NOVEL INTERFACES FOR TRAINING NEUROMOTOR CONTROL OF THE UPPER AND LOWER LIMB

by

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ABSTRACT OF THE DISSERTATION

Neuromotor Control of the Ankle in Unilateral Activities

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One of the primary goals in hemiparetic stroke and cerebral palsy rehabilitation is to improve movement efficiency by correcting typical abnormalities of hemiplegic movement. In order to help patients achieve higher functionality, it is important to not only to develop new equipment, but also to understand how motor learning takes place in those with neuromuscular dysfunction. The hypotheses of this dissertation were (1) A simple interface between muscles with and engaging game interface can be used to promote repetitive task practice. (2) Inter-limb transfer of learning occurs in the lower limbs in accord with their hemispheric specialization.

Specific contributions include the design a force myograpic cuff (FMG), a goniometric ankle platform, a LabVIEW interface that recorded signals from both devices ran games. In addition, four small-scale clinical tests conducted using the two devices, and a clinical test of ankle ILT on 22 unimpaired subjects.

The clinical tests performed with hemiparetic, and unimpaired volunteers revealed that the ankle platform performed better when used with impaired patients than the FMG cuff. However, combining force myography with a mouse emulator to allow patients to play computer games using meaningful upper limb movements appears to be a promising upper limb intervention.

The second part of this project explores how Inter-limb learning transfer (ILT) occurs in the lower limbs and the implications these results may have for lower limb rehabilitation. Twenty-two healthy right-dominant subjects were divided into two groups: half performed the tasks first using the right foot (group RL), and the other half performed it first with the left foot (group LR). Results demonstrated that group LR but not group RL experienced significant ILT of directional as well as positional information in both tasks in a manner reflective of the distinctly different functional roles played by the upper and lower limbs. The present results thus provide clear evidence for the potential benefit to the affected limb afforded by contralateral limb training, and studies are underway to test its efficacy.

ACKNOWLEDGEMENT AND DEDICATION

This thesis is dedicated to my parents Bernadine Mitchell and Harold Sims who have helped me get to this point, and to my husband Albert Morris who has patiently waited for this day. I would like to thank Dr. Craelius and all of those in the Rutgers University Rehabilitation Laboratory who have provided invaluable advice and guidance during my time at Rutgers. A portion of this dissertation has been submitted to and published in the Journal of Experimental Brain Research[1].

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CHAPTER 1

INTRODUCTION

A universal objective in hemiparetic stroke and cerebral palsy rehabilitation is to improve movement efficiency by correcting typical abnormalities of hemiplegic gait. To achieve this goal many have used treatments such as muscle strengthening, biofeedback, pharmaceuticals, and orthotics. Presently the standard approach to ankle rehabilitation is to overcome muscle weakness with the use of an artificial foot orthosis (AFO). This approach allows patients to have limited mobility, but the underlying problem remains unaddressed. New rehabilitation approaches have been developed to increase independent mobility. In order to provide more valuable rehabilitation regimens, it is important to not only to develop new equipment, but also to understand how motor learning takes place in those with neuromuscular dysfunction.

There are a variety of methods used to capture and measure human motion such as human eye or other senses, timing devices, videography or computerized optical systems, goniometry, electromyography, dynamography, and accelerometry. Apart from the video capture systems there are very few commercially available joint measurement devices. Many of the devices used are focused only on the impaired limb and are custom made by individual research groups.

Unilateral hemispheric brain damage can produce both contralateral and ipsilateral motor deficits. In the past, more focus has been placed on contralateral deficits since they are more pronounced. In recent years, researchers have started to explore the nature and extend of ipsilateral motor deficits. Although damage to either hemisphere can lead to comparable ipsilateral deficits left and right hemisphere injuries produce qualitatively different results [2]. Therapists have found that left hemispheric damage tends to produce deficits in the transport component (the manner in which a person moves to a target), and right hemispheric damage tends to produce deficits in the positional component (the manner in which a person holds an object at a desired position).

These patterns are similar to those found in motor learning studies which have shown that the dominant arm is capable of sending endpoint information to the non-dominant arm, and the non-dominant arm is capable of sending trajectory information to the dominant arm. According to the hemispheric dominance theory proposed by Sainburg, this asymmetric transfer of learning occurs because the dominant hemisphere specializes in interpreting trajectory information, and the non-dominant arm specializes in interpreting end-point information [3, 4]. However, the implications of these findings for the lower limb are still unclear.

The goals of this project are to design a device capable of quantifying the motion of the human ankle; and to explore how the brain learns new ankle movements. This will be achieved specifically by studying the nature of cross-hemispheric learning of ankle movements [5] in healthy individuals and individuals with stroke in a manner similar to that of the upper limbs of healthy individuals. I hypothesize that Inter-limb transfer of learning occurs in the lower limbs for healthy individuals.

CHAPTER 2

BACKGROUND

2.1 Ankle Kinematics

The ankle is formed by the union of three bones, the tibia, the fibula, and the talus of the foot. It is responsible for both load bearing and kinematic functions. The tibiotalar, fibulotalar, and tibiofibular joints are the three articulations that make the ankle joint. Another term used to describe the ankle joint is the "mortise"; it is formed by the shape of the three articulations, and the ligaments and muscles crossing the joint.

Figure 1 Ankle joints [6].

The tibiotalar ankle joint acts as a hinge between the spool-like surface of the trochlea of the talus and the concave distal end of the tibia. This joint permits only flexion and extension (dorsiflexion and plantar flexion). Other movements such as inversion and eversion, inward and outward rotation, or pronation and supination occur about the subtalar and transverse tarsal joints of the foot.

There are many muscle groups crossing the ankle. The key plantar flexor muscles are the gastrocnemius, and the soleus muscles, which both attach to the posterior surface of the calcaneus. In addition, the anterior and lateral muscles provide pronation, and supination, and inward and outward rotation of the foot.

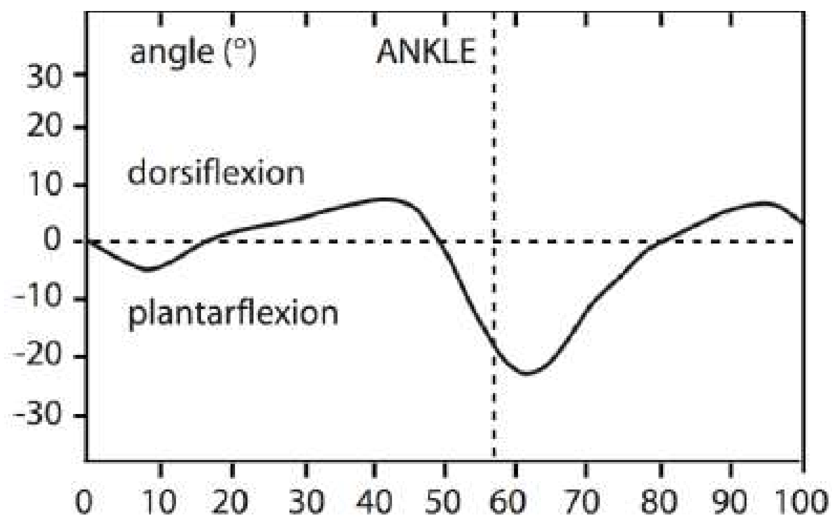

Figure 2 Sagital plane ankle joint angles[7] .

The ankle muscles play a critical role in the gait cycle. Throughout the initial contact phase of the gait cycle, the ankle is close to the neutral position (Figure 2). It then plantar flexes to bring the foot flat onto the ground. During mid-stance, the tibia rotates forward, moving the ankle into dorsiflexion. The gastrocnemius and soleus muscles of the leg are heavily activated during the stance phase in order to push the center of mass of the body forward and upward before the heel strike of the contralateral leg.

Sophisticated computer models have been able to replicate the forces seen by the muscles during walking. According to these models, both the gastrocnemius and soleus muscles develop peak forces at the same time. The peak force seen by the gastrocnemius muscle is approximately 900 N, and the peak force seen by the soleus muscle is 2000N

2.2 Neuromuscular Ankle Dysfunction

Having good control of the ankle joint is important for balancing and ambulation. Gait is a complex motor skill that requires equilibrium and antigravity supporting reactions to maintain a balanced upright posture during locomotion. Impaired ankle motion may be caused by neural factors such as spasticity, muscle weakness, structural adaptations, or non-neural factors such as immobilization [8].

Drop foot is a gait disorder caused by ankle muscle weakness or stiffness. In order to achieve normal gait, a minimum of 10 degrees of dorsiflexion is needed to clear the ground during the swing phase of gait. In moderate cases of drop foot, the front of the foot drops to the ground following heel strike and the toe catches the ground during the swing phase. Thus, drop foot can have a major effect on gait.

Drop foot can occur as the result of an injury to one of the ankle muscles or a neuromuscular disorder such as cerebral palsy or stroke. It can manifest itself in many forms such as ankle equinus or ankle equinovarus. Ankle equinus is a deficiency experienced by many cerebral palsy patients, in which there is limited dorsiflexion of the ankle due to decreased flexibility in the gastrocnemius or soleus muscle groups. Most stroke patients experience equinovarus disorder, which is a weakness in both the ankle dorsiflexion and the ankle eversion muscles. To compensate for a decreased range of motion, the subtalar and midtarsal joints (inversion-eversion) may occasionally be used and could result in a severely pronated foot[9].

2.2.1 Neurological Causes

Neuromuscular dysfunction is often the result of damage to the brain either via an ischemic event or a traumatic injury to the brain. Cerebral ischemia is a common cause of stroke, and is usually caused by sudden blockage of an artery supplying the brain or by low flow to a distal artery that is already blocked or highly constricted. This blockage or constriction may be the result of disease of the arterial wall, embolism from the heart, hematological disorders, or various rare but treatable conditions[10].

Unilateral cerebral-hemisphere lesions cause contralateral hemiparesis. The forehead, tongue, and bulbar musculature are unaffected unless the upper motor neuron lesion is bilateral, however the lower facial musculature are affected. Focal lesions that affect only a portion of the motor cortex produce paralysis of the body part controlled by that point of the motor cortex. Focal motor, or Jacksonian epileptic attacks that affects only one side of the mouth, the finger and thumb, or the great toe, are typical of irritative lesions of the motor cortex. Lesions in the internal capsule, even those that are minor typically lead to complete hemiplegia because the corticospinal-tract fibers are densely packed.

In addition to motor impairments, cognitive and sensory deficits may also result from lesions in the motor cortex. Typically, cortical or subcortical lesions produce cognitive deficits. In contrast, lesions deep in the white matter frequently produce dense hemiplegia without cognitive loss. Lesions in the medial section of the medulla oblongata may produce an ipsilateral hypoglossal nerve deficit and contralateral sensory loss. This loss leads to contralateral hemiparesis and affects the ability to sense vibration and joint position.

2.2.1.1 Muscle Weakness

Patients typically start to notice muscle weakness when they begin to have trouble with walking. Proximal leg muscle weakness typically makes it hard for patients to climb or descend stairs, stand out of a bath or arise from sitting without using the arms. Distal leg muscle weakness causes ankle instability or foot drop, the inability to curl the toes into plantar flexion to keep loose shoes from falling off the foot, or to grip the edge of a swimming pool to dive.

The patterns of weakness and the symptoms present can provide clues concerning the location and severity of a lesion, and the pathology of the impairment. Lesions in the upper motor neuron cause paralysis of the limbs. Severe lesions cause complete paralysis; less severe lesions cause distinctive patterns of weakness. In the lower limbs, hip flexion due to the weakening of the iliopsoas usually occurs the earliest, and hamstring and ankle dorsiflexion weakness is often pronounced. Hemiparesis, affecting both the upper and lower limbs on one side of the body is typical of a cerebral-hemisphere lesion. Paraparesis or weakness of both legs is usually the result of thoracic spinal cord or cauda equine disease. Quadriplegia, the weakness of all four limbs, suggests cervical spinal cord or diffuse neuromuscular disease.

Some now believe that muscle weakness is more harmful to function than spasticity. Muscle weakness is a key component of cerebral palsy[11]. Palsy by definition is weakness originating from the brain. Phelps theorized over 50 years ago, that resistive training would be beneficial to children with CP. However, in the past there was a fear that near maximal effort during resistive training would increase spasticity and muscle tightness, making the problem worst. When this article was written (2002) there was still some opposition to resistance training, despite evidence that supports such exercise.

Muscle weakness is not the only factor hindering movement in those with CP. Other factors such as decreased central input to the muscle, changes in the elastic properties of the muscles themselves, defects in the inhibition pathways of the agonist-antagonist muscle pairs, and increased stretch responses or spasticity also contribute to poor function. Some of these factors may actually be secondary, and thus preventable if the primary factors (i.e. muscle weakness) are addressed. Some caution against trying to correct secondary factors because they fear that doing such may actually increase muscle weakness.

2.2.1.2 Muscle Stiffness

In addition to muscle weakness, muscle stiffness may also lead to decreased joint mobility. Spasticity, rigidity or hypertonia can contribute to increased muscle tone. The tone of a muscle is the response it shows to passive stretching. In practice, tone is assessed by moving a limb and observing the reaction that occurs in the muscles that are being stretched. Muscle tone is regulated by reticulospinal fibers that accompany the pyramidal tract and exert an inhibitory effect upon the stretch reflex.

Abnormal ankle stiffness is a common aliment associated with ankle dysfunction. Ankle stiffness is the relationship between joint position and the torque acting about it, which defines the mechanical behavior of the joint. Ankle stiffness can be separated into two components, intrinsic stiffness, and reflex stiffness. The intrinsic stiffness is due to the mechanical properties of the joint, passive tissue, and active muscle fibers; whereas the reflex stiffness is the stiffness component due to changes in muscle activation due to sensory responses to stretch [12]. The parallel cascade method developed by Kearney et al. has been used to separate the reflex and intrinsic components of stiffness. This method allows investigators to determine whether neural pathways, or mechanical properties of the muscles (i.e. flexibility) are damaged[10].

Galiana et al. [13] conducted a study comparing the intrinsic and reflex stiffness between stroke affected and normal subjects. A bimodal distribution was found within the stroke group, meaning there were some subjects in the stroke group with torques similar to those in the control group, and some subjects in the stroke group with torques much higher than the control group. Due to this difference, the authors decided to separate the stroke group into two subgroups: stroke high reflex torque (SHRT), and stroke low reflex torque (SLRT). The intrinsic stiffness was found to be similar for all groups; however, the reflex stiffness was higher in the SHRT group. In addition, it was also shown that as the angle of dorsiflexion increased, the intrinsic stiffness contribution to the overall stiffness decreases, and the reflex stiffness contribution increases. These results confirmed that increased ankle stiffness in stroke patients is primarily due to the neuromuscular components of stiffness and not to the mechanical properties of the joint.

Despite the serious effect drop foot has on gait, there is very little in the literature concerning its treatment. Even comprehensive textbooks tend to address this issue briefly, typically only advising caregivers to use a brace that will keep the foot in dorsiflexion day and night, with passive ankle flexing to help prevent atrophy [14].

2.2.2 Current Treatment

Current methods used to treat drop foot, include physical and occupational therapy, biofeedback, surgery, drugs, and orthotics. All have been shown to be effective for certain groups. Both function focused, and quality of movement focused therapies have been found to yield similar results.[15] The standard treatment options are currently physical and occupational therapy, and the use of orthotics. Recently, rehabilitation providers have begun using techniques such as biofeedback and functional electrical stimulation to supplement both therapy and orthotics.

2.2.2.1 Strength Training

A major part of physical therapy is muscle training, which changes muscle contractile properties such as strength and contraction speed; as well as the ability of the nervous system to control muscular function[16]. Strength can be increased by an increase in muscle size or an increase in the net neural drive to the muscle. For older persons, hypertrophy may be limited and strength gains may depend on more neural factors [17].

Neural factors are the primary cause of strength gains in the first two to eight weeks of strength training, when subjects are still learning how to exert force effectively. Strength training enhances synchronous muscle fiber recruitment (resulting in the summation of force) and reduces inhibition of motor units. Usually, each motor unit receives signals from several motor units, including some that are inhibitory. By reducing inhibition, more units can become active, thus increasing the force output. Strength training also tends to decrease the activation of the antagonist muscles opposing the desired movement, and increase the activation of the synergist muscles that assist the muscles making the desired movement. However, a significant change in muscle spasticity as a result of strength training has not been found.

Isometric and isokinetic training are the two most common methods of improving muscle strength. Isometric contraction can be used to measure the muscle's maximum strength and is dependent on the length or angle at which the muscle is held. Isokinetic knee extension can change the shape of the force-velocity curve for the quadriceps, which in turn increases the power of the muscle. A dynamometer is used for both types of training.

Isokinetic testing at slow speeds can reliably improve strength in those with CP and those with other spastic disorders for certain muscle groups. The method of producing strength gains in CP appears to be the same as those for other chronic motor disorders. However, due to shortened muscles, alternative test positions may be necessary. During rehabilitation, therapists are careful to avoid positions that inhibit the use of flexor and extensor synergies (muscles that assist the main muscle producing the desired force).

2.2.2.2 Orthotics

It is common for those with drop foot to wear an ankle foot orthosis (AFO) while ambulating. The AFO may be the only orthosis worn or it may act as a basic component for a more extensive orthosis system. AFOs may come as dynamic (DAFO) or molded (MAFO). The DAFO contains a custom contoured foot-plate designed to promote balance of muscle power and to reduce the need to seek stability through compensatory balancing methods. Examples of a DAFO include inhibitory casts and inhibitory orthoses, which are sometimes referred to as inhibitory AFO, tone-reducing AFO or neurophysiologic AFO. The MAFO does not have a custom contoured footplate, and consequently usually does not offer support for the arches or to the toes. However, with some modification it can offer support for the arch of the foot[18].

In addition to differences in fabrication, AFOs vary in the amount of restriction or assistance of ankle movements. AFOs may come as fixed, hinged, dorsiflexion-assist, or ground reaction. A fixed AFO provides support and position for the ankle and the subtalar joint. They are typically used for patients that have little or no voluntary control of dorsiflexion, or excessive knee extension during weight bearing. The trim lines of the fixed AFO can be used to control the amount of forward movement of the tibia. The tendency to flex the knee at initial contact can be limited by setting the trim line at 2°-3° of dorsiflexion. Setting the dorsiflexion to 5°-7° can limit the tendency for knee hyperextension at stance.

A hinged AFO allows dorsiflexion and/or plantar flexion at the ankle. However, the hinged AFO usually blocks ankle plantar flexion while allowing dorsiflexion. The hinged AFO is used for patients that have at least some voluntary control of dorsiflexion, but no control of plantar flexion, or for those with limited voluntary control of both plantar flexion and dorsiflexion. A spring loaded dorsiflexion assist can be used together with a hinged AFO to allow for the passive dorsiflexion of the ankle.

Dorsiflexion assists are generally used for patients that have a sufficient range of passive motion but limited or no voluntary control of dorsiflexion or plantar flexion. A dorsiflexion assist AFO may come in plastic form either in a spiral shape that coils around the shaft of the lower leg and supports the foot or as a posterior leaf-spring orthosis. All of these AFOs except for the spiral are able to limit mediolateral motion by extending the plastic over the sides and the top of the foot.

A ground reaction AFO is molded to fit around the front of the leg. Ground reaction AFOs are used for patients with excessive knee flexion, and ankle dorsiflexion during weight bearing.

The straps on the back of the leg hold the leg and heel in place, and along with the anterior shell, limit the forward movement of the tibia at initial contact and throughout the stance phase of gait. This orthosis may enable the wearer to maintain a full upright posture by limiting the dorsiflexion range [18].

2.2.2.3 Biofeedback

EMG biofeedback has been demonstrated to be an effective form of muscle feedback for both upper and lower extremities. In a study conducted by Durson et al., EMG biofeedback was applied to dorsiflexors and plantar flexors of the ankle joint in CP children who also had dynamic equinus deformities[19]. For ten days, patients received biofeedback thirty minutes per day, plus conventional exercise for two hours per day. The control group received conventional exercise two and a half hours per day. The biofeedback group displayed statistically significant improvements in the tone of the plantar flexor muscles and active range of motion of ankle joints. Both groups showed statistically significant improvements in gait function; however the biofeedback group showed more progress than the controls.

Lyons and colleagues created and tested an EMG controlled biofeedback game system designed to motivate patients to adhere to a therapy regimen. [20]. The system was capable of controlling a space invader game with EMG signals. Computer game play was enabled or disabled, based on the intensity of muscle activity recorded from the muscle of interest similar to an exercise machine in which you can play Tetris or watch TV, as long as peddling continues.

Four adult subjects with cerebral palsy and four healthy adults underwent thirty minutes of conventional therapy in addition to this protocol. Muscle range of motion,and muscle tension were measured by the same trained physical therapist before and after the four-week trial. The linear envelope of the subjects' EMG was compared to a threshold to determine whether to enable or disable game play.

There was no change in spasticity for the control or experimental group. Muscle tension increased on average by 1 for the experimental group, with no changes for the control group. Range of motion increased 5 degrees for the control group and 12.75 degrees for the experimental group. In addition to physical parameters adherence performance was measured by keeping track of the number of contractions and the time spent for each session. The EMG controlled game produced more frequent movement, and thus greater improvements. The average number of contractions per session for the control group was 45.45, and 100.5 for the experimental group. The average time spent per session was 110 seconds for the control group and 129.75 seconds for the experimental group. The authors concluded that this was a promising method of increasing patient physical activity.

2.2.2.4 Functional Electrical Stimulation (FES)

Functional electrical stimulation (FES) is the application of electrical current to excitable tissue to augment or replace function that is lost in individuals with neurological impairments. FES can be used to restore both sensory and motor function. Functional restoration is accomplished by electrically activating the intact lower motor neurons using electrodes placed on or near the innervating nerve fibers. The electrical stimuli elicit action potentials in the innervation axons, causing a muscle contraction. The strength of the muscle contraction is regulated by modulating the pulse frequency, amplitude, and duration. A functional limb movement can be induced by coordinating the electrical activation of several muscles[21] .

The threshold charge needed to elicit action potentials in muscle fiber is much greater than the threshold needed to produce action potentials in the neuron. Therefore, FES applications for motor function typically work by electrically stimulating the nerves associated with the muscles needed to produce a movement. To activate muscles using this form of FES, the lower motor neurons must be intact from the anterior horns of the spinal cord to the neuromuscular junctions. Patients with spinal cord injury, stroke, head injuries, cerebral palsy, and multiple sclerosis are good candidates for neuromuscular electrical stimulation when the lower motor neurons are excitable and the neuromuscular junction and the muscle are healthy.

Stimulators can be designed to regulate current or voltage. Voltage-regulated stimulation is often used for surface stimulation applications in order to minimize the possibility of skin burns that may result from high current densities. However, because the electrode impedance can affect the current delivered by the voltage regulated stimulator the motor response is more variable.

One of the first applications of FES was preventing the foot from dragging on the ground during the swing phase of gait in patients with post-stoke hemiplegia. These systems used surface electrodes positioned on the tibialis anterior and on the common peroneal nerve where it crosses the head of the fibula, a stimulator implanted in the medial thigh region, or a stimulating electrode implanted on or near the peroneal nerve. Stimulation was activated with a heal switch worn in the shoe of the non-impaired foot.

The early work demonstrated the efficacy of FES for foot drop; however, there were many shortfalls with these systems. Those working with the early FES systems experienced problems placing the surface electrodes, false triggering of the stimulation, inadvertent elicitation of reflex spasms in the plantar flexor muscles, pain or discomfort from the stimulation, mechanical failure of the switch or other components, and difficulty achieving balanced dorsiflexion with a single electrode. Currently several foot drop systems are already or close to being commercially available[21].

2.3 Motor Learning

While central motor control of the limbs is primarily contralateral, there is increasing evidence that specific types of learned motor actions can transfer across the hemispheres [22]. Inter-limb transfer [23] of acquired motor skills, i.e., the ability for a limb to perform a task learned by the opposite limb, has been demonstrated for many activities such as drawing, writing, mirror tracing, ball catching, and pointing [24-28].

Research on bilateral transfer has been published as early as 1894, in which Scripture et al. studied the increase in the voluntary force-generating capacity of the opposite untrained limb that occurs as a result of unilateral resistance training [29]. Since this time, researchers have studied the bilateral transfer of a variety of skills. During the 1930's, a series of bilateral transfer studies using a star shaped maze was conducted by Cook. Cook referred to this phenomenon as "cross education, which is traditionally said to be the transference of common elements of a task to another limb. Both cognitive and motor control explanations have been used to explain why bilateral transfer occurs[30].

Bilateral transfer can occur with isometric or dynamic training, and is usually confined to the homologous muscle of the opposite untrained limb[29]. Throughout the years, the bilateral transfer of several skills has been studied in both the upper and lower limbs. Most studies on ILT have been done with the upper limbs, where ILT has been found to be symmetric (bi-directional) or asymmetric, depending on the task studied. Learning of simple tasks, such as grasping, lifting small objects, or anticipatory timing, transfers across hemispheres symmetrically [31]. ILT of more complex learned movements, such as reaching in the presence of visuomotor rotations is asymmetric and correspondingly more complex.

2.3.1 Bilateral Transfer of Strength

There have been several studies conducted to determine the feasibility of using bilateral transfer to increase the strength of an inactive limb. One motivation for studying the bilateral transfer of strength is to find a way to prevent joint stiffness or muscle atrophy in an unused limb. Immobilization for short periods of time can result in undesirable joint stiffness and weakness of the muscles around the affected joint. Numerous methods of limiting the damaging effects of joint immobilization in the knee and other joints have been investigated. Experts have recommended using a hinged cast or brace as soon as possible, continuous passive movement of the joint, electrical stimulation of the muscles surrounding the injured joint, and various strength training programs [32].

Bilateral transfer promises to be a practical means of preserving and improving limb function. Some of the key questions asked of researchers in this area are: (1) how much can strength increase with such a strategy, and (2) how long will these improvements be sustained. This review gives a few examples of studies that have examined protocols that encourage the bilateral transfer of strength.

Munn et al. conducted one of the largest studies used to determine the effect of unilateral resistance training on the strength of the untrained limb, following opposite arm training[33]. One hundred and fifteen untrained healthy subjects were divided into a control group, and four training groups. Following a warm-up, the strength of the elbow flexor muscles was measured on both arms of all subjects using a one-repetition maximum test. The training groups performed either one set at high speed, one set at low speed, three sets at high speed, or three sets at low speed. Training was conducted three times a week, for six weeks.

One set of slow exercise increased initial strength by 25%, and three sets of slow exercise increased initial strength by 48% in the trained arm. Training at the higher speed resulted in an 11% greater strength increase than slow training. For the contralateral arm, one set of exercise did not increase the initial strength, however, three sets of exercise increased initial strength by 7%. Speed had no effect on the initial strength.

Hortobagyi et al. also conducted a fairly large study in which the short-term effects of eccentric and concentric training at equal force levels was examined in the lower limbs[34]. Forty-two healthy female volunteers were randomly placed into an eccentric training group or a concentric training group. All subjects underwent training for six weeks. Strength and neural adaptations prior to and following training were observed. There were no significant strength increases in the contralateral leg for the concentric or eccentric group. The authors believe the cross training effect in this study may contrast to Munn's findings because of the length and intensity of the training.

Uh et al. examined the effects of a single leg strength-training program on the untrained contralateral ankle muscles [32]. Twenty healthy subjects were assigned to a control or training group. Before and after the eight week training session, all subjects underwent isokinetic strength testing at two speeds, 30 and 120 degrees per second, using an isokinetic dynamometer on both ankles. The training group was divided into two subgroups: one group trained the non-dominant ankle and the other group trained the dominant ankle. The subjects in the control group were allowed to resume normal daily activities, and were prohibited from starting a strength-training program for the lower extremities. Strength, measured as peak torque, power, and endurance, were the primary outcome measures for this study.

There was a significant difference in strength among the training groups at both speeds. The subjects who trained with the dominant leg showed an 8.5% improvement in the dominant leg and a 1.5% improvement in the non-dominant leg. The subjects who trained with the non-dominant leg showed a 9.3% improvement in the non-dominant leg and a 3.5% improvement in the dominant. At 30 degrees per second both the dominant and non-dominant groups had significantly higher increases in strength than the control group. At 120 degrees per second, only the nondominant leg group showed an increase significantly higher than the control group. There was a significant difference in power between the control and nondominant leg groups and between the non-dominant and dominant leg groups at 30 degrees per second. At 120 degrees per second there was a significant difference between the change in power between the nondominant leg group and the control group. There were no significant differences between the three groups for the endurance measure.

The improvements seen in the untrained leg may be due to neuromuscular facilitation increasing the recruitment of more muscle bundles per nerve impulse. It is likely that training the muscles around one ankle has a mutual benefit when activating the corresponding muscles of the opposite ankle.

The level of strength increase achieved with bilateral transfer has varied for each study. A meta-analysis conducted on 13 small studies concluded that an 8% increased of initial strength of the homologous muscles of the contralateral side was observed [33]. The frequency and intensity of the training protocols appear to be the two factors most influential to the level of transfer.

2.3.2 Bilateral Transfer of Motor Skills

Knowing how the human body generalizes motor skills is important for developing training methods and rehabilitation techniques [35]. Previous studies have shown that damage in either the dominant or non-dominant hemisphere is capable of producing deficits in the control of the ipsilesional limb [36]. Lesions in the dominant hemisphere have been shown to affect the initial ballistic component of reaching and movement speed; whereas, lesions in the non-dominant hemisphere have been shown to affect the final position of reaching movements. All of these findings suggest that patients with lesions in certain locations of the brain will have a harder time achieving certain aspects of motion, irrespective of the rehabilitation protocol.

The bilateral transfer of motor skills can occur symmetrically or asymmetrically. Previous studies reveal that asymmetry in interlimb transfer can depend on several factors, such as the sequence of the arms in learning the task, the movement parameters being examined, and the type of transformation used during training (i.e. visual-motor rotation vs. dynamic transformation). Groups who have studied the ability to anticipate forces with the upper limbs have found the skill to transfer either in both directions or not at all.

2.3.2.1 Upper Limb Motor Skills

Several studies have used writing and key pressing tasks to study bilateral transfer. When Halsband et al. asked subjects to trace ideograms they found that subjects were able to transfer the skill from the right hand to the left hand [30]. In another study by Parlow and Kinsbourne, it was found that the tactual recognition of brail letters transfers in left handed individuals who write in an inverted position [37]. A key-pressing study by Taylor and Heilman revealed that subjects who performed a key-pressing task were able to transfer the skill from the left arm to the right arm [38]. To the contrary, Rand et al. found that the ability to press buttons sequentially, transfers in both directions[39].

Teixeira et al. found that when exposing subjects to an anticipatory timing task, they were able to transfer the skill to both arms[40]. However, subjects exposed to a task that required fine force control of the wrist showed transfer only from the dominant arm to the non-dominant arm. Salimi et al. did not find bilateral transfer when subjects used the index and thumb to lift objects at their center of mass [41]. Morton et al. found that when subjects caught balls of different weights with as little displacement as possible they were able to perform better with the second arm, despite which arm went first [42].

Researchers have explored the patterns of transfer for various components of reaching and have sought to determine whether the brain seeks to control movements via an intrinsic or extrinsic coordinate system. Many of the outcomes have been consistent. However, there is some debate over the control mechanisms responsible for the observed outcomes.

Sainburg and Wang examined the patterns of transfer of final position and trajectory information between arms during a multidirectional reaching task [43]. Fourteen right dominant subjects were separated into two groups. One group trained with the right arm first and the other trained first with the left arm. Trajectory information transferred better from the left arm to the right arm, and end-point data transferred better from the right arm to the left arm.

Sainburg and Wang conducted another study to determine the effects of initial training with one arm on the performance of the relaxed arm when the two arms were located in different workspaces[44]. Twelve right-handed subjects without neurological impairments were asked to reach perform a reaching task. Three measures of performance were calculated: hand-path direction error at peak tangential arm velocity and at peak tangential arm acceleration, and final position error. Both arms showed improvement in the initial direction control at maximum velocity and maximum acceleration following opposite arm adaptation. Neither arm showed improvement in the final position following opposite arm adaptation.

According to the authors, this suggests that final position information only transfers between the arms when target locations and associated learned hand positions have identical absolute coordinates. When the task is performed in a shared workspace, the motor control system is presented with redundant solutions. Hence, an executive decision-making process must be evoked to allocate task resources to the limb most proficient in the control of the task.

Crisimanger et al. studied bilateral motor transfer in an extrinsic and an intrinsic coordinate system [35]. Subjects were trained to reach for one of three targets while exposed to a clockwise curl field (intrinsic), a counter clockwise curl field (extrinsic) or no field (control). All subjects were tested on a counter clockwise field after the training session. Performance was quantified by computing an adaptation index based on the final position error of the hand.

Bilateral transfer occurred only from the dominant to nondominant arm for those in the extrinsic group. However, no significant transfer occurred for any of the other groups. A split-brain patient, who previously had a complete resection of the callosal fibers, experienced transfer only from the non-dominant to dominant arm under the extrinsic condition. The authors believe that their results occurred because the hemisphere used to learn the task during movements with the contralateral arm also controls the ipsilateral arm.

2.3.2.2 Transfer of Lower Limb Motor Skills

Haaland et al. studied the transfer of a newly learned complex locomotor movement pattern to the mirror condition, and the side specific differences between the right and left leg[30]. Thirty-nine right dominant male players were randomly assigned to an experimental or a control group. Three soccer performance tests (dribble, receiving, and passing) were carried out on a grass soccer field in good weather conditions using either leg. Two standardized foot tapping tests were performed, using the preferred and the non-preferred leg consecutively. The experimental group trained for eight weeks, in all parts of their soccer training except full play, using the non-dominant leg.

Training the non-dominant leg was able to enhance the performance of the dominant and non-dominant leg. The post-test performance of the training group's non-dominant leg matched the pre-test performance of the dominant leg. The control group did not show improvement in either leg. The author believes that training the non-dominant leg leads to increased attention to training, which in turn results in increased learning.

Van Hedel et.al studied the legs ability to avoid obstacles during gait[45]. Twelve healthy subjects were recruited and divided into two groups, RL and LR. They were instructed to step over an obstacle on a treadmill with the lowest clearance possible. Data on the movements were obtained via EMG recordings and force plates to signal toe off and hill strike. Three changes in performance were analyzed: adaptation, training effect (improvements between successive trials with the same leg), and transfer effect.

There were no significant differences in the cross correlations for runs two and three. However, evidence of both a training effect and transfer (between trials one and three) were present. Adaptations occurred in rectus femoris muscle activity for the three successive runs. The amplitude of the knee joint movement trajectories showed strong adaptation changes for the first run, and less pronounced changes for runs two and three. Runs two and three differed significantly from run one, but did not differ significantly from each other, indicating that both training effect and transfer had occurred.

During the second and third trials, optimal foot clearance was established from the start. This may have occurred because the subjects were trying to make the lowest clearance possible. The adaptation changes of this measure did not differ significantly between the second and third runs, which indicated that transfer had occurred.

In contrast, the amplitude of the ankle joint movements showed no clear adaptation changes during the three runs. The cross correlations of the three runs did not differ significantly, indicating that neither training effect nor transfer had occurred. It is possible that this occurred because the knee is the primary joint used for obstacle clearance. The author hypothesized that the lack of adaptation changes for the ankle joint trajectories might be due to the great variability and complexity of the ankle joint movement.

Furthermore, the analysis used to assess the changes in joint trajectories may not have been sufficient. The author stated that the high level of transfer between the legs was not surprising because an inter-leg coupling, substantial for human locomotion is present in the lower limbs. Observations by Hiebert et al. indicate that the flexor half centers of homologous limbs inhibit each other if assessed during walking, whereas the extensor half centers are not directly coupled with each other.

2.3.3 Theoretical Basis of Learning

Both physical therapy and movement science holds that repetition is a major player in motor learning. Sir Fredrick Bartlett once made the statement about tennis "When I make the stroke I do not as a matter of fact produce something absolutely new, and I never repeat something old." Movement scientists believe that cognitive processes that occur during repetition make a larger contribution to motor learning than the physical act of repetition. This is supported by studies that show considerable motor learning can occur by merely observing in the absence of overt physical practice[29]. Watching an unskilled person perform a task is more beneficial than watching a skilled person because the learner is able to view the problem-solving process when observing the unskilled person.

Schmidt's schema theory states that movement entails two fundamentally different representations: invariant characteristics and parameterization schema. Invariant characteristics are the representation for basic details of movement such as ordering, phasing, and relative forces of each action. The parameterization schema is the representation for specific details, such as the individual muscles or muscle groups to be used for a task, as well as force, time, and space details. Implementation of these parameters is based on the schema, which is a learned rule for movement generation. Schema theory provides a more flexible view of motor skill representation, and is used as the basis of discussing the role of cognitive processes in practice.

2.3.3.1 Effects of Practice on Learning

Shea and Morgan conducted a study in which a blocked order, (all practice trials on one pattern were completed before practice on another pattern was undertaken) and a random order (the practice trials on all three of the patterns were conducted intermittently) approach was used[29]. The blocked order group outperformed the random-order group at the end of practice. However, when retention tests were performed 10 minutes and 10 days later, the random-order group performed better than the blocked-order group. The random-order group also performed better than the blocked-order group on tests of transfer using a novel more complex version of the movement pattern.

According to Battig this effect is in keeping with the view that cognitive activities of the performer play an important role in learning. Bernstein addressed the role of repetition years ago, and concluded that

> "practice, when properly undertaken, does not consist in repeating the means of solution of a motor problem time after time, but in the process of solving this problem again and again by techniques which we changed and perfected from repetition to repetition".

Bernstein's insight implies that the evaluation of feedback as well as the formulation of a new plan is essential in these problem-solving activities.

Jacob's analogy: If someone was asked to add two large numbers, the answer would be obtained by going through the process of addition. However, if they were asked to recall the sum again a few seconds later, the solution would be obtained through memory, not as the result of addition. However, if a significant period of time elapses, the process of addition would again be used. The solution can be either a product of a process or retrieval from memory.

2.3.3.2 Transfer of Learning

Two hypotheses have been used to explain the asymmetrical transfer that has been observed in arm reaching studies: (1) learning is asymmetrical because one arm or hemisphere system is superior to the other in learning certain tasks. (2) When learning occurs, each arm has asymmetric access to the memory resources. Both models predict transfer in only one direction. Wang and Sainburg propose that both limbs have equal access to the memory resources. However when the arms share a workspace, a competition between the two hemispheres occurs, resulting in asymmetric transfer (Figure 3).

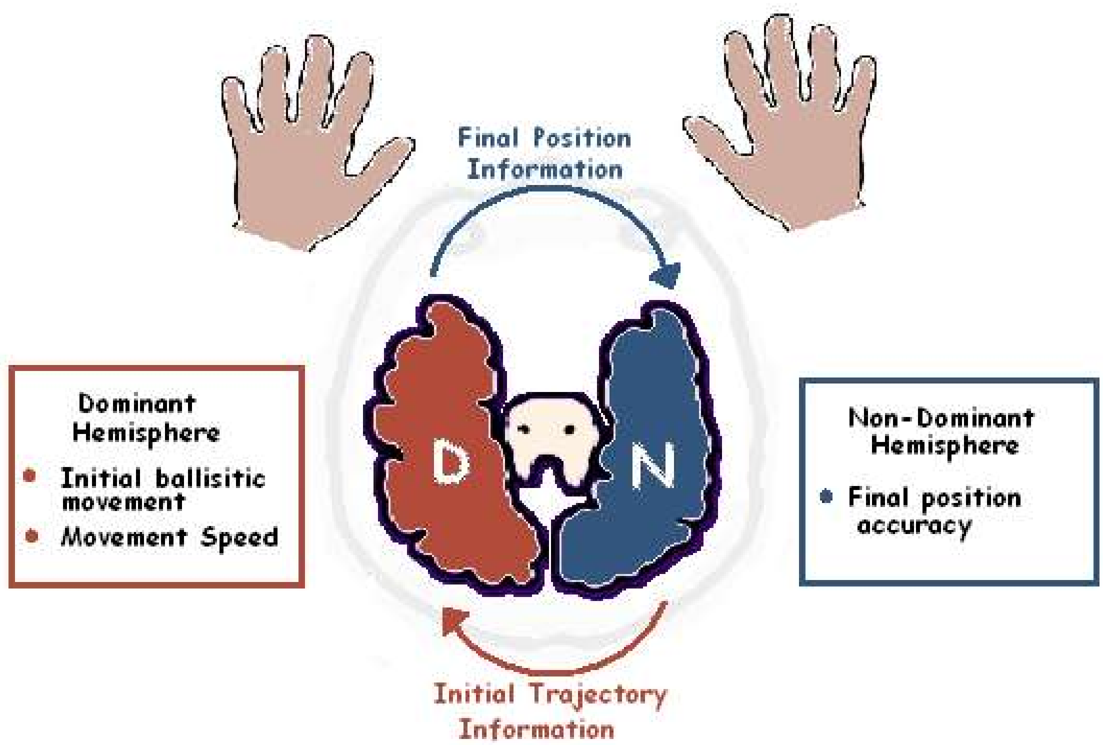

Figure 3 Cross-Hemispheric Learning Diagram.

In contrast Haaland et al. believe that been previous evidence indicates that the dominant hemisphere may have a significant role in controlling the nondominant arm via ipsilateral projections but not vice versa[35]. Therefore, it is possible that when a motion is learned by the dominant arm adaptation is primarily in the dominant hemisphere, which can later assist in performing the task with the nondominant arm. However, adaptation occurring in the nondominant hemisphere cannot be used to control the ipsilateral arm.

2.4 Force Myography

Force myography (FMG), also known as myokinetic imaging, is the measure of surface pressure exerted on the skin as a result of muscle activation. FMG employs the use of force sensing resistors (FSR) in either a voltage divider or Wheatstone bridge circuit to measure muscle surface pressures. A force-sensing resistor is made of a resistive material applied to a film that makes an electrical path between two sets of conductors on another film. When a force is applied to the sensor, a better connection is made between the contacts, which in turn decreases the resistance of the material[46].

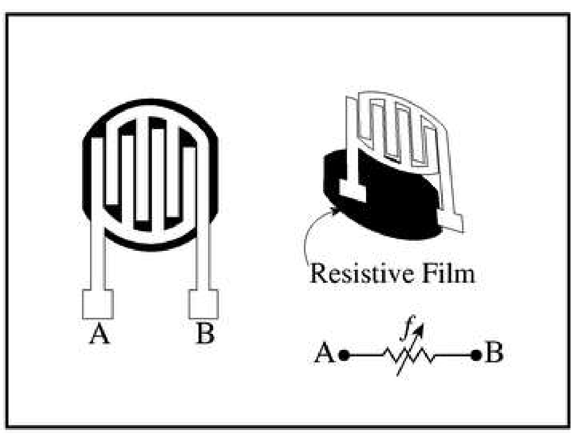

Figure 4 FSR Schematic [46].

When placed over muscles, force sensitive resistors are capable of detecting concentric and eccentric contractions. During concentric contractions, the muscle shortens and increases in diameter; and during eccentric contractions, the muscle lengthens and decreases in diameter. For muscles closer to the surface, these changes in diameter are visible by eye and can be detected with force sensitive resistors. To detect change in muscle diameter with the FSR, the sensors should be secured to the skin with a band or tape that applies a constant pressure. Concentric contractions will cause a decrease in resistance and eccentric contractions will cause an increase in resistance.

Force myography is currently in its infancy and has been implemented by only a few groups. The most optimal way of securing the FSR interface to the skin is still being determined. Amft et al. created a simple circuit that allows for the detection of isometric activity of the lower arm due to grasp activity [47]. This group tried two FSR securing approaches. The first approach was to secure the FSRs to the skin using tape over two large muscle bellies, which allowed FMG signals to be recorded from individual muscles. Furthermore, they were able to show that voltages from the force sensing resistors during four gross motor activities correspond to EMG threshold detector for two individual muscles (figure 5).

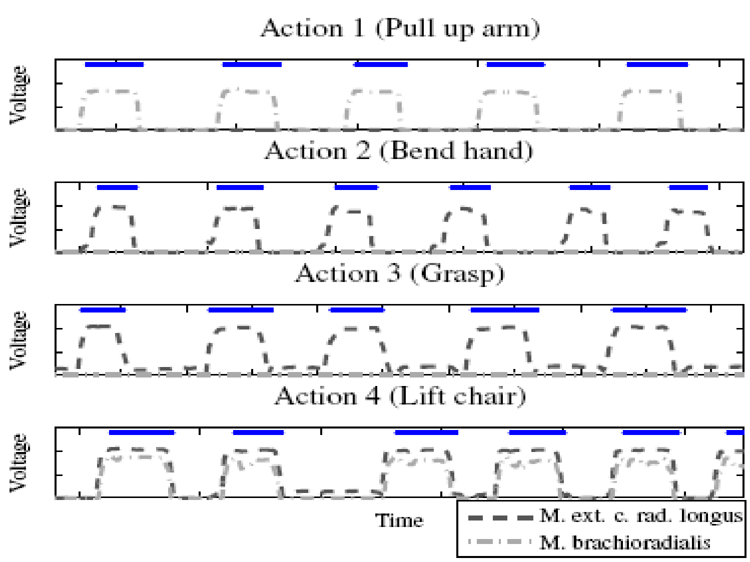

Figure 5 Signals of FSR Sensors placed over the M. extensor cardi radialis longus, and the M. brachio radialis (black) and the corresponding EMG switch (blue)[47].

The second approach used by this group incorporated the use of a fabric developed by Lorussi et al. whose electrical resistance depends on stretch. This fabric is more analogous to a strain gauge, and exhibits hysteresis, making it difficult to obtain repeatable measures. Unfortunately, the fabric was unable to detect the activation on individual muscle groups. If positioned incorrectly the material will detect both agonist and antagonist muscle activations, which may lead to decreased signal detection as illustrated in figure 6.

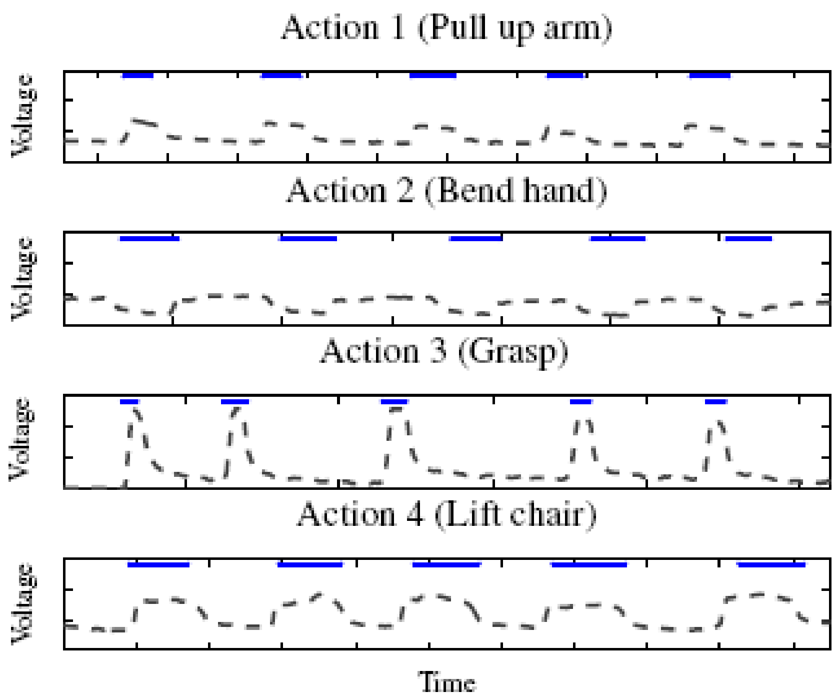

Figure 6 Signals of fabric stretch sensors placed over the M. extensor cardi radialis longus, and the M. brachio radialis (black) and the corresponding EMG switch (blue)[47].

Yungher, et al. used an array of FSRs mounted on a cuff to detect pressures in the leg of a single female subject as she walked on a treadmill [48]. This approach allows the sensors to be quickly and repeatedly applied to the surface of the skin. EMG and FMG data from the leg were summed and filtered to yield composite signals. The results indicate that the FMG signal corresponds to both the foot switch signal (muscle activation) and the actual EMG signal. This occurs because there is more than one motor neuron that can be used to activate the muscle, but very limited ways a muscle can contract to make a specific movement.

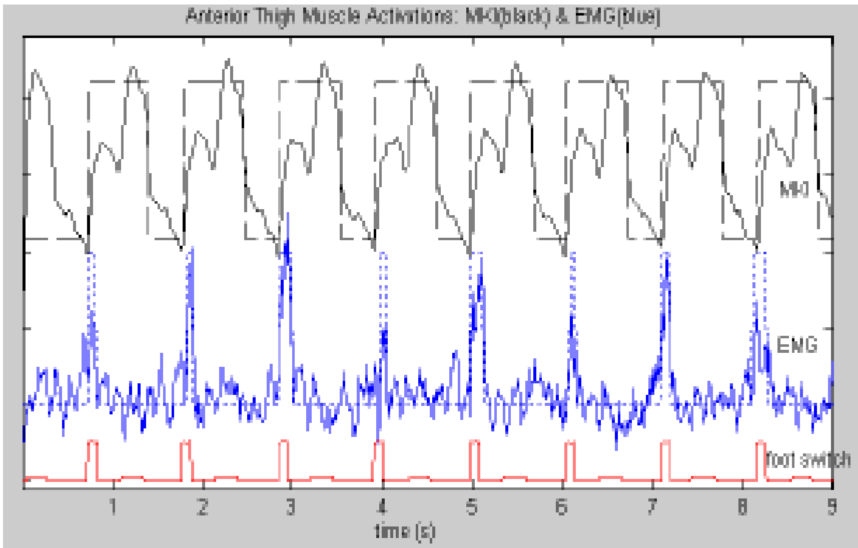

Figure 7 FMG and EMG data recorded while walking on a treadmill [48] .

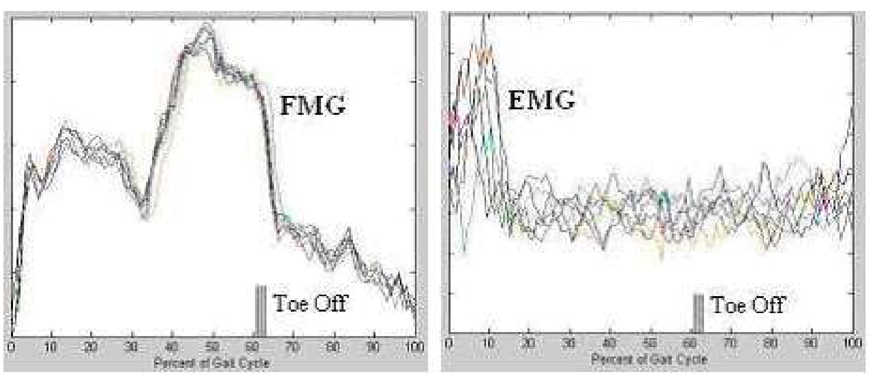

Figure 8 Repeatability of FMG and EMG data while walking on a treadmill [48].
It was found that the FMG signal produced a more repeatable signal for each gait cycle.

Force myography shows promise of being an easy reliable method of detecting muscle activation. One weakness of this technique is that it can only detect activation of the external muscles; but this is still a problem for EMG as well. FMG requires less amplification and data processing than EMG, which will be a very valuable benefit for real time applications.

CHAPTER 3

ROMAR DEVICE DEVELOPMENT

3.1 Joint Measurement Devices

This section will explore some of the devices used to measure joint movements. There are many ways to measure human joint movements including timing devices, videography, goniometry, tilt sensors, and accelerometers. All methods have been or are currently being employed by researchers and doctors. The method chosen by any given practitioner is usually driven by the desired accuracy, and the cost of using the method.

Timing devices and videography are the two least invasive forms of joint measurement. Timing devices such as stop watches, counters, photoelectric cells, and real-time computer clocks are used to record the speed of human body parts by measuring the time taken to travel a fixed distance. Videography and computerized optical systems are a more advanced technology that is used to easily digitize select anatomical markers. Standard digital video cameras can be relatively inexpensive and readily available to the consumer, however high precision systems typically used in gait laboratories can be expensive.

The electrogoniometer is a device that uses a potentiometer placed at the joint center to measure angles. The potentiometer changes the resistance to the flow of electric current in a circuit and can be attached to an oscilloscope or analog to digital board to continuously record movements. The advantages of electrogoniometry include the ability to record the action at the joint when it is not visible to the observer, and its ability to instantaneously portray angular displacement. Recorded measurements can later be used to compute angular velocities and accelerations.

Tilt sensors are typically used for gross angle measures. However, they have been incorporated into more precise measurement devices. An Australian group developed an inclinometer based device shown in Figure 9, that measures the deviation of the base of the foot from horizontal [49]. This device was created by mounting a bicycle wheel next to a footplate. The center of rotation of the footplate and wheel were aligned, allowing the two to rotate together. A rope was extended from the end of the footplate, around the wheel and over a pulley. A weight was hung from the end of the rope to provide a constant 17 Nm dorsiflexion torque.

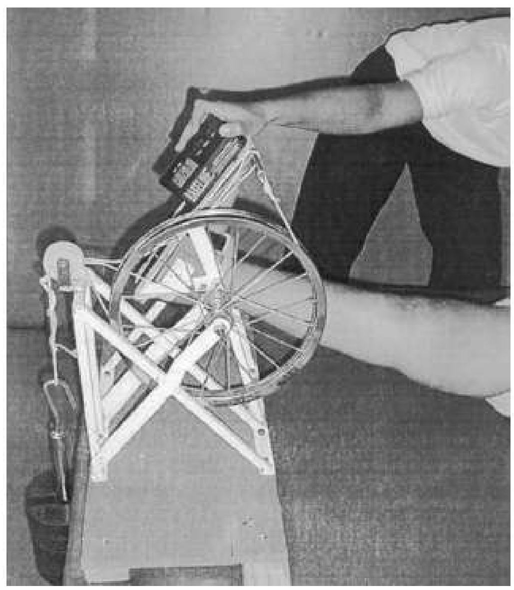

Figure 9 Inclometer Based Device.

This device was created to enable physicians to quickly and easily measure ankle mobility in their daily practice. A study was conducted to determine the effectiveness of an ankle intervention at a Sydney spinal injury unit. The device was able to minimize measurement errors by using an inclinometer instead of a manual goniometer. However, according to the authors, the adjustability of the foot platform can lead to misalignment of the two centers of rotation and thus become a source of potential error.

A Seattle group created a goniometric device called the equinometer [50], that consists of a potentiometer to measure the angle of the ankle and a load cell to monitor and control the amount of torque applied to the foot. It is secured to the patient with a shank component and a foot component. The shank component is fit and aligned to the subject's shank with soft Velcro straps. Device alignment is simplified by using a fixed length parallelogram transfer apparatus attached to the shank component, which permits translation of the foot's axis of rotation without a change in plantar flexion or dorsiflexion. The foot component is comprised of an aluminum plate with a plastic grip that can be positioned under the metatarsal heads. A load cell between the plate and the grip measures the force applied to the metatarsal region of the subject's foot. The hind foot is held in the neutral position in a stiff cuff because its position can be influenced by the subtalar axis. A power supply and signal conditioner unit powers and amplifies the potentiometer and load cell.

Figure 10 Equinometer.

The Equinometer was used by two experimenters to measure the angle of dorsiflexion at 15N-m of torque on five unimpaired subjects. Each subject performed 18 measurements divided into 6 sets. Each examiner removed the device between each set. Prior to the first trial the examiner applied a 30 second passive stretch followed by a 30 second resting period. The measurement errors were found to be less than 1.36 degrees, which is the acceptable clinical level. The authors believe that this error may be due both to the variability between subjects and to the placement of the device by the different experimenters. Furthermore, the authors were concerned that repeated measurements and stretching may affect the results, and recommend a rest period of a few minutes between each trial.

The S720 miniature joint angle shape sensor is a commercially available device created by Measurand Inc. typically used to produce life like animations [51]; however, it may have applications in joint angle measurement. The joint angle sensor can be used to measure any 1 degree of freedom joint such as fingers or toes. It can be mounted with a reusable, flexible adhesive polymer, and can be used directly on the skin or on a glove or biocompatible tape. The sensor is stand alone, and needs a power supply of 5 to 15 volts. There is an optional 12-bit data acquisition system with windows software.

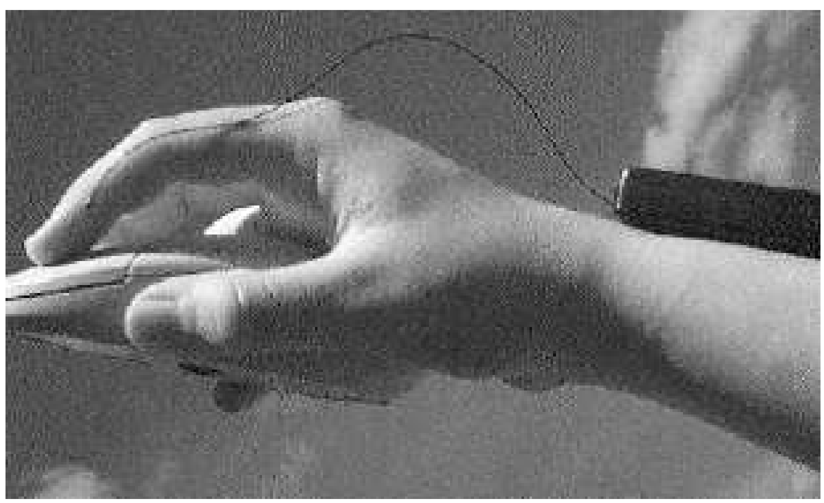

Figure 11 Miniature Joint Angle Sensor.

The devices mentioned are all capable of providing reasonably accurate measurements of joint angles; however they were not suitable for this project. The inclometer based device does not have a computer interface, making it unsuitable for cursor control. The Equinometer, on the other hand, does interface to the computer, but it only measured one dimension of ankle movement. The Miniature joint angle sensor, offers the same disadvantages as the equinometer, although it is possible to use more than one sensor to capture the necessary degrees of freedom. Unfortunately, these sensors are expensive, approximately $900 a sensor. Consequently, it was decided that it would be best to design a device capable of interfacing with the computer and measuring ankle motion in two dimensions.

3.2 Platform Design Parameters

The design objective was to create a device capable of tracking ankle movements while hindering movement as little as possible. Several design options such as a trackball, joystick, and car pedal based device were considered. Since the device would be used on the lower limb by individuals with limited limb control, it was also important that the device would not break or shift when subjected to forces. Finally, it was essential that the sensing mechanism used is capable of giving accurate and repeatable measurements.

3.2.1 Mechanical Design

Initially, an industrial joystick was chosen because it is capable of providing more reliable goniometric measures than a gaming joystick. However, the industrial joystick was not as sturdy as the gaming joystick and fell apart during preliminary testing. Furthermore, the industrial joystick gave too little mechanical resistance, which caused the ankle muscles to fatigue quicker. Therefore the Logitech Attack 3 joystick was used. In order to overcome the reliability issue the circuitry of the gaming joystick was bypassed, and wires were attached directly to the goniometers to get the raw data from the original goniometers. A metal brace was used instead of putting the joystick into a whole drilled into the wooden base. Washers were placed around the joystick before inserting it into the metal brace to prevent the joystick from wobbling.

3.2.2 Electrical Design

The Logitech joystick automatically makes the position of the joystick at the time it is plugged into the computer the zero point. This is unsatisfactory, because measurements will not be repeatable if the device is disconnected or if the computer is turned off, unless the device is in the same position every time. To do this would be tedious, and a possible cause of error, therefore another method of tracking the joystick's position was developed.

The raw joystick potentiometer outputs were wired to the C8051F020 Silicon Laboratories microprocessor development board (SIDB), to communicate with the computer. The microprocessor code was adapted from example code provided by Cygnal Integrated Products Inc. This program utilizes ADC0 which is a 12 bit ADC that contains nine channels. The start-of-conversion to measure the voltages on AIN0 was triggered by Timer3 (16 bit timer) overflows. Data in registers were left-justified. Low power tracking mode was used to insure that minimum tracking times were met when the ADC0 channels were changed. Tracking was continuous when the ADC was enabled unless a conversion was in progress. The internal voltage reference was driven on the VREF pin, which was set to 2430 millivolts. The internal temperature sensor was activated. Serial port communication was handled by an eight bit UART. Timer 1 was used to control the baud rate, which was set to 115200 bits per second. The system clock used a 22.1184 MHz crystal as the clock source.

A decimation filter was added to the ADC0 Interrupt service routine to increase the accuracy of the measurements. This filter uses an oversampling and averaging technique to increase measurement resolution, eliminating the need to resort to a more expensive ADC[52]. It improves the signal to nose ratio and measurement resolution at the cost of increased CPU utilization and reduced throughput. The filter was implemented by taking 1000 ADC0 samples and storing them to a global array. The mean of the array was found, and the result was sent to the ADC output.

The SIDB did not have a resolution high enough to capture enough voltage levels necessary to map the voltage from the joystick to a cursor position. These shortfalls lead to choppy cursor movements. Furthermore, the voltage of the joystick's potentiometers fell into a very narrow range, which lies in the middle of the ADC's range. Therefore, it was necessary to build a supplemental circuit that would provide an offset, and then amplify the voltage. A differential amplifying circuit (Figure 12)was used to offset the joystick's voltage by approximately 1 volt and then amplify it by a factor of 10.

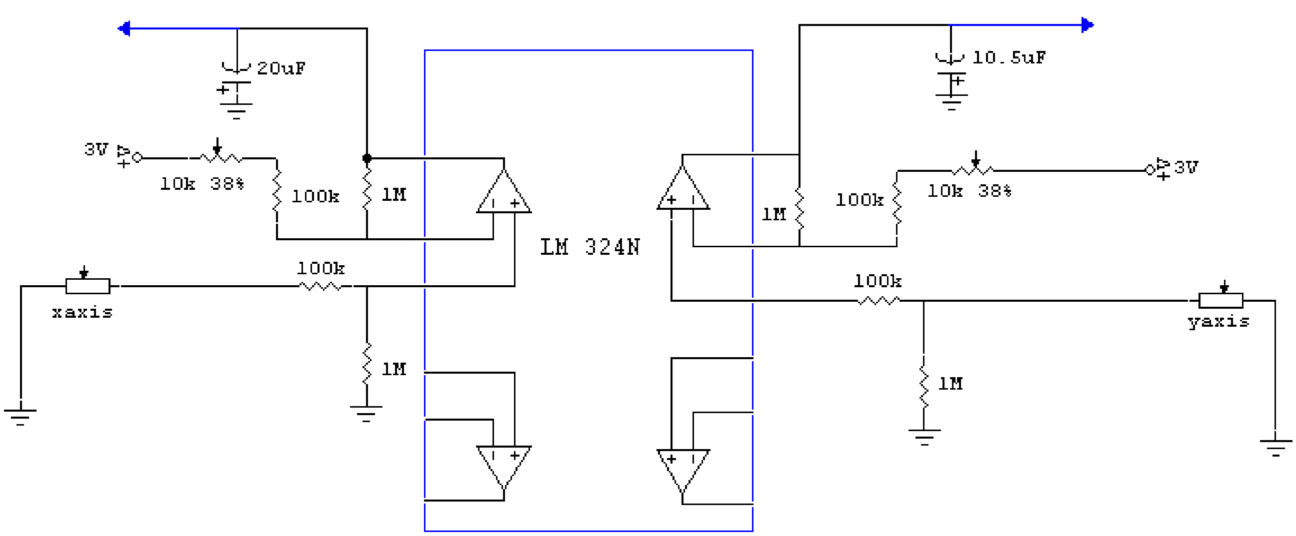

Figure 12 Differential Amplifying Circuit.

The SIDB was also used to power and collect data from the FSR cuffs used in the Cross Hemispheric Learning study. The FSR circuit consisted of a fixed resistor in series with two FSRs in series. The purpose of this circuit was to form a voltage divider pair allowing the voltage to vary as with the changing resistance of the FSRs.

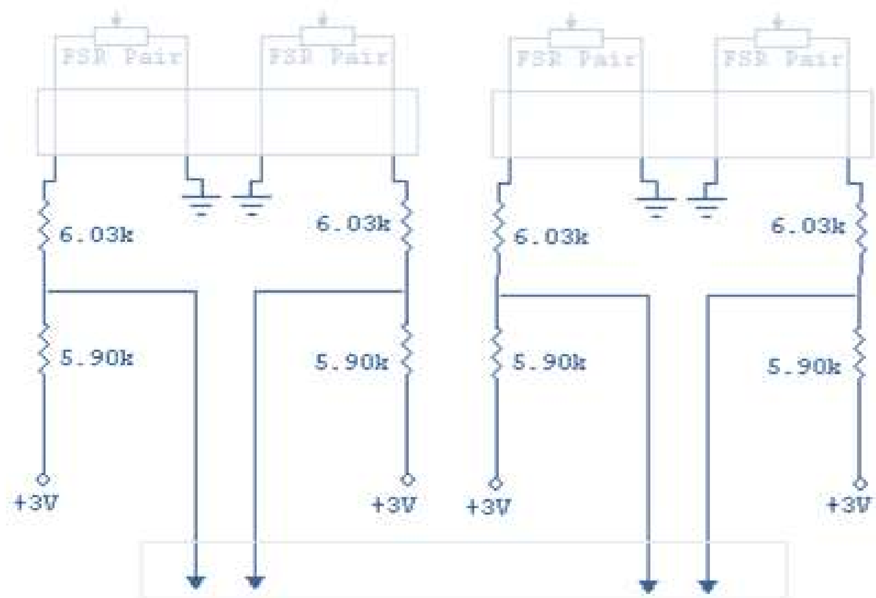

Figure 13 FSR Circuit.

3.2.3 Software

The software interface written in LabVIEW allows the user to play three games; a bird game that allows the user to practice dorsiflexion and plantar flexion while trying to avoid moving blocks; a boat game that allows the user to practice inversion and eversion while trying to avoid moving logs; and a mole game, which involves a combination of all four movements.

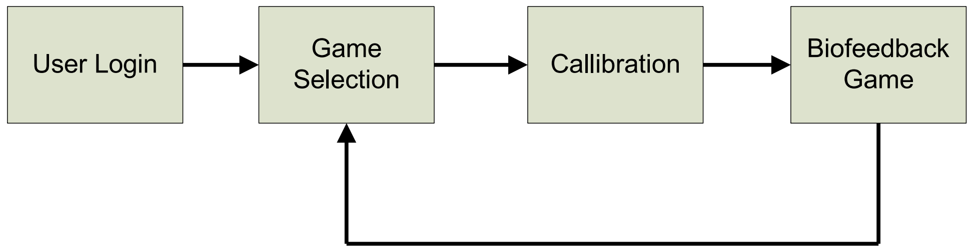

Figure 14 Block Diagram of User Interface.

Prior to each game, the system is calibrated to fit the users' maximum range of motion by having the user perform the required maximum movements. After calibration, the user can decide to use either continuous movement or sustained contraction (in which the user must sustain their maximum inversion, eversion, plantar flexion or dorsiflexion) to play the games. The user also has the option of decreasing the speed of the game if it is too difficult, or increasing the speed to provide a greater challenge.

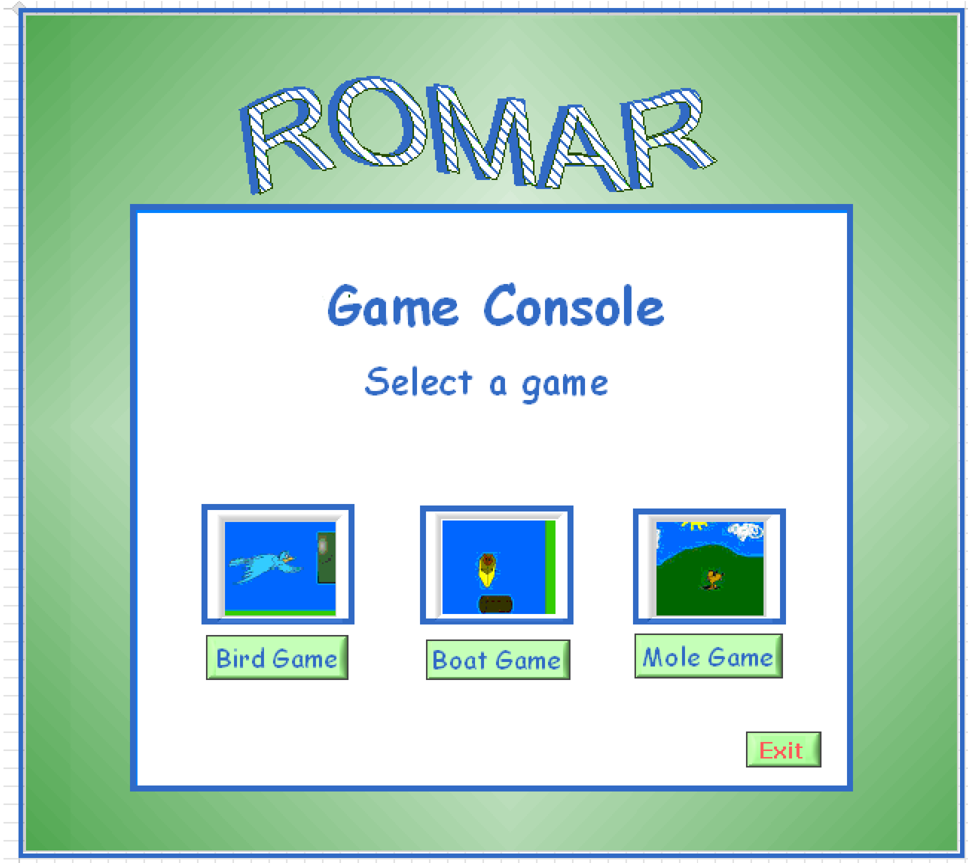

Figure 15. Screenshot of the ROMAR front panel is shown. The user has a choice of three games: The bird game is used to practice plantar flexion and dorsiflexion, the boat game is used to practice inversion and eversion, and the mole game is used to practice all four movements.

3.2.4 Preliminary Testing

A clinical test was performed with Version 1 of the platform. A LabVIEW program was created and used to record the joystick output and convert it to an angle measurement. Six trials were conducted, three trials for dorsiflexion (positive angles) and three trials for plantar flexion (negative angles). Using a protractor ,the joystick was initially positioned at ninety degrees (which is considered zero for the ROMAR device), using the rear most rivet (the approximate location of the ankle) as the pivot point. The joystick was then moved in increments of five degrees, in either the positive or negative direction. The joystick was held at each position for five seconds. A MATLAB program was used to determine the average raw output value for each angle. The best-fit line was calculated for the data using Excel, and later used to determine the angle the ankle moved while the foot was in the ankle device.

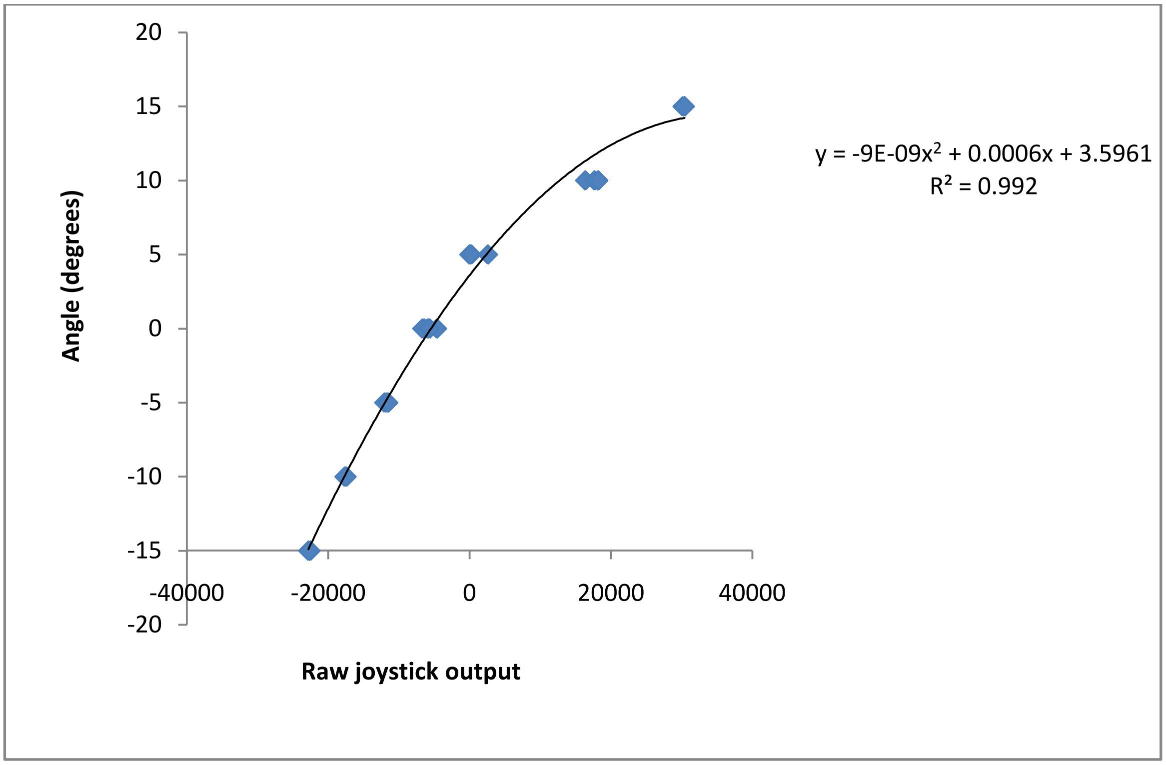

Figure 16Joystick Calibration Curve. A Goniometer was used to calibrate the raw joystick voltage signal to an angle measure. A quadratic curve was found to be the best-fit line.

The ROMAR device was then tested on six subjects, three hemiparetic stroke, and three unimpaired to determine feasibility of unipedal, bipedal, and monopedal protocols. The experimental setup consisted of two platforms placed side-by-side, approximately hip width apart. However, data was only recorded from the platform under the affected foot.

Subjects were asked to play seven rounds of the bird game. The first round was a practice round, meant to allow each person to become familiar with the game. The following six rounds were used to test the different conditions. The game was played at two speeds 30 flexions per min and 70 flexions per min (70 flexions per minute is equivalent to slow walking). For each speed, the game was played to test the following three foot movements, unipedal (one foot), monopedal (two feet in phase), and bipedal (two feet alternating).

The results were analyzed by calculating the mean velocity, acceleration, and normalized jerk (a measure of smoothness). Normalized jerk was calculated using the following formula [53]:

$$NJ = \sqrt{.5\frac{T^5}{D^2}\int j^2(t)dt}$$

where, T=the total elapsed time, D=the total distance traveled, and *j*=jerk (the third derivative of position). An ANOVA was performed to determine whether the difference in means were statistically significant.

Table 1Table 1 Mean plantar flexion and dorsiflexion values. summarizes the mean values calculated from the clinical evaluation of the ROMAR system.

			Unipedal	Monopedal	Bipedal
Plantar flexion	Healthy	Flexion Range	16.88	17.44	16.86
		Velocity	19.25	17.27	19.4
		Acceleration	-20.49	-15.95	-11.79
		Normalized jerk	8.5	2.11	4.66
	Stroke	Flexion Range	13.17	12.03	13.63
		Velocity	12.47	10.34	11.73
		Acceleration	0.28	-30.37	-30.02
		Normalized jerk	12.1	4.05	4.66
Dorsiflexion	Healthy	Flexion Range	12.6	9.54	10.74
		Velocity	18.51	19.17	19.57
		Acceleration	4.58	34.48	22.63
		Normalized jerk	2.86	3.48	4.57
	Stroke	Flexion Range	11.44	8.42	9.81
		Velocity	13.74	9.38	12.37
		Acceleration	4.58	34.48	22.63
		Normalized jerk	12.1	4.05	4.66

Table 1 Mean plantar flexion and dorsiflexion values.

An ANOVA revealed a main effect for the type (unipedal, monopedal, or bipedal) of movement ($p=.01$). To understand the cause of this main effect a pairwise comparison of the three movements were conducted with a Bonferroni correction; since there was an n of 6 the correction causes the new significant p value cutoff to be 1.67 ($.5/\sqrt{6}$). The range of motion was found to be significantly higher for the unipedal movements (13.52 °±1.60°) than for monopedal movements(11.86° ±1.96°) ($p=.02$). The range of motion for bipedal movements (12.76° ± 1.82°) was not significantly higher or lower than the range of motion for the unipedal or monopedal movements. According to the ANOVA, there was a significant interaction between movement and flexion ($p=.03$). A pairwise comparison revealed that the cause of this interaction was a significantly larger ($p<.01$) dorsiflexion range for monopedal movements (12.02° ± 2.26°) than for bipedal movements (8.98° ± 2.77°) . There were no significant differences in the range of motion between the stroke and healthy participants.

The velocity, acceleration, or normalized jerk did not prove to be significantly different between the three types of movements or the two flexions. There was however a significant difference ($p<.05$ for both) in the average velocities and normalized jerk of the two groups. The healthy group had a significantly higher velocity (18.86 ± 1.36) than the stroke group (11.67 ± 1.36). The stroke group had a significantly higher normalized jerk (11.42 ± 1.72) than the healthy group (4.31 ±1.72). There were no significant differences in acceleration between the two groups.

The results suggest that the three different movements did in fact elicit different range of motion behavior from the ankle. Monopedal movements which are not typically used in daily activities of living yielded the smallest range of motion. The lack of practice with this type of movement may be the cause of this reduced range. It is believed that the two groups did not show a difference in range of motion because the movements required by the game were well within the range of the two groups. It is possible that a game that requires the participants to make larger movements would show range of motion differences. As expected the stroke group tended to produce movements that were slower and jerkier than the movements of the healthy subjects. These findings illustrate that the ankle device and game are capable of capturing and quantifying the differences between an impaired and unpaired patient group.

3.2.5 Platform Version 2

The final design choice was a potentiometer based joystick capable of capturing joint angles in two planes. Dorsiflexion and plantar flexion motions were measured on the y-axis, and inversion and eversion motions were measured on the x-axis. A sandal attached to a plastic plate with rivets was used to secure the foot to the platform. A metal brace attached to a wooden base with screws was used to hold the joystick stationary. Rubber feet at the bottom of the wooden base were used to reduce sliding. Latch hooks were attached to both the plastic plate and the wooden place, in order to hold resistance rubber bands, in the event resistance is desired.

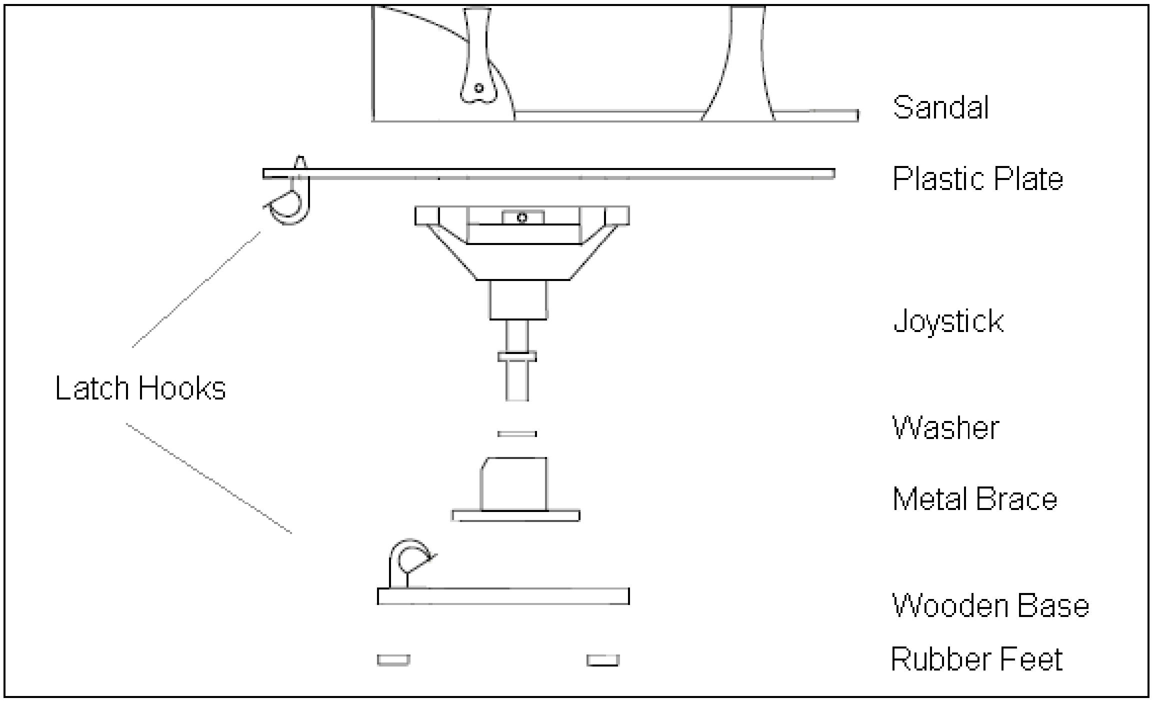

Figure 17 Final Platform Design. This is an exploded view of the final platform design, which includes a leather sandal attached to a plastic plate. The plastic plate is secured to a joystick that is held in place with a metal brace attached to a wooden base.

Improvements were made to the ankle platform to improve shortfalls seen during the testing of the version 1 platform. After version 2 of the platform was created calibration was performed once more to map the raw joystick output to an angle measure. The joystick angle correlation data is summarized in Figure 18. The correlation between raw joystick output and angle is best described by a first order polynomial. The correlation coefficient is .955 indicating that there is indeed a strong correlation between raw joystick input and angle, as expected.

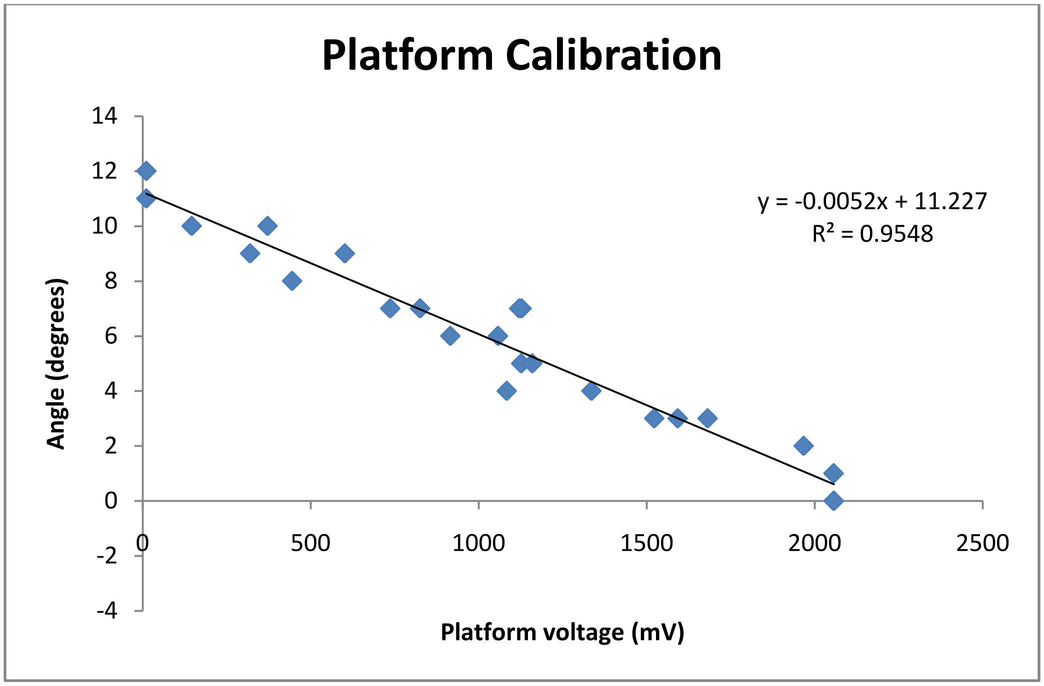

Figure 18 Ankle Platform Calibration. A Goniometer was used to calibrate the raw joystick voltage signal to an angle measure. A straight line was found to be the best-fit line.

Colored markers were placed on the base of the platform to indicate the center of rotation, a point located on the vertical joystick stem and a point located away from the center of the base. A camera was positioned to record the platform as it was slowly dorsiflexed and planar flexed twice as well as a computer monitor that displayed the raw output voltage throughout the movements. Video images were then imported into E-human, which was used to measure the angle between the platform and the horizontal. Frame numbers were used to match the raw voltage and angle values. Excel was used to determine the best fit line and the correlation value.

3.3 FMG Device

Another solution for capturing and encouraging joint movement is to combine a biofeedback interface with a mouse adapter, enabling patients to play preexisting mouse controlled games using wrist, elbow, or ankle flexion motions. In this project I sought to determine whether our current FMG interface is suitable for mouse cursor control. I hypothesize that the current FMG interface is suitable for mouse cursor control, and together with the mouse emulator will provide an engaging environment for physical rehabilitation.

3.3.1 Mechanical Design

The FMG cuff hardware is comprised of two force sensitive resistors mounted onto a plastic sheet and covered with a layer of foam. The sheets are then placed within a wrist support band to secure the sensors to the limb. The FSRs are connected to a silicon laboratories microcontroller board, which is used to power the force sensitive resistors, and to convert the analog voltage signals to a mouse input. A LabVIEW program was created to run games and record the FMG signals.

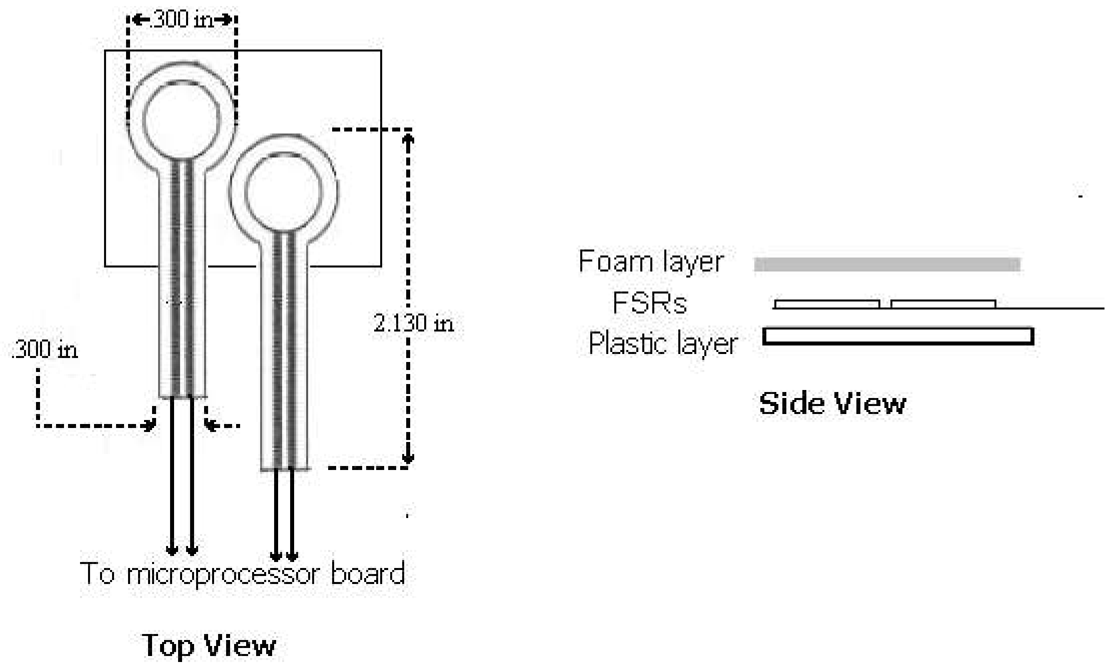

Figure 19 Schematic showing the construction of the FMG sensor. Two FSR were glued to a plastic sheet. A foam layer was placed over the FSR sensors.

The voltage to force relationship of the FSR was studied by placing a FSR in a voltage divider circuit and exposing the sensor to a varying force provided by an Instron machine. Two tests were conducted as the Instron varied forces at a low speed (speed=.01 in/min, hold time=10 sec) and high speed (speed=.01 in/min, hold time=.5 sec). Both speeds yielded similar hysteretic force-voltage curves.

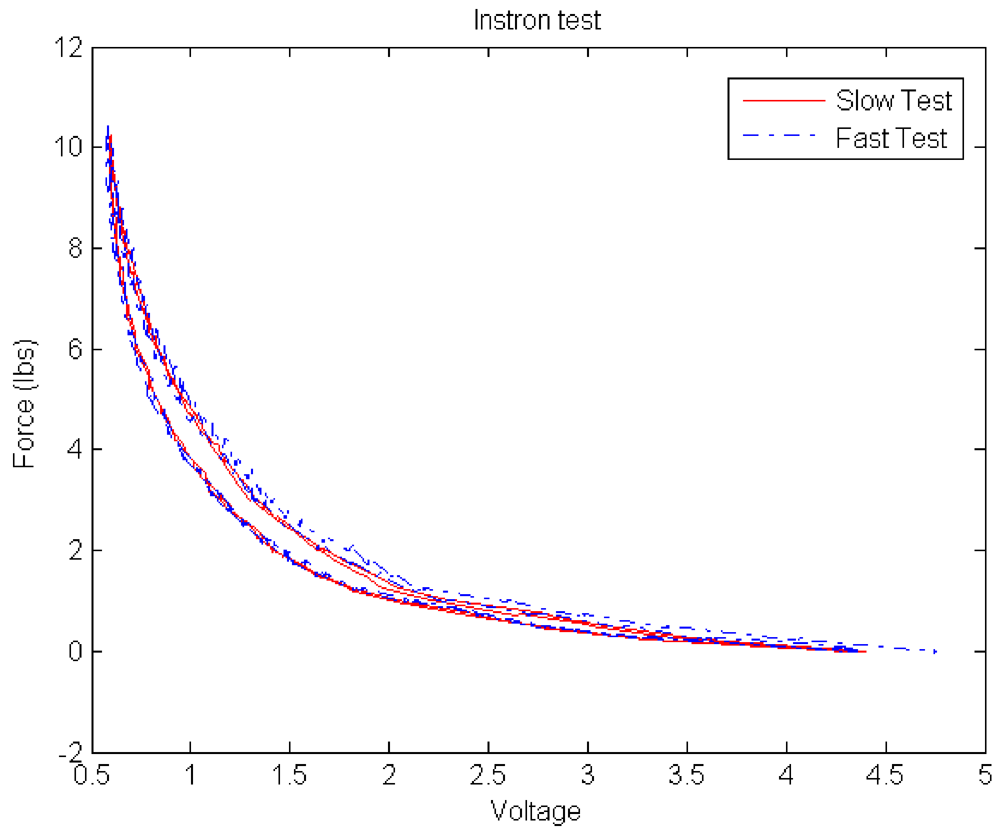

Figure 20 FSR Force Voltage Curve

3.3.2 Electrical Design

Two sensor configurations are used during testing. The one limb configuration (Figure 21) consists of two FSRs and one fixed resistor connected in parallel and mounted as described above. This parallel group is then placed into a voltage divider circuit with a fixed resistor. The output voltage, Vout, is the voltage across the fixed resistor. With this configuration the two resistors operate as one unit by both either increasing or decreasing the output voltage. When connected to the silicon laboratories board this configuration causes the cursor to move towards the right of the screen when pressure is applied to the sensor, and to remain at the left of the screen when no pressure is applied.

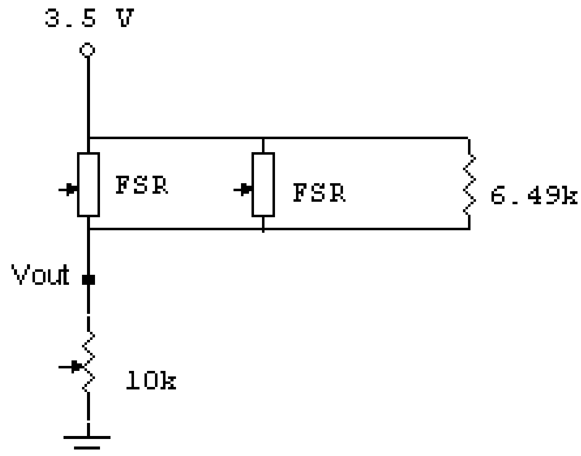

Figure 21: One-leg cuff. Relaxing causes max and exerting causes minimum.

The two limb configuration is constructed by placing two of the single limb groups in series. In this configuration, Vout is the voltage across the second set of FSRs and fixed resistor. When connected to the silicon laboratories board this configuration causes the mouse cursor to move towards the right of the screen when pressure is applied to the top set of FSRs, to move to the left of the screen when pressure is applied to the bottom set of FSRs, and to not move when the pressure applied to both sets are equal.

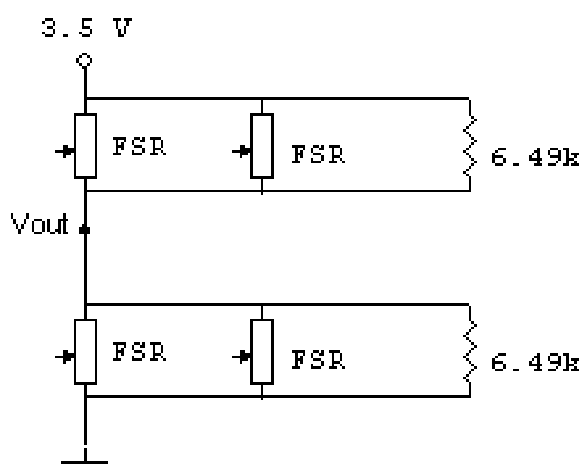

Figure 22: Two leg cuff. Activating top cuff causes minimum and activating bottom cuff causes maximum.

3.3.3 FMG Software

A LabVIEW interface was developed to run and display shockwave flash games while recording mouse cursor movements. With the help of an undergraduate student, six out of one hundred twenty mouse controlled internet games were selected and downloaded from the internet for testing (http://elegans.imbb.forth.gr/games/). Games suitable for this system had to be controlled by only two directions of movement. Since the games chosen are pre-written shockwave flash games they cannot be adjusted, and are therefore better suited for advanced users.

When the program is started, a practice screen appears allowing the user to become familiar with the device by practicing the desired motion. If necessary, gain adjustments can be made during this time by turning the knob on the box. When the subject is comfortable with the device and the administrator is satisfied with device performance the user can advance to the next stage by clicking the “play games” button on the screen with the mouse.

Figure 23 ROMAR Console Screen shot displaying the four top games.

The next screen displays the available games allows the user to select a game and either play the game while recording the mouse cursor movements or simply practice the game with no recording. The games available are a car racing game named “Ponky”, a boulder dodging game named “Watchout”, a bobsledding game named “Ice Racer”, a skateboarding game named “Trickmaster”, a water skiing game named “Wakeboarding”, and an airplane shooting game named “Pearl Harbor”. Recorded movements are saved in a text file, and can be opened at a later time for analysis.

3.3.4 Sensor Performance

Five unimpaired volunteers were instructed to flex and extend the elbow in the horizontal plane while the arm was supported by the MAST (Mechanical arm support tracker). Two sets of eleven repetitions were performed by each participant. Subjects’ arms were secured to the HARI chair, which was used to track elbow position with a goniometer. In addition, both EMG and FMG sensors were placed over the biceps and triceps. Before evaluating the signal the first and last repetition was removed to avoid late start and early finish errors.

All signals were collected at 2 kHz to ensure the Nyquist sampling theorem was satisfied. The EMG and FMG signals were rectified and passed through a 5^{th} order low pass filter with a 4Hz cutoff. Data were then resampled to reduce the number of data points by a factor of 200. Prior to performing the correlation all signals were normalized to have a minimum value of 0 and a maximum value of 1.

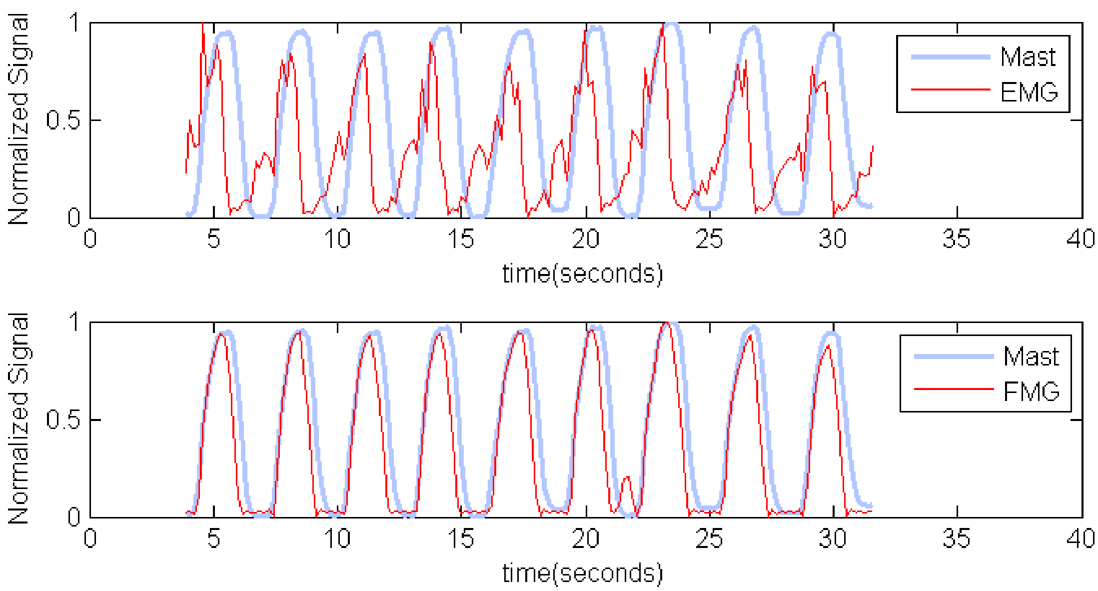

Figure 24 Shows the plots of the MAST, FMG, and EMG signals.

The FMG signal detected by the sensor placed over the biceps was able to match both the flexion and extension movements. Contrarily, the EMG signal from the electrode placed over the biceps was only able to match the flexion motion. Since the EMG signal is more localized, both signals were only correlated to the flexion component of MAST signal.

n=6	Mean rho	Std	p-value
MAST-EMG	.4693	.1930	.0165
MAST-FMG	.9240	.0492	.0000

Table 2 EMG and FMG Correlation to the MAST

There is a moderately strong correlation between EMG and joint angle (rho=.469, p<.05). FMG signals showed a strong correlation with the joint angle (rho=.924, p<.001). This test revealed that FMG signals have a stronger correlation to joint angle than EMG signals. However, the EMG signal is able to isolate flexion from extension.

A test with a single subject was conducted to determine how well the angle of ankle flexion correlated with a FMG signal recorded from the calf muscles (Figure 25). The subject was asked to flex the ankle 12 times while connected to the ROMAR ankle platform. A high correlation, rho=.81, was found between the two signals.

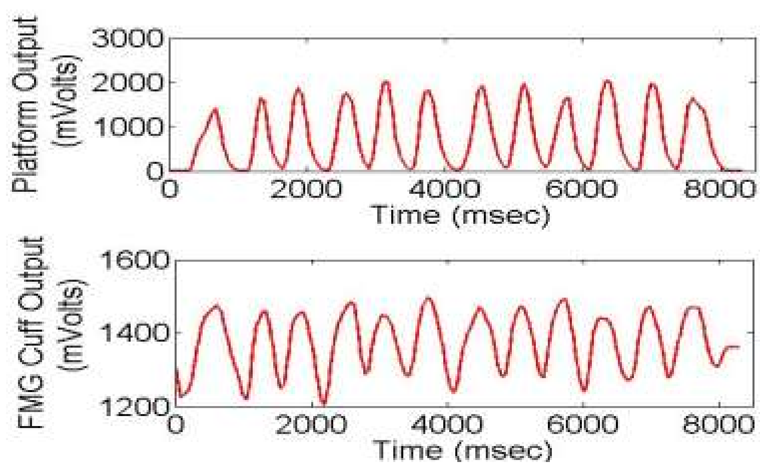

Figure 25 FMG and ROMAR platform signals recorded simultaneously from a single subject.

3.3.4 FMG Clinical Testing

3.3.4.1 Dual Cuff Testing

Six non-impaired subjects and one hemiplegic stroke subject were recruited to test the device. Subjects completed a questionnaire, designed to collect basic subject demographic data such as activity level and the amount of time spent playing computer games. Written informed consent was obtained from all participants, as approved by the Rutgers University IRB.

Figure 26 This figure shows the FMG Cuff placed on the lower portion of the calf muscle. Wires can be seen coming from the four FSRs placed within the cuff.

Each subject used two FMG cuffs, one for each calf. The mouse cursor position was proportional to the FMG signals received from each calf muscle. When the ankles were in a neutral position (feet flat on the floor), the mouse cursor remained in the center of the screen. As the left ankle was dorsiflexed the cursor moved to the left of the screen in proportion to the angle of flexion, and as the right ankle was dorsiflexed the cursor move to the right of the screen in proportion to the angle of dorsiflexion. Before playing the games, participants were familiarized with the device by controlling a cursor on the computer screen. Subjects were allowed to choose a game and either run a practice session to preview the game, or play the game while ankle flexions were recorded.

All health subjects were able to successfully control the mouse cursor using ankle flexion. The stroke subject had difficulty controlling the cursor using ankle flexions, without a mechanical support. Figure 27 shows the FMG signals of a stroke and healthy subject while playing on round of the Ponky car racing game. The y-axis shows the normalized mouse cursor position, with 0 indicating the left most cursor position and 1 indicating the right most cursor position. The stroke subject made fewer flexions and had a shorter round duration (36 seconds) than the healthy subject (100 seconds). In addition, the stroke subject made only one flexion with the affected leg and tended to use the un-affected leg for most of the cursor control.

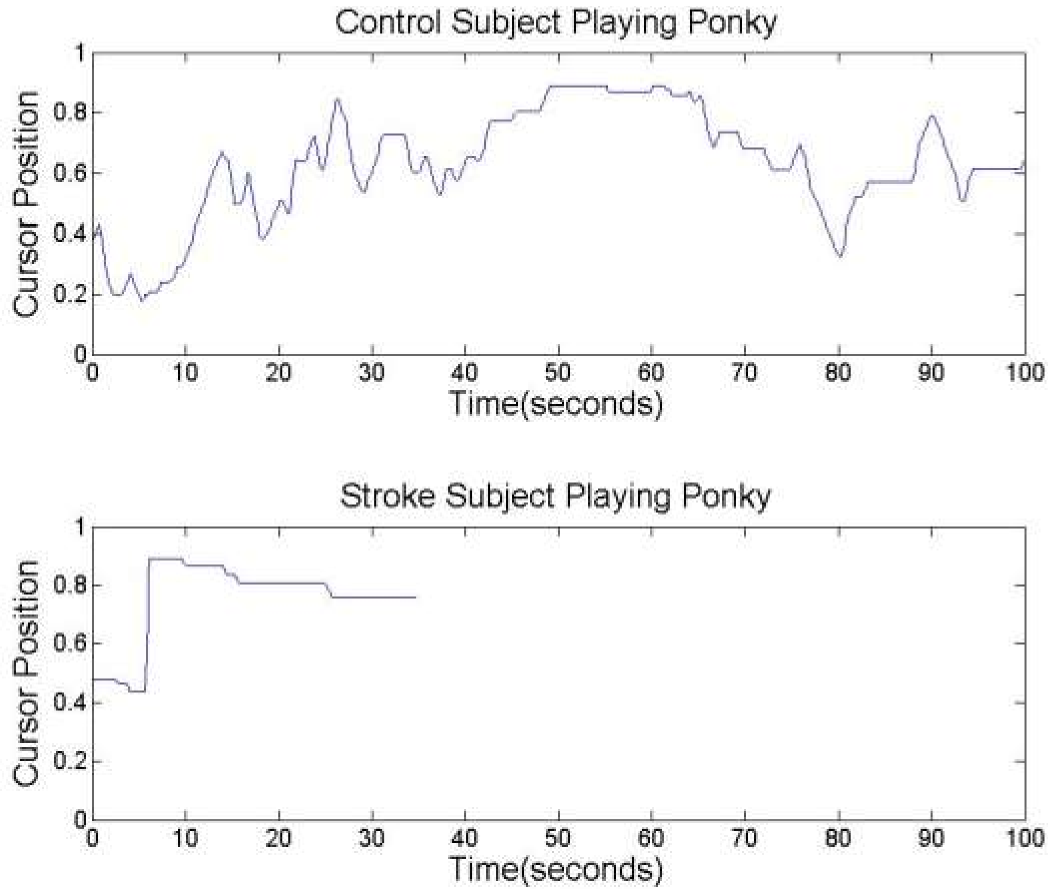

Figure 27 This figure shows sample ankle movement recordings of a healthy control (top) and stroke patient (bottom) playing the ponky car racing game. Plots are the normalized mouse cursor position which was proportional to the FMG signal of the calf muscle. The healthy control played this game for 100 seconds, making several flexions with both legs. The stroke patient played this game for only 35 seconds making flexions primarily with the left (unaffected leg).

The four most popular games, Trickmaster, Ice racer, Watch out and Ponky, were studied in more detail. The relationship between the length of time the games were played and the age of the subject was studied. The games were divided into two categories, sport and juvenile. Sport games were defined to be games that mimicked playing a sport, such as Trickmaster (skateboarding), and Ice racer(bob sledding). Juvenile games were defined to be games that looked more like cartoons, and did not attempt to mimic a real life situation, such as Ponky (a car racing game), and Watchout (a game in which the player dodged falling boulders). Although everyone played both types of games, four out of six of the subjects spent more time playing the juvenile games.

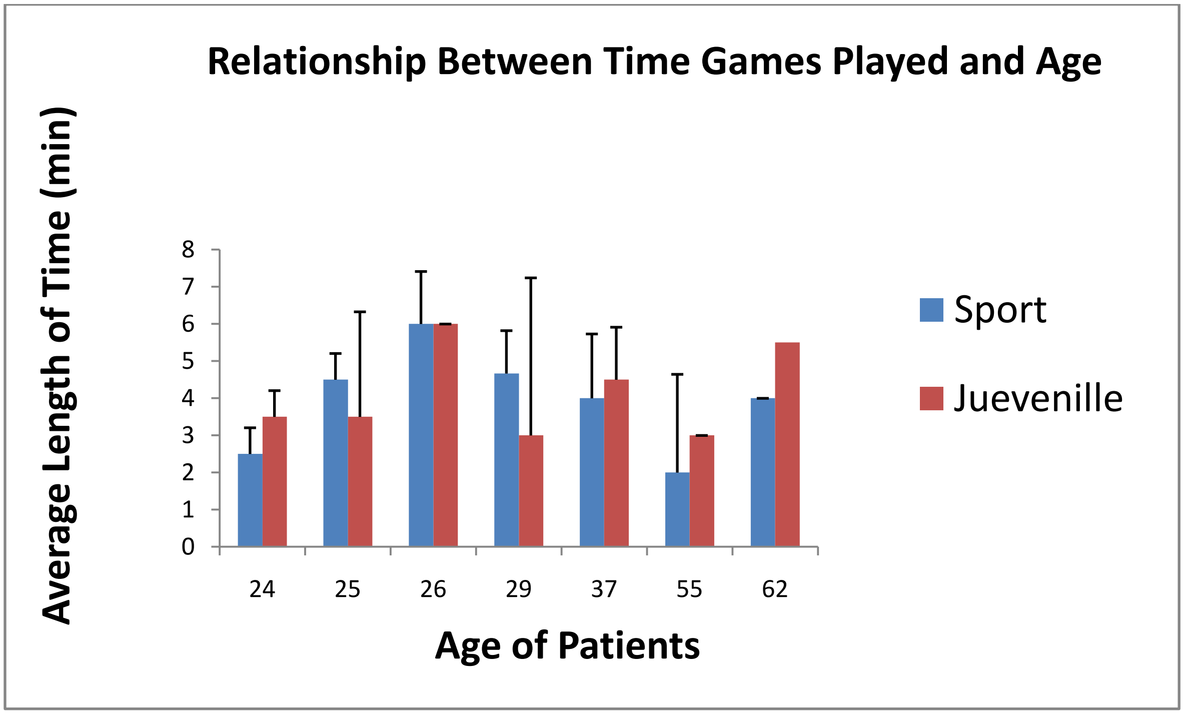

Figure 28 This figure shows the relationship between the time spent playing the two types of games and the age of the subject. Most subjects played the juvenile games longer than the sport games however both types of games were played by all.

MATLAB code was written to compute the range of cursor (ROC), velocity(Vel) and normalized jerk(Njerk) for each round played by the subjects. The cursor location was recorded in pixels with 0 representing the left most position of the screen and 1152 representing the right most position of the screen. Prior to all data analysis these values were normalized by dividing by 1152 to allow the values to range from 0 to 1.

To study the differences in each of the three measures between the unimpaired subjects and the stroke subjects, all of the unimpaired subject data were grouped and averaged for each game. Three sets of box plots were created to compare the measures of the two diagnostic groups for each game. Ponky, Trickmaster, and Ice racer were the only games studied in more detail, because the stroke volunteer chose to play only these three of the four top games.

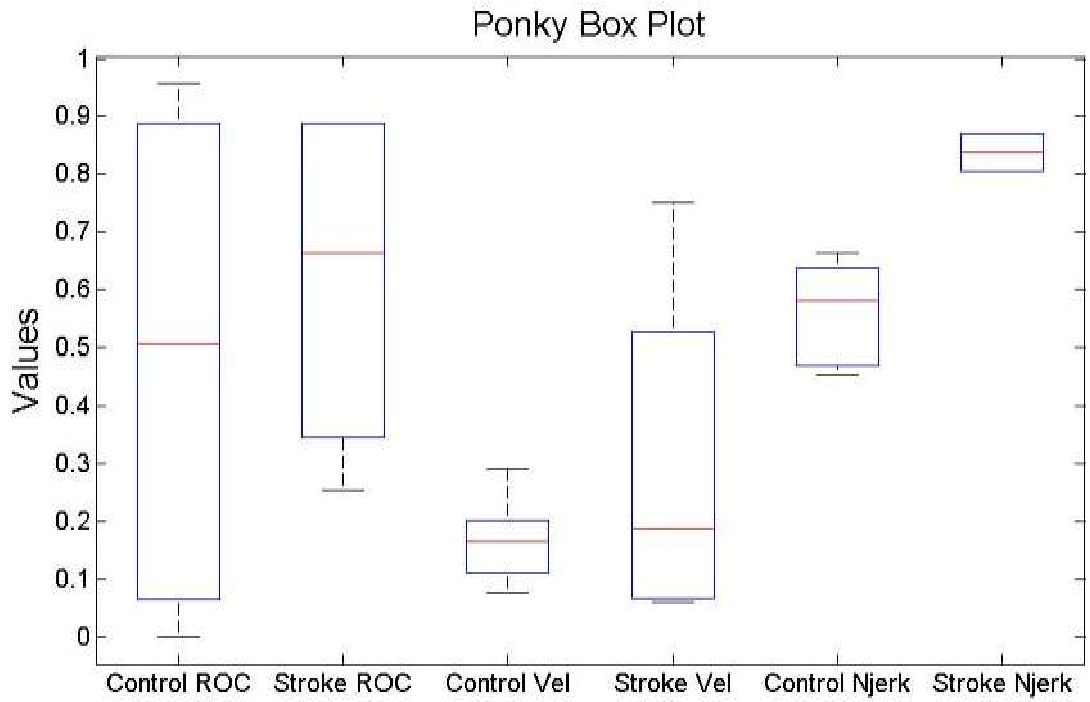

Figure 29 Box Plot quantifying the performance of the two groups while playing Ponky(car racing).

The stroke subject's range of cursor fell within the same range as the values of the control group. Since stroke patients typically have a smaller range of motions than those who are unimpaired this may seem counterintuitive. However, when the sensors are first applied, the gain of the device is adjusted to ensure each player can move the cursor from the left side of the screen the right side of the screen. Thus this adjustment ensures that most players will have a similar range of cursor.

Interestingly the stroke subject had a larger range of velocity values than the control group. This result is a red flag, because typically the variance of an individual is less than the individual group. This plot shows that the stroke subject made both very slow and very fast movements while playing the game. Contrarily, the range of velocities for the control subjects was much smaller indicating that they tended to move at a more constant velocity.

The stroke subject had a higher normalized jerk than the control subjects as expected. It is believed that the large variation in velocity while playing the game contributed to the higher normalized jerk score.

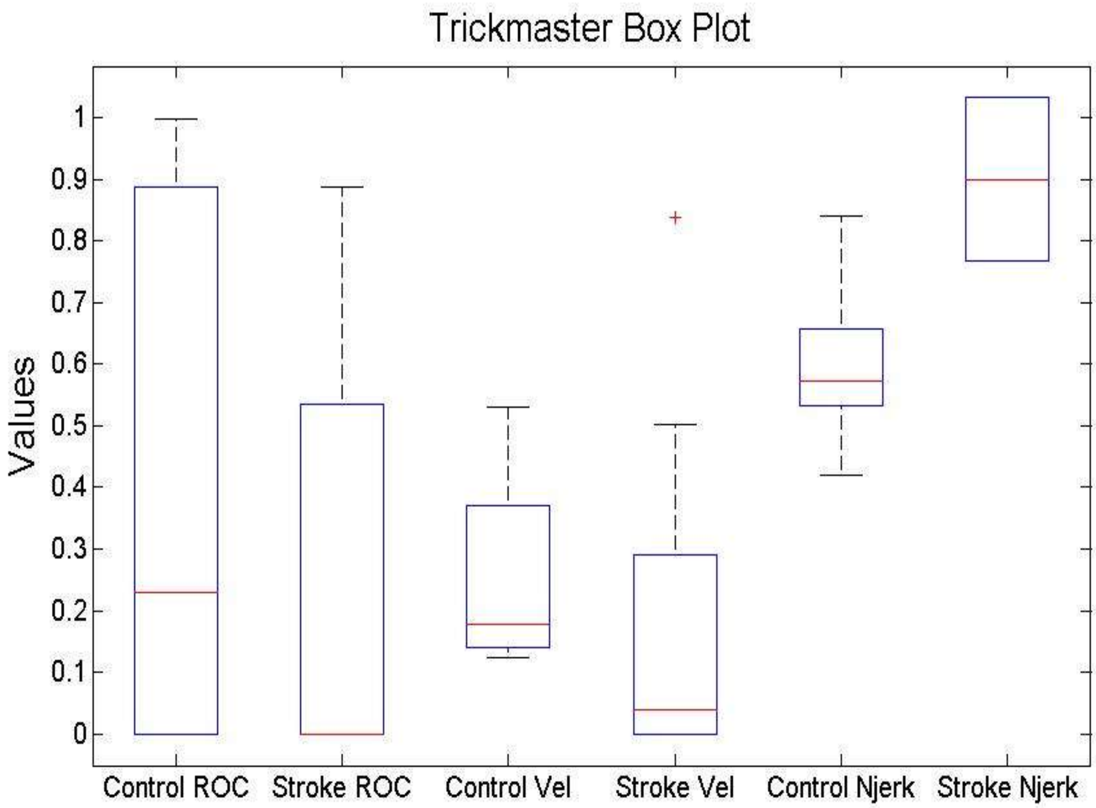

Figure 30 Box Plot quantifying the performance of the two groups while playing Trickmaster (skateboarding)

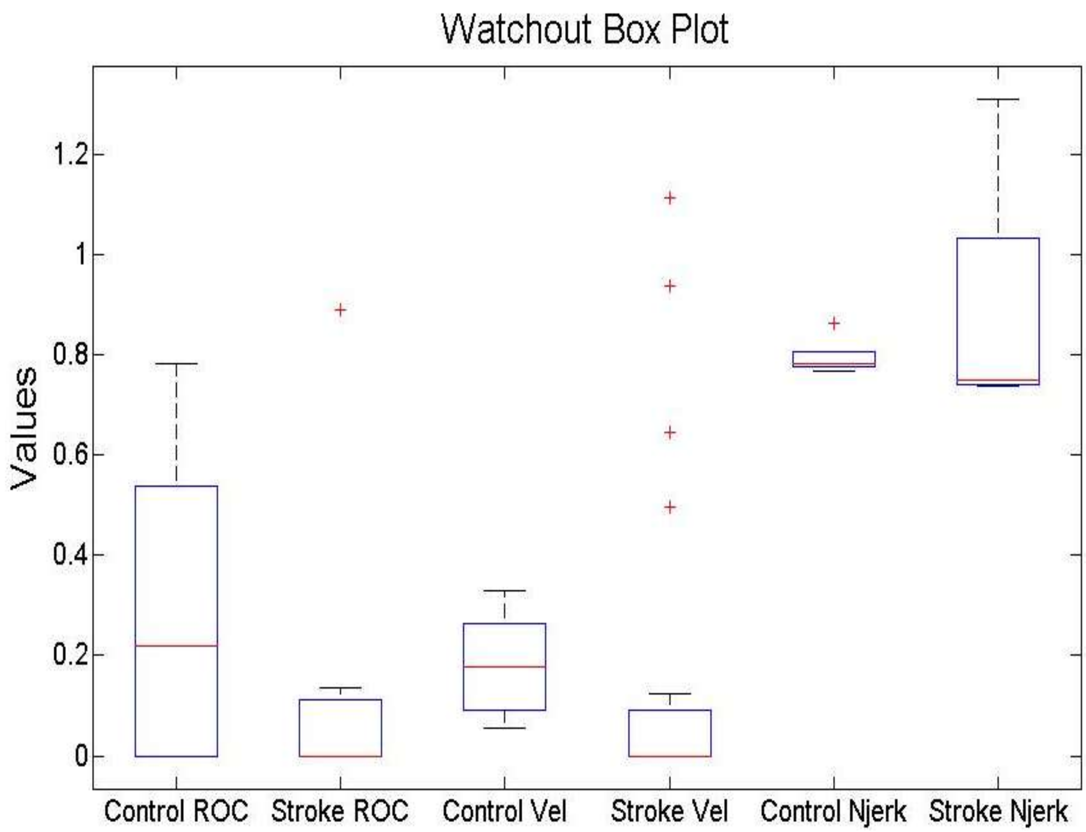

Figure 31 Box Plot quantifying the performance of the two groups while playing Watchout (Dodging boulders)

The stroke and control values for each of the three measures while playing trick master showed a trend similar to the values received while playing Ponky. The ROC values for the stroke subject were within the same range as the ROC values for the control subjects. Although the difference in range size for the velocity measures are not a large as they were for the previous game for the two groups , the stroke subject did have a higher range of velocity values. This decrease in range may be attributed to the removal of an outlier point. The mean normalized jerk for the stroke subject was higher than that of the control subjects. However, the box and whiskers do overlap indicating that this difference is not significant. All of values for the stroke subject were within the same range of the values for the control subject for all measures while playing Watchout.

3.3.4.2 Single Cuff Testing

Four Pediatric therapists and 11 children were recruited to test the device. Each therapist determined the target movements and secured the cuff over the appropriate muscle. The children that tested the device were diagnosed with traumatic brain injury, cerebral palsy, or Guillian-Barre. Each child played the boat game first to become familiar with the operation of the device. Children were then permitted to select from the four games that received the highest ratings from the lower limb testing (ponky, ice racer, trick master , and watch out). Each child was permitted to select any game and play as many times as they wished. Therapist and children were then asked to complete a questionnaire that asked about the ease of use, the repeatability/reliability of the sensors, the ability of the system to hold the patients attention, and whether or not the games were too fatiguing.

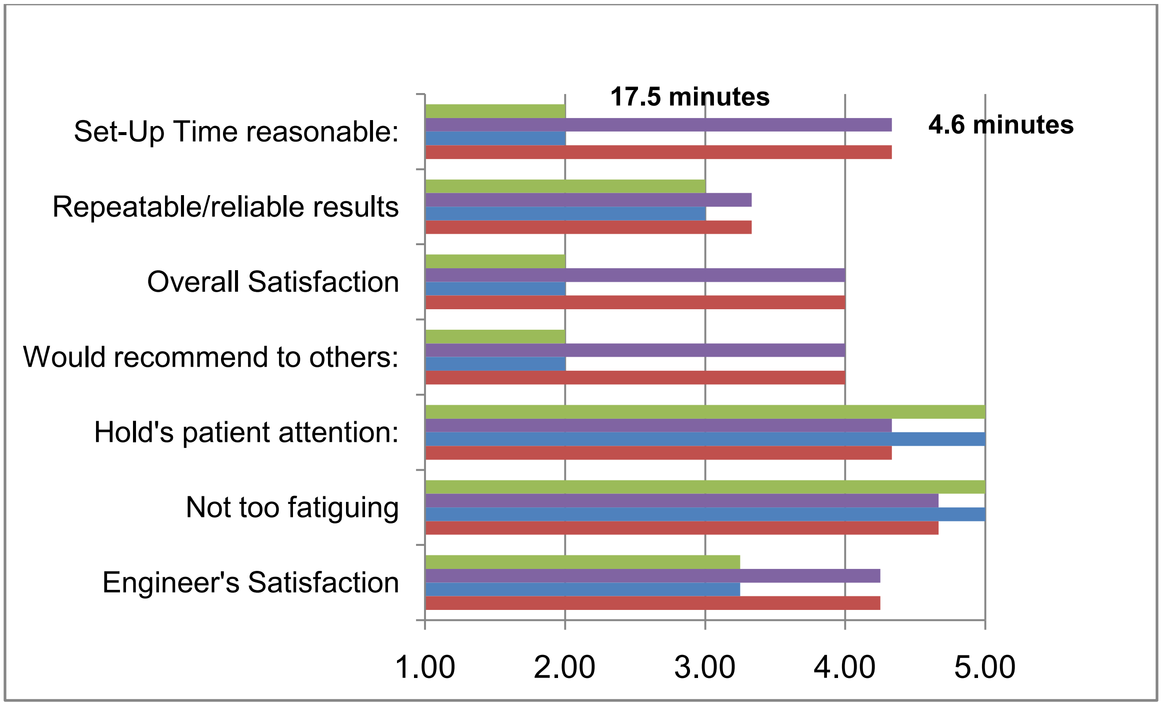

Figure 32 Therapist survey results. After using the device therapist ranked seven categories on a scale of one to five, with five being the most desirable.

The exercises used by the therapists included bicep curls, shoulder extension, and wrist flexion. The therapists were separated into two categories: Those who tested the device on a child with hypotonic muscle, and those who tested the device on a child with non-hypotonic muscle. The therapists that tested the device on children with hypotonic device tended to have lower overall satisfaction with the device than the other therapists (2.0/5 vs. 3.0/5). The time it took to get the device to detect the muscle activations of the children with hypotonic muscle took significantly longer (17.5 minutes vs. 4.6 minutes) than it did for the other children. Both groups rated the system’s ability to hold patients attention and the amount of fatigue caused by the device with high scores.

CHAPTER 4

CROSS-HEMISPHERIC LEARNING

4.1 Methods

4.1.1 Subjects

Twenty-two neurologically intact right-footed adults (mean age 27.1, range 21-36 years, ten males and twelve females) participated in this study; twenty of the subjects completed the entire protocol, and an additional two subjects only performed the neutral protocol. Subjects were randomly divided into either group RL or LR. Group demographics did not differ with respect to age, weight or height. Written, informed consent was obtained from each subject prior to participation, and was approved by the Rutgers University Internal Review Board. The inclusion criterion consisted of being right foot dominant. All subjects had normal ranges of ankle motion and experienced no pain or stiffness.

Footedness was assessed using the Modified Waterloo Footedness questionnaire that addressed the preference for a foot manipulating an object (kicking a ball, smoothing sand, etc.), and the preference for the foot in providing support during a task (hopping on one foot, etc.) [54]. For each question a score of -2 is given for answers of left always, and +2 for answers of right always. Therefore, an overall score of -20 is the maximum left dominant score, and +20 is the maximum right dominant score. For the balance and manipulation categories a score of -10 is the maximum left dominant score, and a score of +10 is the maximum right dominant score. A score of 0 indicates ambidexterity.

A written version of the Waterloo test was followed by an active (physical) version in which subjects were asked to demonstrate use of their feet, according to the written questions. To ensure that only right-dominant subjects were used, only those with foot dominance scores greater than 5 were included.

4.1.2 Experimental Set-up

Subjects sat in a height-adjustable chair with feet hip-width apart, and shank oriented vertically, as depicted in Figure 33. The foot was secured to a custom isokinetic ankle platform, and chair height was adjusted to bring the foot in contact with the platform with minimal vertical load [55]. The ankle platform was mounted on a universal joint fitted with potentiometers for goniometric registration of dorsiflexion and plantar flexion on the y-axis and inversion-eversion motions on the x-axis. The goniometer output was digitized and sent to a computer to control cursor movement. Subjects thus could move the cursor up and down with dorsiflexion and plantar flexion respectively, and left and right by everting/inverting the right foot or oppositely for the left foot. Subjects viewed the cursor and targets on a 16” screen with 800 X 600 pixels resolution updated at 60 Hz. Subjects fixated on the screen, which was head-height, and did not directly observe their feet during testing. Goniometric accuracy was 0.64 ± 0.17 º and data were sampled at 13 Hz.

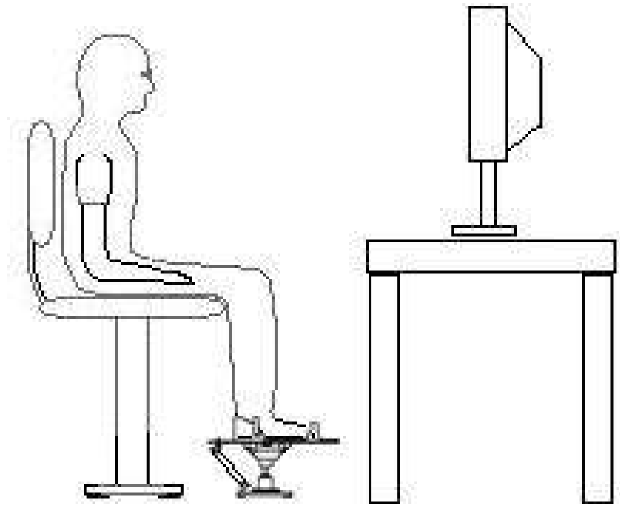

Figure 33 Experimental set-ups for Interlimb transfer study.

4.1.3 Experimental task

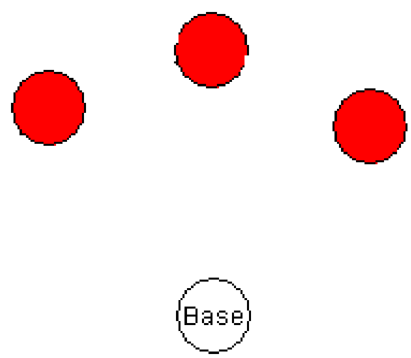

Figure 34 Motor Learning Test Screen. The red circles represent the targets. The base is the white circle located in the center of the screen.

Each subject performed two tasks adapted from Sainburg and Wang (2002), as shown in Table 2. Group RL subjects performed the tasks first with the right foot, and then with the left foot; group LR subjects did the opposite. During task (1), the cursor moved in proportion to the biaxial ankle rotations of the user. Throughout task (2), a VM rotation was imposed, wherein the previously learned ankle motions now moved the cursor 30° counter clockwise (ccw) relative to the starting circle, causing the subject to learn a new motion, that needed to be rotated 30° clockwise (cw) to hit each target. Each task consisted of 24 trials to each of three (random) targets, for a total of 72 trials per task for each foot; each subject performed 288 trials in total. In addition, five subjects were presented with a catch-trial, in which the VM rotation was removed.

Knowledge of results was given to the subjects by a score and audio feedback, which were both based on the final position of the cursor. Errors less than 42 pixels (0.84 º) received 10 points, errors within 42 and 84 pixels (0.84 º and 1.68 º) received 3 points, and errors within 84 and 126 pixels (1.68 º and 2.52 º) received 1 point. The subjects performed the task with each foot first without a VM rotation and then during a VM rotation in which the cursor was rotated 30° counterclockwise relative to the start circle. During VM rotation, target 3 was along the x-axis, and was reached by almost pure inversion by the right foot and eversion by the left; target 1 was centered after rotation.

	Baseline (0°)		**Experimental (30°)**	
Group A	Right Leg	Left Leg	Right Leg	Left Leg
Group B	Left Leg	Right Leg	Left Leg	Right Leg

Table 3 Protocol for the first experimental task.

4.1.4 Evaluation

A custom MATLAB program quantified the movement data using three measures of performance: final position (FP) error, initial direction (ID) error, and final direction (FD) error. FP error was measured by the Euclidean distance between the center of the target and the foot-path position at the end of movement and reported as the percentage ratio of final position error to the distance from the center of the base to the center of each target (250 pixels). ID error, was calculated as the difference between the vector defined by the foot-path position at the start of movement to the center of the target and the vector defined by the start of movement to the point of maximum velocity. FD error was determined by calculating the difference between the vector defined by the foot-path position at the start of movement to the center of the target and the vector defined by the start of movement to the foot-path position at the end of movement.

To standardize subjects according to basic skill level, the z-score, as shown in equation 1, was computed for each of the three measures. Since the goal was to normalize each subject, and not each trial, scores from all 288 trials were pooled to compute one mean and standard deviation value for each subject. Since the distribution of the subjects' data was logarithmic, the log-normal mean, μ, and standard deviation, σ, were computed as shown in equations 2 and 3 respectively.

$$z_i = \frac{x_i - \mu_{log}}{\sigma_{log}}$$ ***Equation 1***

$$\mu_{log} = e^{\mu + \frac{\sigma^2}{2}}$$ ***Equation 2***

$$\sigma_{log} = \sqrt{(e^{\sigma^2} - 1)e^{-2\mu + \sigma^2}}$$ ***Equation 3***

To test for ILT, the transfer (second) foot of one group was compared with the naïve (first) foot of the other group in both tasks. The statistical approach was identical to previous studies [3] and consisted of comparing the performance of the two groups during the first epochs (average of four trials) for each foot using a post-hoc group comparison. The percent of measurement error decrease was computed for both ankles as the ratio of the difference between the group average z-scores to the average of the naïve foot (Equation 4).

$$ILT = \frac{\mu_{naive} - \mu_{OFT}}{\mu_{naive}} * 100\% \qquad \textit{Equation 4}$$

Where $\mu_{naive} u_{naive}$ is the average z-score of motor skill for the first epoch of the first group that performed the task with the ankle of interest, and μ_{OFT} is the average z-score of the first epoch of the group that performed the task with the ankle of interest following opposite foot training [56].

A repeated measure ANOVA was performed using SPSS 15.0 (SPSS Inc, Chicago IL) with group (RL or LR) as the between subjects variable, and target and foot (R or L) as the within-subject factors. Post hoc pair-wise comparisons using a Bonferroni correction were implemented to test the differences among targets, and to determine if there were differences between the naïve foot and that receiving OFT.

To assess subjects' adaptation to the VM rotation, a catch trial was done with five subjects (3 RL and 2 LR) who performed four repetitions in the neutral condition after having completed the VM task with the final foot. Task adaptation was defined as the difference between the initial direction errors of the first post-training trial and those of the final neutral trial.

4.2 Results

4.2.1 Subjects

Subject demographics and selected results from the footedness survey are presented in Table 2. The mean written footedness assessment scores for group RL and LR were 12.5 and 12.0, respectively, and the mean physical footedness scores were 10.9 and 10.1, respectively. These scores are all well above the required score of 5, indicative of right dominance. All subjects completed the neutral task and ten subjects in each group completed both the neutral and rotated tasks.

	Group RL (n=11)	Group LR (n=11)
Age	26.2 ± 5.0	25.7 ±4.3
Weight (lbs)	146.9 ± 35.4	146.1 ±27.7
Height (in)	66.2 ±3.4	66.2 ± 2.7
Physical Footedness Assessment :	10.9± 4.6	10.1 ± 3.2
Written Footedness Questionnaire :	12.5 ± 3.4	12.0 ±3.6

Table 4 Demographics. A written footedness questionnaire was the primary determinant of footedness and physical footedness assessment was used to confirm the written results.

4.2.2 Task Sequence and Overall Performance:

To illustrate the protocol, cursor trajectories for a representative subject from each group during all tasks are shown in Figure 35. The RL subject is outlined in red (solid) and the LR subject is outlined in blue (dashed); trajectories to their respective targets (1, 2, and 3 from left to right) are coded as shades from light to dark. Tracking the RL subject, beginning with the first column, first row, it can be seen that trajectories during the final trial of task 1 were generally aimed near the targets. Note that the initial direction toward target 1 was off course, but corrected quickly.

Accuracy during the first trial after VM rotation was markedly reduced (second column, first row) with increased errors in initial and final direction, as well as final position. Initial performance of his left foot following OFT is shown in the third column, second row, and left (transfer) foot performance after several trials is shown in the fourth column, second row. Targeting accuracy improved by the last VM trial.

To determine if adaptation to the VM counter-clockwise rotation occurred, 5 subjects performed the neutral task immediately following the last VM rotation trial (post-exposure). Most of the trajectories (fifth column, second row) were rotated clockwise to the targets.

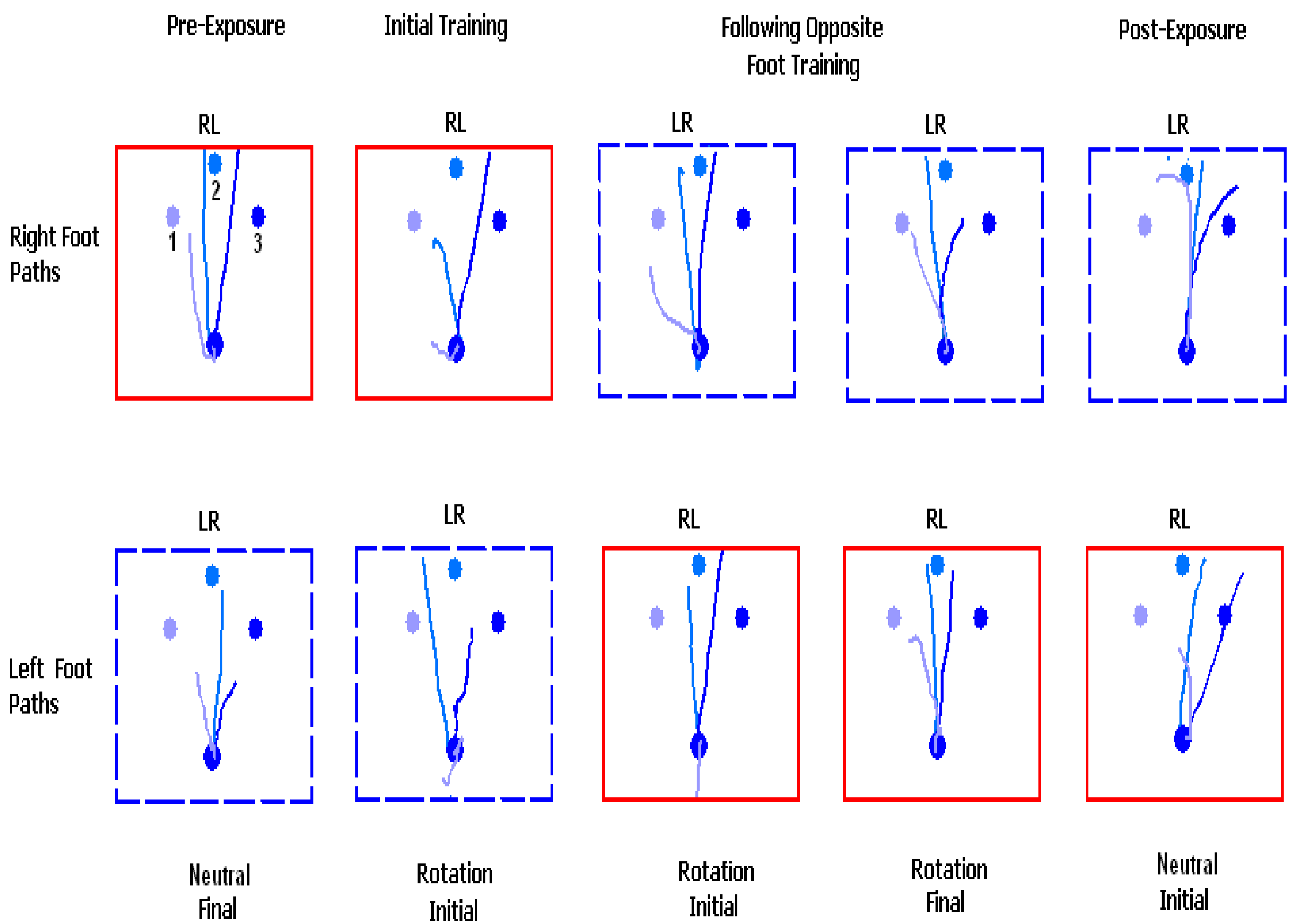
Pre-Exposure
Initial Training
Following Opposite Foot Training
Post-Exposure
Right Foot Paths
Left Foot Paths
RL
RL
LR
LR
LR
LR
LR
RL
RL
RL
1
2
3
Neutral Final
Rotation Initial
Rotation Initial
Rotation Final
Neutral Initial

Figure 35 Sample Movement Trajectories of representative subjects. Trajectories from a subject in Group RL are in solid boxes, and trajectories from a subject in Group LR are in dashed boxes. Trajectories are shaded in accord with the target. The first column shows the last cycle of movements of each foot during the baseline condition. The second column shows the first cycle of movements during the initial training session. The third and fourth columns show the first and last cycle of movements following OFT.

Average performance for both tasks from both groups is summarized in Figure 36; more detailed views and comparisons are presented in later figures. Each data point represents the average and standard error of the z-normalized target errors from twelve trials. Each row of panels shows curves made by the first and second feet in the baseline and VM rotated conditions for both groups. The three rows show performance in terms of FP, ID and FD. One general observation is that the beginning trials of both groups in the neutral condition exhibited the worst performance (highest z-scores) in all three categories, as expected. Also evident from Figure 4 is that all performance curves trended downwardly, meaning that errors decreased from the first to last of the 6 epochs in all tasks.

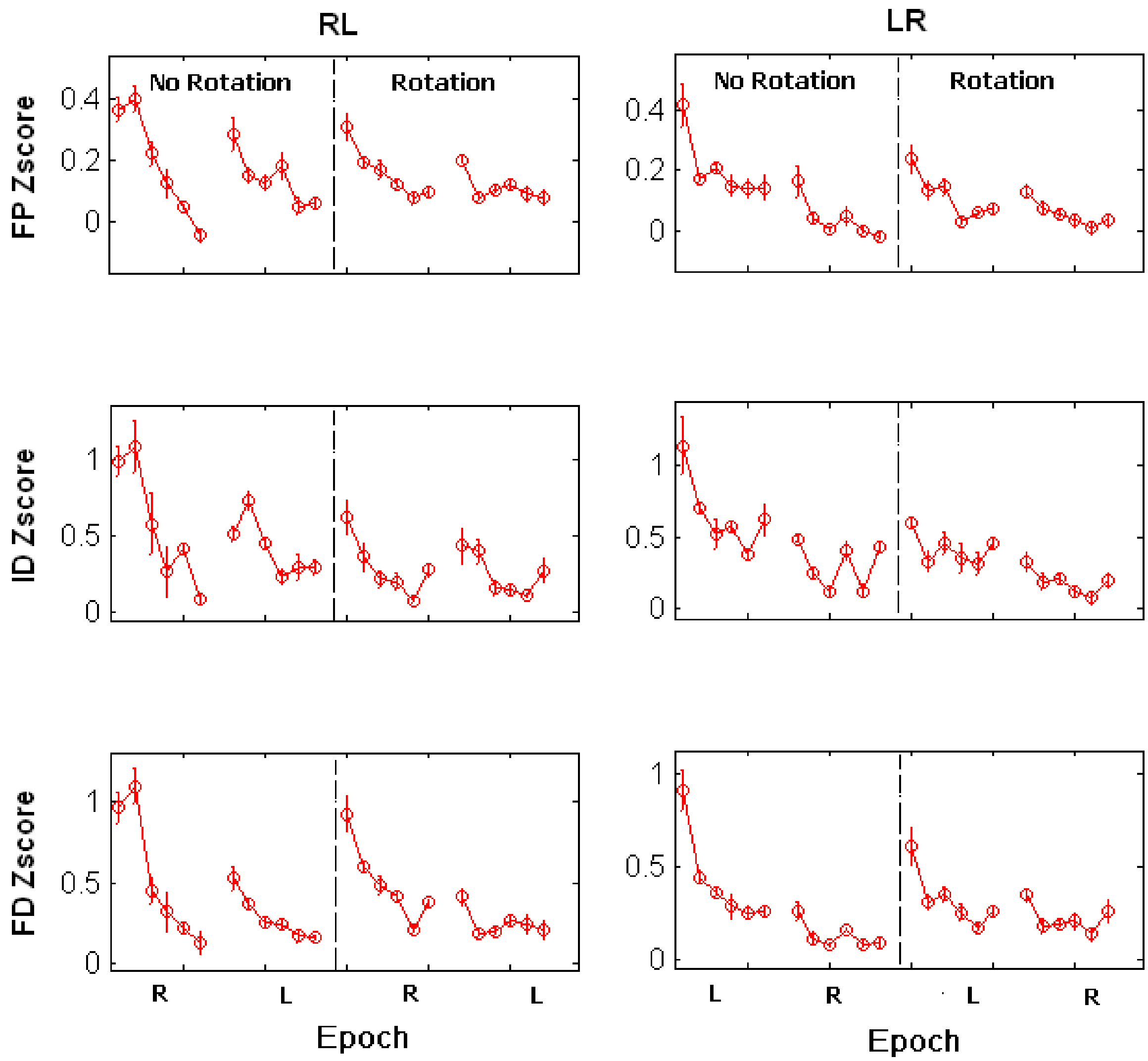

Figure 36 Performance Curves for both groups in chronological order. Each data point represents the average of 12 consecutive trials (mean ± SE). The first two curves in each panel are from the first and second foot that performed the task in the neutral condition.

4.2.3 ILT during Task 1: Neutral

Since the task space of the ankle was in a different plane from the target effector space, and ankle movements were not mapped to that of the cursor in absolute coordinates, subjects needed to make VM transformations for the task, which constituted a novel learning challenge.

Therefore, performance curves of the baseline (neutral) condition were compared to determine possible occurrence of ILT, as shown in Figure 37. ANOVA of all three measures showed a significant interaction between foot and group ($p < .05$), but not between foot and target ($p \gg .05$). Comparing the initial left foot performance of both groups (right panels of Figure 37), it can be seen that the right foot z-scores of group LR are lower than those of group RL for both FP ($0.16 \pm .06$ versus $0.36 \pm .06$, $p<.05$), and FD ($.26 \pm .21$ and $.96 \pm .21$ $p<.05$). Right foot ID scores for group LR were lower than those of group RL but the two scores did not significantly differ ($0.49 \pm .37$ versus $1.0 \pm .37$, $p=.34$). Left foot z-scores showed slight trends for FP ($0.29 \pm .08$ versus $0.42 \pm .08$ $p=.23$), ID ($0.51 \pm .22$ versus $1.1 \pm .22$; $p=.053$) and FD ($.53 \pm .20$ and $.91 \pm .20$ $p=.18$), but the differences between the two groups were not significant. ILT measures are annotated in the upper right of each panel; note that the right ankle improvements in both FP and FD are significant ($p<.05$).

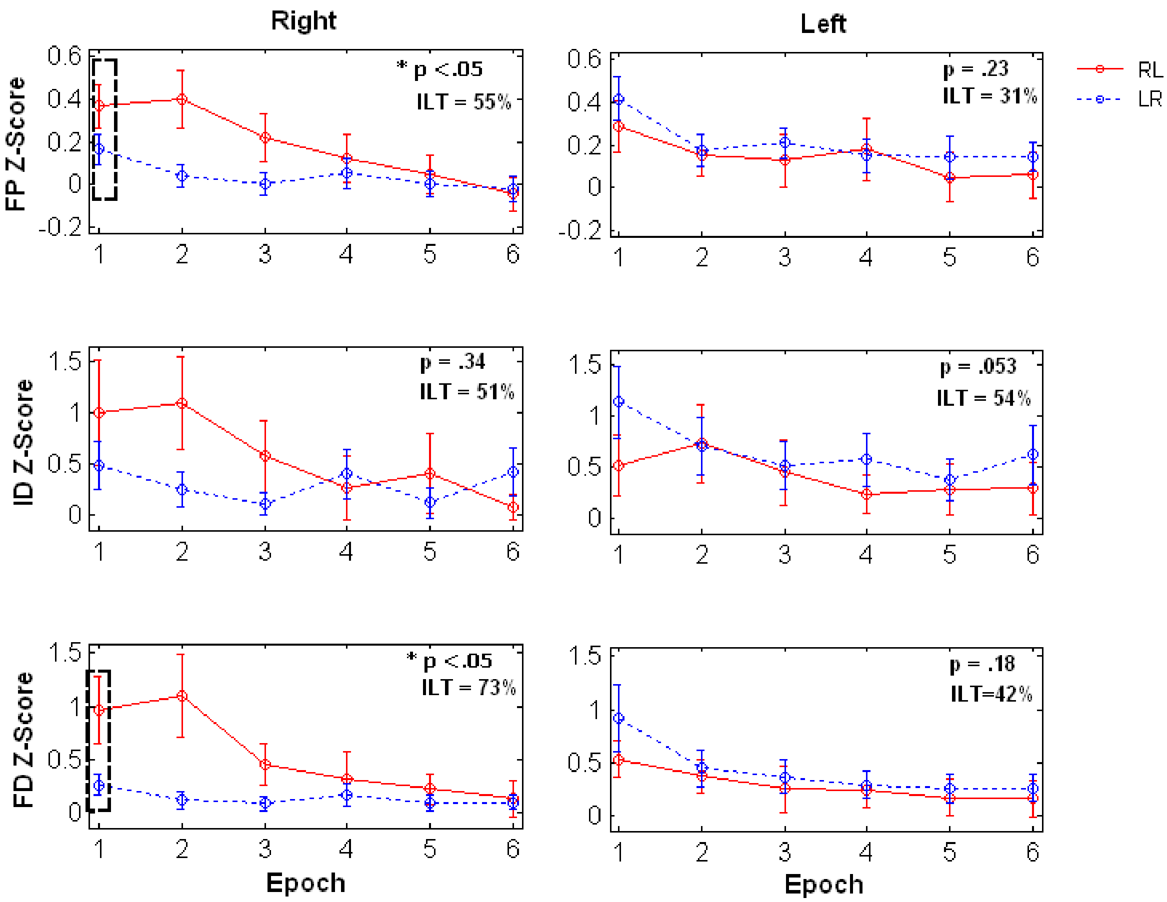

Figure 37 Performance curves showing the mean z-scores from the neutral trials for the right and left feet. Each data point represents the average of 12 consecutive trials (mean ± SE). The performances for Group RL(solid red) and Group LR (dashed blue) are shown separately for the right and left feet. For the right foot, Group RL is naïve to the task; for the left foot, Group LR is naive to the task.

4.2.4 ILT During Task 2: VM Rotation

ANOVA showed a significant interaction between foot and group for the FD measure ($p<.05$), but not for FP ($p=.10$) or ID ($p=.13$). Between-group comparisons were made to study the foot group interaction in more detail (Figure 38). Group LR experienced significant ILT in terms of both FP and FD, but the RL group did not. Average z-score for FP of the right foot of group LR was significantly lower than that of group RL (.14 ± .06 versus .37 ± .06; $p <0.05$). The left foot z-score of group LR, in contrast, did not significantly differ from those of group RL (.27 ± .06 versus .22 ± .06; p =.61). Likewise, FD scores were also lower for the right foot of group LR (.33 ± .15 versus .88 ± .15; p <.05). The difference between the groups is evident also by comparing the performance curves of the right and left feet. In the case of the right foot, the performance curves of the RL group are consistently higher than those for the LR group for all epochs in all measures, unlike the left foot, in which the group curves cross.

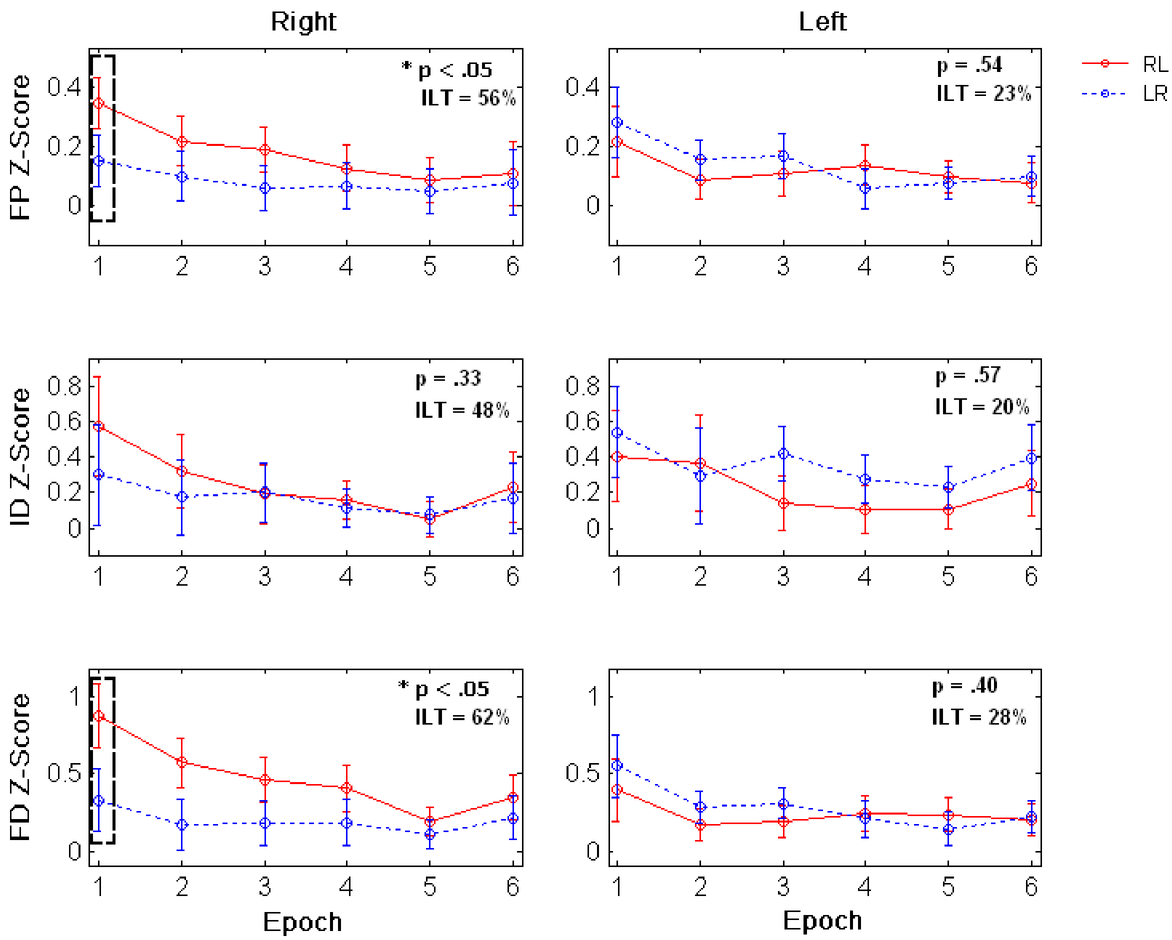

Figure 38 Performance curves showing the mean z-scores from the VM rotation trials for the right and left feet. Each data point represents the average of 12 consecutive trials (mean ± SE). The performances for Group RL (solid red) and Group LR (dashed blue) are shown separately for the right and left feet. For the right foot, Group RL is naïve to the task; for the left foot, Group LR is naive to the task.

Since the off-center targets seemed to be more difficult for the subjects, pair-wise comparisons were done to detect the presence of systematic effects. ANOVA did not show a significant interaction between foot and target for any of the scores ($p>>.05$ for FP and ID, $p=.051$ for FD). The mean scores for all epochs for the three targets are shown in Figure 39 a-c. As seen in Figure 39 a & b, target 2 (center) produced the lowest errors and target 3 produced the highest for both feet in terms of both FP and FD. FP error differed significantly between targets 2 and 3 ($.01 \pm .07$ versus $.50 \pm .10$; $p<.01$) for the left but not the right foot. FD errors (Figure 39c) were also significantly higher at target 3 for the left foot compared to the center target ($.89 \pm .19$ versus $.04 \pm .08$; $p <.01$). Target 3 required the left foot to dorsiflex and evert and the right foot to dorsiflex and invert and thus although left foot eversion was less accurate than right foot inversion, the difficulty with target 3 does not relate to the type of movement, but its position. Note that targets 1 and 3 did not differ significantly for either foot, and hence there is no evidence for movement preference along the x-axis.

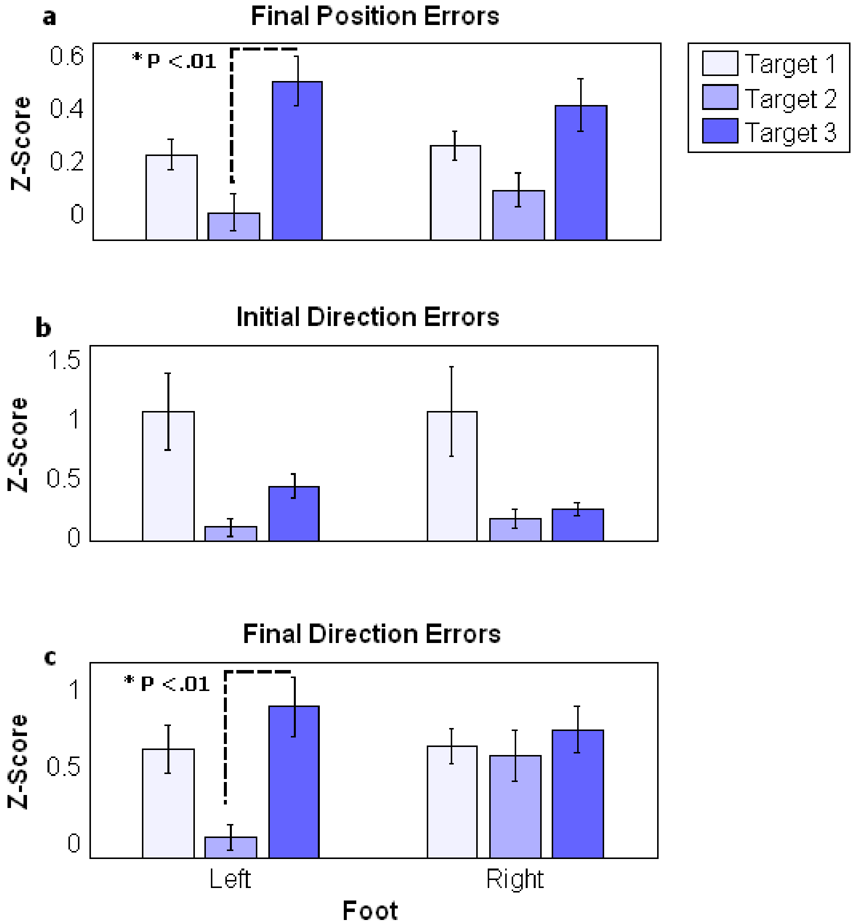

Figure 39 Target Performance Comparison. The mean scores for all visuomotor rotation trials for each foot are shown above. Error bars represent ± SE. a) FP errors; b) FD errors; c) ID errors

4.2.5 Catch Trials

Five subjects were exposed to a catch trial to determine the after-effects of the visuomotor rotation. After the final VM rotation trial, the 30 ° ccw rotation was removed, and four successive trials were run. Changes in ID errors were computed for all three targets for both, with negative differences representing a cw rotation. Mean ID differences were similar for targets 1 and 2 (-27.8 ± 19.9 ° and -27.9 ± 17.7 ° respectively). The mean ID difference was lower for target 3 (-

17.7 ± 23.3°), but there were no significant differences among the three targets. The close approximation to 30° cw errors in the after-effect test indicates that adaptation occurred during VM rotation.

4.3 DISCUSSION

4.3.1 Validity of Study

Herein we report the first systematic study of ILT of learned movements by the ankle. We studied twenty-two healthy subjects, separated into two groups. Using a target reaching VM rotation task similar to that developed for the upper-limb, we demonstrated ILT occurring from the non-dominant (left) to the dominant [57] foot, but not vice-versa. As seen in the between-group comparisons in Figure 38, the learning transfer included information about final movement direction and position ($p < 0.05$). Significant improvements in performance ranging from 55% to 74% were seen following OFT. A test of adaptation in a subset of subjects using a catch trial after removing the 30-degree ccw VM rotation, revealed a cw after-effect in all three targets, with two of the targets having after-effects close to 30°, indicating nearly complete adaptation. Furthermore, we showed ILT occurring for a novel task involving coordinate transformation between foot and cursor motion. In this test, the learning transfer pattern was identical to the VM task, with right foot final movement direction and position errors decreasing after OFT. These results, with preferential L to R transfer in both tasks, are consistent with previous studies in which exercise training of the left leg improved the performance of the right leg, but not vice versa [30].

Thus, ILT can be generalized to the ankle and extend those of van Hedel et al. (2002), who showed that movements learned at the knee to avoid an obstacle on a treadmill are transferred from one leg to the other, non-preferentially. In that study, the ankles did not experience significant ILT and their negative finding was attributed to the greater variability and complexity of the ankles as compared with knees. Herein, by studying isolated ankle movements in a single-joint targeting paradigm, with between-group comparisons, we found positive evidence for ILT of ankle motor control.

Our analysis differs from that of Sainburg et al. (2002), mainly in terms of baseline testing. In their studies, the task and effector space was common, since the cursor was projected onto the hand. Our subjects' were required to make coordinate transformations in all tests since their feet operated near floor level, while the cursor moved at eye-level while seated. Furthermore, cursor movement was in the frontal plane, while foot motion was 3-dimensional. Learning this transformation was an integral part of the experiment, and indeed, we found evidence for significant ILT in the baseline (neutral) trials. It was therefore not possible to use baseline performance as a normalization factor as done by Sainburg et al. (2002). Instead, we computed z-normalization scores for each subject, in order to minimize skill-differences.

The possible influence of musculoskeletal differences between ankles on the ILT results during VM rotation was investigated by analyzing performance difference among targets (Figure 39). For the left foot, target 3 was significantly more difficult, in terms of both FP and FD errors, based on a pair-wise comparison with target 2. Since target 3 required eversion by the left foot, a motion which was amplified by VM rotation (see Figure 35), it could be argued that left foot performance suffered relative to that of the right because eversion was more difficult than inversion, which is the motion the right foot needed to reach target 3. This argument, however, is not strongly supported by the evidence, since, (1) the right foot tallied more errors toward target 3 (inversion), than toward target 1 (although not significantly, Figure 39); (2) there was no significant difference between targets 1 and 3 for either foot; (3) ANOVA showed no interaction between foot and target. Thus, the target difference may not have significantly influenced the ILT asymmetry of our results in terms of VM rotation, and the consistency of the latter with those of the non-rotated test argue against a major influence.

4.3.2 Comparison with Upper-limb ILT

Direct comparison of ILT characteristics of the upper and lower-limbs is not straightforward due to fundamental differences in their function. Upper-limbs specialize in fine manipulation, with infinite degrees of freedom, usually cooperating in the same workspace whereas the lower limbs specialize in relatively gross rhythmic alternations, usually operating in separate workspaces, where stability and balance are the primary objectives. For manipulation tasks, the foot task space, i.e., pedals and levers, is usually removed from the effector space, i.e., vehicles or equipment. Accordingly, motor strategies for upper and lower limbs differ widely, as documented by their relative cortical activities. Upper-limb activity involves primarily the contralateral sensorimotor cortex, and the ipsilateral cerebellum, whereas ankle movements exhibit more ipsilateral activity in the motor and pre-motor cortices and are continually under reciprocal control. [58]. Thus, the upper limbs operate with more independence, and less cross-hemispheric interaction than the lower limbs.

It has been suggested that ILT expresses as asymmetric when there is a large degree of hemispherical interactions [44]. In arm experiments, when workspace is common, ILT is asymmetric; when arms operate in separate workspaces, i.e. both task and effector space are separate, ILT is symmetric [44]. Although our subjects' feet manipulated a common effector (cursor on screen), they were technically in separate workspaces, a situation that is somewhat in between the two protocols used by Sainburg et al. Nevertheless, our finding of asymmetric ILT is consistent with previous interpretation, since lower-limbs have inherently greater cross-hemispheric activity.

Lateralization of the arms, i.e. handedness, manifests as preferential use of the left hand (for right-handers) for holding objects in position while the right arm manipulates a tool, such as in the case of handwriting [3]. This behavioral specialization of the arms, when working in a common space, provides an explanatory context for the asymmetric transfer of information between them. Asymmetry is also reflected in the cortex, wherein left hand motions are much more bilaterally represented in the sensorimotor cortex compared with the right [59].

Lateralization of the feet, i.e. footedness, may be less prominent than handedness, and may vary depending on the context. For example, during unilateral balance, the right foot is the favored postural stabilizer, with the ankle being the most important joint [60]. Thus during gait initiation, the right foot is likely to be the primary stabilizer. Our results on ILT of both position and direction going to the right foot are consistent with the fact that the feet are not involved in common manipulation, but rather each ankle, during the single stance phase, is alternately required to control both position and trajectory. The primary lead ankle, i.e. the right, may be endowed with a more adaptive control system that would be more responsive to ILT. This concept is concordant with upper-limb ILT, where each limb receives benefit only according to its specialty, i.e. the right arm learns trajectory information and the left, position. It should be noted that our experimental condition of sitting, with one ankle moving while the leg was partially weight-supported, simulates a common position for ankle exercise, and does not closely simulate either standing or gait.

Whether the present results represent cross-hemispheric transfer (CHT) of learning, or transfer at a lower level, such as spinal, cannot be concluded. Recent studies have provided strong evidence for CHT by showing that subjects who had left hemispheric damage due to stroke exhibited deficits in arm trajectory, whereas those with right hemispheric damage had deficits in final position accuracy [61]. These specific hemispheric lateralizations correlate well with behavioral specializations of the arms noted above. Further support for CHT was demonstrated in a study showing that muscular strength gained by upper or lower-limb transfers to the opposing limb [33]. Meta-analysis of randomized, controlled studies of limb training transfer revealed that a strength increase of 35% in a trained lower-limb was accompanied by a 7.8% strength increase in the contralateral limb even though it experienced no substantial muscle activity during training, and did not increase cross-sectional muscle area [33]. These 'cross-educational' strength gains were limited to the homologous muscle of the opposite untrained limb, and to the same movement task performed by the trained limb. The specificity of this phenomenon and the lack of detectible morphological changes in muscle suggest that transfer is due to alterations in neural control at a central, possibly hemispheric level [31].

CHAPTER 5

DISCUSSION

These preliminary studies demonstrate that it is possible to use FMG signals to control a mouse cursor. Using FMG signals to control a mouse cursor made it easy to find and include games that satisfied people of different ages. Ankle flexions proved to be very difficult for the stoke subject. Flexing the ankle against gravity is a difficult task for many stroke patients. All stroke subjects were able to flex while using the platform; therefore a mechanical assist may be necessary when working with ankle flexions. However, it may be possible to use wrist or elbow flexions without an assist. When given the option to use both legs, it is likely that the patients will rely more on the unaffected leg for cursor control, as seen with one subject. Therefore, we have designed a system that can control both directions with one limb. We believe that making a system in which relaxing a muscle, and activating a muscle controls the two directions of the mouse cursor may be more effective.

Both the platform and FMG sensor gave similar information. However, during testing it was discovered that the platform is easier to use for ankle testing than the FMG cuff. This is due primarily to the weight of the lower limbs, which is much greater than that of the upper limbs. Although EMG has been proven to be very reliable in applications involving a switch control it is not well suited for systems which require proportional control. Though it is possible to adapt the software to use a switch control, this method would hinder the recording of the actual movements and make it difficult if not impossible to quantify velocity, acceleration, and movement smoothness.

Understanding the hemispheres that control the different aspects of motion is important when evaluating the efficacy of a rehabilitation regimen for those with brain damage. The inter-limb transfer study revealed that movements learned in the left ankle are able to transfer to the right ankle. The results have important implications for functional rehabilitation of clients with hemiparesis due to stroke, CP or other central injury. Most hemiparetics cannot fruitfully exercise their affected leg due to severe control deficiency, and therefore do not generally participate in directed physical therapies. Therapeutic options are further limited since the affected ankle is generally immobilized in an orthosis in order to restore a semblance of gait. If the affected limb could be improved by sustained exercises of the contralateral limb, this could ameliorate the complications caused by disuse and maximize the effectiveness of rehabilitation. In particular, restoring even a limited degree of ankle control could restore un-assisted gait and/or postural balance to many clients. The present results thus provide suggestive evidence for the potential benefit to the affected limb afforded by contralateral limb training, and studies are underway to test its efficacy [55].

APPENDIX A

WATERLOO FOOTEDNESS QUESTIONNAIRE [54]

Instructions: Answer each of the following questions as best you can. If you *always* use one foot to perform the described activity, circle RA or LA (for **right always or left always**). If you **usually** use one foot circle RU or LU, as appropriate. If you use **both** feet **equally often, circle EQ.** Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

1. Which foot would you use to kick a stationary ball at a target straight in front of you?	**LA LU EQ RU RA**
2. If you had to stand on one foot, which foot would it be?	**LA LU EQ RU RA**
3. Which foot would you use to smooth sand at the beach?	**LA LU EQ RU RA**
4. If you had to step up onto a chair, which foot would you place on the chair first?	**LA LU EQ RU RA**
5. Which foot would you use to stomp on a fast-moving bug?	**LA LU EQ RU RA**
6. If you were to balance on one foot on a railway track, which foot would you use?	**LA LU EQ RU RA**
7. If you wanted to pick up a marble with your toes, which foot would you use?	**LA LU EQ RU RA**
8. If you had to hop on one foot, which foot would you use?	**LA LU EQ RU RA**
9. Which foot would you use to help push a shovel into the ground?	**LA LU EQ RU RA**
10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first?	**LA LU EQ RU RA**

11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities? (choose one)	**Yes No**
12. Have you ever been given special training or encouragement to use a particular foot for certain activities? (choose one)	**Yes No**

13. If you have answered YES for either question 11 or 12, please explain:

APPENDIX B

FOOTEDNESS EVALUATION

Subject ID: ____________________ Date:_____________
Date of Birth: ______________________ Time:______________
Gender: _________________________

Instructions: Have subjects perform each activity four times; create a tally mark under the appropriate column. Use the scoring key to assign the appropriate score to each task.

	Left Leg	Right Leg	Score
1. Have subject kick 4 stress balls straight ahead, in between the legs of a chair.			
2. Have subject balance on one leg for 10 seconds. And relax for 5 seconds in between.(4X)			
10. Relaxed standing (which leg is slightly bent?)(4x)			
3. Have subject smooth the sand in the sand box.(4x)			
4. Have subject step on to the step ladder. Which foot went first? (4x)			
5. Roll stress ball from right to left and have subject stomp on the ball to prevent it from reaching a target to their left. (2x) Repeat, this time rolling the ball from left to right. (2X)			
6. Have subject balance on one foot on a metal beam for 10 seconds. (4x)			
7. Have subject pick up 4 marbles with his or her toes.			
8. Have subject hop on one foot and then relax. (4x)			
9. Have subject use a foot to press the foam alphabet into the appropriate slot.(4x)			
	Score Total:		
11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities? (choose one)	Yes	No	
12. Have you ever been given special training or encouragement to use a particular foot for certain activities? (choose one)	Yes	No	

13. If you have answered YES for either question 11 or 12, please explain: ____

____

SCORING:

Right always	=	2
Right usually	=	1
Equal	=	0

APPENDEX C

MICROCONTROLLER CODE

Description:

This program shows an example of using ADC0 in interrupt mode using Timer3 overflows as a start-of-conversion to measure the voltages on AIN0 through AIN7 and the temperature sensor. The voltages are calculated from the resulting codes and are transmitted out UART0. Assumes a 22.1184MHz crystal is attached between XTAL1 and XTAL2. The system clock frequency is stored in a global constant SYSCLK. The target UART baud rate is stored in a global constant BAUDRATE. The ADC0 sampling rate is stored in a global constant SAMPLERATE0. The voltage reference value is stored in a constant VREF0, and is used to convert the resulting codes from the ADC0 measurements into a voltage.

Key Global Definitions:

```
#define SYSCLK 22118400 // SYSCLK frequency in Hz
#define BAUDRATE 115200 // Baud rate of UART in bps
#define SAMPLERATE0 2000000
```

Global VARIABLES:

```
bit TX_Ready;                          // '1' means okay to TX
char *TX_ptr;                          // pointer to string to transmit
long result[8];                        // AIN0-7 output
char message;
bit RX_Ready;                          // '1' means RX string received
char idata RX_Buf[RX_LENGTH];          // receive string storage buffer
```

```
//char idata TX_Buf[RX_LENGTH];                  // transmit string storage buffer
```

MAIN Routine:

```
void main (void) {
        long voltage;                            // voltage in millivolts
        int i;                                   // loop counter
        double dmvolts;
        int done;
        WDTCN = 0xde;                            // disable watchdog timer
        WDTCN = 0xad;
        SYSCLK_Init ();                          // initialize oscillator
        PORT_Init ();                            // initialize crossbar and GPIO
        UART0_Init ();                           // initialize UART0
        Timer3_Init (SYSCLK/SAMPLERATE0);        // initialize Timer3 to overflow at
                                                 // sample rate
        ADC0_Init ();                            // init ADC
        AD0EN = 1;                               // enable ADC
        EA = 1;                                  // Enable global interrupts
        done=1;

        while (done)
        {
        message=getchar();
          if (message== 'Q')
                done = 0;
          else
                {
```

```
                if (message== 'S')
                  {
                   printf("/n");
                   for (i = 0; i < 8; i++)
                     {
                     EA = 0;                      // disable interrupts
                     voltage = result[i];         // get ADC value from global variable
                     EA = 1;                      // re-enable interrupts
                                                  // calculate voltage in millivolts
                     voltage =voltage * VREF0;
                     dmvolts=voltage;
                     dmvolts=dmvolts/4096;
                     voltage=dmvolts;
                     printf("%ld/n",voltage);     //print voltage only
                      }                           //end of for loop
                  }                               //end of if
              }                                   //end of else
          }                                       //end of while loop

}                                                 //end of main loop
```

Key Subroutines:

SYSCLK_Init:

This routine initializes the system clock to use an 22.1184MHz crystal as its clock source.

void SYSCLK_Init (void)

```
{
int i;                                  // delay counter
OSCXCN = 0x67;                          // start external oscillator with
                                        // 22.1184MHz crystal
for (i=0; i < 256; i++) ;               // XTLVLD blanking interval (>1ms)
while (!(OSCXCN & 0x80)) ;              // Wait for crystal osc.  to settle
OSCICN = 0x88;                          // select external oscillator as
                                        //SYSCLK
                                        // source and enable missing clock
                                        // detector
}
```

ADC0_ISR

ADC0 end-of-conversion interrupt service routine. Here we take the ADC0 sample and store it in the global array <result>. We also select the next channel to convert. During this routine, we take the ADC sample, add it to a running total <accumulator>, and decrement our local decimation counter ,int_dec>. When <int_dec> reaches zero, we calculate the new value of the global variable <result>, which stores the accumulated ADC result.

```
void ADC0_ISR (void) interrupt 15
{
static unsigned char channel = 0;       // ADC mux channel (0-8)
static unsigned int_dec=INT_DEC;
static long accumulator=0L;
AD0INT = 0;                             // clear ADC conversion complete
                                        // indicator
accumulator+=ADC0;
```

```
int_dec--;
if(int_dec==0)
  {
        int_dec=INT_DEC;
        result[channel]=accumulator/INT_DEC;
        accumulator=0L;
        channel++;                              // change channel
        if (channel == 8)
           {
           channel = 0;
           }
           AMX0SL = channel;                    // set mux to next channel
}
  }
```

APPENDIX D

MOTOR LEARNING LABVIEW CODE

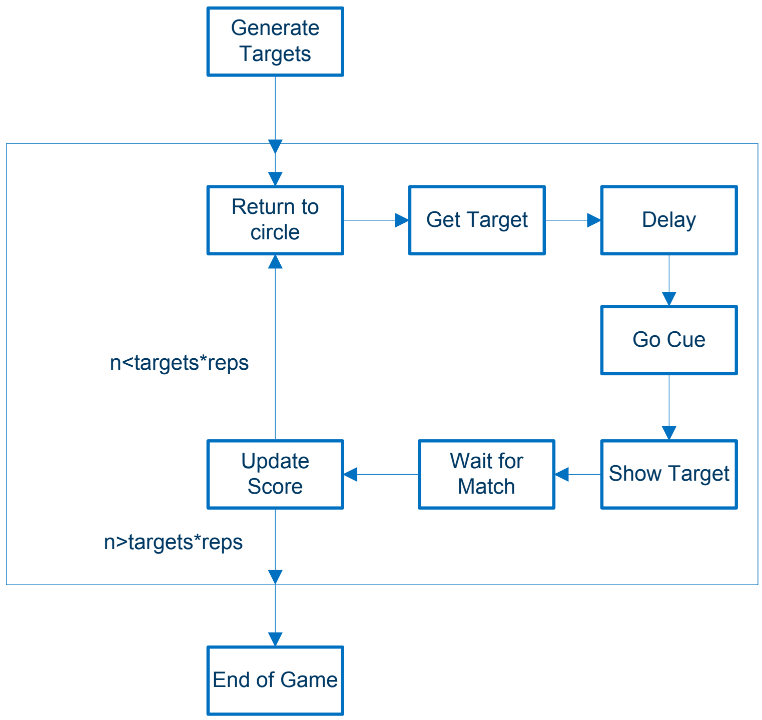

Figure 40 Flow Chart for Motor Learning LabVIEW Code

Figure 40 Illustrates the logic used to design the LabVIEW code for the motor learning experiments described earlier in this thesis.

APPENDIX E

ENGINEERING DRAWING OF ANKLE DEVICE

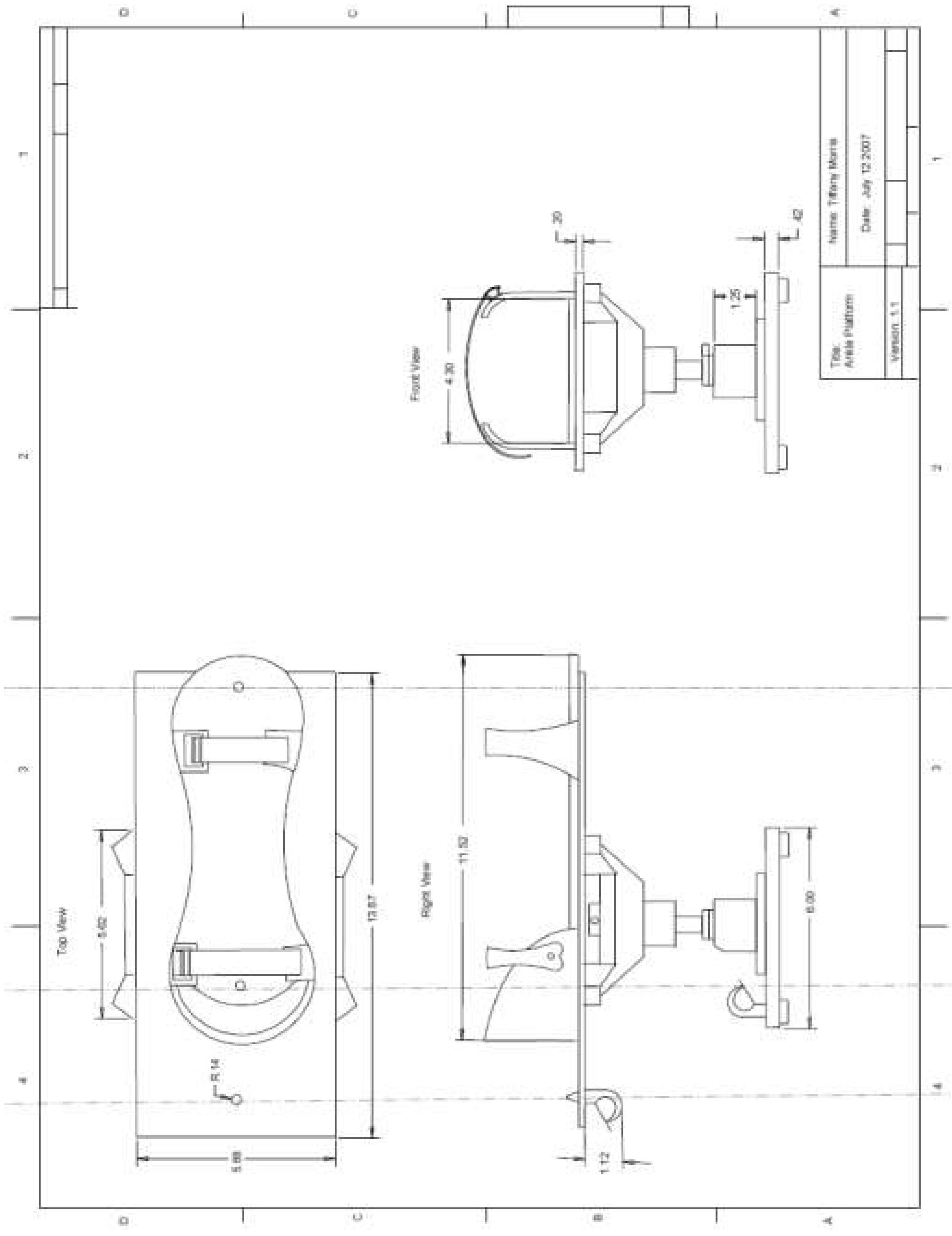

Figure 41 Ankle Platform Drawing

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