Lower Limb Motor Coordination and Rehabilitation Facilitated through Self-Assist

by

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*Haven't we **all** earned it?*

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ABSTRACT

The shortage and expense associated with human therapists is a limiting factor in the time that patients spend in therapy. Although robotics may alleviate a portion of this problem, the autonomous robotic controllers have not yet demonstrated the adaptability or success of a human in administering rehabilitation assistance. Self-assisted rehabilitation through patient-operated telerobots provides a means to offer accessible, adaptable, task-specific rehabilitative practice to patients with neurological injury. Upper limb therapy augmented with robotic bimanual self-assist has been clinically shown to demonstrate greater improvements in range of motion and functional recovery when compared with traditional therapy alone. This dissertation generalizes bimanual assistance through rehabilitation telerobots to applications for lower limb rehabilitation. Using electromechanical devices under real-time control, we have built a scientific foundation supporting the hypothesis of lower limb motor facilitation through upper limb involvement.

A critically relevant concern to assistive rehabilitation is the degree to which the lower limbs are capable of coordinating with assistance to limb movement. To evaluate self-assist and traditional techniques, we compared subject coordination with exogenous forces commanded by the patient, a computer, or a human therapist, and applied to the lower limbs through a telerobot (n = 12). We found that subjects exhibit improved anticipation and compensation when the assistive (or disturbance) forces are self-

generated. This performance enhancement demonstrates that subjects can coordinate their muscle activation with self-generated assistive forces in a more appropriate manner than is possible when coordinating with a secondary agent. The centralized control inherent in self-assist provides a expectation of forces, through efference copy, that is more intuitive than an expectation of therapist or computer assistance, which is generally developed slowly from previous experiences and communication. It must be noted that motor control of both the upper and lower limbs represents a significantly increased cognitive load than either alone. We evaluated subjects performing coordinated motor tasks where the upper and lower limbs simultaneously manipulate a single dynamic object (n = 7 separated into two experiments). We demonstrated that neurologically intact subjects are capable of sustaining the increased cognitive demand provided that the motions of both effectors are sufficiently similar, spatially and temporally.

In another motor adaptation study, hemiparetic subjects (n = 15) practiced dorsiflexion of the impaired ankle (toe raises) with upper limb assistance offered through a teleoperator and directed by an automatic controller, an experimenter, or the patients themselves. During training blocks with assisted practice, subjects demonstrated improved task performance. When aware of the assistance provided, as from self-generation, subjects maintained high levels of lower limb muscle activation while improving performance. Assistance provided from an outside agent (an experimenter or a computer) resulted in diminished muscle activation implying reduced effort on the part of the subject. In addition, subjects demonstrated an overall improvement in the range of motion and smoothness of motion during unassisted active dorsiflexion following the short term assisted practice.

CHAPTER I

Introduction

This dissertation presents a robotic rehabilitation paradigm intended for lower limb therapy following neurological injury. The shortage and expense associated with human therapists is a limiting factor in the time that patients spend in therapy. This shortage might be partially alleviated by integrating robotic technology into rehabilitation. Robots can be made widely available by their reproducible nature and ability for continuous operation. However, autonomous robotic controllers have not yet demonstrated the adaptability or success of a human in administering rehabilitative assistance. It is possible to combine the hardware benefits of robotics with the controlling benefits of a human. Instead of using autonomously controlled robots, therapy could be administered through a teleoperator. A teleoperated robot consists of a *slave* electromechanical device that follows a force/motion plan generated by a human manipulating a separate *master* device. Human-directed telerobots provide a means to offer accessible, adaptable, task-specific rehabilitative practice to patients with neurological injury. Furthermore, the human in the best position to administer the rehabilitative assistance might very well be the patient himself, essentially offering *self-assist*. Upper limb therapy augmented with robotic bimanual self-assist has demonstrated some clinical success in terms of range of motion and functional recovery when compared with traditional therapy alone. Although self-assist has focused on bimanual applications, it is generalizable to applications for lower limb rehabilitation. This document presents a scientific foundation supporting the generalization to lower limb self-assist.

This chapter presents the structural framework from which the idea of generalized self-assist was conceived. We explore some current methods of neurorehabilitation in clinical use and demonstrate how traditional therapy techniques inspired robotic substitutions that mimic those techniques. The review continues with a presentation of alternative control strategies in robotics, including bimanual teleoperators. Finally, we make an argument for the benefits and generalizability of self-assist, illustrating how our work contributes to the existing literature.

1.1 Prevalence of neurological trauma in the United States

According to the American Heart Association, there are 5.8 million stroke survivors living in the United States, with approximately 780,000 new or recurrent strokes annually. Upwards of 48% of this group suffers lasting hemiparesis and nearly half of those are non-ambulatory. The long term effects of stroke, or cerebral vascular accident (CVA), may include decreased motor function, muscle co-contraction and spasticity, weakness, and abnormal movement synergies [1, 2]. Individuals with spinal cord injuries (SCI) constitute another large population with chronic neurological impairment. The National Center for Injury Prevention and Control reports that nearly 200,000 people are living with a disability related to a SCI and approximately 11,000 people sustain a new SCI annually [3]. Very few of these patients, less than 1%, achieve complete neurological recovery by their initial hospital discharge while the remaining population sustains some level of paraplegia. The total annual cost of CVA and SCI in the United States is estimated at $53 billion where up to half of that may be attributed to indirect costs from disability, low productivity, and other factors [3, 4]. In addition to the financial costs to the individual and society, the loss of motor function, ambulation, and independence can drive many of these patients into clinical depression.

Intense acute and chronic rehabilitation efforts are centered on returning motor control and functionality to patients who have suffered a neurological trauma. In the past, it was commonly believed that damage to the nervous system was irreparable; however, recent studies have demonstrated that with appropriate intervention, recovery may be possible [5, 6]. Functional recovery after neurological injury depends in large part on central nervous system plasticity and the reorganization of neural pathways [6, 7]. Neuroplasticity refers to the change in organization or synaptic strengths of the brain or the spinal cord as a result of experience or rehabilitative recovery [8, 9]. With appropriate training following a neurological injury, the nervous system can reallocate resources in the brain or spinal pathways to adopt functions previously performed by the now damaged portions. Recent research is revealing that both the spinal cord and the motor cortex are capable of a great deal more plasticity than traditionally thought [8, 10]. The key to increasing motor recovery after neurological injury is task-specific training that induces activity-dependent plasticity. Researchers are developing new techniques that provide patients with functional exercises to promote activity-dependent plasticity and maximal recovery. These strategies have taken many different forms; constraint-induced upper limb therapy and therapist assisted locomotor training are two of the new therapies that are demonstrating positive outcomes [11, 12].

1.2 Therapist assisted locomotor training

Task-specificity and repetition appear to be dominant training characteristics to promote reorganization of the neural pathways [6]. Therapist assisted body weight supported treadmill training encompasses both task-specificity and repetition. In manually assisted bodyweight supported treadmill training, or locomotor training, partial bodyweight support is provided while a team of physical therapists manually assists the patient's lower limbs through a stepping motion on a treadmill. The resulting gait-specific sensory feedback, including ground reaction forces, stimulates muscle activity patterns and neurological recovery [12]. Clinical studies have reported that after several weeks of locomotor training, participating patients were able to support more of their own body weight, increase treadmill walking distance and speed, and, in many cases, improve over ground walking capabilities and reduce dependence on walking aids [13, 14]. Although this therapy has had great success, it requires a team of specially trained, highly skilled therapists. In addition, the therapists must endure intense labor demands that can result in fatigue and repetitive stress injuries.

1.3 Robot-assisted locomotor training

The limitations of therapist assisted treadmill training have inspired a number of research groups to explore the use of robots for providing the mechanical assistance necessary for locomotor training [15, 16]. Robot-assisted locomotor therapy, if successful, would enable delivery of locomotor training to much larger patient populations. The electromechanics of robots also allow therapists to precisely administer reproducible aid while quantitatively monitoring the progress of the patient [17]. There are a number of robotic locomotor devices used in research and in clinical studies, including the Pelvic Assist Manipulator (PAM), the Lokomat, and the Mechanized Gait Trainer (MGT) [18]. All of these systems operate in a similar way: the patient's limbs, while strapped into the device, are driven through a stepping pattern by computer-controlled robotic actuators. In theory these robotic devices appear to be ideal replacements for the human therapists; a robot is not limited by fatigue or biomechanical injury. However, due to fundamental differences between robot-assisted and traditional locomotor training, the clinical results for the robotic devices have shown limited success [19, 20].

Patient passivity offers one possible explanation for the reduced success of the devices. Studies have shown that active involvement in the production of a motor pattern results in greater motor learning and retention than passive movement [21-23]. When in operation, these large robots dominate the motion of the lower limbs. This results in an unchanging gait pattern regardless of subject participation. This is in contrast to traditional body weight supported treadmill training where the assistive effort offered by the therapist is continuously adjusted to the needs of the patient. The limits of upper limb strength of a person may prevent a therapist from driving the patient's lower limbs in a non-backdrivable fashion.

A human therapist is able to adapt the assistance patterns to the changing behavior of the patient's limbs. The therapist's hands can accommodate and adapt both in the short-term (over the course of a single movement) and in the long-term (over the course of a therapy session or longer). Underlying the short-term ability to adapt is the smaller mechanical impedance, or equivalently, the greater amount of compliance used by the therapist compared to a robot. In part because the therapist uses a lower impedance to impose motions on the patient's lower limbs, he is better able to detect the patient's muscle action. The therapist can then respond by adapting his assistance over time, gradually decreasing assistance as he observes the muscle action increasing appropriately. The robot, unlike a human therapist, cannot easily accommodate such modifications and progression on the part of the patient. When a therapist uses his hands to guide a patient's limb through a movement or task, the goal is to promote independent motor function. Thus the therapist will wean away mechanical input and let the patient take over. The therapist carefully gauges, then reduces and removes the manual assist. The therapist also complements the provisional manual assist with verbal and other forms of communication. This will encourage independent, coordinated activity on the part of the patient and thereby guide the activity-dependent neural plasticity.

A number of groups have recognized the potential benefit of greater compliance and adaptation in rehabilitation robots and are currently pursuing research in this area [24-26]. Most notably, Riener and colleagues are working on ways to adapt the Lokomat to incorporate patient-cooperative control. They have begun experimenting with three methods of automatic control that incorporate the intentions of the patient. These strategies include adding compliance to the fixed reference trajectory, adapting the robot behavior based on forces applied by the patient, and providing visual performance feedback to the patient [25].

1.4 Robotic alternatives in upper limb stroke rehabilitation

The rehabilitation robots that we have thus far discussed generally execute high impedance, feedforward motion plans to drive the patient's legs. In upper limb stroke rehabilitation, robots are frequently integrated without feedforward execution of a recorded pattern, but by engaging the patients with low impedance interactive games and virtual environments. In Volpe, et al. [17] subjects received robot training to supplement standard post-stroke multidisciplinary rehabilitation. In this study, an MIT-MANUS was used to assist and guide the subject's arm in a target reaching game. Patton and colleagues use a planar manipulandum to execute a number of strategies in rehabilitation games, including force resistance, error amplification, and error reduction [27, 28]. Researchers concluded that these novel therapeutic approaches can reduce impairment and improve motor learning.

Upper limb rehabilitation telerobots can be used to administer force/motion plans to the impaired limb while maintaining low impedance control. The motion plan is formulated and executed by a human directing the master device, while the slave device drives the impaired limb. In a novel application of this control structure, the patient himself is positioned to command the master, thus offering *self-assist*. This is specifically applied for the upper limb rehabilitation of stroke patients, where the less affected arm controlling the master directs the assistance provided to the paretic arm through the slave. The Mirror Image Motion Enabler (MIME) used by Lum, Burgar, van der Loos, and co-workers has examined bilateral self-assist in stroke patients during planar reaching motions [29]. The slave device for this system is a commercially available PUMA 560 robot. The PUMA 560 is a large, high impedance robot, much like the gait trainers discussed previously. However, in the teleoperated system, the low impedance and adaptability of the control is derived from the human operating the master. Clinical outcomes using MIME are quite promising; subjects demonstrated improvements in Fugl-Meyer scores, strength, and range of motion when robotic self-assist augmented traditional therapy [30]. Similar clinical results were found by Hesse and colleagues during supination/pronation motion of the wrist using the BiManuTrack [31]. These promising results are derived in part from the involvement of neural coupling across the two hemispheres of the brain. Later in this chapter we will further review other applications of bimanual coordination and their impact on rehabilitative success.

1.5 Neural coupling and inter-limb coordination

Neural coupling in part refers to the neural synergies that exist between two limbs. This coupling is particularly apparent during voluntary actions requiring interlimb coordination. Kelso and colleagues have presented strong evidence to support neural coupling during rhythmic bimanual tasks in neurologically intact subjects [32]. Subjects are able to simultaneously flex/extend both wrists when the motion is temporally and spatially similar. Tasks that can be performed easily with a single hand become very difficult or even impossible when the second hand is added in an asymmetrical fashion [33]. The MIME rehabilitation studies, with non-rhythmic motion, have further supported the existence of bilateral neural coupling. Using MIME, it appears that the involvement of the less affected limb improves activation and performance of the more affected limb [29, 34] by taking advantage of this neural coupling. Neural coupling is not limited to bimanual movements; it also exists during coordinated motions of the upper and lower limbs [33, 35, 36]. Baldissera and colleagues have produced a series of studies on healthy subjects examining the neural coupling between the upper and lower limbs in simple rhythmic motions such as coordinated, single-joint flexion/extension [33, 36, 37]. They have demonstrated that due to neural coupling, coordinated motions of an ipsilateral hand and foot are easier when they are spatially similar. There might also be benefits in performing similar, coordinated motor tasks with the upper and lower limbs over performing the same tasks with the limbs independently. Huang and Ferris showed increased lower limb muscle activation in a cyclic task when the upper limbs were involved [38], compared to when subjects used the lower limbs alone. It is not clear how neural coupling between limbs is affected by cerebral vascular accident, but it has been suggested that this coupling could actually be exaggerated following stroke [39]. The neural coupling between the upper and lower limbs might prove advantageous in rehabilitation schemes that involve both sets of limbs, including patient self-assist between the upper and lower limbs.

1.6 Generalized self-assist in rehabilitation

Bilateral neural coupling can be argued as one of the underlying mechanisms explaining the clinical success of bimanual self-assist. This principle, paired with the extensive evidence for other types of neural coupling, particularly between the upper and lower limbs, suggests a generalized model for self-assist rehabilitation where the neurologically intact or more functional portion of a patient's motor system directs the assistance offered to the neurologically impaired portion. This generalization would be predominantly useful in lower limb rehabilitation, including locomotor therapy. A person who has suffered an incomplete spinal cord injury affecting the lower limbs would use his unimpaired upper limbs to command assistance, while a hemiparetic stroke subject would assist his paretic lower limb with the contralateral upper limb. In the case of hemiparetic locomotor rehabilitation, one might suggest using the less affected leg to assist the impaired leg, in a bilateral manner that directly parallels bimanual therapy. This may not offer a practical solution. Locomotion is a rhythmic task where the legs progress through the gait cycle with different phase. Thus, the less affected leg is otherwise occupied and unavailable to offer assistance.

In addition to neural coupling in the motor centers of limb motion, there exist other physiological mechanisms by which we can reason that generalized self-assist may benefit rehabilitative therapy over robot-assisted therapy, and even traditional therapist-assisted therapy. Currently in robot-assisted locomotion, the movement of the robot does not adapt to the activity of the patient and will likely dominate the contribution of the patient. This leads to reduced mental and physical effort on the part of the patient and contributes to the limited success of fully autonomous locomotor robots. If the patient has active control over the mechanical assistance and full knowledge regarding the amount of assistance applied, it is unlikely that he will dominate the cooperative movement with the assistance. Also, patients will necessarily remain cognitive participants in their rehabilitation because they are responsible for generating and executing the motor plans for both the lower limb motion and the proffered assistance. Recall that the active production of motion facilitates learning and motor memory [21, 22, 40]. Self-assist encourages increased exercise of both the mental and physical muscles involved in motor control.

The human nervous system is capable of controlling a complex system by integrating multiple sensory inputs, previous experiences, and intuition. This type of control is updated in a highly adaptive manner that cannot readily be duplicated by a series of computational control laws. We maintain that in determining a replacement for a physical therapist, it is better to select another human (the patient) over an automatic controller because this biological system will provide assistive control that is capable of matching the degree of adaptation inherent in therapist-patient interactions.

While external mechanical assistance from a therapist or robot may help guide a patient's limb through the correct motion, there is the possibility that the nervous system will interpret the mechanical assistance as a perturbation. An assessment of a perturbation could lead to reflexive reactions, confuse the nervous system regarding movement dynamics, and lead to difficulty forming an internal model for neural control [41, 42]. A host of papers by Wolpert and colleagues have shown that humans attenuate self-induced afferent stimulation to a greater extent than externally induced afferent stimulation [43-45]. Self-assist centralizes the generation of neural command signals for the lower limbs and for the assistance given to the central nervous system of the patient. The centralized access to efference copy and afferent sensory feedback will allow subjects to treat the mechanical assistance as part of the motor command rather than as an external perturbation. Patients will further benefit from centralized motion control by improving the temporal and spatial coordination of muscle activity with assistance. This will ensure productive combination of the forces driving the limb motion, that is the muscle force and the assistive force. It will further prevent the assistance from fighting the productive component of volitional muscle effort in limb motion. Patients will simultaneously experience execution of their motor plan with successful completion of the desired motion.

In practice, self-assist will be realized through teleoperators. Teleoperators eliminate the constraints imposed by the physical limitations of the assisting and assisted limbs. A hand would not be able to comfortably reach the leg to provide assistance, nor would it be strong enough. A telerobot could easily resolve the strength mismatch and proximity issues associated with the hand assisting the leg motion. The force-motion profiles of the hand would still drive the force-motion profiles externally applied to the leg. Independent master and slave interfaces for the assisting and assisted limbs allow for the independent adjustment of the control gains, force feedback, or position feedback to each device. These parameters could be continuously adjusted over the course of several therapy sessions, within a single session, or even within a single exercise. This gives greater flexibility in the design of the rehabilitation scheme in terms of the type of exercise or task performed and the assistance possible.

Self-assist implemented with telerobots stands to offer several advantages over other robotic control strategies. The key advantages discussed in this chapter can be summarized as the following:

- Generating and executing motor plans in more than one limb simultaneously takes advantage of neural coupling in the control centers to encourage muscle activation [38, 46]
- Self-assist mandates that patients remain cognitively active, which promotes motor learning and retention [21-23, 40].

- Patients can be encouraged to apply an "as needed" strategy for aid, thus subjects will receive assistance when necessary and push the limits of their abilities.
- Efference copy will allow subjects to appropriately interpret afferent sensory feedback [44].
- Centralized control provides subjects with more information, such as the amount of assistance provided, the timing of its application, and the desired force trajectory of the assistance [42].
- Implementation through telerobots offers all of the benefits associated with autonomous robot therapy such as quantitative measures of performance, virtually limitless strength and endurance, and reduced risks for therapist injury.

Based on the literature pertaining to neural coupling, motor control, motor learning and relearning, and upper rehabilitation techniques, we hypothesize that these benefits may be present during generalized self-assist. However, there are many questions left to answer in developing this foundation. Even within the body of work pertaining to bilateral upper limb movements, there is no consensus on immediate effects of this type of assistance in terms of facilitation and muscle recruitment, nor on the resulting after effects when this type of assistance is used during training.

1.7 Bilateral training in stroke rehabilitation

Earlier in this chapter, we described a few examples of bimanual self-assist that reported clinical success when augmenting traditional therapy (ex. MIME and BiManuTrack). Bimanual self-assist is one example of the more general bilateral training, or rehabilitation practice with simultaneous activation of both arms. Many research groups are investigating the efficacy of bilateral training for stroke rehabilitation. A review by Cauraugh and Summers [47] proposes three mechanisms by which bilateral training promotes neural plasticity. First, bilateral movement may reduce interhemispheric inhibition. When performing upper limb unilateral motions, the ipsilateral hemisphere is inhibited to prevent mirrored motions of the opposite upper limb. The inhibition affecting the paretic side may be greatly exaggerated in stroke subjects compared to neurologically intact subjects. Bilateral training may act to normalize this increased inhibition. The second proposed mechanism relates to neural crosstalk. During bilateral movements the arms coordinate as a single unit with a central regulatory mechanism dominating the organization and control of both limbs. Thus identical motor plans are specified for both limbs and are reinforced through interhemispheric crosstalk. Finally, it has been suggested that sensorimotor integration may be key during motor rehabilitation following stroke. Bilateral training involves increased sensory feedback and integration of afferent information.

The clinical studies investigating bilateral training for stroke rehabilitation have yielded mixed results. There exists an open debate in the literature questioning which patient population might stand to benefit from bilateral training, what is the best regimen for prescribing it, and even if bilateral training facilitates motor recovery at all. The inconsistent findings across bilateral movement studies have limited overall conclusions about the efficacy of bilateral training.

Cunningham and colleagues studied discrete elbow extensions and found facilitation during bimanual conditions in all but one subject [48]. Subjects demonstrated improvements in smoothness of motion, reduced peak velocities and velocity oscillations during bilateral motions. Although there was a trend for a slight increase in movement times in the bilateral condition, this was not significant compared to unimanual conditions. Positive training effects have also been reported by Luft and colleagues in a study involving 21 chronic stroke subjects [49]. Subjects were divided into two groups where the first group (n = 9) received bilateral arm training with rhythmic auditory cues while a second group (n = 12) served as a dose-matched control. The bilateral group performed linear rhythmic movements in the transverse plane. The motions were coordinated and either in-phase or anti-phase. Six of the nine subjects who received bilateral training demonstrated increased fMRI brain activation, representing a significant improvement for the group. As a group, no functional improvement was discernable; however, when examining only the six subjects with increased fMRI response, the bilateral training improved arm function to a greater extent than the control therapy.

Other studies have been unable to demonstrate facilitation during bilateral training. Tijs and Matyas reported no positive effects during or following bilateral training [50]. Using three copying tasks, no facilitation was observed in terms of both spatial and temporal performance metrics that included jerkiness, speed, task duration, accuracy and postural positioning. Lewis and Byblow reported detrimental consequences of bimanual coordination for both neurologically intact and stroke subjects during circle-drawing tasks [51]. Mixed results have even been reported within research groups. In earlier studies, Mudie and colleagues reported that task specific, homologous bilateral training demonstrated rapid and significant decreases in abnormality of movement during subsequent unilateral performance [52]. However in a subsequent study using the same shoulder abduction and wrist extension tasks, they were unable to demonstrate significant differences between bilateral and unilateral training in persons with densely paralyzed upper extremities [53].

A systematic review and meta-analysis performed by Stewart and colleagues concludes that despite the mixed results, a conservative look at the literature supports bilateral training as an effective protocol for stroke rehabilitation [54]. The success of bilateral training appears highly dependent on the particular population tested and the specificity of the training task. It is unclear if the training necessarily requires symmetrical in-phase motions or if anti-phase or some other phase offset could also produce a positive result [55]. Success of bilateral training might also be due to the verbal feedback/instruction and general encouragement, or increased task-specific training that generally accompany bilateral protocols [50]. The lasting effects and impairments associated with stroke are greatly varied. It is therefore unlikely that functional recovery will be uniformly observed from a single intervention. Given the small sample sizes, the breadth of lesion type, time since stroke, and impairment level among subjects, and the variety of task specifics and complexities, it is not surprising that studies have yielded mixed results [47, 54].

1.8 Summary of experiment goals

Given the mixed results and controversy surrounding bilateral training in stroke rehabilitation, the efficacy of bimanual or generalized self-assist demands further research and exploration before definitive conclusions can be drawn. A summary of potential benefits and theorized mechanisms supporting generalized self-assist was presented at the end of section 1.6. The aim of this dissertation is to further examine the scientific underpinnings that will provide a foundation to support lower limb rehabilitation with upper limb self-assist. The goals and contributions of this thesis are as presented below.

Aim 1

Demonstrate that the increased cognitive load from multi-limb control does not outweigh the advantages gained from coordinated multi-limb control in dynamic object manipulation.

There is an inherent increased cognitive load associated with multi-limb control over independent limb control. In Chapter 3 we present two studies that explore how this increased cognitive load affects motor performance in dynamic object manipulation tasks. Multi-limb control, when compared to independent limb control, may involve more concentration, the development of more complicated motor plans, and activation of more muscle groups, which could all be detrimental to motor performance. At the same time, coordinated control in a single dynamic task may offer improved authority, more afferent sensory information, and the development of an improved mental model of the task. These benefits could lead to improved motor performance. This dissertation addresses how these combine to affect motor performance in dynamic tasks. We hypothesize that subjects will be able to sustain the increased cognitive load associated with multi-limb motion and that the benefits will outweigh the detrimental effects. If subjects can accommodate the increased cognitive demand associated with non-homologous multi-limb control, then it stands to reason that subjects maintain the cognitive capacity for self-assist. That is, they will be able to handle performing a lower limb task while directing assistance with the upper limbs.

Aim 2

Demonstrate that subjects develop anticipatory adjustments in the lower limbs that can be used to improve coordination when compensating for externally applied loads.

It has been shown that people will develop motor actions in anticipation of expected disturbances when those disturbances are self-triggered [56, 57]. In Chapter 4 we present a study that examines the comparative development of anticipatory adjustments in the upper and lower limbs in response to an externally applied load disturbance when that disturbance is triggered a) without warning, b) in a predictable fashion, or c) by the subject through a volitional action at the hand. We hypothesize that subjects will anticipate imminent disturbances with their lower limbs when they self-generate the disturbance, even through an action as small as pressing a handheld trigger. If subjects are able to accommodate load disturbances at the lower limbs better when they are involved in generating those disturbances, subjects will likewise be able to accommodate load assistance at the lower limbs better when it is self-generated. This would suggest that subject efforts can be more easily integrated and coordinated with self-assist than with assistance from an outside agent such as a therapist or autonomous robot.

Aim 3

Design a functionally relevant task and determine if short term assisted practice in that task, including practice with interlimb coordination, will benefit subsequent unassisted capabilities in hemiparetic stroke subjects.

In Chapter 5 we present an experiment that examines the active dorsiflexion capabilities of subjects who have suffered a stroke and maintain lasting hemiparetic effects. We present an exercise wherein subjects must dorsiflex the impaired ankle to avoid colliding with a series of obstacles. This task allows subjects to practice lifting the impaired toe, a challenge that arises during locomotion. We provide subjects with various types of assistance in completing a training period. We hypothesize that given time to practice with assistance, subjects will improve the unassisted active dorsiflexion capabilities of the impaired limb including range of motion and muscle activation.

Aim 4

Assess the immediate effects of self-assist, computer-assist, and experimenter-assist on task performance and coordination between patient and assistance for the impaired lower limbs of hemiparetic stroke subjects.

In rehabilitation therapy, assistance can be offered from a therapist, an autonomous robot, or the patient through a telerobot, as in our definition of self-assist. In Chapter 5 we present a study that models these three types of rehabilitation as experimenter-assist, computer-assist, and self-assist, respectively, in performing a functionally relevant task: ankle dorsiflexion. Participants in the experiment have all suffered a stroke and demonstrate lasting hemiparetic effects. We examined if self-assist offers positive effects in terms of better coordination between the lower limbs and the assistance when compared to assistance from an external agent. Bimanual studies report mixed results regarding facilitative effects of self-assist. We hypothesize that self-assist will encourage muscle activation over conditions where the applied assistance is occluded such that the subject is unaware of how much assistance is applied. It is a significant challenge for us to determine the effect of assistance type on the coordination between the efforts of the subject and the assistance applied, especially in a functionally relevant task.

In this chapter we have presented motivation and support for self-assist, generalized for lower limb rehabilitation with upper limb assistance. The remainder of this document will continue this theme by developing a scientific foundation for generalized self-assist and address the aims outlined above. Chapter 2 discusses the hardware devices used in the experiments that will be discussed. Chapter 3 presents two separate studies, both focused on addressing the effects of increased cognitive load during coordinated multi-limb motor control in neurologically intact individuals. Continuing with neurologically intact subjects, Chapter 4 examines anticipatory adjustments in the upper and lower limbs in response to upper limbs volitional actions. A final experiment is presented in Chapter 5 where we examine various methods of assistance application during dorsiflexion in a stroke subject population. Finally, the dissertation is concluded in Chapter 6 with a discussion of the major contributions of this work.

CHAPTER II

Technical aspects of hardware and controller design

Self-assist between the upper and lower limbs is most easily executed through teleoperated robots. Separate master and slave devices allow greater distance between and comfort for the assisting and assisted limbs. The computer modulation of electromechanical devices allows them to be force matched to the capabilities of each limb. Quantitative performance measurements assist patients and therapists in tracking progress over any time duration. Haptic and visual feedback can be scaled to best encourage recovery. Teleoperators also enable other guiding features such as virtual walls to limit motions that fall outside normal ranges, damping properties to control involuntary spasms, and additional safety features to prevent accidental injury.

A fully realized self-assist apparatus for clinical locomotor rehabilitation would likely entail a multi-joint exoskeleton powering ankle, knee, and hip assistance. The master device could resemble either an intuitive hand-held joystick or an upper extremity exoskeleton that parallels the lower extremity one. Such devices involve complicated multi-degree of freedom motions and would require intricate engineering design and construction. Before anyone can assume such an endeavor, it is necessary to create a scientific foundation for upper/lower self-assist. We therefore examined motor control in simple, single joint, movements to test the hypothesis that centralized, patient-controlled assistance offers benefits to lower limb rehabilitation. We selected the ankle joint as the single joint of interest because of its importance in standing and walking.

Two distinct teleoperated devices were constructed to test ankle motor control with self-assist. The first apparatus is capable of presenting loads comparable to those felt by the ankles during standing. The second apparatus is smaller in scale but portable for transportation to patient centers. Both sets of hardware can be arranged in configurations to meet particular experiment and subject needs. Both are motorized with computer modulation to allow different motor tasks. This chapter presents the technical aspects of the hardware and software that were designed for the experiments presented in this dissertation.

Figure 2.1 Device 1: Ankle-strength-matched platform. This reconfigurable device limits the subject's motion to plantar flexion and dorsiflexion of the ankle. It is capable of presenting programmable loads to the ankle on the same scale as those felt during everyday tasks, such as standing. In the figure, the device is ready for a seated subject but can be reconfigured for standing as well.

2.1 Device 1: Ankle-strength-matched platform

The first test apparatus, Device 1 (Figure 2.1), was custom designed to present programmable loads to the subject's upper limbs, lower limbs, or a combination of upper and lower limbs. This device offers extreme flexibility in its configuration for testing a number of motor control hypotheses (Figure 2.2 and Figure 2.3). It is capable of presenting loads to the subject's feet approximately equivalent to what a healthy, neurologically intact adult would expect to experience during standing (around ±150 Nm about the ankle). The apparatus, in its entirety, can be broken down into the following hardware elements and the following software elements.

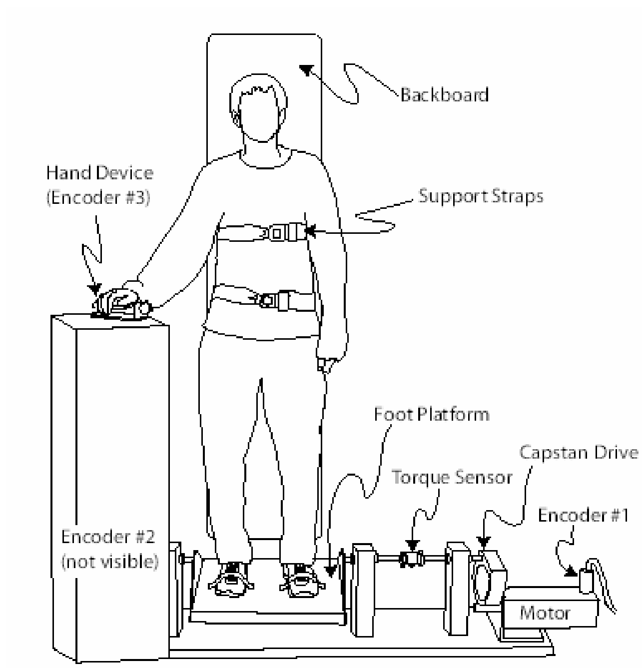

Figure 2.2 Device 1: Standing configuration. In one configuration of Device 1, the subject stands on the footplate and is secured to the wall through a backboard. This provides support and grounding while the subject performs motor tasks with his ankles. This illustration also shows how the user would operate a motorized hand interface.

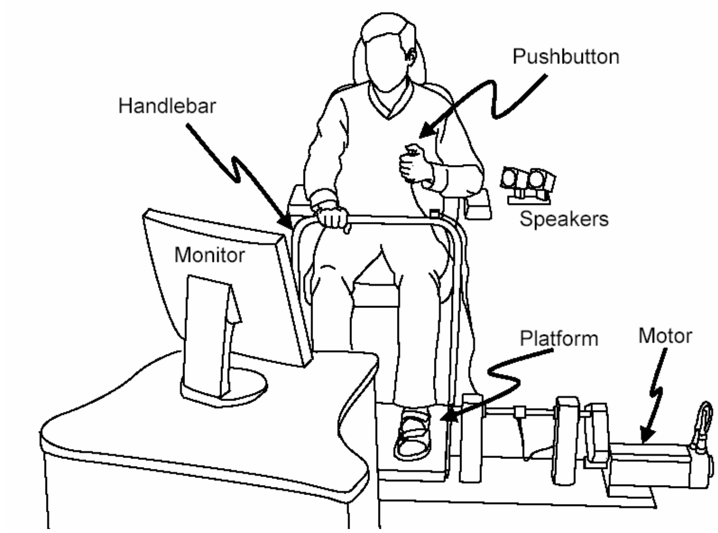

Figure 2.3 Device 1: Seated configuration. In a second configuration, the subject sits in an adjustable chair. Interaction with the footplate is possible through either his feet secured directly to the footplate or through his hand grasping a handlebar rigidly attached to the footplate. In addition, audio feedback is available through the speakers. Visual feedback is available through animation presented on the monitor. A handheld pushbutton can be used to register discrete input from the user.

2.1.1 Major mechanical components and sensors

The central element of Device 1 is the footplate or platform (Figure 2.4). Constructed of aluminum, wood, and Plexiglas, the footplate is the direct interface between the subject's feet and the programmable loads. The user's feet are secured through snuggly fitting, interchangeable shoes fixed to the footplate. This rigid component swings only in the sagittal plane of the subject with the axis of rotation approximately aligned with the ankle axis of rotation. This single axis motion allows only plantar flexion/dorsiflexion of the ankle and restricts off-axis motion.

The footplate is actuated with a low inertia brushless servomotor, the Kollmorgen Goldline Series 406-B (Figure 2.4), coupled with a ServoStar power supply/amplifier. This 7.4 hp motor offers continuous stall torque of 18.6 Nm with short duration peak torques of up to 49.5 Nm possible. The transmission between the footplate and the motor consists of steel shafts, aluminum couplings, and a capstan drive. The zero backlash, cable driven transmission provides a mechanical advantage of 7.3, allowing continuous stall torques of 136 Nm with much larger short peaks possible.

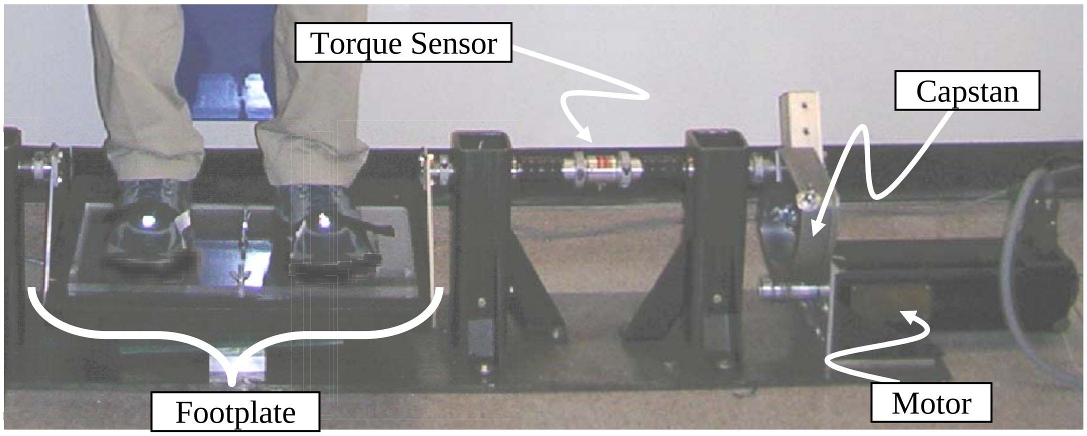

Figure 2.4 Major hardware components of Device 1. The footplate acts as the direct interface between the subject's feet and the applied loads. Loads are transmitted from the motor through the capstan transmission to the user. The torque sensor is used in feedback to compensate for some of the transmission dynamics.

The sensor suite for this device includes both torque and position measurements. A single axis torque transducer (GS Sensors, Model CS-1060-A2), positioned inline with the transmission between the footplate and the gearing, measures the actual load applied to the footplate. The angular displacement of the motor is measured by a resolver (2048 counts per revolution) embedded within the motor. An externally mounted optical quadrature encoder (BEI, Series E25 Incremental Optical Encoder; 2540 counts per revolution) measures the angular displacement of the footplate. This device also features a moveable single axis linear accelerometer (Crossbow Model CXL04), with a range of ±4g. The accelerometer position can be adjusted depending on the device arrangement for the particular experiment.

While the footplate provides the only lower extremity interface for Device 1, there are several upper extremity options. An aluminum handlebar rigidly attached to the footplate allows the subject to directly manipulate the position of the footplate with his upper extremity (Figure 2.3). While grasping the handlebar, the subject can use multi-joint motion of his arm to move his hand, and therefore the bar, through an arc of 0.7 m radius. This will in turn cause the footplate to pivot. This mechanical connection also enables programmable upper limb loads to be presented from the Kollmorgen motor. The sensors for the footplate can be used for position and torque measurements of the handlebar. This handlebar was used in a study to be presented in Chapter 4.

Haptic interaction with the subject's hands can also be achieved through a separate motorized single-axis hand device (Figure 2.2). With this device, subjects use single joint motion of the wrist (flexion/extension) to closely parallel the single joint motion of the ankle. An optical quadrature encoder (US Digital, Model HEDS-9100; 1024 counts per revolution) measures the angular position of the handle. This device, without the force-feedback, was used for an experiment presented in Chapter 3.

Visual information can be provided to the users through a 17" Dell LCD monitor. Experiment instructions, task animations, and performance feedback can all be presented through the monitor. Auditory information can be provided through one of two sets of computer speakers. One set of speakers is driven by an analog voltage output from the DAC board and presents real-time auditory feedback. This is important when the exact timing of the audio cues is critical. The second set of speakers operates in the Windows environment and presents auxiliary auditory information. This might include tones to indicate a switch in operating conditions or signifying the end of an experiment. Device 1 also has the capability to accept discrete input from the user through a handheld pushbutton.

2.1.2 Electrical interface and software

Data acquisition and control is handled by a Sensoray 626 data acquisition board embedded in a Pentium II target PC. The 626 board offers sixteen 14-bit analog inputs, four 14-bit analog outputs, and six encoder inputs with quadrature decoding. The designated PC runs the QNX real-time operating system at 1000 Hz. Development of the real-time control software is managed on a separate Pentium 4 host computer by MATLAB/Simulink with autocode generation by RT-LAB 6.2 Solo developed by Opal-RT Telnologies. For the experiments presented in this document using this device, the input data were collected on the target computer at 100 Hz, though higher frequencies are possible. All animations and data processing are handled by the host computer.

The target PC, amplifier, and other electrical elements are housed in an electronics cabinet. A central ground bus was used for signal grounds. This helped to prevent floating grounds, ground loops, and the associated signal noise. An inductor noise block was implemented to attenuate high frequencies and help guard against further noise issues. The power lines from the motor power supply were coiled around ferrite toroidal cores (Amidon Associates, Inc.) to create inductors, thus low pass filtering the lines.

The motor is controlled through a ServoStar 600 power supply/amplifier. The ServoStar software package allows easy control of the motor in position, velocity, and torque modes. For the experiments presented, the amplifier was operated in torque mode, where an input DC voltage designated the desired motor output torque. A built-in PID controller modulated the power supply current loop to generate desired output torque at the motor. Factory settings for the control loop gains were maintained. Because the mechanics of the hardware, specifically the cable driven capstan, present significant dynamic effects, a torque controller was implemented with feedback from the torque sensor. The general model of the system and this controller are presented in Figure 2.5.

The control architecture acts to increase the system bandwidth, force low frequency and stead-state torque matching, and effectively counter some of the hardware dynamics.

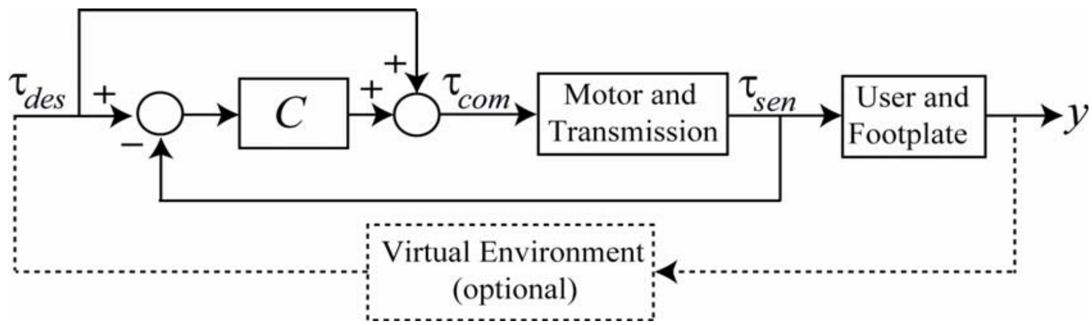

Figure 2.5 General motor control loop. The commanded motor torque (τ_{com}) is a function of desired torque (τ_{des}) and measured torque (τ_{sen}). The controller *C* acts to cancel some of the hardware dynamics inherent in the motor and the transmission by adjusting the motor torque such that the measured torque to tracks the desired torque. The measured torque is what is felt by the combined user and footplate, resulting in motion or footplate position (*y*). Optionally, the desired torque can be determined by the equations of motion of a virtual environment that accepts position *y* as an input.

2.1.3 Configurations

When standing, a person is able to maintain upright posture and modulate sway by adjusting his ankle torque (among other postural adjustments). When interacting with Device 1, the subject's feet are no longer in contact with a mechanical ground, the unmoving floor. Mechanical ground is provided through the subject's torso instead of his feet. Device 1 offers two possible grounding configurations: standing and seated. For a standing subject (Figure 2.2), the torso is secured to a rigid backboard (LSP Xtra EMS Backboard) with straps across the chest and hips; the backboard is then grounded to the wall. This prevents falling, restricts the torso, and isolates the ankle motion. In addition, it alleviates the need for ankle torque to maintain postural balance and allows subject interaction with programmable loads. The backboard allows the legs to be straight in a joint configuration that resembles standing, but does not provide any body-weight support. Subjects must be able to support their own weight.

Device 1 can also be used in a seated configuration (Figure 2.3). When seated, the biomechanics of the lower limb muscles are significantly different than when standing. Although this configuration is not an appropriate model for human standing balance, it removes the need for subjects to maintain their own body weight and allows the testing of other ankle motor tasks with smaller loads. The seated configuration employs a custom built fully adjustable chair (Figure 2.6). The height and fore/aft position adjustments make it possible to standardize body position for different sized subjects within an experiment. For example, the experiment presented in Chapter 4 required 90° knee flexion with a vertical shank.

This apparatus can be configured in a number of ways depending on the goals of the particular experiment; Figure 2.2 and Figure 2.3 present two of the possibilities. These two figures illustrate configurations used in the experiments presented in Chapters 3 and 4, respectively. They show a subject both standing and seated, both the haptic handle and the handlebar, and a number of the auxiliary elements such as the pushbutton input, the monitors for visual output, and the speakers for auditory output.

Figure 2.6 Adjustable chair. This SolidWorks model illustrates the fully adjustable subject chair. The height and fore/aft positions can be easily changed to suit individual subject anthropometry.

2.2 Device 2: Portable single hand/foot cradle

The second custom built apparatus, Device 2 (Figure 2.7), was designed to offer a more portable alternative to Device 1. Similar to Device 1, this set of equipment can be easily configured to interface with one or both hands, one or both feet, or any combination. Device 2 consists of two types of motorized interfaces: one for use with a single hand and one for use with a single foot.

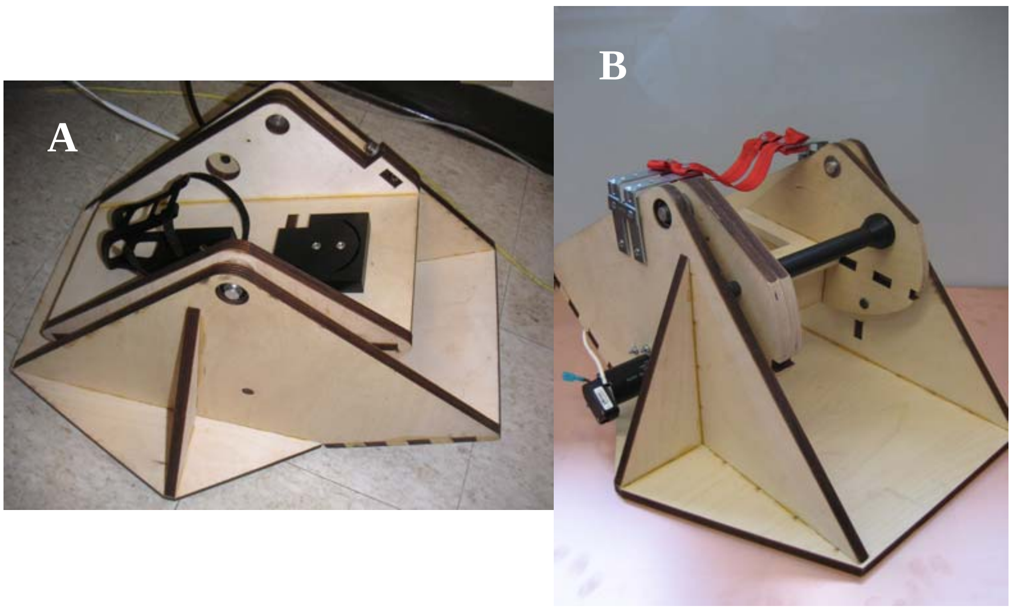

Figure 2.7 Device 2: Portable hand/foot cradles. The motorized interface for the foot (A, above) allows plantar flexion/dorsiflexion of the ankle, while the one for the hand (B, right) allows flexion/extension of the wrist.

The foot cradle is a motorized platform that is primarily constructed out of wood. Similar to the footplate of Device 1, this cradle allows plantar flexion and dorsiflexion with the cradle axis of rotation approximately aligned with the ankle axis of rotation. The subject's shod foot is secured in a toe clip fixed to the pivoting base (Figure 2.8A). The hand cradle operates very similarly to the foot cradle. With his forearm supported, the subject grasps a bar with his wrist approximately aligned with the cradle axis of rotation. This allows flexion and extension of the wrist (Figure 2.8B).

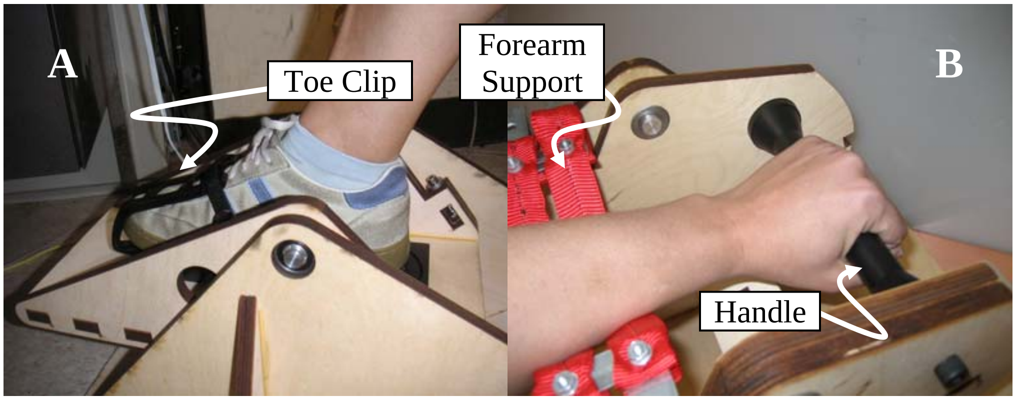

Figure 2.8 User interface of Device 2. The foot is secured to the device with straps and a toe clip (A). The upper limb interface requires the subject to voluntarily grasp a handle (B). The forearm is supported on soft straps to allow comfortable wrist flexion/extension.

The hand and foot cradles are independently driven by low inertia Maxon R40 DC brushed motors. These motors offer a maximum continuous torque of 0.18 Nm and a peak stall torque of 2.29 Nm. Using a cable driven transmission with a mechanical advantage of 7, we are able to achieve continuous torques about the ankle or wrist of 1.26 Nm and peak torques up to 15 Nm. On each interface, an optical quadrature encoder (US Digital, Model HEDS-9100; 1024 counts per revolution) mounted on the back of the motor provides measurements of the motor shaft angular position.

Each interface is powered and driven by an amplifier box designed in the Haptix Laboratory at the University of Michigan. A DC brushed servo amplifier (Copley Controls Corp, Model 4122D) provides controlled current proportional to an input voltage. This device uses the same computer hardware and software as Device 1 to handle real time control, data acquisition, and data processing. However, the data were collected at 200 Hz for the experiments that used this device.

The pendular nature of the hand and foot cradles gives them significant dynamics that are easily felt by a user. Both elements have a natural frequency very close to 1 Hz. The system identification of the open loop transfer function from commanded torque to output position for the foot cradle is presented in Figure 2.9. The low frequency gain (or the spring effects due to gravity) contributes the dominant dynamics felt by a human user. Therefore, gravity compensation was included for each interface. This made the device feel “light” to a user.

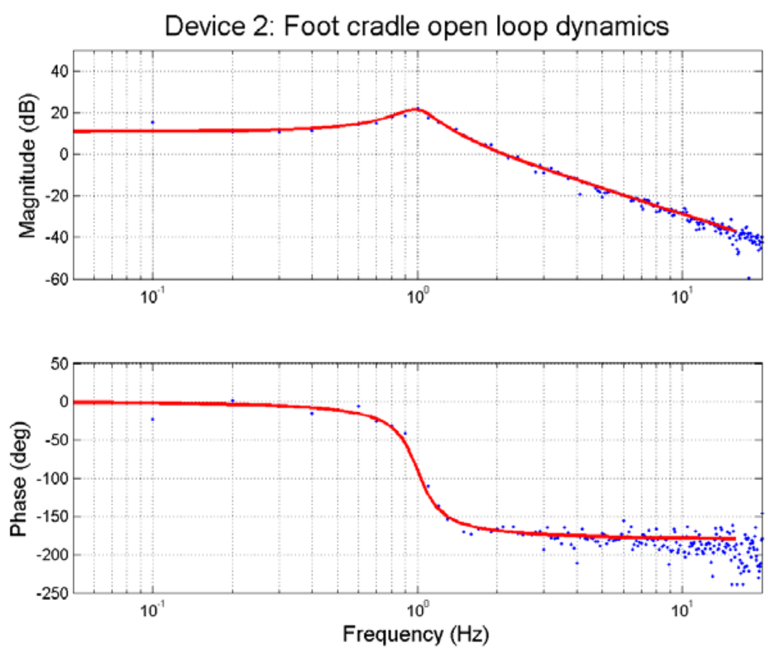

Figure 2.9 Device 2: Foot cradle open loop dynamics. The experimental transfer function (dots on the plot) can be modeled as a linear second order underdamped system (smooth line). Note the resonant peak around 1 Hz and the low frequency gain.

To provide the most flexibility in experiment design, we constructed two hand cradles and two foot cradles. The interfaces are not side specific meaning a hand or foot cradle can be used on a subject’s left or right side. Each of these can be used independently for interaction with a virtual environment or in combination to form a teleoperator. The experiment presented in Chapter 5 examined self-assist with a teleoperator utilizing one foot cradle and one hand cradle.

CHAPTER III

Tradeoffs of coordinated multi-limb motor control

We have suggested that self-assist offers a number of possible benefits to rehabilitation schemes including increased cognitive participation on the part of the subject. Active participation promotes motor learning [22, 23], while passivity may lead to loss of motivation and effort. The nature of self-assist requires subjects to remain cognitively active; however, it does so in a manner that simultaneously increases the cognitive load. In employing self-assist, subjects will be required to perform multi-limb control which may involve developing and executing two motor plans for two sets of muscle groups. The tradeoffs associated with multi-limb control are as yet unclear. On one hand subjects may have greater control and better sensory information from two effectors in performing a single task through self-assist. On the other hand, the increase in cognitive load associated with larger motor demands may be detrimental to performance. This chapter discusses two experiments with neurologically intact subjects involving dynamic object manipulation tasks through which we explore the tradeoffs associated with multi-limb control.

There are a number of mechanisms that influence physical activity and motor control during single- and multi-limb activation. Many of the observed effects relate to the activation of homologous muscle groups. For example, strength training in one limb will affect the homologous muscles in the untrained contralateral limb [58]. It has been found that in untrained individuals, the maximum contraction capabilities of a single limb are decreased during bilateral contractions as compared to unilateral contractions [59]. This bilateral deficit can be as large as 45% during rapid contractions [58]. With appropriate training, it is possible for the bilateral deficit to be eliminated, or even converted into a bilateral facilitation. This phenomenon, whether deficit or facilitation, appears to only be present during activation of homologous muscle groups and not observed in non-homologous muscles [60]. It is clear that these neural effects will influence bilateral self-assist, but it is not clear how these effects will influence generalized self-assist.

In the first chapter, we discussed the existence of bilateral and ipsilateral neural coupling. We proposed that neural coupling will act to strengthen the effectiveness of rehabilitative exercises; however, it is not obvious under what circumstances the benefits of neural coupling can be exploited. The rhythmic coordination studies by Kelso (for bilateral upper limb motions) and Baldissera (for ipsilateral upper/lower limb motions) primarily discuss performance limitations while two effectors are involved [32, 33]. These papers suggest that neural coupling is responsible for challenges that arise in performing tasks that are spatially or temporally dissimilar. People have great difficulty in coordinating rhythmic multi-limb motions that are not at the same or integer frequencies of each other. Interlimb coordination appears to present an even greater challenge following a stroke [61]. Compared to unilateral rhythmic movements, subjects demonstrated reduced movement amplitude and increased cycle duration during interlimb coordination. This effect was even more pronounced during non-homologous coordination than homologous coordination. These studies all involve simple cyclic motions and focus on low-level neural crosstalk which could present interference during multi-limb coordination [47]. Generalized self-assist requires multi-limb coordination of non-homologous muscle groups; this literature suggests possible detrimental effects during such motor coordination.

In executing even a single limb motor task, the central nervous system is responsible for generating and playing out an appropriate motor plan for the portion of the body involved. This action requires a certain level of attention. Self-assist requires subjects to maintain that level of attention while also generating and executing the appropriate motor plan for the assistance provided. The added responsibility surely increases the cognitive demand on the subject. It is important to determine if subjects can handle this increase in cognitive stress. We have suggested that self-assist allows the generation of a single motor plan for both the assistive and assisted elements. However, the motor plans might look very different for activation of non-homologous muscle groups, such as the upper and lower extremities.

The challenges that might be associated with generalized self-assist between the upper and lower limbs include interference from low-level neural crosstalk and increased high-level cognitive load in generating motor plans. However, there are a number of potential benefits associated with coordinated multi-limb control. Performing a single task with multiple effectors might offer subjects access to increased or improved sensory information, as from afferent channels. Developing and executing similar motor plans for each effector will also utilize high-level neural crosstalk (connections in the corpus collosum) to enable improved coordination [55]. In addition, it has been demonstrated that homologous and non-homologous multi-limb coordination during rhythmic motion with loads on the limbs can actually increase muscle recruitment [38].

This chapter discusses two experiments designed to assess the tradeoffs that are associated with multi-limb motor control. Both experiments include neurologically intact subjects presented with spatial and temporal challenges. Using tasks that require object manipulation of systems with underactuated dynamics, the experiments both compare conditions with input from the feet alone, the hand alone, and the feet and hand together. The first experiment examines end point tracking of a sprung virtual pendulum. The second experiment examines excitation of a virtual resonant system. Together, these experiments examine if the detrimental affects of increased cognitive load outweigh the benefits associated with multi-limb control.

3.1 *Experiment 1: Combined effort in pursuit tracking*

In this study we examined the ability of subjects to use their hands and feet together in controlling and manipulating a single dynamic object. Subjects were asked to control a virtual pendulum to track a reference signal. We compared the tracking performance during operating conditions that included using the hands alone, using the feet alone, and using both the hands and feet together. [62]

3.1.1 Methods

For this experiment, we used Device 1 configured with subjects standing (Figure 2.2). Through the footplate, subjects could interact haptically with the dynamic task. The electromechanical single hand interface was used to allow upper limb interactions.

For this experiment, it provided input from the user to the virtual environment but did not offer force feedback to the hand.

Three neurologically intact subjects (1 male, 2 female) between the ages of 22 and 24 participated in the study. All subjects were free of any upper or lower limb malformations or impairments and reported right hand dominance and normal or corrected-to-normal vision. Subjects were not compensated for their participation. All subjects provided informed consent in accordance with University of Michigan IRB policies.

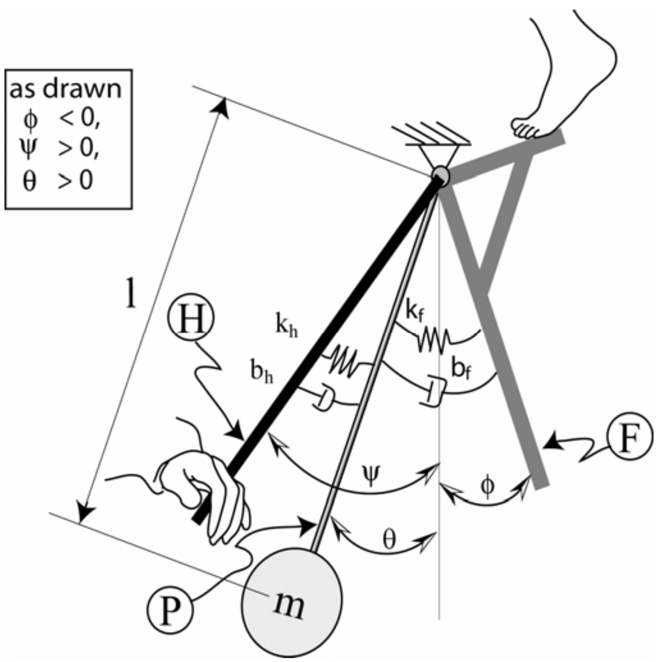

Figure 3.1 Physical representation of the virtual environment. The virtual couples, represented here as spring/damper pairs, connect the subject through the hand- and foot-plates to the virtual pendulum. Force inputs from the hand to the handplate (H) and feet to the footplate (F) are transmitted through these couplings to drive the pendulum (P).

Description of the virtual pendulum

Subjects were asked to track a pseudorandom signal with the endpoint position of underactuacted virtual pendulum. The physical representation of this configuration is illustrated in Figure 3.1. Virtual spring/damper pairs couple the pendulum to the footplate and/or the handplate. The pendulum *P* is modeled as a point mass *m* at the end of a massless bar of length *l* pivoted to ground through a horizontal axis. Let θ describe the angular displacement of the pendulum from the vertical; let ϕ describe the displacement of the footplate *F*; and let ψ describe the displacement of the handplate *H*. The stiffness and damping of the virtual coupler between the pendulum and the handplate are k_h *and* b_h, respectively. Similarly, the stiffness and damping of the virtual coupler between the pendulum and the footplate are k_f and b_f, respectively. The equations of motion for the pendulum and the reaction torque τ applied to the feet are:

$$\begin{aligned} ml^2\ddot{\theta} + (b_h + b_f)\dot{\theta} + (k_h + k_f)\theta + mgl\sin(\theta) \\ = b_h\dot{\psi} + k_h\psi + b_f\dot{\phi} + k_f\phi \end{aligned} \quad \textbf{(3.1)}$$

$$\tau = -b_f(\dot{\phi} - \dot{\theta}) - k_f(\phi - \theta) \quad \textbf{(3.2)}$$

Operating Conditions

The hardware configuration used in this experiment is capable of accepting input from either the feet or a hand and providing force feedback to the feet. The potential operating conditions include every combination of these options. Figure 3.2 illustrates with schematic switches how each input and the output can be turned on or off. By opening or closing the three switches, eight distinct operating configurations are possible.

Table 3.1 summarizes the eight operating conditions, where FC denotes foot control, HC denotes hand control, HFC denotes both hand and foot control, and Fdbk signifies haptic feedback at the feet.

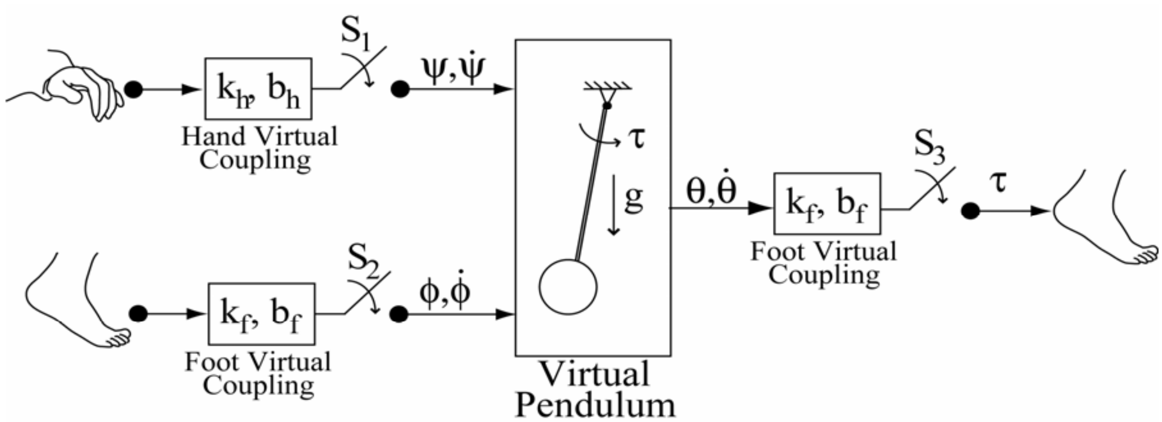

Figure 3.2 Schematic of input-output configurability. By changing the state of the three switches (S1, S2, and S3) we can select which effectors will control the virtual pendulum and the haptic feedback at the feet.

Table 3.1 Possible input-output operating conditions

Condition	S_1	S_2	S_3	Description
1	Off	Off	Off	Null Case
2	Off	Off	On	No Input
3	Off	On	Off	FC, No Fdbk
4	**Off**	**On**	**On**	**FC, Fdbk**
5	**On**	**Off**	**Off**	**HC, No Fdbk**
6	**On**	**Off**	**On**	**HC, Fdbk**
7	On	On	Off	HFC, No Fdbk
8	**On**	**On**	**On**	**HFC, Fdbk**

*Note: Conditions 4, 5, 6, and 8 are tested in this experiment

The first two configurations described in this table have no practical significance for manipulating a virtual pendulum since without input, the pendulum will simply remain stationary. The other six configurations represent control by the hand only, the foot only, or by both. Note that the various configurations can also be realized directly from the equations of motion by adjusting the stiffness and damping coefficients. For instance, by setting k_h and b_h equal to zero, we essentially turn off the virtual coupling between the hand and the pendulum, allowing no transfer of power between the two. In this experiment, we used four of the eight possible operating condition; the selected conditions correspond to conditions 4, 5, 6, and 8 in

Table **3.1**, and correspond to control through the feet alone with force feedback, combined control through the feet and hand with force feedback at the feet, control through the hand alone without force feedback to the feet, and control by the hand with force feedback to the feet.

Table 3.2 Parameters of the virtual pendulum system

Pendulum Mass	*M*	*35 kg*
Pendulum Length	*L*	*0.9 m*
Virtual Coupling Stiffness	k_h, k_f	*900 Nm/rad*
Virtual Coupling Damping	b_h, b_f	*100 Nms/rad*

Protocol

The parameters for the virtual pendulum remained unchanged throughout the experiment. Table 3.2 summarizes the physical equivalent of these parameters. Even though the parameter values remained unchanged, the system dynamics are necessarily affected when certain elements are removed from the system, such as when one effector controls the pendulum as opposed to two. The natural frequencies of the free virtual pendulum ω_p, of the virtual pendulum coupled to one effector ω_1, and the virtual pendulum coupled to two effectors ω_2 are:

$$\begin{aligned} \omega_p &= \sqrt{g/l} = 3.3rad/s \\ \omega_1 &= \sqrt{(k+mgl)/(ml^2)} = 6.5rad/s \\ \omega_2 &= \sqrt{(2k+mgl)/(ml^2)} = 8.6rad/s. \end{aligned} \tag{3.3}$$

Subjects were asked to manipulate the pendulum endpoint to track a pseudo-random signal comprised of the sum of seven sine waves with unevenly distributed (without harmonic inter-relationships) frequencies and unevenly distributed phase shifts. The maximum frequency was set to less than half the natural frequency of the virtual pendulum to enable relatively simple tracking despite the presence of the resonant frequency in the pendulum. The target tracking signal was presented on an oscilloscope-type display on a computer monitor. Subjects received visual feedback of the current pendulum position overlaid on the reference signal. The display showed the time history of both the pendulum position and the reference signal but provided no preview of the reference. In the applicable conditions, subjects also received haptic feedback at the feet and proprioceptive feedback of the hand and feet positions.

Each subject was given two minutes of unrecorded practice time per condition. After the practice period, three-minute trials were run where each trial tested only a single operating condition. Three replicates were recorded for a total of twelve three-minute trials per subject. The twelve trials were administered in a random order. Subjects were instructed that they could pause or terminate the test at any time if they felt discomfort or fatigue.

3.1.2 Data analysis

Performance in the pursuit tracking was assessed by comparing the actual position of the pendulum endpoint to the reference signal presented to the subjects. A sample portion of the pendulum position and the reference signal for one subject operating in two of the four conditions tested is presented in Figure 3.3. By qualitative inspection of the time history traces, visual distinctions are readily apparent in performance between the operating conditions.

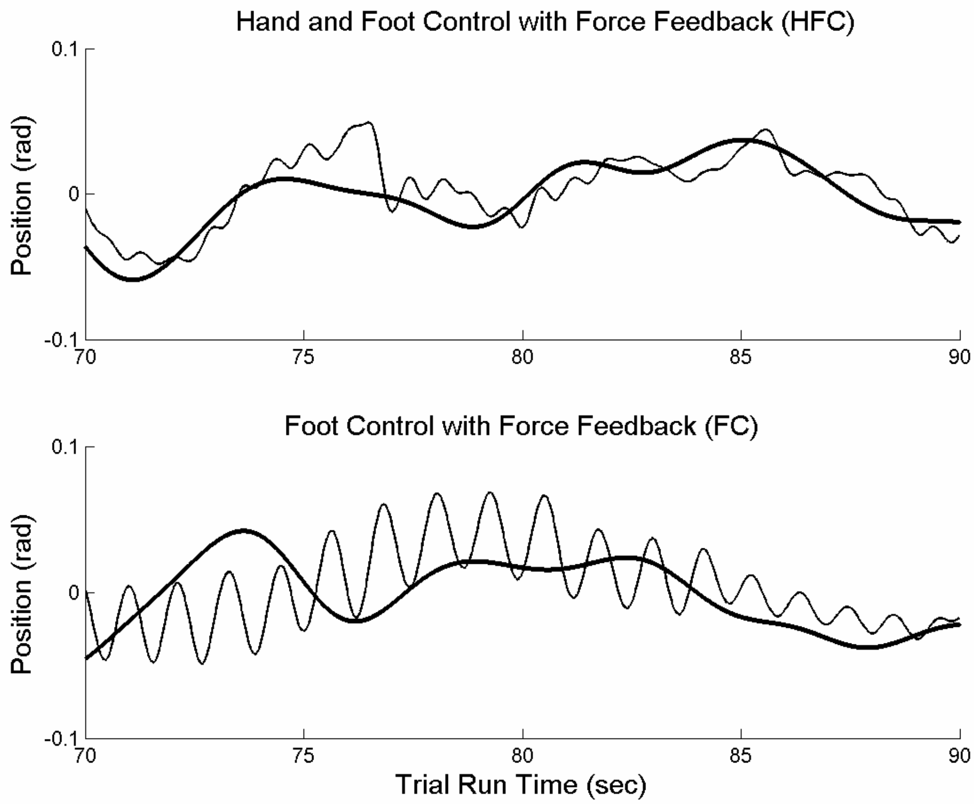

Figure 3.3 Sample portion of reference signal and endpoint position. The two plots show a middle 20 second interval (time 70-90 sec) of two separate trials for a single subject in the combined hand and feet control (HFC, top) and the feet alone control (FC, bottom). The pendulum endpoint position, presented as the thin trace, follows the reference signal, presented as the heavy dark line.

The total root mean squared (RMS) error between the actual and reference positions was used as the performance metric to quantitatively compare conditions. A smaller RMS error would indicate better performance. We calculated this performance variable for each three minute trial. Statistical analyses were performed in SPSS (Chicago, IL) using a repeated measures linear mixed model with condition as a fixed effect and subject as a random effect. The threshold for statistical significance was set at

$\alpha = 0.05$. Significance was determined for both subject and condition. The data was further processed by normalizing the RMS errors for each subject by the subject mean performance across all conditions. Pairwise comparisons were then made for each of the conditions. The Bonferroni method of adjustment for multiple comparisons was used for correcting significance levels in the pairwise comparisons.

3.1.3 Results

Subjects performed significantly better when hand control was involved. The mean normalized RMS error across all subjects for the four conditions is presented in Figure 3.4. Control by the feet alone (FC) demonstrated significantly worse performance than the other three conditions ($p < 0.01$ for each of the three paired comparison). The general trend shows that the condition with combined hand and feet control (HFC) achieved the best performance results. This condition was not significantly different than the other two conditions involving the hand (HC). The combined hand and feet control demonstrated a 23% reduction in RMS error over the feet alone condition ($p = 0.007$).

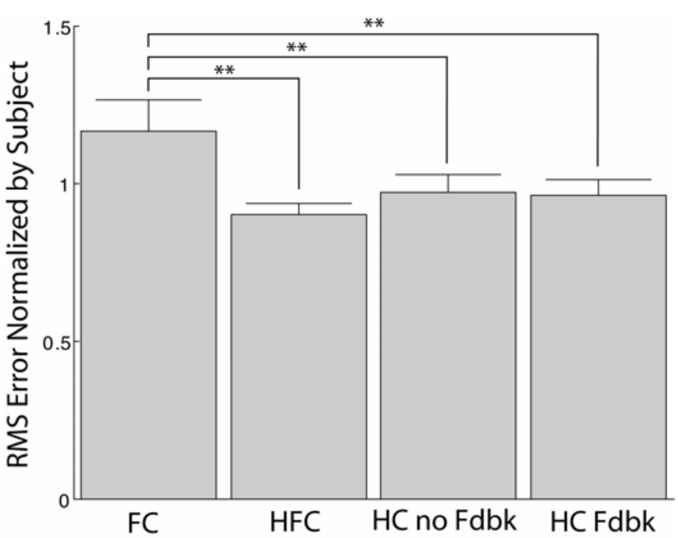

Figure 3.4 Mean normalized RMS error. In comparing the mean RMS error, normalized by subject, pendulum control by the feet alone (FC) demonstrates the worst performance while control by the combined efforts of the hand and feet (HFC) demonstrates the best performance. The ** denotes significant differences in mean ($p < 0.01$). The error bars denote 95% confidence interval of the mean.

3.1.4 Discussion and significance

The results of this experiment suggest that for manipulation of a sprung mass in this tracking task, the control efforts of the hand and feet may be combined to achieve superior performance than control effort applied by the feet alone. The combined hand and feet control outperformed of the two hand alone-methods, but only slightly and not significantly so. It is important to recognize that the addition of the feet to hand control did not degrade performance. Subjects are capable of allotting appropriate cognitive capacity to operate both the upper and lower limbs in the same or a similar position control task to achieve a single goal.

We examined three conditions that included hand control, but ultimately our objective is to comment on methods for administering lower limb rehabilitation. Therefore, the most interesting comparison is between the two cases that include feet control. Input from the hand was not designed to dominate the input from the feet in controlling the pendulum; both effectors influenced the virtual pendulum equally. The appearance of improved performance when both inputs were used suggests that the two limbs were working together.

Even in neurologically intact subjects, the upper extremity may offer additional benefits in training the lower extremity. The hands my have increased dexterity which can supplement that of the feet to achieve better performance in a tracking task. Two subjects independently reported that the combined control condition felt easier; that their strategy evolved to use the hand for broad motions and the feet to damp the system oscillations. This study supports the notion of using the upper limbs to assist the lower limbs in completing a difficult task for the purpose of improving the duration and quality of the practice that the lower limbs receive.

3.2 Experiment 2: Combined effort in resonance excitation

Having examined control of underactuated states for endpoint tracking in Experiment 1, we turn our attention to examine excitation of underactuated states in a simple resonant system. In this experiment we present subjects with a virtual spring-inertia-damper and ask the subjects to excite maximum oscillations. This resonant task is designed to assess the ability of subjects to identify system characteristics and execute appropriate timing. Success at this task requires both spatial and temporal movement considerations on the part of the subject. Previous work has demonstrated that people are capable of exciting systems with unknown resonance using supination/pronation of the hand, given haptic and/or visual feedback of the system [63]. Here, we examine this task using flexion/extension of the wrist and ankle. Once again, we compare the differences in control and performance with the upper limbs, the lower limbs, and the combination.

3.2.1 Methods

In this experiment, we examined each subject's ability to identify the natural frequency of a resonant system and subsequently maintain maximum oscillations of that system. Using Device 2 described in Chapter 2 (Figure 2.7), a subject interacted with the virtual environments through his dominant hand, his dominant foot, or his dominant hand and foot simultaneously. Four neurologically intact subjects (2 male, 2 female) between the ages of 27 and 31 participated in the study. All subjects were free of any upper or lower limb malformations or impairments and reported right hand/foot dominance and normal or corrected-to-normal vision. Subjects were not compensated for their participation. All subjects provided informed consent in accordance with Northwestern and the Rehabilitation Institute of Chicago IRB policies.

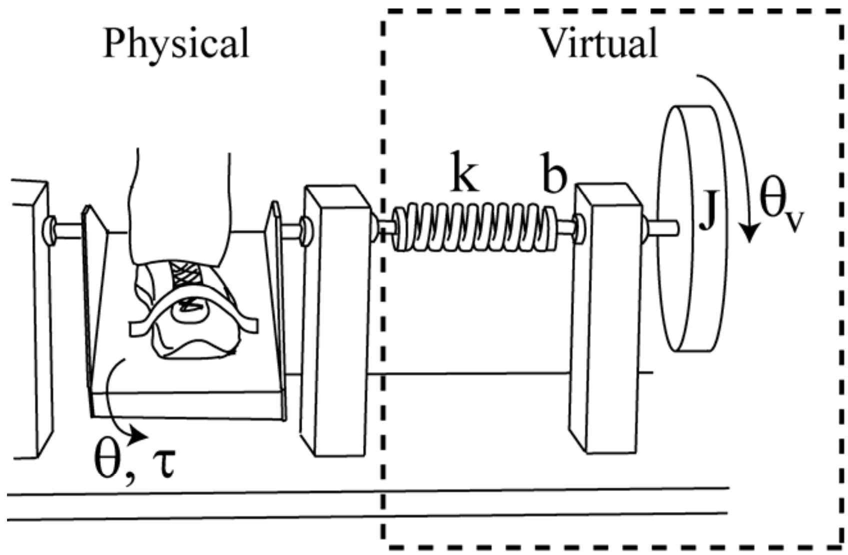

Figure 3.5 Physical representation of the resonant system. For a single foot input, shown here, the foot cradle position (θ) drives the virtual resonant system to position θv. The virtual system is comprised of a torsional spring (with stiffness k), a rotary inertia (with moment of inertia J), and damping to ground (with damping b). The subject feels a reaction torque from the system at his foot (τ).

Description of the resonant system

As with Experiment 1, this study used an underactuated spring-mass system. Unlike the virtual pendulum, the spring-mass system implemented in this study is not influenced by the effects of gravity. The virtual system consists of a rotational inertia (with moment of inertia J) coupled to the user interface (hand or foot cradle) through a torsional spring (with stiffness k) and damping to ground (with damping b). Figure 3.5 presents an illustration of the physical representation of the virtual system interfaced with a subject's foot. The angular displacement of the user input, or the device cradle (θ in the generic single input case; θ_h for the hand; θ_f for the foot), and the angular displacement of the virtual inertia output (θ_v) can be related by the equations

$$\ddot{\theta}_v + 2\zeta\omega_n\dot{\theta}_v + \omega_n^2\theta_v = \omega_n^2\theta_v \quad \textbf{(3.4)}$$

where $\omega_n = \sqrt{\frac{k}{J}}$ and $\zeta = \frac{b}{2\sqrt{kJ}}$

The output torque felt by the user through the cradle is equivalent to the spring torque and is defined as

$$\tau = -k(\theta_v - \theta) \quad \textbf{(3.5)}$$

During this experiment, each subject manipulated the virtual spring-mass in three possible configurations: hand, foot, and combined. The three configurations are presented in Figure 3.6; however, for visual simplicity, the rotary system has been transformed into a linear system. For the single effector input (Figure 3.6A and Figure 3.6B), the equations of motion and output torque are given by equations 3.4 and 3.5, respectively.

The virtual system changes slightly when we consider two input effectors. Essentially, the virtual inertia manipulated by the hand is rigidly fixed to the virtual inertia manipulated by the foot. The hand and foot are free to move independently with different angular displacements (Figure 3.6C). The equations of motion and output torque for this joint system become

$$(J_h + J_f)\ddot{\theta}_v + (b_h + b_f)\dot{\theta}_v + (k_h + k_f)\theta_v = k_h\theta_h + k_f\theta_f \qquad \textbf{(3.6)}$$

$$\begin{aligned}\tau_h &= -k_h(\theta_v - \theta_h)\\ \tau_f &= -k_f(\theta_v - \theta_f)\end{aligned}$$

These dynamics no longer represent a second order underactuated system. They are more complicated. However, if the user manipulated both inputs such that they hand and foot move synchronously (Figure 3.6D), the dynamics simplify considerably. When θ_h equals θ_f, the equations of motion reduce to

$$\ddot{\theta}_v + 2\zeta\omega_n\dot{\theta}_v + \omega_n^2\theta_v = \omega_n^2\theta_v \qquad \textbf{(3.6)}$$

where $$\omega_n = \sqrt{\frac{k_h + k_f}{J_h + J_f}}$$

Provided that the natural frequencies of the uncoupled systems are equal (ω_n), the natural frequency of this reduced system will also be ω_n. The motion of the coupled virtual inertia driven by two synchronized effectors is equivalent to the system described by equation 3.1. Therefore, the coupled dynamic task will be simpler and easier for the subject if he can maintain synchronous motion of both effectors.

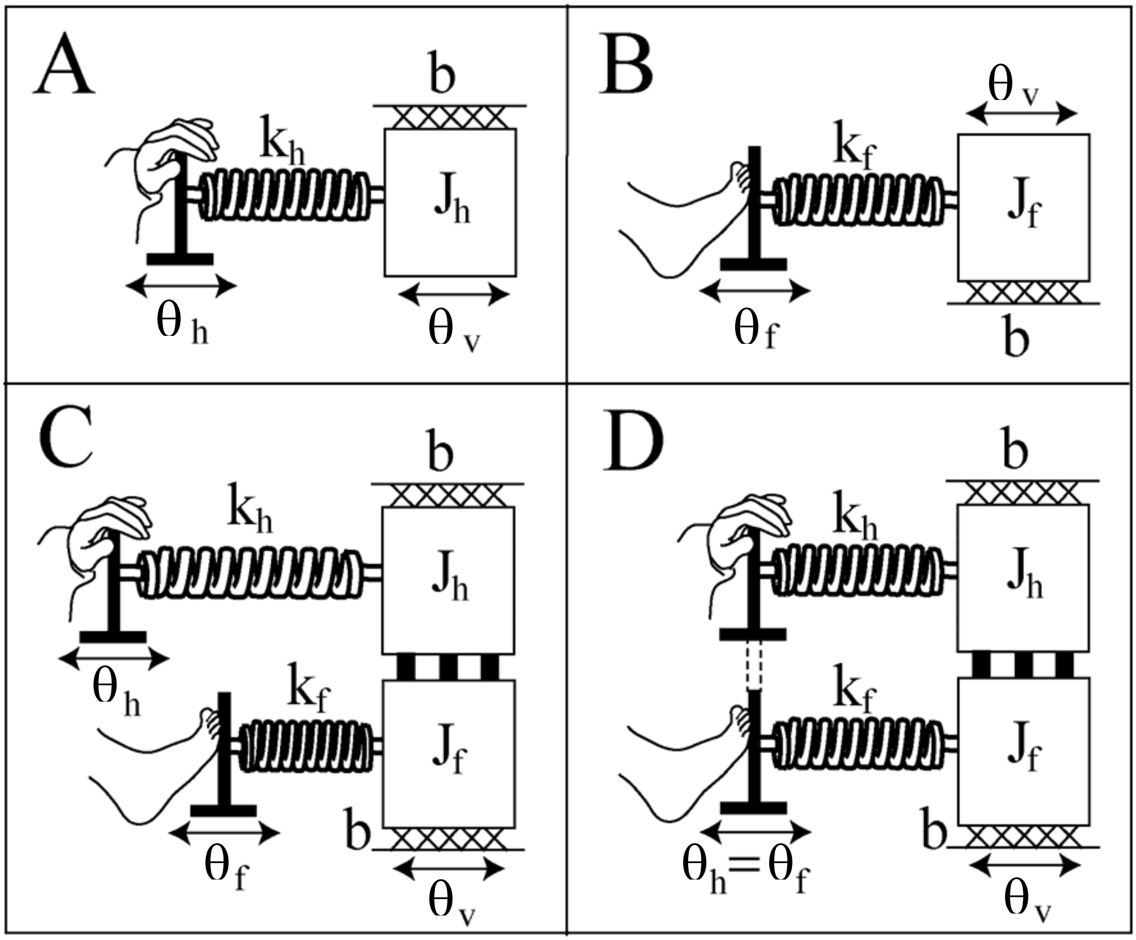

Figure 3.6 Configurations for hand and foot interfaces with resonant system. A single input from the hand (A) or foot (B) drives the virtual inertia in some of the experiment trials. For the combined hand and foot control (C), the two inertias are rigidly connected but the input effectors can still move independently. The system is greatly simplified if the user moves the hand and foot synchronously (D) such that $\theta_h = \theta_f$.

Subjects were presented with several spring-mass virtual systems having different natural frequencies. Certain parameters remained unchanged while others varied between the systems. Table 3.3 summaries the physical equivalents of the parameters used in this experiment. The different systems necessarily feel different to subjects in terms of reaction forces felt and response timing of the dynamics. We chose to maintain constant spring stiffness and damping ratio while varying the inertia to achieve the desired natural frequency. Alternatively, we could have opted to hold inertia constant and adjust stiffness to achieve the various natural frequencies. By maintaining constant spring stiffness, we generate a set of systems that all present the same magnitude gain from position to force at the resonant frequency.

Table 3.3 Parameters of the virtual resonant system

Damping Ratio	ζ	*0.1*
Hand Spring Stiffness	k_h	*0.15 Nm/rad*
Foot Spring Stiffness	k_f	*0.2 Nm/rad*
Natural Frequencies	ω_n	*6, 7, 8, 9, 10 rad/s*

Protocol

Subjects were asked to excite and maintain maximum amplitudes of the virtual spring-mass system under three conditions: with hand control through wrist flexion/extension, with foot control through plantar flexion/dorsiflexion, or with both hand and foot control. Haptic feedback of the spring torque was provided for the hand and foot when applicable. In addition, subjects were given real-time visual feedback in the form of an abstract performance gauge. The needle on the gauge moved in proportion to the kinetic and elastic potential energy in the virtual system, defined by

$$E = \frac{1}{2}J\dot{\theta}^2 + \frac{1}{2}k(\theta - \theta_v)^2 \tag{3.7}$$

As subjects increased the oscillation amplitude of the virtual inertia, the energy of the system increased moving the needle higher.

Prior to the initiation of the evaluated and recorded trials, subjects completed a period of familiarization and training with the device. They practiced with resonant systems different than those presented during the experiment. Subjects were required to demonstrate a minimum level capability before proceeding with the experiment. To demonstrate minimum capabilities, subjects were required to excite and maintain oscillations of three resonant systems similar to those presented during the experiment. The natural frequencies of the sample systems (ω_n = 6.5, 7.5, and 9.5 rad/s) were unique from those presented during the experiment.

During the data collection, subjects were presented resonant systems with five different natural frequencies (ω_n = 6, 7, 8, 9, and 10 rad/s). Subjects were asked to excite each of these systems in the three operating conditions: hand, foot, and both. Each system and condition combination was presented with four repetitions. Thus, the experiment consisted of sixty 30 s trials (5 frequencies x 3 conditions x 4 repetitions = 60 trials). The frequencies were presented in randomized order.

3.2.2 Data analysis

Trial performance is based on the subjects' ability to excite maximum oscillations of the virtual inertia. This performance varied greatly trial by trial across all subjects, input conditions, and natural frequencies. Figure 3.7 presents input-output sample data from a few trials. These plots illustrate the dissimilarities between trials in finding and maintaining excitation at the appropriate natural frequency.

The use of two independently actuated electromechanical devices allows us to determine the input from each effector, even when they are acting on the same virtual system. We are able to compare the performance of the foot when driving the resonant system independently to the contributions of the foot when driving the resonant system in combination with the hand. Likewise, we can compare the performance of the hand when operating alone to the performance of the hand when operating in combination with the foot. Our objectives do no include deciphering if the hand or the foot is better at this particular task, but rather how the capabilities of either are changed when the other is also involved.

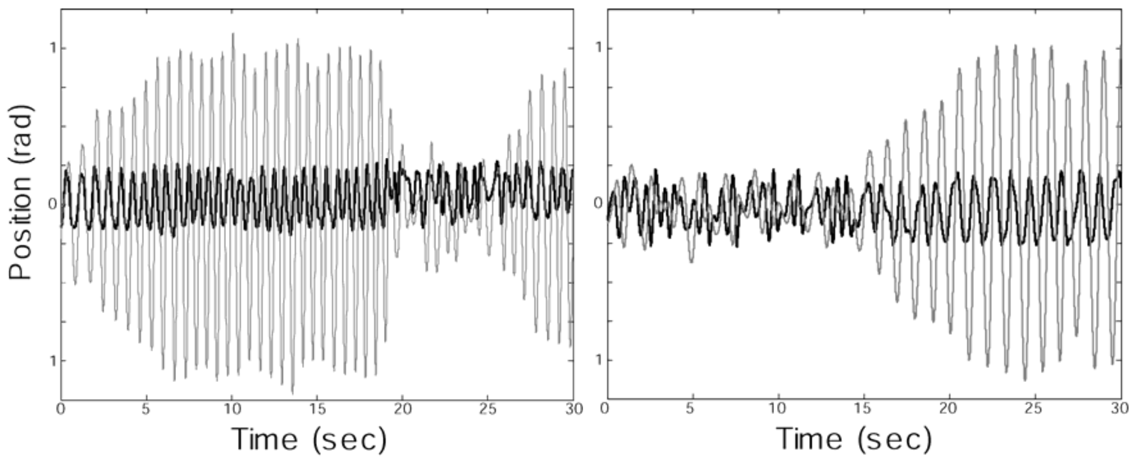

Figure 3.7 Exp 2 input-output position sample traces. The input cradle position (heavy black line) drives the output virtual inertia position (light grey line). These sample plots demonstrate performance differences between trials. In the trial presented on the left, the subject quickly found the appropriate excitation frequency but could not maintain it for the entire trial; he lost the large oscillations around second 20. In the sample presented on the right, the subject struggled to select an appropriate input frequency until about second 15.

The performance metric used for this experiment is the amount of work done on the virtual system by the subject. The total work is indicative of how successful the subject was at pumping energy into the resonant system, thus exciting oscillations as per the experiment instructions. The total work for a single trial was calculated as the integral of power (rate of work) over the course of the trial. The work rate is defined as the force exerted by the subject multiplied by the velocity of the input motion. Larger positive power indicates more energy into the system and increases the total work done on the subject by the system.

The data were independently analyzed for the hand and foot performances, where performance is the total work per trial. We initially performed an analysis of variance (ANOVA with repeated measures). The statistical model included the main effects of condition (independent or combined), natural frequency (ω_n = 6, 7, 8, 9, or 10 rad/s), and subject. The model also included the interaction effects of condition by frequency. Post-hoc paired t-tests were performed to determine the condition effect while controlling for subject differences. The significance level α was set at 0.05.

3.2.3 Results

Combined hand and foot control significantly increased the amount of work performed by the foot while significantly reducing the positive work performed by the hand as compared to the respective single limb alone conditions. The initial repeated measures ANOVA with work done by the foot as the performance variable yielded subject ($p < 0.001$) and condition ($p = 0.011$) as the significant input factors. The significant input factors for work done by the hand as the performance variable included all main effects: subject, condition, and frequency ($p < 0.001$). Table 3.4 summarizes the ANOVA significance results.

Table 3.4 Summary of significance from repeated measures ANOVA

Factor significance	Subject	Condition	Frequency	Cond*Freq
Work done by foot	< 0.0001*	0.0114*	0.0546	0.2071
Work done by hand	< 0.0001*	< 0.0001*	0.0001*	0.9063

* denotes significance

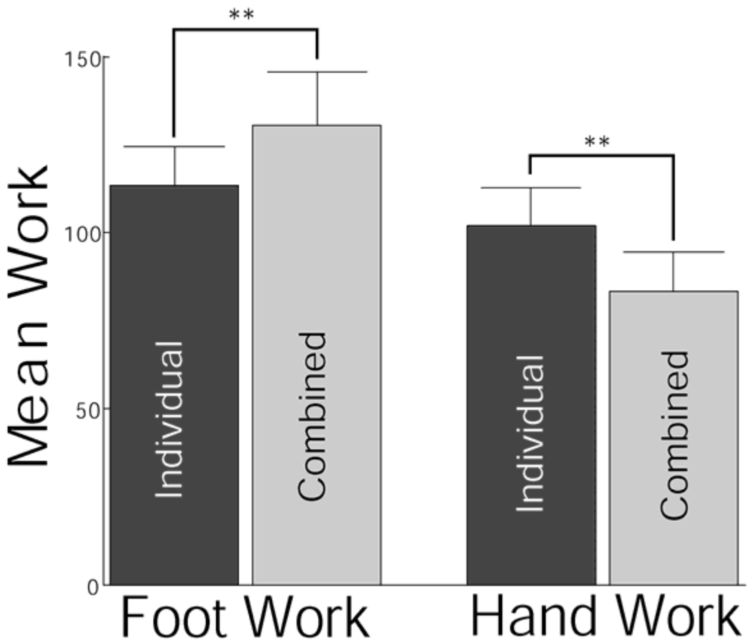

Figure 3.8 Average of total work done on the virtual system. The total work done by the foot was 15% larger when the hand was also contributing to the resonant task (Combined – dark grey) as compared to when the foot performed the task alone (Individual – light grey). However the addition of foot control was detrimental to the hand performance, reducing the total work done by the hand by 18%. The error bars denote 95% confidence intervals of the mean and the (**) denote significant difference between the means ($p < 0.01$).

The paired t-test demonstrated significant differences in the amount of work performed for both the hand and the foot in the combined condition when compared to the alone conditions. The mean for each subject and frequency pairing was computed over the trial replicates for the analysis. The mean total work per trial by the hand and the foot in both the individual and combined conditions, averaged over all subjects and all frequencies, are presented in Figure 3.8. The control condition, whether independent or combined, influenced the amount of work subjects were able to do on the virtual system by each effector. However, the hand and foot were affected differently by the addition of the other. Combined control acted to increase the effectiveness of the foot input; total work performed by the foot during the combined control condition was 15.1% higher than during the independent foot control. The hand performance degraded with the introduction of simultaneous foot control. During the combined control condition, the work performed by the hand was an average of 18% less than when the hand acted alone.

3.2.4 Discussion

Combined control of the resonant system improved the foot performance while degrading the hand performance. Despite identical visual feedback and the spatial and temporal similarity of the task, the challenges presented to each effector were quite different biomechanically. Different muscle groups comprise the control for the wrist and ankle having different strength, speed, and dexterity. Using total work done on the virtual system, it is not possible to directly compare the hand and foot performances to each other. It appears that the hand may have been more adept in this particular rhythmic task than the foot. Thus, when the two effectors operated in unison to complete the presented task, the level of performance of each seemed to meet in the middle.

Exciting a resonant system in the fashion presented is a relatively obscure objective for people, especially using their ankle. It turns out that the task was quite difficult for subjects to understand and become comfortable with, as reported qualitatively by the subjects. In designing the experiment, our goal was to present a challenge for neurologically intact subjects. This rhythmic task was selected because people are familiar with the concept of resonance without being well practiced at it. Repetitive, rhythmic tasks are often chosen in multi-limb experiments because they, on a very superficial level, bear resemblance to locomotion [64]. In daily life, the ankle has little exposure to smooth, continuous rhythmic motion; even in locomotion, the joint primarily turns on and off during pushoff and swing, respectively. The upper limbs are more practiced at dealing with resonant systems. For example, dribbling a basketball requires an understanding of the ball dynamics and formulation of appropriate input motion. Another easily understood activity involving a system with an associated natural frequency is pushing a child on a swing. Although the interface between the child and the pusher is through the upper extremity, the whole body is frequently involved in the timing and excitation. Ultimately, it was observed that this task may have been too difficult for subjects even though the population tested demonstrated some success at accomplishing it. Because it was not based on a functional equivalent, it was not intuitive to understand.

The virtual systems presented to the hand and foot were the same in terms of natural frequencies, but were scaled to accommodate the strength differences of the two effectors. Even with the force scaling, the haptic feedback may not have been equivalent in relative magnitude for the hand and foot. The loads presented to the foot were likely undersized compared to the capabilities of the ankle. Therefore, the hand may have had access to more information about the resonant task than the foot. Just as we would expect in using multi-limb control with a neurologically impaired population, one of the limbs would have an advantage in terms of superior motor control and/or sensory information over the other. This discrepancy in the quality of afferent information may have contributed to the noted performance differences between the hand and the foot in the individual and combined conditions. During identification of and motor adaptation to the unknown resonant system, the central nervous system (CNS) uses sensory feedback to develop a model of the dynamics. During combined control, the CNS has access to the visual feedback as well as the haptic feedback from both the upper and lower limbs. A mental model of the system would be developed with afferent information from the hand that could be used to update the motor plan for both the upper and lower limbs. Essentially, in the combined condition, the lower extremity motor control can be formulated by utilizing the superior upper extremity sensory information. The discrepancy in sensory information that exists between the two limbs may be critical to the facilitation observed in the less capable limb.

The cognitive load increase associated with controlling both the upper and lower limbs may contribute to the detrimental effects witnessed in upper limb performance. The CNS is not able to devote as much attention to the control of each effector when both are involved. However, the improvement in lower limb performance might indicate that the benefits gained by the hand sensory feedback and dexterity outweigh the cost of higher cognitive load. Self-assist may prove beneficial when applied to lower extremity motor learning in rhythmic tasks; however, the upper extremities are unlikely to benefit from lower extremity involvement in the same rhythmic task. Self-assist seems to offer the most reward to the extremity that has some sort of disadvantage compared to the other. Therefore, in a rehabilitation setting, self-assist may prove beneficial in the motor recovery of an impaired limb, provided that the assisting limb offers advantages in those abilities that the assisted limb is lacking.

3.3 Comparison discussion and significance

Coordinated multi-limb motions are necessarily different than simple single joint motions. They may be easier or harder, requiring different attention levels, different amounts of time for motor plan development, different adaptation responses. The differences could result from increased cognitive load, neural coupling, more sensory information (additional afferent channels), larger efferent signals out (bigger could be better), or some other unexplained phenomenon. The two experiments presented in this chapter demonstrated obvious changes in the motor task when it moved from single limb to multi-limb motor control. Subject strategy appeared to adjust, accommodating the altered cognitive and motor demands that accompanied control by two input effectors. The resulting performance also noticeably changed. In both experiments, endpoint tracking and resonant excitation, performance was improved when both the hand and foot or feet were involved in control as compared to the foot or feet alone. There is an increased cognitive load on the subject when he must control multiple limbs. However, the lower limb improved performances during combined control demonstrate that the benefits of multi-limb control outweigh the detriments associated with greater cognitive load.

We recognize that the tasks, meaning the dynamics of the virtual systems were altered by adding a second effector to the control. This may account for some of the performance differences that we observed in the experiments. These dynamic changes are an unavoidable consequence of combined control. We designed the virtual environments in both experiments to maintain as much consistency as possible between the independent and combined control so as to not favor either scenario over the other. In rehabilitation practices, the strengthening or training exercises performed with physical therapist assistance will present different dynamics than independent execution of a similar functional task. The primary objective is recovery of neuromuscular capabilities through an effective means, even if in an altered form of normal function.

In Experiment 1, we saw improved tracking performance of the combined upper and lower limb control over lower limb alone. The contributions of the feet could not be separated from those of the hand in influencing the endpoint control of the pendulum. The combined control may have offered improved performance because of the increased authority offered to the lower limb by the upper limb. The combined condition also tended towards better performance than the two independent hand controlled cases. For this task, unlike the resonance task in Experiment 2, we do not see a degradation of upper limb performance when the feet were added. The additional cognitive demand of lower limb motor control (as in the hand and feet controlled case, HFC) or of processing added lower limb afferent sensory feedback (as in the hand control with lower limb feedback, HC Fdbk) was not detrimental to the upper limb performance. On the contrary, the combined control of the resonant system reduced the hand performance. This contradiction may be related to the nature of the two tasks. Event based motions are easier to coordinate than rhythmic ones. Even though pursuit tracking is a continuous task, this type of position control can be thought of as a series of distinct path movements. Because event based motions are naturally easier to coordinate, increase in cognitive load associated with this multi-limb motion may not have been as great and thus did not affect the independent hand motor control as heavily. Rehabilitative exercises often include elements from both rhythmic and even-based tasks. The combination of these two experiments addresses the breadth of the types of exercises expected during therapy.

In these two experiments, we show that collaborative control between two effectors may offer advantages to the capabilities of one or the other. The weaker or less adept effector will reap the rewards of coordinated control, while the other may actually be compromised. There does appear to be an additional burden associated with multi-limb motor control. Tasks that are easy to coordinate, such as spatially and temporally similar event based motions, appear to reduce the effects of increased cognitive load.

Finally, multi-limb coordination appears to be worthwhile when one limb can offer something that the other is lacking, such as dexterity, strength, or better feedback. We demonstrated that the tradeoffs associated with multi-limb control, including detrimental effects to the more adept effector (the hand), favor improved performance for the lower limbs. The cognitive burdens that may accompany generalized self-assist do not present an inhibitive obstacle for its implementation. Further exploration in this topic might include studying the effects of coordinated motor control after long term training. The neural mechanisms responsible for muscular bilateral deficit could be trained into facilitation. After great practice, we might be able to further exploit multi-limb motion to benefit individual limb capabilities.

CHAPTER IV

Cuing and efference copy in disturbance rejection

We have suggested that self-assisted rehabilitation through patient-directed telerobotics offers a number of advantages over traditional therapy. The possible benefits include increased patient cognitive involvement, reduced subject passivity, better coordination between the efforts of the subject and the assistance, and decreased physical demand on the therapists. This chapter endeavors to explain and gather support for the motor control mechanism that supports one of these benefits: improved coordination of efforts between the patient and the assistance.

During an assisted motor task, the actual motion of a particular limb is the result of efforts from both the patient and the outside agent, such as a therapist or autonomous robot. The block diagram presented in Figure 4.1A represents a simplistic view of how the efforts of the patient and outside assistive effort, as from a therapist, sum to influence the resultant motion of the limb. If a patient is to coordinate his muscle activity with this assistance, he must possess an appropriate feedforward motor plan that incorporates an expectation of the assistive force. In the case when the assistive force is generated by an external agent (Figure 4.1B), the expectation of applied force is learned in a gradual manner and updated based on the successes and failures in previous exposures. This expectation must consider both the intent of the outside agent directing the assistive effort and the dynamics of the actual force application. This can be a difficult system to model since updates may be slow and the intentions of the therapist may be time varying. When the assistance is patient-generated, the assistive force prediction is developed in a distinctly different manner (Figure 4.1C). Essentially, efference copy of the intended assistance replaces the slowly formed and stored model of therapist intention. Subjects are now left to model only the mechanism by which their intent is transmitted. For the case of teleoperated self-assist, this would be the dynamics of the patient-controlled telerobot.

The role of efference copy in coordinating personal efforts with outside forces to control limb motion is extensively examined in the literature on anticipatory adjustments. We therefore appeal to this literature to help explain how self-assist might offer coordination benefits over therapist- or computer-assist. Although this literature does not exactly fit the self-assist scenario that we have discussed, these experiments generally examine tasks in which subjects counteract perturbations or reject disturbances rather than incorporate assistance. With some reinterpretation, we can alternatively view assistive forces as perturbations to the motor actions of the patient. In fact, examination of the motor control block diagram (Figure 4.1) reveals that the outside assistance enters the system as an exogenous signal exactly as a disturbance would. We see that the feedforward motion plan of a subject must counteract the effects of perturbations, whether assistive or detrimental, while still producing the desired motion. Therefore, results from the literature that explain the role of efference copy in the development of anticipatory adjustments for disturbance rejection tasks can be explored as a parallel mechanism to explain the role of self-assistance in coordination of assistive forces.

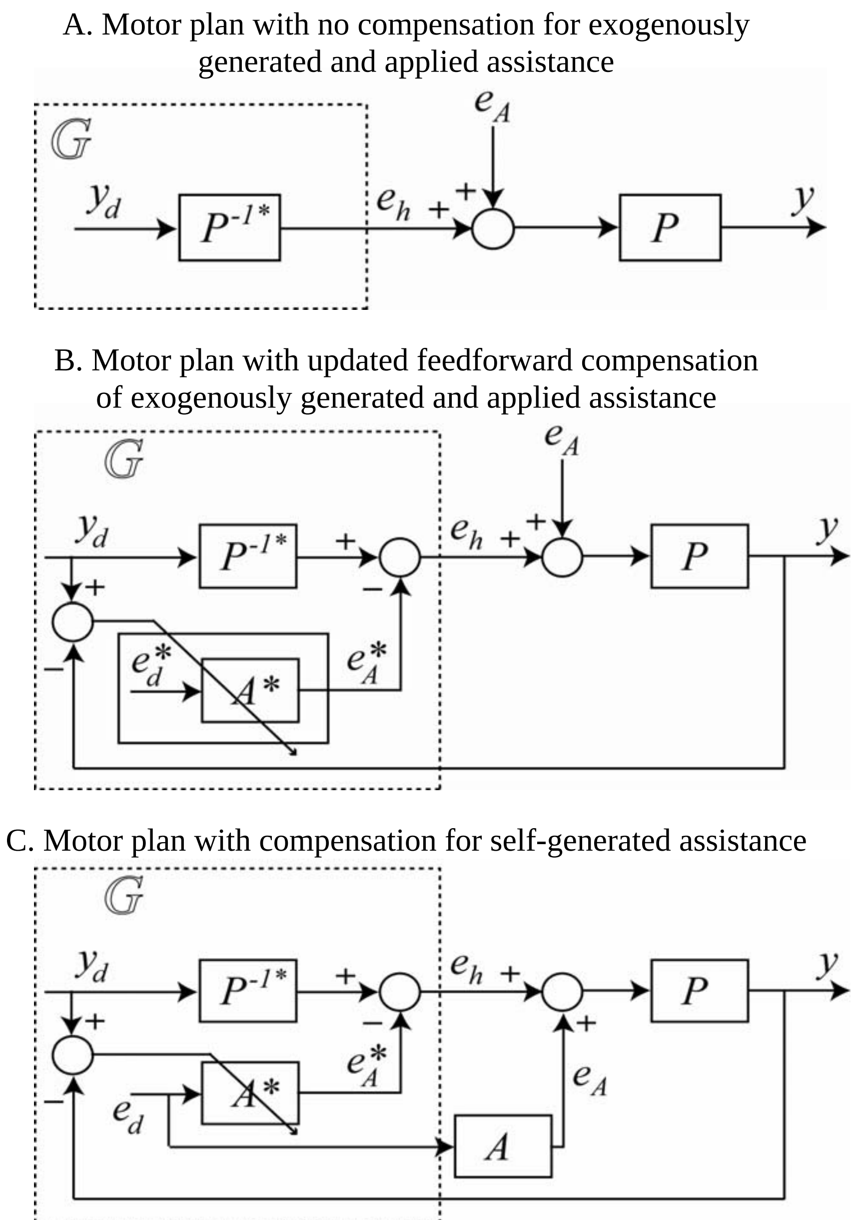

Figure 4.1 Block diagram of human and assistive efforts. The effort applied by the human subject (e_h) sums with the assistive effort (e_A) to act on the limb (P) and generate limb motion (y). Subjects generate a motor plan for the desired limb motion (y_d) using an inverse model of the limb (P^{-1}*), developed over years of use. In all diagrams, the * represents a model or expectation of an unknown element. Block diagram A illustrates the case when no compensation for outside assistive effort is considered in the development of the motor plan or generation of human effort (e_h). In B, the subject attempts to cancel the assistive force e_A using an updated model of the expected forces. This model includes a prediction of the desired assistive force (e_d*), that is the intention of the assisting agent such as a therapist or an autonomous controller. In addition, the model must account for the minimally changing actuator dynamics through which the force is applied through (A*), such as the therapist's body or robot dynamics. In the case of self-generated assistance (diagram C), the subject compensates for the assistive forces by generating and updating only a model of the application dynamics (A*), since the desired assistive effort (e_d) is known through efference copy.

4.1 Previous work on anticipatory adjustments

The central nervous system uses anticipation together with feedforward motor actions to compensate for the effects of disturbances with known characteristics. When a person self-generates a disturbance, it is easier for the nervous system to identify the properties and timing of the disturbance than when another agent produces the disturbance. Anticipatory compensation for self-generated disturbances has been demonstrated in lower limb muscle activation of standing subjects maintaining postural stability [56, 65, 66], as well as in upper limb muscle activation of seated subjects [67-69]. If a disturbance is a result of a voluntary action by the person, activation in the affected muscles will occur prior to the disturbance onset. This muscle activation is generated by the central nervous system to minimize the effects of predictable perturbations and essentially acts as the biomechanical equivalent of feedforward cancellation [56]. Anticipatory postural adjustments in the legs and hips have been found preceding the initiation of leg movements [66, 70], trunk movements [71], and arm movements [65]. Lower limb anticipatory adjustments are also present when the disturbance is caused by a change in load imposed on the body rather than only movement of body segments [56, 72]; this includes the application or removal of an external load on the upper limbs [73, 74]. In general, studies concur in the finding that anticipatory adjustments occur when subjects are responsible for the perturbation through a volitional action, but not when the perturbation is generated by an outside source, such as a computer or the experimenter. In addition, anticipatory adjustments in postural muscles have been shown to depend on the type and magnitude of the generating volitional action, the magnitude of the disturbance, the subject's postural position, and subject's muscular fatigue [56, 75, 76].

These motor behavior principles have also been examined using upper limb disturbance rejection tasks with perturbations of various levels of predictability. During voluntary upper limb coordination, muscle activity in the arms has been shown to precede self-generated load disturbances. These anticipatory adjustments have been explored in the form of modulated grip forces [45, 77] and stabilization forces of the hand or arm [69, 78].

Researchers have postulated that humans utilize efference copy and internal models to accommodate for external load disturbances generated through volitional actions. In a bimanual unloading task, subjects demonstrated feedforward deactivation of stabilizing arm muscles when they directly remove the load [67]. Further studies showed that subjects were capable of anticipatory actions when the volitional action was as small as a button push or trigger; in such cases, however, anticipatory actions appeared only after an extended learning period [57, 78]. In all experiments, subjects were unable to generate precisely timed anticipatory adjustments in an upper limb unloading task if the load was removed with cuing [57] or in an otherwise predictable manner [45].

Mechanical coupling between the two upper limbs usually takes place through an object grasped, held, or manipulated between the hands. As such, bimanual studies have examined disturbances applied to an object in a grasping or stabilizing hand, as through a load application or removal. In the postural stability studies, however, the disturbances to the lower limbs are mechanically transmitted internally through the body segments from the original location of the disturbance. For example, when a load is released at the hands, as in [73], that external disturbance will dynamically propagate through the body to cause a postural perturbation. For both types of disturbance application, external applied or internally transmitted, neural communication and the utilization of efference copy must exist between the limbs involved for the coordinated feedforward control signals that accompany subject initiated perturbations.

Lower limb anticipatory adjustments have thus far only been examined in the context of postural stabilization, that is, perturbations from mechanical interlimb coupling. Coordination of upper and lower limb disturbance applications involving tasks with directly applied loads or coupling through an object have not yet been explored. This type of directly applied force is precisely the type we are interested in when discussing rehabilitation applications. Frequently, lower limb rehabilitation involves exercising with applied loads, assistive or resistive, or exercising in other precision lower limb tasks when locomotor or postural practice is not possible or desired.

The experiments described in this chapter were designed to explore how anticipatory actions are developed for consistent, repeatable, externally applied disturbances, especially in the lower limbs. We aim to determine if the efference copy associated with an upper limb action could be used to benefit performance in an upper or lower limb task. Specifically, we tested if healthy subjects could better compensate for a disturbance applied at the lower limbs when that disturbance was self-generated by their own upper limbs. In addition, we replicated other results from the literature by comparing upper limb disturbance rejection to bimanually triggered perturbations. We hypothesized that subjects would demonstrate reduced peak accelerations when they self-initiated unloading than when the computer initiated unloading. We suspected that subjects could more fully integrate internally generated information in the development of motor commands than externally provided cues. Furthermore, we hypothesize that self-generated assistance would be used more effectively and better understood by the patient than assistance directed by another agent. A key aspect of the study is that subjects performed both upper limb and lower limb trials in a non-postural configuration.

4.2 Methods

These experiments used hardware described in Chapter 2 (Device 1, Figure 2.3), configured for subjects to be in a seated position. Subjects interacted with the programmable motor through either the foot platform or the handlebar. In addition, the handheld pushbutton was used as a means for accepting user input.

4.2.1 Protocol/Experimental Task

Subjects were presented with a simple motor task: to minimize motion of the handlebar or footplate when subjected to a large and sudden change in load. Within a given trial, or single execution of the task, the subject first opposed a slowly applied load until a predetermined magnitude was reached. Depending on the experiment configuration, the maximum load amplitude W was set to 14 N, 20 Nm, or 10 Nm for experiments with one hand, both feet, or one foot, respectively. The subject held his hand, feet, or foot steady while the load was rapidly removed. This function produced a smooth unloading such that 95% of the load was removed within the first 200 ms following the release time. The motion of the footplate was displayed as a moving horizontal bar on the monitor. Subjects were instructed to maintain a steady position before the weight release and to minimize motion following the unloading.

Within a particular experiment, the condition tested was the means for triggering the unloading function. These were *Uncued*, *Self-Triggered*, or *Computer-Cued*. In the Uncued condition (considered the control) the subject received no visual or auditory warning; the release occurred after a randomly generated delay once the load was applied and motion had settled. For the Self-Triggered condition, the subject was in direct control of the load release through a pushbutton held in one hand. Once the load was applied and motion had settled, the subject was permitted to press the button at any time to release the load. For the Computer-Cued condition, the subject was cued prior to the load release with a sequence of four colored lights on the computer monitor accompanied by four short tones on the speakers. Although the subject had no control over the trigger of the load release in the Computer-Cued condition, he was forewarned of the timing with the regularized visual and auditory cues.

The experiments were all structured in similar fashion to ensure many repetitions of the task in each condition and to limit the effects of ordering. Each experiment included three blocks of Uncued trials and four blocks each of the Computer-Cued and Self-Triggered trials. Blocks of the Computer-Cued and Self-Triggered trials consisted of twelve standard task completions. These two blocks were presented in alternation. Pilot testing indicated small variability among trials in the Uncued condition; therefore these blocks included only five trials. In addition, the three Uncued blocks were presented at the beginning, middle and end of the block sequence. Table 4.1 summarizes the presentation sequence for the blocks during each experiment.

Intermixed with the trials already described, we included a small number of *catch* trials. A catch trial behaved like a standard trial in all respects except that no unloading actually occurred. When the unloading function was triggered (whether by the computer or the subject) the load remained constant instead of releasing. Catch trials were included in all of the Computer-Cued and Self-Triggered blocks. To discourage alterations in strategy based on expectations of catch trials, we presented subjects with only one catch trial per block, randomly positioned among the twelve standard trials. The subjects were forewarned of the existence of occasional catch trials, but were given no indication of when one might occur.

Table 4.1 Summary of conditions and experimental presentation sequence

Block	Condition	Std Trials	Catch Trials
1	**Uncued**	5	None
2	**A** (ex. Self-Triggered)*	12	1
3	**B** (ex. Computer-Cued)*	12	1
4	**A**	12	1
5	**B**	12	1
6	**Uncued**	5	None
7	**B**	12	1
8	**A**	12	1
9	**B**	12	1
10	**A**	12	1
11	**Uncued**	5	None

* For Experiments 1 and 2, subjects were grouped randomly to receive either Self-Triggered or Computer-Cued as the first test condition, thus defining the A and B conditions for the experiment. For Experiment 3, subjects were grouped randomly to receive either Ipsilateral or Contralateral Self-Triggered as the first test condition.

This apparatus and protocol were used for three distinct experiments: *Upper Limb Unloading*, *Lower Limb Unloading*, and *Ipsilateral/Contralateral Triggering*. Table 4.2 provides a summary of the three experiments, the conditions associated with each, and the number of trials presented in each.

Table 4.2 Summary of experiments and associated conditions

Condition	# Blocks	Standard Trials per Block	Catch Trials per Block
Experiment 1: Upper Limb Unloading			
Uncued	3	5	None
Computer-Cued	4	12	1
Self-Triggered	4	12	1
Experiment 2: Lower Limb Unloading			
Uncued	3	5	None
Computer-Cued	4	12	1
Self-Triggered	4	12	1
Experiment 3: Ipsilateral/Contralateral Triggering			
Uncued	3	5	None
Ipsilateral Self-Trig.	4	12	1
Contralateral Self-Trig.	4	12	1

4.2.2 Experiment 1: Upper Limb Unloading

The first experiment used an upper limb unloading task to test the dependence of anticipatory actions on the type of initiation: whether the unloading event is self-triggered or computer-triggered. Each subject opposed the 14 N load by pushing the handlebar with his dominant hand, while operating the pushbutton with his non-dominant hand. For this experiment, the Uncued condition acted as the control while the two test conditions were Self-Triggered and Computer-Cued. During all conditions, the subject propped his feet on a stationary footrest. The Self-Triggered and Computer-Cued blocks were presented in alternation and the first condition presented to each subject was assigned in a balanced fashion.

4.2.3 Experiment 2: Lower Limb Unloading

The second experiment used a lower limb unloading task applied directly at the feet to test the dependence of lower limb anticipatory actions on whether the unloading event is self-triggered by the upper limbs or computer-triggered. The experimental task was completed with the subject using both feet to stabilize the footplate against a rapidly released load, with maximum amplitude *W* of 20 Nm at the ankle joint. The load was applied in a single direction requiring plantar flexion to oppose it before release. As with Experiment 1, the two test conditions were Self-Triggered (with the non-dominant hand acting through the pushbutton) and Computer-Cued. Once again, the test condition presented first was assigned to subjects in a balanced fashion.

4.2.4 Experiment 3: Ipsilateral/Contralateral Triggering

The third experiment used an unloading task in which one upper limb was used to trigger a motor response in one lower limb. We examined the anticipatory actions developed in the single lower limb when the release was triggered by the ipsilateral hand and when the release was triggered by the contralateral hand. For this experiment a single foot, the dominant foot, was used for the unloading task. Correspondingly, the maximum load *W* was set to 10 Nm, half the load used in Experiment 2 for both feet. The Uncued condition acted as the control, and the test conditions were both a type of Self-Triggered condition. Specifically, one condition was *Ipsilateral* Self-Triggered, where the subject triggered the unloading with the pushbutton held in the hand ipsilateral to the controlling foot (the dominant hand). The other condition was *Contralateral* Self-Triggered, where the subject triggered the unloading with the contralateral (non-dominant) hand. During these trials, the unused hand and non-dominant foot rested on the armrest and footrest, respectively. The test condition presented first was assigned to subjects in a balanced fashion.

4.2.5 Participants

The same subject pool of twelve individuals (8 male, 4 female) between the ages of 23 and 32 participated in each of the three experiments. All subjects were free of any upper or lower limb malformations or impairments and reported right hand/ right foot dominance and normal or corrected-to-normal vision. Subjects were not compensated for their participation. All subjects provided informed written consent in accordance with University of Michigan IRB policies.

4.3 Data analysis and performance metrics

The data were analyzed by first aligning all measurements for each trial to the initialization of the load release, setting initial trial time to $t = 0$ in the unloading function. We evaluated subject performance based on the acceleration profile that followed the release. We expected that small acceleration magnitude would be an indication of better compensation for the load release. Figure 4.2 illustrates a sample torque load trace and associated acceleration response for both a standard and a catch trial. As the primary metric for performance comparisons, we used the peak acceleration in the first 200 ms following load release, defined as peak acceleration $\mathcal{P}$ (noted as the star in Figure 4.2). This peak acceleration was generally positive for the standard trials and negative for the catch trials; therefore the performance $\mathcal{P}$ of the standard trials and catch trials were analyzed separately for comparisons between conditions. Statistical analyses were performed in SPSS (Chicago, IL) using a repeated measures linear mixed model with condition and block as fixed variables and subject as a random variable. The threshold for statistical significance was set at $\alpha = 0.05$. The Bonferroni method of adjustment for multiple comparisons was used for correcting significance levels in the pairwise comparisons of the three conditions.

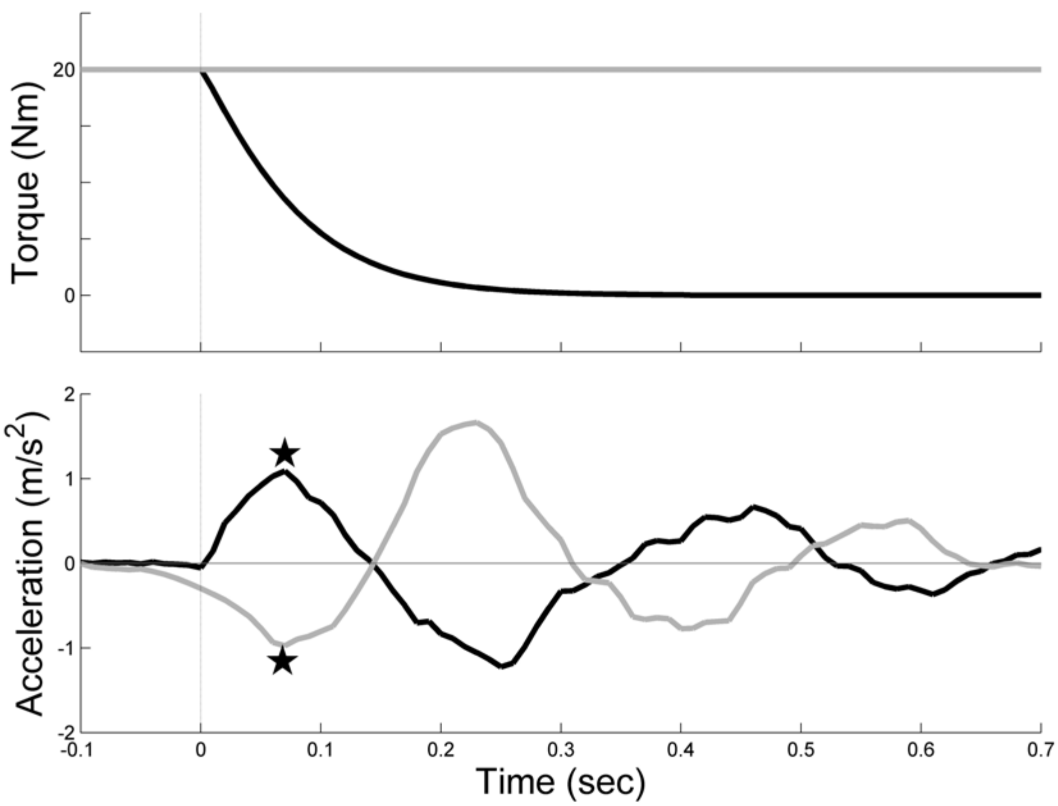

Figure 4.2 Sample traces of load disturbance and resulting acceleration. The upper plot shows a sample time history of the torque applied to the feet in a standard trial (black line) and a catch trial (grey line). The corresponding accelerations, shown in the lower plot, highlight the general differences in the standard and catch trial response waveforms. The acceleration profiles for all non-catch trials contain a sharp positive acceleration that results from slight mistiming between the release of the external load and the deactivation of the muscles that had been opposing that load. The initial negative acceleration apparent in the catch trial results from the deactivation of the stabilizing muscles in anticipation of the load release when the load, in actuality, remained. The magnitude of this initial peak (or valley in the case of catch trials), demarcated with the stars, is used as the primary performance metric.

As a second performance metric, we examined the release acceleration $\mathcal{R}$. This is defined as the acceleration at time $t = 0^-$, or just before the onset of the load release. The release acceleration occurs before the divergence between standard trials and catch trials, therefore $\mathcal{R}$ was computed with the aggregate of all trials for comparisons in the three conditions (Uncued, Computer-Cued, and Self-Triggered). The same statistical model and significance level were used for this second performance metric.

4.4 Results

The degree to which subjects were able to compensate for the load disturbance depended on the condition triggering the disturbance. When using their upper limbs to oppose the load (Experiment 1), subjects were able to compensate for the load release better when the release was self-triggered than when it was triggered by computer. When computer-triggered, subjects were able to compensate better when a timing cue was presented. Acceleration waveforms exhibited a characteristic sequence of peaks and valleys (Figure 4.2) for all trials. The mean acceleration waveforms, computed by averaging across the 12 subjects and the 48 trials for the Self-Triggered and Computer-Cued conditions, and the 15 trials in the Uncued condition, were ordered in overall magnitude by condition, with Self-Triggered exhibiting the smallest, Computer-Cued the intermediate, and Uncued the largest levels (Figure 4.3a). Smaller acceleration magnitudes indicate less motion of the platform following the load release and therefore better disturbance compensation. The mean waveforms of the three conditions in standard trials are clearly distinct in magnitude, despite similarly shaped waveforms. Post hoc analyses with multiple comparison procedures indicate significant differences in the peak acceleration $\mathcal{P}$, as defined in Section III (Figure 4.4). Compared to the Uncued condition, the Computer-Cued condition produced smaller $\mathcal{P}$ by 4.7% (0.11 m/s^2 with $p = 0.006$) while the Self-Triggered condition produced smaller $\mathcal{P}$ by 21.9% (0.53 m/s^2 with p

< 0.001). The peak acceleration $\mathcal{P}$ for the Self-Triggered condition was also significantly smaller than for the Computer-Cued condition (p < 0.001).

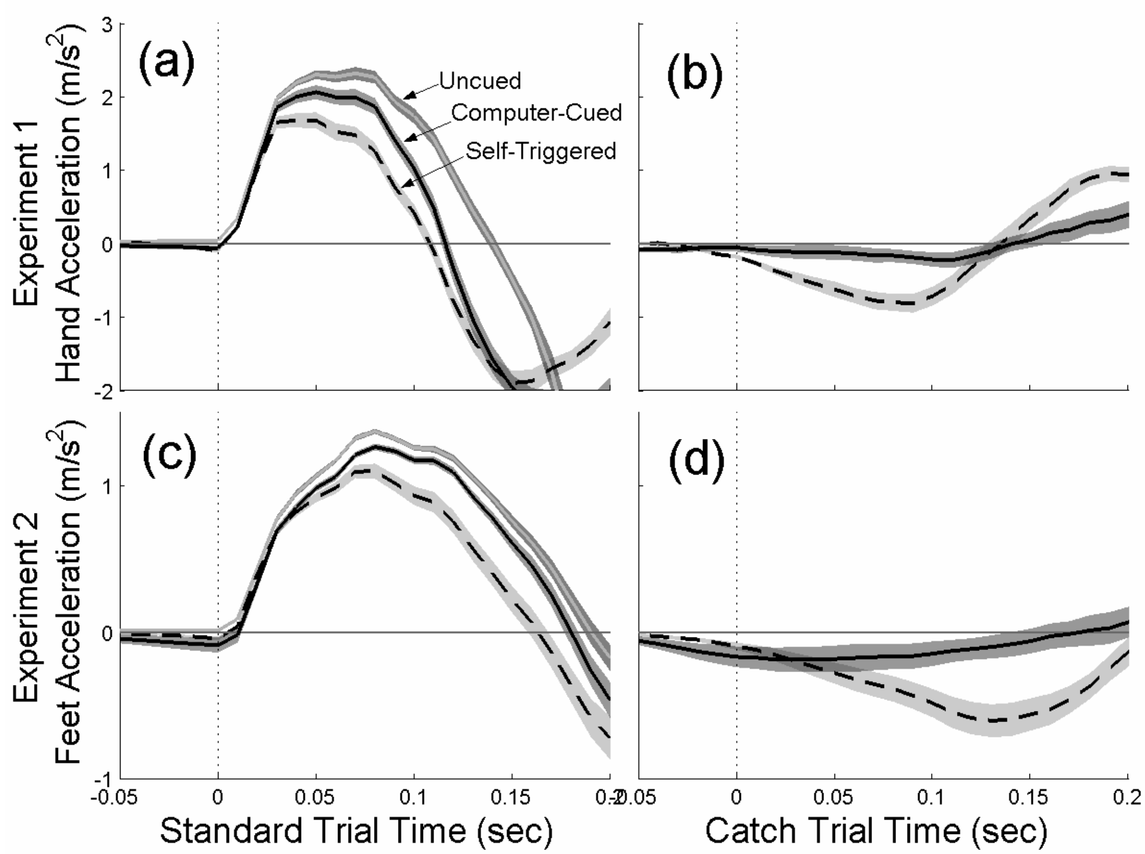

Figure 4.3 Mean time history of accelerations. The mean acceleration responses, averaged across trials and subjects, for the upper limbs (a and b) and lower limbs (c and d) show differences in behavior by condition (Uncued – grey trace, Computer-Cued – solid black trace, and Self-Triggered – dashed black trace). The shaded regions indicate between-subject standard error. In standard trials (a and c), trials under the Self-Triggered condition exhibit reduced accelerations compared to those under the Computer-Cued and Uncued conditions. In catch trials (b and d), responses show larger accelerations for the Self-Triggered catch responses compared to the Computer-Cued catch responses.

Subjects exhibit more active compensation for the expected but missing load release during the catch trials for the Self-Triggered condition than for the Computer-Cued condition. The acceleration profile for the catch trials took one of two possible forms. In some trials, there appeared to be essentially no change in acceleration before and after the load release. However, in many trials, the acceleration pattern resembled the inversion of the acceleration profile produced in standard trials (Figure 4.2). The combination of these two shapes resulted in the catch trial mean acceleration waveforms shown in Figure 4.3. The accelerations (and the platform motion) were in the opposite direction relative to standard trials. Although the mean magnitude is smaller for catch trials than standard trials, distinctions by test condition still exist. Just as the catch trial mean acceleration waveform is generally an inversion of the standard trial mean waveform, the peak accelerations $\mathcal{P}$ for the catch trials are minima rather than maxima (peaks). Subjects have larger $\mathcal{P}$ magnitudes in the Self-Triggered catch condition than in the Computer-Cued catch condition. The repeated measures statistical analysis of $\mathcal{P}$ magnitudes indicates a significant difference between these two conditions (Figure 4.4).

In Experiment 2, subjects demonstrated similar capabilities and patterns in disturbance compensation using their lower limbs as they demonstrated in Experiment 1 using their upper limbs. The mean accelerations for standard trials were smaller for the Self-Triggered condition than for the Computer-Cued condition and larger for the Uncued condition than either of the other two (Figure 4.3c). The peak accelerations $\mathcal{P}$ for the Computer-Cued condition were 4.5% (0.06 m/s^2) smaller than for the Uncued condition ($p = 0.042$); while the peak accelerations for the Self-Triggered condition were 15.6% (0.22 m/s^2) smaller ($p < 0.001$). The catch trial data for the lower limbs also show results consistent with the findings in the upper limb performance (Figure 4.3d). The peak accelerations $\mathcal{P}$ for the Self-Triggered conditions were significantly larger than those for the Computer-Cued condition ($p = 0.0018$).

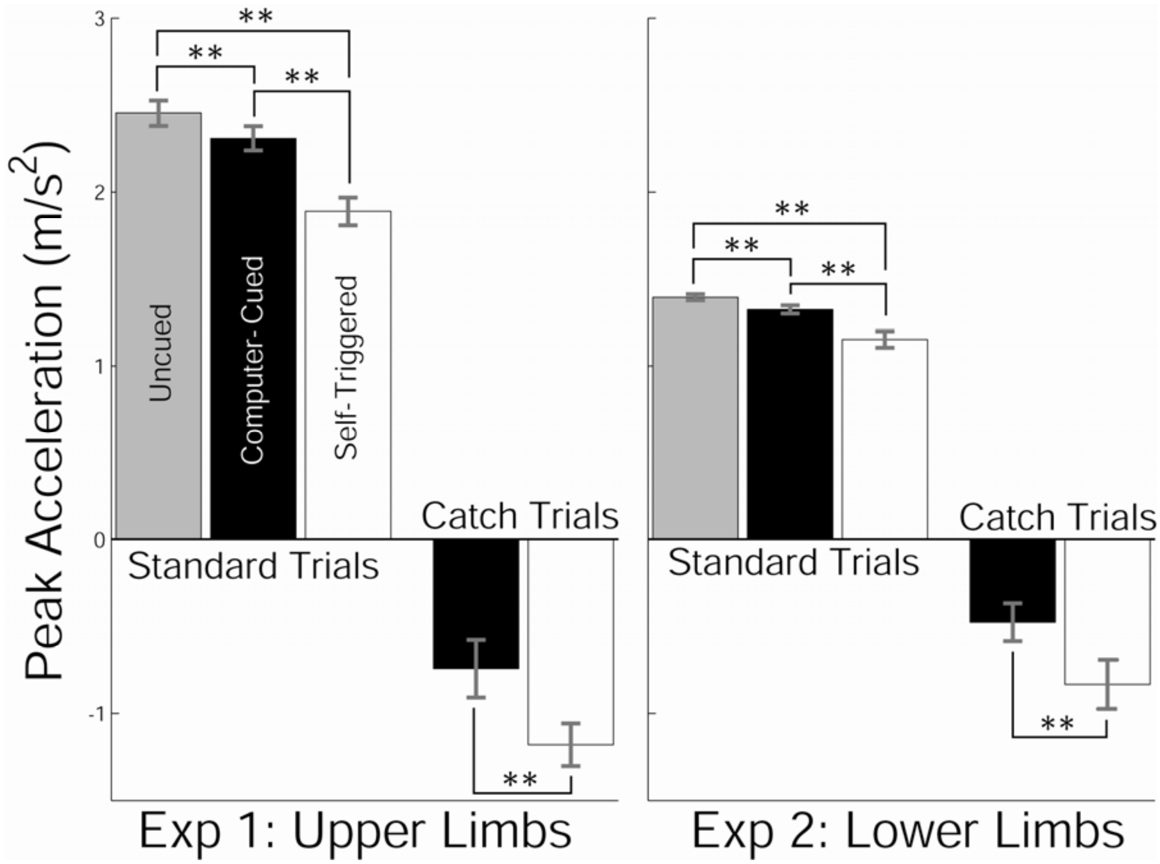

Figure 4.4 Peak accelerations summary. The peak accelerations P for Experiment 1 (upper limbs) and Experiment 2 (lower limbs) demonstrate a significant difference between conditions (Uncued – gray, Computer-Cued – black, Self-Triggered – white) for both the standard and the catch trials. The error bars represent the between-subject standard error and the double asterisks (**) denote significant differences between means with $p < 0.05$.

Examination of $\mathcal{R}$, the release acceleration immediately before the unloading, also shows distinct differences by condition. For all conditions in both experiments, the release acceleration was very close to zero; however, some distinctions do exist with release acceleration tending to demonstrate negative magnitude, meaning in the opposite direction of the imminent disturbance, when information regarding the disturbance was available (Figure 4.5). In Experiment 1, with upper limb unloading, the Uncued condition demonstrated accelerations not significantly different from zero (0 m/s^2), whereas the Computer-Cued and Self-Triggered conditions resulted in non-zero $\mathcal{R}$ values

(-0.07 m/s^2 and -0.11 m/s^2, respectively). For Experiment 2, with the lower limb unloading, the mean accelerations of the Uncued and Self-Triggered conditions were not different from zero (0 m/s^2 and 0.04 m/s^2, respectively) while the Computer-Cued condition yielded negative mean accelerations (-0.09 m/s^2). Although the differences in the mean $\mathcal{R}$ values for the Self-Triggered and Computer-Cued conditions may not be of practical significance, there exists a significant difference in the variability of the two ($p < 0.05$, for both experiments; Levene's test with Bonferroni adjustment); the Computer-Cued condition shows significantly larger between-subject and within-subject variance than the Self-Triggered condition (Figure 4.5).

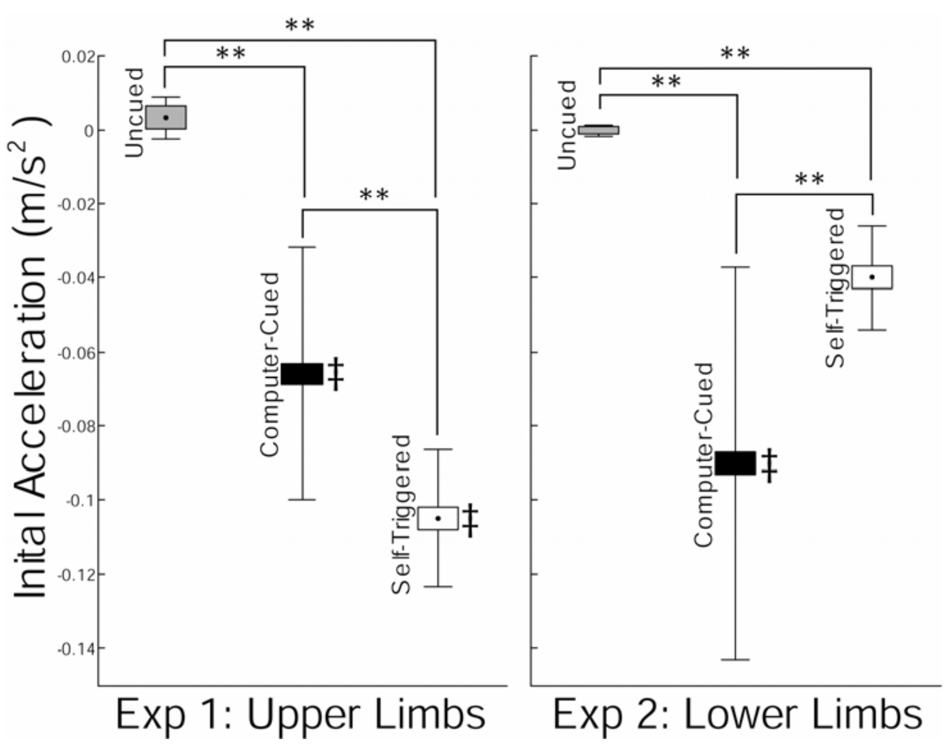

Figure 4.5 Release acceleration summary. The mean release accelerations at time t = 0- show that in the Computer-Cued and Self-Triggered conditions, subjects tended to accelerate to oppose the imminent disturbance. In addition, each of the three conditions demonstrated significantly different variances ($p < 0.05$ Levine's test, demarcated by **). The error bars represent the between-subject standard error and the ‡ signifies the 95% confidence interval of the mean is significantly different from zero.

4.4.1 Experiment 3

Unlike the first two experiments that compared responses to disturbances triggered either by the subject or by a computer, Experiment 3 compared the ability of a single foot to compensate for Self-Triggered disturbances initiated by the subject's ipsilateral or contralateral hand. Subjects exhibited no difference in compensatory abilities when the disturbance was triggered using the ipsilateral versus contralateral hand. The peak accelerations $\mathcal{P}$ for the Ipsilateral Self-Triggered condition and the Contralateral Self-Triggered condition were distinct from the Uncued condition (both $p < 0.001$), but not from one another ($p = 0.67$). Additionally, non-zero mean $\mathcal{P}$ for the catch trials appears to demonstrate active compensation for both types of Self-Triggered conditions, whether Ipsilateral or Contralateral. As with the standard trials, there was no demonstrable distinction between the two conditions in mean $\mathcal{P}$ for the catch trials ($p = 0.64$). The platform release accelerations immediately preceding the disturbance (the $\mathcal{R}$ metric) demonstrate trends similar to those for the Self-Triggered condition previously discussed for Experiments 1 and 2. For both test conditions, $\mathcal{R}$ is less than the nominal acceleration resulting from the Uncued condition ($p < 0.001$).

4.5 Discussion

Results from this study confirm the basic findings from previous bimanual unloading experiments: self-generation of a disturbance is associated with improved compensation, ostensibly through the processes of efference copy, and feedforward control. In Experiment 1, our results show that self-generation of a disturbance through a button press yields reduced object motion compared to a computer-triggered disturbance

(Figure 4.3a), similar to the results of [57]. We extended these results through Experiment 2 to demonstrate that lower limb anticipatory adjustments develop as a feedforward means of countering direct force perturbations. Further, this lower limb cancellation was most appropriately timed when the disturbance is self-generated from an upper limb volitional action (Figure 4.3b). Despite characteristic differences that exist between the upper and lower limbs, the performance changes across condition generated for the lower limb unloading task were comparable to those produced in the upper limb unloading task (Figure 4.4). Our results also indicate that although audio-visual cuing of a disturbance does not produce as strong an anticipatory action as a self-generated disturbance, the additional information received through the cuing allows subjects to achieve somewhat improved performance compared to no cuing.

Anticipatory adjustments for disturbances generated by a button press or trigger may develop differently than anticipatory adjustments for disturbances generated by a gross motion. Our experimental protocol involved only a triggered disturbance. In studies by [57, 78], the development of anticipatory adjustments required an extended learning period. Results from [73], however, suggest that a trigger will elicit anticipatory adjustments without learning but the size will be scaled according to the magnitude of the motor action that caused the perturbation. Our results fall in line with the latter studies; that is, apart from the initial few trials of the testing sequence there were no apparent trends across or within subjects toward performance enhancement or degradation as the testing sequence progressed. Furthermore, pilot studies involving extended practice did not indicate marked improvement for any unloading condition, and the differences between the Self-Triggered and Computer-Cued conditions persisted robustly.

After we verified bimanual anticipatory adjustments in an unloading task using our apparatus and protocol, we presented the same task to the lower limbs. The literature has established the existence of lower limb anticipatory adjustments for tasks involving postural stability. Note that in postural tasks, disturbances generally originate at body segments remote to the lower limbs and are transmitted by virtue of mechanical interconnection of body segments. This type of postural perturbation is distinctly different from those that have been used in bimanual tasks, where the perturbations arise from loads applied through external objects or directly to the limb in which the anticipatory adjustments are examined. We were interested in testing a non-postural task in the lower limbs to further explore relationships between anticipatory adjustments shown for postural maintenance and for upper limb stabilization. We thus compared the existence of anticipatory adjustments in the upper limbs (associated with an upper limb triggered disturbance) to anticipatory adjustments in the lower limbs (likewise associated with an upper limb triggered disturbance), where subjects were seated for both experiments. Note that there exist many points by which one would expect differences in the anticipatory adjustments for the upper and lower limbs; these differences include morphology of the limbs, musculoskeletal anatomy, biomechanics of the joints involved, and the length and structure of neural pathways. Despite these characteristic differences between the limbs, the performance changes across condition generated for the lower limb unloading task were comparable to those produced in the upper limb unloading task.

Feedforward control and anticipatory adjustments offer a viable explanation for our observed dependence of disturbance rejection on the predictability of the disturbances. However, we must consider whether our subjects used muscular co-contraction, another available strategy for disturbance rejection. Indeed, co-contraction may have been used by our subjects; however, if this were the only strategy employed, we would expect to see no motion in the catch trials. Although for some catch trials we observe little to no motion, many trials exhibit acceleration profiles very similar to an inversion of the standard trails. This is especially true for the Self-Triggered trials, where the mean across all subjects and all trials (including those with no motion) maintains the characteristic inverted shape, but at slightly reduced magnitudes. The Computer-Cued trials also present non-zero mean acceleration profiles, but lose almost all features of the inverted shape (Figure 4.3). The catch trials of both conditions indicate that there exists some component of feedforward control. However, only in the mean acceleration traces of the Self-Triggered condition can we identify a profile that would act to cancel a portion of the expected accelerations as established in standard trials. The presence of a compensating strategy in the Self-Triggered catch trials, but absence of a compensating strategy in the Computer-Cued catch trials suggests that subjects employ either more aggressive or more consistently timed anticipatory actions when the disturbance is Self-Triggered rather than Computer-Cued.

The results from the $\mathcal{R}$ metric, the release acceleration at time $t=0$, indicate that anticipatory actions in both the upper limb and lower limb tasks are often ill-timed for the Computer-Cued trials. The Self-Triggered and Computer-Cued conditions for the upper and lower limbs generated mean release accelerations across all subjects that are distinct from zero and in the direction opposite of the acceleration expected immediately following the load release. The instantaneous acceleration difference at the point of release for the two conditions is small; however, the difference in the variances of the data sets is worth noting. For the upper limb and lower limb experiments, the variance in $\mathcal{R}$ for the Computer-Cued condition is significantly larger than for the Self-Triggered condition. The larger variance indicates that compensatory actions were less focused. Although subjects may have attempted to account for the unloading disturbance given cues, the compensation was less precisely applied than when the disturbance was self-triggered.

In Experiment 3, we investigated whether characteristics of anticipatory adjustments in a single lower limb are dependent on the upper limb that generated the neural cue. Our results indicate no difference in performance between disturbances triggered from the ipsilateral hand and disturbances triggered in the contralateral hand. Neural coupling has been shown to exist bilaterally in the upper limbs [32] and ipsilaterally in the upper and lower limbs [33, 64] for rhythmic tasks. Studies have not discussed the comparative strength of the bilateral and ipsilateral coupling nor have they examined contralateral coupling of the upper and lower limbs. There is indication in the literature that anticipatory adjustments are stronger when the hand and foot are positioned isodirectionally, that is, in-phase motion of the two limbs occurs in the same direction spatially rather than biomechanically [37]. Our results from Experiment 3 cannot be used alone to determine if neural coupling is different for ipsilateral upper and lower limbs than for contralateral upper and lower limbs. However, we can deduce that neural coupling exists between the upper and lower limbs for both the contralateral and ipsilateral cases as demonstrated by the similarity in performance and development of anticipatory adjustments under both conditions. Perhaps further studies that employ gross motor actions producing similar workspace motion of the upper and lower limbs could illuminate finer differences between bilateral (contralateral) and ipsilateral limb coupling [38, 64].

4.6 Significance

Results from Experiment 2 indicate that seated subjects develop anticipatory adjustments in the lower limbs while performing an unloading task. The effects observed were similar in character and magnitude to those demonstrated in bimanual unloading tasks performed on the same apparatus (Experiment 2). Lower limb neuromuscular commands appear to incorporate predictions of interaction forces from the upper limbs. The ability to accurately predict disturbances, as occurs when the disturbances are self-generated, facilitates coordination and integration of those outside loads with the motor plan development and execution by the subject. This coordination and integration of assistance is likely one of the primary mechanisms underlying the success of bimanual self-assist. The extension of this mechanism to a lower limb task supports a generalization of bimanual self-assist, where the control and application site of assistance might span non-homologous muscle groups, provided that the prescribed assistance is self-generated. Therefore, self-assist with the upper limbs guiding the lower limb motion may be a productive means of neurological rehabilitation [79]. In contrast, interaction forces from outside sources (i.e. robots or therapists) are not easily anticipated and therefore are not readily incorporated into lower limb neuromuscular control during therapy. This type of outside agent assistance may be limited when compared to self-assist because of the reduced patient involvement and coordination.

CHAPTER V

Active dorsiflexion in chronic stroke subjects

In previous chapters we have discussed in depth some of the hypothesized benefits of self-assist in a rehabilitation setting. We have presented support from the literature concerning various motor control mechanisms that we theorize will be exploited during self-assist. Increased patient involvement, additional afferent sensory feedback, and coordination enabled by efference copy will all contribute to motor recovery in rehabilitation. We also claim, and have demonstrated with the experiments presented in Chapter 3, that the cognitive challenges that arise from self-assist do not outweigh the benefits it may offer to lower limb capabilities. This, however, was only addressed for neurologically intact subjects. Even though motor control in healthy subjects provides a representative model for motor recovery, there exist many differences between motor adaptation and motor rehabilitation. Impairment following a neurological trauma can take many forms. Some subjects may suffer from aphasia where the speech centers of the brain are primarily affected. Other subjects may maintain most of their strength and motion capabilities, but require much more concentration to execute motor functions on the impaired side. Subjects may have difficulty in coordinating desired motions or in maintaining necessary levels of concentration. In addition, some research suggests that neurological trauma can alter the nature of the coupling that exists between the upper and lower limbs. Knikou showed that arm motions can act to suppress lower limb spinal reflexes in patients with a spinal cord lesion [35]. Kline et al. demonstrated increased neural coupling between the upper and lower limbs of stroke subjects [39]. The implications of multi-limb coordination and control might prove significantly different in an impaired population as compared to a neurologically intact population.

This chapter compares self-assist with other methods of assistance in lower limb motor control of an impaired population. We look at subjects who have suffered a cerebral vascular accident (CVA) in performing a simple, functionally motivated motor task. Subjects were required to dorsiflex their ankle, i.e. activation of the tibialis anterior (TA), to avoid a series of moving obstacles. This task is motivated by the dorsiflexion requirements that accompany locomotion. During the swing phase of gait, a person must activate the TA, holding the toes up, to avoid tripping. Reduced dorsiflexion is a common impairment following a CVA and is commonly referred to as drop-foot. Development of a motor exercise that allows neuromuscular training of the TA is a secondary motivation for the self-assist study presented in this chapter.

5.1 Drop-foot as a functional motivation

Drop-foot is an often chronic disability that accompanies hemiparesis in up to 20% of persons with history of stroke [80]. This motor deficiency is caused by lack or loss of motor control, often manifested as weakness, in the tibialis anterior (TA) resulting in reduced dorsiflexion capabilities of the ankle. Drop-foot is primarily characterized by foot-slap and toe drag. Foot-slap occurs when the foot slaps the ground uncontrolled just after heel strike in the gait cycle. Although there are no immediate detrimental effects of this condition, it is often accompanied by a noticeable and undesirable noise. Toe drag refers to the inability to lift the toe during the swing phase of gait, thus the toe drags on the ground. Toe drag interrupts the progression of the gait cycle and greatly increases the risk of tripping.

Physical and occupational therapists work with patients intensely during the acute phase following CVA to regain as much motor function as possible. However, the long term rehabilitation and exercise regimens do not generally include accommodations for focused attention and strengthening of the TA. The TA (an ankle dorsiflexor) is a smaller muscle than the gastrocnemius (an ankle plantar flexor), and is generally responsible for only lifting the weight of the foot during locomotion. It controls the foot during heel strike (preventing slapping) and holds the foot out of the way during swing (preventing toe drag). The gastrocnemius and other plantar flexors, on the other hand, play a large role in locomotion, contributing nearly all of the power during push-off [81].

Drop-foot is generally treated by prescribing an ankle-foot orthosis (AFO) to immobilize the foot. This rigid mechanical brace holds the foot in a neutral position, preventing toe drag during swing. There are both benefits and limitations to this solution. Some studies report that AFO use induces increased step frequency, step length, velocity and ankle angle [82]. Others indicate no improvement in gait velocity or step length [83]. This passive device does nothing to improve drop-foot when not worn, nor does it address foot-slap. An alternative to the passive AFO is an active AFO. Active AFOs provide mechanical actuation in addition to stabilizing support [83].

Another active alternative to the AFO is a portable peroneal stimulator. This battery operated device applies an electrical stimulation to the peroneal nerve inciting continuous muscle contractions and active dorsiflexion [80, 84]. The electrical signal can originate from surface or implanted stimulators and can be triggered by a foot switch or other sensor to determine shank motion or orientation. The peroneal stimulator takes a large amount of custom adjusting and instruction from a trained professional before it can be effectively used by a subject. In addition, some subjects are unable to successfully use the device regardless of the tuning and training received. These stimulators may greatly improve the effects of drop-foot resulting in a more normal appearing gait; however, they offer no significant improvement in recovery, i.e. when the device is not in use [84]. Prolonged use of peroneal stimulators is often accompanied by patient discomfort, especially skin irritation.

Researchers are also beginning to explore therapeutic interventions addressing ankle mobility that employ robotic technologies. If successful these could benefit a large number of patients and reduce the need for long term assistive aids, such as the AFO [85]. Drop-foot might be further addressed through a mechanized exercise system to rehabilitate ankle strength, range of motion, and coordination, particularly using robot-enabled therapies that feature self-assist modes. This chapter presents a motor task that could be further cultivated into an exercise regime for the neuromuscular development of the TA, thus addressing drop-foot.

5.2 Methods

We developed a simple task for subjects to raise their toes by dorsiflexing their ankles to avoid obstacles. With this simple collision avoidance task, we looked at dorsiflexion capabilities (range of motion and timing) of chronic stroke subjects. This experiment was conducted using Device 2 described in Chapter 2 (Figure 2.7). The foot cradle was used to monitor the position of the subject's foot and provide assistance in certain conditions. The hand cradle was used as the input interface for commanding assistance when it was directed by either the subject or the experimenter. Each device is equipped with an optical quadrature encoder to measure the angular position of the cradle. The data were recorded at 200 Hz on the target computer that processes the real time operation and control. In addition to these sensors that were described in Chapter 2, this experiment also included collection of electromyography (EMG). We recorded surface EMG of the tibialis anterior (TA) using bipolar surface electrodes. The amplifier (Grass Technologies, model CP511) analog bandpass filtered the signal with a 30 – 1000 Hz bandwidth. Data were collected at 1000 Hz and were rectified and filtered in post processing.

5.2.1 Obstacle avoidance task

The general task was for subjects to use ankle rotation to raise and lower the toes, represented as a moving green dot marker on a computer monitor. Using only one foot in the foot cradle, the subjects were instructed to move the marker up and down to avoid touching any of a series of colored obstacles that scrolled horizontally across the screen. The obstacles were all the same width and moved with a constant speed. Each obstacle was previewed for 10 s with a duration width of 0.67 s. The heights of the obstacles and frequency of occurrence varied with the different sections of the experiment. Obstacle heights were distributed randomly about each subject's own active range of motion (ROM) with a standard deviation of 5% of that ROM. Active ROM was determined during the initial assessments and will be further explained in the description of the protocol below. The foot marker remained fixed horizontally on the screen, only moving vertically with the ankle motion; the ankle plantar flexion and dorsiflexion represents a single degree of freedom. Subjects were required to dorsiflex the ankle until the marker was at a position higher than the next obstacle and hold the mark above the obstacle as it moved past. Figure 5.1 presents the visual display that the subjects watched during the experiments. If the marker collided with the obstacle, the experiment proceeded but subjects were made aware of the collision through visual and haptic feedback. While in contact with an obstacle, the green marker turned red and vibrotactile feedback was displayed through the foot cradle and hand cradle (when applicable). Subjects were also provided with a numerical score following the passing of each obstacle. The numerical score was visually displayed on the screen and labeled *Leg Effort Gauge*. This rough score was computed by averaging the absolute value of the raw EMG signal voltage for the window of time during and preceding the obstacle by 50ms. This average was scaled by 150 and rounded to present an integer score. In addition to visual displays of marker (foot) position, the obstacle position (current and preview), and the leg effort score, subjects were also given written textual instructions on the screen guiding them through the experiment.

Unlike the tasks presented in Chapters 3 and 4, in this experiment, the task did not involve underactuated dynamics or other external forces to be manipulated. In fact, the motor provided compensation for major hardware dynamics. The effects of gravity on the foot cradle were removed; however, we did not include dynamic compensation to counter the inertial effects of the cradle. The foot motions were performed at low frequencies and relatively slowly, therefore we do not believe that the inertia of the device was a significant consideration. This task was designed to give subjects practice with unweighted dorsiflexion, where they were responsible for lifting the weight of their own foot as would be expected during locomotion. In certain conditions though, the subjects received lifting assistance in the form of forces applied to the foot from the motor through the cradle. These assisting conditions are described below.

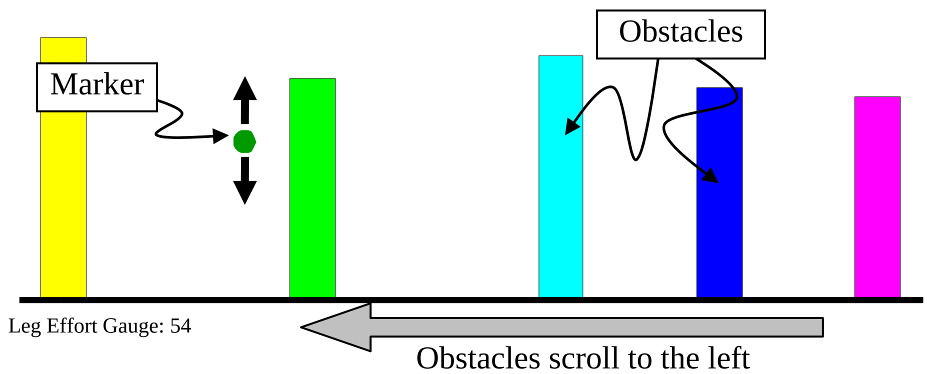

Figure 5.1 Visual display of collision avoidance task. Subjects moved the position of the circular marker up and down (depicted as black arrows) by raising and lowering their toe with ankle motion. The obstacles scrolled left across the screen and the lateral position of the marker did not change. Subjects were instructed to raise the marker higher than the obstacle height as to avoid collision when the obstacle moved by. The "Leg Effort Gauge" provided an abstract, numerical score for each obstacle based on EMG activity.

5.2.2 Operating conditions and controllers

Subjects performed the collision avoidance task in one of four possible conditions: *No Assist*, *Self-Assist*, *Experimenter-Assist*, and *Computer-Assist*. When operating in the *No Assist* condition, subjects used only a single foot in the foot cradle to raise the marker on the screen, with no additional external forces applied through the cradle. The *Self-Assist* and *Experimenter-Assist* conditions involved using the hand cradle teleoperating the foot cradle. The *Computer-Assist* used an automatic controller employing position feedback from the foot cradle encoder and information about future obstacles to command assistance to the user.

We implemented the teleoperator for the *Self-* and *Experimenter-Assist* conditions using standard position-position proportional-derivative control architecture. The block diagram presented in Figure 5.2 illustrates how the basic teleoperator controller combines with the hardware compensation to determine the commanded motor torque. For the position-position control, the position of each device was measured and compared. A virtual spring-damper pair coupled the two devices to mutually track the position of one another. As the motions differ, the motors apply torques in an attempt to rectify the discrepancy. Therefore, it was possible to provide force assistance to the foot by moving the hand device. Likewise, it was possible to monitor the motion of the foot by lightly touching the hand cradle and feeling its movement. The same controller was used for the *Self-* and *Experimenter-Assist* conditions with the hand cradle operated by the subject or

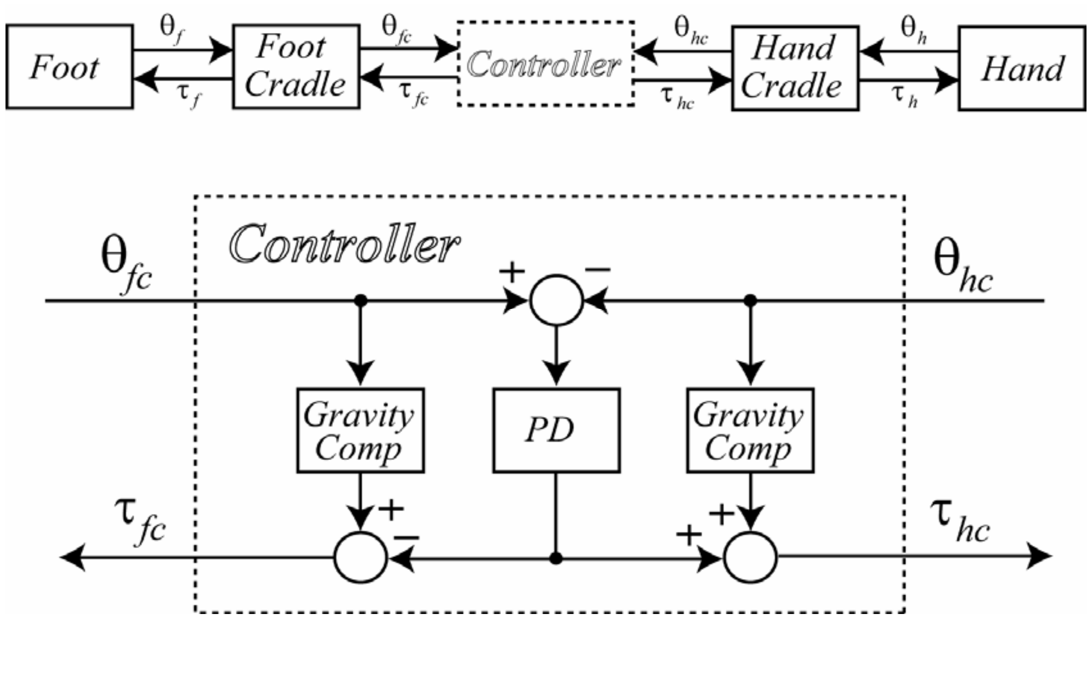

Figure 5.2 Block diagram of teleoperator control architecture. The general structure of the teleoperator connects the motions of the hand θ_h through the hand cradle to the motions of the foot θ_f through the foot cradle. The controller determines the torques commanded to the motors of the hand and foot cradles (τ_{hc} and τ_{fc}, respectively) which include effort from the teleoperator PD controller and the gravity for the devices.

the experimenter, respectively. When applicable, subjects operated the hand cradle with the less affected upper limb to assist the impaired contralateral lower limb. For all subjects, the same human experimenter operated the hand cradle during the *Experimenter-Assist*.

The automatic controller designed for the *Computer-Assist* condition also used proportional-derivative control as its foundation. Where the teleoperator controller used the error signal between the two interfaces for control, this automatic controller used the error between the foot cradle position and a time-varying reference position. The equation for assistive torque is given by

$$\tau = \begin{cases} k(\theta - \theta_r) + b\dot{\theta}, & \text{for } \theta \leq \theta_r \\ 0, & \text{for } \theta > \theta_r \end{cases}$$

where θ is the foot position, θ_r is the reference position, and k and b are constants of the PD controller. The controller stiffness k was set at 5 Nm/rad and the controller damping b was set at 0.2 Nm·s/rad. Note that the controller only provides assistive torque if the foot position is less than the reference position, otherwise the torque is zero. In other words, the controller only helps if the foot is not high enough to clear the obstacle without assistance. The predetermined path of the reference signal θ_r was based on the distance from the marker to the subsequent obstacle and the height of that obstacle. Figure 5.3 depicts a physical representation of how this controller would function. The dashed line in the figure illustrates the desired minimum height for the foot marker. The initial ramp for θ_r begins 0.67 seconds prior to the obstacle while the final ramp concludes 0.33 seconds following the obstacle. The maximum reference height was set at 0.25 radians greater than the obstacle height. The virtual spring-damper pair applies assistive forces to the foot, pulling the marker closer to the reference position.

The motor driving the foot cradle is limited in the magnitude and duration of torque that can be applied. Therefore, the torque administered in any of the three conditions that offer assistance was restricted to quantities possible for the motor to generate and those that would reduce the risk of overheating and damaging the motor. Despite the software saturations applied to the commanded torque, two motors did overheat and consequently burned out during the course of data collection. Unfortunately, this issue was only diagnosed until after completing the experiment. Therefore, some subjects were excluded during analysis of differences by condition.

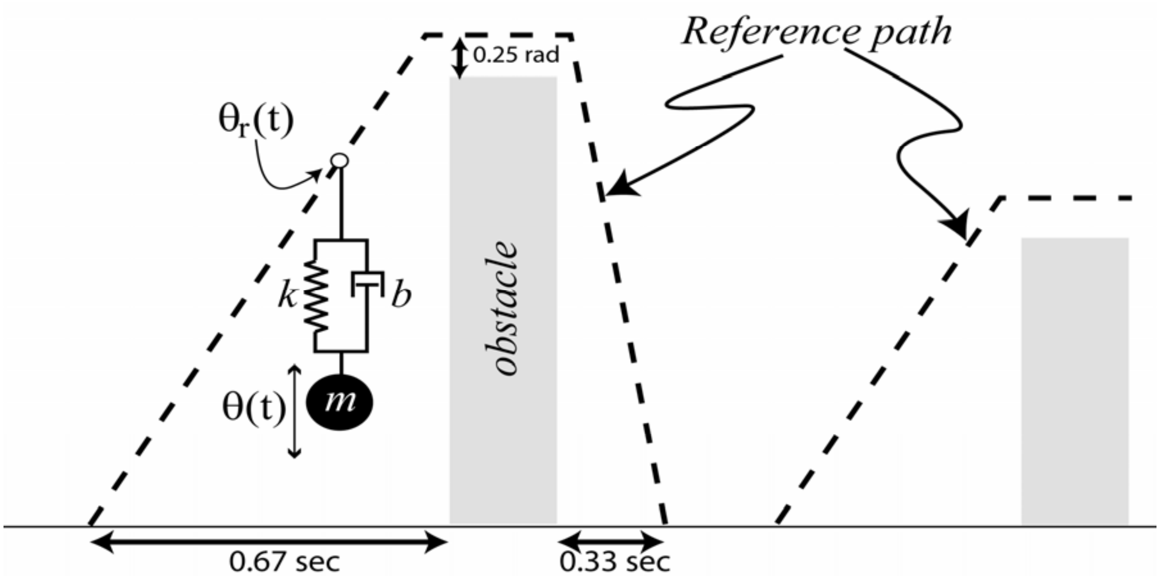

Figure 5.3 Physical representation of *Computer-Assist* controller. The automatic controller applies forces to pull the foot marker position θ to the reference position θr with a virtual spring-damper pair. The reference position is determined by the distance to and height of the upcoming obstacle. The predetermined reference position for the obstacles illustrated is depicted by the dashed reference path.

5.2.3 Participants

A total of 18 subjects (11 men, 7 women) volunteered to participate in this study. All subjects had a history of cerebral vascular accident with lasting hemiparetic effects. Participants ranged in age from 27 to 67 with a mean age of 48 years (sd 9 years). Table 5.1 summarizes the demographic information about the subject population. Subjects were recruited through the voluntary stroke registry maintained by the Rehabilitation Institute of Chicago. Participants reported having normal or corrected to normal vision. Each provided written informed consent in accordance with the human subject protection policies of Northwestern University and the Rehabilitation Institute of Chicago. Subjects were monetarily compensated for their time.

Although 18 subjects volunteered for participation in the study, one subject was unable to complete the full experiment and the data files for another subject were corrupted. One wheelchair bound subject was unable to achieve the minimum muscle control required to complete the task, even with assistance. She was therefore excluded from all data analysis. Hardware equipment malfunctions, specifically the motor, interrupted consistent administration of the experiment conditions for three subjects, making a portion of the collected data invalid. Finally there is no EMG data for one subject due to an amplifier malfunction; therefore, those data were not included in the analysis concerning EMG performance metrics. The details of the performance metrics will be discussed later in this chapter. A total of 12 subjects were included in analysis of performance by condition during training (Section 5.3) and a total of 15 subjects were included in analysis of performance during the initial and final assessments of the experiment (Section 5.4).

Table 5.1 Summary of subject demographics

	Mean ± Std Dev	Range
Age (years)	48.4 ± 9.4	27 – 67
Post CVA (years)	6.7 ± 5.0	2 – 19
	Abs quantity	% of total
Total	18	100 %
Male	11	61 %
Left side affected	12	67 %
AFO use	8	44 %
Assistive aid*	8	44 %

* cane or quad cane for 7 subjects, wheelchair for 1

5.2.4 Protocol

The experiment was conducted in a single session lasting approximately 1.5 hours. It was sectioned into three parts, each primarily centered on the collision avoidance task described above. The sections included an initial assessment, a training stretch, and a final assessment. Subjects were given water and encouraged to rest at frequent intervals throughout the experiment to reduce fatigue.

Initial Assessment

The initial assessment portion began with subjects first using their less impaired foot in the foot cradle to command the marker position. With no torque assistance, subjects completed a block of collision avoidance that included 21 evenly spaced obstacles. The obstacles were presented at a constant frequency in which there was a period of 4 s between the initialization of each obstacle. This first block served to allow the subjects to become familiar with the hardware, the visuals, and the task. For most subjects, the collision avoidance task using the less impaired lower limb was a relatively trivial task, the same as would be expected for neurologically intact individuals.

Following the familiarization block, subjects were then asked to complete a series of ten toe raises with the impaired lower limb in the foot cradle. During the toe raises, subjects were not presented with any obstacles or goal heights for dorsiflexion; they were asked to raise the impaired toe as high as possible and hold it for a few seconds, all with no outside assistance. Subjects were told when to lift, but did not need to maintain the height for any appreciable time, and could rest between lifts. This exercise served to identify the initial active range of motion (ROM) for each subject. The ROM was defined as the average of the ten toe lifts and was used as the baseline to set obstacle heights in subsequent collision avoidance blocks.

The final portion of the initial assessment portion included completion of three blocks of unassisted collision avoidance with 21 obstacles each. As with the familiarization block, the obstacles were consistently spaced with 4 s between the leading edges of two adjacent obstacles. These three blocks in the *No-Assist* condition served to guarantee subject comfort with the equipment and task using the impaired limb. In addition, they can be used to establish baseline performance before training in the task.

Training

During the training portion of the experiment, subjects were asked to complete sixteen blocks of collision avoidance. Each block consisted of 21 obstacles presented in three groups of seven obstacles at each of three different frequencies. The periods for the obstacle spacing were 2.5 s, 4 s, and 6 s (f_1, f_2, and f_3, respectively). The sixteen blocks were partitioned into four repetitions each of the four assistance conditions (*No Assist*, *Self-Assist*, *Experimenter-Assist*, and *Computer-Assist*). Any single block only included one assistance condition. The presentation of the sixteen training blocks followed a Latin

Square experiment design. Table 5.2 presents the generalized ordering of this type of design. The use of the Latin Square design was intended to guarantee even distribution of treatment type and reduce ordering effects. There were no particular nuisance variables assigned to the rows and columns of the square beyond presentation order. Subjects were required to take a break of at least two minutes upon the completion of each row in the Square although other breaks were permitted as needed.

Table 5.2 Latin Square experiment design

A	B	C	D	Rest	Conditions randomly assigned to treatments A, B, C, and D
C	D	A	B	Rest	
D	C	B	A	Rest	
B	A	D	C	Rest	

Final Assessment

The final assessment section was very similar to the initial assessment section, mirroring many of the same elements. Subjects first completed one block of unassisted collision avoidance with 21 evenly spaced obstacles using the impaired lower limb. Subjects were then asked to repeat the series of ten self-paced toe raises with the impaired limb. Finally, subjects once again used the less impaired foot to complete a block of unassisted, evenly spaced obstacles.

5.3 Training analysis and results by condition

During the training portion of the experiment, subjects were tasked with collision avoidance for a series of obstacles presented continuously. We developed a number of metrics to compare the four assistance conditions (*No Assist*, *Self-Assist*, *Experimenter-Assist*, and *Computer-Assist*). These metrics are varied to capture differences in task-oriented performance, as well as differences in control strategies and muscle recruitment capabilities of the subjects on a per condition basis. We examined subject success in the task, using collisions observed, and subject success in exercising the impaired leg with EMG activity and assistance administered.

5.3.1 Performance metrics and data analysis

The first two metrics related to the abilities of subjects to complete the presented task, i.e., avoid hitting the on-screen obstacles with the marker controlled by ankle motion. For each trial, or series of 21 obstacles, we determined the number of obstacles that were hit one or more times by the marker (*Col #*). This count metric gives an integer value out of 21 possible. We used a second metric based on the collisions during each trial. We computed the total amount of time that the marker was in contact with any obstacle during a trial (*Col Time*).

The metrics related to task performance, *Col #* and *Col Time*, are determined on a trial by trial basis. This means that each metric delivers a single score per trial, totaling sixteen observations per subject (4 conditions by 4 repetitions). For the statistical analysis of *Col #*, we used a generalized linear model. This model used subject and condition (*No Assist*, *Self-Assist*, *Experimenter-Assist*, or *Computer-Assist*) as input factors and *Col #*, a non-normally distributed integer count out of 21 possible, as the outcome. For analysis of *Col Time*, we used a linear mixed model with subject as a random variable and condition as a fixed variable. For both metrics, the observation order was determined to be insignificant. Data from twelve subjects were included in these analyses. Three subjects were excluded due to problems with the DC motor during data collection which rendered the consistency of the administered assistance unreliable. Corrections for pairwise comparisons were made with the Bonferroni adjustment.

The analysis also included assessing the capabilities of the impaired leg in performing the task. The hardware we used is not capable of directly measuring the force contribution of the subject's muscles in lifting the foot; however, we could determine how much assistive force was provided by the foot cradle to help raise the foot. In the *Experimenter-Assist* and the *Computer-Assist* conditions, the assistive force is a measure of how much help the subject needs to clear the obstacle, or get closer to clearing the obstacle. In the *Self-Assist* condition, this force is the amount the subject perceives he needs to clear the obstacle, since it is unregulated by the experiment. The assistive force is necessarily always zero for the *No Assist* condition, since no assistive effort was available. As a quantitative score for this metric, we determined the average assistive torque provided in a set window of time around each obstacle (τ-*assist*). We used a one second window that preceded the leading edge of each obstacle by 0.75 s and ended 0.25 s into the obstacle. The window chosen captures the time when subjects are actively lifting the marker and the critical time in maintaining the clearance height.

We also looked at the EMG activation in the TA to gauge dorsiflexor muscle recruitment. The raw EMG signal recorded at 1000 Hz during the experiment was rectified and filtered during data post processing. The voltage for each subject was normalized by EMG voltages measured during a series of maximum voluntary contractions performed after the experiment completion. The normalized data was then filtered with a 60 Hz notch filter, to remove noise at the utility frequency, and with a 50-500 Hz 4^{th} order Butterworth bandpass filter. Next, we took the root mean square (RMS) of the signal using a 100 ms window. Finally, the data was down-sampled by a factor of 5, to give a processed EMG signal at 200 Hz. For each obstacle presented, in all trials of all conditions, we determined the average EMG (*EMG*) using the same one second averaging window described for the τ-*assist* performance metric.

The performance metrics τ-*assist* and *EMG* were calculated at the obstacle level, meaning that each trial yielded 21 observations, one for each obstacle. Analyses were performed using a repeated measures linear mixed model. The fixed effects were condition type (*No Assist* was not included in τ-*assist*) and obstacle frequency (three levels: slow, medium, and fast). Subject was considered a random effect. The obstacle height was included as a covariate in the analysis (using change from mean for each subject). Observation order and trial number were not statistically significant factors when included in the model and were therefore eliminated for the final analysis. The same twelve subjects that were used for the trial-level analysis were used in analyzing τ-*assist*. Due to corrupted EMG data, one additional subject was excluded in analyzing *EMG* leaving eleven subjects. For both metrics, Bonferroni's adjustment was used to correct for pairwise comparisons.

5.3.2 Results

As one might expect, subjects demonstrated better performance at the obstacle avoidance task when they were provided with some type of assistance. Histograms of the distribution of *Col #*, grouped by condition are presented in Figure 5.4A. The quantity of collisions observed *Col #* was the largest for the *No-Assist* condition, averaging 3.08 collisions per trial. The *Self-Assist* condition trended towards fewer collisions with an average of 1.7; however this difference was not statistically significant. Both the *Experimenter-Assist* and the *Computer-Assist* conditions demonstrated significantly improved performance over the *No-Assist* condition (p = 0.08 and p = 0.09, respectively).

There was no significant difference between any of the three assist conditions, including *Self-Assist*. Figure 5.4B presents a comparison of the means and 95% Wald confidence intervals for the four training conditions. Averages were taken across all subjects and all trials by condition.

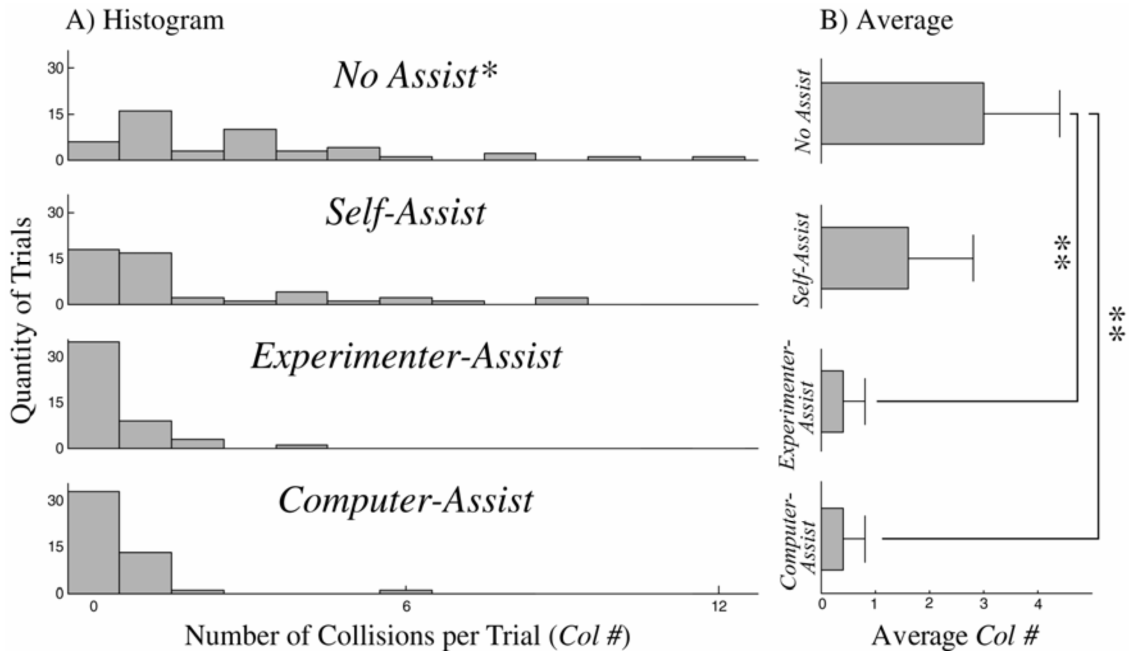

Figure 5.4 Summary of *Col #* by condition. The histogram of *Col #* distribution (A) shows that most trials with assistance incur no or one collision, while the *No Assist* condition is much more varied in the *Col #*. Note, one observation in the *No Assist* condition was excluded from the histogram at *Col #* 20. The figure on the right (B) compares the mean collisions per trial for each condition. *Experimenter-Assist* and *Computer-Assist* are demonstrate significantly fewer collisions than *No Assist* (denoted with **, $p < 0.01$).

The trends for the amount of time per trial that a subject spent in contact with any obstacle, *Col Time*, followed essentially the same patterns as *Col #*. During the *No-Assist* condition subjects were in collisions for an average of 1.23 seconds per trial. This represents a significantly longer duration than any of the other three conditions: *Self-Assist* ($p = 0.04$), *Experimenter-Assist* ($p < 0.01$), or *Computer-Assist* ($p < 0.01$). Figure 5.5 presents a histogram of duration distribution (A) and the mean time in collision per trial (B), both grouped by condition.

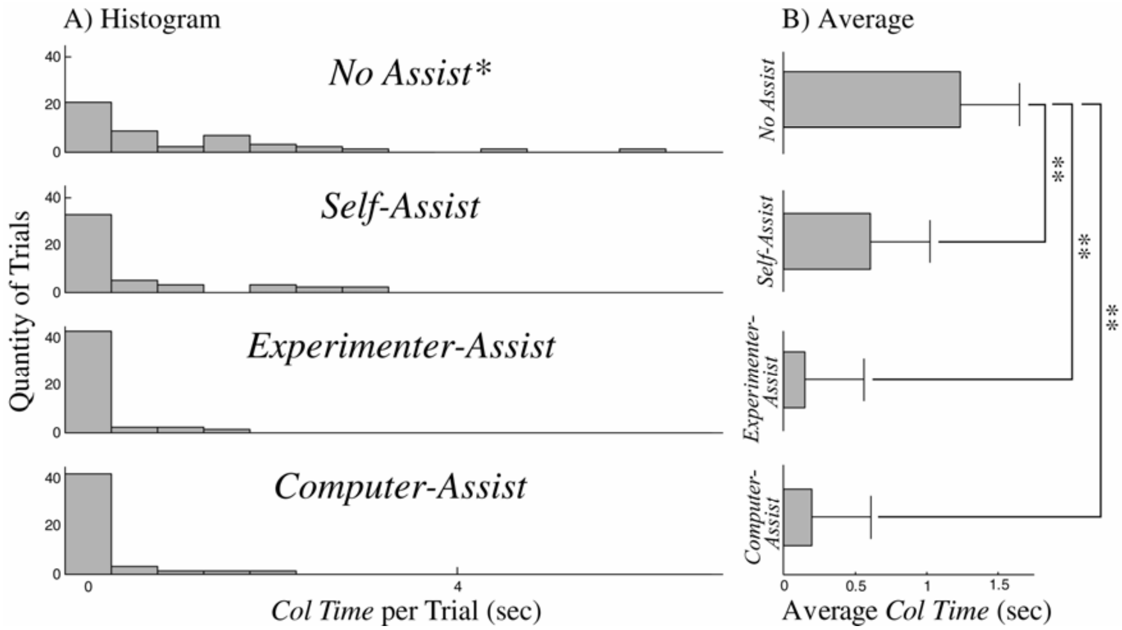

Figure 5.5 Summary of *Col Time* by condition. For the assisting conditions, the majority of trials had *Col Time* less than 0.5 s, as is evident by the histogram of the distribution (A). The mean *Col Time* was significantly longer for No Assist than any of the other three conditions (denoted by **, $p < 0.01$).

The metrics designed to capture the capabilities of the impaired leg (τ*-assist* and *EMG*) demonstrated performance distinctions between the conditions. The linear model included many factors that were determined to be significant in addition to the assisting condition. The significant fixed effects influencing the assistive torque (τ*-assist*) as an outcome are the intercept, the assistance condition (*Self-Assist*, *Experimenter-Assist,* and *Computer-Assist*), the obstacle frequency, the obstacle height (change in height from the norm for each subject, delta), and the interaction effects between condition and delta height. Each of these was significant with $p < 0.001$. The muscle activation, *EMG*, was influenced by the same factors with the addition of *No Assist* in the conditions, also with significance for each of $p < 0.001$.

One might expect that better performance might be indicated by less assistive torque required in dorsiflexion to raise the marker above the scrolling obstacles. The pairwise comparisons of τ-*assist* reveal that subjects received the least amount of assistive torque in the *Computer-Assist* condition as compared to either of the other two assistive conditions ($p < 0.01$). In addition, *Self-Assist* administered less assistive torque than *Experimenter-Assist* ($p = 0.001$). Regardless of condition, subjects also required more assistance when the obstacles came at a more frequent pace. The highest frequency (f_1) demanded significantly more assistive torque than either of the other two ($p < 0.01$), which were statistically identical to each other ($p = 1.0$). Figure 5.6 presents the mean τ-*assist* for all subjects separated by condition and by frequency.

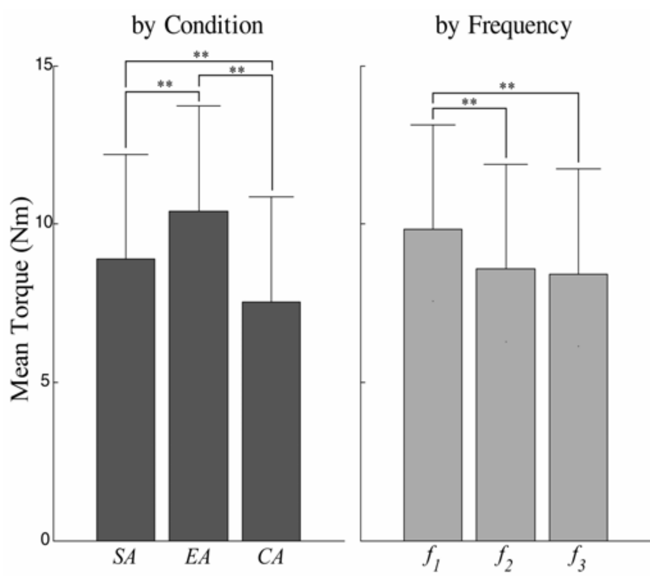

Figure 5.6 Estimated marginal means of τ-*assist*. The figure on the left presents the means by condition: *Self-Assist* (*SA*), *Experimenter-Assist* (*EA*), and *Computer-Assist* (*CA*). The figure on the right presents the means by frequency where f_1, f_2, and f_3 represent the spacing periods 2.5 s, 4 s, and 6 s, respectively. The marginal means were estimated using SPSS using the nominal obstacle height for each subject. The error bars represent the 95% confidence interval for between subject variations and the ** denotes significance with $p < 0.01$.

Muscle activation of the tibialis anterior was captured in the *EMG* performance metric. Higher *EMG* values indicate more activation which suggests better exercise for the impaired limb. Figure 5.7 presents the mean *EMG* averaged across all subjects and separated by condition and by obstacle frequency. As was observed with the metric for required assistive torque τ-*assist*, performance degraded when there was less time between obstacles; that is, the highest frequency (f_1) demonstrated significantly lower *EMG* than the other two frequencies ($p < 0.001$). When comparing the assistive conditions, the general trend of *EMG* was in order from highest values to lowest: *No-Assist*, *Self-Assist*, *Computer-Assist*, and then *Experimenter-Assist*. However, pairwise significance was only found for *No Assist* compared to *Computer-Assist* ($p = 0.001$), for *No Assist* compared to *Experimenter-Assist* ($p < 0.001$), and for *Self-Assist* compared to *Experimenter-Assist* ($p = 0.002$).

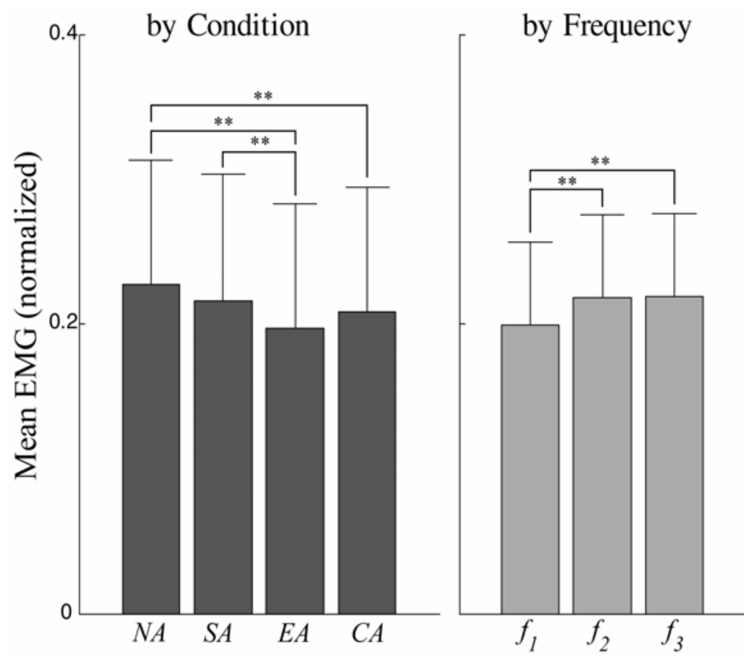

Figure 5.7 Estimated marginal means of *EMG*. The means are taken across all subjects and by condition (left) or frequency (right) with estimates for nominal obstacle height (delta = 0). The error bars represent the 95% confidence interval for between subject variations and the ** denotes pairwise significance with $p < 0.01$. Overall, the *No Assist* condition demonstrated the highest EMG activity, but it was not significantly greater than the *Self-Assist* condition.

5.3.3 Discussion

Subjects are able to integrate external assistance with personal effort to increase dorsiflexion of the paretic ankle following a CVA. Given any of the three means of assistive force application, performance at completing the obstacle avoidance task improved; subjects had fewer obstacle collisions and reduced time in collision per trial. Of the four operating conditions, subjects trended towards the highest EMG activity during the *No Assist* condition, but not significantly more than in *Self-Assist*. When subjects self directed the provided assistance, they maintained their level of effort while improving their performance at the task. The *Computer-Assist* condition proved the best in terms of task performance (*Col #* and *Col Time*) while also requiring the least amount of assistive torque to enable the effect. The automatic controller appeared to be the most efficient at task success by consistently raising the impaired foot without expending unnecessarily high torques. This success, however, was at the expense of subject effort as gauged by the *EMG* metric.

Each of the two metrics used to assess task success, *Col #* and *Col Time*, were limited in certain respects. *Col #* simply gives a count of the number of obstacles that were touched in any way and for any amount of time by the marker. If a subject clipped the corner of the obstacle before reaching a clearance height, the collision number would increment the same as if the collision had been more egregious. In the case of a minor collision, the subject may still have achieved good dorsiflexion and for all intents and purposes succeeded at the task. Assessing time in collision captures the distinction between these two types of collisions. Casual observation of the subjects during data collection reveals a weakness of the *Col Time* metric. It was noted that occasionally following a collision subjects would discontinue efforts to surmount that particular obstacle. The subject may have viewed any collision as a failure, thus giving up on the entire obstacle and focusing on the next. This defeatist strategy would inflate the *Col Time* metric in an unpredictable manner, based in part on desire or motivation rather than capability. Despite the potential shortcomings of the two metrics, the combined results present a solid measure of task performance.

The obstacle avoidance task was designed to be relatively difficult for subjects. The heights of the obstacles were set such that approximately 50% would be higher than the subject’s voluntary range of motion as determined during the initial assessment toe raises. Therefore, subjects needed assistance, provided through the motorized foot cradle, to surmount many of the obstacles. Unfortunately, the motor actuating the foot cradle was not always able to provide sufficient torque assistance for the subject to succeed. Essentially, the device was not strong enough in all circumstances. In some instances the foot cradle, despite the best efforts of the assistance controlling mechanism, may have been unable to lift the foot to the appropriate height, especially for a few subjects that had very stiff ankles from disuse. In addition, data collection was interrupted for at least two participants due to over exertion and consequential destruction of the driving motor. The deficiencies of the equipment may have caused additional unmodeled effects on the task performance.

We predicted that when subjects were responsible for administering the assistance provided to their impaired limb, they would adopt an "as-needed" strategy. That is, we believed that subjects would only apply assistance when they needed it and only as much as they needed, thus resulting in less overall aid than in other assisting conditions. However, this effect was not observed with the metrics that we employed. The average assistive torque applied during *Self-Assist* was statistically significantly larger than during the *Computer-Assist*. The larger τ*-assist* values observed may be contributed to different strategies employed by the human versus the computer controller. The automatic computer controller used a proportional-derivative control to gradually increase torque based on marker-to-clearance error as the obstacle approached. Although each subject used the teleoperator with slightly different methods, one strategy was very common. For each obstacle, subjects would initially administer no aid but try to independently complete the task. Once the subject accepted that the obstacle was insurmountable, he would aggressively slam on the assistance, operating more like a bang-bang controller as compared to the proportional controller implemented for *Computer-Assist*.

An important distinction between the three methods of directing the assistance provided is the type of information available to the subjects during the trials. In all instances, subject were provided visual feedback of the marker and obstacles, proprioceptive feedback of ankle position, haptic feedback of the interaction forces between the foot and the foot cradle, and a rough gauge of their muscle activation. The *Leg Effort Gauge* was in place to provide subjects with some biofeedback, albeit crude, of their contribution to the task. Subjects were not required to maintain a minimum effort level but were encouraged to monitor the score and focus on using their own leg, as opposed to the assistance, to dorsiflex the foot. In the *Self-Assist* condition, subjects received additional feedback, haptic and proprioceptive, from their hand on the hand cradle. This information may have helped subjects to monitor the foot motion. More importantly, this feedback, along with efference copy of the hand motor plan, correlated exactly to the applied assistive effort (magnitude and timing). Subjects were given an additional information path to monitor the effort of the impaired limb, allowing segregation of ankle effort with torque assistance to raise the foot. Although subjects may have received verbal communications from the experimenter during *Experimenter-Assist*, in general only the haptic feedback at the impaired foot provided information about assistive effort during both the *Computer-Assist* and *Experimenter-Assist* conditions. The torque loads provided by the motor were large enough to push the range of motion of the ankle, but they were not strong enough to completely overpower it. This limited the intensity and discernable resolution of the haptic feedback at the foot.

Essentially, during the *Computer-Assist* and *Experimenter-Assist* conditions, subjects were unaware of how much aid they received. In fact, many times subjects were unable to recognize that any assistive torque was offered. The inability to quantitatively or qualitatively gauge external assistance might partially explain the observed decrease in *EMG* during the *Computer-Assist* and *Experimenter-Assist* conditions. If subjects could not discern how much of the lifting success could be attributed to the external aid and how much could be attributed to their efforts, they may not have been inspired to push quite as hard. Essentially, the external aid made the task easier without subjects fully understanding why; they were able to succeed without recognizing that it was not completely attributable to personal effort.

5.4 Initial and final assessment analysis and results

During the initial and final evaluations, subjects completed a series of ten toe raises to determine their active dorsiflexion range of motion (ROM). During these toe raises, subjects initiated the movement and raised the marker as high as possible without cuing from any visual goal. Sample traces of the foot cradle angular position during the initial and final toe raises for a single subject are presented in Figure 5.8. We developed performance metrics based on the active ROM before and after the training. These metrics allow us to discuss the effects of limited practice with assistance on ankle abilities and activation.

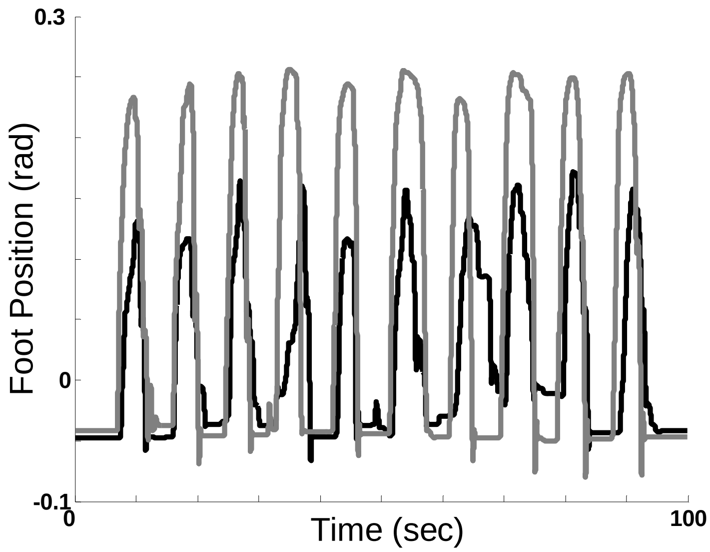

Figure 5.8 Sample traces of toe raises. The foot cradle position during the initial (black) and final (grey) range of motion assessment can be compared for changes in active dorsiflexion abilities before and after training. Presented are the traces for a single sample subject.

5.4.1 Performance metrics

We developed three performance metrics to compare the active ROM at the beginning and end of the experiment. Of the three metrics, two are based on the motion of the foot cradle and the height achieved during the lifts. The third metric is based on the EMG activity of the TA. The two position-based metrics are 1) the maximum absolute position reached (peak) and 2) the displacement between the resting position of the foot and the peak position (change) (see Figure 5.9). Although ideally two metrics would be identical, in practice, the resting position of the foot was not necessarily constant between the initial evaluation and the final evaluation. There are various reasons for this discrepancy. The stiffness of the ankle may have changed slightly due to use (warming up or fatigue) changing the equilibrium position; subjects may have shifted their posture slightly through the experiment; or subjects may simply be holding the foot at a different resting angle. The third metric captured the EMG activity of the TA during the lift. The normalized, filtered and rectified EMG signal during the toe raises was further reduced by taking the maximum of the rolling average for each lift. The averaging window was set at 0.5 seconds where each lift generally lasted 3-5 seconds.

The three metrics each provided a single numerical score for each of the ten lifts performed in the active ROM assessment. These ten trials were averaged for each subject. Using the means by subject, performance was analyzed using two-tailed paired t-tests with $\alpha = 0.05$. Data from all 15 eligible subjects was included in the analysis with the position based metrics. Analysis of the EMG metric included data from 14 subjects; recall that there was no available EMG data for one subject.

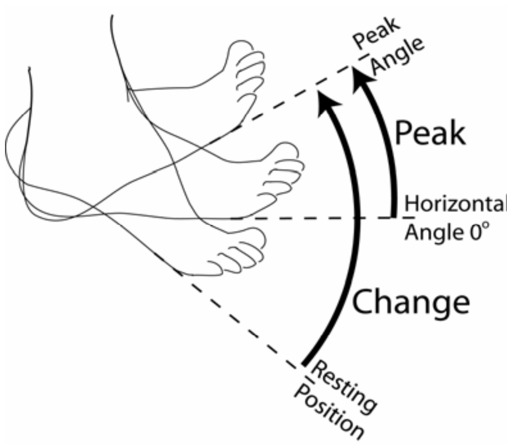

Figure 5.9 Position based metrics. The active range of motion is determined by the maximum absolute angle (peak) reached by the foot and by the difference between the resting position and the maximum (change). We used both metrics because throughout the resting position in the initial and final assessment was not always the same. By using both metrics, we are guaranteed to capture differences in the dorsiflexion abilities of the subjects before and after the training, no matter what the cause of the altered resting position.

5.4.2 Results

On average, subjects demonstrated improved active ROM following the obstacle avoidance training when compared to pre-training. Looking at performance metrics, not all subjects improved, though the vast majority either remained unchanged or improved. Table 5.3 summarizes the distributions in subject performance changes. As a population, subjects demonstrated significantly larger absolute peak dorsiflexion angles, averaged per subject, after completing an ankle exercise task. The means for all subjects determined during the initial assessment toe raises and during the final assessment toe raises are illustrated in Figure 5.10A. Also presented in this figure is the mean difference between the initial and final, calculated per subject then averaged. The final peak represents an average of 26.34% improvement over the initial (two-tailed paired t-test, p = 0.0291). The data for all 15 subjects was included in the analysis; however, one subject improved greatly from virtually no initial motion. Therefore, to prevent abnormally skewed values, the percent increase for this subject was not included in the average percent improvement.

Similar to the absolute peak angles, the displacement between resting angle to peak angle (change) was also improved after training as compared to before training. Figure 5.10B presents the means across all subjects of the initial, final, and difference in the ankle change performance metric. The final assessment using this measure of active range of motion demonstrates an 18.65% increase over the initial assessment (two-tailed paired t-test, p = 0.0228). Once again the outlier was removed in calculating this percentage.

Table 5.3 Summary of initial-final performance difference

Active ROM	Number of Subjects			Group Difference		Sig
	Increased	Unchanged	Decreased	Absolute	Percent*	P
Peak Angle	10	3	2	1.97	26.3%	0.0291
Angle Change	7	7	1	2.13	18.7%	0.0228
Peak EMG	7	5	2	0.08	29.2%	0.0239

* One outlier removed

In addition to the two position-based metrics, subjects also demonstrated more EMG activity in the TA in the final assessment toe raises than in the initial assessment toe raises (Figure 5.10C). Only 14 of the 15 subjects generated valid EMG data for analysis. In the final assessment, those 14 subjects increased the peak EMG generated by

29.18% over the initial assessment (two-tailed paired t-test, p = 0.0239). Again, the outlier was removed in calculating this percentage.

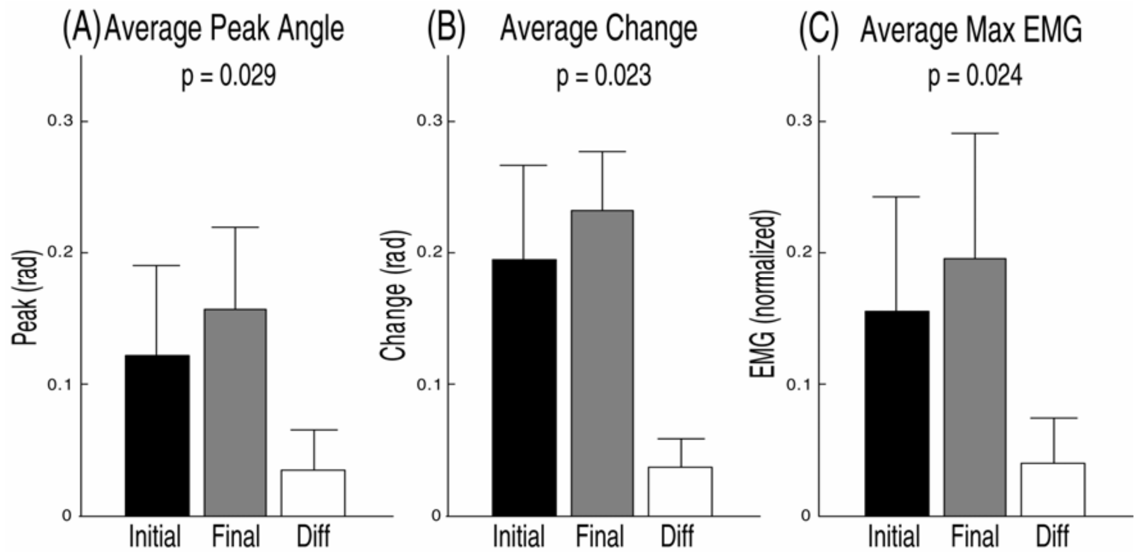

Figure 5.10 Quantitative comparisons of initial and final assessment performance metrics. The results from the peak angle (A), displacement change from rest (B), and peak EMG (C) are presented in the graphs. The means across all subjects of each performance metric show that on the whole subjects improved from the initial assessment (black) to the final assessment (grey). The error bars represent one standard deviation of between-subject error. The mean difference per subject, the statistical significance, and the percent change referenced from initial performance are also all presented in the graphs (two-tailed paired t-test). The error bars on the difference represent the 95% confidence interval of the effect size. One outlier was omitted in all calculations of percent change.

5.4.3 Discussion

Subjects demonstrated improvements in active dorsiflexion capabilities after a short, single session training period. The ankle range of motion, as measured with the peak angle and displacement change metrics, was significantly improved in the final assessment when compared to the initial assessment. Subjects also increased the muscle activation in the TA, as demonstrated by increased EMG. The analysis of the initial and final assessments can be used to comment on the benefits of the obstacle avoidance task as an exercise for improving dorsiflexion. From this analysis, we cannot determine which assistance condition is the most effective, or even if assistance is necessary during training exercises. However, we were able to determine that during a short duration of TA exercising, subjects were able to increase their active range of motion and dorsiflexion capabilities.

One important distinction between the initial and final toe raises and the training task is the existence of a specific goal. During training, subjects were charged with an objective: raise the marker to surmount an obstacle with a particular height. Goal oriented tasks present different motivations and strategies than best effort or capability-type tasks. During the toe raises, subjects were asked to perform the best they could at their own pace. There was no time pressure or obvious success/failure gauge, such as a collision. Therefore, subjects were able to concentrate their efforts on lifting. The initial and final evaluations present a better measure of dorsiflexion capabilities than the within training performance metrics, which may be influenced by other factors such as obstacle height and frequency as well as assisting condition.

5.5 Significance

The ability to dorsiflex the ankle, raising the toe, is an important component to safe locomotion. We developed a functionally motivated task to exercise the frequently neglected dorsiflexion muscles. After a short duration of practicing the task, subjects were able to improve the range of motion of the ankle and the activation of the tibialis anterior. In fact, two subjects who initially reported having no voluntary dorsiflexion capabilities demonstrated significant motion and consistent lifting abilities by the final assessment toe raises. Most subjects were able to complete the obstacle avoidance task with assistance and were able to independently demonstrate dorsiflexion in toe raises.

In addition to demonstrating the potential of self-assist in designing exercises for an impaired population, this functional task was well received by the subjects. Subjects expressed excitement at completing an exercise that was both engaging and useful. Most subjects reported having little or no experience exercising the TA and were disappointed to find that standard rehabilitation gym equipment did not address the muscle. The comments collected from the subjects regarding the equipment and the task include the following:

"I wish we had this at the gym"
"Really challenging"
"Great exercise"
"Don't really work that muscle much and it really needs it"
"If you can make it a little tougher – resist more – I'll take one for home!"
"I could sit and do this while watching TV"

Unlike assistance provided by an external agent, such as an experimenter or a computer, self-assist can appear somewhat unregulated. Thus, one of the concerns of self-assist is that the assistance itself might become a crutch for the subject, where the subject depends on the assistance to complete the task without extending effort on the part of the impaired limb. The results from the analysis during the training portion of this experiment indicate that subjects were able to improve their performance in the collision avoidance task without using self-directed assistance as a crutch. If subjects depended fully on the assistance, we would expect to see the *EMG* decrease dramatically; however, subjects maintained high levels of muscle activation during the *Self-Assist* condition. Self-assist allows subjects to augment their own capabilities while monitoring the amount of assistance applied. Self-assist might present a compromise between offering subjects a means for improving their performance at a task, thus experiencing larger ranges of motion and the kinematics of correct completion, and not diminishing their efforts by taking over.

CHAPTER VI

Discussion

In this dissertation we have presented a model for lower limb rehabilitation using upper-limb self-assist. In a generalized structure for self-assist, any impaired limb may be assisted through a patient-directed telerobot using an unimpaired limb. With support from the literature, we have theorized mechanisms by which we suppose generalized self-assisted rehabilitation will offer benefits to patients. These include increased patient involvement and active learning [22], increased access to informative feedback through additional afferent sensory channels [63], and improved coordination of effort and assistance through efference copy [57]. Bimanual self-assist has demonstrated some clinical success [31, 34], but is not yet universally accepted as a beneficial practice. Generalized self-assist, on the other hand, remains almost universally untested and primed for debate. Through the work presented in this dissertation, we present a scientific foundation for lower limb rehabilitation through generalized self-assist. In this chapter, we further discuss and summarize our results, addressing the aims presented in Chapter 1.

6.1 Multi-limb motor control can offer performance benefits

Aim 1

Demonstrate that the increased cognitive load from multi-limb control does not outweigh the advantages gained from coordinated multi-limb control in dynamic object manipulation.

Before delving into the complications of generalized self-assist for lower limb rehabilitation, we first addressed the tradeoff that exists between motor control benefits and the disadvantages of increased cognitive load, both of which may accompany multi-limb control. We examined this tradeoff through performance in object manipulation tasks involving underactuated dynamics. These tasks provided appropriate cognitive and motor challenges because extrinsic state can be very difficult to control even for neurologically intact individuals. Using two separate experiments, presented in Chapter 3, we demonstrated that neurologically intact subjects can use the combined input from their hands and their feet to achieve better performance than their feet alone. For the dynamic tasks that we presented, the increased cognitive demand associated with multi-limb motion, especially of non-homologous muscle groups, does not appear to outweigh the lower limb befits reaped from the coordinated control of two effectors.

In both experiments, we were unable to show improvements in upper limb performance when the lower limbs were involved. In fact, the results from the second experiment demonstrated a degradation of hand performance, in terms of positive work performed. It appears that for the chosen tasks, the additional dexterity and sensorimotor sensitivity of the hands may assist in the motor plan development and execution at the feet, while the feet may not have any reciprocal benefits to offer the hands. This imbalance in capabilities between the upper and lower limbs for neurologically intact subjects offers a good model for the imbalance in capabilities that exists between the affected and less affected limbs of neurologically impaired subjects.

We are interested in developing strategies for lower limb rehabilitation. Therefore, it is desired to demonstrate methods for promoting lower limb performance even if at the expense of upper limb performance. It is understandable that the hands were better equipped to succeed at the pursuit tracking and resonant excitation tasks. The hands are more dexterous with higher resolution afferent feedback; people are accustomed to using their hands and fingers for precisions tasks such as typing and tying shoe laces. The feet, on the other hand, are generally involved in gross motions involving larger forces. It is possible to design a task where the lower limbs are better equipped for success than the upper limbs. The feet might offer the hands something in the way of force and endurance capabilities. However, in the design of the experiments presented in Chapter 3, we considered and compensated for the force mismatch between the two effectors. Using a telerobot provided for independent scaling of forces to the hands and feet, essentially removing an advantage the feet might otherwise have had. Similarly, it would be possible to exploit the scalability of the telerobot to compensate some of the disadvantages the feet might have compared to the hands. For example, the position scaling could be altered to increase the lower limb motion, essentially increasing the resolution of the ankle proprioception in comparison to the hand.

We have demonstrated that there exists a tradeoff between the benefits of multi-limb control and the difficulty of increased cognitive load. However, with careful design, it is possible to create a task that balances this tradeoff to improve some element of motor performance. Multi-limb control does not necessarily benefit both limbs. There appears to also be a tradeoff between improving the performance of an inferior effector and degrading the performance of a superior one. The benefits that a subject receives from multi-limb control might only become evident in circumstances where one effector is superior to the other at the presented task. Self-assist, therefore, might prove successful only in the circumstances when there is a dominant or unimpaired limb directing the assistance to a non-dominant or impaired limb. In designing a self-assisted protocol, it will be important to consider the imbalance in capabilities between the limbs.

6.2 Self-assist yields superior anticipation and coordination

Aim 2

Demonstrate that subjects develop anticipatory adjustments in the lower limbs that can be used to improve coordination when compensating for externally applied loads.

The results from the disturbance rejection experiment presented in Chapter 4 demonstrated that subjects could better anticipate and compensate for a known disturbance to the lower limbs when they were directly responsible for triggering it with the upper limbs. Another significant component of this experiment was the reproduction of upper limb disturbance rejection results as previously demonstrated by Diedrichsen and colleagues [57]. We showed that using the same protocol and equipment, the lower limbs exhibited reductions in peak acceleration, meaning better disturbance compensation, similar to the upper limbs. The performance improvement for the upper limbs during self-triggered disturbance generation as compared to uncued disturbances was slightly greater than for the lower limbs: 24% reduction compared to 16% reduction. For both cases, the coordination was between non-homologous muscle groups. However, the upper limbs are more practiced at coordinated tasks even if each is performing distinct motions, such as tying a shoelace. This divergence in experience might account for the minor performance differences that favored the upper limbs.

This experiment used a discrete pushbutton trigger to initiate the disturbance. Therefore the results pertain to timing coordination between muscle activation and external loads. We demonstrated that this timing coordination is better within a subject than between a subject and a computer, even with audiovisual cuing. We have shown that for timing coordination, the lower limbs demonstrate similar positive responses to upper limb volitional actions as the upper limbs. Previous studies on bimanual coordination have shown that compensation for external loads is even further improved when subjects generate the shape of the force profile as well as the timing of it [45]. Generalized self-assist through telerobots would more fully exploit the coordination benefits of self-generated force profiles. The efforts of the impaired limb would coordinate better temporally and spatially with assistive forces generated by the patient than through some outside agent.

6.3 Short term exercise increases active dorsiflexion

Aim 3

Design a functionally relevant task and determine if short term assisted practice in that task, including practice with interlimb coordination, will benefit subsequent unassisted capabilities in hemiparetic stroke subjects.

Mobility is one of the most important factors in maintaining personal independence. Therefore, locomotor rehabilitation is frequently a significant focus following a neurological injury. It is likewise our primary motivation in researching lower limb motor control. However, in developing motor tasks to evaluate principles for generalized self-assist, we chose to focus on simple single-axis motion over gait. By examining relatively simple biomechanical movements, we reduced the number of confounding factors that might have masked discernable results. However, as a feature of the body of work presented in this dissertation it was important to demonstrate that we could develop a functionally relevant exercise for stroke subjects using single axis ankle motion.

The obstacle avoidance task presented in Chapter 5 represents an engaging means of exercising the tibialis anterior to promote improved dorsiflexion. The stroke subjects that participated in the obstacle avoidance study expressed an overall positive reaction to the exercise presented. Based on conversations with the participants, there was a general consensus that the ankle dorsiflexion muscles were difficult to exercise given the available equipment. Subjects noted that the task was both challenging and rewarding.

After a single training session on the equipment, subjects were able to increase their active range of motion and their muscle activity as measured with EMG. Range of motion increased an average of 19%-26%, depending on the specific position metric used. The normalized EMG increased for subjects an average of 29%. This experiment and subsequent results cannot be used to determine if self-assist was a necessary or sufficient component in improvements observed. However, we can comment on the benefits of researching simple lower limb motor control. By examining a small yet significant component of gait, we were able to demonstrate performance improvements in stroke subjects that have relevance to larger motor tasks.

6.4 Immediate effects of assistance type on performance

Aim 4

Assess the immediate effects of self-assist, computer-assist, and experimenter-assist on task performance and coordination between patient and assistance for the impaired lower limbs of hemiparetic stroke subjects.

The experiment presented in Chapter 5 discussed a training protocol wherein we compared the facilitative effects of self-assist, computer-assist, and experimenter-assist on dorsiflexion in hemiparetic subjects. We demonstrated that all three types of assistance increased the ability of subjects to avoid colliding with the obstacles. Fewer collisions in an indication that subjects moved their feet through larger ranges of motion and accomplished better timing with regard to the oncoming obstacles.

Muscle activation, as measured with recorded EMG, indicated effort on the part of the subject to complete the task. We hypothesized that self-assist would encourage muscle activation through facilitation over conditions when no assistance is provided. However, we found that subjects demonstrated the highest levels of muscle activation when they were not provided any outside assistance in lifting the foot. Of the assisting conditions, only during self-assist was performance in terms of EMG not significantly reduced. Both the computer-assist and the experimenter-assist appeared to act as a crutch in completing the task. That is to say that subjects were better at avoiding obstacles but used less personal effort in accomplishing it. In terms of efficiency, this appears to be a positive result: better performance with less cost. However, in terms of rehabilitation, it is necessary for the subjects to push themselves to improve. Self-assist offered a balance between encouraging subjects to stay motivated and try while allowing them to experience more success during the exercise. We believe that the performance differences between self-assist and assist from an outside agent can be primarily attributed to subject awareness of aid. During self-assist, the subject is fully aware of the timing and magnitude of assistance that he provides from the hand device. When assist is provided from an outside agent, such as the experimenter or the computer, the subject may not be able to develop a good model for applied aid. It is possible that subjects do not even recognize that assistance was provided. Therefore, this type of assistance can become an unintentional crutch, whereas self-assist demands subject awareness.

Although we saw certain trends across many of the subjects, there were large differences in the results. Due to the highly individualized nature of impairment following CVA, it is common for results, even within a study, to be mixed [47]. Factors such as the size and area of the legion, the time since the trauma, the age of the subject, and the pre-stroke hand/foot dominance are just a few of the factors that can contribute to inconsistent results. Our study only controlled for the severity of the hemiparesis and that all subjects were considered in the chronic stage of recovery. There is a high level of individualism in the motor effects following a CVA. Future studies could include discerning what specific group of subjects would most likely benefit from self-assist. This would involve examining the effects between groups where homogeneity of impairment is as controlled as possible.

6.5 Accomplishments

This dissertation presented a body of work that developed the concept of generalized self-assist motivated from clinical success of certain neurological rehabilitation treatments [6, 14, 29]. We presented a series of experiments conducted using custom built hardware to test concepts in lower limb motor control and adaptation.

We have shown that people are capable of performing multi-limb motor control to achieve improved performance over single limb operation. Also, that this multi-limb control can lead to better temporal coordination, afferent information, and lower limb muscle recruitment. This document does not answer all questions about generalized self-assist, but it does address some of the initial fundamental concerns. The results that we have obtained encourage further investigation into generalized self-assist as a viable and beneficial means of administering lower limb rehabilitation.

6.6 Future Work

The work presented in this document inspires many further questions on the details and specifics of generalized self-assist as well as questions pertaining to the particulars of multi-limb motor control. There are two primary experiments that should be conducted to complement the work presented here. The first experiment would more completely address the questions of cognitive load initiated in Chapter 3. The second experiment would further explore the causation of the increased active dorsiflexion capabilities observed in the stroke population presented in Chapter 5.

The primary results of Chapter 3 indicate a tradeoff between the benefits of multi-limb control in a dynamic task and the difficulty associated with executing multifaceted and more complicated motor commands. The experiments presented demonstrate that despite the increased mental demands of multi-limb control, the addition of the upper limbs offers benefits to lower limb performance. Future work should include a series of experiments to further define and/or quantify the cognitive load that is associated with multi-limb control. In addition, we should also examine how increases in cognitive load from secondary tasks affect performance in dynamic motor tasks. These questions can be explored by combining upper limb, lower limb, and combined upper and lower limbs motor tasks with various secondary cognitive tasks. These secondary tasks should be verbally executed and may include in-depth conversations, simple math problems, memory retention, or verbal responses to visual cues. The experiments would include assessment of motor task performance as the secondary task changes or becomes more difficult, as well as secondary task performance as the motor task changes from single limb to multi-limb. Motor task performance metrics may include ones similar to those already presented in this document, such as work done, tracking performance, and accelerations. Secondary task performance metrics may include reaction times and accuracy of the responses.

The work presented in this dissertation would be nicely complemented with another experiment with a stroke population to further examine the causation and retention of the increased active dorsiflexion noted in Chapter 5. This experiment demonstrated that after a short, single session training period, stroke subjects significantly increased their active dorsiflexion range of motion as well as the EMG activity in the tibialis anterior. The experiment, however, was not controlled to indicate which of the training conditions was responsible for the dramatic improvement. Given the positive results in both task performance and muscle activation during the *Self-Assist* training condition, we believe that it maybe have been at least partially responsible for the overall effect. A carefully designed experiment should be conducted to test this hypothesis. This experiment should involve dividing subjects into different groups, where each group receives a different training treatment. By comparing the pre- and post-training performances of each subject, we can determine if one type of training has a larger effect than another. Given the variance in impairment among stroke subjects, each group should include 12-15 participants. The most important comparison is between a *Self-Assist* group and a *No Assist* group. This would address the effects of assisted practice compared to a controlled practice with no assistance. After that distinction is made, then the specific type of assistance can be addressed.

A follow-up to this experiment would include investigation of the long term affects of dorsiflexion training. We should further examine how much of the dorsiflexion improvements are retained after different periods of time: a day, a week, a month. Further, we should examine if the performance improvements and the retention can be increased by repeated training sessions. An experiment that includes a series of visits by each subject should be conducted. The visits would involve pre-evaluation, training, and post-evaluation. Further, subjects would be divided into groups to receive only a single type of training condition. This experiment would address retention, by comparing the pre-evaluation to the post-evaluation of the previous session. It would also address long term training effects by tracking subject performance over the course of the entire experiment. The combination of a controlled experiment to determine the comparative effects of each training condition and a long term experiment to determine the effects of extended training and retention will provide a nice supplement to this dissertation. The results could greatly influence rehabilitation training techniques for the treatment of drop-foot. In addition, they may further justify generalized self-assist and encourage additional research projects on the topic.

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