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Development of a Closed-Loop Force Reduction Mechanism in a Gait Rehabilitation Device

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A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science at Virginia Commonwealth University.

By

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Abstract

DEVELOPMENT OF A CLOSED-LOOP FORCE REDUCTION MECHANISM IN A GAIT REHABILITATION DEVICE

By Jeffrey A. Frankart, B.S. Mechanical Engineering Technology

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science at Virginia Commonwealth University

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Elliptical trainers are prescribed in rehabilitative exercise but difficult to implement in populations with significant functional gait deficits. Typical elliptical machines do not mimic normal gait and therefore require modifications for clinical rehabilitation. This research builds on previous modifications of an elliptical trainer designed to simulate level-surface walking. This design differed from a commercial version. It included articulated footplates and an *electromechanically-driven virtual-cam* to control footplate position. Ankle dorsiflexion elicited lower-extremity muscle spasticity which produced an unwanted gait variant during stroke patient testing. Spasticity is a hyperexcitable stretch reflex causing inefficient gait. This project's purpose was to develop an *autonomous cam-profile adjustment* to optimize the device's rehabilitation potential. Foot-to-footplate forces were measured in stroke patients and compared to normative data. Greater than normal forces were considered spastic. An embedded controller was designed to reduce footplate forces via real-time cam-profile attenuation. A *simulated spastic dorsiflexion load* successfully proved the algorithm's efficacy.

Keywords: adaptive control, embedded control, spasticity, stroke, virtual camming, rehabilitation, elliptical trainer.

Introduction

Stroke is a disabling and potentially fatal medical emergency affecting the patient and everyone around them. Rehabilitation is intense and exhausting, and is designed to return the patient to a more functional status. Unfortunately, there is not a perfect rehabilitation protocol that fits every stroke victim. The options for rehabilitation vary in the timing of the intervention, the amount of human labor required, and the equipment needed.

This project consisted of modifying an elliptical exercise device intended for use in the rehabilitation of stroke patients to improve its efficacy. These modifications were necessary in order to reduce the impact of an unanticipated side effect resulting from the previous design. The previous design articulated the footplates of the elliptical trainer to mimic a normal lower extremity gait pattern. This is a useful tool in the rehabilitation of patients who have suffered stroke since it helps them to practice (and reinforce) a pattern that is similar to normal walking. The unfortunate side effect of this design was muscle spasticity. This is characterized as a velocity dependent reaction to muscle stretch that produces a hyper-reflexive contraction. It was hypothesized that these aberrant contractions would be evident in ground reaction forces, so the design was modified by adding load cells to measure these forces. Using this force measurement as a control variable, an algorithm was designed to tailor the movement of the footplates to the patient and reduce spasticity. The improved device should be more efficient in helping patients recover from stroke.

Stroke and Chronic Effects of Stroke

Seven hundred eighty thousand people suffer strokes each year (1). Many of these stroke victims die, but the survivors have long lasting effects requiring them to relearn common tasks. The effects of a stroke depend on which area of the brain was affected and the severity of the injury. A loss of strength or an inability to control fine motor movements on only one side of the body, a condition known as hemiparesis, is typical of stroke victims. The hemi- prefix means half, indicating that only half of the body is paretic. The paretic side is opposite to the stroke site in the brain.

There are two main categories of strokes, occlusive and hemorrhagic. Regardless of classification, a stroke deprives a portion of the brain of the normal blood flow. Blood carries oxygen and nutrients to the brain and carries waste products, such as lactic acid, away from the brain (2). Without the appropriate blood flow, toxins accumulate in that portion of the brain. The accumulation of toxins and lack of nutrition and oxygen cause brain matter to die, a condition known as infarction (2).

Occlusive strokes result from ischemia, a lack of blood flow to the brain due to a blockage. The blockages are the result of either a thrombus or an embolism. A thrombus is a buildup of deposits within the artery to the point of complete closure. The blockage occurs at the site where the buildup occurred. An embolism is a deposit which breaks off from the site where the buildup occurred and blocks an artery downstream as the arteries branch into smaller diameters. The other main category of stroke is the hemorrhagic stroke. A hemorrhagic stroke is the result of blood starvation because of bleeding from the arteries which carry blood to the brain. This is usually caused by hypertension or aneurysms (2).

Stroke Rehabilitation

Current research focuses on designing the most effective therapy to help the patient gain independence following stroke. Task-specific training has proven to be the most effective indicating that specific tasks should be rehabilitated instead of general muscle movements (19). Medical costs cause concern for patients and their families, employers, and insurance companies. Insurance companies are reluctant to cover unproven therapies, making experimental treatment burdensome for the family.
Basic tasks, such as walking, require rehabilitation for hemiparetic stroke patients. A task-specific, cost-effective training protocol does not exist. A commercially available elliptical trainer is the closest device to level surface walking available, but it typically keeps the foot in a toe-down or plantarflexed position throughout the gait cycle. This is not acceptable for stroke rehabilitation as it does not meet the task-specific training condition shown to be effective (19). The patient reaps the maximum benefit from gait rehabilitation with a protocol or device capable of facilitating a dynamic ankle motion (plantarflexed or dorsiflexed position) during the appropriate gait phases (19).

Gait

Gait is defined as the act of walking consisting of a sequence of distinct events and phases. The starting point for a gait cycle is initial contact (IC). IC is the point in the gait cycle in which the heel of a foot (foot 1) strikes the walking surface, while the other foot (foot 2) is still in contact with the ground. IC marks the beginning of the first double-support (DS1) phase and the beginning of the stance phase for foot 1. Weight is shifted from foot 2 to foot 1 during the DS1 phase, an event known as weight-acceptance (WA). The flat foot (FF) event follows WA and coincides with the toe-off (TO) event for foot 2. This is the end of DS1 and the beginning of the first single-support (SS1) phase of the gait cycle. The next event in the cycle is heel-off (HO) where the heel of foot 1 comes off the ground but the toe remains in contact. Contralateral ground contact (CGC) occurs after HO of foot 1. This event is marked by IC of foot 2 which ends the SS1 phase and begins the second double-support phase (DS2). Foot 1 then reaches the TO event, marking the end of DS2 and entering the swing phase for foot 1 to prepare for the next IC.

Each foot is on the ground 60% of the time and off the ground for 40% in a normal gait cycle. The overlapping stance phases (DS1 and DS2) account for the disparity between stance and swing time.

In this research, an elliptical trainer was modified to simulate level-surface walking with closed-loop force data to create a cost-effective gait rehabilitation device. The modified elliptical trainer footplates move the foot into the desired position consistent for each gait phase and event. The footplate movement pattern is learned by the patient over time and the brain relearns the appropriate muscle firing patterns.

Muscle Control

Each portion of the brain specializes in a particular function, such as speech, motion, or memory. The function performed by that area is lost or hampered when an infarction impacts an area. For example, there is an area of the brain known as Broca's area, which controls speech. It is located in the posterior portion of the frontal lobe. An infarction to Broca's area does not prevent the subject from speaking, but can alter word choice and prevent the subject from completing complex sentences or phrases (6). Similarly, an infarction in the motor cortex, also located in the frontal lobe, will not cause hemiplegia, but rather hemiparesis. Patients with hemiparesis are able to move the affected limb or muscle, but only in a jerky, uncoordinated fashion. Patients with hemiplegia are unable to move the affected body parts in any way.

It is possible for patients with hemiparesis to regain some of the muscle control with rehabilitation. The patient learns how to use their muscles and perform tasks with the affected side of the body during rehabilitation. The part of the brain which controlled the muscles prior to the stroke is no longer viable, but the brain is capable of providing an area unaffected by the stroke to perform tasks practiced during rehabilitation. The function or task previously handled by this area of the brain is lost or degraded as a result. This is known as remapping and demonstrates a concept known as neuroplasticity. The brain's neuroplastic properties allow it to remap itself in the presence of an appropriate stimulus (11). Neuroplasticity is exploited for gait rehabilitation by externally moving both limbs in a pattern similar to walking. The patient uses their own muscles to control the nonparetic limb. External manipulation of the paretic limb is required, however.

Hebb's Theory

Repeatedly moving a patient's paretic limb 180 degrees out of phase with the nonparetic limb creates a pattern of afferent and efferent nerve firing. Hebb's Theory suggests a repeated firing synapse in close temporal proximity to another firing synapse will create a relationship between the two synapses such that the firing of the first will result in the firing of the second. In other words, the neurons that fire together, wire together (13). The tendency to reassign lost function was validated by Castro-Alamancos, et al. in 1992 using Wistar Rats and strategically placed legions to the brain (14). Hebb's theory, as applied to gait rehabilitation, suggests that the paretic limb will learn to follow the nonparetic limb by 180 degrees, similar to walking. This walking behavior is a learned action.

Edward Taub and the Silver Spring Monkeys

Behavior is learned and reinforced. A stroke patient who tries to use a paretic limb will find the motion uncoordinated. This is negative reinforcement and teaches the patient that the limb does not work well. The patient then learns it is easier to rely upon the unaffected side to perform tasks. This learned behavior strengthens and reinforces the inappropriate response of using the unaffected side exclusively (18).

Edward Taub demonstrated this behavior phenomenon with the controversial Silver Spring monkeys. These monkeys had the sensory nerves of one or both arms deafferented. The monkeys with only one arm deafferented used the unaffected limb exclusively while the deafferented arm was unused and atrophied, labeled “learned nonuse.” Both arms were deafferented in subsequent experiments. The monkey continued to use both arms as a means to survive in this case. Taub theorized that the monkeys simply learned that the deafferented limb was not as effective as the unaffected limb, so they learned to use the unaffected limb exclusively. The monkeys were able to use either limb when they had no choice. A condition known as spinal shock was thought to cause some post-surgery effects. Taub conducted a further experiment to evaluate the effects of spinal shock. Similar to previous experiments, one of the monkey’s arms was deafferented. This limb was then constrained after the surgery instead of allowing the monkey to use it. The constraint was removed 3 months after the surgery and the monkey used the deafferented limb just as he had before the surgery. The effects of spinal shock were never observed by the monkey so he did not know that his deafferented limb did not work well (18).

Taub conducted further constraint induced therapy work with stroke patients. This work demonstrated that forcing the patient to use their paretic limb caused a growth in the brain area dedicated to that muscle group in addition to reversing the atrophy started during the learned nonuse period after the stroke (18).

Spasticity

Stroke rehabilitation focuses on teaching the patient how to use their paretic limbs. A complication found during the rehabilitation process is spasticity. Spasticity is an inappropriate activation of the stretch reflex. The stretch reflex is a latent protective mechanism intended to keep muscles from stretching too rapidly. Information from the afferent system signals the muscle to contract in a situation where a muscle is stretched too quickly. The knee-jerk response, observable during a typical medical examination, is an example of a stretch reflex. The motor cortex inhibits the stretch reflex under normal conditions. This inhibition is overcome when the muscle lengthening velocity is greater than the threshold, established as a balancing act between the spinal cord's constant tone signal and the motor cortex's inhibition signal. In patients with an upper motor neuron lesion, the inhibition signal is effectively reduced, so the muscle lengthening velocity required to elicit the stretch reflex is lower (16). A muscle which is contracting at the same time it is lengthening is said to be eccentrically contracting.

An eccentric contraction is not inherently abnormal or indicative of spasticity or other pathological condition. The timing of the contraction within the gait cycle and the magnitude of the contraction characterize spasticity.

Focus of the Study

A therapeutic approach is one in which the patient's muscle is lengthening at a velocity just below the onset of spasticity. This approach maximizes therapy effectiveness while attaining maximum reinforcement of the positive aspects of the movement.

A conservative approach to therapy limits the intensity to eliminate the spastic response. A more aggressive approach can exceed the spasticity threshold, limiting the effectiveness of the therapy or reinforcing poor behavior.

When the lessons learned from the Silver Spring monkeys are applied to stroke victims, the concept of spinal shock is analogous to the stroke event itself and the immediate result is known as cortical shock. The paretic limb is equivalent to the deafferented limb. The patient inappropriately learns that the paretic limb does not work well because he attempts to use it soon after the stroke event itself, demonstrating the “learned nonuse” discussed previously.

Gait Rehabilitation

Stroke rehabilitation has traditionally required a team of medical specialists to work together to evaluate each patient individually and develop a recovery plan. Physical therapists are an integral part of this team. The role of the physical therapist consists of evaluation and rehabilitation of the gross motor skills. According to Hebbian theory, a simulation of level-surface walking through manual manipulation of the hemiparetic limbs has the potential to recreate the muscle-firing pattern necessary to walk (13). Physical therapists provided this manipulation in a team setting with the patient on a treadmill. One of the therapists manually moved the hemiparetic limb in a walking-like pattern while the other therapist moved the patient’s pelvis in an appropriate pattern. The patient’s body weight was supported by an overhead harness during the Body-Weight Supported Treadmill Training (BWSTT). This technique is effective, but labor-intensive for the physical therapists (3). A commercially available device, known as the Lokomat™ (Hocoma, Inc., USA, Rockland, MA), is used in some rehabilitation settings (seen in Figure 1) as an alternative to the physical therapist assisted gait rehabilitation. The Lokomat™ robot attaches to the patient's legs and moves them over a treadmill in a walking motion while the patient's body weight is supported by an overhead harness. This device is effective (15), but expensive and therefore availability is limited (6). Current pricing as of August 2012 is $345,000 according to the Hocoma sales team. Despite the expense and limited availability, the Lokomat™ also has its limitations and

Figure 1 Lokomat™ Robotic Gait traineris expensive, has limited availability, and less than ideal joint kinematics

shortcomings. The joint kinematics are not identical to the kinematics found in normal, level-surface walking. Ankle articulation is not controlled with the Lokomat™ and it does not allow for normal rotation of the hip joint during the gait cycle (5). The Lokomat™ is capable of producing an identical gait pattern during each step. This reinforces the positive attributes, but does not allow for any variation which is inevitable during level surface walking, especially in an outdoor environment with rocks, potholes, and other obstacles (5).

The Lokomat™ addresses the main shortcoming of the therapist-assisted BWSTT approach, but introduces its own shortcomings in the abnormal kinematics and high cost, which leads to limited availability. The shortcomings of the therapist-assisted BWSTT and the Lokomat™ are addressed by another device known as an Elliptical Based Robotic Gait Trainer (EBRGT). The EBRGT bridges the gap between the labor-intensive physical therapist team option and the expensive Lokomat™ robot. The EBRGT is a relatively low-cost device which provides gait-like manipulation of the patient's lower extremity kinematics through distal joints. Movable footplates are fitted to the skis of a commercially available NordicTrack™ elliptical trainer and connected to MPP-series servo motors (Parker, Cleveland, OH) and worm drive 60:1 reduction gearboxes (Boston Gear, Charlotte, NC). This drivetrain is controlled by Parker Aries 04CE motor controllers (Parker, Cleveland, OH) with input from a flywheel-mounted encoder (Dynapar, Gurnee, IL). This encoder provides the flywheel position to the motor controller which moves the footplate into the appropriate position for each flywheel position. The encoder also generates an index signal. An index signal is a voltage signal which changes each time the flywheel reaches a specified point. Bradford showed this gait rehabilitation device to be effective for stroke-based hemiparesis (17).

Specific Aims

Specific aims of this project are to develop, model, and simulate a mechanism through which a spastic response can be detected in real-time and apply a corrective measure to reduce or eliminate the spastic response while maximizing the rehabilitative effect on the EBRGT. The gait phase of interest is the swing phase. Figure 2 diagrams the research logic and the scope of this project.

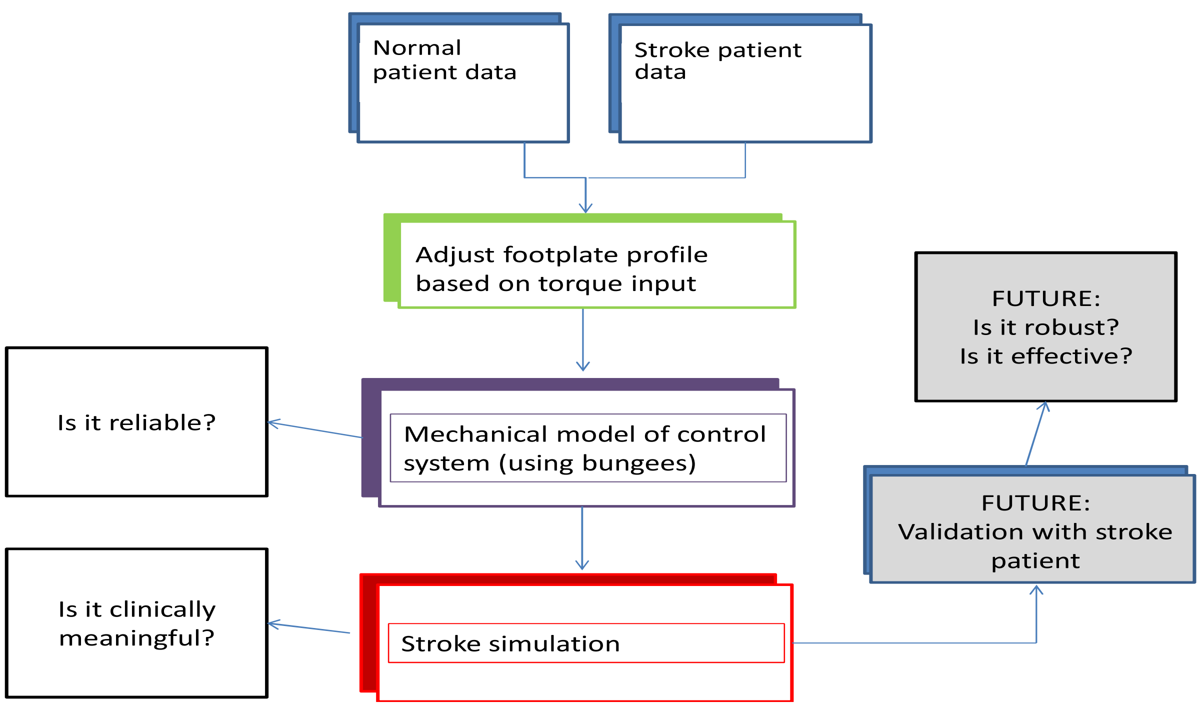

Figure 2 Scope of current research and proposed future research

Methods

THE EBRGT was tested on both healthy, normal subjects (n=20) and post-stroke subjects (n=6). The first data set recorded was a baseline EMG level to capture the latent signal in the subject's muscles. During the subsequent testing, both EMG and video data were captured while the subject ambulated on the EBRGT. Video data were captured at 120 Hz to assess joint kinematics using high speed cameras (Bassler Scout, Bassler Inc., Exton, PA) placed perpendicular to the EBRGT and patient at a distance of 10 feet.

The subjects were prepared for the collection of the video data by placing reflective markers on the subjects with double-sided tape at known points using bony landmarks. The bony landmarks allowed uniform placement across subjects. The landmarks are the acromion (shoulder), greater trochanter (hip), fibular head (knee), lateral malleolus (ankle), heel, and 5th metatarsal. The kinematic data stream was divided into individual gait cycles starting with initial contact of the heel using the index signal powered light. The data was averaged and time-normalized into 100 points per gait cycle. Lengthening velocities of the muscles were calculated from the changes in joint angles using the techniques developed by Winter & Scott (8).

The EMG data collection was performed with a Myosystem 1200 (Noraxon USA, Scottsdale, AZ) EMG device with a 12-bit Analog-to-Digital converter sampled at 1000

Hz and stored with ***The MotionMonitor*** software V7.0 (Innovative Sports Training, Chicago, IL) on a Dell laptop computer. The EMG data was captured through surface electrodes placed on the subjects and included the EMG data readings from the vastus lateralis (VL), tibialis anterior (TA), biceps femoris (BF), and lateral gastrocnemius (LG). These muscles are intended to represent their respective areas as table 1 shows. These data were divided into individual gait cycles starting from heel strike on the right side based on the index signal from the flywheel encoder. Each gait cycle was resampled to 100 points and the results were averaged over each 30 second data collection period.

The kinematic and EMG data were synchronized using the EBRGT flywheel encoder index signal. Only the data points during the swing phase with EMG levels above the baseline level and positive lengthening velocities were analyzed.

Table 1 The muscles instrumented for the EMG data represent muscle groups

Muscle	Abbreviation	Muscle group	Location
Vastus Lateralis	VL	Quadriceps	front of upper leg
Tibialis Anterior	TA	Shin	front of lower leg
Biceps Femoris	BF	Hamstring	back of upper leg
Lateral Gastrocnemius	LG	Calf	back of lower leg

When using the EBRGT on patients who had suffered stroke, it was noted that their heel sometimes left the footplate during the gait cycle. This was indicative of a spastic response of the ankle plantar flexors and presented a problem that could hinder the rehabilitation of the normal gait cycle the EBRGT was trying to facilitate.

Using the post-processing technique developed by Lamontagne, et al., a positive correlation between the lengthening velocities and EMG signal confirms the spastic response. A positive correlation indicates that the muscle is more electrically active as the muscle stretches faster (19). In the study by Lamontagne, et al., the correlation coefficient for the paretic limb was 0.62 compared to 0.52 in this project. The healthy control subject in the study by Lamontagne, et al. had a correlation coefficient of -0.81 compared to the -0.45 found in the control subject in this project. Finally, the nonparetic limb of the same patient in Lamontagne's study, had a correlation coefficient of -0.31 compared to the -0.53 in this study. The disparity in the correlation coefficients between this project and the data in Lamontagne's study is likely due to a smaller sample size. Lamontagne, et al. also averaged the values after time normalizing, but created only 50 data points per gait cycle instead of the 100 in this data set.

Figures 3 and 4 are representative EMG and muscle lengthening velocity correlation plots of one of the six patients with a history of stroke and spasticity. Figure 5 is the same data from a healthy, normal subject. The stroke patient's right side is paretic. Figure 3shows the right (paretic) side. The positive correlation(r=0.52) between EMG activity and muscle lengthening velocity is indicative of a spastic response. Figure 4 shows the stroke patient's left (nonparetic) side. The negative correlation (r=-0.53) between the EMG activation and the muscle lengthening velocity is normal and similar to Figure 5 in which a healthy patient also exhibits a negative correlation (r=-0.45) between EMG activation and muscle lengthening velocity.

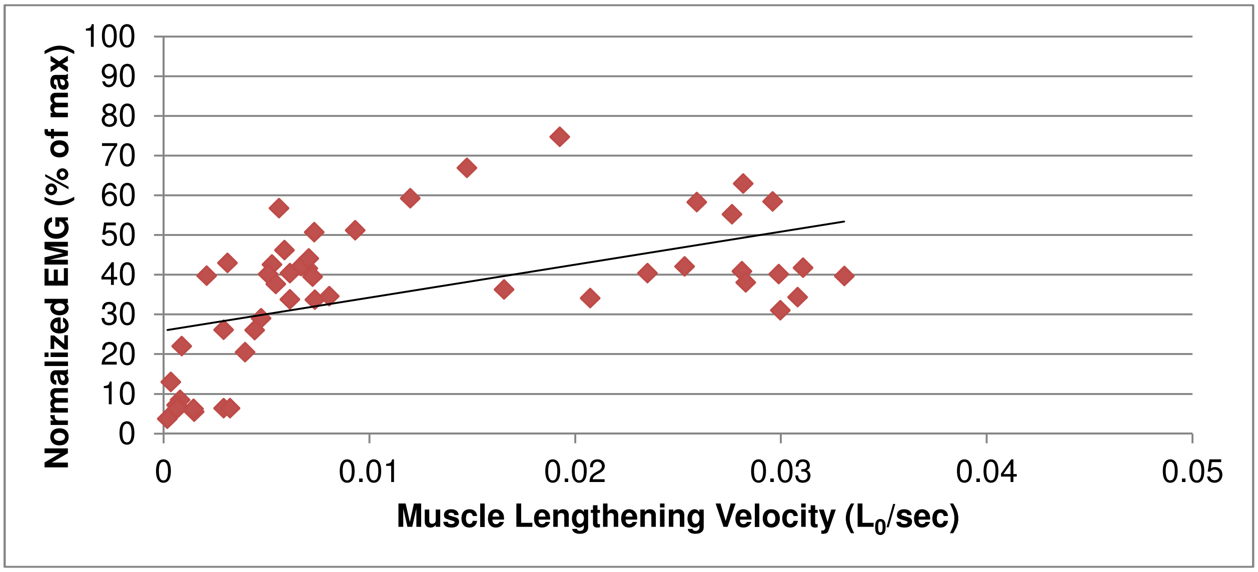

Figure 3 Positive correlation demonstrates a spastic response

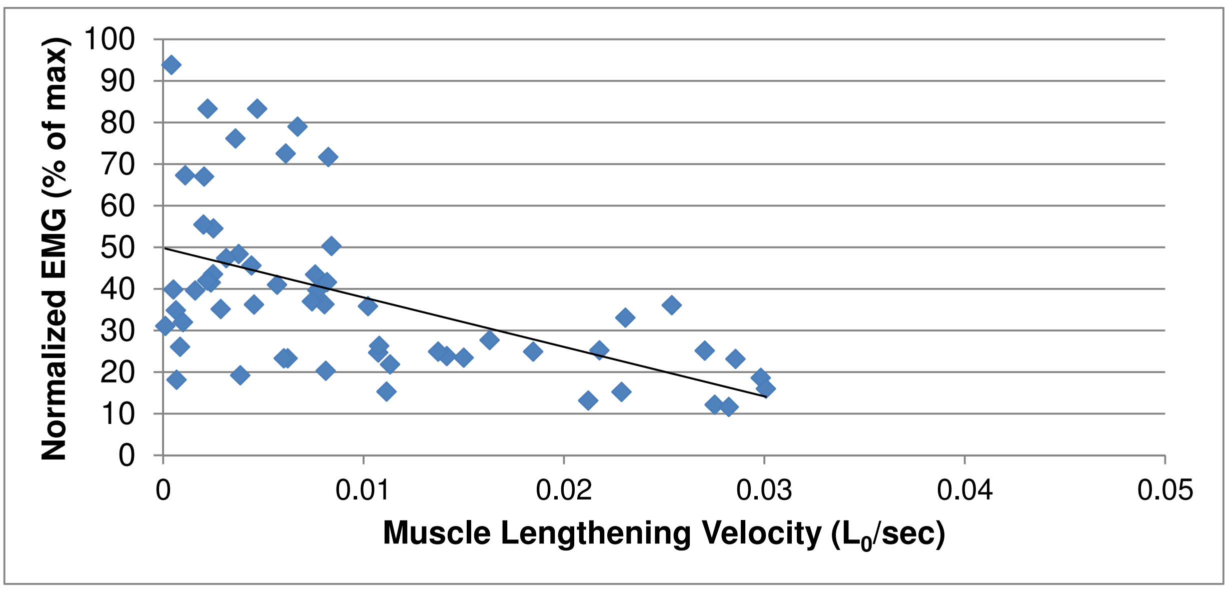

Figure 4 Negative correlation demonstrates a normal, non-spastic response

These plots clearly demonstrate the spastic response but a real-time intervention using EMG data is not possible because of the extensive post-processing required. A method of detecting spasticity that can provide feedback in near real time is needed. A patient exhibiting spasticity on the EBRGT will learn and reinforce a spastic pattern if a corrective action is not implemented.

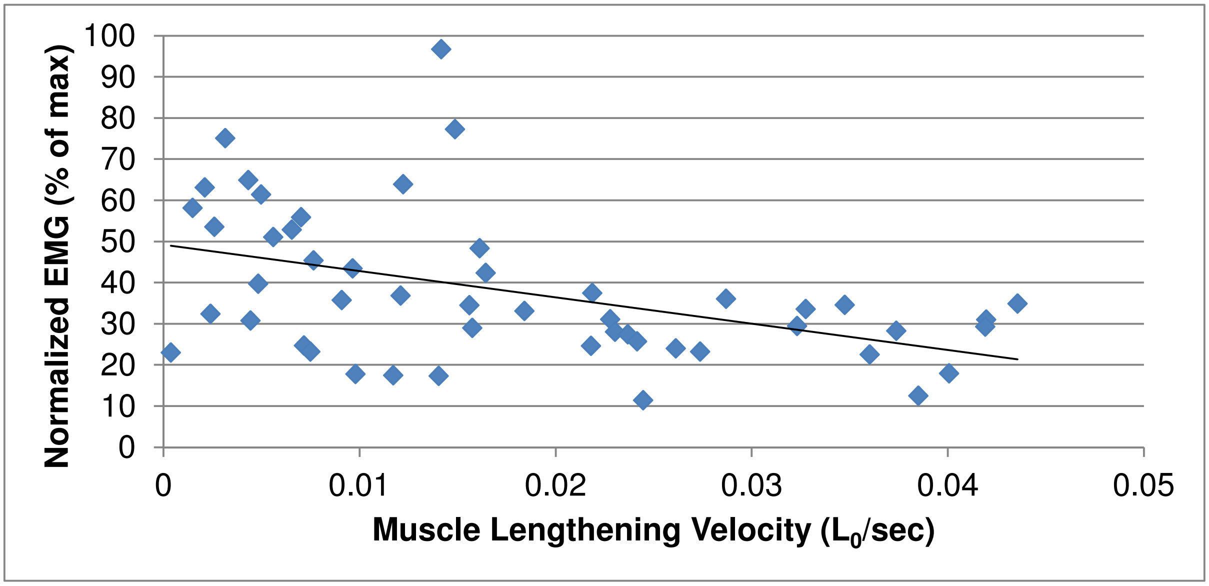

Figure 5 Negative correlation demonstrates a non-spastic response from a healthy, normal subject

The electrical activity of the muscle detected by the EMG signal is correlated with force generation within the muscle. A stronger EMG signal is indicative of a higher level of force. A high level of force from a muscle at a point in the gait cycle when it should be stretching is indicative of spasticity, but this was only detectable during the post-processing of the EMG signal, kinematic video data, and the index signal from the flywheel encoder. In post-stroke subjects, these data showed that the calf muscle generated a stronger EMG signal while the muscle was lengthening at a faster rate.

It was found that the pre-programmed pattern of the footplate stretched the muscle too quickly eliciting the spasticity. The muscle activated to resist the stretch. If this force can be detected in real time, the pre-programmed movement profile can be altered to reduce the speed at which the footplate moves and stretches the muscle. To detect a force at an inappropriate time, a load cell is needed within the EBRGT system. To accomplish this, a load cell (Interface SSM-AJ-500, Scottsdale, AZ) was added in the pushrod between the motor-driven gearbox and the pivot arm for the footplate (identified by the purple box in the lower left corner of Figure 6). Based on the lever type motion of the footplate, this load cell is more appropriately termed a torque cell. The load cell signal, conditioned by the amplifier (Interface SGA, Scottsdale, AZ), is converted by the 12-bit A/D card and captured by ***The MotionMonitor***® software on a Dell laptop PC.

The modified pushrods with inline load cells required careful placement to avoid interfering with the pivoting footplates and allowing the existing EBRGT hardware used to determine the fore and aft limits to remain in place and unencumbered (footplates outlined in orange, swing arm circled in red, limit switches circled in yellow, Figure 6). The swing arm and end-of-travel switches are in place to prevent the motor from moving the footplate too far. This would cause damage to the ski and footplate from the mechanical interference. After the load cells were mechanically installed, the amplifiers were connected to supply power to the load cells and condition the output signal from the load cells. These amplifiers are powered by a 24V DC power supply already on the EBRGT.

Torque Variables

The output from the load cells is a voltage proportional to the amount of force exerted by the pushrods in either tension (the pushrods are trying to pull apart from each other) or compression (pushrods squeezing the load cell between them). Because the load cell is acted on by a lever arm (the footplate and swing arm), it is actually measuring torque.

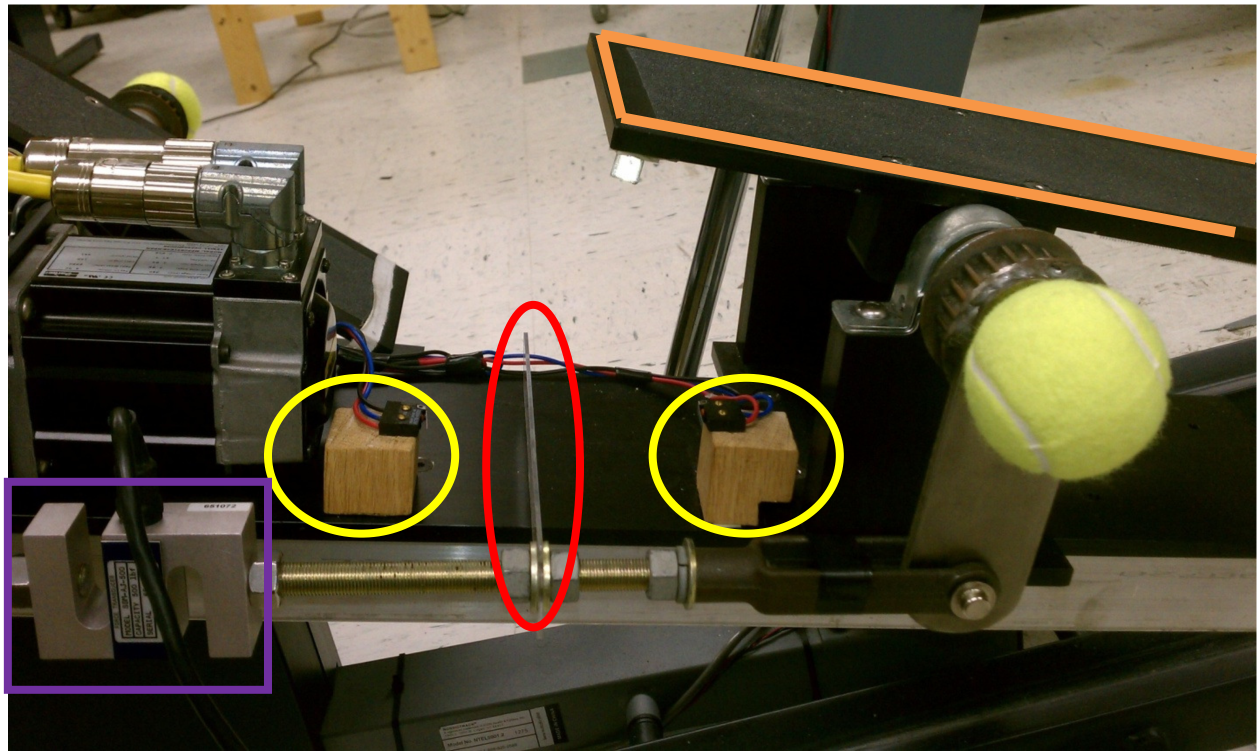

Figure 6: The location of the inline load cells (purple box) added during this project was carefully considered to prevent interference with the swing arm (circled in red) and end of travel switches (circled in yellow). The load cell measures the tension or compression in the pushrod generated by footplate (orange) torque.

Torque is a measure of rotational force found by multiplying the force by the distance at which the force is applied. More torque is generated with the same force when it is applied at a greater distance. This is relevant to the EBRGT load cells when comparing subjects with varying foot size. Uniform foot placement practices were adopted to minimize intrasubject and intersubject reliability.

Calibration of the load cells was required before use to verify range, sensitivity, and zeroing of the load cells and amplifiers. This process is outlined in Appendix F.

A spastic response is seen by the EBRGT as a force exerted at an inappropriate time during the gait cycle. The shape of a force curve during a healthy, normal gait cycle is not clearly known, however. This requires a control group to establish the basis for normal against which an unknown subject can be tested.

Establishing Normal

Data were collected with 10 healthy, normal subjects using the EBRGT. Subjects were allowed to run on the device for several minutes to become accustomed to the kinematics of the device. The kinematics of the EBRGT are slightly different than conventional, level-surface walking because of the 50/50 stance/swing gait cycle instead of the conventional 60/40 stance/swing cycle. The load cell signals, sampled at 1 kHz, provide a force curve over the course of a typical gait cycle. Data were collected three times each on ten healthy subjects for 30 seconds each time.

During preliminary data analysis, fore-to-aft foot placement on the EBRGT footplate was determined to be an error source. A foot placement protocol was instituted to minimize intrasubject and intersubject reliability. Subjects' foot size ranged from European size 36 to 47 (9.25 inches to 11.375 inches), making a designated toe or heel point infeasible. Instead, the lateral malleolus (bony projection on the outside of the ankle) was aligned with the pivot point of the footplate. A foot placement jig (see Figure 7) was used to ascertain the distance from the back of the foot to the aft edge of the footplate during the first visit for each subject. The distance of the foot placement jig was recorded with the subject's other pertinent information (shoe size and weight). This allowed more consistent foot placement from fore to aft during subsequent visits.

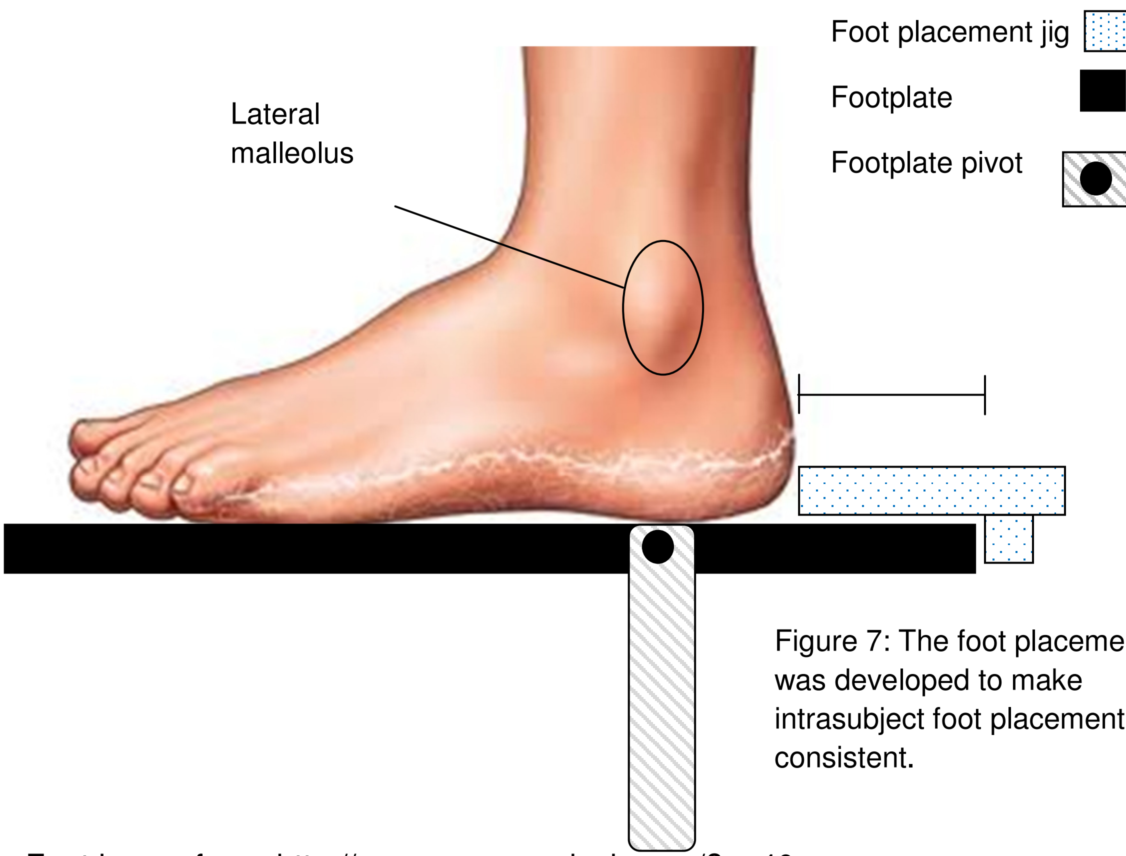

Figure 7: The foot placement jig was developed to make intrasubject foot placement consistent.

Foot image from: http://www.nycnewsdesk.com/?p=40

The hemiparetic subject with a probable spastic response while walking on the EBRGT returned for additional data collection. Before his arrival, however, changes were made to the programming code for the EBRGT footplate virtual cam profile. These changes allowed an adjustment to the virtual cam profile to reduce the amount of movement. For these trials, the virtual cam profile was reduced by a given percentage. Kinematic and EMG data on the subject's paretic side were collected. Load cell data was also collected on both sides at 100%, 60%, and 20% of the normal virtual cam profile.

The footplate motion pattern does not change uniformly throughout the gait cycle. The footplate pattern is modified only in the swing phase but not evenly throughout the swing phase. Figure 8 shows the virtual cam profile at various attenuation levels starting with the initial contact event starting the stance phase. The first portion of swing phase is identical for all levels of attenuation. The swing phase consists of 34 points in the virtual cam array. The virtual cam percentage indicates the level of reduction from the 100% profile at the most dorsiflexed footplate position. The cam values are reduced gradually to maintain a seamless footplate movement.

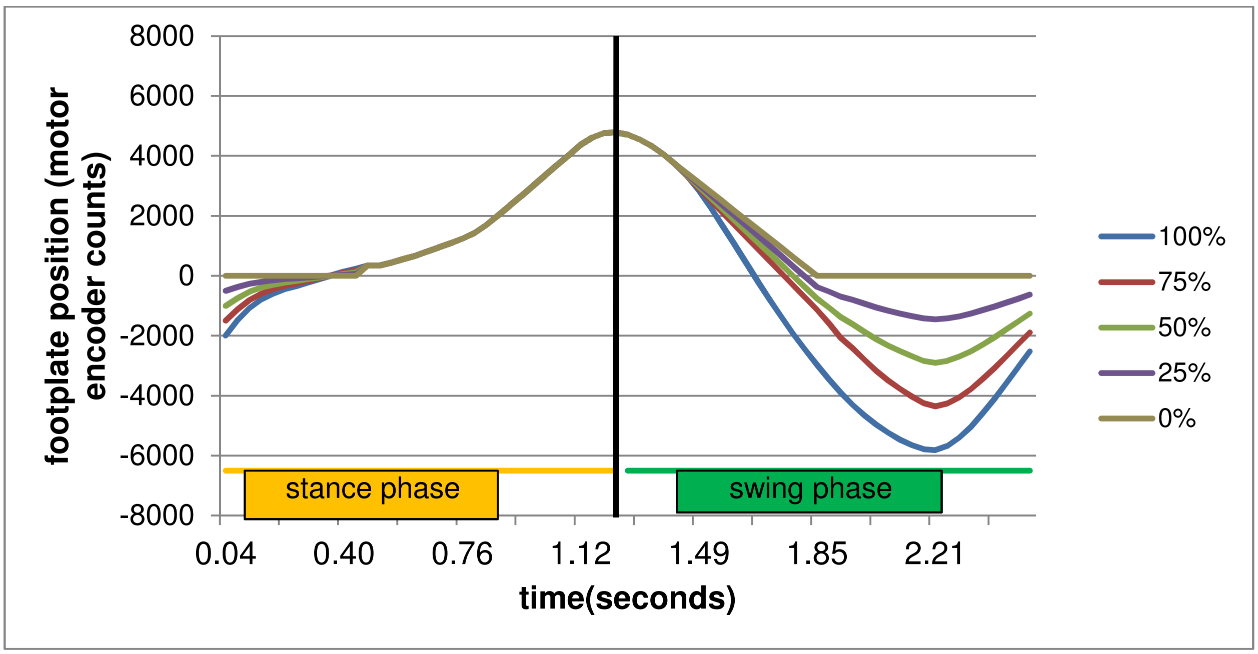

Figure 8 Attenuated footplate patterns. The swing phase is the portion of the gait pattern changed during the closed loop control.

Closed-Loop Control

The EBRGT must reinforce good habits and negatively reinforce bad habits in order to become a viable rehabilitation device. A spasticity detection and reduction algorithm was added to the EBRGT to accomplish this. This necessitated additional hardware and software. The previous circuitry provided a single input to the Parker controllers for each relevant input (Appendix C). The wiring of the EBRGT was modified to incorporate the load cell signals into a separate microprocessor (mbed™ NXP LPC1768) because of the limited number of inputs available on the existing Parker controller. In the new wiring, the stop button was wired in series with the aft travel switch to make an input available. The green button was wired into the mbed™. This provided two inputs to the Parker Controller to be used by the mbed™. Using a combination of the two bits (input 0 and input 3), a total of 4 combinations are available. This provides for the possibilities shown in the table in Appendix D.

The Parker motor controller software was modified to allow the motor controller to interpret the combination of these bits as discrete events. During the startup process, the motor controller starts with a 100% virtual cam. The user can select a different virtual cam profile as a percentage of the original profile through the Parker ACR-View software. A separate option for closed-loop control is also available for selection which allows the Parker motor controller to independently modify the cam profile based on the force feedback from the load cells, instead of the user. This creates a virtually adaptive cam. The logic used in the closed loop control of the EBRGT is depicted in Figure 9.

During actual operation of the EBRGT in the closed-loop mode, the Parker motor controller will decrease the virtually adaptive cam profile in 5% increments when it detects a toe-down force in excess of the programmed limit during the swing phase of gait. In addition, an index signal is generated by the Parker motor controller and available for recording by ***The MotionMonitor***® software. This index signal, in addition to the load

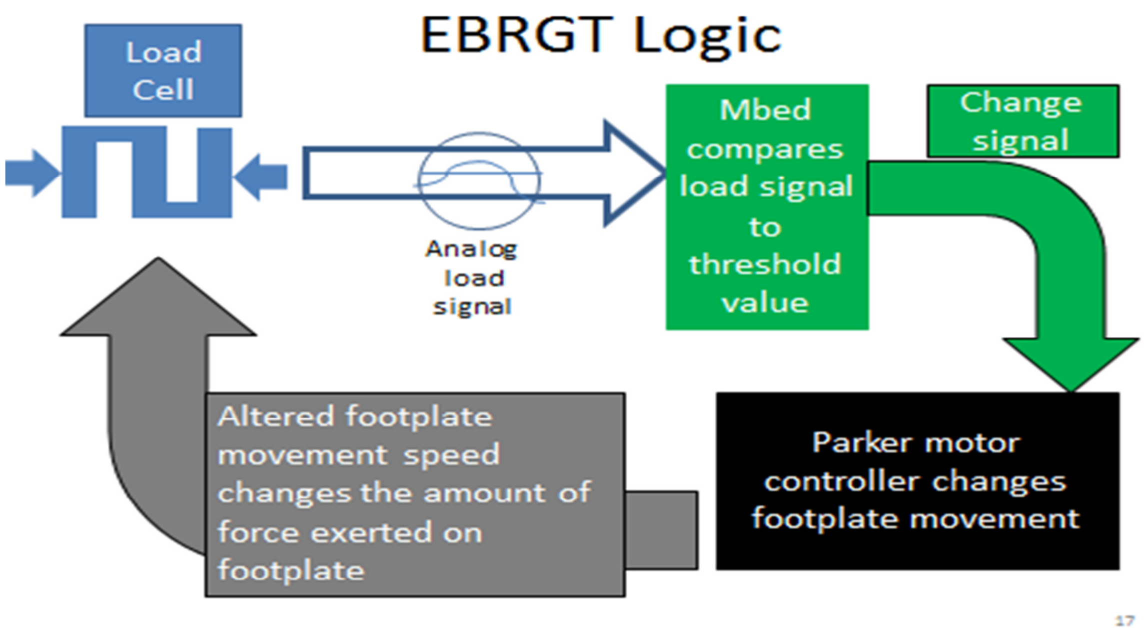

Figure 9 A visual description of the EBRGT closed-loop control logic. The load cell provides a real time control signal. The mbed™ (embedded controller) determines the direction of foot plate attenuation and provides a command signal to the Parker motor controller.

cell signals and encoder index signal, synchronizes the data to identify the point in the gait cycle at which the forces occurred and the gait cycle in which the Parker motor controller detected the excessive toe-down force.

The virtually adaptive cam profile reduction algorithm is promising, but the stretching of the patient's muscle should be kept as fast as possible without inducing spasticity. For this reason, while the EBRGT is in the closed-loop mode, the virtual cam profile will increase by 1% when excessive toe-down force is not detected for five consecutive revolutions of the flywheel. This will keep the stretching velocity as high as possible, but still be sensitive to the spasticity.

Quantifying Spasticity

A desire to quantify the amount of spasticity exhibited by hemiparetic subjects and validate the EBRGT's ability to detect a spastic response resulted in additional programming changes to the EBRGT's motors. These changes resulted in a test intended to simulate the Modified Ashworth test performed manually by physical therapists as a way to assess patients. This new test is referred to as the standing perturbation test in the user interface with the Parker ACR-View software. It allows the computer user to select from three speeds for the perturbation to be experienced by the subject standing on the device. It should be noted that this test is not intended to be performed while the EBRGT is being used for gait training. It should only be used with the footplate in a neutral position. Failure to abide by these guidelines will result in interference between the travel limit switches and the footplate pushrod swing arms which will interrupt the test. Starting with the footplate in a neutral position, such as during the mid-stance phase, the computer user will select the standing perturbation test from the menu of options displayed on the screen. The computer will again prompt the user to select from fast, medium, or slow speeds. The footplate will then move into a toe-down position. At that point, the footplate will stop, the motor movement profile is changed to reflect the proper speed (invisible to the computer user and the test subject), and the footplate will move at the proper speed to a toe-up position. ***The MotionMonitor***® software collects the data which will reflect the amount of force exerted by the pushrod onto the footplate and give an indication of the amount of spasticity or resistance to movement exhibited by the test subject.

Logic Circuit

The additional hardware needed for the virtually adaptive cam and standing perturbation test consists of an mbed™ NXP LPC1768 microprocessor and an assortment of resistors, transistors, switches, capacitors and wiring (see wiring diagrams in Appendix G). The green button mounted on the front of the EBRGT is a push button switch used to initiate the homing process. Prior to modifications, this button's output connected directly to the Parker motor controller input. Because of the limited bandwidth available on the Parker motor controller, the switch was connected to the mbed™ microprocessor. This created a voltage compatibility problem which was overcome by using a transistor to change the voltage to an acceptable level. A similar arrangement on the output side of the mbed made communication between the mbed™ microprocessor and the Parker motor controller possible. The load cell amplifier outputs are configured so they can be connected directly into the mbed™ microprocessor. Two switches are used as inputs to the mbed™ microprocessor to determine if a change signal will be sent to the Parker motor controllers and for which side. This is to prevent arbitrary changes to the unaffected side of the patient or test subject.

Results

The resulting torque curve is plotted as a function of time, starting with the event of right-side initial contact. Positive values indicate toe-down force and more negative values indicate less toe-down force or heel-down force. The healthy, normal subjects were instructed to maintain a constant speed of 1.3 MPH as indicated on the gait trainer's LCD readout. A small variance during each cycle and from one cycle to the next is expected. All gait cycles for each subject were resampled to 1500 points using MATLAB. Each 30 second trial produced about nine gait cycles. These force plots can be compared to each other for intrasubject and intersubject reliability and correlation.

The force curves for the healthy, normal subjects were plotted and compared to each other as well as compared for intrasubject reliability. A source of error found in early trials was related to foot placement. The distances from the pivot point of the footplate to the fore and aft ends of the foot determine the magnitude and the scaling of the signal as demonstrated by the moment equations (Appendix F). Figure 10 shows a typical torque plot of a test subject with data collected 2 days apart. The shape of the curves is similar, but scaling and zero offset is different. Both curves decrease slightly after initial contact, increase for the first 0.5 second of stance phase, and then decrease again until the midpoint of swing phase. The torque then increases again and the footplate moves into a dorsiflexed position to prepare for initial contact. The plot starts with initial contact and shows one complete gait cycle, or stride.

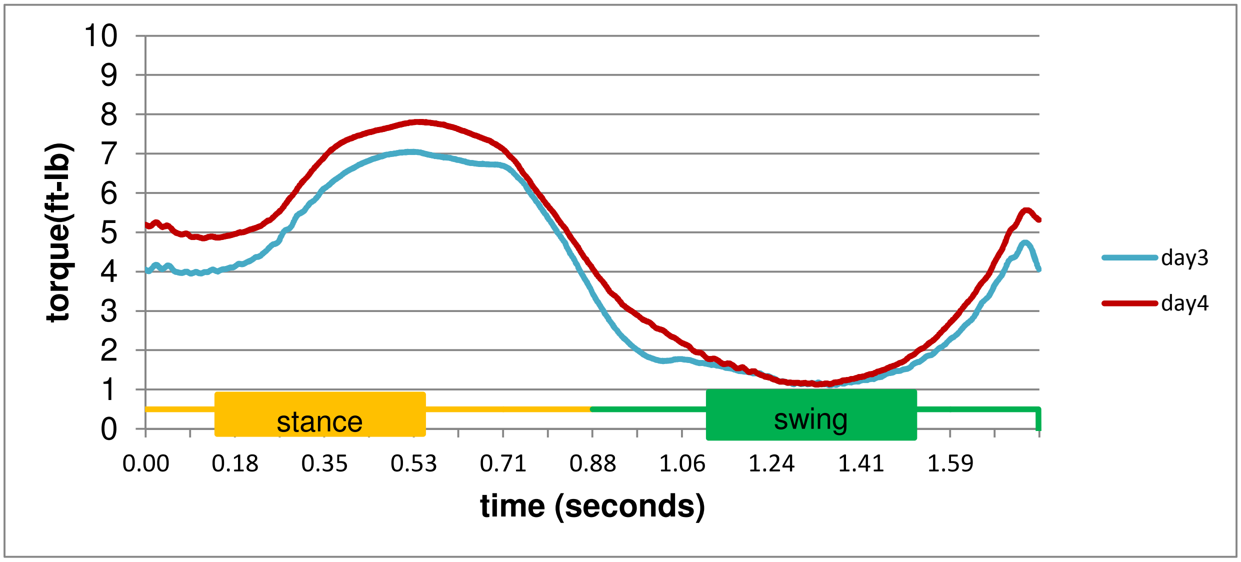

Figure 10: Footplate force pattern from normal subject demonstrating poor intrasubject reliability. r = 0.66

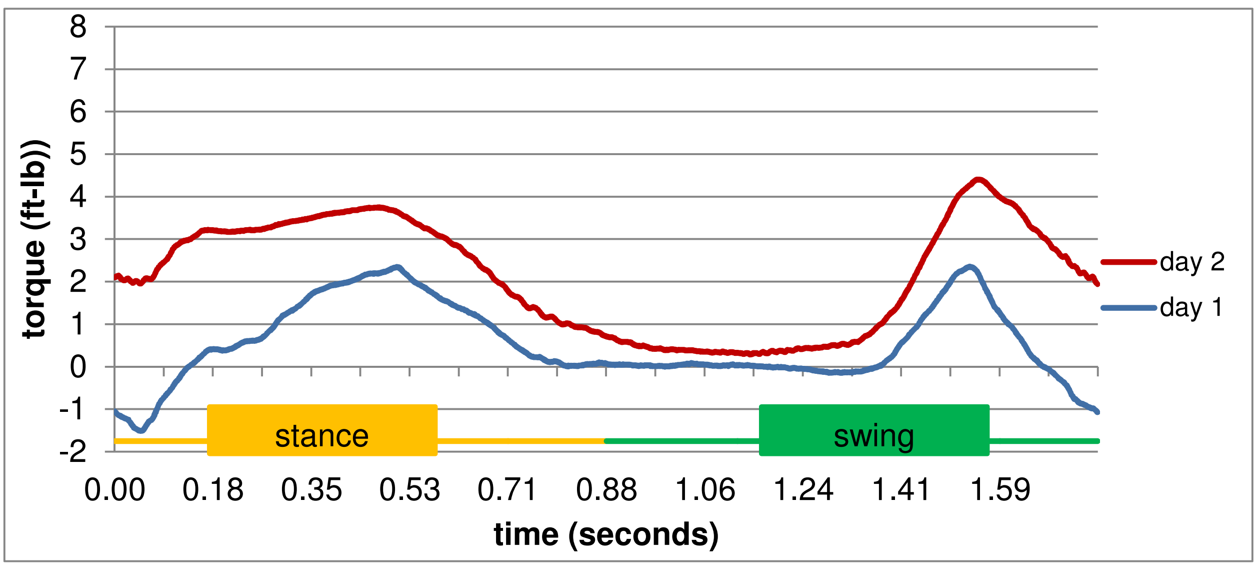

Figure 11: Footplate force pattern from the same subject with new foot placement procedure demonstrating better intrasubject reliability. r = 0.99

The expected force exerted on the load cell increases with subject weight and foot size. To accommodate for these variables, the subject weight multiplied by the length of the foot in front of the lateral malleolus will provide the maximum torque exerted about the footplate pivot. This torque value divided by 4 inches (the length of the pushrod to footplate arm) provides the force exerted on the load cell itself.

After the data collection on healthy, normal subjects was completed the stroke patient with the previously exhibited spastic response returned for additional data collection. Data were collected with the virtual cam set at 3 different steady-state articulations – 100%, 60%, and 20%. The resulting force curves (Figure 12) show a significant difference in the shape of the force curve during the swing phase (far right side of the plot) of gait for his paretic (right) side. This is evidence that spasticity is effectively reduced with muted cam profiles and the resultant decreased footplate angular velocity.

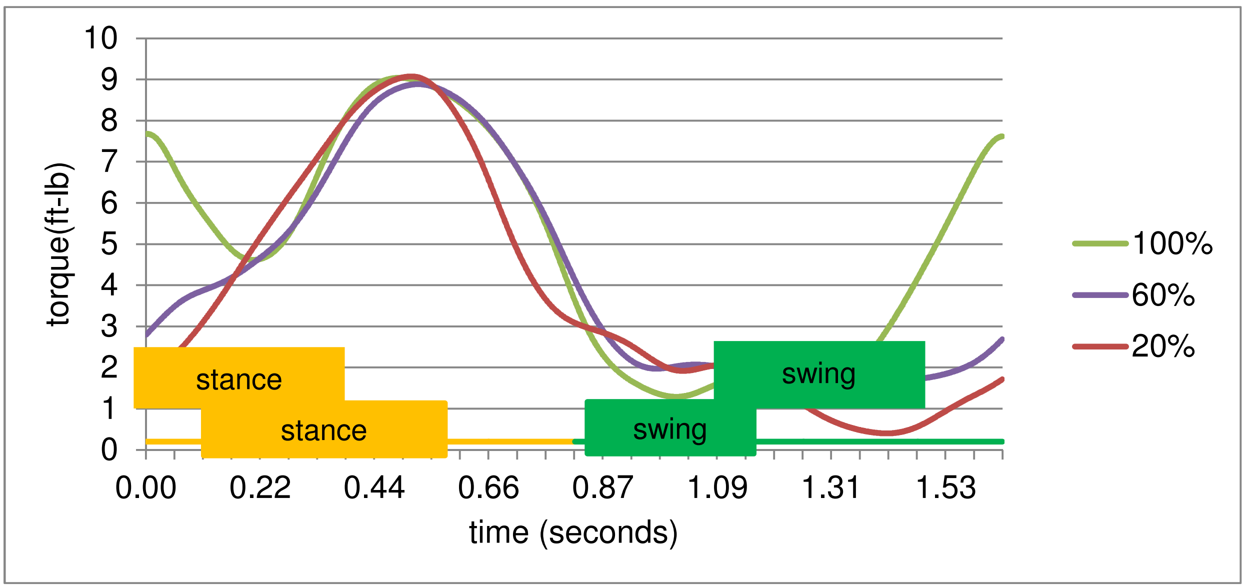

Figure 12: Footplate force pattern from stroke patient with history of spasticity shows a decrease in the level of force generated with decreased footplate movement in the late swing phase.

Modeled Closed-Loop Control

After the stroke patient data was collected and analyzed, additional changes were made to the EBRGT to allow it to autonomously adjust the virtual cam to adapt to the loads seen during the swing phase. The load cell data was provided as an input to the mbed™. The mbed™ software communicated with the Parker motor controller when the toe-down force exceeded a pre-defined threshold. The Parker motor controller software used the information from the mbed™ and the flywheel encoder data to adjust the virtual cam to increase or reduce the footplate motion on the following revolution. In the event the toe-down force exceeds the threshold, the Parker motor controller software is written to adjust the cam profile down by 5% in the following revolution. After 5 consecutive revolutions of no changes, the Parker motor controller software will automatically increase the cam setting by 1% without input from the mbed™. The Parker motor controller software also sends a signal to ***The MotionMonitor***® software to indicate when a virtual cam is changing. This allows the load cell data and flywheel position data to be synchronized with time when the virtual cam changed.

A rubber bungee cord arrangement was developed to test the ability of the mbed™ and motor controller to detect and reduce spasticity. This bungee cord applies a high toe-down force similar to the spastic response from the stroke patient during the dorsiflexed portion of the swing phase as shown in Figures 13 and 14. With this arrangement, toe-down force is increased at greater levels of dorsiflexion, similar to a spastic response. Although it does not accurately simulate the force pattern throughout the entire gait cycle, the rubber bungees create the highest toe-down force during the swing phase just before initial contact, similar to the patient exhibiting spasticity whose data is displayed in Figure 12. Figure 15 shows the force profile of a healthy, normal subject plotted with a representative sample of rubber bungee arrangement and the same spastic stroke patient data at 100% footplate motion as in figure 11. All three data sets were sampled at 100% normal cam profile.

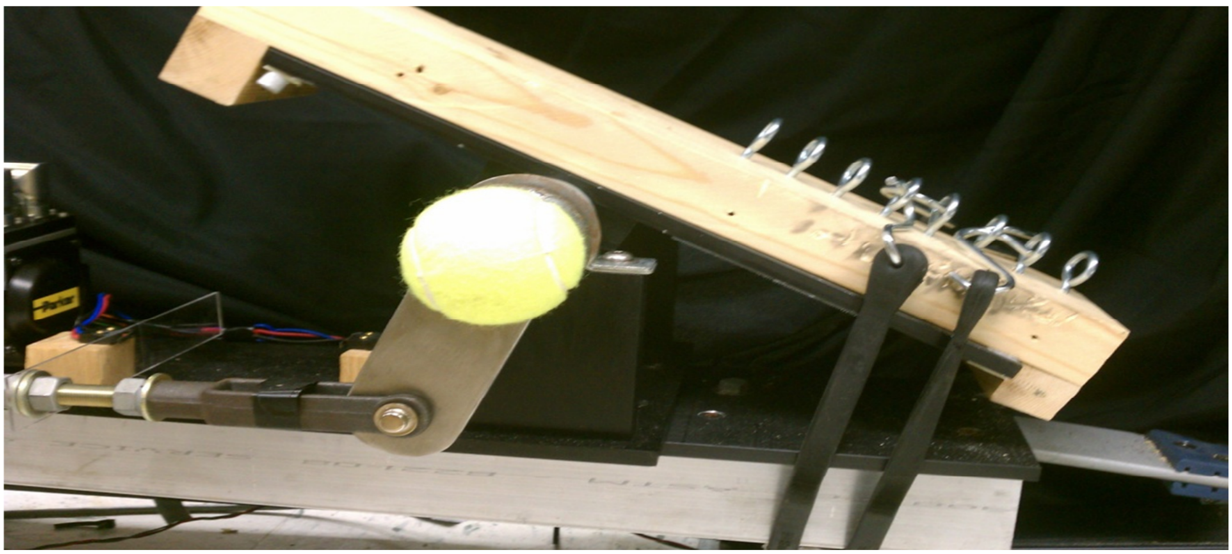

Figure 13: Rubber bungee arrangement used to model a spastic response in the toe-down (plantarflexed) position on the right side footplate at the heel off event in late stance phase.

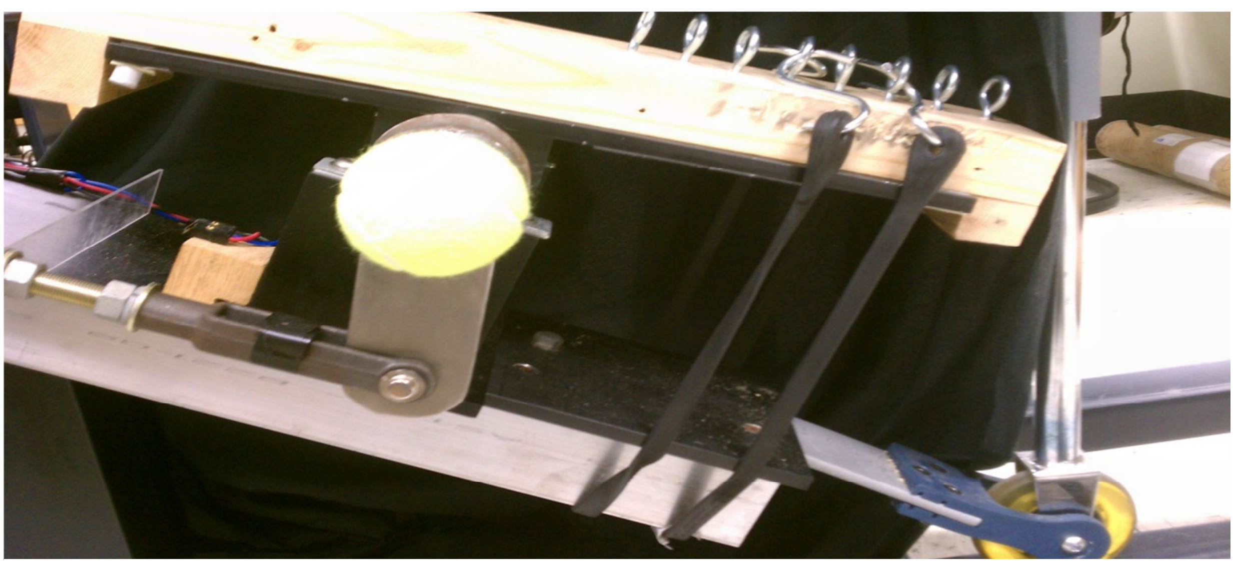

Figure 14: Rubber bungee arrangement in the late swing phase with the footplate in a dorsiflexed position preparing for initial contact. This bungee arrangement simulates increased dorsiflexion force, simulating spasticity.

Like the other force plots, Figure 15 shows the force plot with respect to time. The toe-down force in the late portion of the swing phase at the extreme right end of the plot is sufficient to model a spastic response.

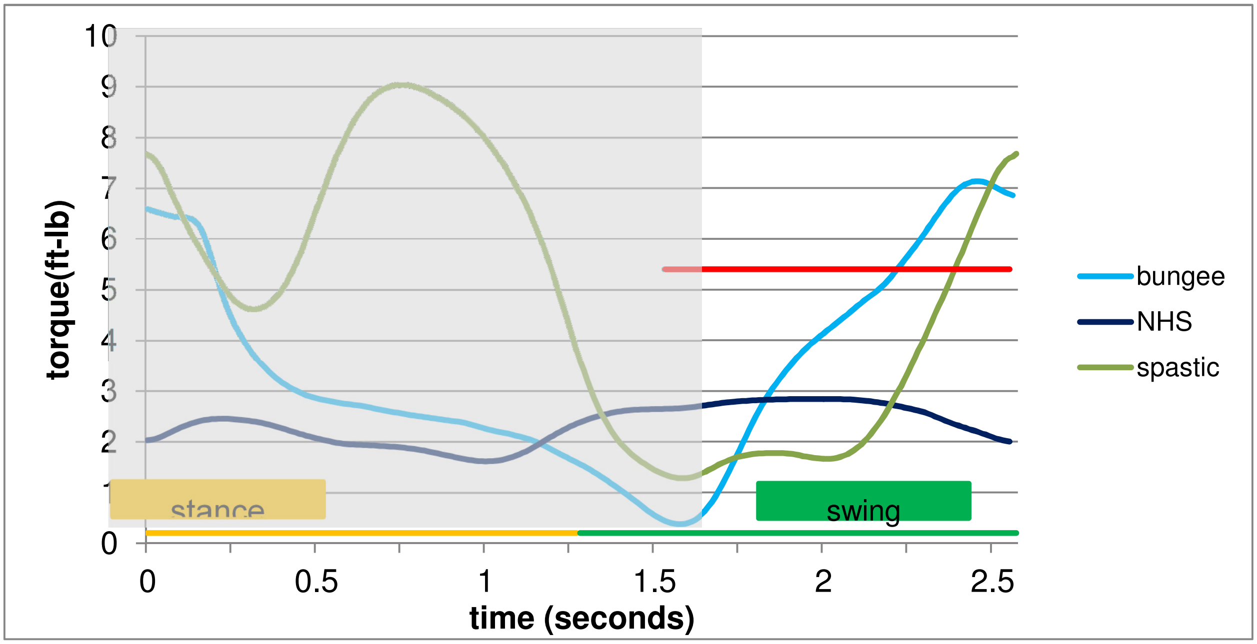

Figure 15: Force profiles of the EBRGT with the modeled rubber bungee arrangement, a healthy, normal subject, and the spastic patient at 100% footplate movement. The threshold line (red) is only active in the late portion of the swing phase and is exceeded by both the spastic patient and the bungee cord modeling. Values close to zero indicate greater toe-down force.

The desired end state of this project is the development of a device that detects and takes action to correct a spastic response. The toe-down force is the critical variable which must be monitored. The virtual cam setting is the variable which must be adjusted when the toe-down force exceeds threshold. In Figure 16a, a higher torque value will result in the virtual cam changing to a lower value on the next cycle. Automatic cam adjustments over four minutes demonstrate the ability of the mbed™ and Parker motor controller to make the 1% increases and 5% decreases to keep the torque levels at or near the preset threshold.

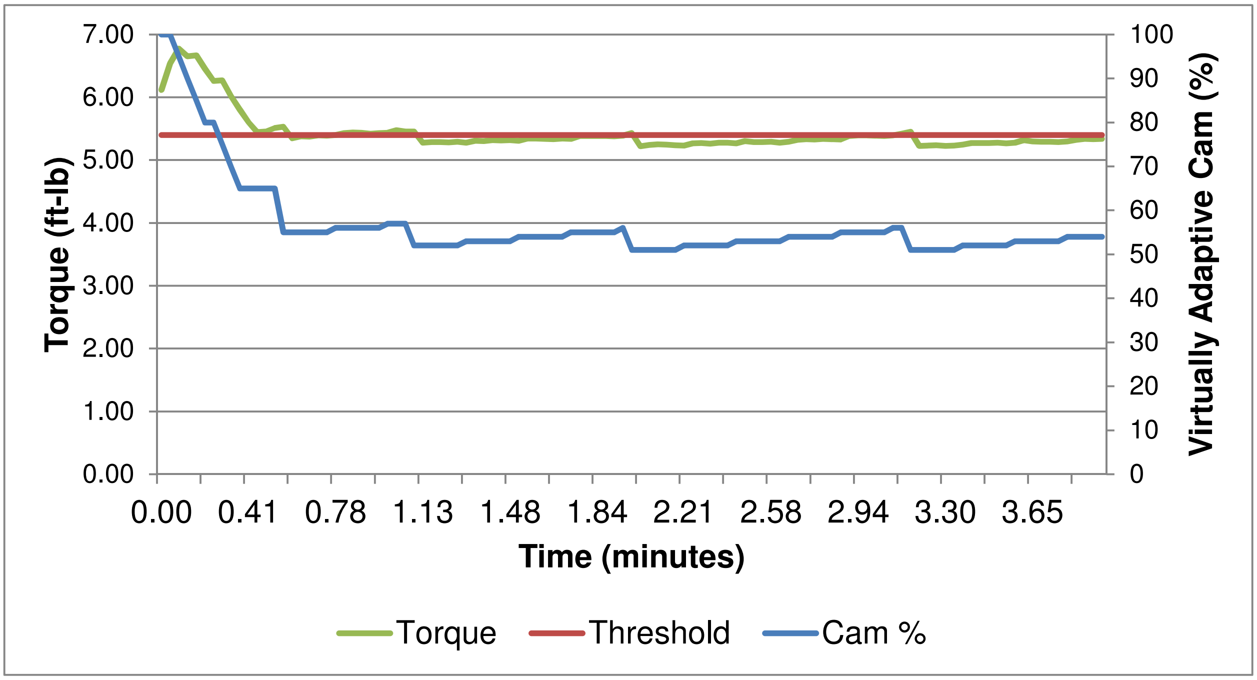

Figure 16a: A record of the cam gain and peak recorded torque for each gait cycle with the rubber bungee modeling and the EBRGT in the closed loop mode.

Figure 16b shows a small sampling of the data from figure 16a, annotated with the adjustment points for the cam increasing or decreasing. This plot shows the virtual cam profile changing and the corresponding change in the peak torque.

Figure 17 shows the same torque values as figure 16a with the threshold value subtracted. This provides a way to view the error in the amount of torque generated. It also demonstrates more clearly the adaptive way in which the closed-loop control works by reducing in increments of 5% and slowly increasing in increment of 1% until the threshold is reached.

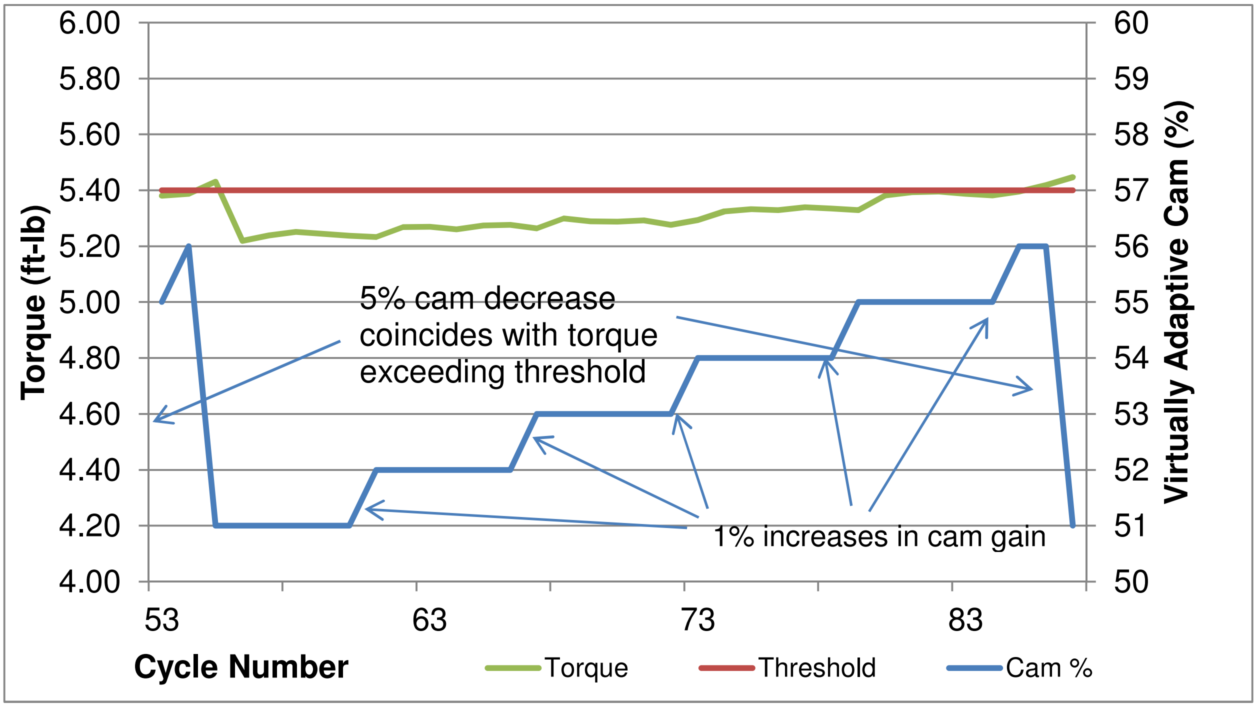

Figure 16b: A sample of the entire data set in figure 16a with magnified x-axis and y-axis for clarity. The peak torque for each cycle increases each time the cam gain is increased. When the peak torque exceeds threshold, the cam gain decreases by 5% and starts to increase bv 1% everv 5 cvcles.

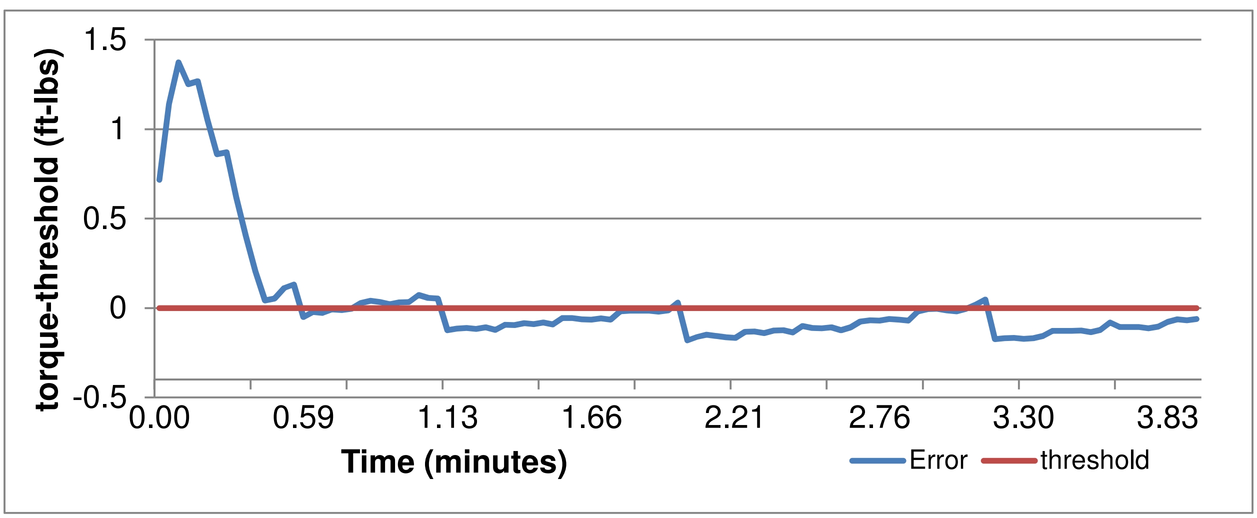

Figure 17: The blue line represents the peak torque value for each cycle with the torque threshold subtracted from it.

Discussion

Preliminary data analysis was conducted during the course of data collection in each phase of this project. Error were found and addressed making the next round of data collection more accurate. This was the case with the preliminary healthy, normal data prior to standardizing the foot placement. Data collected with random foot placement cannot be compared to subsequent trials because the force profile is shifted and magnified in different phases of the gait cycle depending on the effective lever arm (see appendix F). The foot placement standardization protocol resolved the issue with unpredictable scaling and zero offset.

The data from the patient with spasticity was compared to the healthy, normal subject population to find the difference in the force profile. The gait phase of interest in this project is the swing phase where this patient exhibited the spastic response during his initial testing. This spastic response presented itself as the heel of his foot rising off the footplate into a plantarflexed position in reaction to the toe of the footplate rising to prepare for initial contact. Figure 12 shows a difference in the last 0.3 seconds of the gait cycle for the 100% footplate cam force profile compared to the 60% and 20% cam profiles. These last 0.3 seconds of data represent the time just before initial contact when the foot should be in the most dorsiflexed position. The toe-down force was much higher in these last 0.3 seconds than in the 60% and 20% cam profile data samples.

Figure 18a shows the swing phase of simulated data samples at varying attenuation levels to demonstrate the changes in force as a result of the automatic closed-loop attenuation of the footplate.

In the last 0.3 seconds of the gait cycle, the load curves diverge based on the level of attenuation. The cam profiles with less dorsiflexion produce less torque. This is the desired and anticipated reaction to a reduced footplate profile when demonstrated on a patient with spasticity. This is more easily seen in Figure 18b, where only the last 0.4 seconds of the gait cycle are plotted.

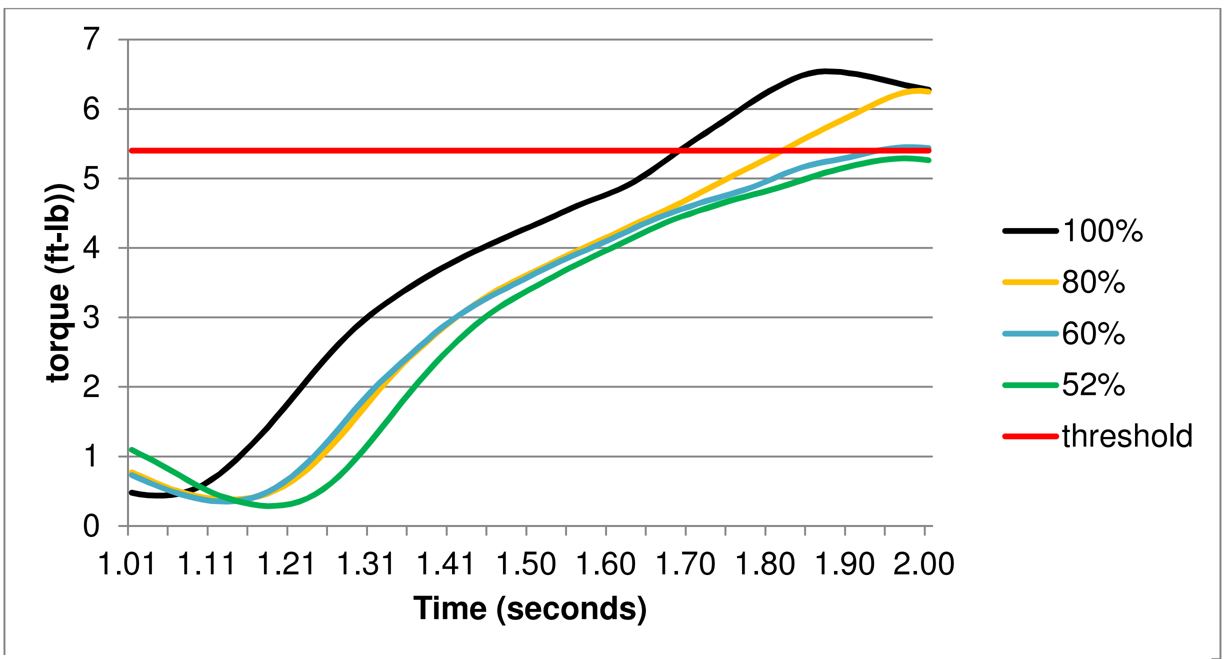

Figure 18a: Simulated force curves using the rubber bungee arrangement for the swing phase of gait only. The 100% and 80% cam profile curve go well above the threshold, eliciting a change signal from the mbed™.

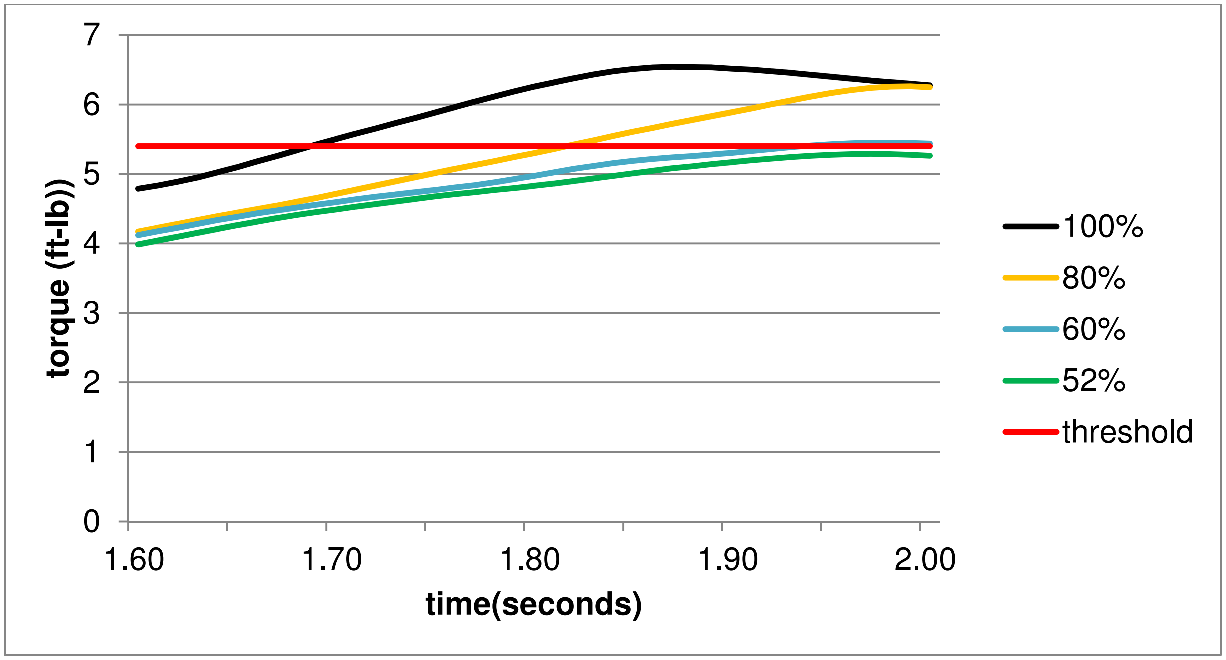

Figure 18b: The diverging force curves can be more easily seen in the last 0.4 seconds of the gait cycle. The 100% and 80% curves are well below the threshold line, causing a change signal to reduce the footplate motion and consequently, the force level for the next cycle.

The closed-loop control was tested with rubber bungee bands to simulate a high toe-down force during the gait cycle immediately prior to initial contact. The mbed™ interprets this load signal and provides an output to the Parker motor controller to indicate if the load is too high, too low, or within an acceptable range. The Parker software was originally written with a 1 second delay in the closed-loop routine to prevent a malfunction with the print screen command. This caused the mbed™ communications to be missed by the Parker motor controller in some cases because the mbed™ communications were only active for the time in the gait cycle when the torque was above the threshold level.

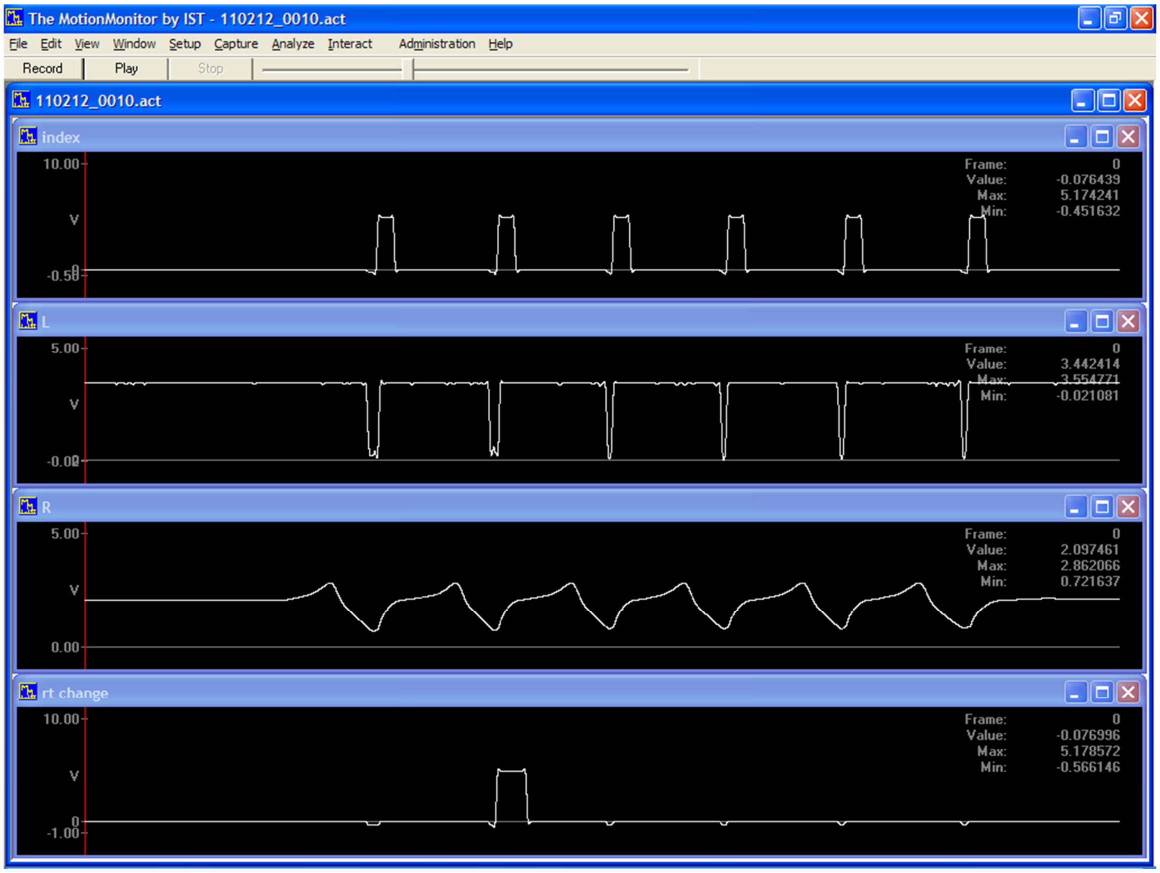

Figure 19 screen output from ***The MotionMonitor™*** during simulated spastic trial

Figure 19 shows a data series where the mbed™ output was not observed by the Parker controller. In order from top to bottom, the data streams are:

1. flywheel index (0V normal, goes to 5V at initial contact on right side and stays at 5V for 0.5 second)

2. mbed™ change pattern (3.3 V normal, drops to 0 when toe-down force exceeds threshold)

3. right side load cell signal (0 to +5V analog signal, no load is +3V, more positive values indicate heel load, values closer to 0 indicate toe load)

4. Parker change signal (0V normal, goes to 5V for 0.5 second when increasing or decreasing)

Figure 19 shows the toe-down force threshold was exceeded 6 times in this 30 second data sample which consisted of 6 gait cycles. The Parker motor controller detected the mbed™ change signal only on the second occasion, despite the load signal exceeding the threshold each cycle. This was addressed by changing the Parker software to eliminate the print status commands which created the communication error without the delay. The print status command was moved to a separate if-then statement within the closed-loop routine to print only when the flywheel encoder was past the initial contact event in the gait cycle.

In addition to the footplate 5% movement reduction algorithm, the Parker motor controller increases the footplate movement by 1% when the flywheel cycles 5 consecutive times without an excessive toe-down force causing a reduction. Figures 16 and 17 demonstrate the torque adaptation and reduction algorithm working properly. The cam profile started at 100% with the rubber bungees in place. The cam profile was decreased to as low as 51% (shown by the blue line in figure 16 and plotted on the secondary axis on the right side). The cam profile increased by 1% each time the flywheel completed five consecutive revolutions without a change. The torque developed (green line in figure 16) also increased with the increase in footplate movement. The cam profile decreased by 5% when the torque exceeded the threshold. The decrease in cam profile (and footplate movement) caused the torque to decrease.

The end result is a combination of hardware and software which works together to detect excessive toe-down force during the swing gait phase and implement a footplate movement reduction algorithm which reduces the spastic response while maintaining a therapeutic approach.

Conclusion

The EBRGT was designed to simulate level-surface walking while maintaining proper joint kinematics lacking with the Lokomat™. The nature of an elliptical machine limits the swing-stance relationship to a fixed 50-50 split, unlike the natural 60-40 stance-swing characteristic of normal gait.

The physical design of the footplates, gearboxes, and pushrods constrain the footplate movement. The standard cam profile is similar in the angular displacement during dorsiflexion and plantarflexion to level surface walking and the EBRGT is capable of accommodating that motion profile. A rehabilitation protocol requiring an exaggerated movement of the footplates on one or both sides would be limited by the supporting ski structure.

The latest changes in this project consisting of the closed-loop control hardware and software were not validated with a patient exhibiting spasticity. Although the data collected from the spastic trial can be interpreted and acted upon by the algorithm, the final outcome is unknown without this final step in the validation process.

Future work would consist of a trial using a patient with a history of spasticity to serve as validation for this project to prove that the spasticity demonstrated in the early part of the session would be reduced by a reduction in the footplate motion. Assuming that is successful, additional patients with spasticity would be needed to validate that the EBRGT is not tuned for the individual. The standing perturbation experiment also requires further development. Data needs to be collected on a patient with spasticity using the standing perturbation test. Data on healthy normal subjects would also be required in conjunction with the results from a physical therapist's interpretation of the Modified Ashworth test. Finally, if there is a correlation between the results of the Modified Ashworth test and the standing perturbation test, the virtual cam setting would start at a level close to the individual patient's spasticity threshold.

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List of References

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Appendices

Appendix A

Electrical Pin Chart

The mbed™ box houses the 37-pin connector, mbed™, LM 7805 voltage regulator, capacitors, and switches. The wires from the 37-pin connector attach to screw terminals inside the box. The terminals attach to a wire under the circuit board which connects to the relevant component. These charts show the name of the signal for each screw terminal and what it connects to inside the box. The far right column shows the mbed™ pin which feeds or is fed by the signal.

Screw Terminals			
L1	10+	LM7805	mbed™2
L2	Green button signal	transistor 7	mbed™ 7
L3			mbed™ 9
L4			mbed™ 10
L5			mbed™ 11
L6	right side selector switch	10k resistor	mbed™ 13
L7			
L8	Left load		mbed™ 15
L9	Right load		mbed™ 16
L10	left side selector switch	10k resistor	mbed™ 18
Screw Terminals			
R1	ground	ground bus	mbed™ 1
R2			mbed™30
R3			mbed™ 29
R4			mbed™ 28
R5			mbed™ 27
R6			
R7	Parker R7	transistor 24	mbed™ 24
R8	Parker R1	transistor 25	mbed™ 25
R9	Parker L7	transistor 22	mbed™ 22
R10	Parker L1	transistor 21	mbed™ 21

Appendix B

Electrical Pin Charts

The 37-pin connector on the mbed™ box has 9 pins in place. This chart details the location of each pin, the wire color, the signal name, there screw terminal to which it attaches inside the box, the wiring run destination, and the mbed™ pin affected by that signal.

	Box Connector 37-pin round				
Connector position	wire color	signal	screw terminal	transistor	mbed™
1	blue	Parker L1	R10	transistor 21	mbed™ 21
2	blue/white	Parker R7	R7	transistor 24	mbed™ 24
3	blue/black	Parker R1	R8	transistor 25	mbed™ 25
4	blue/red	Parker L7	R9	transistor 22	mbed™ 22
5					
6	orange/black	ground	R1		mbed™ 1
7	orange/red	+10V	L1		mbed™ 2
8					
9					
10	green	right load	L9		mbed™ 16
11					
12					
13					
14					
15					
16					
17					
18					
19	red	left load	L8		mbed™ 15
20					
21					
22					
23					
24					
25	white/red	green button	L2	transistor 7	mbed™ 7

Pins 26-37 not used

Appendix C

Parker™ Motor Controller Electrical Connector

The footplate motors are controlled by Parker™ motor controllers with input from the flywheel-mounted encoder, the mbed™ microprocessor, and switches to designate end of travel and emergency stop.

Signal	Pin	Previous Function	New Function
Input 0+ Input 0-	1 14	Red button	mbed™ input
Input 1+ Input 1-	2 15	End of travel -	End of travel - aft
Input 2+ Input 2-	3 16	End of travel -	End of travel - forward
High-Speed Input High-Speed Input	4 17	Encoder	Encoder
High-Speed Input High-Speed Input	5 18	Encoder	Encoder
High-Speed Input High-Speed Input	6 19	Encoder	Encoder
Input 3+ Input 3-	7 20	Green button	mbed™ input
Reserved (future) Reserved	8 21		
Output 32+ Output 32-	9 22	Flywheel	Flywheel encoder index
Output 33+ Output 33-	10 23		Change index signal
Output 34+ Output 34-	11 24		
Output 35+ Output 35-	12 25		
Not used	13		

Appendix D

mbed™ to Parker™ Motor Controller Bit Pattern

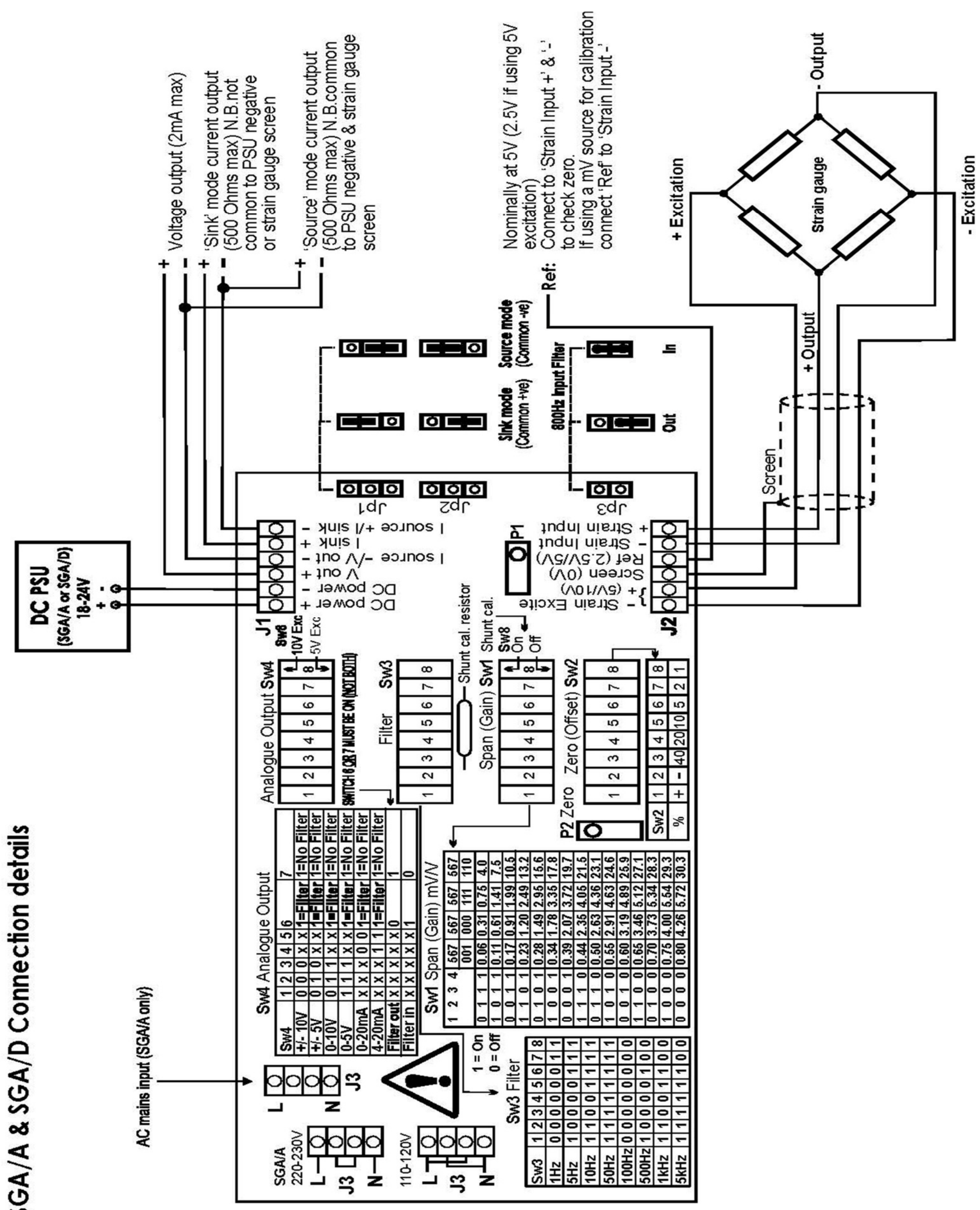
The mbed™ microprocessor communicates with the left and right side Parker™ motor controllers through a series of bit patterns. The combination of bits signifies the action needed by the motor controller when the mbed™ senses a toe force exceeding threshold or when the start button on the EBRGT is pressed.

	Input 0	Input 3
No changes	0	0
Load cell too low	0	1
Load cell too high	1	0
Green button pushed	1	1

Appendix E

SGA Diagram

The signal amplifier provides conditioned power for the load cells and conditions the load cell output signal for zero offset and sensitivity. Switches and potentiometers are set based on the desired output.



Appendix F
Adjustment of Load Cell Signal Amplifiers

A schematic of the amplifier is found in Appendix E. The first step was properly zeroing the output. This was accomplished by collecting data on the device with ***The MotionMonitor***® with no load or motion on the EBRGT. After each 30 second data collection period, the average value was calculated and an adjustment was made inside the amplifier to switch SW2 for gross adjustments and potentiometer P2 for fine adjustments. Switch SW2 has a total of 7 DIP switches which can be selected on or off to specify a positive or negative offset and a percentage of offset. Potentiometer P2 has a small screw which is turned to make small adjustments.

After zeroing the amplifiers, the sensitivity was established to determine the relationship of force placed on the footplate of the EBRGT to the voltage produced by the load cell. This was accomplished using known weights placed a known distance from the pivot point of the footplate.

In order to validate the data collection method and torque measurements, a comparison between the measured static load and the calculated load was performed. The force in the pushrod varies throughout the gait cycle with a constant weight at a fixed distance from the pivot point. This is a result of the changing angle of the footplate as it articulates to simulate level surface walking. As the footplate moves to a position closer to vertical, the, the effective moment arm is decreased. This force was measured at each 5 degree increment of flywheel rotation with a known weight placed on the bar of a bar clamp affixed to the footplate. With the perpendicular distance of the bar clamp and the angles of the footplate and pushrod relative to horizontal known, the theoretical force in the pushrod is calculated. This is done through the following calculations.

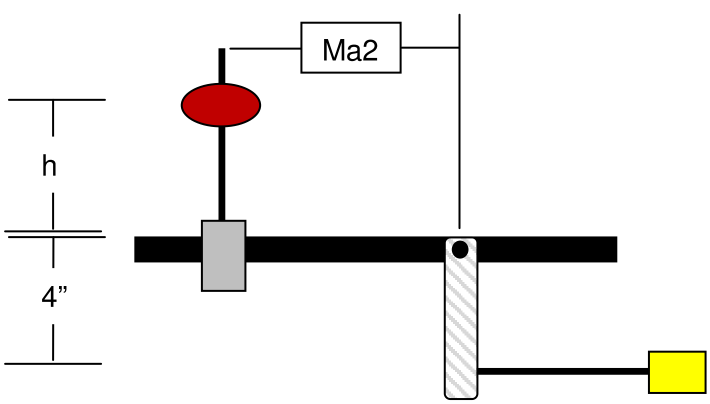
Ma1 =horizontal distance from footplate pivot point to attachment point of bar clamp

Ma2 = horizontal distance of center of mass to edge of ma1

Ma3 = sum of ma1 + ma2

FP= angle between the footplate and horizontal

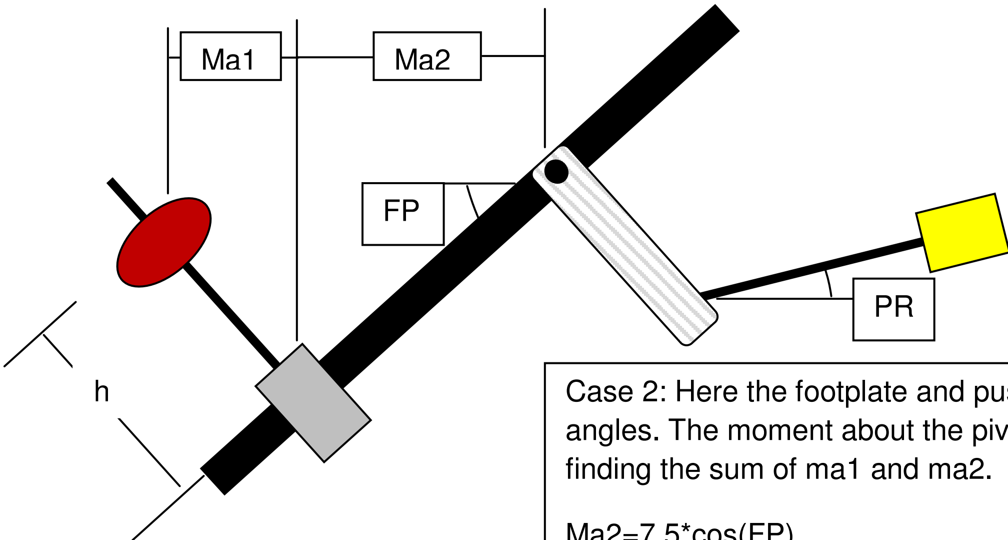
PR = angle between the pushrod and horizontal



Case 1: In this case, the footplate and pushrod are parallel to each other and to the ground. The moment about the footplate pivot point is equal to the weight multiplied by the horizontaldistance from the pivot point to the bar clamp (ma2=7.5 inches). The force in the pushrod is equal to the moment divided by the length of the crank arm (4 inches).

M=W*Ma2

F(pr)=M/4



Case 2: Here the footplate and pushrod are at different angles. The moment about the pivot point is calculated by finding the sum of ma1 and ma2.

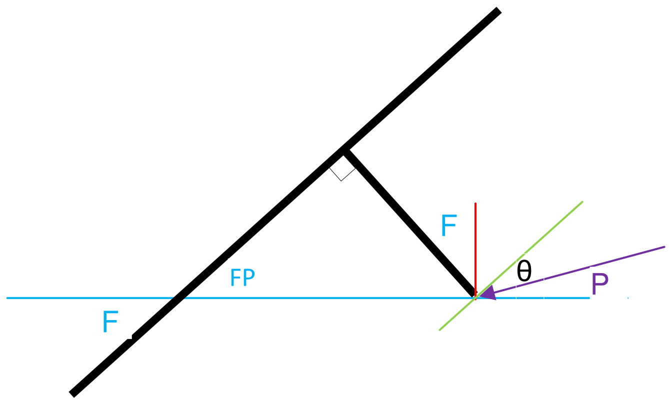
Ma2=7.5*cos(FP)

Ma1=h*sin(FP)

Ma3=Ma1+Ma2

M=W*MA3

The force in the pushrod is a function of the interacting angles between the footplate and the pushrod. This is further described in the next drawing.



The footplate and crankarm are shown here by the bold, black lines. The footplate is drawn at an angle of 42 degrees. The pushrod force (depicted by the purple arrow) is drawn at 14.5 degrees. This represents the position of these components at toe off.

The force in the pushrod is a composite of two component forces. The first force is acting at a right angle to the crank arm, or parallel to the footplate. This force must be equal to the moment about the pivot (M from the previous page) divided by 4 inches (the length of the crank arm). The other force is perpendicular to the first force and acts parallel to the crank arm.

The angle between the pushrod and the first composite force is known as θ. We can find θ with following equation:

$$\begin{aligned} 90 + \theta + \mathrm{PR} &= 90 + \mathrm{FP} \\ \theta + \mathrm{PR} &= \mathrm{FP} \\ \theta &= \mathrm{FP} - \mathrm{PR} \end{aligned}$$

Now that we know θ and the first component force, we can find the total force in the pushrod (F_{pr}) with trigonometry. This is done by the following equation:

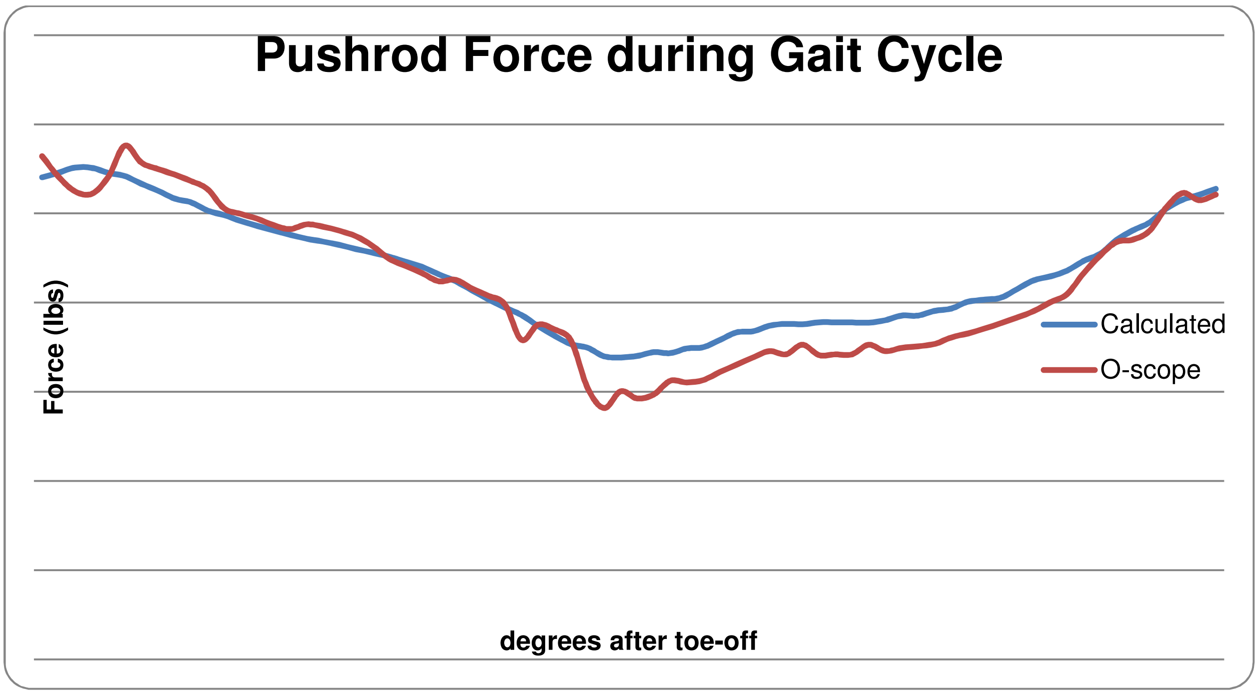
$$\frac{M}{4} = F_{pr} * \cos(\theta)$$

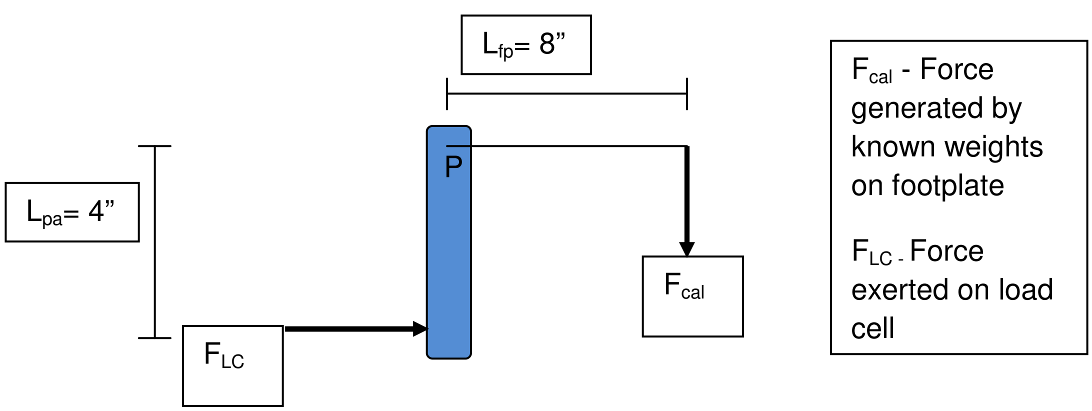
Solving for F_{pr}, we find :

$$F_{pr} = \frac{M}{4 * \cos(\theta)}$$

Calculating the force in the pushrod with a known mass located at a known distance from the pivot point is valuable. Because the angles between the pushrod and the footplate are known at 72 points during the gait cycle (5 degree increments of flywheel rotation), the actual moments can be found when a force is measured in the load cell with a dynamic patient on board the device.

This technique of calculating pushrod force was validated by comparing the expected pushrod force to the voltage readings taken with an oscilloscope directly from the signal amplifier. The following plot shows the two lines.





Equation 1: $\sum M_P = 0$

When the footplate is not accelerating, the moments about the pivot point of the footplate P must be equal to zero. Moment is found by multiplying the length of the moment arm by the force exerted at the end of that moment arm. In this case, the lengths of the two moment arms (L_{fp} and L_{pa}) are known and the magnitude of F_{cal} is also known. Only 2 forces produce a moment about point P, F_{cal} and F_{LC}. These forces, multiplied by their respective moment arms and substituted into Equation 1 are:

Equation 2: $0 = (F_{LC} * L_{pa})+(F_{cal} * L_{fp})$

Assuming counter-clockwise motion is positive, Equation 2 becomes:

$0 = (F_{LC} * 4 \text{ inch}) – (65Lb_f * 8 \text{ inch})$

$F_{LC} = 520 \text{ in-}Lb_f / 4 \text{ in}$

$F_{LC} = 130\ Lb_f$

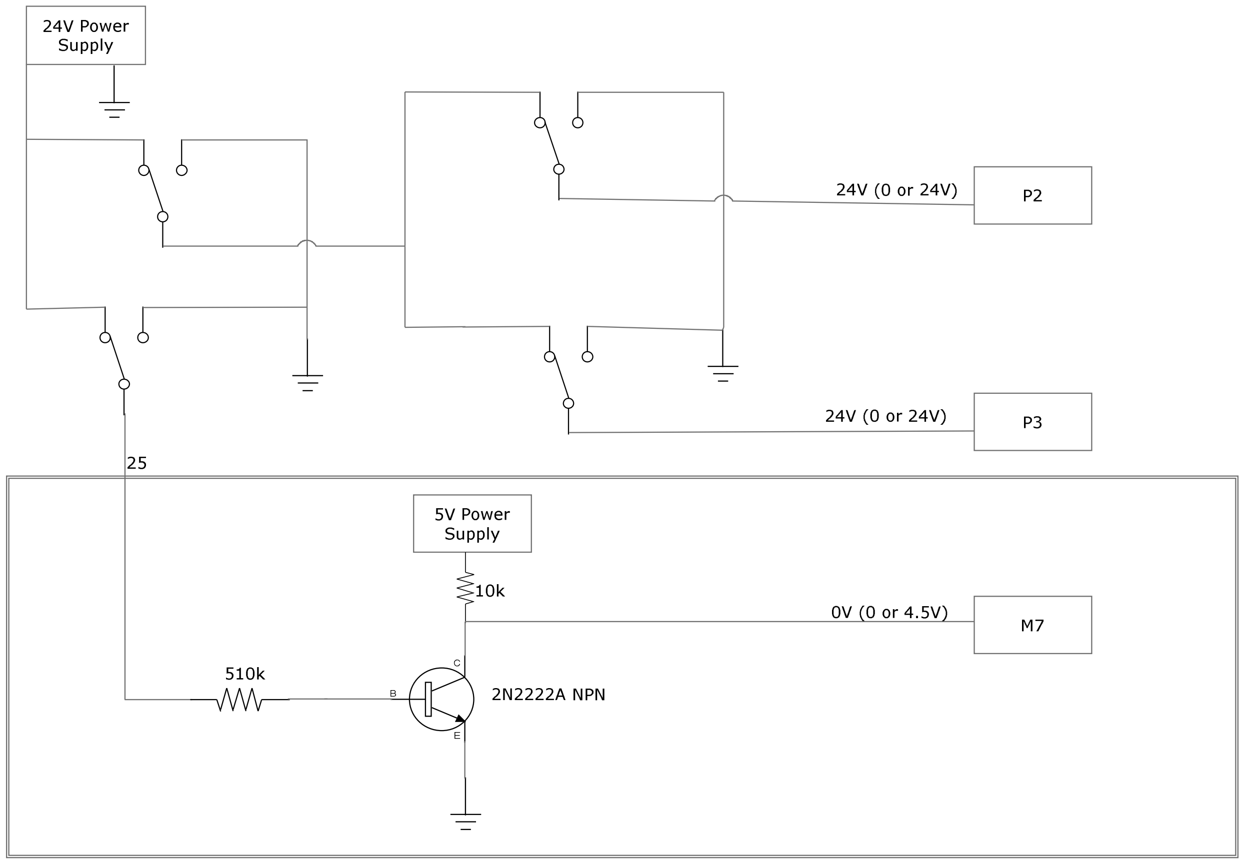
These calculations show that the force exerted on the load cell is equal to 130 Lb_f for every 65 Lb_f exerted on the footplate at a distance of 8 inches from the pivot point. This multiplication of force is a result of the longer moment arm to the center of loading on the footplate than from the pivot point to the pushrod. The point in the gait cycle when maximum force is attained is at the instant of toe-off. The gain was set to 3.46 using a series of DIP switches in the amplifier. This gain value combined with an output scale of 0 to 5 volts and a zero load value of 3V permits a load of up to 351 LB_f to be exerted on the tip of the footplate before the output is saturated. The maximum anticipated test subject weight is 200 Lb_f. This entire load will not be exerted on the tip of the footplate, but rather distributed over the surface of the footplate. The magnitude of the torque is estimated by using the weight and foot size of each subject. With consistent foot placement, the maximum allowable output will be below the maximum range of the amplifier.

Appendix G

Wiring Diagrams

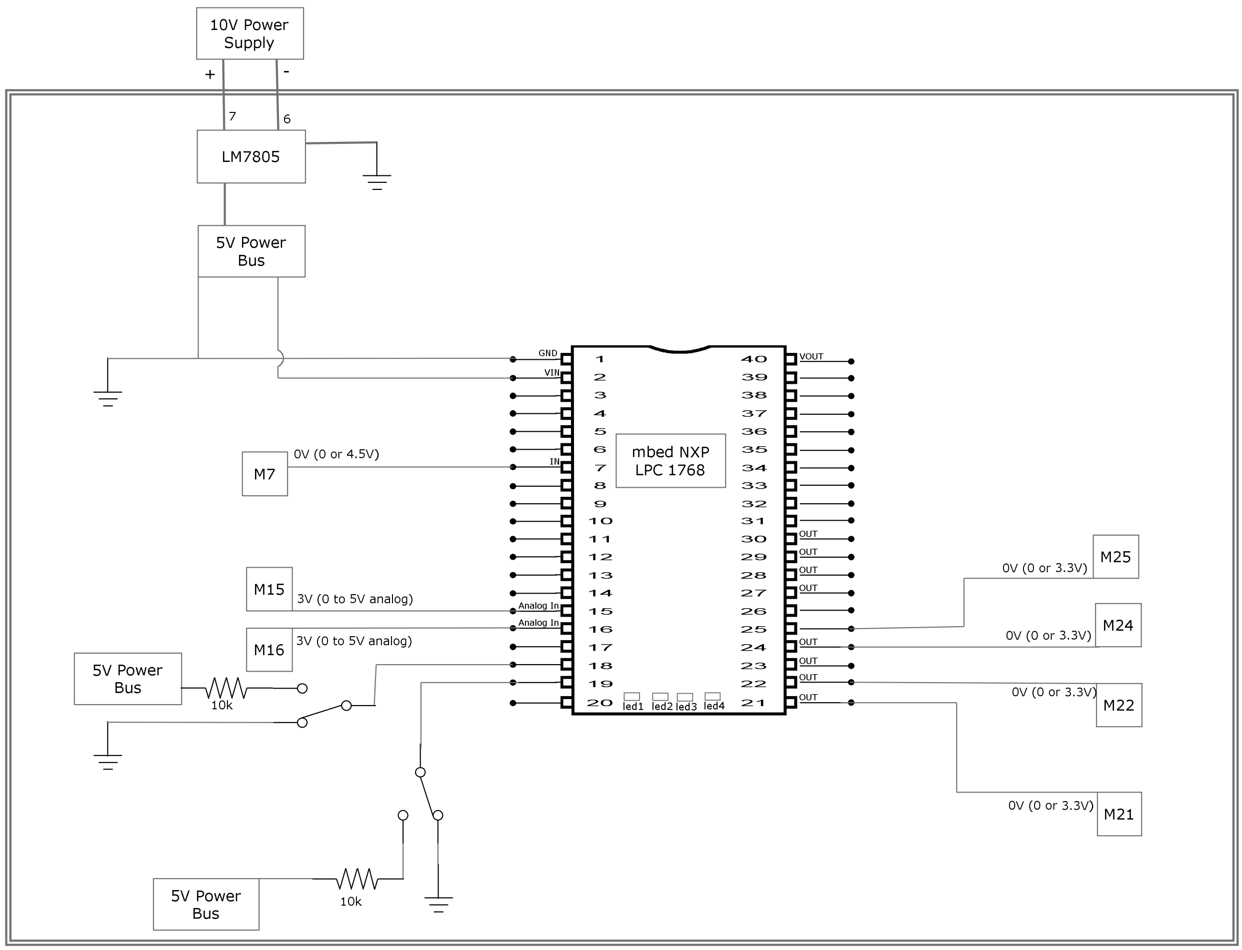
The red emergency stop button is a 24V normally-closed push button switch wired in series with the footplate end-of-travel switches. These switches provide an input to the Parker controllers, setting a bit which must be set before movement can start. This provides a fail-safe arrangement to prevent the EBRGT from operating if one of the wires was broken, just as if the switch was opened.

The green button output runs into the mbed™ box and provides the base signal to a NPN transistor which switches a signal going to the mbed™. The transistor is required because the mbed™ is only 5V tolerant.



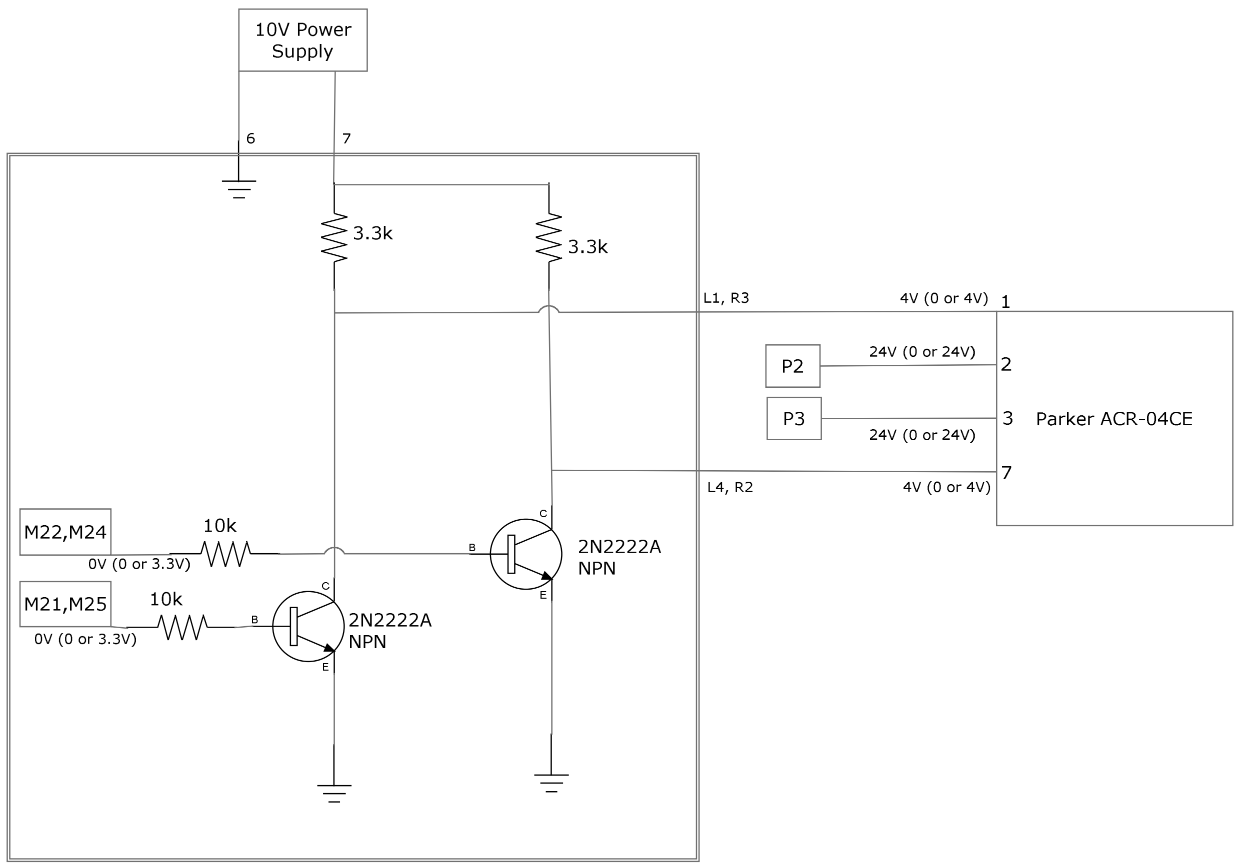
The double lined box indicates the components are located inside the enclosed box. The number outside the box indicates the pin location in the 37-pin connector.

The 10V DC power supply provides power to the mbed™ box which is reduced to 5V with a LM7805 voltage regulator. The output from the LM7805 provides 5V to the circuit board power bus to power the mbed™ and provide the voltage for the signals switched by the transistors.



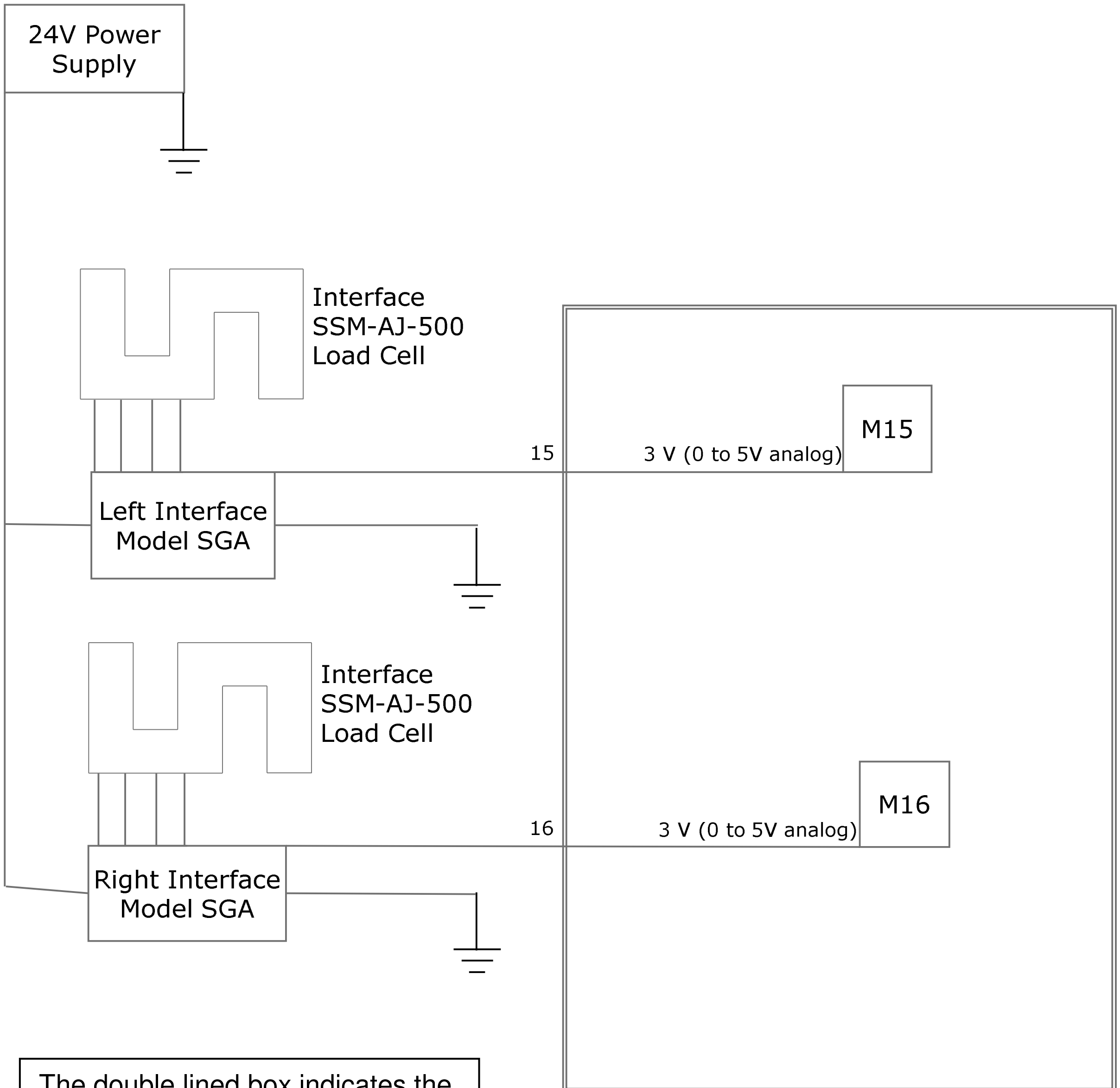
The double lined box indicates the components are located inside the enclosed box. The number outside the box indicates the pin location in the 37-pin connector.

The mbed™ output is 3.3V, which is not enough for the Parker controller to detect reliably. The 3.3V mbed™ output is used as a transistor base input to switch the higher 10V signal from the DC power supply.



The double lined box indicates the components are located inside the enclosed box. The number outside the box indicates the pin location in the 37-pin connector.

The load cells receive power through the amplifiers which also condition the load cell outputs. The amplifiers are set to a 0 to +5V output which can go into the mbed™ without further conditioning.



The double lined box indicates the components are located inside the enclosed box. The number outside the box indicates the pin location in the 37-pin connector.

Appendix H

mbed™ Code

```
#include "mbed.h"
#define low (0.9 / 3.3)
#define high (3.3 / 3.3)

DigitalIn green(p7);
DigitalIn leftselect(p13);
DigitalIn rightselect(p18);
AnalogIn loadleft(p15);
AnalogIn loadright(p16);
DigitalOut left1(p21);
DigitalOut left2(p22);
DigitalOut right1(p25);
DigitalOut right2(p24);
DigitalOut led1(LED1);
DigitalOut led2(LED2);
DigitalOut led3(LED3);
DigitalOut led4(LED4);

int main() {
   led1=led2=led3=led4=0;
   left1=left2=right1=right2=0;
   led1=led2=led3=led4=1;
   wait(0.5);
   led1=led2=led3=led4=0;

   while(1) {
      while (green==1){
         left1=left2=right1=right2=led1=led2=led3=led4=1;
         }
      if (leftselect==1 and rightselect==0){
         left2 = led2 = (loadleft < low);
         left1 = led1 = (loadleft > high);
         }
      else if (leftselect==0 and rightselect==1){
         right1 = led3 = (loadright > high);
         right2 = led4 = (loadright < low);
```

```
        }
    else if (leftselect==1 and rightselect==1){
        right1 = led3 = (loadright > high);
        right2 = led4 = (loadright > low);
        left1 = led1 = (loadleft > high);
        left2 = led2 = (loadleft < low);
        }
  }
}
```

Appendix I

Left Parker Code

Program 0:

```
PROGRAM
 PBOOT
 DETACH
 ATTACH MASTER0
 ATTACH SLAVE0 AXIS0 "L"
 PPU L8000
 AXIS0 EXC(5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation,
less than .1 for footplate)
 SET BIT8469: REM enable EXC response
 TLM L7 : REM set torque limit to +- 2 V
 REM Axis Gains values
 AXIS0 PGAIN 0.008
 AXIS0 IGAIN 0
 AXIS0 ILIMIT 0
 AXIS0 IDELAY 0
 AXIS0 DGAIN 0.0001
 AXIS0 DWIDTH 0
 AXIS0 FFVEL 0
 AXIS0 FFACC 0
 AXIS0 TLM 10
 AXIS0 FBVEL 0
 REM Axis Limits
 AXIS0 HLBIT 1
 AXIS0 HLDEC 100
 HLIM L3
 'SET BIT16144
 SET BIT16145
 CLR BIT16146
 SET BIT16148
```

```
SET BIT16149
AXIS0 SLM(20,-20)
AXIS0 SLDEC 100
SLIM L3
SET BIT16150
SET BIT16151
REM MOTION PROFILE
REM the desired master acceleration
ACC 100
REM the desired master deceleration ramp
DEC 100
REM the desired master stop ramp (deceleration at end of move)
STP 250
REM the desired master velocity
VEL 10
REM the desired acceleration versus time profile.
JRK 0
JOG VEL L1
JOG ACC L25
JOG DEC L25
REM BEGIN HOMING SEQUENCE
CLR BIT136
clr bit137
clr bit0
clr bit1
clr bit2
clr bit3
clr bit1920
clr bit1921
PRINT "Press green button To start homing, press red button To stop at any time"
'
_MAIN1
IF (NOT BIT1 OR NOT BIT2) THEN SET BIT1920 REM RED BUTTON OR ANY EOT
SWITCH
IF (BIT1920) THEN SET BIT8467
IF (BIT 1920) THEN CLR BIT136
IF (NOT BIT0 AND NOT BIT3) THEN SET BIT1921  REM 0001 GREEN BUTTON
IF (BIT1921 AND NOT BIT136) THEN GOTO HOMING
IF (BIT136 AND NOT BIT137) THEN GOTO CAMMING: REM IF BIT 136 (USER
DEFINED = HOMING COMPLETE) IS SET, START CAMMING
```

```
IF (BIT8467) THEN CLR BIT136 REM IF A KILL ALL MOTION FLAG IS SET (8467)
THEN CLEAR BIT 136 AND TURN THE CAM OFF
 IF (BIT1921 AND BIT136 AND BIT137) THEN GOTO CHANGE
 GOTO MAIN1
 '
 _HOMING
 PRINT "BEGIN HOMING"
 BIT798= 0 : REM CHECK JOG LIMITS WHEN JOGGING FWD/REV
 JOG VEL L1 : REM SET JOG VELOCITY TO 1 REV/S
 DRIVE ON L
 CLR 8467
 JOG FWD L
 PRINT " JOGGING IN POSITIVE DIRECTION "
 INH -792 : REM WAIT UNTIL MOTION HAS STOPPED
 PRINT " POSITIVE LIMIT SWITCH FOUND "
 CLR 8467 : REM CLEAR KILL ALL MOVES FLAG THAT IS SET WHEN A LIMIT IS
REACHED
 JOG REV L
 PRINT " JOGGING IN NEGATIVE DIRECTION "
 INH -792
 PRINT " NEGATIVE LIMIT SWITCH FOUND "
 PRINT " ZERO POSITION AT NEG SWITCH "
 CLR 8467
 JOG INC L6.08334
 PRINT " MOVING TO OFFSET POSITION "
 INH -792
 PRINT " AT OFFSET POSITION"
 JOG RES L0
 RES L0
 PRINT " ZERO POSITION REGISTER AT HOME POSITION "
 SET BIT136
 CLR BIT137
 CLR BIT1921
 clr bit1936
 GOTO MAIN1
 '
 _CAMMING
 AXIS0 EXC(5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation,
less than .1 for footplate)
 DIM LA(4) : REM Dimension 4 long arrays
```

```
DWL 0.5
DIM LA0(69) : REM LA0 has 69 elements
DWL 0.5
DIM LA1(69)
DWL 0.5
DIM LA2(69)
DWL 0.5
DIM LA3(69)
DWL 0.5

LA0(0) = -1388
LA0(1) = -1940
LA0(2) = -2464
LA0(3) = -2969
LA0(4) = -3451
LA0(5) = -3894
LA0(6) = -4299
LA0(7) = -4659
LA0(8) = -4970
LA0(9) = -5237
LA0(10) = -5466
LA0(11) = -5645
LA0(12) = -5790
LA0(13) = -5815
LA0(14) = -5679
LA0(15) = -5404
LA0(16) = -5044
LA0(17) = -4583
LA0(18) = -4103
LA0(19) = -3588
LA0(20) = -3054
LA0(21) = -2521
LA0(22) = -2000
LA0(23) = -1490
LA0(24) = -1077
LA0(25) = -791
LA0(26) = -595
LA0(27) = -444
LA0(28) = -341
LA0(29) = -218
```

LA0(30) = -98
LA0(31) = 24
LA0(32) = 138
LA0(33) = 239
LA0(34) = 340
LA0(35) = 444
LA0(36) = 556
LA0(37) = 666
LA0(38) = 803
LA0(39) = 939
LA0(40) = 1077
LA0(41) = 1241
LA0(42) = 1425
LA0(43) = 1693
LA0(44) = 2005
LA0(45) = 2336
LA0(46) = 2672
LA0(47) = 3007
LA0(48) = 3356
LA0(49) = 3691
LA0(50) = 4028
LA0(51) = 4364
LA0(52) = 4611
LA0(53) = 4767
LA0(54) = 4782
LA0(55) = 4706
LA0(56) = 4553
LA0(57) = 4336
LA0(58) = 4060
LA0(59) = 3726
LA0(60) = 3330
LA0(61) = 2848
LA0(62) = 2272
LA0(63) = 1669
LA0(64) = 1058
LA0(65) = 428
LA0(66) = -201
LA0(67) = -804
LA0(68) = -1388

LA2(0) = -1388
LA2(1) = -1940
LA2(2) = -2464
LA2(3) = -2969
LA2(4) = -3451
LA2(5) = -3894
LA2(6) = -4299
LA2(7) = -4659
LA2(8) = -4970
LA2(9) = -5237
LA2(10) = -5466
LA2(11) = -5645
LA2(12) = -5790
LA2(13) = -5815
LA2(14) = -5679
LA2(15) = -5404
LA2(16) = -5044
LA2(17) = -4583
LA2(18) = -4103
LA2(19) = -3588
LA2(20) = -3054
LA2(21) = -2521
LA2(22) = -2000
LA2(23) = -1490
LA2(24) = -1077
LA2(25) = -791
LA2(26) = -595
LA2(27) = -444
LA2(28) = -341
LA2(29) = -218
LA2(30) = -98
LA2(31) = 24
LA2(32) = 138
LA2(33) = 239
LA2(34) = 340
LA2(35) = 444
LA2(36) = 556
LA2(37) = 666
LA2(38) = 803
LA2(39) = 939

```
LA2(40) = 1077
LA2(41) = 1241
LA2(42) = 1425
LA2(43) = 1693
LA2(44) = 2005
LA2(45) = 2336
LA2(46) = 2672
LA2(47) = 3007
LA2(48) = 3356
LA2(49) = 3691
LA2(50) = 4028
LA2(51) = 4364
LA2(52) = 4611
LA2(53) = 4767
LA2(54) = 4782
LA2(55) = 4706
LA2(56) = 4553
LA2(57) = 4336
LA2(58) = 4060
LA2(59) = 3726
LA2(60) = 3330
LA2(61) = 2848
LA2(62) = 2272
LA2(63) = 1669
LA2(64) = 1058
LA2(65) = 428
LA2(66) = -201
LA2(67) = -804
LA2(68) = -1388

DIM LV(5)
 LV0=0
 LV3=100
 LV4=0

 PRINT "SLOWLY MOVE FLYWHEEL FORWARD UNTIL THE FOOTPLATES BEGIN
MOVING"
 INTCAP AXIS0 10 : REM arms capture of axis0 position when HS inp 4 rises
(designated by 10)
 INH 777 : REM wait for flag 777 to be set (flag 777 is set when inp 4 trips intcap)
```

```
ENC1 RES -2912 : REM resets encoder to -3700 so it is zero at BDC on the right.
set bit 138
PRINT "Index detected. Encoder reset."
CAM DIM L1 : REM Define 1 cam segments
CAM SEG L(0,10000,LA0) : REM Define cam segment range and source
CAM SCALE L(1/1000) : REM scales cam output back to revolutions
CAM SRC L1 : REM Define cam source as ENC1
CAM SRC RES : REM resets the cam source to 0
SET BIT137
'
_loop
IF  (P6160 = 0) THEN CAM ON L
IF  (BIT790) THEN GOTO MAIN1: REM Start camming
GOTO loop
'
_CHANGE
PRINT "Change Left Footplate Pattern"
INH 3
DIM DV(2)
DIM $V(2,6)
PRINT "Which Program?"
PRINT "1 Normal Camming"
PRINT "2 Attenuated Camming"
PRINT "3 Auto Attenuation"
PRINT "4 Standing Pertubation"
PRINT "5 incremented/decremented camming"
PRINT "6 closed loop camming"
'
INPUT; $V0
PRINT $V0
LV4 = VAL($V0)
PRINT "LV4=";LV4
'
IF (LV4=1) THEN PRINT "1 Normal Camming, BACK TO MAIN PROGRAM"
IF (LV4=2) THEN PRINT "2 Attenuated Camming"
IF (LV4=3) THEN PRINT "3 Auto Attenuation"
IF (LV4=4) THEN PRINT "4 Standing Pertubation"
IF (LV4=5) THEN PRINT "5 incremented/decremented camming"
IF (LV4=6) THEN PRINT "6 closed loop camming"
```

```
 IF (LV4=1) THEN GOTO MAIN1
 IF (LV4=2) then goto ATT
 IF (LV4=3) then goto AUTO
 IF (LV4=4) then goto SP
 IF (LV4=5) then goto INCREMENT
 IF (LV4=6) then goto CLOSED

 PRINT "ERROR! BACK TO MAIN PROGRAM!"
 GOTO MAIN1

_CLOSED
CLR BIT1921
DWL 1
DV0=P6160/10000
LV2=DV0
DV1=DV0-LV2
PRINT "CLOSED LOOP CONTROL"
IF (LV1=LV2) THEN PRINT "WAITING"
IF (LV1<>LV2 and BIT0 AND NOT BIT3 AND (DV1>0.9 or DV1<0.3))
        LV1=LV2
        SET BIT1924
        SET BIT1927
        PRINT "TOE FORCE TOO HIGH"
        PRINT "Current CAM = " ; LV3
        LV4=LV3-5
  GOTO absolute
ELSE IF(LV1<>LV2 and NOT BIT0 AND BIT3 and (DV1>0.9 or DV1<0.3))
        LV1=LV2
        SET BIT1925
        SET BIT1927
        PRINT "TOE FORCE TOO LOW"
        PRINT "CURRENT CAM= "; LV3
        LV4=LV3+5
        GOTO absolute
ELSE IF (LV1<(LV2-5))
        LV1=LV2
        SET BIT1925
        SET BIT1927
        PRINT "Adjusting up by 1%"
        PRINT "CURRENT CAM= "; LV3
```

```
          LV4=LV3+1
          GOTO absolute
ELSE
  PRINT "LOAD OK"
  PRINT "CAM STABLE AT " ;LV3
ENDIF
IF (NOT BIT0 AND not BIT3) THEN GOTO MAIN1
GOTO CLOSED

_INCREMENT
PRINT "Current attenuation is" ; LV3
PRINT "Press 1 for gross decrease"
PRINT "Press 2 for fine decrease"
PRINT "Press 3 for fine increase"
PRINT "Press 4 for gross increase"
PRINT "PRESS 5 TO ESCAPE"
INPUT; $V0
LV4=val($V0)
PRINT LV4
IF (LV4=1)
 LV4=LV3-5
 GOTO absolute
ELSE IF (LV4=2)
 LV4=LV3-1
 GOTO absolute
ELSE IF (LV4=3)
 LV4=LV3+1
 GOTO absolute
ELSE IF (LV4=4)
 LV4=LV3+5
 GOTO absolute
ELSE IF (LV4=5)
 GOTO MAIN1
ELSE
 PRINT "ERROR"
GOTO INCREMENT
'
_ATT
'user-directed attenuated camming
LV0=0
```

```
'prompt and wait for user input before proceeeding to automatic attenuation to 50%
PRINT "Type desired attenuation %:"
INPUT; $V0
 PRINT $V0
 LV4=VAL($V0)
 IF (LV4<1)
  PRINT "invalid, attenuation must be greater than 0"
  GOTO ATT
 ELSE IF (LV4>100)
  PRINT "invalid, maximum attenuation is 100"
  GOTO _ATT
 ELSE
  PRINT "Valid input"
ENDIF
'check for difference between current attenuation and desired attenuation and modulate

_absolute
IF (LV3>99 AND BIT1925)
 PRINT "LV3 ALREADY 100, NO CHANGE"
 LV3=100
 CLR BIT1925
 GOTO CLOSED
ELSE IF (LV3<1 AND BIT1924)
 PRINT "LV3 ALREADY 0, NO CHANGE"
 LV3=0
 CLR BIT1924
 GOTO CLOSED
ELSE IF ((ABSF (LV3-LV4)) <=10)
 LV3=LV4
 PRINT "Attaining " ;LV4
ELSE IF (LV4>LV3)
 LV3=LV3+10
 PRINT "increasing to " ;LV3
ELSE
 LV3=LV3-10
 PRINT "decreasing to " ;LV3
ENDIF
LV0=0

FOR LV0=34 TO 59 STEP 1
```

```
  LA1(LV0)=LA2(LV0)
NEXT

 LA1(3)=0.51*LA2(3)*(LV3/100)
 LA1(4)=0.61*LA2(4)*(LV3/100)
 LA1(5)=0.71*LA2(5)*(LV3/100)
 LA1(6)=0.75*LA2(6)*(LV3/100)
 LA1(7)=0.8*LA2(7)*(LV3/100)
 LA1(8)=0.85*LA2(8)*(LV3/100)
 LA1(9)=0.89*LA2(9)*(LV3/100)
 LA1(10)=0.92*LA2(10)*(LV3/100)
 LA1(11)=0.95*LA2(11)*(LV3/100)
 LA1(12)=0.98*LA2(12)*(LV3/100)
 LA1(13)=LA2(13)*(LV3/100)
 LA1(14)=LA2(14)*(LV3/100)


FOR LV0=60 TO 68 STEP 1
  LA1(LV0)=LA1(59)-(LV0-59)*((LA1(59)-LA1(3))/12)
NEXT

LA1(0)=LA1(68)

FOR LV0=1 TO 2 STEP 1
  LA1(LV0)=LA1(59)-((LV0+68)-59)*((LA1(59)-LA1(3))/12)
NEXT

FOR LV0=15 TO 32 STEP 1
 LA1(LV0)=LA2(LV0) * (LV3/100)
NEXT

LA1(33)=250

LV0=0

CLR BIT 159
WHILE (NOT BIT 159)
DV0=P6160/10000
LV2=DV0
DV1=DV0-LV2
```

```
'DV0=P6160/10000  'flywheel rotations
'LV2=DV0   'whole number of fw rotations
'DV1=DV0-LV2  'fraction of fw rotation
 IF(DV1 > 0.90 and DV1 <0.91)
     FOR LV0=0 TO 68 STEP 1
      LA0(LV0) = LA1(LV0)
     NEXT
     SET BIT 159
   ELSE IF (BIT 160)
     PRINT "!"
   ELSE
     CLR BIT 160
   ENDIF
WEND
IF (LV3<>LV4) THEN GOTO absolute
IF (BIT1924)
 CLR BIT1924
 GOTO CLOSED
ELSE IF (BIT1925)
 CLR BIT1925
 GOTO CLOSED
ELSE PRINT "Going to main1"
ENDIF
GOTO MAIN1
'
_SP
Print "Select time frame for Ashworth test"
Print "1 Fast (1 sec)"
Print "2 Med (1.5 sec)"
Print "3 Slow (2 sec)"
Input ; $V0
LV4=Val($V0)
Print LV4

DWL(10)
JOG VEL L5
JOG ACC L 50
JOG DEC L50
JOG ABS L5.5
```

```
PRINT "JOGGING TO TOE DOWN"
DWL (2)

If (LV4=1)
 JOG VEL L18
 ELSE IF (LV4=2)
  JOG VEL L9
 ELSE IF (LV4=3)
  JOG VEL L6
 ELSE PRINT "Invalid input, try again"
  GOTO SP
ENDIF
JOG ACC L 500
JOG DEC L500
JOG ABS L-5.5
PRINT "JOGGING TO DORSIFLEXION"
DWL(2)
PRINT "STANDING PERTUBATION COMPLETE"
JOG OFF
DWL (0.5)
JOG VEL L5
JOG ACC L 50
JOG DEC L50
JOG ABS L0
JOG VEL L1
JOG ACC L25
JOG DEC L25
goto MAIN1
'
_AUTO
PRINT "AUTO LOOP, PRESS GREEN BUTTON AGAIN"
INH -3
PRINT "GB PUSHED"
INH 3
PRINT "GB RELEASED, STARTING AUTO"
FOR LV1=10 TO 50 STEP 10
 LV3=100-LV1
 PRINT LV3
 LV0=0
 FOR LV0=0 TO 68 STEP 1
```

```
 LA1(LV0)=(LA2(LV0)*(LV3/100))
NEXT
LV0=0
CLR BIT 159
WHILE (NOT BIT 159)
 DV0=P6160/10000
 LV2=DV0
 DV1=DV0-LV2
 IF(DV1>0.90 AND DV1<0.91)
  FOR LV0=0 TO 68 STEP 1
   LA0(LV0)=LA1(LV0)
  NEXT
  SET BIT 159
 ELSE IF (BIT 160)
  PRINT "!"
 ELSE
  CLR BIT 160
 ENDIF
WEND
NEXT
PRINT "MAIN 1"
GOTO MAIN1
'
ENDP
```

Appendix J

Right Parker Code

Program 0:

```
PROGRAM
PBOOT
DETACH
ATTACH MASTER0
ATTACH SLAVE0 AXIS0 "R"
PPU R 8000.0000
AXIS0 EXC(5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation,
less than .1 for footplate)
SET BIT8469 : REM enable EXC response
TLM R6 : REM set torque limit to +- 2 V
REM Axis Gains values
AXIS0 PGAIN 0.008
AXIS0 IGAIN 0
AXIS0 ILIMIT 0
AXIS0 IDELAY 0
AXIS0 DGAIN 0.0001
AXIS0 DWIDTH 0
AXIS0 FFVEL 0
AXIS0 FFACC 0
AXIS0 TLM 10
AXIS0 FBVEL 0
REM Axis Limits
AXIS0 HLBIT 1
AXIS0 HLDEC 100
HLIM R3
SET BIT16144
SET BIT16145
CLR BIT16146
SET BIT16148
```

```
SET BIT16149
AXIS0 SLM (20,-20)
AXIS0 SLDEC 100
SLIM R3
SET BIT16150
SET BIT16151
REM MOTION PROFILE
REM the desired master acceleration
ACC 100
REM the desired master deceleration ramp
DEC 100
REM the desired master stop ramp (deceleration at end of move)
STP 250
REM the desired master velocity
VEL 10
REM the desired acceleration versus time profile.
JRK 0
JOG VEL R 1
JOG ACC R 25
JOG DEC R 25
REM BEGIN HOMING SEQUENCE
CLR BIT136
CLR BIT137
clr bit0
 clr bit1
 clr bit2
 clr bit3
 CLR BIT1920
 clr bit1921
PRINT "Press green button To start homing, press red button To stop at any time"
'
_MAIN1
 IF (NOT BIT1 OR NOT BIT2) THEN SET BIT1920 REM RED BUTTON OR ANY EOT
SWITCH
 IF (BIT1920) THEN SET BIT8467
 IF (BIT 1920) THEN CLR BIT136
 IF (NOT BIT0 AND NOT BIT3) THEN SET BIT1921  REM 0001 GREEN BUTTON
 IF (BIT1921 AND NOT BIT136) THEN GOTO HOMING
 IF (BIT136 AND NOT BIT137) THEN GOTO CAMMING: REM IF BIT 136 (USER
DEFINED = HOMING COMPLETE) IS SET, START CAMMING
```

```
 IF (BIT8467) THEN CLR BIT136 REM IF A KILL ALL MOTION FLAG IS SET (8467)
THEN CLEAR BIT 136 AND TURN THE CAM OFF
 IF (BIT1921 AND BIT136 AND BIT137) THEN GOTO CHANGE
 GOTO MAIN1
'
_HOMING
PRINT "BEGIN HOMING"
BIT798=0 : REM CHECK JOG LIMITS WHEN JOGGING FWD/REV
JOG VEL R1 : REM SET JOG VELOCITY TO 1 REV/S
DRIVE ON R
CLR 8467
JOG FWD R
PRINT " JOGGING IN POSITIVE DIRECTION "
INH -792 : REM WAIT UNTIL MOTION HAS STOPPED
PRINT " POSITIVE LIMIT SWITCH FOUND "
CLR 8467 : REM CLEAR KILL ALL MOVES FLAG THAT IS SET WHEN A LIMIT IS
REACHED
JOG REV R
PRINT " JOGGING IN NEGATIVE DIRECTION "
INH -792
PRINT " NEGATIVE LIMIT SWITCH FOUND "
PRINT " ZERO POSITION AT NEG SWITCH "
CLR 8467
JOG INC R6.58
PRINT " MOVING TO OFFSET POSITION "
INH -792
PRINT " AT OFFSET POSITION"
JOG RES R0
RES R0
PRINT " ZERO POSITION REGISTER AT HOME POSITION"
SET BIT 136
clr BIT 137
clr bit1921
clr bit1936
GOTO MAIN1
'
_CAMMING
AXIS0 EXC (5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation,
less than .1 for footplate)
DIM LA(4) : REM Dimension 1 long arrays
```

```
DWL 0.5
DIM LA0(69) : REM LAO has 69 elements
DWL 0.5
DIM LA1(69)
DWL 0.5
DIM LA2(69)
DWL 0.5
DIM LA3(69)
DWL 0.5

LA0(0)=339.226928166667
LA0(1)=443.273956666667
LA0(2)=555.969912833333
LA0(3)=665.807622
LA0(4)=802.524442333333
LA0(5)=938.461863
LA0(6)=1076.0084055
LA0(7)=1240.614497
LA0(8)=1424.72406666667
LA0(9)=1692.78635833333
LA0(10)=2004.310565
LA0(11)=2335.63963833333
LA0(12)=2671.48274333333
LA0(13)=3006.37583833333
LA0(14)=3355.59110666667
LA0(15)=3690.29117333333
LA0(16)=4027.62613166667
LA0(17)=4363.927885
LA0(18)=4610.572525
LA0(19)=4766.083325
LA0(20)=4781.586975
LA0(21)=4705.83735166667
LA0(22)=4552.33108666667
LA0(23)=4335.92759333333
LA0(24)=4059.82390166667
LA0(25)=3725.80573333333
LA0(26)=3329.39902333333
LA0(27)=2847.62354833333
LA0(28)=2271.33554333333
LA0(29)=1668.67754166667
```

LA0(30)=1057.03093866667
LA0(31)=427.2144505
LA0(32)=-200.624312166667
LA0(33)=-803.5657265
LA0(34)=-1387.36292633333
LA0(35)=-1939.90853166667
LA0(36)=-2463.29105166667
LA0(37)=-2968.41557
LA0(38)=-3450.05389666667
LA0(39)=-3893.47668
LA0(40)=-4298.401745
LA0(41)=-4658.95064666667
LA0(42)=-4969.22602833333
LA0(43)=-5236.16728166667
LA0(44)=-5465.644645
LA0(45)=-5644.59219333333
LA0(46)=-5789.04185666667
LA0(47)=-5814.93966666667
LA0(48)=-5678.261535
LA0(49)=-5403.76587
LA0(50)=-5043.02838
LA0(51)=-4582.52410166667
LA0(52)=-4102.50969833333
LA0(53)=-3587.18502
LA0(54)=-3053.40115166667
LA0(55)=-2520.79283
LA0(56)=-1999.80411333333
LA0(57)=-1489.19091416667
LA0(58)=-1076.852008
LA0(59)=-790.619221166667
LA0(60)=-594.294135333333
LA0(61)=-443.957026833333
LA0(62)=-340.140709
LA0(63)=-217.485689666667
LA0(64)=-97.7279608333333
LA0(65)=23.0387716666667
LA0(66)=137.848582666667
LA0(67)=238.5903135
LA0(68)=339.226928166667

LA2(0)=339.226928166667
LA2(1)=443.273956666667
LA2(2)=555.969912833333
LA2(3)=665.807622
LA2(4)=802.524442333333
LA2(5)=938.461863
LA2(6)=1076.0084055
LA2(7)=1240.614497
LA2(8)=1424.72406666667
LA2(9)=1692.78635833333
LA2(10)=2004.310565
LA2(11)=2335.63963833333
LA2(12)=2671.48274333333
LA2(13)=3006.37583833333
LA2(14)=3355.59110666667
LA2(15)=3690.29117333333
LA2(16)=4027.62613166667
LA2(17)=4363.927885
LA2(18)=4610.572525
LA2(19)=4766.083325
LA2(20)=4781.586975
LA2(21)=4705.83735166667
LA2(22)=4552.33108666667
LA2(23)=4335.92759333333
LA2(24)=4059.82390166667
LA2(25)=3725.80573333333
LA2(26)=3329.39902333333
LA2(27)=2847.62354833333
LA2(28)=2271.33554333333
LA2(29)=1668.67754166667
LA2(30)=1057.03093866667
LA2(31)=427.2144505
LA2(32)=-200.624312166667
LA2(33)=-803.5657265
LA2(34)=-1387.36292633333
LA2(35)=-1939.90853166667
LA2(36)=-2463.29105166667
LA2(37)=-2968.41557
LA2(38)=-3450.05389666667
LA2(39)=-3893.47668

```
LA2(40)=-4298.401745
LA2(41)=-4658.95064666667
LA2(42)=-4969.22602833333
LA2(43)=-5236.16728166667
LA2(44)=-5465.644645
LA2(45)=-5644.59219333333
LA2(46)=-5789.04185666667
LA2(47)=-5814.93966666667
LA2(48)=-5678.261535
LA2(49)=-5403.76587
LA2(50)=-5043.02838
LA2(51)=-4582.52410166667
LA2(52)=-4102.50969833333
LA2(53)=-3587.18502
LA2(54)=-3053.40115166667
LA2(55)=-2520.79283
LA2(56)=-1999.80411333333
LA2(57)=-1489.19091416667
LA2(58)=-1076.852008
LA2(59)=-790.619221166667
LA2(60)=-594.294135333333
LA2(61)=-443.957026833333
LA2(62)=-340.140709
LA2(63)=-217.485689666667
LA2(64)=-97.7279608333333
LA2(65)=23.0387716666667
LA2(66)=137.848582666667
LA2(67)=238.5903135
LA2(68)=339.226928166667

DWL(0.5)
DIM LV(5)
LV0=0
LV3=100
LV4=0

PRINT "SLOWLY MOVE FLYWHEEL FORWARD UNTIL THE FOOTPLATES BEGIN
MOVING"
INTCAP AXIS0 10 : REM arms capture of axis0 position when HS inp 4 rises
(designated by 10)
```

```
INH 777: REM wait for flag 777 to be set (flag 777 is set when inp 4 trips intcap)
enc1 res -2912 : REM resets encoder to -2912 so it is zero at BDC on the right.
set bit 138
PRINT "Index detected. Encoder reset."
CAM DIM R1 : REM Define 1 cam segments
CAM SEG R(0,10000,LA0) : REM Define cam segment range and source
CAM SCALE R (1/1000) : REM scales cam output back to revolutions
CAM SRC R1 : REM Define cam source as ENC1
CAM SRC RES : REM resets the cam source to 0
set bit 137
'
_loop
IF (p6160 = 0) THEN CAM ON R
IF (BIT 790) THEN GOTO MAIN1: REM Start camming
GOTO loop
'
_CHANGE
PRINT "CHANGE PATTERN"
DIM DV(2)
DIM $V (2,6)
PRINT "Which program?"
PRINT "1 Normal Camming"
PRINT "2 Attenuated Camming"
PRINT "3 Auto Attenuation"
PRINT "4 Standing Pertubation"
PRINT "5 incremented/decremented camming"
PRINT "6 closed loop camming"
 '
INPUT; $V0
PRINT $V0
LV4=VAL($V0)
PRINT "LV4=";LV4
'
IF (LV4=1) THEN PRINT "1 Normal Camming"
IF (LV4=2) THEN PRINT "2 Attenuated Camming"
IF (LV4=3) THEN PRINT "3 Auto Attenuation"
IF (LV4=4) THEN PRINT "4 Standing Pertubation"
IF (LV4=5) THEN PRINT "5 incremented/decremented camming"
IF (LV4=6) THEN PRINT "6 closed loop camming"
```

```
IF (LV4=1) THEN GOTO MAIN1
IF (LV4=2) THEN GOTO ATT
IF (LV4=3) THEN GOTO AUTO
IF (LV4=4) THEN GOTO SP
IF (LV4=5) then goto INCREMENT
IF (LV4=6) then goto CLOSED
'IF (not bit3) THEN GOTO MAIN1
PRINT "ERROR! BACK TO MAIN PROGRAM"
GOTO MAIN1
'
_CLOSED
CLR BIT1921
DV0=P6160/10000
LV2=DV0
DV1=DV0-LV2
IF (LV1<>LV2 and BIT0 AND NOT BIT3 AND (DV1>0.4 and DV1<0.8))
  LV1=LV2
 SET BIT1924
   SET BIT33
   'SET BIT1927
    PRINT "TOE FORCE TOO HIGH"
    PRINT "Current CAM = " ; LV3
    LV4=LV3-5
    GOTO absolute
ELSE IF(LV1<>LV2 and NOT BIT0 AND BIT3 and (DV1>0.4 and DV1<0.8))
  LV1=LV2
   SET BIT1925
   SET BIT33
   'SET BIT1927  ' program 4 tells motion monitor change in progress
   PRINT "TOE FORCE TOO LOW"
   PRINT "CURRENT CAM= "; LV3
   LV4=LV3+5
   GOTO absolute
ELSE IF (LV1<(LV2-5))
 LV1=LV2
   SET BIT1925
   SET BIT33
   'SET BIT1927
   PRINT "Adjusting up by 1%"
   PRINT "CURRENT CAM= "; LV3
```

```
   LV4=LV3+1
   GOTO absolute
ELSE IF (DV1>0.9)
          PRINT "LOAD OK"
          PRINT "CAM STABLE AT " ;LV3
          DWL 0.5
ENDIF
IF (NOT BIT0 AND not BIT3) THEN GOTO MAIN1
GOTO CLOSED
'
_INCREMENT
PRINT "Current attenuation is" ; LV3
PRINT "Press 1 for gross decrease"
PRINT "Press 2 for fine decrease"
PRINT "Press 3 for fine increase"
PRINT "Press 4 for gross increase"
PRINT "PRESS 5 TO ESCAPE"
INPUT; $V0
LV4=val($V0)
PRINT LV4
IF (LV4=1)
 LV4=LV3-5
 GOTO absolute
ELSE IF (LV4=2)
 LV4=LV3-1
 GOTO absolute
ELSE IF (LV4=3)
 LV4=LV3+1
 GOTO absolute
ELSE IF (LV4=4)
 LV4=LV3+5
 GOTO absolute
ELSE IF (LV4=5)
 GOTO MAIN1
ELSE
 PRINT "ERROR"
GOTO INCREMENT
'
_ATT
'user-directed attenuated camming
```

```
LV0=0
'prompt and wait for user input before proceeeding to automatic attenuation to 50%
PRINT "Type desired attenuation %:"
INPUT; $V0
PRINT $V0
LV4=VAL($V0)
IF (LV4<1)
   PRINT "invalid, attenuation must be greater than 0"
   GOTO ATT
ELSE IF (LV4>100)
   PRINT "invalid, maximum attenuation is 100"
   GOTO _ATT
ELSE
   PRINT "Valid input"
ENDIF

'check for difference between current attenuation and desired attenuation and modulate
_absolute
IF (LV3>99 AND BIT1925)
  PRINT "LV3 ALREADY 100, NO CHANGE"
  LV3=100
  CLR BIT1925
  clr bit33
  GOTO CLOSED
ELSE IF (LV3<1 AND BIT1924)
  PRINT "LV3 ALREADY 0, NO CHANGE"
  LV3=0
  CLR BIT1924
  clr bit33
  GOTO CLOSED
ELSE IF (ABSF (LV4-LV3) <=10)
  LV3=LV4
  PRINT "Attaining " ;LV4
ELSE IF (LV4>LV3)
  LV3=LV3+10
  PRINT "increasing to " ;LV3
ELSE
  LV3=LV3-10
  PRINT "decreasing to " ;LV3
ENDIF
```

```
LV0=0
FOR LV0=0 TO 20 STEP 1
  LA1(LV0)=LA2(LV0)
NEXT
LA1(21)=(LA2(21))
LA1(22)=(LA2(22))
LA1(23)=(LA2(23))
LA1(24)=(LA2(24))
LA1(25)=(LA2(25))

LA1(37)=0.51*LA2(37)*(LV3/100)
LA1(38)=0.61*LA2(38)*(LV3/100)
LA1(39)=0.71*LA2(39)*(LV3/100)
LA1(40)=0.75*LA2(40)*(LV3/100)
LA1(41)=0.8*LA2(41)*(LV3/100)
LA1(42)=0.85*LA2(42)*(LV3/100)
LA1(43)=0.89*LA2(43)*(LV3/100)
LA1(44)=0.92*LA2(44)*(LV3/100)
LA1(45)=0.95*LA2(45)*(LV3/100)
LA1(46)=0.98*LA2(46)*(LV3/100)
LA1(47)=(LA2(47))*(LV3/100)
LA1(48)=(LA2(48))*(LV3/100)

FOR LV0 = 26 TO 36 STEP 1
  LA1(LV0)=LA1(25)-(LV0-25)*(LA1(25)-LA1(37))/12
NEXT

FOR LV0=49 TO 66 STEP 1
  LA1(LV0)=LA2(LV0)*(LV3/100)
NEXT

LA1(67)=0.70*LA2(68)*(LV3/100)
LA1(68)=LA1(0)
LV0=0

CLR BIT 159
WHILE (NOT BIT 159)
  DV0=P6160/10000
  LV2=DV0
  DV1=DV0-LV2
```

```
   IF(DV1 > 0.90 and DV1 <0.91)
     FOR LV0=0 TO 68 STEP 1
      LA0(LV0) = LA1(LV0)
     NEXT
     SET BIT 159
   ELSE IF (BIT 160)
     PRINT "!"
   ELSE
     CLR BIT 160
   ENDIF
WEND
IF (LV3<>LV4) THEN GOTO absolute
IF (BIT1924)
  CLR BIT1924
 clr bit33
 GOTO CLOSED
ELSE IF (BIT1925)
  CLR BIT1925
  clr bit33
  GOTO CLOSED
ELSE PRINT "Going to main1"
ENDIF
GOTO MAIN1
'
_SP
Print "Select time frame for Ashworth test"
Print "1 Fast (1 sec)"
Print "2 Med (1.5 sec)"
Print "3 Slow (2 sec)"
Input ; $V0
LV4=Val($V0)
Print LV4

DWL (10)
JOG VEL R 5
JOG ACC R 50
JOG DEC R 50
JOG ABS R5.5
PRINT "Jogging to toe down"
DWL (2)
```

```
If (LV4=1)
 JOG VEL R18
 ELSE IF (LV4=2)
  JOG VEL R9
 ELSE IF (LV4=3)
  JOG VEL R6
 ELSE PRINT "Invalid input, try again"
  GOTO SP
ENDIF
JOG ACC R 500
JOG DEC R500
JOG ABS R-5.5
PRINT "JOGGING TO DORSIFLEXION"
DWL(2)
PRINT "STANDING PERTUBATION COMPLETE"
JOG OFF
DWL (0.5)
JOG VEL R5
JOG ACC R 50
JOG DEC R50
JOG ABS R0
JOG VEL R1
JOG ACC R25
JOG DEC R25
goto MAIN1
'
_AUTO
PRINT "auto loop, press green button again to start auto attenuation"
INH -3
PRINT "GB pushed"
INH 3 'wait for green button to be pressed and released again before starting automatic
attentuation
PRINT "GB released, starting AUTO"
FOR LV1 = 10 TO 50 STEP 10
 LV3=100-LV1
 print LV3
 LV0=0
  FOR LV0 = 0 TO 68 STEP 1
  LA1(LV0)=(LA2(LV0)*(LV3/100))
```

```
  NEXT
 LV0=0
 CLR BIT 159
 WHILE (NOT BIT 159)
  DV0=P6160/10000
  LV2=DV0
  DV1=DV0-LV2
  IF(DV1 > 0.90 and DV1 <0.91)
   FOR LV0=0 TO 68 STEP 1
   LA0(LV0) = LA1(LV0)
   NEXT
   SET BIT 159
  ELSE IF (BIT 160)
   PRINT "!"
  ELSE
   CLR BIT 160
  ENDIF
'PRINT "NEW VALUES SET"
 WEND
 'FOR LV0=0 TO 68 STEP 1 LA0(LV0)=LA1(LV0) NEXT:
 'WHILE (LV0<=68)
 '   DV0=P6160/10000
 ' LV2=DV0
 ' DV1=DV0-LV2
 ' IF (DV1 = (31*148/10000)) THEN LA1(LV0)=(LA2(LV0)*(LV3/100)):LV0=LV0+1
 'WEND
 'PRINT LA0(68)
NEXT
PRINT "main1"
GOTO MAIN1
ENDP
ENDP
```

Program 3:

```
PROGRAM
'Program 3
'TODO: edit your program here
PBOOT
_main
```

```
intcap axis0 10
inh 777
P0=P6916
while (P6916<(P0+500))
 set 32
wend
clr 32
goto main
ENDP
```