A Vibrotactile Display Design for the Feedback of External Prosthesis Sensory Information to the Amputee Wearer

by

Andrea W. Chew

Bachelor of Science,
Stanford University, 2003.

Submitted to the Program in Media Arts and Sciences,
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Abstract

This thesis documents the development of a vibrotactile display to be incorporated into a powered ankle-foot prosthesis. Although existing devices have addressed the need for tactile and proprioceptive feedback in external prostheses, there has not yet been an attempt to develop and clinically evaluate a comprehensive vibrotactile display and signaling schematic for use with an active myoelectric prosthesis.

The development and evaluation of two different hardware solutions are presented including an array of vibrating pancake motors embedded into the exterior of a carbon fiber prosthetic socket and an array of vibrating pancake motors embedded into a silicone socket liner. Three haptic mappings were designed based on previous work in psychophysics, haptics, and HCI. These schematics include a spatial discrimination pattern, an amplitude modulated pattern, and a gap detection pattern.

To assess the effectiveness of the system, lower-limb amputees were asked to learn the three haptic mappings and use the feedback system to control a virtual ankle to a desired ankle position using a physical knob interface. Results show an overall recognition rate of 85% for all three haptic mappings and error response averages ranging from 8.2 s to 11.6 s. The high recognition rates and lack of variance between the mappings suggest that the three vibration parameters of spatial discrimination, amplitude modulation, and gap detection may be successfully used to represent different ankle parameters. However, the overall successful integration of the vibrotactile display ultimately depends on the interaction between the components of the whole prosthetic system.

Thesis Advisor: Hugh Herr
Title: Associate Professor of Media Arts and Sciences

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1. Introduction

According to statistics collected between 1988 and 1996 there were approximately 1.2 million persons living in the United States with limb loss. These include the loss of toes, fingers, feet, hands, legs, and arms. The cause of amputation can be categorized into four causes: congenital, cancer, trauma, and vascular problems. Congenital and cancer patients respectively compose 0.7% and 0.9% of the amputee population. The majority of amputations are due to vascular disease which account for approximately 82% of these amputations, many of the cases associated with diabetes. These patients are typically older with decreased mobility and relatively inactive lifestyles. In contrast, trauma patients who make up 16% of the amputation cases are young, active, and athletic (Dillingham 2002).

After recovery from surgery and adjustment to daily life with their prostheses many of these younger patients seek out ways to return to their previous athletic activities. As a result, a host of specialized prostheses have been designed to mimic the human body during running and cycling. Along the same biomimetic technological trend, a number of computer controlled knees such as the Ossur Rheo Knee™ have been designed to alter damping properties depending on amputee gait, imitating the behavior of the healthy human knee. On the output end, EMG-controlled prosthetic arms have been introduced as an alternative to cosmetic and static hand/arm replacements, functional hook hands, and mechanical body-powered prostheses. Electromyography (EMG) is a medical technique that detects the electrical potential produced in the muscle during contraction. With myoelectric arms, surface electrodes are placed on the residual limb to detect muscle flexion as a signal to move the external prosthesis. Unfortunately, the limitations of EMG signal processing restrict the granularity of amputee control and limit the prosthesis to a binary mode of operation (open/closed). Additionally, the lack of sensory feedback which is essential in motor control further reduces the effectiveness of these myoelectric prostheses. This thesis proposes a feedback system to be incorporated into an EMG-controlled prosthetic system to close the sensorimotor control loop and improve performance and granularity of artificial limb movement.

2. A Powered Ankle-Foot Prosthesis

Figure 1: An existing commercially available ankle prosthesis mimics the power and stiffness properties of the natural human ankle at a fixed slow walking speed through a carbon fiber leaf spring structure.

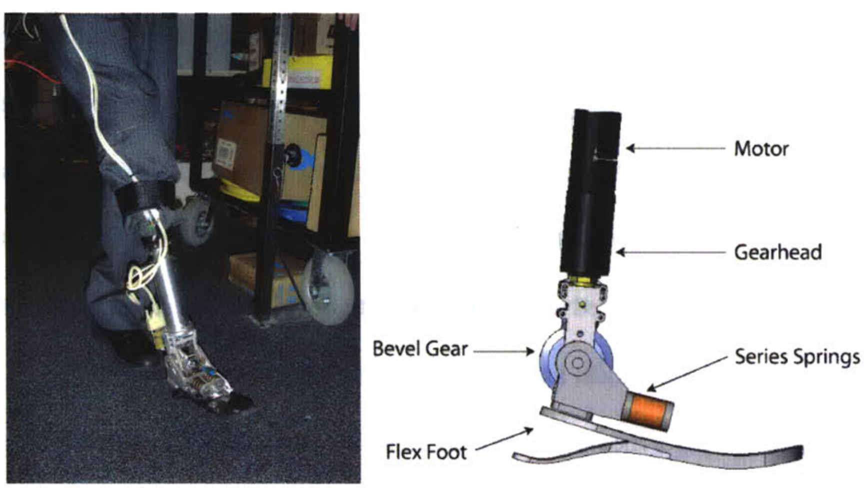

Figure 2: Using active mechanical components the Biomechatronics active-ankle prosthesis supplies the power and stiffness characteristics of the natural human ankle under dynamic walking conditions to restore gait speed and metabolism in leg amputees to near normal levels.

A team of researchers in the Biomechatronics Group at the MIT Media Lab are developing the world's first powered ankle-foot prosthesis capable of restoring gait speed and metabolism in leg amputees to near normal levels. In contrast to athletic prostheses, this device will address the needs of both the younger active amputees and the relatively larger population of less mobile older amputees. Human gait studies show that the ankle provides power and varies stiffness depending on walking speed and gait phase. Existing commercially available ankle prostheses are passive devices composed of carbon fiber leaf spring structures to mimic the ankle-foot, and possibly a shock absorbing device positioned between the ankle-foot and the human or prosthetic knee. The carbon fiber ankle-foot has been designed to supply the power and stiffness characteristics for a fixed walking speed in an undisturbed environment. With the development of the powered ankle-foot, we hope to design a dynamic biomimetic system that varies stiffness and power depending on walking velocity and environmental conditions (Au & Herr 2005). In parallel to the mechanical and control system design, the Biomechatronics Group is investigating EMG as a method of detecting user intention. With the control of joint impedance, motive power, and joint position in the active-ankle, a multidimensional sensory feedback system is necessary for the effective control of these additional features.

3. Feedback in Prostheses

3.1 Feedback Systems in Existing Prostheses

Previous studies have stated the necessity of incorporating non-visual feedback to improve the functionality of external prostheses. Researchers have proposed a range of alternative sensory feedback systems for use with artificial limbs. Cable tension and harness position in body-powered arm prostheses conveys information about device actuation through extended physiological proprioception (Light 2002, Bertels 2003). For this reason, many users have chosen these mechanically-actuated systems over myoelectric arms. OttoBock offers a commercial hand called the SensorHand® which incorporates a slip sensor into the hand of a myoelectric arm. The hand automatically increases grip when slip is detected (Cranny 2005, Puchhammer 2000). However, this grip adjustment system is independent of user control and the user is not otherwise signaled about a change in sensory input.

Figure 3: (Left) The Otto-Bock SensorHand® SPEED is an EMG controlled prosthetic hand with integrated slip sensors. Grip force is automatically increased when a grasped object begins to slip in the hand. (Right) A traditional body powered prosthesis conveys proprioceptive information through cable tension and harness position.

3.2 Proposed Feedback Systems for Prostheses

Researchers have also attempted to map proprioceptive and tactile information from the external prosthesis to another sensory modality in order to continue using the amputee's internal motor control system. Kawamura suggests the development of a sound schematic to provide feedback in a myoelectric limb (Kawamura 1990). Studies showed the efficacy of the sound system through the improvement of amputee performance during gripping tasks with an upper-limb prosthesis and position and velocity recognition with a lower-extremity prosthesis. Utilizing a more direct method of feedback, Yoshida explored inferential current as a means of providing sensory information to the user (Yoshida 2001). Psychophysics experiments confirmed the use of inferential current to synthesize a low frequency in the body as a useful method of signaling prosthesis users. Another study examined the usefulness of a vibrotactile display used during the amputation rehabilitation period. In an initial short-term study, this display was used to provide feedback to improve weight bearing and gait symmetry (Sabolich 1994). As compared to traditional approaches, the training programming utilizing proprioceptive feedback enhanced patient performance during prosthetic fittings and trainings. In the force-feedback realm, the Rutgers Ankle Orthopedic Rehabilitation Interface provides a platform to improve strengthening, stretching, and balancing by using virtual reality based games to motivate patients to engage in rehabilitative exercises. A flight simulator is controlled by resistive strength building and range of motion exercises to provide parallel feedback to stimulate and enhance proprioception. Initial case studies showed clinical improvements for strength, range of motion, and balance ability (Deutsch 2001). Although this device does not directly replace missing or injured feedback systems, the task of controlling the virtual simulations reinforces proprioceptive feedback.

3.3 Proposed Invasive Feedback Systems for Prostheses

With the advancement of neuroscience and neuroengineering, scientists have begun to investigate more invasive means of sensory feedback. Cineplasty is a surgical procedure that externalizes the force and extension of muscle by physically connecting the muscle to a prosthetic device. This system can be likened to power steering in a car whereby a small movement at the muscle triggers an amplified corresponding movement in the artificial limb (Weir 2001). In a related surgical practice, Kuiken has developed a technique to transfer the residual nerves of amputees to spare muscles in or near the residual limb. Because the transplanted nerves were controlling the same function in the external prosthesis that they controlled in the natural arm, the signaling system was easier and faster to learn (Kuiken 2005). There has been extensive research into the use of neuroprostheses for the direct electrical stimulation of the nerves. Electrodes are integrated into the peripheral nervous system and are used to record efferent signals and simulate afferent signals. Preliminary short-term research has shown this technology to be useful in implementing an artificial closed-loop control system (Loeb 2000, DiLorenzo 2003, Edell 1996, Moxon 2001, Struijk 1999, Yoshida 1996). However, many complications need to be resolved before these devices can be safely and reliably used by human subjects. Implanted objects need to be robust enough to withstand the internal chemical and mechanical processes that occur as a result of normal functioning and also avoid triggering adverse immunological attacks. Furthermore, any invasive methodology requires years of research to confirm and refine the safety of the technology.

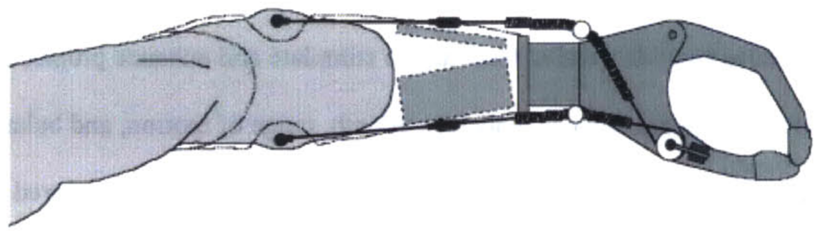

Figure 4: A schematic of cineplasty, a surgical procedure that externalizes the force and extension of muscle by physically connecting the muscle to a prosthetic device.

Figure 5: Jessie Sullivan gripping a glass with his prosthetic arm controlled using EMG signals collected from nerves transplanted from his amputated limb to his chest.

3.4 Vibrotactile feedback systems in prostheses

Most recently, Pylatiuk and Sabolich have suggested the development of a vibrotactile display for use as a sensory feedback system in a myoelectric hand prosthesis and a passive lower limb prosthetic, respectively (Pylatiuk 2004, Sabolich 2002). Pylatiuk initiated vibratory psychophysics studies in amputees and reported a frequency discrimination profile similar to non-amputees. Similarly, Sabolich has proposed the use of vibratory feedback as indicators of pressure or contact on the sole of a passive prosthetic foot. Although existing devices have addressed the need for feedback in prostheses, there has not yet been an attempt to develop and test a comprehensive display and signaling schematic for use with an active myoelectric prosthesis.

4. A Sensory Feedback System for an Active Lower-Limb Prosthesis

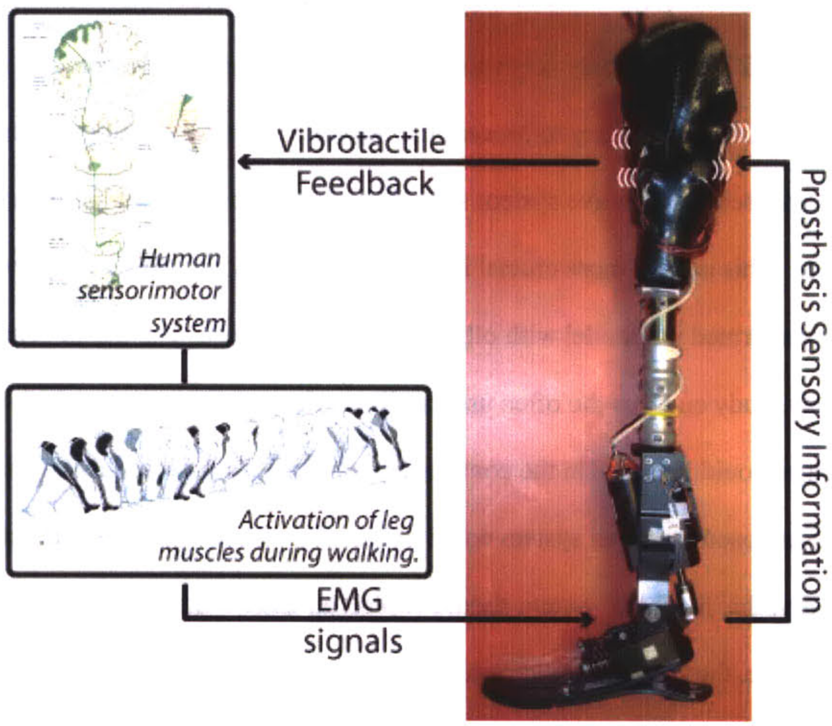

Figure 6: A schematic of the complete active-ankle system including active-ankle prosthesis, EMG control, and vibrotactile feedback. The components of this feedback control loop are necessary to emulate the motor control system of the healthy human.

As discussed in the previous section, past research has focused on various aspects of feedback in prostheses including extended physiological proprioception, user-independent feedback, sound and current feedback, rehabilitation aid, invasive implantation of feedback devices, and haptic feedback in passive devices. Although these studies have not directly investigated the usefulness of vibrotactile feedback in an active myoelectric prosthesis, their findings lay the theoretical groundwork for the development of a multidimensional vibrotactile feedback system for a prosthesis. This document details the development and evaluation of a vibrotactile display to be incorporated into a prosthesis – more specifically a myoelectric powered ankle-foot system. Two different design approaches have been pursued: (1) an array of vibrating pancake motors embedded into the exterior of a carbon fiber prosthetic socket and (2) an array of vibrating pancake motors embedded into a silicone socket liner.

4.2 Design Specifications

There are a number of design specifications an active lower-limb prosthetic sensory feedback system needs to fulfill. The display needs to peripherally and subtly communicate to the user to reduce the cognitive load necessary to process the feedback. Unlike existing implementations of myoelectric prostheses, an effective system should incorporate non-visual indicators to convey limb parameters. This is even more crucial in a lower-extremity prosthesis where walking is almost always performed in parallel with other cognitive activities. The use of sound as feedback in the Kawamura study engages the often used auditory channel (Kawamura 1990). Normal listening activities could interfere in the communication ability of the feedback system, or conversely, the imposed feedback system could potentially distract from normal listening activities. Ideally, this feedback system should exploit an underused sense other than audition or vision. Furthermore, the system should be able to communicate multiple channels of information to correspond to the multiple parameters of control in the ankle. Integration of the control and feedback into a unified module directly connected to the user forms a closed-loop and correspondingly more accurate control system. To encourage adoption of the technology, the device should be integrated into an existing prosthetic structure. By incorporating the system into the prosthetic interface, package design can be optimized for weight and obtrusiveness. Although inferential current as detailed in Yoshida's work does not require invasive surgery, the use of output electrodes can conflict with EMG control and should be avoided in myoelectric prostheses (Yoshida 2001). This feedback system needs to be implemented and actualized in parallel with the active-ankle prosthesis to reciprocally improve and verify the performance of the prosthetic device. Accordingly, the technology should be non-invasive in order to avoid the delays involved with neuroprosthetic research developments and be ready for immediate testing with ankle prototypes.

4.3 Haptic Displays

Haptic displays have been designed for a variety of applications including interfaces for blind users, teleoperation, virtual reality, and HCI. Pylatiuk and Sabolich have demonstrated the potential usefulness of vibrotactile feedback in amputee rehabilitation and passive prostheses (Sabolich 1994, Pylatiuk 2004, Sabolich 2002). By using the sensory channels of the skin and the related kinesthetic senses, information can be subtly transmitted in parallel with and also in replacement of traditional audio/visual information displays. Mechanoreceptors in the skin communicate sensations of vibration, temperature, pain, and pressure while receptors in the muscles convey muscle stretch, muscle tension, and joint flexion. This multidimensional sensing capacity lends itself to creating a multidimensional signaling schematic mapping active ankle impedance, position, and power to force, pressure, and vibratory feedback. Haptic devices have been built for both mobile and stationary applications and the range of compact actuators available (including small motors and electroactive polymers) can be easily integrated into the prosthetic structure. Also, the mechanical nature of haptic feedback precludes electrical interference with EMG control. In contrast to neuromedical engineering, the study of haptics and the supporting field of psychophysics have matured sufficiently to develop immediately useful applications to test with the active-ankle. In conclusion, the properties of haptics correlate well to the design requirements of a powered ankle-foot feedback system.

The field of haptics encompasses both proprioceptive and tactile senses. Force feedback displays stimulate proprioceptive mechanoreceptors such as the Golgi tendon organs and the muscle spindles to impart a large-scale sense of shape and force as they would be perceived by muscles and joints. In contrast, tactile feedback displays convey small-scale distributed sensations that activate the mechanoreceptors in the skin. These physical quantities include vibrations, small-scale shape or pressure distribution, and thermal properties. In parallel to these characteristics, tactile displays can be divided into three subcategories: vibrotactile, shape, and thermal (Howe 2002).

As a system which largely conveys proprioceptive information about the ankle, it would seem appropriate to use a force feedback display to close the feedback loop. However, a true representation of these sensations would require either the stimulation of amputated and thus non-existent mechanoreceptors or the invasive direct stimulation of the nerves. Furthermore, the hardware necessary to build a force feedback display is too bulky to be incorporated into a portable prosthetic structure. There has been evidence that vibration of the muscle belly activates muscle spindles which signal muscle stretch velocity. During standing and walking studies, subjects will change their posture during vibration as if adjusting balancing posture to compensate for stretches in the leg muscle (Ivanenko 2000, Sorenson 2002, Verschuren 2002 & 2003). Using this simple external vibratory simulation to activate the proprioceptive receptors would be the closest one-to-one mapping of prosthesis feedback using non-invasive stimulators. However, the restraints of portability restrict the integration of the large mechanisms needed to produce the 70–90 Hz, 1–2 mm vibrations and again require the stimulation of amputated muscles and joints. Likewise, shape displays require bulky actuators and extensive power unreasonable for a portable display and thermal displays would further disrupt the already difficult to control temperature conditions inside a prosthetic socket.

Any imposed haptic schematic would be an artificial mapping to proprioception. Therefore it becomes a matter of designing a portable and easy to learn system which can be integrated into the prosthetic structure. Vibrotactile displays which operate at skin sensitivity (versus muscle spindle sensitivity) are the most mobile in terms of size and power. Actuators require as little as 1.5 V at 60 mA and measure as small as 12 mm in diameter and 3.5 mm in thickness. As evidenced by sensory impaired patients, alternative sensory substitution mappings can be learned to compensate for the loss of sensory input. The ability of these users to learn and adapt in behavior and performance in every day tasks is further supported by the restructuring of the brain to process these newly learned sensory mappings (Flor 1995, Davis 1998). A large number of these devices are vibrotactile and have been successfully used to replace both hearing and seeing for the sensory impaired and general remote sensing for teleoperators. The efficacy of these systems suggests that a vibrotactile display may be effectively used to replace the sensory feedback of an active-ankle.

5. Sensory substitution systems

A wide range of sensory substitution systems have been designed and implemented to replace the lack or impairment of another sense. As a complex system able to communicate multidimensional parameters, the sense of touch has been used for vision substitution, auditory substitution, and remote tactile sensing. In conjunction with, distal attribution or the ability to perceive an event as occurring at a location other than the physical stimulation site, haptic displays have proved successful in providing sufficient sensory input to replace or improve the performance of previously impaired tasks.

5.2 Tactile Vision Substitution

Braille is a tactile representation of the alphabet based on a matrix of six small raised dots. Using their fingertips to scan the text, visually impaired users can read at an average rate of 125 words/min; approximately half the rate of visual readers. Braille readers achieve the same conceptual analysis and mental imagery from reading as visual readers. Furthermore, studies on cross-modal plasticity have shown that areas associated with visual processing in the brain are reassigned to the sense of touch in the visually impaired (Bavelier 2002). This remapping of the cortex provides physical evidence of the human brain to associate new sensory inputs with external physical stimuli.

The Optacon (Optical to Tactile Converter) is a dynamic matrix of 6 by 24 vibrotactile elements used to convert letter outlines recorded by a camera to vibrotactile outlines communicated to the fingers. Users can read at an average of 28 words/min and up to 90 words/min. A similar 2-D matrix of vibrotactile stimulators connected to a camera was designed to translate visual images into a haptic image. The intensity or amplitude of individual actuators corresponded to the intensity of light pixels recorded by the camera. Experienced users were able to recognize common objects and faces. As a result of distal attribution, subjects were able to perceptually project the haptic image from their abdomen or back into the space in front of the camera. Unfortunately, masking effects, limited spatial resolution, and narrow dynamic range restrict the system's efficacy in communicating cluttered images. Pin displays convert visual images into haptic images by translating pixel intensity into pin height. Actuators such as solenoids and electroactive polymers have been used to drive the pins. However, the power consumption and bulky drive mechanisms of these displays render them unreasonable as portable devices.

A few devices have been designed to make the non-linear 2-D space of GUIs accessible to the visually impaired. Immersion Corporation manufactures a haptic mouse which resembles a normal mouse but includes vibrotactile elements to signal events or obstacles in gaming environments. The Moose is a force feedback manipulandum which translates a GUI interface into a haptic interface by mapping visual elements on the screen into force effects. Objects can be located, identified, and manipulated by feeling the display. For example, window edges are represented by a groove and selecting and dragging icons impart a virtual mass to the user (O'Modhrain1997). A lower resolution portable analog of this device was proposed by Schneider. Activation of integrated electromagnets attracts the mouse toward a ferromagnetic mouse pad thereby increasing the frictional force between the two (Schneider 2004). The modulation of this friction can be used to convey different tactile events.

5.3 Tactile Auditory Substitution

Fingerspelling and American Sign Language make up two forms of gestural languages perceived respectively tactually by the deaf-blind and visually by the deaf. The rate of fingerspelling recognition appears to be limited by the production of signs. However, experienced users can read at 80% accuracy at speeds up to 6 letters/s. American Sign Language conveys concepts at the same rate as speech. Tadoma is another tactile language substitution system whereby the receiver places their hands on the face and neck of the speaker to monitor lip and jaw movements, airflow at lips, and vibration at neck. Tadoma users show the highest rate of speech comprehension in accuracy and speed as a result of the richness, quantity, and variety of tactile information conveyed to the receiver. Furthermore, these tactile characteristics correspond closely to the auditory properties of speech such as volume, rhythm, and speed.

The Tactaid was developed as an aid in lip reading and operates on principles similar to Tadoma. Sound from a hand-held microphone is translated into vibratory patterns at frequencies and amplitudes at which skin is most sensitive. These haptic effects communicate sound characteristics so that the user may understand rhythm, duration, and intensity of environmental noise, speech, and music. Lip and jaw movement observations are replaced by lip reading.

In 1955 Békésy observed that the human ear localizes frequency-processing of incoming sounds at selective regions of the cochlea. Based on these findings, scientists developed the Tacticon which modulates the intensity of 16 electrodes which correspond to sound intensity of 16 different passbands. The device was shown to improve speech clarity, auditory discrimination, and comprehension. The Audiotact is a similar prosthesis with 32 channels of communication. An evaluation of these vocoder devices showed that body location (finger v abdomen), dimensionality (1-D v 2-D), or stimulation mode (electrotactile v vibrotactile) do not have any effects on performance. However, compared to Tadoma users, vocoder users do not perform as well.

5.4 Remote Tactile Sensing

Teleoperation tasks have proliferated in space exploration, hazardous environment operations, minimally invasive surgery, and surgery simulation. Human motor control studies have shown the role of kinesthetic and proprioceptive feedback in regulating and planning motion. Sensory gloves developed for patients with reduced hand sensation and astronaut space suits transmit pressure information from the fingertips and palms to vibrotactile stimulators on other parts of the body such as the abdomen. NASA has shown that the addition of force feedback in teleoperation tasks reduces task time from 92 s to 63 s (Hannaford 1992). A barehanded time of 14 s indicates the possibility of further improving performance with tactile feedback

The essential ingredient in the success of these sensory substitution systems is the formation of a perceptual model of received sensory information (afference) as a function of motor control over sensed environment. The accurate transmission of simulated information that changes appropriate with the subject's hand, head, and overall body movements helps to formulate a sense of distal attribution or telepresence. Users were able to remap visual, auditory, proprioceptive, and cutaneous information to vibratory feedback. Learning these sensory substitution systems may be similar to the learning process of healthy children with normal sensory motor system, adults learning a foreign language, or the deaf learning sign language. Through prolonged use, the information extraction process becomes an automatic, unconscious, and integral part of the sensorimotor neurocontrol loop.

6. Lower-Limb Prosthetic System

To better understand the integration of the vibrotactile feedback system with the larger prosthetic system, the three elements of lower-limb prostheses are detailed in the following section. The three major parts of this system are composed of the socket and socket liners, suspension, and components.

6.2 Socket and Socket Liners

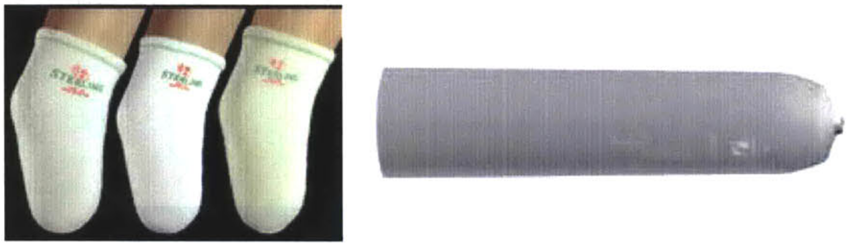

Figure 7: Prosthetic socks and socket liners protect the soft tissue of the residual limb from the rigid surfaces of the prosthetic socket.

The human body is designed to bear weight through the skeletal system, however, with amputees load carrying is transferred to the soft tissue of the residual limb in the socket. The increased stress to the skin in combination with poor blood circulation, lack of ventilation, and abnormal fluid regulation results in an environment susceptible to perpetual soreness and irritation. The fit and comfort of the socket and socket liner is the most important feature in prosthetic systems. Without a comfortable interface between the body and the prosthetic device, the most technologically advanced components would be unusable. The socket provides the means to attach the prosthesis to the body and the socket liners and socks manage the distribution of pressure and friction forces.

Modern sockets are custom formed from flexible plastics, laminates, and carbon fiber to adapt to walking and sitting. To cushion the limb against the relatively rigid surface of the socket, socks and socket liners provide a resilient barrier to protect the skin against pressure and friction (Smith 2004). Prosthetic socks are worn next to the skin and are manufactured in various materials including cotton, wool, and synthetics. These sleeves help to pad the limb, absorb perspiration, and adjust the volume of the socket as the residual limb gains and loses volume. Socket liners made of silicone, urethane, and thermoplastic elastomer gel provide added protection. These liners come in a range of standard sizes with either uniform or distributed wall thicknesses providing additional padding at bony protuberances. They help to provide even pressure distribution as the gel material flows from areas of high pressure to areas of lower pressure (Uellendahl 2001). The comfort of the prosthesis user is dependent on the proper selection and fit of these socket components.

6.3 Suspension

Figure 8: Prosthetic suspension systems. (Left) A suction valve socket affixed to a prosthetic ankle. (Right) A locking liner suspension system – the distal pin locks into a mechanism at the bottom of the socket.

The suspension is the method used to attach the prosthesis to the body. In the traditional suction valve system, as the residual limb is pushed into the socket, air is forced out through a valve at the bottom creating a vacuum to hold the limb in place. A recent innovation using a high vacuum suspension system was developed to adapt to the volume changes of the residual limb. As a consequence of pressure from the socket, fluids are continually pushed out of the residual limb throughout the day causing a fluctuation in the volume of the limb and thereby compromising the proper fit of the static socket. Furthermore, amputation of the lower extremity disturbs the normal pattern of blood and lymph fluid control resulting in additional volume regulation problems. With the high vacuum system, the entire surface of the residual limb is used for weight bearing and volume fluctuations are reduced to 1 percent or less. (Caspers 2003). Additional benefits include the reduction of sweating and optimization of blood flow which enhances overall skin health of the limb. Another commonly used suspension system is the locking liner. A pin integrated into the distal end of a socket liner fits into a locking mechanism at the bottom of the socket. Because the pin concentrates forces at the end of the residual limb, a pistoning-suctioning effect may produce swelling and tenderness. Additional reinforcement from either the socket liner or the sock is necessary to reduce these effects. For users who have problems with either of these more modern suspension systems, an older system of straps and buckles may be more effective in attaching the prosthesis.

6.4 Components

The components of the prosthetic system are the devices which allow the amputee to walk. Depending on the height of the amputation, components include the knee, ankle, foot, and intermediate leg structures. The functionality of these devices ranges from simple static structures to complex computer-controlled adaptive technologies. Details of the powered active-ankle foot prosthesis were reviewed earlier.

7. Physiology of Human Touch

The human skin forms a major interface between the body and the world yet has been overlooked in the field of human computer interaction. To better understand the mechanisms of touch and how they may be used in communicating information an overview of the physiology of skin is detailed in the following section. Touch is mediated by four types of mechanoreceptors corresponding to different sensations including flutter, stroking, pressure, texture, vibration, and skin stretch. The location and shape of the specialized end organs surrounding the nerve terminal determines the sensitivity to different types of mechanical displacement.

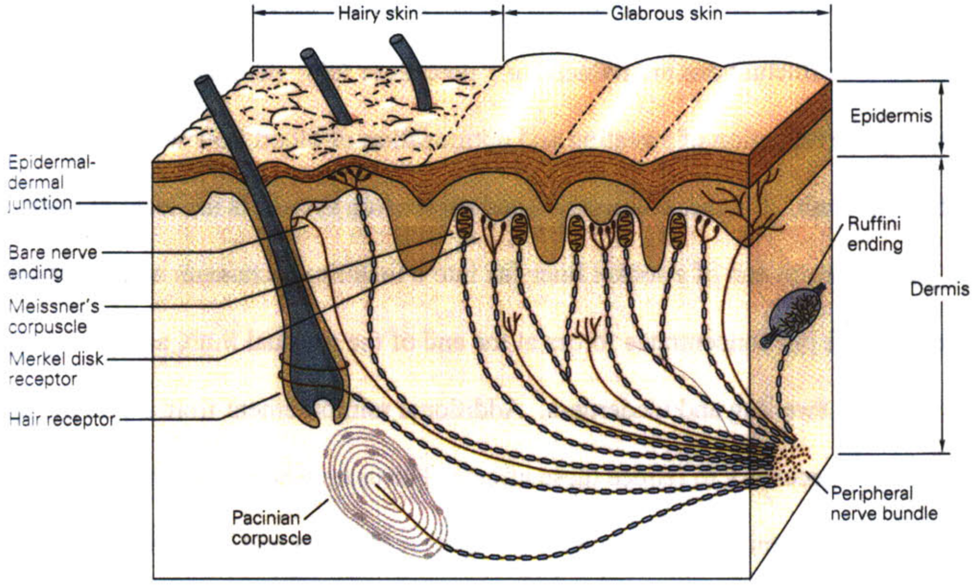

Figure 9: The location and morphology of mechanoreceptors in the hairy and glabrous skin in the human hand (Kandel 2000).

The Meissner's corpuscle and the Merkel disk receptor are located in the superficial layers of the skin. These receptors are associated with the mechanical disturbance of the papillary ridges in glabrous (hairless) skin on the palms, fingers, foot soles, and lips. The small size and high density of these receptors form small distinct receptive fields that confer fine mechanical sensitivity. The Meissner's corpuscle is a rapidly adapting receptor that detects flutter or stroking and the Merkel disk receptor is a slowly adapting receptor that detects pressure or texture. The other two mechanoreceptors, the Pacinian corpuscle and the Ruffini ending, can be found in the deep subcutaneous tissue. Compared to the Meissner's corpuscle and the Merkel disk receptor, these receptors are larger and less numerous. Accordingly, their receptive fields are larger with indistinct borders which sense macroscopic properties of sensed objects. The Ruffini ending is a slowly adapting receptor associated with stretch of the skin. These receptors contribute to our perception of the shape of objects. The Pacinian corpuscle responds to vibrations at the skin surface and nearby joints and muscles. Additional mechanoreceptors are found in the non-glabrous (hairy) areas of the skin that cover most of the body surface. The receptors are located in the hair follicle and respond to displacement of the down, guard, and tylotrich hairs.

8. Psychophysics of Vibration Perception

Psychophysics studies have detailed the human perception of vibrotactile stimuli. The following section presents a summary of Gescheider's overview of vibration perception (Gescheider 1992).

8.2 Temporal Domain

8.2.1 Frequency

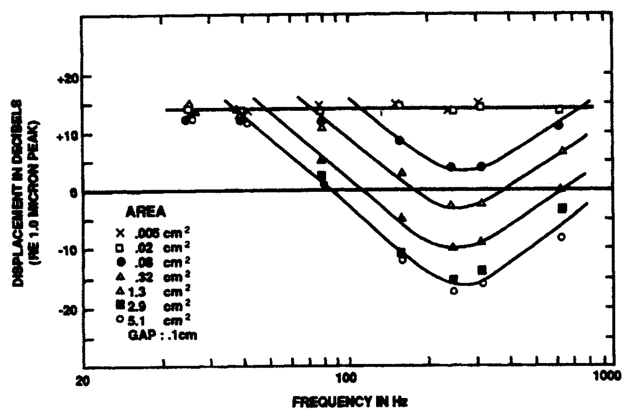

Figure 10: Thresholds for the detection of vibrotactile stimuli measured as function of frequency. Contactor size is the parameter (Verillo 1963).

Psychophysics studies have shown that skin is most sensitive to vibration ranging from 40 Hz to 1000 Hz with a maximum sensitivity at approximately 250 Hz. At this frequency, subjects are able to detect displacements as small as 0.1 μm (Verillo 1992). The resulting U-shaped curve describing these detection thresholds typically indicate activation of two separate receptor mechanisms in biological systems. This observation supports the idea that human touch is mediated by four independent receptor channels corresponding to the four different types of mechanoreceptors located in the skin (Bolanowski 1988). Detection levels for contactors smaller than .005 cm2 show insensitivity to variations in frequency (Verillo 1992). These findings are important to consider during actuator design and selection.

8.2.2 Frequency Discrimination

Human skin shows a marked inability to detect differences between stimuli of different frequencies. Mowbray and Gebhard showed that subjects perform better in discrimination tasks at lower frequencies than higher frequencies using a pulsating rod held between the fingers (Mowbray 1957). Goff reports a difference limen of 30% between frequencies (Goff 1967). The reduced ability of humans to detect discrete levels of frequency limits the usefulness of frequency as a design parameter in haptic mapping schematics. However, some qualities of frequency variation may be used in parallel with other vibration parameters to produce more complex patterns. Studies show that pulse frequency differences are easier to recognize than sine wave frequency levels. Furthermore, subjects report a buzzing sensation for frequencies lower than 100Hz compared to a smoother sensation at higher frequencies. Although limited in scope, frequency discrimination may be somewhat useful.

8.2.3 Stimulus Duration

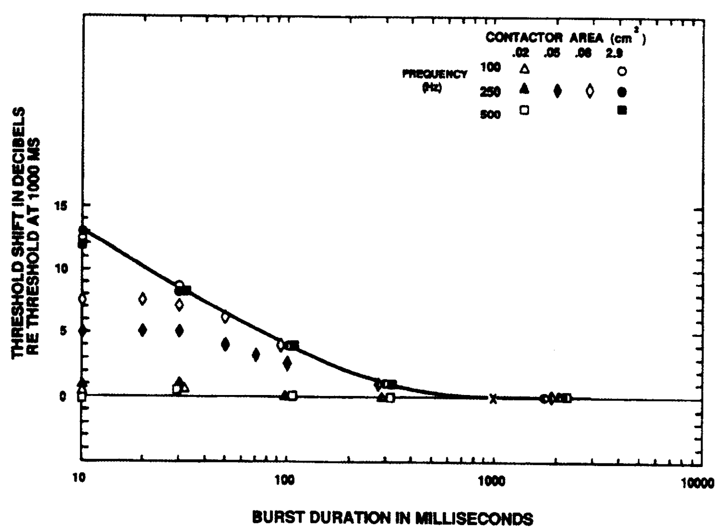

Figure 11: Vibrotactile detection thresholds as a function of stimulus duration. Contactor size is the parameter (Verillo 1965).

There is generally a trade off between duration and intensity of stimulation in perceptual sensitivity. Duration thresholds decrease with decreasing contactor size and sensitivity improves as an orderly function of signal duration with a 3dB per doubling of duration for durations up to 200 ms. This evidence shows that the tactile system is capable of integrating stimuli energy over a limited period of time. This temporal summation is absent when small contactors are used and at low frequencies of vibration.

8.2.4 Gap Detection

Gap detection or the ability to detect a pause in stimuli is a common measure of auditory temporal capabilities. In the tactile realm, Gescheider showed that detection improves as a function of the time interval and the intensity of a pair of separated tactile clicks (Gescheider 1966, 1967). This gap threshold measures approximately 10 ms but has been reported to be as low as 5 ms for highly damped pulses. In a comparison of waveforms, pauses between sinusoids were easier to discern than gaps between noise. Sinusoids are qualitatively described as smooth signals and intervening pauses are interpreted as a click whereas noise conveys a feeling of roughness and the gaps are detected as a modulation of amplitude.

8.2.5 Amplitude Modulation

Weisenberger conducted some experiments exploring the skin's sensitivity to amplitude modulated vibrotactile signals. Amplitude modulated signals are the result of multiplying a sine wave of one frequency with another waveform of a second frequency. The modulated or base waveforms referred to as carrier waveforms included sinusoids of 25 Hz, 50 Hz, 100 Hz, and 250 Hz; wide-band noise; and narrow-band noise. Modulation frequencies ranged from 5-200 Hz. The use of a "smooth" carrier such as a sinusoid appears to enhance perception of amplitude modulation. Users were most sensitive to modulation between 20-40 Hz (Weisenberger 1986).

8.2.6 Temporal Order

The ability to determine the temporal order of multiple stimuli presented to the skin is important to consider in the presentation of complex patterns on distributed areas of the body. Hirsh and Sherrick reported a 20 ms interval between the onset of two brief stimuli. This threshold increases with the number of stimuli and is reported to be approximately 500 ms for 5 or 6 stimuli. However, if the task is to discriminate between two temporal sequences rather than to identity temporal order, increasing the number of stimulus has little effect on discrimination thresholds are below 100 ms.

8.3 Spatial Domain

8.3.1 Contactor Size

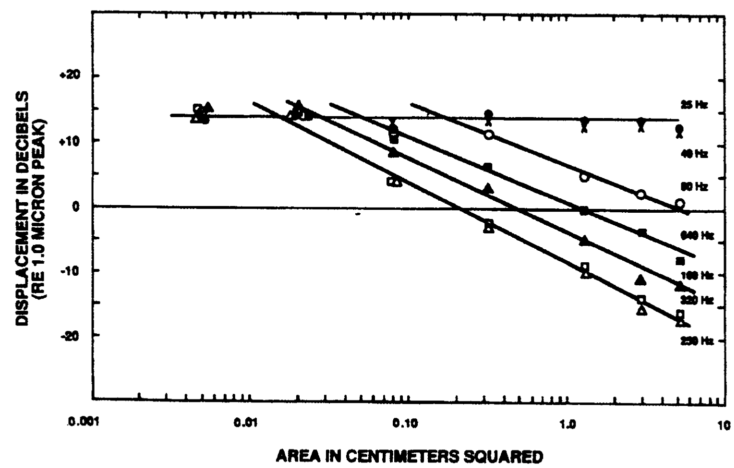

Figure 12: Vibrotactile detection thresholds at different frequencies as a function of contactor size (Verillo 1963).

In the U-shaped section of the frequency sensitivity curve, decreasing contactor size indicates decreasing sensitivity to vibration. As mentioned previously, sensitivity to small contactors operate independently of frequency. At frequencies ranging from 80-320 Hz, sensitivity increases directly with the size of the contactor at a rate of 3 dB per doubling of contact area. As a result of spatial summation which will be discussed in greater detail later, this sensitivity-contact area relation holds true for both increasing contactor size and increasing number of contactors. At frequencies below 40 Hz, detection threshold is independent of contactor size and therefore spatial summation does not occur.

8.3.2 Body Site

Body site of actuator placement is important to consider in haptic display design. The density of mechanoreceptors in the body varies from site to site and the types of receptors differ between glabrous (non-hairy) and non-glabrous skin. The most responsive areas of the body include the hands and face with decreased sensitivity in the rest of the body. One measure of sensitivity is the two-point discrimination threshold. Researches have recorded a distance of 11-18 mm on the back for vibrotactile stimuli. According to data on electrotactile stimuli and static touch, the threshold for the thigh and the calf are slightly higher than the back. Accordingly, a threshold of approximately 10 – 21 mm may be extrapolated to the thigh and calf. However, the validity of this measure as a design parameter has been questioned as the threshold has been shown to decrease with practice and deteriorate with fatigue or distraction. Furthermore, patterns and dynamic or moving stimuli produce greater resolution sensitivity as compared to static individual points.

Biggs showed that the glabrous and non-glabrous skin showed different sensitivities to tangential and normal disturbances. When the actuator is limited in terms of peak displacement (e.g. the strain of a ceramic piezoelectric actuator), then tangential stimulation is superior for both glabrous and non-glabrous skin. When the actuator is limited in terms of peak tangential force (e.g. the stall torque of a DC micromotor), then tangential stimulation is preferred for glabrous skin while normal stimulation is better for non-glabrous skin (Biggs 2002).

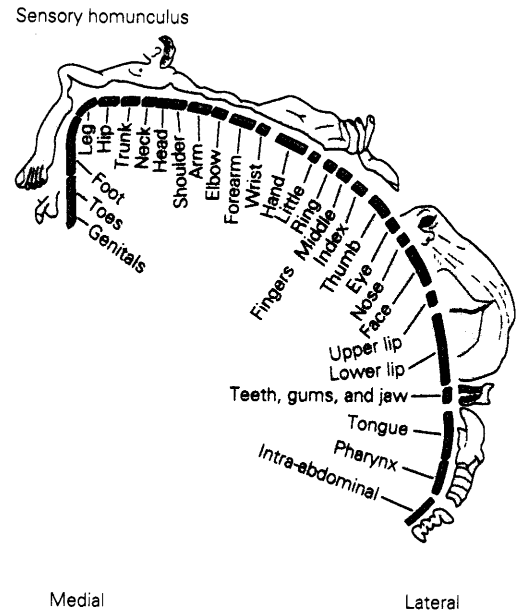

Figure 13: The sensory homunculus illustrates the location and amount of cortical area dedicated to the sensory input from particular areas of the body (Kandel 2000).

8.4 Intensity

8.4.1 Subjective Magnitude

A set of equal sensation magnitude curves have been published for the sense of touch which detail the physical intensity a stimuli must have at any given frequency to feel as loud as a specified one at a different frequency. Lower frequency stimuli must be delivered at a higher intensity to impart the same sensation as higher frequency stimuli at a lower intensity. These measurements are important when it is necessary to balance or emphasize the perception of a range of stimuli at different frequencies (Verillo 1992).

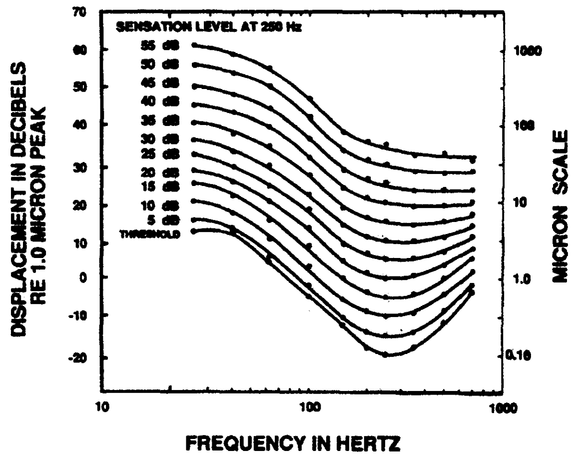

Figure 14: Equal sensation curves plotted as a function of frequency (Verillo 1969).

8.4.2 Intensity Discrimination

The cutaneous mechanoreceptors detect an intensity range of approximately 55 dB above detection threshold beyond which stimulation becomes painful. Studies have measured a detectable difference range from 0.4 dB to 2.3 dB between discernable discrete intensity levels. Discrimination detection is improved when the increment is imposed on a continuous background pedestal of vibration compared to pedestals of brief duration. These differences average about 0.7 dB. Ability to detect intensity steps is insensitive to the type of stimuli and remains constant in response to wide-band noise, narrow-band noise, and sinusoids of varying frequencies.

8.4.3 Adaptation

Adaptation is the reduction of sensitivity induced by prolonged exposure to a stimulus. The recovery time of tactile adaptation ranges from a few seconds to a few minutes depending on the duration and intensity of the conditioning stimulus. Adaptation within one tactile channel does not affect the sensitivity in other channels and is not manifested across Pacinian and non-Pacinian channels.

8.4.4 Effects of Multiple Stimulation

Stimuli presented in close temporal proximity can interfere with intended haptic effects. The different types of perceptual phenomena which may occur are documented below.

Masking	Masking is the reduced ability to detect a signal in the presence of a masking stimulus. This occurs in simultaneous presentation of stimuli or when the masking signal precedes or follows the target signal by a brief time interval. Effects disappear when the two signals excite two different mechanoreceptor systems or when the masker and target signal are situated in opposing frequencies (high v low).
Enhancement	Enhancement increases the perceived intensity of a stimulus in the presence of a brief enhancing stimulus. This phenomenon is present when signals occur within 500 ms of each other and is a decreasing function of interstimulus time intervals. Unlike masking, enhancement is a produced during the activation of a single mechanoreceptor system (Pacinian or non-Pacinian) and is present when both signals are characterized by high or low frequencies.
Summation	Summation is an integration of combined sensation magnitude. The opposite of enhancement, summation involves the stimulation of two systems.
Suppression	Suppression, or masking from remote sites, is the depreciation of ability to detect the presence of a stimulus at one body site in the presence of a stimulus from another part of the body. This happens when the two signals are located on different parts of the body or contralaterally on the same part of the body. This mechanism is triggered when the two stimuli are presented within 75 ms of each other

9. Tactile capabilities of amputees

In working with amputees, additional factors that may affect the efficacy of a haptic display need to be considered. According to statistics, approximately 82% of amputees have lost their limbs as a result of dysvascular problems typically associated with diabetes. In particular, lower-limb amputations accounted for 97% of limb loss discharges (NLLIC Staff 2004). As a result, many lower-limb amputees have decreased levels of vascularity in their residual limb and associated decreased levels of skin sensitivity. Furthermore, the interface between the soft tissue of the residual limb and the rigid surfaces of the prosthetic socket induce sores which would further reduce the sensitivity of the skin. These factors could reduce an amputee's ability to utilize a haptic display. In the work described in this thesis, subjects were chosen from a pool of young, healthy amputees with minimal vascular problems to minimize these adverse effects. Additionally, recent advances in socket technology may help alleviate the skin problems associated with prosthesis use.

Phantom sensation and phantom pain are common side effects experienced by patients after amputation. The amputee experiences the sensation that the amputated limb continues to exist as an extension of the body. Studies have shown that this may be a result of cortical reorganization in the brain and serves as an adaptive compensatory mechanism by restoring activity in a zone deprived of its afferent input. Areas formerly associated with the amputated limb are reassigned to both the face and areas of the residual limb (Flor 1995, Davis 1998). This remapping phenomenon could be exploited by stimulating the residual limb and thereby indirectly activating the parts of the brain formerly associated with the amputated limb.

10. Hardware Design

10.1 Tactile Actuators

There are two categories of vibrotactile actuators that have been used in vibrotactile displays. The first type has been developed specifically for use in vibrotactile displays and operates within the frequency and amplitude range of greatest skin sensitivity. These devices may use voice coil (acoustic-speaker) technology or an inertial transducer. These include the Tactaid and the Engineering Acoustics C2 Tactor. The Tactaid is a tactile aid for the hearing impaired and is typically composed of a microphone attached to two vibrating elements. Sound is translated into vibratory patterns so that the user may understand rhythm, duration, and intensity of environmental noise, speech, and music. Amplitude and frequency are independently controlled and these voice coils have been designed to operate at the amplitude and frequency at which skin is most sensitive. The Tactaid measures approximately 26 mm by 19 mm by 11mm while the C2 Tactor measures 31 mm in diameter and 8 mm in thickness.

Figure 15: (Left) The Tactaid, a vibrotactile aid for the deaf translates speech information into rhythm, duration, and intensity in vibration pattern. (Right) The C2 Tactor from Engineering Acoustics Inc. nominal center frequency at 250 Hz designed for vibrotactile displays.

Small DC motors have been developed for use in pagers and mobile phones to communicate simple messages. A small counterweight mounted on the shaft of the motor produce vibrations detectable by the human skin. Amplitude and frequency are linked and controlled by a single voltage. These motors draw approximately 60 – 120 mA and 1.5 – 3 V and vibrate at frequencies around 130Hz. They are produced in two form factors: cylindrical and pancake. The pancake motors are favored for the maximum transmission of tangential forces against the skin. These measure approximately 12 mm in diameter and 3.5 mm in thickness.

In consideration of size, I have chosen pancake motors as the tactile actuator. The thickness of the specialized tactile actuators prevents the easy integration of the display into existing prosthetic sockets and socket liners. Although the pancake motors operate at a less favorable frequency and amplitude, they function well within the limits of vibration sensitivity. This characteristic will be important in designing the haptic schematic. As mentioned earlier, the tangential stimulation supplied by the pancake motors is superior to normal stimulation on the glabrous skin. The ability of humans to discriminate between relatively few discrete levels of amplitude and frequency offset the correspondingly linked parameters. Furthermore, the haptic schematics have been designed to compensate for this lack of control.

10.2 Socket

As the contact point between the prosthesis and the body, the socket and socket liner were chosen as the site of vibrotactile display integration. To preserve the comfort of the user, it seemed logical to embed the rigid actuators into the rigid socket. With the help of a trained prosthetist, cavities were molded exterior to a custom-fit carbon fiber below-knee prosthetic socket. Nine flat vibrating pancake motors typical to pager motors, cell phones, and other vibrotactile displays were grounded within these cavities to maximize the transmission of vibration to the socket. These types of actuators have been repeatedly used for other vibrotactile displays and have been shown to fall within the range of human skin sensitivity. These specific devices are rated for a maximum of 4 V at 70 mA and measure exactly 12 mm in diameter and 3.5 mm in thickness. The placement of the contactors was determined by the optimization of the following parameters: contact between the flat motors and flatter parts of the residual limb, interval spacing above extrapolated 2-point limen threshold, and maximum number of allowable actuators. The motors are distributed on both bony and fleshy parts around the length and diameter of the residual limb at intervals of 40 – 50 mm, well above the extrapolated 10 – 21 mm 2-point limen threshold for vibratory stimuli on the calf and thigh.

A series of exploratory pilot studies was conducted using the carbon fiber socket to gain some insight into haptic mapping design and feedback on the actuator placement. Tests and fittings with a single subject helped to refine the hardware design. The experiments explored the ability of the subject to discriminate between different levels of frequency, amplitude, duration, and body location. A GUI presented varying levels of each parameter to the user. The same set of experiments was used to evaluate each design iteration.

Initially, the motors were held in place by vacuum formed caps screwed into the socket. The intent was to loosely hold the actuators in place so that the amplitudes of the vibrations would not be damped out by the rigid socket and the greater residual limb mass. However, instead of transmitting vibrations to the limb, the motors moved around in the too-large cavities and transmitted vibrations to the caps. As a result, the vibrations were barely perceptible by the amputee and it was impossible to distinguish any discrete stimuli levels. Next, a small piece of foam was inserted between the cap and the actuator in an attempt to ensure that the motor stayed in contact with the wall of the socket. Again, this was unsuccessful and the haptic signals were difficult to detect and discriminate. An alternative idea suggested the complete elimination of the wall of the socket where the actuators were placed so that the contactors were in direct contact with the socket liner. Unfortunately, the resulting Swiss cheese-like structure would have compromised the structural integrity of the socket and eliminated the use of suction valve suspension systems. To preserve structural integrity, a smaller hole could be used to transmit vibrations through the socket from a small pin attached to the vibrator. However, this solution fails to resolve the loss of the vacuum suspension system. Finally, the motors were grounded to the inside of the cavity using an epoxy. Transmission improved enough so that different stimulation sites were easily distinguishable. In comparison, the subject performed poorly in frequency, amplitude, and duration discrimination tasks. The poor detectability of these parameters can be attributed to the limited ability of skin to distinguish between different levels and the difficulty in labeling abstract tones. Furthermore, the damping properties of the stiff carbon fiber and the comparatively large mass of the residual limb reduced the transmission of the vibrotactile signals. Despite slightly improved transfer of vibration with the grounding of the motor, the reduced signal intensity invalidated the integration of the vibrotactile display into the carbon fiber socket as a viable design.

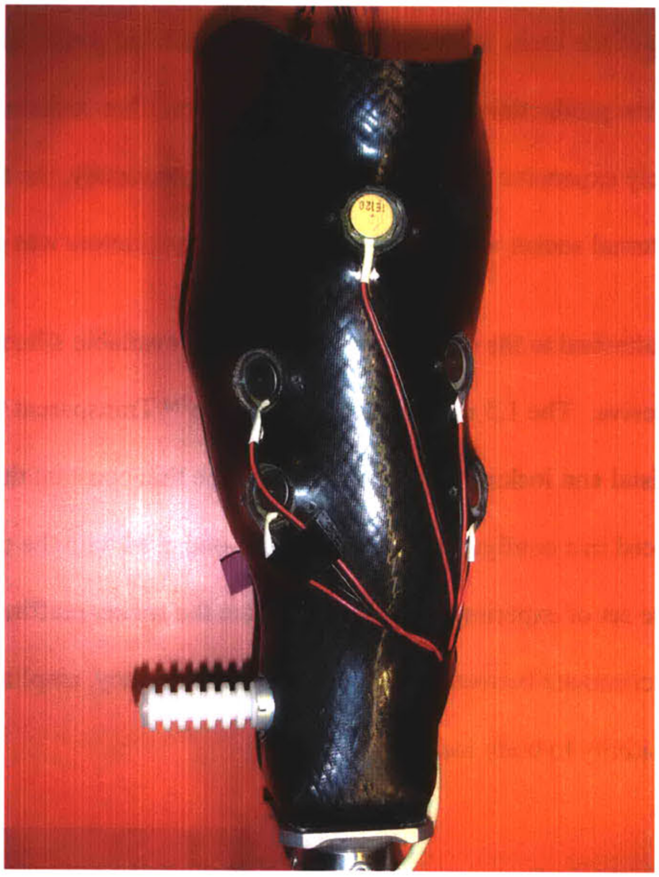

Figure 16: The socket embedded sensory feedback vibrotactile display.

10.3 Socket Liner

Attempts to incorporate the vibrotactile display into the socket showed that the vibrators needed to be in closer contact with the residual limb and that an intervening rigid wall damped out most signals to the point of imperceptibility. As the outermost resilient layer in the socket system, the socket liner was chosen as the next site of integration in the succession of design iterations. To maintain the protective layers provided by both the sock and the socket liner, attempts were made to hollow out cavities on the exterior of a thicker socket liner to preserve the subject's original socket wall thickness between the actuator and the limb. The elasticity and resilience prevented the easy removal of uniform hollows which would have compromised the consistence of signal transmission from contactor to contactor and across experimental liners. Next, the motors were affixed to the exterior of the socket liner and cast over with a layer of silicone with the intent to cushion the actuators on both sides to reduce contact with the damping socket and to maintain the protective cushioning of the limb. However, the use of this thicker socket liner with a socket would have required the production of multiple custom carbon fiber sockets. This process would have been prohibitively expensive and time consuming. Consequently, the socket liners were tested without the external socket and the need for silicone-cast motors was eliminated.

Motors were simply attached to the exterior of commercially available silicone socket liners using a silicone rubber adhesive. The 1.5 mm thick Alps Clearpro™ Transparent Silicone Suction Socket without the distal end locking pin was chosen for the thin constant thickness liner walls. The actuators are placed in a configuration similar to the one used with the carbon fiber prosthetic socket. Tests with the set of experiments used to evaluate the socket reaffirmed the subject's reduced ability to discriminate between different levels of frequency, amplitude, and duration and relatively higher sensitivity to body location.

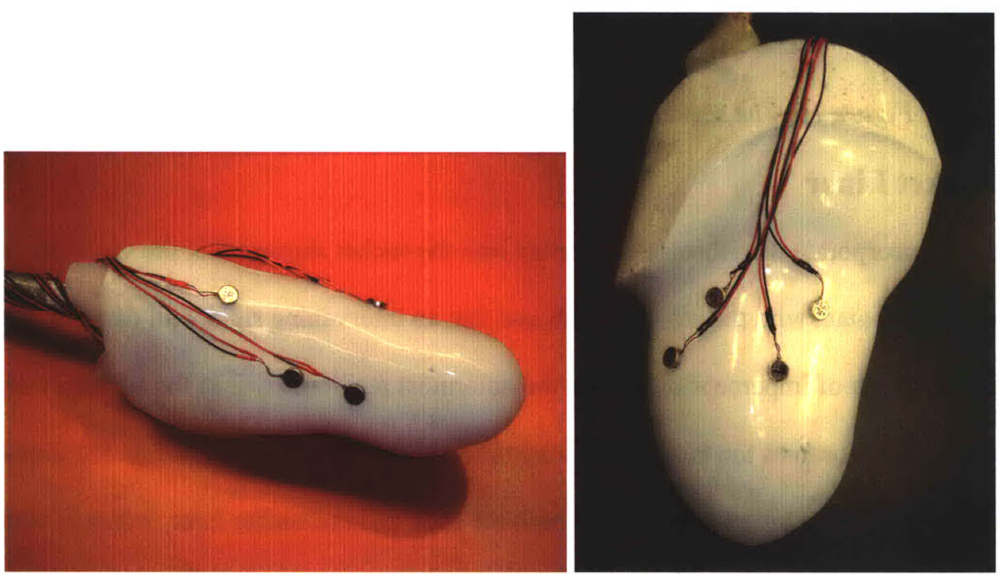

Figure 17: The socket liner embedded sensory feedback vibrotactile display. (Left) The front of the below-knee socket liner. (Right) The back of the above-knee socket liner. Both liners have similar actuator placements.

10.4 Data Acquisition Card: Hardware-Software Interface

A PCI-DAS1602/16 Multifunction Analog & Digital I/O Board from Measurement Computing was used to interface the vibrotactile actuators and potentiometer controlling the virtual ankle to the MATLAB Simulink control software. Nine of the sixteen digital I/O ports were used to output to the ankle. Exploiting the inertia of the motor, a PWM signal was used to simulate variable voltages. An analog input was used to connect the potentiometer to the control software.

11. Haptic Schematic Design

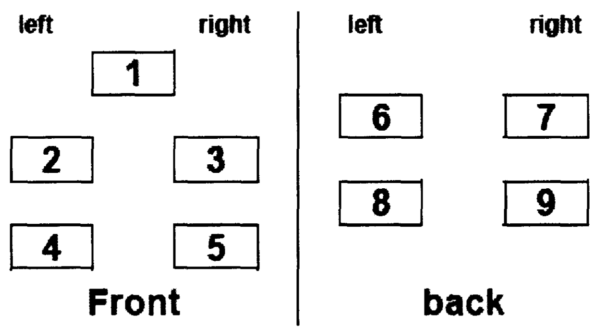

Figure 18: Placement of the actuators on the socket and socket liner.

A haptic mapping framework was formulated based on the constraints of the actuators and the perception of vibration as detailed in the psychophysics review. Most psychophysics studies focus on threshold values of various parameters. However, to be most effective across a range of skin sensitivities, suprathresholds levels should be used in haptic schematic design.

As a visually identifiable and familiar characteristic which can be quantified into angular units, ankle position was chosen as the initial mapping parameter. Limited by the nine actuators and the spatial pattern (which will be described in a later section) the ankle movement space ranged from 0° to 90° with each signal representing 10°.

11.2 Tactons

Within the field of HCI, Brewster has taken previous psychophysics research and proposed guidelines for the design of tactons or structured abstract messages that can be used to communicate messages non-visually (Brewster 2004). Tacton properties are formulated and classified as frequency, amplitude, waveform, duration, rhythm, body location, and spatiotemporal patterns. In an initial investigation into the effectiveness of tactons, Brewster proposed the combination of these basic parameters into more complex patterns in order to encode more information. In addition to finding parameters people can distinguish, it is important to find parameters that people are able to label and remember. In parallel audio work with earcons Brewster showed that using musical timbres was more effective than using simple tones such as sine waves and square waves. This was attributed to the difficulty of labeling abstract tones which aren't present in everyday auditory activities.

Amplitude modulation is the result of multiplying a sine wave of one frequency with another waveform of a second frequency. As reported by experimental subjects, different modulations produced different feelings of "roughness". Using two vibrotactile devices resonant at 250 Hz, the TACTAID and the C2 Tactor, Brewster tested the ability of subjects to differentiate a 250 Hz sinusoid modified by no modulation, 20Hz, 30 Hz, 40Hz, and 50 Hz modulation. With the TACTAID, no modulation, 40 Hz, and 50 Hz can be differentiated in terms of roughness. Using the C2 Tactor, either no modulation, 20 Hz, 40 Hz, and 50 Hz or no modulation, 30 Hz, 40 Hz, and 50 Hz formed discrete distinguishable tactile levels. 20 Hz and 30 Hz modulation were indistinguishable from each other.

11.3 Haptic Mapping

11.3.1 Spatial Pattern

Observations from the pilot study and previous work suggest that spatial discrimination may be successfully used as a haptic signal. In accordance with Brewster's definition of an effective haptic symbol, body loci are easily detectable and can be labeled using familiar spatial tags such as left, right, top, and bottom and body parts such as shin, calf, and knee. Accordingly, the following spatial pattern was designed. Each of the 9 actuators on the socket or socket liner was mapped to a specific angle for a total of 9 different angle levels. Starting at the top front of the socket, consecutive angle levels were mapped to the actuators in the order front to back, top to bottom, left to right. The actuators were run at the data acquisition card maximum voltage of 2.5 V for the highest possible amplitude and frequency.

11.3.2 Amplitude Modulation

The linked parameters of frequency and amplitude in the motors prohibit a true amplitude modulation where a sinusoid modulates a fixed frequency and amplitude base waveform. Regardless, a pseudo amplitude modulation imposed on these pager motors was hypothesized to convey similar sensations of roughness. Three modulations were chosen based on the Brewster's tacton work: 20 Hz sinusoid, 50 Hz sinusoid, and no modulation. The actuators were again run at a maximum voltage of 2.5 V. To match the 9 different angle levels, the front of the socket was divided into 3 levels. Level 1: actuator 5, Level 2: actuator 3, Level 3: actuator 1. Angles 10°, 20°, and 30° are represented respectively on Level 1 actuators as 20 Hz sinusoid, 50 Hz sinusoid, and no modulation. Angles 40°, 50°, and 60° are represented on Level 2 using the same modulations and the pattern is repeated for angles 70°, 80°, and 90° on Level 3 actuators.

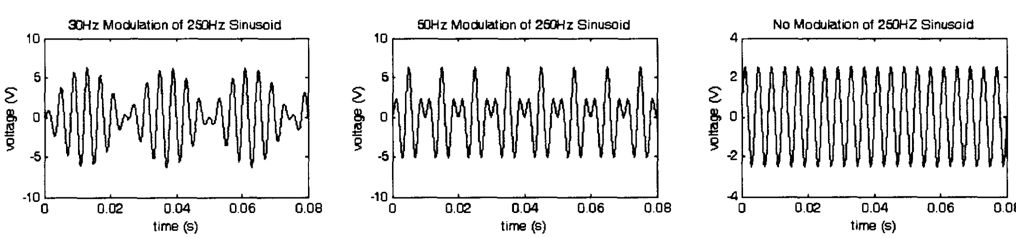

Figure 19: Amplitude modulation vibration patterns.

11.3.3 Gap Detection

Subjective observations of differing sensations of roughness and smoothness from Gescheider's work on gap detection are similar to those reported in Weisenberger's study on amplitude modulation (Weisenberger 1986). This suggests that gap detection may be a useful parameter in designing spatiotemporal vibrotactile patterns. The gap threshold measures approximately 10 ms and any pattern using interspersed pauses should consider this limit. Similar to the amplitude modulation patterns, three gap detection patterns were formulated based on a pilot study. Pauses of 0.1 s, 0.3 s, and no gap interspersed between 2.5 V actuator activation of respectively 0.4 s, 0.2 s, and 0.5 s. These signals are based on a time period of 0.5 s or the 500 ms duration threshold required to determine the temporal order of 5 or 6 stimuli. The front of the socket is divided into 3 levels as in the amplitude modulation pattern and the representation of angles 10° to 90° is repeated for increasing levels of actuator height and gap detection pattern.

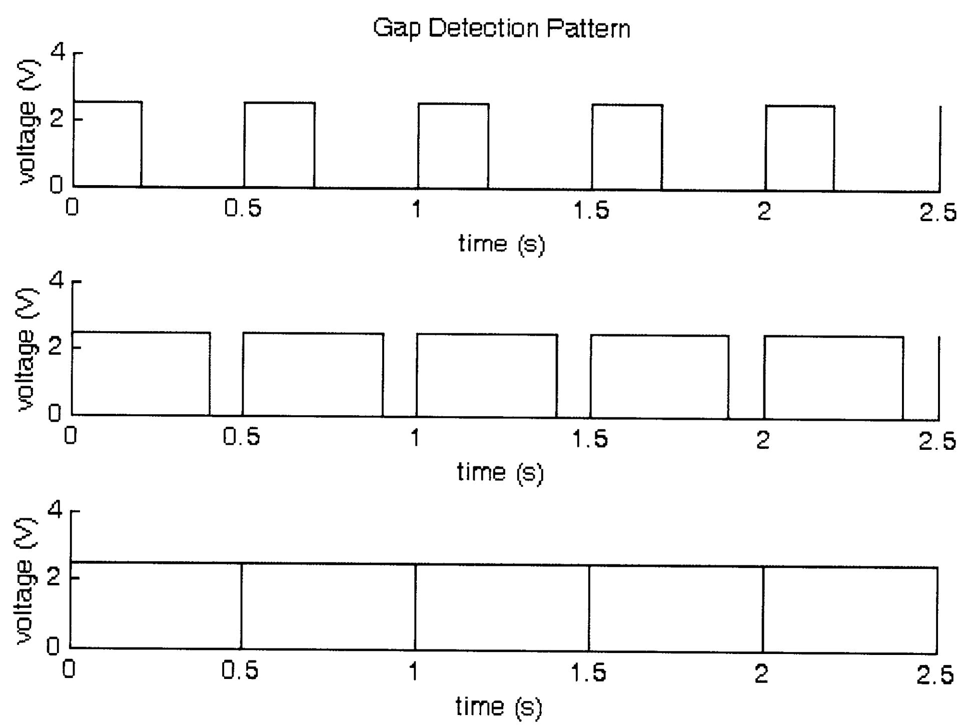

Figure 20: Gap detection vibration patterns

12. Experimental Setup

12.1 Subjects

Three amputees participated in the experiment – one below-knee double amputee (subject 1), one below-knee right limb amputee (subject 2), and one above-knee right limb amputee (subject 3). Subject 1 had been a subject in the pilot studies and identified as an experienced subject. Subject 2 and 3 had never had any previous experience with the feedback system and are identified as novice subjects. All three subjects were young, active, healthy males whose amputations were all a result of trauma. With the exception of the distal end and posterior areas of scar tissue on the above-knee residual limb, the three tested residual limbs showed relatively good skin sensitivity and vascularization.

12.2 Experimental Procedure

The experiments were divided into three stages: learning-exploration, learning-reinforcement, and evaluation. Subjects were asked to don an appropriately sized prosthetic socket liner embedded with 9 small vibrator motors in a seated position in front of a desktop computer. During the positioning of the socket liner on the above-knee amputee, efforts were made to ensure that the contactors avoided the partially insensate areas on the distal end and the posterior areas of scar tissue. Each of the three stages was repeated for each of the three haptic mappings. To minimize the effects of fatigue, each set of experiments was separated by a short 5-10 minute break. Each of the haptic mappings was presented in a different order to each of the three subjects.

12.2.2 <u>Learning-Exploration</u>

In the first stage, subjects were presented with a grid divided into angles 0° – 90° at 10° intervals. As the 'Next' button is clicked, the GUI scrolls through the successive vibrations in each vibrotactile mapping pattern. The angle each vibration pattern represents is indicated by the blue line on the graph and the large number in the upper right corner of the screen. The subjects were asked to experiment and explore until they feel that they had learned which vibration pattern matched which ankle angle.

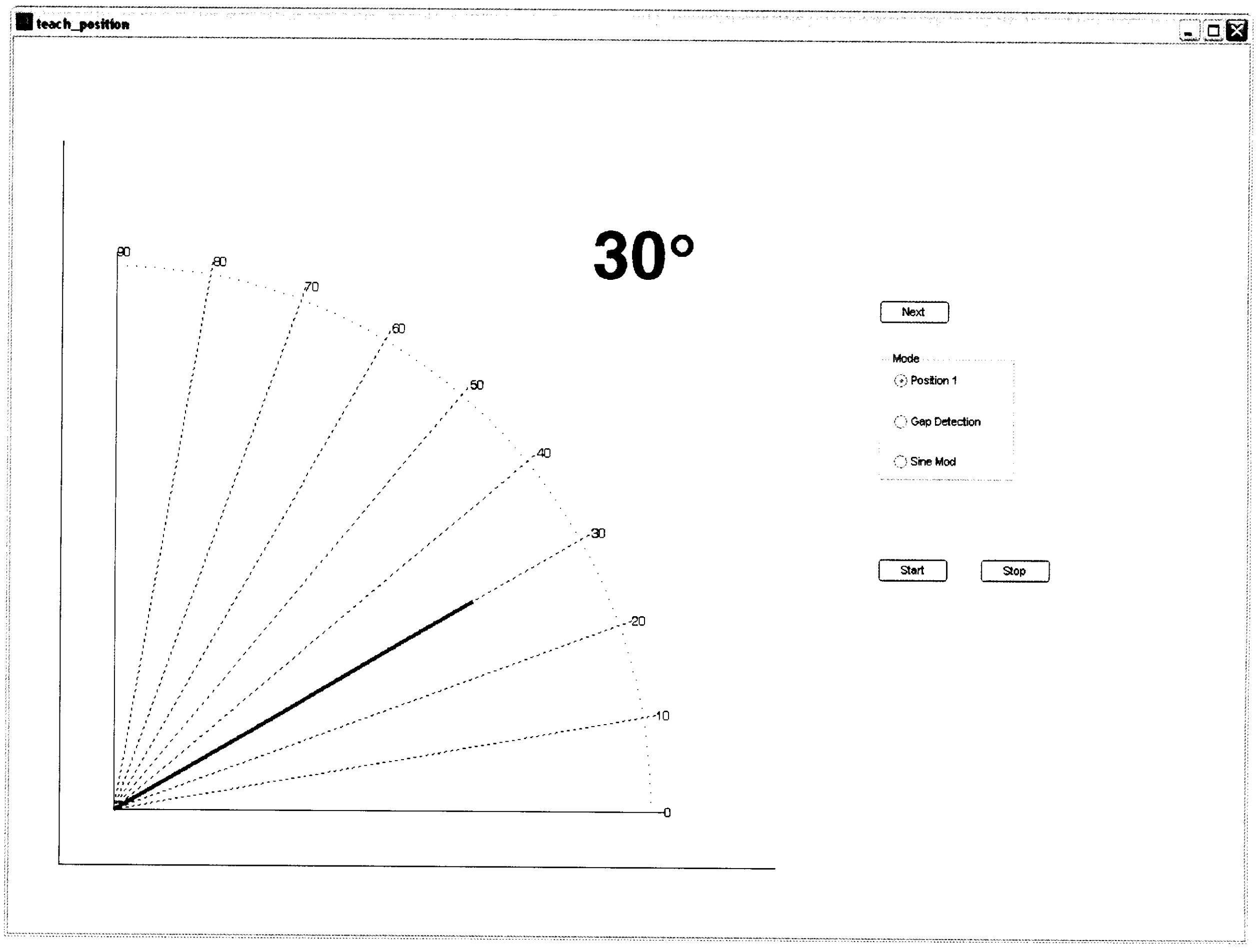

Figure 21: GUI used in the learning-exploration phase. By clicking on the 'Next' button, the system sequentially scrolls through the angle to vibration mappings of each haptic schematic.

12.2.3 Learning-Reinforcement

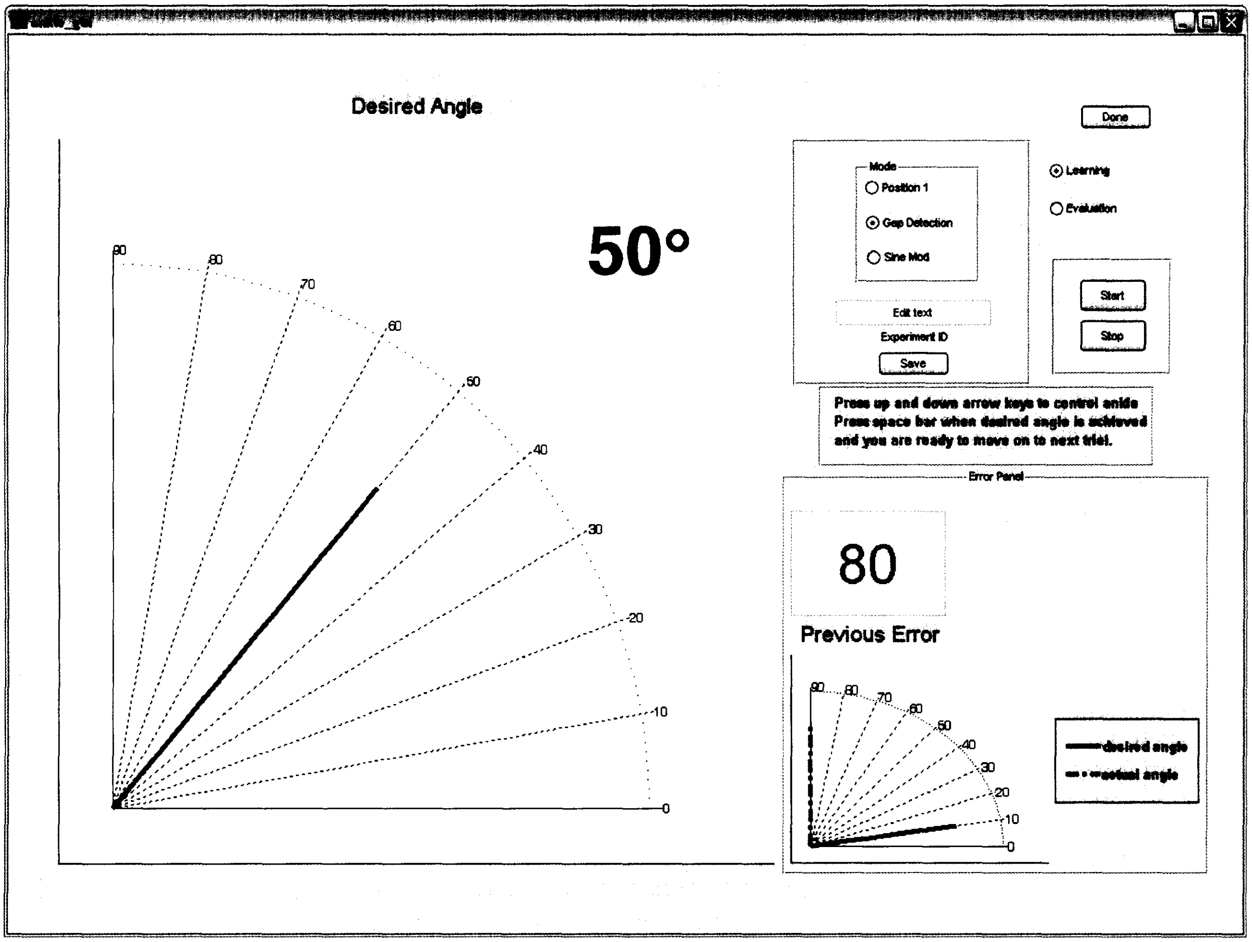

Figure 22: GUI used in vibrotactile feedback display evaluations.

During the learning-reinforcement phase, the subject's understanding of the angle to vibration mapping was reinforced. A small handheld knob controlled a virtual ankle. As the knob is turned, the position of the virtual ankle moves and the vibrations in the socket liner indicated what position the virtual ankle was in. The subject was directed to control the ankle to the desired angle as represented by a blue line on the graph and a number in the upper right corner of a graph similar to the one used in the learning-exploration phase. The 'Done' button was clicked when the user believed the desired angle was reached. To ensure that the subject wasn't simply counting vibration levels up to the desired angle from 0°, the virtual ankle started at a different random position at the start of each new trial. The performance of the previous trial was shown in an error panel in the bottom right corner of the GUI. The desired angle was represented by the solid blue line and the angle the subject controlled the ankle to was represented by the dotted red line. The difference between the desired and actual angle was shown above the previous error graph. The experiment concluded when the user had completed 100 trials.

12.2.4 Evaluation

The evaluation phase was similar to the previous phase except the error of the previous trial was not shown. Again the task was to control the ankle to the desired angle using feedback from the socket liner embedded vibrotactile display. The experiment concluded when the user had completed 100 trials.

13. Results

As mentioned previously, the haptic mappings were presented in a different order for each subject. The haptic mapping order of each subject proceeded as follows:

- Subject 1: Spatial Pattern (or Position), Amplitude Modulation, Gap Detection
- Subject 2: Amplitude Modulation, Gap Detection, Spatial Pattern
- Subject 3: Gap Detection, Spatial Pattern, Amplitude Modulation

During the experiment, collected data included the response time of each trial, the subject controlled angle, the desired angle, and the error. Time response outliers were calculated to account for disruptions during the experiment including technical adjustments and experiment clarification. Outliers were defined as three standard deviations away from the mean. Time response outliers and the corresponding trial data were discarded. The data from the evaluation phase of the amplitude modulation session of subject 3 was removed due to technical malfunction during the experimental session. Figure 22 shows the error level across trials of each of the different sessions of each subject. Each row represents a subject and each column represents the reinforcement-learning and evaluation phase of each haptic mapping. The angle mappings 0°-90° were normalized to angle levels 0-9. Except for the amplitude modulation session of subject 2, the error rates remain close to constant at a mean of 0 with little variation across trials, from learning-reinforcement phase to evaluation phase, and across haptic mappings.

Figure 23 depicts the length of time response across trials for each of the different sessions of each subject. The red line represents the least squares fit for the data to approximate the general trend in time response across trials. With the exception of the learning amplitude modulation session of subject 3, each session shows a slight decrease in response time across trials. The average response time for each session ranged from 7.0 ± 3.3 s to 16.2 ± 8.4 s.

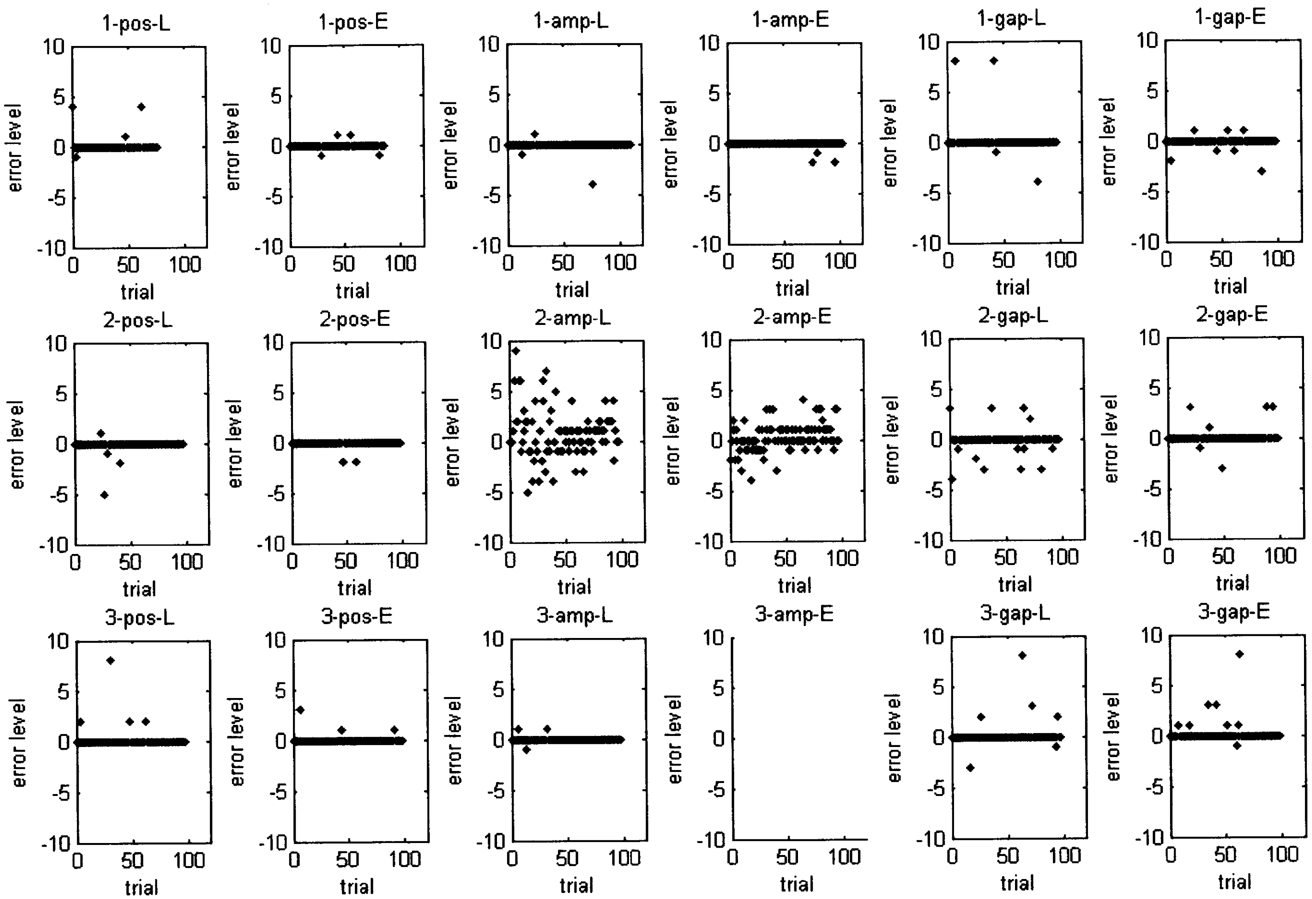

Figure 23: Error over trials. The angles 0°-90° have been normalized to the error levels 0-9. Each row represents the six sessions of one subject. Each of the six sessions consists of the learning-reinforcement stage of each the three haptic mappings. The evaluation amplitude modulation session of subject three has been removed due to technical difficulties during the experimental session.

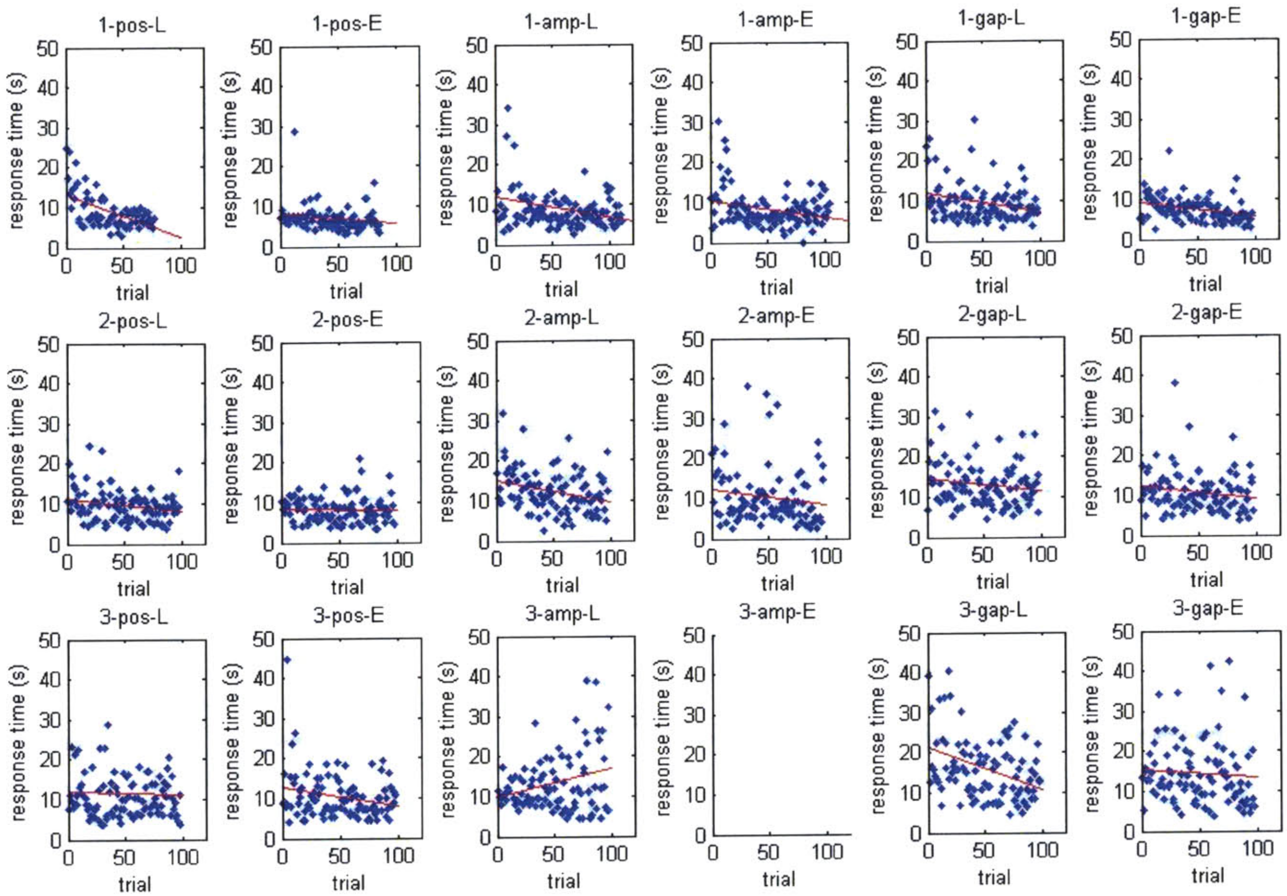

Figure 24: Response time over trials. The red line represents the least squares line to approximate the general change in time response across trials. Each row represents the six sessions of one subject. Each of the six sessions consists of the learning-reinforcement stage of each the three haptic mappings. The evaluation amplitude modulation session of subject three has been removed due to technical difficulties during the experimental session.

The aggregated time response and error data of all three subjects were then analyzed using analyses of variance (ANOVAs) to determine differences in performance across sessions. The percentage of correct responses for each session is shown in Figure 25. The spatial pattern showed the highest recognition rates for both the learning-reinforcement phase and the evaluation phase with correct response percentages of 95.5% and 96.5%. In contrast, the amplitude detection pattern showed the poorest recognition rates for the learning-reinforcement phase and the evaluation phase with correct response percentages of 61.8% and 69.3%. There were significant differences between the three haptic schematics within both the learning-reinforcement phase ($p = 0$) and the evaluation phase ($p = 0$) with the amplitude modulation differing significantly from the spatial pattern and the gap detection pattern. There were no significant differences between the two phases of the spatial pattern ($p = 0.611$). The percentage of correct responses in the amplitude modulation pattern showed a significant decrease between the learning reinforcement phase and the evaluation phase ($p = 0.1168$). The percentage of correct responses in the gap detection pattern showed a significant increase between the learning reinforcement phase and the evaluation phase ($p = 0.4491$).

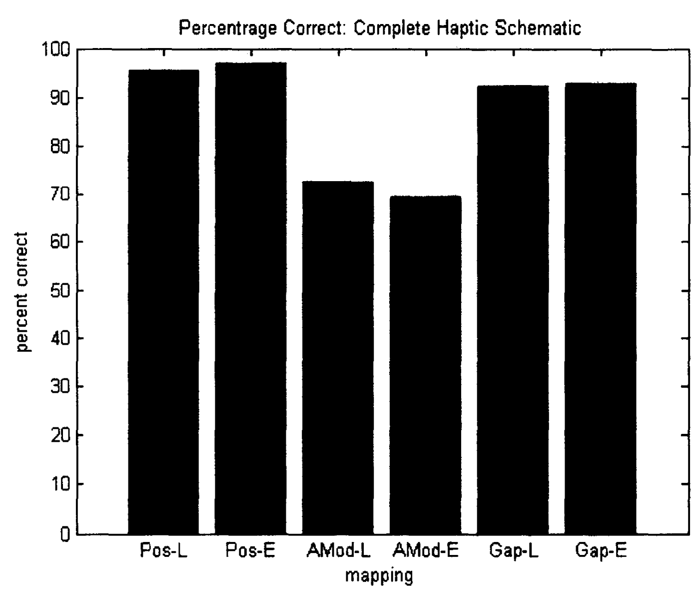

Figure 25: The percentage of correct responses of each haptic schematic across the learning-phase and the evaluation phase for all 3 subjects.

The average response time for each session is shown in Figure 24. Overall, the average response time for the spatial recognition pattern showed the lowest value for both the learning-reinforcement phase and the evaluation phase with response times of respectively 9.5 s and 8.2 s. The gap detection pattern showed the highest average response time for the learning-reinforcement phase and the evaluation phase with response times of respectively 11.6 s and 9.2 s. There were significant differences between the three haptic schematics within the learning-reinforcement phase with the response time of the spatial pattern differing significantly from both the amplitude modulation and the gap detection (p = 0.0008). However, the three mappings showed no significant differences during the evaluation phase (p = 0.0656). The response time improved from the learning-reinforcement phase to the evaluation phase within each haptic mapping. However, only the spatial pattern (p = .0022) and the gap detection (p = 0) showed a significant difference.

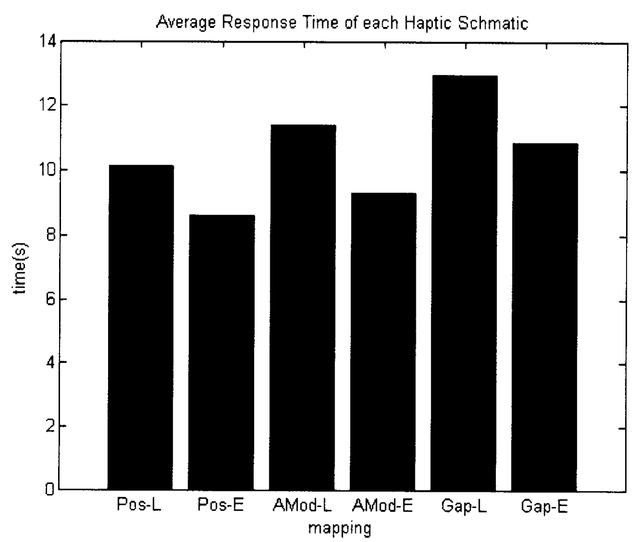

Figure 26: The average response time of each haptic schematic across the learning-phase and the evaluation phase for all 3 subjects.

During testing, all three subjects began to show signs of fatigue and boredom from the length of each session and the number of sessions. This loss of interest and reduced attention may have affected performance over time. One subject reported an initial difficulty in spatially localizing and remembering the position of the actuators. However, the spatial pattern sessions helped to reinforce his perception of vibration location. In terms of adaptation, only one subject reported an increasing insensitivity with time which was not as apparent with the spatial pattern. The same subject also experienced an increase in temperature within the socket liner probably as a result of friction between the liner and residual limb caused by vibration. Again, the effects were less apparent with the distributed vibration in the spatial pattern as compared with the concentrated three actuator vibration in the amplitude modulated and gap detection patterns.

14. Discussion

Results for the correct identification between the haptic signal to angle mapping are very good with an overall recognition rate of 85%. These results compare extremely well with Brown's tacton work showing an overall 71% recognition rate (Brown 2005). With the exception of the amplitude modulation mapping of subject 2, the lack of change in error over time as shown by the individual session mappings and the overall comparison of the learning-reinforcement phase and the evaluation phase for each haptic schematic indicates that subjects are able to immediately learn the haptic mappings. The amplitude modulation mapping was presented as the first schematic to subject 2 and the greater variance in error across trials may be a result of an initial system familiarization period. The significantly lower overall recognition percentage of the amplitude modulation pattern may be a result of the learning curve of subject 2. Additional tests need to be run to determine if amplitude modulation is more difficult to learn, perceive, and remember. Regardless, these results suggest that all three mapping are easily learned and detected and may be successfully used as sensory substitution vibration signals.

In terms of time response, the data indicated a slight improvement across time as evidenced by the trend lines in the individual session mappings and in the decrease in response time from learning-reinforcement phase to evaluation phase for the spatial pattern and gap detection session. Furthermore, time differences between the haptic schematics during the learning-reinforcement phase disappeared in the evaluation phase which implies that with learning and practice, each mapping produces similar time responses. The lower response times and corresponding decreased variance in response times of subject 1 (experienced subject) provides further evidence that response time decreases with continuing use of the feedback system. The time response averages ranging from 8.2 s to 11.6 s is orders of magnitude higher than the transmission of sensory information in the human neural system. Integration of the feedback system with the active-ankle and the EMG control system may further reinforce the feedback system and with time and practice response times may decrease to the point where they can provide useful and timely information as part of the overall prosthetic system.

15. Conclusions and Future Work

The high recognition rates and lack of variance between the mappings suggest that the three vibration parameters of spatial discrimination, amplitude modulation, and gap detection may be successfully used to represent different ankle parameters such as force and stiffness. However, the overall successful integration of the vibrotactile display ultimately depends on the interaction between the components of the whole prosthetic system.

There are still many issues to consider in integrating this vibrotactile sensory feedback display into an active-ankle prosthetic system. The purpose of this thesis was to construct a viable sensory feedback system which included a device design and a corresponding haptic mapping in the first stage of the parallel development of an active-ankle mechanism and EMG control processing. As the various components are integrated, the following design considerations should be addressed.

15.2 Integration with EMG

As stated in the introduction, the purpose of the sensory feedback system is to improve the granularity and effectiveness of EMG control. As the EMG mappings are formulated based on healthy human readings, the sensory feedback system may be integrated with the system and evaluated for improvement in accuracy, response time, and learning time. With regard to device placement on the body, the vibration signals in the socket may interfere with the ability of the electrodes to collect signals on the residual limb.

Design iterations may be necessary to ensure that these two systems can operate without compromising performance. Alternative body sites include integration into an extended portion of the socket liner which lies above the socket but still encases a higher part of the residual limb. This solution preserves the original intent to incorporate the vibrotactile display into the existing prosthetic structure to avoid additional donning and doffing of prosthetic components.

15.3 Effects of Weight Bearing and Walking

The experiments in this thesis were performed while the subjects were sitting without load bearing at the socket interface or the compression of the rigid socket onto the socket liner. Tests need to be run to determine whether the comfort of the amputee is compromised by the integration of the actuators into the socket liner during standing and walking. Directly casting the actuators into the liner may be necessary to provide a uniform cushioning around and in between the motors, limb, and socket. Besides the range of mass produced socket liners, prosthetics companies also offer custom silicone services which may be useful in creating a socket compatible liner. The effects of weight bearing may alter the perception of the vibratory signals. As mentioned in the previous section, the display could also be integrated into an extended portion of the socket liner beyond the socket to compensate for any adverse effects of weight bearing, standing, and walking.

15.4 Stimulus Timing and Duration

Prolonged exposure to a single stimulus may lead to sensitivity adaptation during moments when the prosthesis is inactive. Even if adaptation doesn't occur, the user may become annoyed with the constant stimulation. Options such as shut-off mode controlled by the user or activated when the prosthesis is static for a specified period of time should be considered in the design.

15.5 Multidimensional Parameter Mapping

Once EMG and the effects of weight-bearing have been resolved, the number of useful ankle parameters need to be determined. Three types of receptors in the muscles and joints mediate the sense of proprioception or the sense of body posture and movement. Muscle spindles embedded within the belly of the muscle detect magnitude and velocity of stretch. Golgi tendon organs are located at the junction between muscles and tendons. These receptors sense the tension of a muscle and correspondingly the exerted contractile force. Receptors in the joint capsules signal the flexion or extension of a joint. In conjunction with visual, vestibular, and cutaneous feedback these three mechanoreceptors close the feedback loop in healthy human motor control. However, it is not known exactly what parameters are communicated and how they are transmitted and processed. The individual attributes of position, velocity, acceleration, and force may be sensed directly at the muscles and joints, embedded in the rate of transmission, or calculated in the cerebromotor cortex. Regardless, the adoption of artificial prosthesis components and corresponding EMG control system may require an adjusted feedback system to compensate for the differences between the healthy human motor control loop and the replacement system.

With the inclusion of multiple parameters, the haptic mapping schematic needs to be expanded to communicate these attributes in parallel. Brewster's work with tactons encoded two pieces of information about a cell phone call using amplitude modulation and rhythm (Brewster 2005). A similar approach may be used in the prosthetic feedback system. To ensure perceptually different stimuli, MacLean proposes multidimensional scaling (MDS) as a method to differentiate between haptic signals of different parameters. MDS is a tool for analyzing complex scenarios and provides a means of representing complex perceptual data by placing haptic stimuli in a Euclidean space of specified dimension such that interstimuli distances approximate specified dissimilarities (MacLean 2003). This method will be useful as the number of haptic signals necessary to communicate ankle state increase. In both the amplitude modulated pattern and the gap detection pattern three of the nine actuators were used to successfully communicate nine different meanings. This leaves an additional six vibrators to simultaneously communicate other ankle characteristics using other signals. If these multiple stimuli are presented in either closer temporal or spatial proximity multidimensional stimulation effects need to be considered to ensure that the individual messages are not confounded by masking or suppression.

15.6 Electroactive Polymers

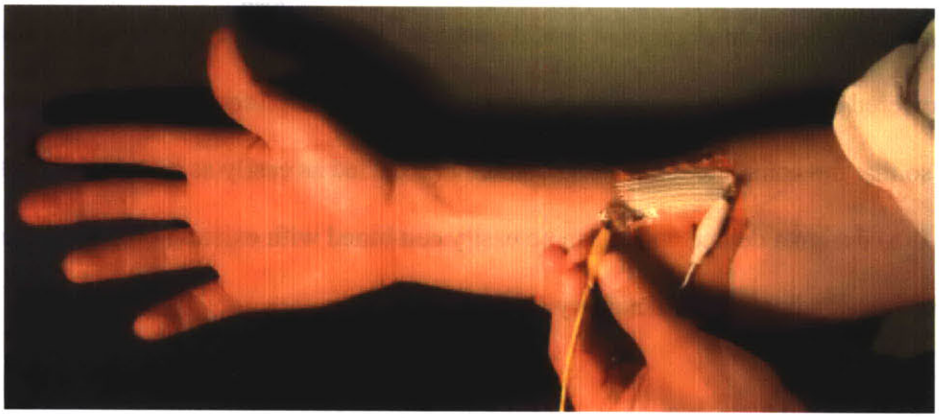

Figure 27: First prototype of an arm mounted tactile display using an electroactive polymer (Bolzmacher 2005).

Different types of actuators may be useful in conveying different physical parameters of the prosthesis. Recent work shows that these electroactive polymers may be viable as tactile actuators in portable haptic displays. EAPs (electroactive polymers) are polymers that change size and shape in response to electrical stimulation. Several characteristics of EAPs are advantageous in designing wearable haptic displays. These include the weight, volume, compliant form factor, speed, efficiency, and achievable strain (Herr & Kornbluh 2004, Bolzmacher 2005). They may be used to impart variable sensations of both force and vibration of different intensities and frequencies. However, these actuators are still being produced in experimental conditions and have not been thoroughly characterized or optimized for performance. Future iterations of the prosthetic feedback system may include the successful integration of these actuators.

15.7 Portable System

To improve the ease of testing with mobile subjects, steps need to be taken to convert the display into a portable system. Currently, the system is connected to a data acquisition card hardwired into a desktop computer running Matlab Simulink. Intermediate experimental platforms may be controlled by a small computing module such as the PC104 worn on a waist module worn by the subject while maintaining the ease of prototyping with Matlab Simulink. As the vibrotactile mappings are formalized into a more permanent structure the control software can be transferred onto a PIC embedded with the rest of the ankle electronics. The ideal configuration would integrate the actuators with a wireless communication unit and be powered by a local rechargeable battery. All the hardware should be completely incorporated into the liner so that the complete unit can be donned and doffed as easily as a normal socket liner. The ultimate goal is to design a device which can be easily combined with existing prosthetic components without requiring specialized complementary components.

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