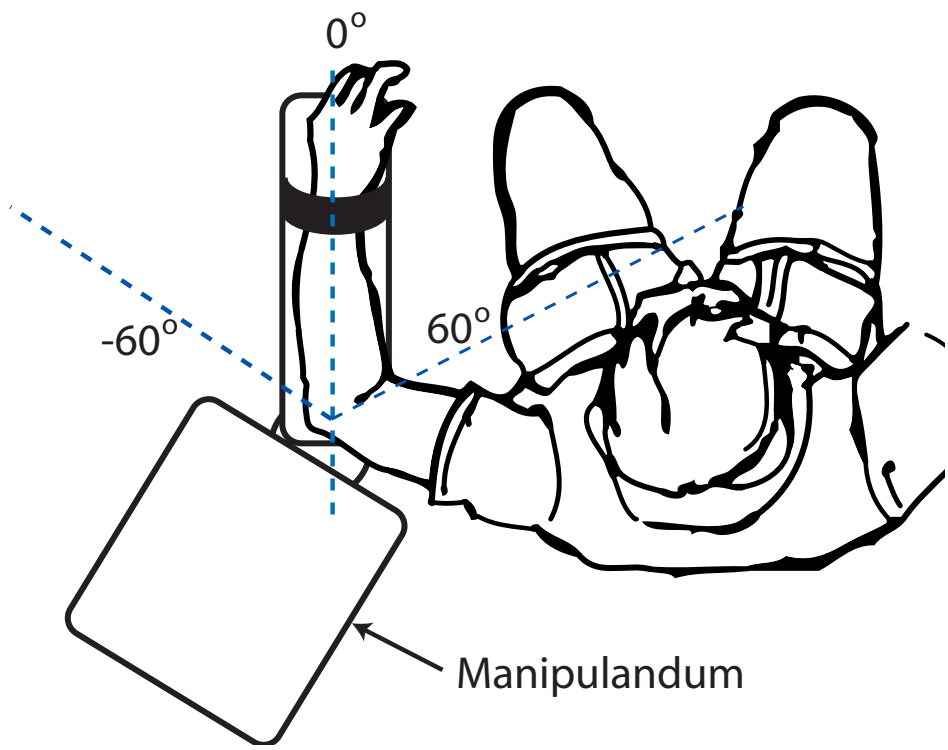

Figure 5.4: Schematic view of elbow manipulandum used to provide contralateral proprioceptive feedback.

Calibration

Prior to the experiment, the participants underwent two phases of calibration. First, they were asked to grip the handle but otherwise relax the arm. EMG activity from both muscles was recorded for five seconds, and the mean of this resting data was subtracted from each channel during later trials. Second, the participant was asked to pull up and push down on the handle while EMG signals were recorded. They were not instructed to use maximal force but a large force that would be comfortable to apply during the experiment. The respective gains for the bicep and tricep muscles were calculated by dividing the maximum filtered EMG signal recorded during the calibration by a constant, in this case 5. The participants were then allowed to move the virtual arm on the screen with visual feedback to determine if the calibration was satisfactory. The calibration was considered satisfactory if the participant had some control of the speed of the arm in both directions and approximately equal effort was required to move the arm in both directions. The calibration was repeated as necessary until the participant was satisfied with the controller. This took no longer than four attempts. In a few cases during the experiment, some drift was encountered due to a change in the DC bias. In such cases the calibration was repeated as needed.

Participants

Fifteen volunteers participated in the studies. Nine were male, six female, ages 19-49, mean age 27.3. Nine of the fifteen participants (6 male, 3 female) completed the targeting trial with contralateral proprioception. These trials were added to the study after six participants had already completed the skin stretch and no feedback trials in an attempt to provide a best-case baseline for performance. The studies took a total of 45-120 minutes to complete, including setup and calibration. The study was approved by Stanford's Institutional Review Board.

5.2.3 Targeting Study

The subjects first participated in a targeting accuracy study where no real-time vision feedback was provided. Prior to each set of trials a brief training period was provided. Each participant was allowed to move the arm through the workspace with visual feedback for two minutes and then was given 10 practice trials, identical to the experimental trials. In the no feedback case, the training allowed them to learn open-loop strategies, and in the haptic and contralateral proprioception feedback cases it allowed them to learn the mapping of arm position to feedback sensations. In each experimental trial, the starting position of the arm was displayed on the screen chosen from [-60,0,60] degrees. An instruction to move ± 30, 60 or 90 degrees was also displayed. Movements were constrained such the desired ending position was either -30, 0 or 30 degrees. This resulted in a total of eight starting/ending combinations, each of which was repeated four times in random order for a total of 32 trials per participant. A sample starting screen is shown in Figure 5.5.

After this screen appeared, the participant moved the virtual arm without seeing the motion of the arm on the screen. The participant was instructed to press the button when they felt the target was reached. After the trial, the participant was shown the actual and desired ending position of the arm. The targeting error was recorded for each trial, i.e. the absolute value of the difference between the desired and actual ending position. The 32 trials were performed with both skin stretch feedback and no feedback. Seven participants did the no feedback trials first; eight did the skin stretch trials first. The nine participants who completed the trials with contralateral proprioception did these last.

5.2.4 Visual Occlusion Study

In the second study we assessed participants' visual demand with and without the skin stretch device using the visual occlusion method [48, 105]. In this case, the task was to keep the virtual arm in a desired range that moved in a quasi-random fashion.

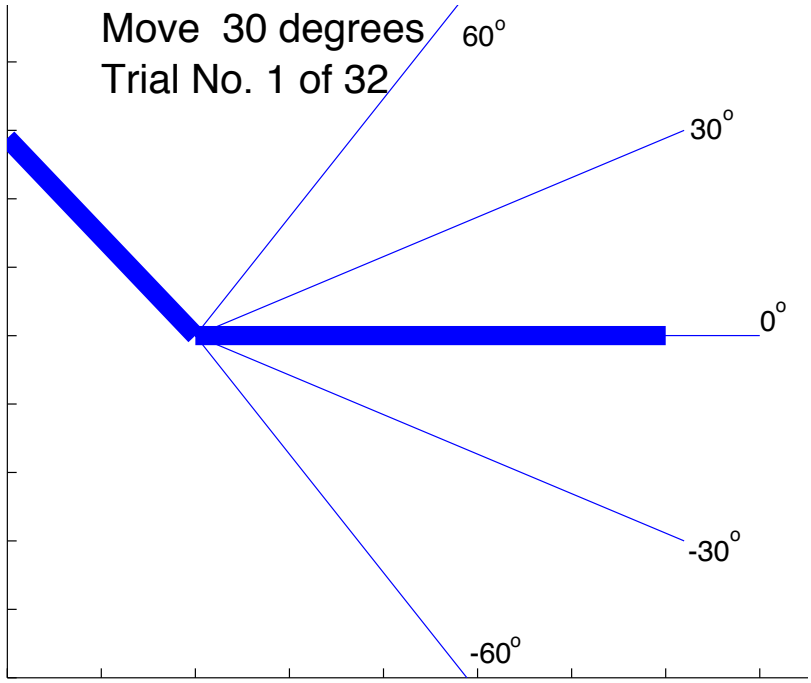

Figure 5.5: Visual display of virtual arm at beginning of a sample trial with starting position equal to zero degrees. The thick diagonal line represents the upper arm and does not move.

Specifically, the participant was instructed to keep the virtual arm in a 30 degree window, displayed on the screen as two lines that moved together (thick outer lines in Figure 5.6).

The desired arm position range was generated by creating a desired position and drawing lines (limits) at $\pm$ 15 degrees from this position. The desired position was determined in pilot trials such that the task was relatively easy with constant visual feedback and very difficult without. A sum of 11 sine waves that appeared random to the participants was chosen with amplitudes and frequencies based on guidelines for human tracking in [93] and [44]. The frequencies of the sine waves (ω_k) were relative prime multiples of the fundamental frequency 0.003 Hz using the prime multipliers 2, 3, 5, 7, 11, 17, 23, 37, 59, 87, and 131. The amplitudes (A_k) were defined with an exponential function of frequency as follows

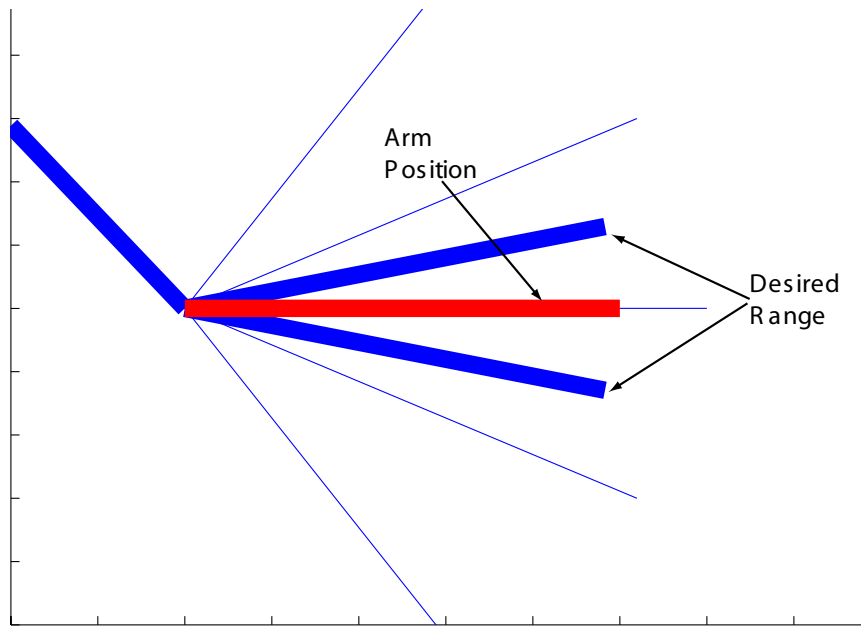

Figure 5.6: Visual display of virtual arm at beginning of visual occlusion trial. Thick outer lines represent the desired range, thick horizontal line in the center is the virtual arm which is only visible for one second after a button press.

$$A_k = 0.3 \sum_{k=1}^{11} e^{-0.12(k-1)} \tag{5.2}$$

The scaling (0.3) and decay rate (0.12) values were determined such that sufficient coverage of the workspace was generated while not exceeding the range. The phase angles of the sine waves were randomized before each trial to prevent the participants from memorizing the inputs. A sample of the desired range and the virtual arm's position for a successful trial are shown in Figure 5.7.

Visual feedback of the virtual arm's position was provided for one second each time the participant pressed the button on the handle. Participants were instructed to press the button whenever they felt visual feedback was required to complete the task. If the arm left the desired range, the trial ended and restarted. The trial was successful if the participant kept the arm in the desired range for the duration of the trial (40 seconds). Trials were repeated until the participant had three successful

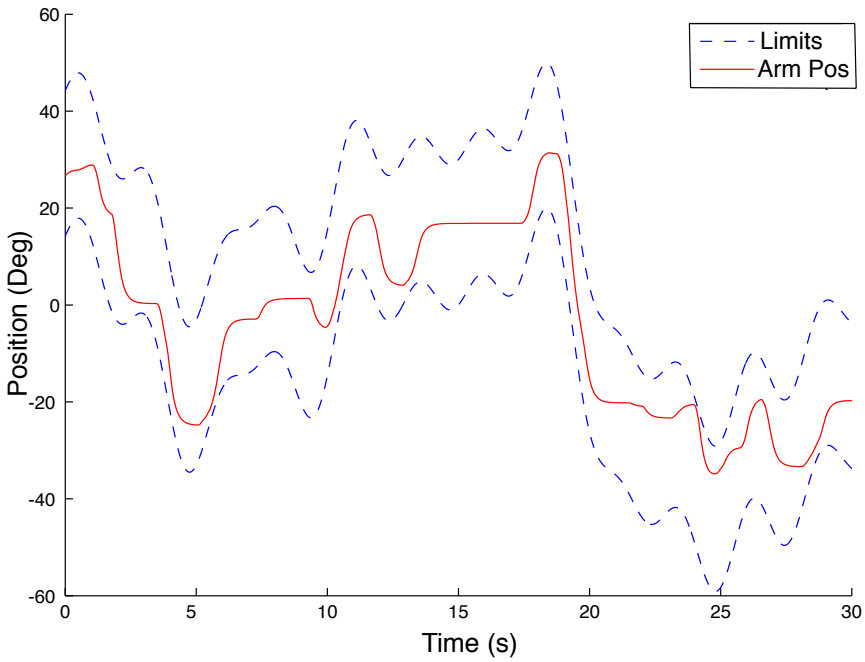

Figure 5.7: Virtual arm position and task position limits for 30 seconds of a sample trial with skin stretch feedback.

trials for the no feedback and skin stretch cases. The order of the trials (no feedback and skin stretch) was randomized. Prior to the experiment, participants were given one full practice trial with constant vision feedback for each feedback case. Over the course of the trials (including failures) participants were able to determine an appropriate strategy of button press frequency that allowed them to complete the task.

For each successful trial the average visual demand was calculated. The instantaneous visual demand (VisDem) can be calculated at each button press using the following relation

$$VisDem = \frac{1.0}{t_i - t_{i-1}} \tag{5.3}$$

where the numerator is the duration of the feedback (1 sec.) and t_i and t_{i-1} are the times corresponding to the current and previous button presses. Visual demand

calculations for a sample no feedback and skin stretch trial are shown in Figure 5.8. The average visual demand is calculated by averaging these values over the trial or equivalently by dividing the number of button presses by the duration of the trial. The average visual demand is the fraction of time that visual demand was provided over the course of the trial. It is a measure of how often the participant felt they needed to glimpse at the arm to complete the task. In the data analysis, the average over all three successful trials was used to compare the haptic and no feedback cases. As a secondary measure, the number of task failures before completing the three trials was also recorded.

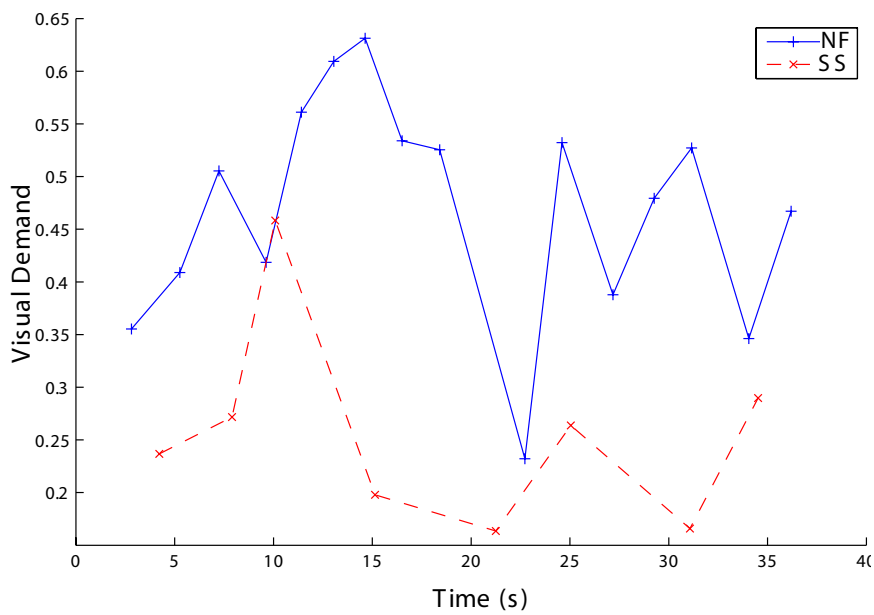

Figure 5.8: Visual demand calculations at each button press for a sample no feedback (NF) and skin stretch (SS) trial.

5.3 Results

5.3.1 Targeting Study

In the targeting study, the primary measure of interest is the average targeting error (across the 32 trials). A box plot of the average position errors for each of the participants is shown in Figure 5.9. Repeated measure analysis of variance (ANOVA) with pairwise comparisons was performed for this performance measure. For all 15 subjects a repeated measure ANOVA with two levels (no feedback and skin stretch) was performed. Significant main effects were found due to both feedback condition, $F(1, 14) = 36$, $p = 3.3 \cdot 10^{-5}$ and participant $F(14, 14) = 544$, $p = 1.3 \cdot 10^{-12}$. Errors with skin stretch were significantly lower compared to errors with no feedback with $p = 3.3 \cdot 10^{-5}$. For the nine subjects who participated in the contralateral proprioception feedback case a separate repeated measure ANOVA with three levels was performed. Significant main effects due to feedback condition, $F(2, 8) = 29$, $p = 6.6 \cdot 10^{-4}$, and participant $F(8, 8) = 57$, $p = 6.8 \cdot 10^{-5}$ were again found. Post-hoc pairwise comparisons with Bonferroni correction found significant differences between all feedback cases with skin stretch lower than no feedback with $p = 0.015$ and contralateral proprioception lower than skin stretch with $p = 2.9 \cdot 10^{-3}$.

While variations between participants were significant, trends were consistent across participants as all 15 had lower average errors with skin stretch than no feedback, and all nine participants who completed the contralateral proprioception trial had lowest errors in that case.

In order to get a sense of how the participants were utilizing the feedback that was provided, we also examined the movement profiles. In the no feedback condition, participants typically moved in a single motion, with a single burst of EMG data. When feedback was provided (contralateral proprioception or skin stretch) we often observed submovements with delays of 0.4-1.0 seconds between them. A sample motion for each case is shown in Figure 5.10. In many cases, the submovements were not all in the same direction. Given the delays associated with the feedback

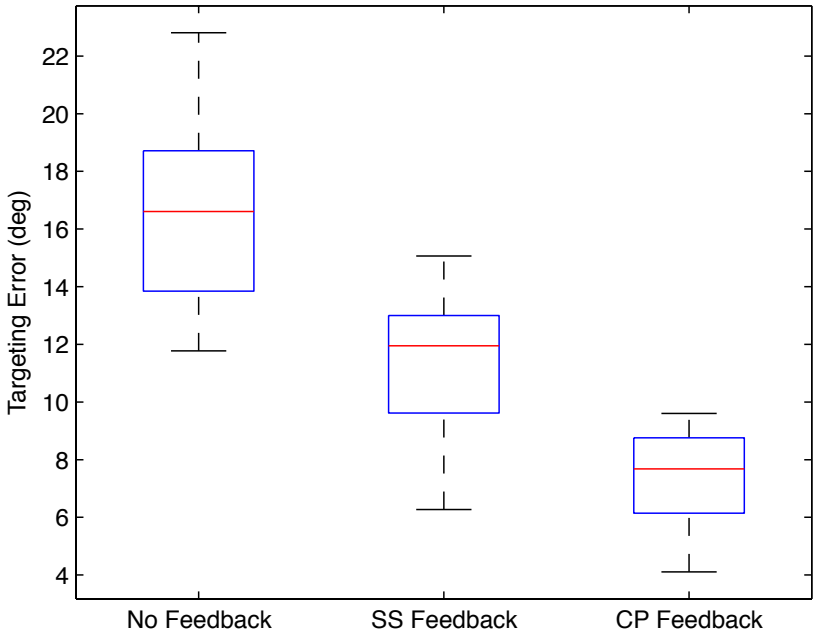

Figure 5.9: Box plot of average targeting errors for participants with no feedback and skin stretch (SS) feedback (n=15) and contralateral proprioceptive (CP) feedback (n=9) .

(probably around 100-200 ms) it seems that subjects would move and then wait to get a sensation of the position before deciding whether to make another movement to correct for error. The number of submovements was calculated by differentiating the arm position and looking for peaks in this velocity profile (with a minimum temporal separation of 0.3 seconds). With no feedback, participants had an average of 1.2 submovements per trial. With contralateral proprioceptive feedback, an average of 2.8 submovements was found. With skin stretch, the result was again between these extremes with an average of 2.1 submovements per trial. These differences were significant with p-values below 0.01.

5.3.2 Visual Occlusion Study

The primary measure in the visual demand study was the average visual demand over the three successful trials for each feedback condition. A box plot of these

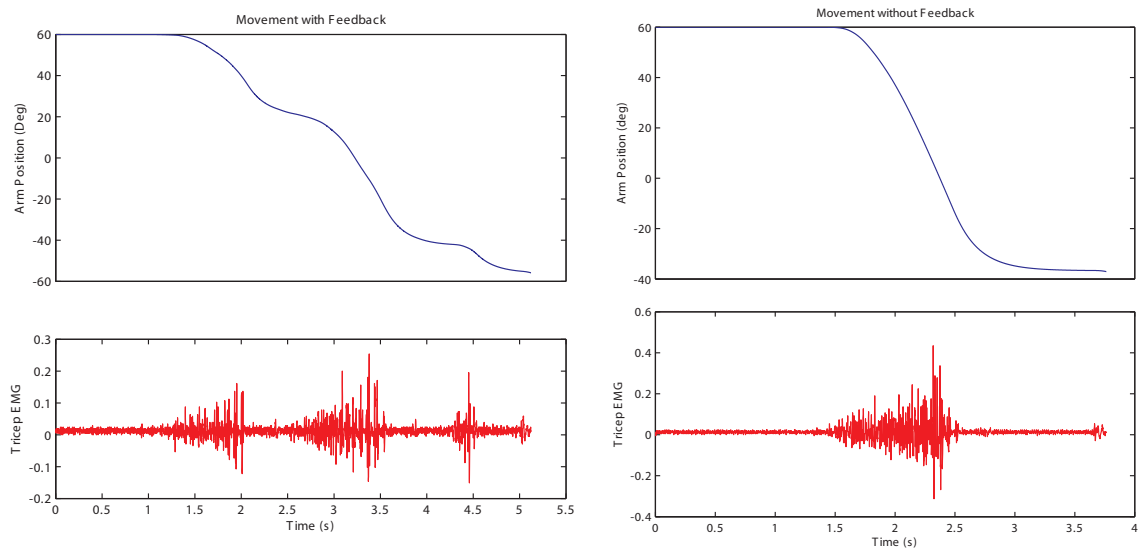

Figure 5.10: Sample of a movement profile for one subject with skin stretch feedback (left) and no feedback (right). Three submovements are evident in the case with feedback. The desired target was -60 degrees in both cases.

data is shown in Figure 5.11. While there was significant participant-to-participant variation in the visual demand in both feedback cases, participants generally had lower visual demand with the haptic device (Figure 5.12). With skin stretch feedback, the average visual demand was reduced 23 percent (from 0.53 with no feedback, to 0.41). A repeated measure ANOVA was performed for average visual demand over the three trials. Significant main effects were found due to both feedback condition, $F(1, 14) = 31$, $p = 7.2 \cdot 10^{-5}$ and participant $F(14, 14) = 184$, $p = 1.9 \cdot 10^{-9}$. Visual demand was significantly lower with skin stretch compared to the no feedback case with $p = 7.2 \cdot 10^{-5}$.

Participants also had fewer failures before completing the three trials with skin stretch feedback. The mean number of failures before completing the no feedback trials was 12.2 (range 5-24), and with skin stretch the mean was 8.2 (range 2-21). The difference in means was significant in a paired t-test with $p < 1 \cdot 10^{-6}$.

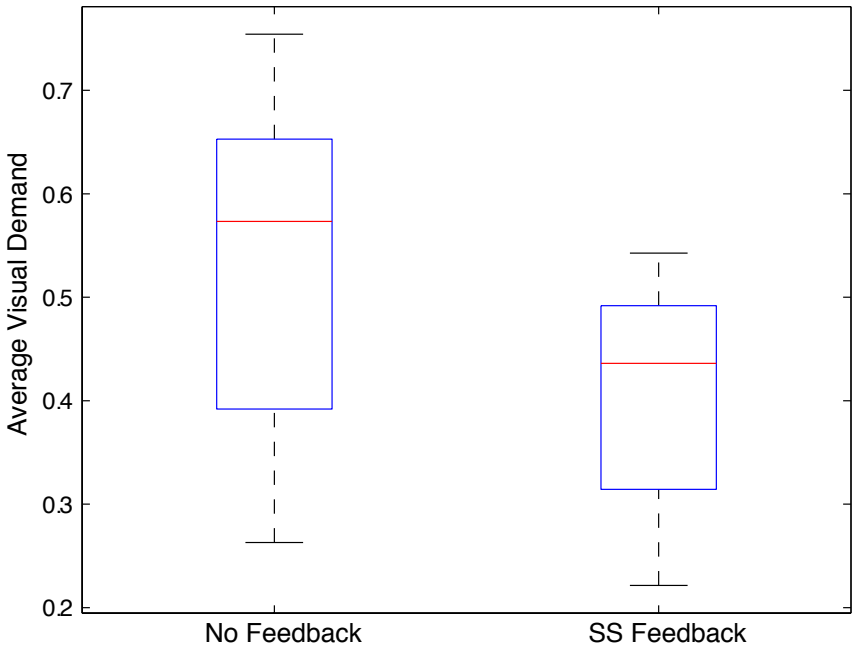

Figure 5.11: Box plot of average visual demand over three successful trials for all 15 participants with no feedback and skin stretch (SS) feedback.

5.4 Discussion

Participants performed better in both user studies with the skin stretch device than with no feedback. This indicates that the device may have value as a proprioceptive feedback device in myoelectric prostheses. When no feedback was provided, participants were forced to rely on open-loop strategies for controlling the arm's position. Presumably due to the non-linearities in the virtual arm model and the noisy nature of the EMG signal, this was difficult, making even a relatively coarse sense of position feedback useful. Targeting errors with skin stretch were approaching, but still significantly larger than errors with proprioceptive feedback from the contralateral arm, which we consider to be the best case scenario for blind targeting as it relies on intact physiological sensors while maintaining the other aspects of the experiment.

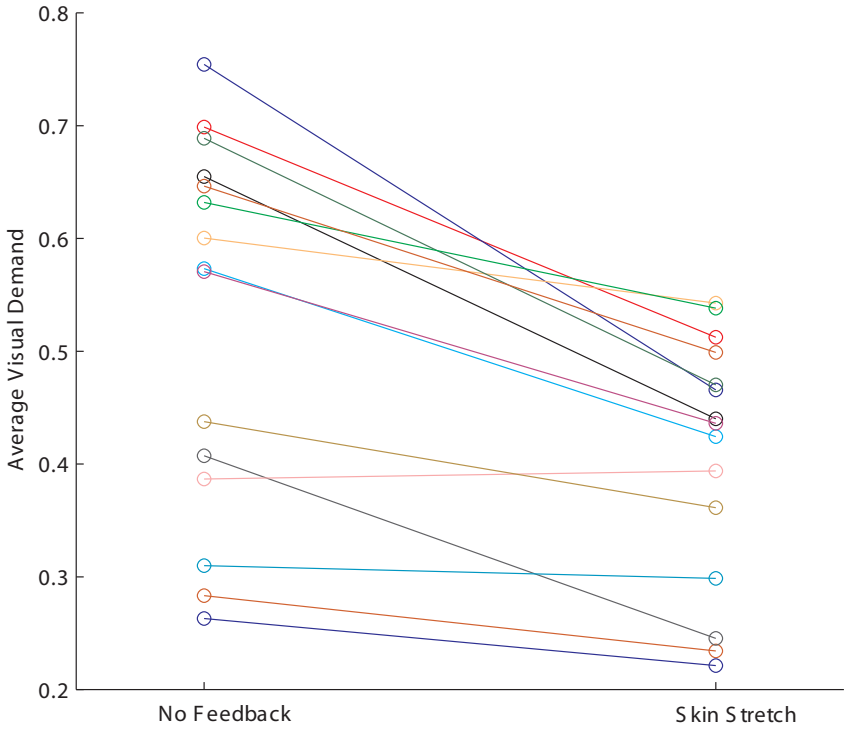

Figure 5.12: Average visual demand for each of the 15 subjects with no feedback and skin stretch.

In addition to reducing targeting errors, skin stretch feedback reduced the visual demand required to track a moving angular window. The lower visual demand indicates that providing a rough sense of position could allow the arm to be used (if position accuracy is not critical) while vision is devoted elsewhere, making the prosthetic arm a more natural extension of the person. The addition of haptic feedback could effectively close the control loop and require only occasional glances from the user, reducing the need for visual attention.

In both studies, only a short training period was provided. Additional training may improve performance with no feedback and with the haptic device; however, we expect minimal training benefit in the contralateral proprioception case. Though we did not directly assess the effects of training as uniform training was provided

in all cases, a simple regression analysis was performed to determine if subjects' performance improved over the time. In the targeting study, we fit a regression line to the absolute value of the errors versus trial number across all participants and found a significant negative slope for the skin stretch case ($p = 0.034$), providing evidence that errors decreased as the trial progressed. However, when no feedback is provided, the slope is not significantly different from zero. Individually, four of fifteen participants had significant negative slopes with skin stretch while two had significant slopes with no feedback. While future studies are needed to evaluate the effects of training, this indicates that participants could improve with more training, particularly with the haptic device.

While the device used in this study is wearable, it is not yet portable. We are developing devices that are smaller and lighter, and that can be worn on more parts of the body. We envision future devices that include on-board power, processing and wireless communication for portable use. All of the primary components of the device, including the motor, driving electronics, and sensors are amenable to portable use. The device consumes power only while changing its position. For powered prosthetic arms, it may be possible to integrate the device into the electronics of the limb, minimally increasing the size, weight and power consumption. Placement of the haptic device should be customizable depending on the wearer, as with most prosthetic components. In some cases, it may even be possible to integrate the device into the prosthetic socket directly, eliminating the need for straps. In the studies described here, in which the user wore the device for no more than 2 hours, negligible slipping between the device and the skin was observed. Future studies will evaluate whether the current attachment method is adequate for longer term use (i.e. a full day). Due to the relatively small amount of rotation applied in the study, subjects did not complain of skin irritation or fatigue despite nearly continuous use for up to 30 minutes at a time. While increasing the range of rotation may improve performance, it is unclear how this will affect comfort. We expect that each user will have a different preferred range.

This paper has presented a proof of principle for rotational skin stretch as a means of providing proprioceptive feedback in applications, such as prosthetic limbs, where it is otherwise absent. The results of the experiments indicate that rotational skin stretch can be used for closed-loop positioning in vision-impaired circumstances. Looking ahead, we acknowledge some differences between the experiments described in this paper and clinical applications, which may affect the performance of skin stretch device. First, in amputees both the MES and feedback properties may differ from unimpaired users. When the muscles on the residual limb are activated, more significant conformal changes may result, which could affect the performance of the skin stretch device in some cases. While subjects in this study did not have a gross sense of joint motion, it is possible that they had some low-level afferent feedback from tendons and other structures that amputees would lack. The effect of these factors is unclear and will be evaluated in future studies. Another difference between the present case and the proposed application is the virtual arm and controller dynamics. These were not specifically designed to match any specific prosthetic arm but were intentionally simplified so that participants could learn to use the virtual arm with minimal training. In reality, prosthetic arm dynamics would likely include more significant non-linearities, control irregularities, and exogenous disturbances that would make the arm more difficult to control in an open-loop fashion (which would increase targeting errors in the absence of feedback). We conjecture that the addition of proprioceptive feedback would be more useful when controlling an arm with more realistic dynamics. The fact that it was beneficial even with these relatively simple dynamics (while still maintaining the basic properties of the MES) is encouraging. There is also the potential for improving the quality of the feedback. The device configuration and mapping used in this study were effective but likely non-optimal. We are evaluating potential improvements in these areas that may increase performance.

Chapter 6

Conclusions

The user studies presented in this thesis showed benefits of wearable tactile feedback for a few applications. Those applications were carefully chosen based on an understanding of the inherent advantages and disadvantages of tactile feedback compared to other modalities. We also relied on insights about the role of proprioceptive feedback in motor control to guide appropriate use of the technology. Previous studies used for motion feedback often tested users in tasks that were unrealistically slow compared to the target applications [16, 75]. While the studies we presented showed feasibility of the approaches, more work should be done in order to create clinically beneficial systems. This chapter summarizes the main findings of this work and presents guidelines for future work.

6.1 Summary of Results

The primary contribution of this work is the evaluation of wearable tactile feedback for specific clinical applications. Related to motion training, we successfully trained subjects to modify their gait (a complex dynamic task) to reduce a potentially damaging knee joint load with both visual and vibration feedback. Skin stretch was also

evaluated for this task in pilot trials. However, as the feedback was largely discontinuous (updated on each step) and the information did not have a direction or motion associated with it, the benefits of skin stretch were not exploited in this case. We expect that performance would have been similar to vibration but there was not sufficient justification for the added complexity of the device. It was somewhat surprising that vibration had similar performance to visual feedback in many respects (with the exception of the longer training time). It seems that the ability of the subjects to control the knee force was limited enough that a high resolution feedback method was not critical. The subjects could lower the load on the average but could not adjust it with a high resolution. While vision is adequate for feedback in a laboratory setting, vibration feedback could be used in a portable system if sensors could be developed to calculate the knee joint loads. This would require a force (or perhaps pressure) sensor under the foot to estimate the ground reaction force and center of pressure as well as a method for estimating the relative location of the knee joint center. The clinical problem addressed by this work is a significant public health problem and could benefit greatly from non-invasive techniques such as this.

The other clinical issue addressed by this work involves feedback for prosthetic limbs. While a number of researchers have attempted to implement force feedback with tactile displays, very little work has focused on proprioceptive feedback. Arguments were made that this type of feedback was not needed as vision was adequate feedback or feed-forward strategies were effective (e.g. [54]). We tested both of these hypotheses (Chapter 5) and found that feedback was indeed beneficial when vision was not provided compared to feed-forward control for myoelectric systems. Additionally, we showed that the visual demand required to perform movements with coarse accuracy can be reduced with artificial proprioception. In Chapter 4, we showed benefits of artificial proprioception when controlling an external object with position-dependent dynamics.

In addition to these specific applications, we can also draw more general conclusions about the use of wearable tactile feedback for motion applications. For motion

training, the feedback is most likely to be beneficial when the task is very slow (which may be unrealistic) or when the task is repetitive. In repetitive tasks, feedback on each repetition is used to build effective feed-forward strategies for future movements. The use of tactile feedback should complement vision appropriately. It can be localized to present an intuitive sense at a particular joint for instance. For many degrees of freedom, 2D vision can be ambiguous. Tactile feedback is also appropriate for portable systems as it is not desirable to look at a display continuously. For replacement of proprioceptive feedback, the feedback seems most appropriate when feed-forward strategies are unreliable and vision is occupied elsewhere.

The use of skin stretch seems most appropriate when it is desired to create a sense of motion as well as position or magnitude and the information is inherently continuous. To reduce direction ambiguity, it seems most appropriate when used in a closed-loop system in which the subject has control over its movement. Vibration is attractive due to its small size and ease of application. They are easily arrayed to create a sense of localized feedback. They have been used extensively in commercial devices to convey exogenous event cues. In this work we also use them to provide intrinsic feedback based on the users' actions. For complex systems, a combination of tactile feedback types along with vision and audition may be an effective way to maximize communication bandwidth.

6.2 Open Questions and Future Work

The studies we presented here tested healthy subjects to show feasibility of the approaches. To establish clinical relevance, studies with patients should be carried out. For osteoarthritis a similar study to the one presented in Chapter 3 could be performed with patients. It should address retention of the new gait as well. Learning or retention may be improved by using adaptive feedback as well as some explicit suggestions (which could be conveyed with tactile displays as well). An immediate clinical impact could potentially be made with minimal modifications from the present study.

As discussed in Chapter 5, studies with amputees should be performed to establish the effectiveness of the feedback for clinical use. The study in that chapter used a virtual elbow joint with electrodes on bicep and tricep muscles for simplicity. However myoelectric hands are more common and often more difficult to control due to the fact that the muscles used to actuate them are smaller. The feedback may be even more beneficial in this case as the EMG signal characteristics are not as good. It may also be possible to provide a sense of grip force in a hand with the same device in a mode-switching paradigm.

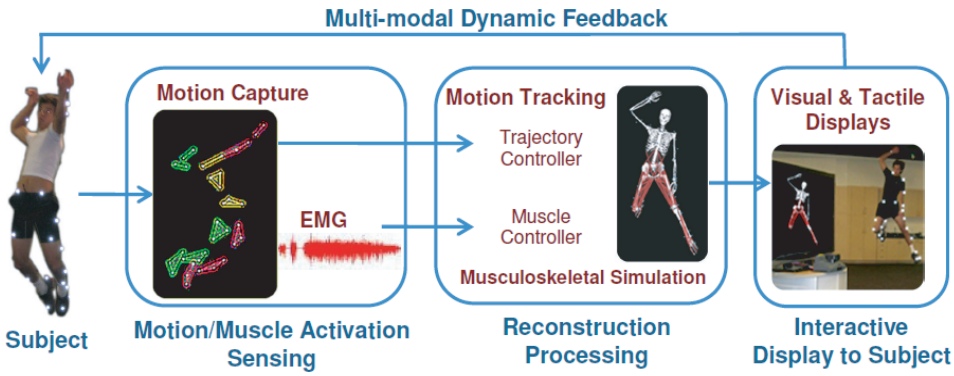

Figure 6.1: Depiction of a system with motion capture, musculoskeletal simulation and real-time feedback for motion guidance.

The studies presented here used a single channel of feedback although in some cases multiple joints contributed to the movements. How the performance will scale to multiple channels is an open question. Lieberman et al. [75] tested vibration feedback on five joints of the arm and found that the feedback they provided was most effective for hinge joints (i.e. elbow, wrist flex/extension and ad/abduction) and not as effective for rotational joints (i.e. forearm and upper arm rotation). This is likely due to the non-intuitive nature of vibration for these motions. Skin stretch, which as implemented in this work is rotational, could provide an effective feedback modality for such joints. An interesting future direction for this work is the measurement of human motion, real-time analysis of musculoskeletal parameters (e.g. joint loads, muscle forces) and real-time feedback (visual, audio or tactile) to guide motions (Figure 6.1).

Because this may involve complex information over many joints being conveyed to the user, studies should be performed to assess how to do this in a way that maximizes the bandwidth and intuitiveness of the feedback.

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