PERFORMANCE SYMMETRY AND MAXIMUM JOINT TORQUES DURING RECOVERY FROM A SIMULATED TRIP

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(ABSTRACT)

Tripping causes a significant number of falls in the elderly. These falls often result in medical costs, hospitalization, disability, decrease in quality of life, and sometimes death. Knowledge of why trips occur and the mechanics of successful recovery from a trip is critical to increasing knowledge of how to prevent falls due to trips. Two separate studies are reported in this thesis. The first study assessed if men recover from a trip equally well when stepping with their dominant or non-dominant lower limbs. An experimental model of tripping was used to determine each subject's trip recovery capability when stepping with the dominant or non-dominant lower limb. Although most subjects were able to recover better when stepping with one lower limb compared to the other, there was no recognizable trend across the subjects. Based on these results, there is insufficient data to recommend the preferential investigation of the dominant or non-dominant lower limb in future trip research. The second study investigated peak joint torques after stepping to recover from a simulated trip. The same protocol as the first study was used for simulating trips. Increasing trip severity resulted in increased ankle plantarflexor torque in young subjects and increased hip extensor torque in both young and older subjects. Older men used higher hip extensor torques and lower knee extensor torques compared to young men. Implications to falls from trips are discussed.

Dedication

To my Mom. Thank you for all the love and support throughout my life. Thank you for the all the sacrifices you made for my education. I love you.

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Chapter 1: Literature Review

I. Introduction to Trips and Their Significance to Our Society

Falls are a common cause of injury and death in the elderly. Approximately 30% of people over age 65 fall once a year. Of these people, approximately 24% will sustain a serious injury, 48% become afraid of falling, and over 25% reduce their activity level because of their fear of falling (Tinetti et al. 1988). Falls in 1995 had an overall total cost of $64 billion when medical, rehabilitation, hospital costs, and costs of morbidity and mortality were considered (Englander et al. 1996). One of the most serious injuries resulting from a fall is hip fracture. In 1990-1991 89% of hip fractures were the result of a fall (Cumming and Klineberg 1994). Half of people over age 75 who suffer a hip fracture die within a year of the injury (Hendrich 1988).

Trips cause up to 53% of falls in the community-dwelling elderly population (Blake et al. 1988). Trips occur when a person's foot unexpectedly impacts a foreign obstacle during gait. This typically causes the body center of mass to translate anteriorly beyond the base of support. With the center of mass anterior to the base of support, a forward step is necessary to extend the base of support to prevent a fall.

There are two general approaches to preventing the morbidity and mortality associated with trips. The first approach is to prevent trips from occurring. Modifying extrinsic factors by improving lighting and removing throw rugs, for example, decreases the environmental factors that contribute to trips. Therefore, the second approach to preventing the morbidity and mortality associated with trips is to improve the ability of elderly adults to recover from a trip. To accomplish this, it is first important to understand the biomechanical factors that are critical to trip recovery. This is the overall

focus of this literature review and thesis: to improve our understanding of the biomechanical factors that are critical to trip recovery. Once these factors are understood, effective interventions can be developed to improve the ability of the elderly to recover from a trip.

This literature review is organized as follows. First, published studies investigating factors critical to trip recovery are presented a single study at a time. This knowledge is then summarized to remind the reader of the key conclusions of the published research. Next, the background and motivation for the first study of this thesis is presented, which investigates possible asymmetries in trip recovery between the dominant and non-dominant lower limbs. Lastly, the background and motivation for the second study of this thesis is presented, which investigates peak extensor torques in the ankle, knee, and hip after stepping to recover from a trip.

II. Findings of Past Trip Research

Researchers have used several different experimental methodologies to investigate different factors of trip recovery. This first part of the literature review has been organized by experimental methodologies.

A. Simulating Trips by Release from a Forward-leaning Position

Early research in trip recovery begins with Do et al.'s (1982) study of the biomechanical process of falling. They released subjects unexpectedly from a forward leaning posture (maintained by a waist level belt and a horizontal cable) and told them to recover their balance by walking after being released. The initial forward leaning posture simulated a trip by placing the subject's center of mass anterior to their base of support, and the unexpected release simulated the surprise element of a real trip. Subjects were

tested at lean angles randomly chosen between 5 degrees and 35 degrees. After analyzing forceplate data they found a "preparation process" that preceded the lift off of the recovery foot and was not affected by lean angle. More specifically, the timing of the minimum and maximum forces of the recovery foot as well as the timing of toe off did not vary with lean angle. The timing of the heel of the recovery foot leaving the force platform wasn't as consistent but likely resulted from the uncomfortableness of keeping the heel flat on the floor in a forward leaning position. After the "preparation process," an "adaptable recovery process" occurred which was studied with step length and velocity measurements. Step length and velocity were both found to increase with an increase in lean angle, showing that subjects adapted this part of the recovery to the seriousness of the challenge presented to them.

Thelen et al. (1997) used a similar methodology to study 10 young and 10 older men recovering from successively increased lean angles. They studied the effect of age on single step trip recovery. Subjects were limited to a single step recovery with their right foot, which was also their dominant foot. Subjects wore a safety harness to prevent falling in case of unsuccessful trip recovery. Thelen et al. (1997) quantified trip recovery capability in these subjects by determining the maximum lean ($Lean_{MAX}$) that subjects were able to successfully recover balance upon release. Thelen et al. (1997) found that young men recovered to significantly higher magnitudes than older men, and found that this difference in capability was based on recovery speed and not sensory or motor processes in the response initiation. This is consistent with Do et al.'s (1982) result that the "preparation process" is consistent but the "adaptable recovery process" changes. Older adults (Thelen et al. 1997) had slightly longer reaction times (on the order of 10

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ms) that were not believed to be biomechanically significant. Weight transfer time, defined as the time from the minimum force to the lift off of the stepping foot, and step velocity were found to correlate strongly with maximum lean magnitude. The changes in step velocity are consistent with Do et al. (1982) but the weight transfer time is part of the "preparation process" and therefore changes in weight transfer time as lean magnitude increased were unexpected. Interestingly, all young men reached failure by putting excessive weight on the safety harness while all older men who completed the protocol failed by taking more than one step.

Wojcik et al. (1999) expanded Thelen et al.'s (1997) research by using the data of Thelen et al. (1997) and adding 10 young and 10 older women to the study. Wojcik et al. (1999) hypothesized that older women would have poorer trip recovery capability than young women and older men. The same protocol as Thelen et al. (1997) was used, with minor modifications. Wojcik et al. (1999) found that older females had significantly poorer balance recovery capability than all other groups studied (young females, young males, and older males). Older females were also found to have a significantly longer reaction time than young females and took longer steps than young females on successful trials. Wojcik et al. (1999) also found that all older females and all but one young female failed by taking multiple steps, the failure method common for older men from Thelen et al. (1997).

In a follow-up study, Wojcik et al. (2001) examined joint torques and joint range of motion before and during stepping after releasing subjects from forward-leaning postures. They hypothesized that age and gender differences in joint torques would explain the differences in trip recovery capability found by Wojcik et al. (1999). Ten

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subjects from each age/gender group (young female, young male, older female, older male) participated. Ankle plantarflexor and hip flexor torques were measured using isometric tests as well as isokinetic tests at 120 degrees/second because they also hypothesized that maximum joint torque limitations may explain poorer trip recovery in older adults. Flexor and extensor torques not measured for the ankle, knee, and hip were found in the literature. Torques calculated did not indicate that available torque was a limitation in trip recovery for young men, young women, or older men. Conversely, older women had calculated ankle plantarflexor, knee flexor, hip flexor, and hip extensor torques that were equal to or greater than those that were thought to be available from the research or their measurement. Young males used significantly larger hip flexor and extensor torques than older men, but they also achieved much larger $Lean_{MAX}$. Wojcik also reported low levels of correlation between maximum volitional strength and $Lean_{MAX}$ suggesting that muscle strength may not be a limiting factor in trip recovery capability.

Thelen et al. (2000) used the forward lean protocol to study the muscle activities used in recovery from a trip. They measured the electromyogram of 12 lower extremity muscles in 14 young and 12 older healthy subjects. Practice trials were used initially, and then leans were increased with successive trials in fixed increments until the subject could no longer recover in a single step. Myoelectric latencies were found to be different between the age groups in only 2 of the 12 muscle groups studied, and these differences are not believed to be biomechanically significant. The consistency of these myoelectric latencies also supports Do et al.'s (1982) "preparation process" that is consistent across lean angles for most subjects. Older adults had higher myoelectric activity levels than

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younger adults at equivalent lean angles in the vastus lateralis, rectus femoris, and biceps femoris (all thigh muscles) in the stance leg and the tibialis anterior and soleus (both are calf muscles) in the stepping leg. Muscle activation timing also showed that older adults used more time in deactivating three stance leg muscles (soleus, gastrocnemius, and biceps femoris) and in activating two step leg muscles (vastus lateralis and rectus femoris). All muscle activation timing differences occurred during stepping, again consistent with Do et al.'s (1982) "adaptable recovery process." Different muscle activation patterns in the young versus older caused Thelen et al. (2000) to conclude that the different age groups may have different recovery strategies.

Reaction times measured during the forward-lean protocol are on the order of 60 msec (Do et al. 1982, Thelen et al. 1997, Wojcik et al 1999) while research accepts that 200 msec is the normal amount of time required for a person to react to a stimulus before their body produces forces or torques (Thelen et al. 1996). One reason for this discrepancy may be different definitions of reaction time. However, Do et al. (1988) noted that the reaction time from their previous study (Do et al. 1982) seemed short and used EMG measurements in the soleus and tibialis anterior muscles to study if this reaction time was based on stretch receptors in the soleus. Twelve healthy subjects participated. Using lean angles from 5 degrees to 20 degrees or more with unexpected release, they found that the soleus latencies were 59 ± 6 milliseconds and no relation existed between muscle latency and lean angle. Four subjects were studied with both announced and unannounced releases, and soleus latency was 62 ± 11 milliseconds for the announced releases and 63 ± 13 milliseconds for the unannounced releases. Do et al. (1988) concluded that the early responses in the forward fall are not caused by the stretch

receptors in the soleus, but may be caused by the receptors in the abdominal or lumbar level, which sense the drop in pressure caused by the release of the belt before the body would otherwise react to a forward fall.

There are many limitations to the release from a forward-leaning posture method of studying trips. Motivation level may vary between subjects. Subjects do not have momentum at the beginning of the simulation due to the static starting position. The release from the initial position is not completely unexpected as in a real trip, so subjects may initially respond to the change in pressure from the equipment instead of the actual sensation of falling. Most studies also restrict recovery to a single step and the simulation may result in less complicated strategies than would occur in a real trip. However, this method, especially when recovery is restricted to a single step, provides a controlled, reproducible method of simulating trips in the laboratory setting. The single step constraint is also considered a concern by some researchers. Both Luchies et al. (1994) and McIlroy and Maki (1996) found that with backwards pulls, younger subjects often recovered with a single step while older subjects tended to use multiple steps. While the constraint may be "unnatural" to some subjects, it allows for even and consistent testing for all subjects.

B. Simulating Trips by Impeding the Foot during the Swing Phase of Gait

Pavol et al. (2001) induced trips in harnessed older adults by causing a hidden metal obstacle on a walkway to suddenly protrude during the mid-to-late swing phase of gait. Kinematics, choice of recovery strategy, motor response, and possible buckling of the recovery limb were studied with the hypothesis that fallers would react in a manner less suitable to successful trip recovery with regards to these variables than non-fallers.

Subjects were instructed to walk at their own pace and continue walking after being tripped. Trips were classified as a recovery (the subject recovered their balance), fall (the subject relied completely on the harness to catch them), rope assist (the subject was found during post hoc analysis to have used the rope to recover to the extent they did not recover independently), and miss (the subject used the obstacle for recovery). Different recovery strategies were also recognized. A lowering strategy occurred when the tripped foot was lowered and became the support limb while the contralateral limb was the recovery limb. An elevating strategy occurred when the tripped limb was raised and became the recovery limb, with flexion occurring at multiple joints in the recovery limb. A reaching strategy occurred when the tripped limb was raised and became the recovery limb but the limb flexion occurred mainly at the hip. The lowering strategy was more common in the later phases of gait.

Two distinct groups of fallers were found. "During-step" fallers fell within 80 ms of the recovery foot contacting the ground, and "after-step" fallers fell after they completed a recovery step, which was over 470 ms after recovery foot ground contact. "During-step" fallers were walking faster and took longer to lower their recovery foot than those who successfully recovered. Buckling in the recovery limb after stepping in the "after-step" fallers suggests that more knee extensor strength would prevent falls in this group. However, after comparing data for subjects who did and did not fall, Pavol et al. (2001) concluded that walking quickly is a great risk to old people for falling. They suggest that walking speed normalized to body height be used to judge this criteria instead of absolute speed.

van den Bogert et al. (2002) used Pavol et al.'s (2001) data and specifically analyzed "during-step" fallers. They were interested in finding if there was a critical body angle at the time the recovery foot contacts the ground beyond which recovery is not possible. Using an inverted pendulum model, they determined this critical angle and studied the effects slowing walking velocity or reducing response time would have increasing trip recovery success. They found that the critical angle from which people could no longer recover is between 23 degrees and 26.3 degrees. They note that these values are consistent with Wojcik et al.'s (1999) work, although the tasks are different due to the different methodologies. van den Bogert et al. (2002) also found from their model that a higher safe walking velocity could be achieved by decreasing response time. The authors of this study concluded that response time is more important than walking velocity for predicting the outcome of a trip, in contrast to the results of Pavol et al. (2001). Due to the nature of the definition of reaction time in this study, these results do not lend themselves to simple comparison to the results of Do et al. (1982) and others who studied trips from forward leans.

While this method allows for subjects to walk at their own pace and does not leave them in anticipation of the trip to the level of forward-lean testing, this method does not allow for more than one set of data from each subject due to the subject's knowledge of the location of the obstacle after their trial.

C. Simulating Trips by Sudden Acceleration on a Treadmill

Owings et al. (2001) studied 65 older subjects in safety harnesses who stood on a treadmill that was suddenly accelerated without warning. Trip recovery methods were examined kinematically and multiple trials were employed to assess if subjects who

initially fell would improve and subsequently not fall. The acceleration caused their base of support to move behind their center of mass, simulating a trip. With the exception of one subject who withdrew, each subject had five recoveries, with the first recovery being the "untrained" response. Subjects were instructed to regain their balance and continue walking after the treadmill was accelerated.

Of the 65 subjects, the 42 who recovered successfully on their first attempt also recovered successfully from all of their other attempts. Eighteen of the 23 subjects who fell on the first attempt recovered in the following 4 attempts, and the remainder were considered multiple fallers. All subjects except one who withdrew recovered from at least two trips. Those who didn't recover on their first attempt had slower reaction times, a more anterior center of mass at toe off, took a significantly shorter step, generally had their center of mass anterior to their ankle marker, and had greater trunk flexion and trunk velocity at stepping than those who recovered on the first try. Subjects who fell then corrected these problems in their successful trials. When fallers did recover, the kinematic variables just discussed improved to the level of non-fallers with the statistically significant exceptions of the fallers taking a longer step and having a more posterior center of mass during their successful recoveries.

Owings et al. (2001) conclude that this method may be helpful for training older adults to prepare for falls. Those who fell were able to try again and learn to coordinate their movements to succeed without the risk of injury from a real trip. Owings et al. (2001) also report that because the subjects improved without outside training, subjects may already have all the necessary skills to recover from a trip, but may simply need the opportunity to safely bring these skills together.

D. Simulating Trips by Pulling on an Ankle Cuff during Gait

      Smeesters et al. (2001) simulated a trip with young adults by having them walk at a specified pace with an ankle cuff on their dominant lower limb (LL) that was unexpectedly held in position by a cable during mid-swing phase of gait. The ankle cuff was held in position for 100 milliseconds in the first trial and this amount of time was increased in increments of 100 milliseconds until the subject could no longer recover their balance. They studied the effect that LL strength, reaction time, step time, step distance, and step velocity from a previous protocol had on the amount of time that movement of the dominant LL could be stopped during walking before they could no longer recover. During the tripping protocol, subjects were instructed to recover their balance with a single step when the tension in the cable was released. Holding the ankle in a position for a specified amount of time simulated an impulse to the toes for that amount of time during contact with an obstruction. The maximum threshold trip duration subjects could recover from was 681 ± 169 milliseconds. Threshold trip duration was larger for subjects with larger LL strength and for subjects with a smaller reaction time. Smeesters et al. (2001) concluded that trip related falls may be related to reduced LL strength and increased reaction time.

III. Summary of Current Trip Recovery Knowledge

      Thelen et al. (1997) found that young men have a larger trip recovery capability than older men. Wojcik et al. (1999) found that the same relationship exists between young and older women. Wojcik et al. (1999) also found that older women recover the poorest of any of the age/gender groups (young and older men and young and older women) studied.

The research cited in the previous section frequently concludes that trip recovery capability differences between young and older adults is because of differences in the performance of compensatory stepping to increase the base of support and not differences in reaction time or response initiation. Do et al. (1982) first hypothesized this by identifying a uniform "preparation process" that occurs after trip initiation and is followed by an "adaptable recovery process." The results of Thelen et al. (1997) generally agree with this, and the electromyogram data from Thelen et al. (2000) also agrees with Do et al.'s (1982) division of trip recovery into two separate processes. Schultz et al. (1997) suggest from a review of the literature that old people are deficient in strength and speed of muscle contraction and not in sensory processing or motor planning for movements that require speed and strength, such as balance recovery. This also agrees with Do et al.'s (1982) uniform "preparation process" and "adaptable recovery process" definitions.

Pavol et al. (2001) concluded that walking speed was the best predictor of falls in the elderly. They also found that buckling of the recovery limb occurred in falls that occurred after a step was taken, suggesting insufficient knee extensor torque as a mechanism of falling from a trip. van den Bogert et al. (2002) used a model to conclude that shortening reaction time would better increase trip prevention and also allow for an increase in gait speed. However, the practical implications of this study are limited because gait speed may be easily increased while reaction time is likely not easy for an elderly person to modify.

Smeesters et al. (2001) concluded that trips may be related to low LL strength and increased reaction time, although they used reaction time data from a separate protocol

and did not measure it during their trip protocol. Grabiner and Jahnigen (1992) found that non-fallers had a faster reaction time than fallers in achieving isometric knee extensor torques. Wolfson et al. (1997) found that fallers living in a nursing home had less than half the knee and ankle strength of non-fallers.

Owings et al. (2001) found that 22 of 23 subjects who fell on their initial trial of an unannounced sudden acceleration of a treadmill trip simulation were able to learn to recover their balance in subsequent trips, indicating that the elderly may already have the knowledge and physical capabilities to recover from a trip and may simply need the opportunity to safely practice trip recovery to improve their trip recovery capability.

In conclusion, trip research and knowledge have greatly improved in the past two decades, but much more remains to be studied. The two studies discussed in Chapters 2 and 3 were performed to increase the field of knowledge of trip recoveries. The next section will explain the need for research discussed in Chapter 2, and the section following will explain the need for research discussed in Chapter 3.

IV. Introduction to Study #1: Is Trip Recovery Capability Equal When Stepping with the Dominant and Non-Dominant Lower Limbs?

Studying trips in the laboratory often requires the investigators to specify the limb used for recovery (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001, Thelen et al. 2000) or the limb to be tripped (Pavol et al. 2001, van den Bogert et al. 2002, Smeesters et al. 2001). These constraints are due to instrumentation constraints (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001, Thelen et al. 2000, Pavol et al. 2001, van den Bogert et al. 2002).

Current research indicates that the performance of both lower limbs during gait may be asymmetrical (Sadeghi et al. 1997, Sadeghi et al. 2000, Sadeghi 2003). A recent review by Sadeghi et al. (2000) documented significant asymmetries between dominant and non-dominant LL kinematics and kinetics during able-bodied gait. For example, Sadeghi et al. (1997) found that one LL had a propulsion function while the contralateral limb had a supporting function. In a separate study, Sadeghi (2003) reported that the LLs have local asymmetry at the hips, knees, and ankles indicated by an unequal distribution of generating propulsion and coordination between same joints on each LL and unequal peak torques. Maupas et al. (2002) studied the range of motion used in both knees of 40 young subjects while walking on a level surface and walking up and down stairs. They found a significant difference in the range of motion used between the knees of 25 of these 40 young subjects. These differences did not correlate to handedness or footedness. They also found no significant differences for isokinetic knee flexor and extensor strength between the right and left LLs, even if subjects were grouped by footedness. Skelton et al. (2002) tested older women in the community for leg strength, power, and asymmetry of strength and power, and found significant asymmetry between legs for both strength and power for fallers and non-fallers. Fallers also had greater asymmetries in power. Asymmetries were not found to be consistent to a pattern of dominance, and the level of asymmetry between legs did not seem to relate to the number of falls in the falling group.

Another study not investigating gait, but examining LL strength and the kinematics associated with taking a rapid step on cue, also found performance asymmetries between the lower limbs (Smeesters et al. 2001). Only adults aged 18-36

years participated, and before the previously described trip protocol symmetry between legs was tested by having subjects take a single volitional step on cue, and strength was also measured for both LLs. The kinematic variables studied from the rapid step were reaction time, step time, step distance, and step velocity, and no significant differences indicating asymmetry were found in any of these variables. However, the dominant LL was found to be significantly stronger than the non-dominant LL. Even with the significant difference in strength, Smeesters et al. (2001) only tested the dominant LL during their trip protocol.

This research indicates that strength, power, and gait may not be consistent across both LLs in some individuals. If similar asymmetries exist between the LLs during stepping to recover from a trip, then individuals may be better able to recover from a trip by stepping with either the dominant or non-dominant LL. However, trips, unlike gait, are unexpected and may require the non-preferred LL to step to recover balance. Previous research which selected the dominant LL for tripping or trip recovery may have inadvertently allowed subjects to "put their best foot forward." As a result, researchers may not have fully appreciated the age-related declines in trip recovery capability (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001, Pavol et al. 2001). Therefore, the first study (Chapter 2) was designed to investigate if the ability to recover from a trip differs when stepping with the dominant or non-dominant LL for trip recovery. Knowing if individuals are better able to recover from a trip by stepping with the dominant or non-dominant LL may allow researchers to make more informed methodological decisions when designing future trip studies.

V. Introduction to Study #2: Peak Extensor Torques After Stepping to Recover from a Simulated Trip

Age-related reductions in muscle strength are frequently cited as an explanation for the elevated fall rates of older people (Schultz et al. 1997, Pavol et al. 2001, Wojcik et al. 2001). Many studies have attempted to quantify strength in older people. In men, muscular strength peaks during their 20s and 30s but then remains unchanged until their 50s, when it begins to decline by 35-45% by their 80s (Hurley 1995). Women appear to reach peak strength levels sooner and start to lose strength sooner, but with slightly less strength loss than men (Hurley 1995). Thelen et al. (1996) found a 24% loss in isometric plantarflexor torque in older subjects compared to younger subjects. Wolfson et al. (1995) found that nursing home residents who fell had less than half the knee and ankle strength of the non-falling residents. The ankle dorsiflexors were the most extreme case, with fallers having only one tenth the strength of the non-fallers.

Decreased muscle strength causes decreased joint torques. LL joint extensor torques are necessary after stepping to recover from a trip. Pavol et al. (2001) suggested that buckling of the stepping LL may occur despite taking a step after a trip because of poor knee extensor strength. While joint torques have been studied before and during stepping (Wojcik et al. 2001), currently no research addresses joint torques after stepping. It is after stepping that this buckling would occur. Joint torques after stepping may be higher than those reported in the previous literature (Wojcik et al. 2001), and thus indicate that a higher level of strength than previously believed is necessary to recover from a specified trip. Using a forward-leaning trip protocol, increases in joint torque with increases in trip severity would indicate that more strength is needed to recover balance.

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Increased severity of trips can be quantified with increased lean magnitude using this method. If in fact more strength is required to recover from more severe trips, subjects may reach their Lean$_{MAX}$ due to this limitation. Therefore, the second study (Chapter 3) was designed to determine the effect of trip severity on joint torques after stepping to recover balance. Knowing if strength was a limitation in the trip recovery capability of older adults would better allow researchers to improve methods to prevent trips.

In addition to declines in muscle strength with age, age-related changes also occur in gait. For example, older adults use more hip extensor torque and less knee extensor and ankle plantarflexor torque than younger adults (DeVita and Hortobagyi 2000). These changes in strength and gait indicate that joint torques during trip recovery may also change due to age. The second purpose of the second study (Chapter 3) is to determine if older adults use different joint torques than young adults in recovering from a trip. Knowing if different age groups use different amounts of torque at different joints would aid researchers in improving methods to increase trip recovery capability in older adults by better understanding the torques older adults use in trip recovery and how these torques differ from those used by young adults.

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Chapter 2: Is Trip Recovery Capability Equal When Stepping with the Dominant and Non-dominant Lower Limbs?

I. Abstract

Background. Tripping causes a significant number of falls in the elderly. These falls often result in medical costs, hospitalization, disability, decrease in quality of life, and sometimes death. The purpose of this study was to determine if individuals are able to recover from a trip equally well when stepping with their dominant or non-dominant lower limb. This information could be useful during the design of future trip-related research.

Methods. To simulate a trip, ten young (20.6 ± 1.3 years) and ten older (74.0 ± 6.5 years) male subjects were released unexpectedly from a forward-leaning posture and instructed to recover their balance with a single step. This process was repeated using progressively larger lean magnitudes to determine the largest lean from which subjects could successfully recover their balance upon release. The largest lean magnitude from which subjects could successfully recover their balance was used as a measure of trip recovery capability. The experiment was performed twice, once while subjects stepped with the dominant lower limb, and once while subjects stepped with the non-dominant lower limb to determine if trip recovery capability is equal when stepping with the dominant or non-dominant lower limb. Several kinematic variables were also measured and compared across lower limbs.

Results. No significant difference in trip recovery capability was found between when stepping with the dominant or non-dominant lower limb. Most subjects did exhibit an asymmetry in trip recovery capability (i.e. they were able to recover their balance better with one lower limb compared to the other). However, the magnitude and direction of

this asymmetry was inconsistent, so the means exhibited no significant difference. Interestingly, differences in stepping kinematics were found between the lower limbs despite no significant difference in trip recovery capability. Subjects exhibited a shorter total balance recovery time (p=0.044) and a faster step velocity (p=0.014) when stepping with their dominant lower limb to recover their balance. Young subjects analyzed separately exhibited a shorter step time (p=0.022) and a faster step velocity (p=0.017) when stepping with their dominant lower limb. Older subjects analyzed separately did not exhibit any kinematic differences between the dominant and non-dominant lower limbs.

Conclusions. Although most subjects were able to recover better from a simulated trip with one lower limb compared to the other, there was no recognizable trend across the subjects. Because of this, there is insufficient data from this study to recommend the preferential investigation of the dominant lower limb or the non-dominant lower limb in future trip research.

II. Introduction

Falls are a common cause of injury and death in the elderly. Approximately 30% of people over age 65 fall once a year. Of these people, approximately 24% will sustain a serious injury, 48% become afraid of falling, and over 25% reduce their activity level because of their fear of falling (Tinetti et al. 1988). Falls in 1995 had an overall total cost of $64 billion (Englander et al. 1996). Eighty nine percent of hip fractures are the result of a fall and 38% are the result of a trip (Cumming and Klineberg 1994). Half of people over age 75 who suffer a hip fracture die within a year of the injury (Hendrich 1988). Trips cause up to 53% of falls in the community-dwelling elderly population (Blake et al.

1988) As such, a growing number of studies are aiming to understand the causes behind the high number of trip-related falls in the elderly.

Studying trips in the laboratory frequently requires the investigator to specify which foot will be tripped (Pavol et al. 2001, Smeesters et al. 2001) or which foot will be used for trip recovery (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001). Previous trip studies (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001, Smeesters et al. 2001) have selected the dominant lower limb (LL) for these functions presumably by default due to lack of data to do otherwise. Sadeghi (2003) reported that the LLs have local asymmetry at the hips, knees, and ankles indicated by an unequal distribution of generating propulsion and coordination between same joints on each LL and unequal peak torques. If similar asymmetries exist between the LLs during stepping to recover from a trip, then individuals may be better able to recover from a trip by stepping with either the dominant or non-dominant LL. Previous research which selected the dominant LL for tripping or trip recovery may have inadvertently allowed subjects to "put their best foot forward." As a result, researchers may not have fully appreciated the age-related declines in trip recovery capability (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001, Pavol et al. 2001). Therefore, the purpose of this study was to investigate if the ability to recover from a trip differs when stepping with the dominant or non-dominant LL for balance recovery. Knowing if individuals are better able to recover from a trip by stepping with the dominant or non-dominant LL may allow researchers to make more informed methodological decisions when designing future trip studies.

III. Methods

Subjects

Twenty subjects participated including ten young men (20.6 ± 1.3 years) and ten older men (74.0 ± 6.5 years). All subjects reported to be free of musculoskeletal injury. Older subjects were required to have passed a medical screening to rule out cardiac, respiratory, neurological, otological, and musculoskeletal disorders. All subjects verbally identified themselves as being right hand and right foot dominant. Anthropometric characteristics of the subjects are reported in Table 2-1. The study was approved by the Institutional Review Board at Virginia Polytechnic Institute and State University, and all subjects signed informed consent forms prior to participation.

Protocol

The protocol and experimental measures were adapted from Thelen et al. (1997). Tripping commonly occurs during forward gait and causes the body center of mass to translate anteriorly beyond the base of support. To simulate a trip, subjects were unexpectedly released from a forward-leaning posture and asked to recover their balance using a single step of the LL specified by the investigator (Figure 2-1). The release from the forward leaning posture simulates a trip by placing the center of mass of the subject anterior to their base of support. Upon successful recovery, another trial was performed using a larger lean magnitude. Upon failed recovery, a second trial was performed at the same lean magnitude. This process was repeated until subjects failed to recover their balance with a single step for two consecutive trials at the same lean magnitude. Trip recovery capability was defined as the maximum lean ($Lean_{MAX}$) from which subjects could recover their balance with a single step after being released. Subjects then repeated the experiment using their opposite LL to recover their balance to compare trip recovery capability between the dominant and non-dominant LLs. A crossover design was used to

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balance any learning effects between the group first stepping with the dominant LL and the group first stepping with the non-dominant LL.

Subjects were held in the forward-leaning posture using a lean control cable. One end of the lean control cable was attached to the back of a belt worn by subjects, and the other end was held in a releasable clasp. To release subjects, the investigator activated a solenoid that released the clasp. Lean magnitude was quantified by requiring subjects to maintain a consistent foot position and measuring tension in the lean control cable (units of percent body weight (BW)) with an in-line load cell (Cooper, Warrenton, Virginia). Lean magnitude was controlled by adjusting the length of the lean control cable. This method of measuring lean magnitude, as opposed to measuring lean angle, provides a consistent method to measure forward lean across subjects of different sizes, can be measured directly during testing, and does not require that body segments be exactly in-line as for angle measurements.

To prevent falls to the ground in the event of an unsuccessful trip recovery, all subjects wore a full-torso harness that was tethered to a ceiling-mounted track using a fall-prevention lanyard. The length of the lanyard was adjusted so that when subjects reached for the ground, there was approximately 2 inches between their fingertips and the ground. This prevented any part of the subject's body from touching the ground except the feet. A load cell (Cooper, Warrenton, Virginia) was mounted in-line with the lanyard to measure force applied to the subject through the harness.

To start each trial, subjects stood with their feet at a toe line and leaned forward to a position where their body was straight except for dorsiflexion at the ankles. Heels were required to touch the ground and subjects were asked to distribute their weight equally

across both feet. Subjects were asked to fold their arms across their chest throughout each trial. Once they were in position at the correct lean magnitude, subjects were verbally reminded to recover with the LL being tested. Subjects were then released unexpectedly 0-10 seconds after the completion of this statement. The initial lean magnitude corresponded to 12% BW in the lean control cable. Subjects were allowed two attempts at each lean magnitude, and lean magnitude was increased by 4% BW after each successful recovery until the subject failed to recover their balance with a single step. Failure could occur three ways: 1) when the non-stepping foot took a step longer than 30% of the subject's body height (BH), 2) when more than one step was taken by the LL tested, and 3) when a subject put more than 30% BW on the harness at any point during trip recovery. A step less than 30% BH was allowed to aid in lateral stability, while a step larger than this was assumed to contribute to trip recovery. The 30% BW threshold in the harness distinguished trials in which the subject would not have recovered their balance if they were not wearing a harness. These failure criteria were defined by Wojcik et al. (1999).

All subjects practiced single step balance recovery prior to the start of the experiment. Music was played throughout the trials to cover the noises associated with data collection.

Data Collection

Body segment position data were sampled at 200 Hertz using an Optotrak optoelectronic motion analysis system (Optotrak, Waterloo, Ontario). Infrared markers were placed on one side of the body at the fifth metatarsal head, heel, lateral malleolus, lateral femoral epicondyl, greater trochanter, and acromion. Subjects initially stood with

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each foot on one forceplate (AMTI, Watertown, Massachusetts). Upon release, subjects

stepped onto a third forceplate (Bertec, Columbus, Ohio) during trip recovery. This

forceplate was oversized (90 x 90 centimeters) so that subjects did not need to alter their

natural stepping patterns to achieve proper foot placement on the forceplate. Forceplate

data and harness load cell data were sampled at 1000 Hertz. Harness load cell data was

subsequently filtered with a 10 Hertz low-pass, zero-phase-shift Butterworth filter. All

experiments were recorded on videotape for post hoc confirmation of trip recovery and/or

failure criteria.

Data Analysis

The independent measure was LL used for trip recovery (dominant versus non-

dominant). The primary dependent measure studied was $Lean_{MAX}$. Additional dependent

measures included kinematic measures that describe trip recovery (Figure 2-2).

Kinematic measures were computed for each subject at the largest lean magnitude

achieved by both LL. These measures are: reaction time (time from release to the

minimum ground reaction force from the LL tested), weight transfer time (time from

minimum ground reaction force to lifting of the LL off the ground), step time (time the

LL uses between leaving the ground and contacting the ground after stepping), landing

time (time from release to contacting the ground after stepping), step length (length of

step in the sagittal plane measured in percent BH), and step velocity (step length divided

by step time). The Wilcoxon Signed-Ranks Test was used to test for differences in

$Lean_{MAX}$ and kinematic variables between the dominant and non-dominant LLs. A

significance level of $\alpha = 0.05$ was used.

IV. Results and Discussion

One older subject could not recover from the minimum lean magnitude with either LL. All other subjects completed at least the minimum lean magnitude for both LLs.

Results did not show a difference in $Lean_{MAX}$ between dominant and non-dominant LL ($p > 0.05$). To determine if age-related differences in trip recovery capability influence this result, separate tests were performed using only young or older subjects. These tests also found no significant differences in either age group ($p > 0.05$). Although most individual subjects did exhibit asymmetry in their trip recovery capability (Figure 2-3) there was no recognizable trend across all subjects.

Differences between the dominant and non-dominant LLs were found in landing time ($p=0.044$) and step velocity ($p=0.014$). The young group had differences in step time ($p=0.022$) and step velocity ($p=0.017$). The older group showed no kinematic differences when tested separately. All data are summarized in Table 2-2.

Differences in trip recovery kinematics were unexpected given that no differences in trip recovery capability were found. One potential explanation for these differences may be a different stepping strategy between the LLs. Smeesters et al. (2001) studied reaction times, step times, step distances, and step velocities between the dominant and non-dominant LLs in taking a single rapid volitional step on cue. They found no asymmetry in these kinematic variables. Kinematic differences in this study all favored the dominant LL, with faster velocities resulting from shorter balance recovery. Research in gait symmetry indicates that gait may be asymmetrical (Sadeghi et al. 2000). For example, Sadeghi et al. (1997) found that one lower limb had a propulsion function while the contralateral limb had a supporting function. If the LLs function differently in gait, it

is possible that they could use different stepping strategies during recovery from a trip. Investigating joint kinetics during trip recovery while stepping with the dominant and non-dominant LL may help determine if different stepping strategies are implemented.

Another potential explanation for the differences in joint kinematics may be strength differences between the dominant and non-dominant LLs. Smeesters et al. (2001) found that hip flexor strength was significantly greater in the dominant LL for subjects aged 18-36 years. More research on the methods each LL uses to recover from a trip would help better assess the extent of possible asymmetries in balance recovery.

This study had several limitations. First, subjects did not have an initial forward momentum at the start of each trial. Actual trips occur during walking when subjects have a forward momentum proportional to their walking speed. Second, some subjects may have been more motivated than others and had a greater fear of falling. However, previous research (Thelen et al. 1997, Wojcik et al. 1999, Wojcik et al. 2001) indicates this method is acceptable for quantifying trip recovery capability and kinematics during trip recovery. Third, younger individuals tend to recover from a balance perturbation with a single step while older individuals tend to use multiple steps (Luchies et al. 1994, McIlroy and Maki 1996). Constraining trip recovery to a single step may force older subjects to use an unpreferred strategy. However, the effects on this study should be minimal as the trip recovery capability between LLs was compared. Fourth, all kinematic analyses were limited to the sagittal plane, which may have slightly altered step length and step velocity data. Fifth, only male subjects participated in this study, so results may not apply to women.

V. Conclusions

The results from this study provide insufficient data to recommend the specific use of the dominant or non-dominant LL in trip recovery research. Kinematic differences between the dominant and non-dominant LL may reflect differences in stepping strategy or strength between LLs. More research is recommended including a LL kinetic analysis of trip recovery to find the nature and explain the kinematic differences measured between the dominant and non-dominant LL.

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Table 2-1. Mean (standard deviation) of subject age, height, and weight.

	Young	Old
N=	10	10
Age (yr)	20.6(1.26)	74.0(6.50)
Height (m)	1.76(0.068)	1.71(0.058)
Mass (kg)	71.3(14.0)	77.0(14.0)

Table 2-2. Mean (standard deviation) of kinematic variables. A * indicates a significant difference (p<0.05) between the dominant and non-dominant LLs.

	Age	Dominant LL	Non-dominant LL
Lean Maximum (%BW)	Young	34.4(5.06)	32.4(5.80)
	Old	20.0(6.93)	21.3(5.66)
	All	27.6(9.42)	27.2(7.95)
Reaction Time (ms)	Young	89.3(14.9)	84.7(10.4)
	Old	76.2(10.3)	70.9(9.6)
	All	83.1(14.3)	78.2(12.1)
Weight Transfer Time (ms)	Young	186.1(27.6)	195.2(34.9)
	Old	239.0(24.4)	253.3(35.7)
	All	211.2(37.2)	222.7(45.4)
Step Time (ms)	Young	185.1(23.1)	205.7(26.5)*
	Old	209.8(28.6)	213.1(34.1)
	All	196.8(28.1)	209.2(29.7)
Landing Time (ms)	Young	460.5(30.3)	485.6(44.1)
	Old	525.7(43.1)	536.7(55.7)
	All	491.4(49.0)	509.8(55.1)*
Step Length (%BH)	Young	56.7(5.74)	58.5(6.63)
	Old	43.9(8.72)	43.1(8.63)
	All	50.6(9.65)	51.2(10.86)
Step Velocity (BH/s)	Young	3.09(0.449)	2.87(0.358)*
	Old	2.12(0.451)	2.07(0.535)
	All	2.63(0.666)	2.48(0.599)*
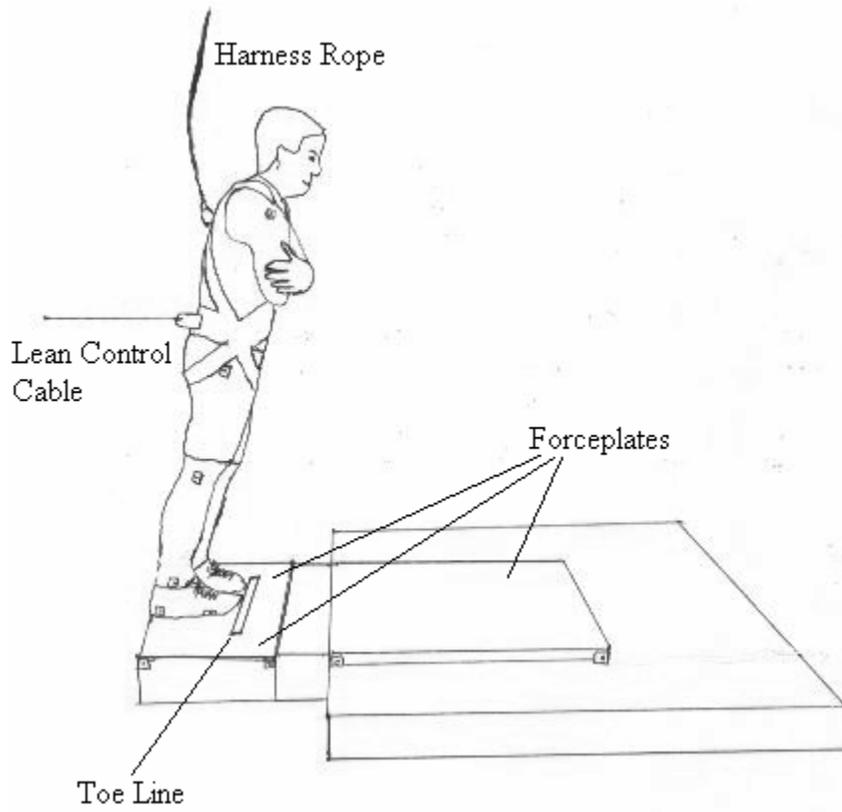
Figure 2-1. Subject ready to be released.
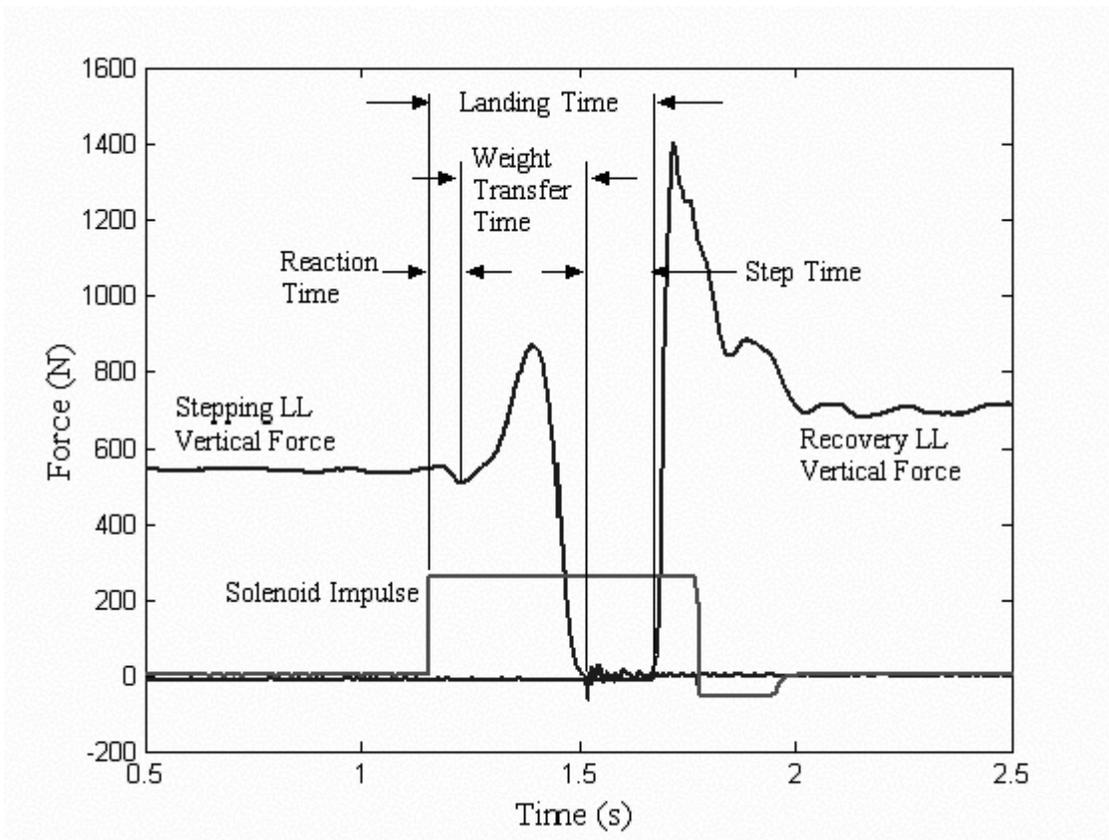
Figure 2-2. Illustration of different kinematic times.
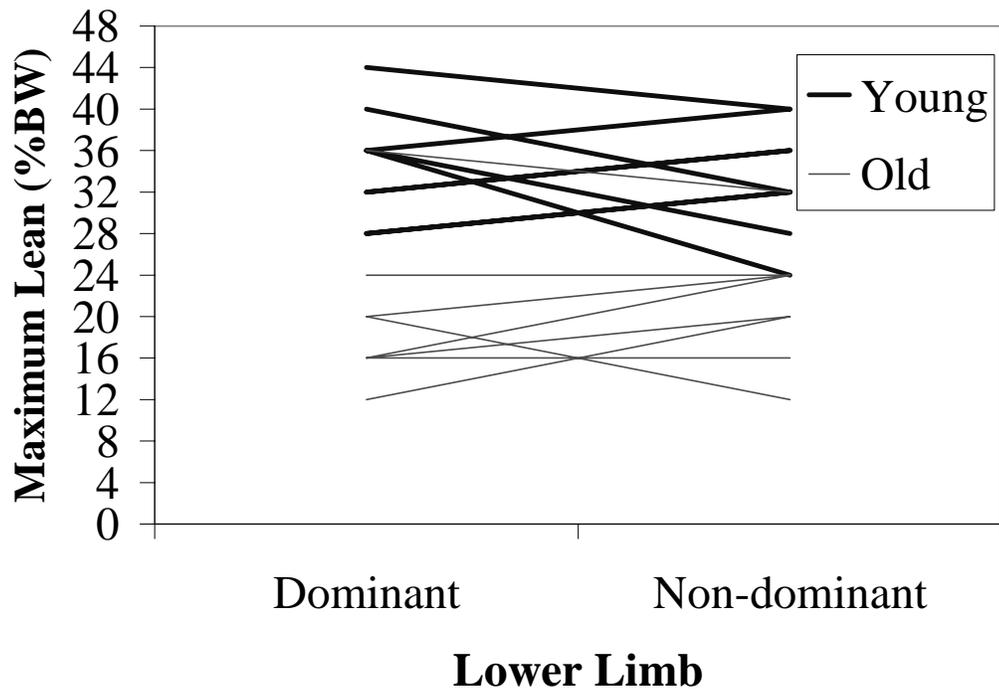
Figure 2-3. Maximum Lean (Lean$_{MAX}$) (%BW) for the dominant and non-dominant lower limbs. Lines sloping upwards indicate better trip recovery capability with the non-dominant LL, lines sloping downwards indicate better trip recovery capability with the dominant LL, and horizontal lines indicate equal trip recovery capability for both LLs. Note that some subjects had the same Lean$_{MAX}$ for both LLs as other subjects so there are fewer lines than subjects.

Chapter 3: Peak Extensor Torques after Stepping to Recover from a Simulated Trip

I. Abstract

Background. Tripping causes a significant number of falls in the elderly. These falls often result in medical costs, hospitalization, disability, decrease in quality of life, and sometimes death. Previous research has shown one mechanism of falling from a trip is buckling of the stepping limb after the completion of a recovery step. This may be due to insufficient muscle strength in the lower limb to resist flexion at the ankle, knee, and hip. No studies, however, have investigated joint kinetics during trip recovery after stepping to recover balance. The current study examines peak extensor torques at the ankle, knee, and hip after stepping to recover from a simulated trip. The two hypotheses were: 1) peak extensor torques at the ankle, knee, and hip after stepping would increase as trip severity increased; and 2) age-related differences in peak extensor torques would exist between young and older men.

Methods. To simulate a trip, ten young (20.6 ± 1.3 years) and ten older (74.0 ± 6.5 years) male subjects were released unexpectedly from a forward-leaning posture and instructed to recover their balance with a single step. This process was repeated using progressively larger lean magnitudes until subjects could no longer recover their balance in a single step. Joint torques in the stepping lower limb were computed with an inverse dynamics analysis for each lean magnitude from which subjects recovered successfully.

Results. Increasing trip severity by releasing subjects from larger lean magnitudes resulted in increased ankle plantarflexion torque in young subjects and increased hip extensor torque in both young and older subjects. However, peak extensor torques did not appear to approach the maximum torque capacity of the subject populations, except

for hip extensor torques in the young subjects. Age effects were found for the knee and hip extensor torques. Specifically, older men used higher hip extensor torques and lower knee extensor torques compared to young men.

Conclusions. Our results suggest increasing trip severity requires larger extensor torques after stepping to prevent LL buckling. Age-related reductions in extensor strength may limit these torques and contribute to higher trip-related fall rates in the elderly. However, comparing measured torques to strength estimates from the literature did not support this. In addition, an age-related difference in joint torques indicates that older men may use a different strategy to prevent LL buckling after a trip.

II. Introduction

Falls are a common cause of injury and death in the elderly. Approximately 30% of people over age 65 fall once a year. Of these people, approximately 24% will sustain a serious injury, 48% become afraid of falling, and over 25% reduce their activity level because of their fear of falling (Tinetti et al. 1988). Falls in 1995 had an overall total cost of $64 billion (Englander et al. 1996). Eighty nine percent of hip fractures are the result of a fall and 38% are the result of a trip (Cumming and Klineberg 1994). Half of people over age 75 who suffer a hip fracture die within a year of the injury (Hendrich 1988). Trips cause up to 53% of falls in the community-dwelling elderly population (Blake et al. 1988). As such, a growing number of studies are aiming to understand the causes behind the high number of trip-related falls in the elderly.

Age-related reductions in muscle strength are well documented (Cahalan et al. 1989, Hurley 1995, Thelen et al. 1996, Schultz et al. 1997, Wojcik et al. 2001). For example, Thelen et al. (1996) found a 24% loss in isometric plantarflexor torque in older

subjects compared to younger subjects. These differences have been implicated as a contributing factor to the higher fall rates in the elderly. However, the biomechanical mechanism by which muscle strength contributes to falls has not been substantiated.

Pavol et al. (2001) found that after an induced trip, some elderly subjects fell despite taking a step to recover their balance. These falls were due in part to buckling of the stepping limb, which they attributed to weakness of the knee extensors. These results suggest the poorer trip recovery capabilities of older adults (Thelen et al. 1997, Wojcik et al. 1999) may be due, at least in part, to decreased muscle strength limiting joint extensor torques after stepping to recover from a trip. However, no studies have calculated joint torques during trip recovery after stepping to recover balance. Therefore, the first purpose of this study was to calculate lower limb (LL) extensor torques after stepping to recover from a trip. It was hypothesized that LL extensor torques after stepping to recover from a trip would increase as the severity of a trip increases.

Another factor that may contribute to LL buckling during recovery from a trip is an age-related difference in joint extensor torques. A study of gait has shown that older adults use more hip extensor torque and less knee extensor and ankle plantarflexor torque than younger adults (DeVita and Hortobagyi 2000). If changes in joint torques during gait occur due to age, it is also possible that changes in joint torques during trip recovery may occur due to age. The second purpose of this study was to examine the effect of age on LL extensor torques. It was hypothesized that older subjects would use different LL joint torques than young subjects in recovering from a simulated trip.

III. Methods

Subjects

Twenty subjects participated including ten young men (20.6 ± 1.3 years) and ten older men (74.0 ± 6.5 years). All subjects reported to be free of musculoskeletal injury. Older subjects were required to have passed a medical screening to rule out cardiac, respiratory, neurological, otological, and musculoskeletal problems. All subjects verbally identified themselves as being right hand and right foot dominant. Anthropometric characteristics of the subjects are reported in Table 3-1. The study was approved by the Institutional Review Board at Virginia Polytechnic Institute and State University, and all subjects signed informed consent forms prior to participation.

Protocol

The protocol was adapted from Thelen et al. (1997). Tripping commonly occurs during forward gait and causes the body center of mass to translate anteriorly beyond the base of support. To simulate a trip, subjects were unexpectedly released from a forward-leaning posture and asked to recover their balance using their right LL. The release from the forward leaning posture simulates a trip by placing the center of mass of the subject anterior to their base of support. Upon successful recovery, the experiment was repeated with a larger lean magnitude. Upon failed recovery, a second trial was performed at the same lean magnitude. This was repeated until subjects failed to recover their balance with a single step for two consecutive trials at the same lean magnitude.

Subjects were held in the forward-leaning posture using a lean control cable (Figure 3-1). One end of the lean control cable was attached to the back of a belt worn by subjects, and the other end was held in a releasable clasp. To release subjects, the investigator activated a solenoid that released the clasp. Lean magnitude was quantified by measuring tension in the lean control cable (units of percent body weight (BW)) with

an in-line load cell (Cooper, Warrenton, Virginia). Lean magnitude was controlled by requiring subjects to maintain a consistent foot position and adjusting the length of the lean control cable. This method of measuring lean magnitude, as opposed to measuring lean angle, provides a consistent method to measure forward lean across subjects of different sizes, can be measured directly during testing, and does not require that body segments be exactly in-line as for angle measurements.

To prevent falls to the ground in the event of an unsuccessful trip recovery, all subjects wore a full-torso harness that was tethered to a ceiling-mounted track using a fall-prevention lanyard. The length of the lanyard was adjusted so that when subjects reached for the ground, there was approximately 2 inches between their fingertips and the ground. This prevented any part of the subject's body from touching the ground except the feet. A load cell (Cooper, Warrenton, Virginia) was mounted in-line with the lanyard to measure force applied to the subject through the harness.

To start each trial, subjects stood with their feet at a toe line and leaned forward to a position where their body was straight except for dorsiflexion at the ankles. Heels were required to touch the ground and subjects were asked to distribute their weight equally across both feet. Subjects were asked to fold their arms across their chest throughout each trial. Once they were in position at the correct lean magnitude, subjects were verbally reminded to recover with their right LL. Subjects were then released unexpectedly 0-10 seconds after the completion of this statement. The initial lean magnitude corresponded to 12% BW in the lean control cable. Subjects were allowed two attempts at each lean magnitude, and lean magnitude was increased by 4% BW after each successful recovery until the subject failed to recover their balance with a single

step. Failure could occur three ways: 1) when the left (non-stepping) foot took a step longer than 30% of the subject's body height (BH), 2) when more than one step was taken by the right LL, and 3) when a subject put more than 30% BW on the harness at any point during trip recovery. A step with the left foot of less than 30% BH was allowed to aid in lateral stability, while a step larger than this was assumed to contribute to trip recovery. The 30% BW threshold in the harness distinguished trials in which the subject would not have recovered their balance if they were not wearing a harness. These failure criteria were defined by Wojcik et al. (1999).

All subjects practiced single step balance recovery prior to the start of the experiment. Music was played throughout the trials to conceal the noises associated with data collection.

Data Collection

Body segment position data were sampled at 200 Hertz using an Optotrak optoelectronic motion analysis system (Optotrack, Waterloo, Ontario). Infrared markers were placed on the right side of the body at the fifth metatarsal head, heel, lateral malleolus, lateral femoral epicondyl, greater trochanter, and acromion. Body segment position data were filtered with a 7 Hertz low-pass, zero-phase-shift Butterworth filter. Subjects initially stood with one foot on each of two forceplates (AMTI, Watertown, Massachusetts). Upon release, subjects stepped onto a third forceplate (Bertec, Columbus, Ohio) with the right LL. This forceplate was oversized (90 x 90 centimeters) so that subjects did not need to alter their natural stepping patterns to achieve proper foot placement on the forceplate. Forceplate data were sampled at 1000 Hertz and filtered with a 100 Hertz low-pass, zero-phase-shift Butterworth filter. Harness load cell data

were sampled at 1000 Hertz and filtered with a 10 Hertz low-pass, zero-phase-shift Butterworth filter. All trials were recorded on videotape for post hoc confirmation of trip recovery and/or failure criteria.

Data Analysis

A link-segment model was implemented to model the human body. The segments included the foot, shank, and thigh of the stepping LL as well as a single segment representing the head, arms, and torso (HAT). Mass and inertial characteristics of the body segments were defined using an anthropometric model (Hanavan, 1964). Joints were represented as frictionless ball and socket joints. Other assumptions are made in this model, including that each LL segment has a fixed mass that can be represented with a center of mass, the center of mass locations remain constant, and the moment of inertia of each segment remains constant. The model was used to perform an inverse dynamics analysis as described by Winter (1990). Only peak extensor torques after stepping are reported here. Peak extensor torques were normalized by height (meters) and mass (kilograms) to account for anthropometric differences in subjects.

A two-way repeated measures ANOVA was used to test for the effects of lean magnitude and age on peak extensor torques. The independent measures were lean magnitude and age. The dependent measures were peak extensor torques at the ankle, knee, and hip after stepping to recover balance. A significant lean magnitude effect was followed by a one-way repeated measures ANOVA of each age group separately. Pairwise comparisons were performed using least square means comparison to find lean magnitudes that had significantly different peak extensor torques after stepping.

IV. Results

Young subjects were able to achieve a maximum lean magnitude of $34.4 \pm 5.06\%$ BW, compared to older subjects who achieved $20.0 \pm 6.93\%$ BW. This indicates that the young subjects have an improved ability to recover from a trip, as expected, compared to older subjects. The percentage of each age group succeeding at each lean magnitude is shown in Figure 3-2.

Peak extensor torques at the ankle, knee, and hip are shown in Figures 3-3, 3-4, and 3-5 respectively. Lean magnitude was found to have significant effects on peak extensor torques at the ankle and hip. Peak plantarflexor torque increased with increasing lean magnitude for young subjects. Peak hip extensor torque increased with increasing lean magnitude for both young and older subjects.

Pairs of lean magnitudes with significantly different peak extensor torques are listed in Table 3-2. Young subjects' peak ankle plantarflexor torque in the first two lean magnitudes are significantly less than peak ankle plantarflexor torques generated in the remainder of the lean magnitudes. Peak ankle plantarflexor torques of young subjects therefore initially increase at low lean magnitudes but do not continue to increase after the 16% BW lean magnitude. Peak hip extensor torques for young subjects continued to increase up to the 28% BW lean magnitude, after which no significant differences were found. Older subjects showed significant differences between the peak hip extensor torque at the first lean magnitude (12% BW) and the remaining lean magnitudes (16% - 24% BW).

Age effects were found for the knee and hip extensor torques. Age differences indicated that young subjects used more peak knee extensor torque and less hip extensor torque than the older subjects. Age effects were not significant at the ankle.

Several trials were not included in the analysis due to problematic data. Eight trials were missing marker data. Of the joint torques calculated, all older subjects and most young subjects used ankle plantarflexor, knee extensor, and hip extensor torques immediately following stepping from a simulated trip. However, a small percentage of trials (9.6%) did not exhibit an extensor torque at the hip. These data were therefore not included in the analyses because they would skew the data from the majority of the subjects. One older subject could not recover from the minimum lean magnitude, and as a result could not be included in the analysis. Analysis was performed at lean magnitudes with a minimum of two data points.

V. Discussion

It was hypothesized that peak extensor torques at the ankle, knee, and hip after stepping would increase as trip severity increased. Plantarflexor torques at the ankles of young subjects and extensor torques in the hips of young and older subjects agree with this hypothesis. It was also hypothesized that differences in peak extensor torques would exist between young and older men. Differences in knee and hip extensor torques between young and older subjects agree with this hypothesis.

Increases in peak extensor torques with increases in lean magnitude indicate that more strength is needed to recover balance as trip severity increases. Young subjects generated a significantly smaller ankle torque in the first two lean magnitudes than in the remaining lean magnitudes, indicating that they may improve their response within a few trials to a level at which ankle torques do not significantly differ from each other. Young subjects generated continually increasing peak hip extensor torques as lean magnitude increased. Older subjects did not continually increase their hip extensor torques nor

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reach the hip extensor torques that young subjects generated during lean magnitudes for which older subjects did not recover. Not continuing to increase peak hip extensor torque as young subjects did may contribute to LL buckling during trip recovery in older adults. The highest peak hip extensor torque for older subjects in comparison to those generated by younger subjects at larger lean magnitudes also suggests that older adults may be limited in their trip recovery capability by an inability to generate the necessary amount of torque to prevent LL buckling during more severe trips.

Age-related differences in joint extensor torques indicate that older subjects generate less knee and more hip extensor torque than young subjects. Ankle plantarflexor torques were comparable between the two age groups. The smaller knee extensor torque generated by older subjects is consistent with Pavol et al.'s (2001) suggestion that poor knee extensor torque may cause buckling of the stepping limb. However, increased hip extensor torque may counter the decreased knee extensor torque in preventing buckling in the stepping limb. The smaller knee extensor torques and larger hip extensor torques in older subjects are also consistent with differences in joint torque during gait between young and older subjects (DeVita and Hortobagyi 2000).

Comparing peak joint torques after stepping with peak joint torques before and during stepping (Wojcik et al. 2001) indicate that ankle plantarflexor and knee extensor torques are higher after stepping while hip extensor torques appear to be similar. Wojcik et al. (2001) concluded muscle strength did not contribute to limitations in trip recovery for male subjects. However, this conclusion was based on torque measurements before and during stepping and not after stepping, as in the present study. The higher ankle and

knee torques measured in the present study indicate that muscle strength may be more important than concluded by Wojcik et al. (2001).

To determine if subjects used joint torques near their maximum joint torque capability, the peak extensor torques measured in this study were compared to the muscle strength measurements in the literature. Only peak hip extensor torques in young subjects appear to be limited by muscle strength. Either isometric or low speed isokinetic muscle strength measurements were used. Available ankle plantarflexion torque (Wojcik 1997) does not appear to limit balance recovery in young or old subjects, as they use 53.0% and 63.8% of their maximum available plantarflexion torque at most, respectively. Available knee extensor torque (Knapik et al. (1983), Clark et al. (1997)) also does not appear to limit balance recovery in young or old subjects, as they use 75.4% and 62.1% of their maximum available torque at most, respectively. The graphs of ankle and knee extensor torques (Figures 3-3 and 3-4) also support that available torque does not limit trip recovery as the torques do not consistently increase with increasing lean magnitude. This same pattern also occurs with hip extensor torques (Cahalan et al.1989) for the older subjects, as they only used 59.3% of the torque estimated to be available to them (Figure 3-5). Young subjects show a continual increase in hip extensor torque as lean magnitude increases and use 83.9% of the estimated torque available to them, indicating that the maximum hip extensor torque they can generate may limit their trip recovery capability. It is difficult to make strong conclusions from these comparisons because strength was not measured in the study participants, and because of the large variability of muscle strength data reported in the literature. This comparison could best be made in a future

study where maximum joint torque was measured for each subject prior to the trip protocol.

This study had several limitations. Subjects did not have an initial forward momentum at the start of the protocol, although actual trips occur during walking when subjects have a forward momentum proportional to their walking speed. Some subjects may have been more motivated than others and had a greater fear of falling. Younger individuals tend to recover from a balance perturbation with a single step while older individuals tend to use multiple steps (Luchies et al. 1994, McIlroy and Maki 1996). Constraining trip recovery to a single step may force older subjects to use an unpreferred strategy. However, constraining recovery method is necessary to compare kinetic data across age groups. All torque calculations were limited to the sagittal plane, which may have slightly altered peak values. Torque values were calculated with estimates of segment masses, center of mass locations, and moments of inertia. Only male subjects participated in this study, so results may not apply to women.

Future research comparing the temporal location of peak joint torques after stepping may reveal more about differences in trip recovery between young and older subjects, particularly if joint torque generation rate is a cause of decreased trip recovery capability. A trip recovery study that directly measures maximum available torque at the ankle, knee, and hip for each subject would better test if peak extensor torques limit trip recovery. An exercise program that increases knee and hip extensor torque, also suggested by Pavol et al. (2001) for solely the knee extensors, combined with using the trip protocol implemented in this study before and after the exercise program would show if increases in these torques increase trip recovery capability.

VI. Conclusions

Our results suggest increasing trip severity requires larger extensor torques after stepping to prevent LL buckling. Age-related reductions in extensor strength may limit these torques and contribute to higher trip-related fall rates in the elderly. However, comparing measured torques to strength estimates from the literature did not support this. An age-related difference in joint torques indicates that older men may use a different strategy to prevent LL buckling after a trip. These differences in joint torques are consistent with age-related differences in joint torques during gait.

VII. Sources

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Table 3-1. Mean (standard deviation) of subject age, height, and weight.

	Young	Old
N=	10	10
Age (yr)	20.6(1.26)	74.0(6.50)
Height (m)	1.76(0.068)	1.71(0.058)
Mass (kg)	71.3(14.0)	77.0(14.0)

Table 3-2. Lean magnitudes with statistically different peak joint torques.

Joint	Age	Minimum Lean Magnitude	Other Magnitudes
Ankle	Young	12	20*, 24, 28**, 32**, 36, 40*
		16	20, 28*, 32*, 40
Hip	Young	12	24, 28**, 32***, 36***, 40***
		16	24, 28**, 32***, 36***, 40**
		20	28*, 32**, 36**, 40**
		24	28, 32*, 36*, 40*
		28	40
	Old	12	16*, 20*, 24

* p<0.01

** p<0.001

***p<0.0001
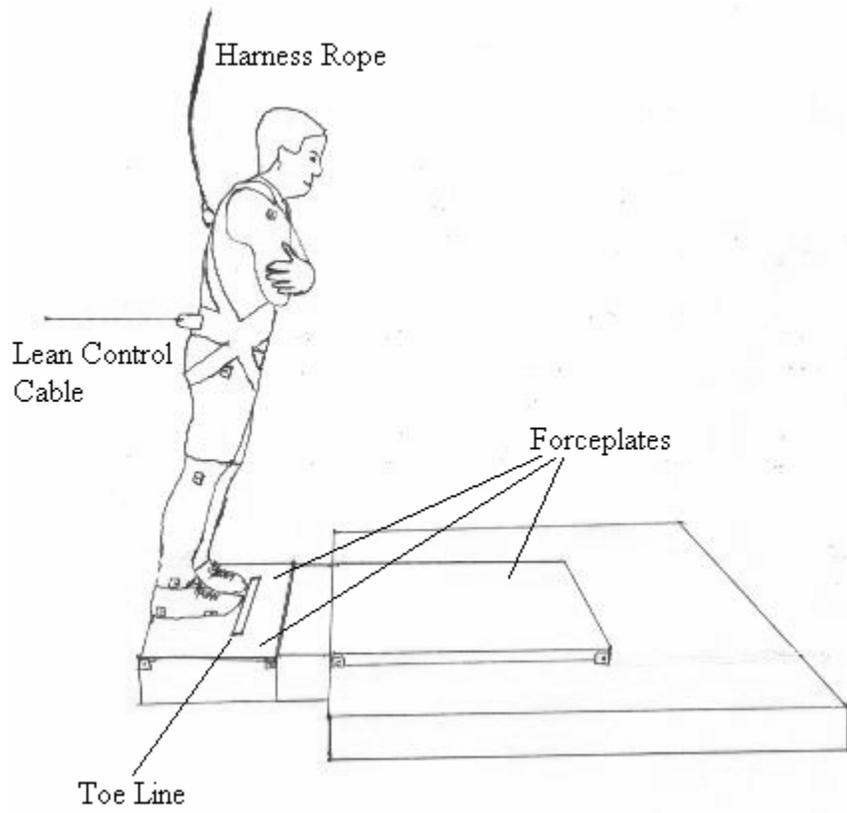
Figure 3-1. Subject ready to be released.
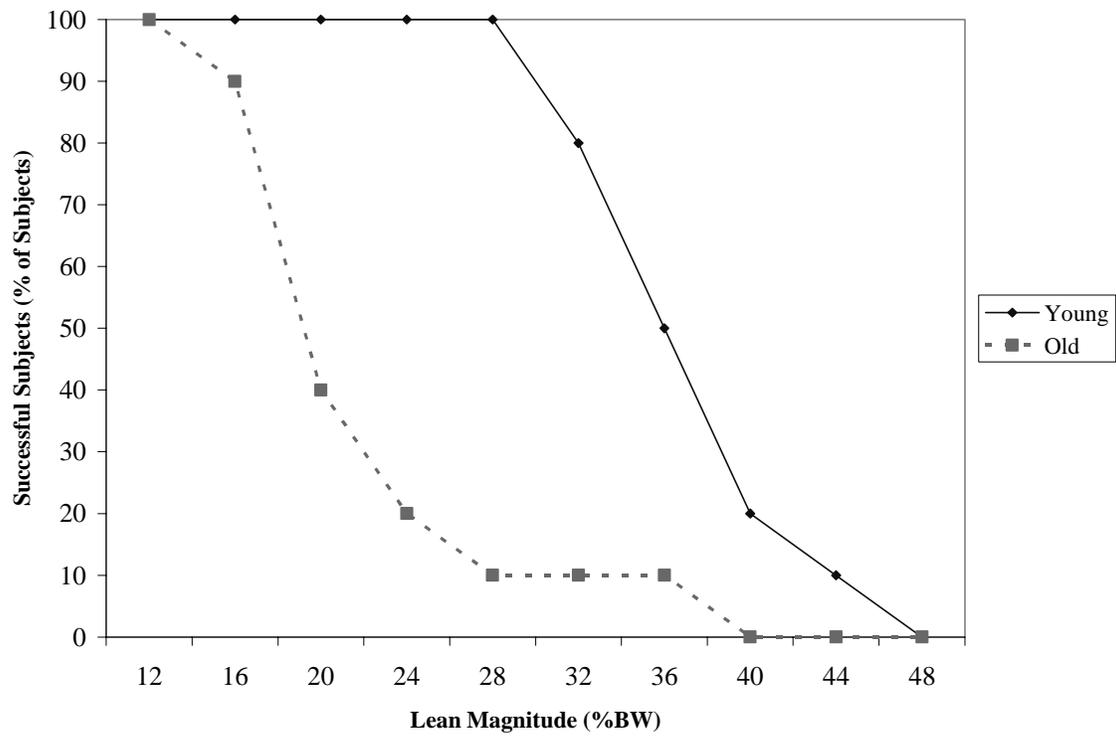
Figure 3-2. Percentage of subjects succeeding at each lean magnitude..
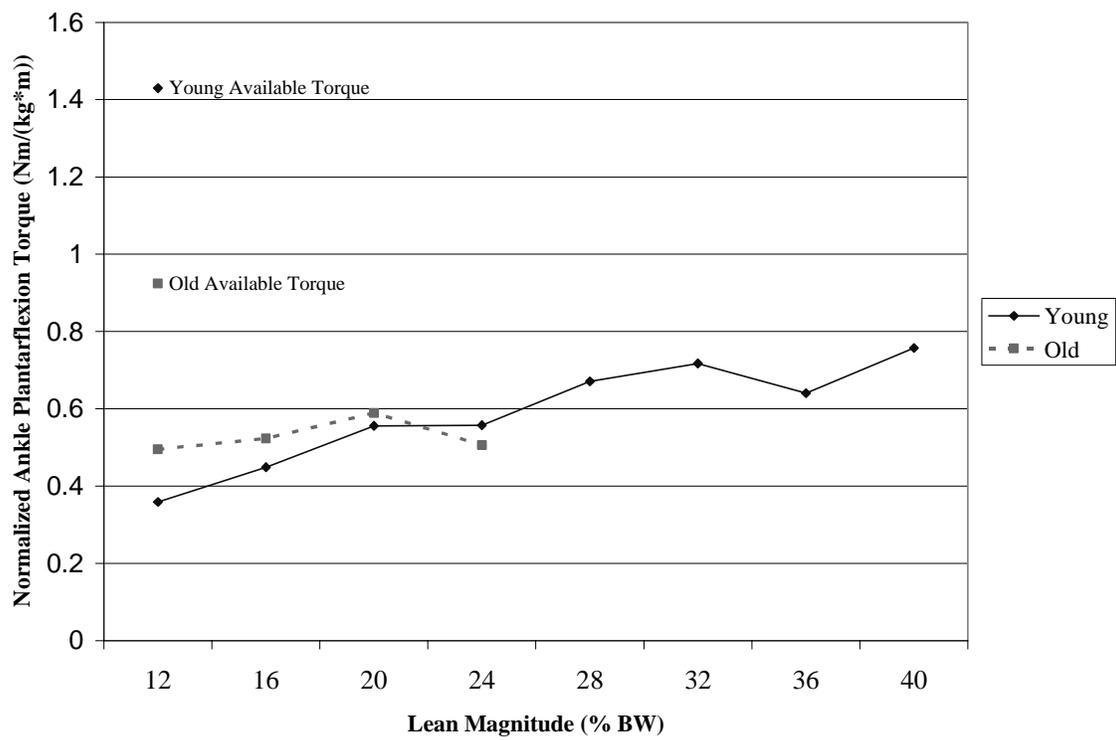
Figure 3-3. Peak ankle plantarflexion torques after stepping.
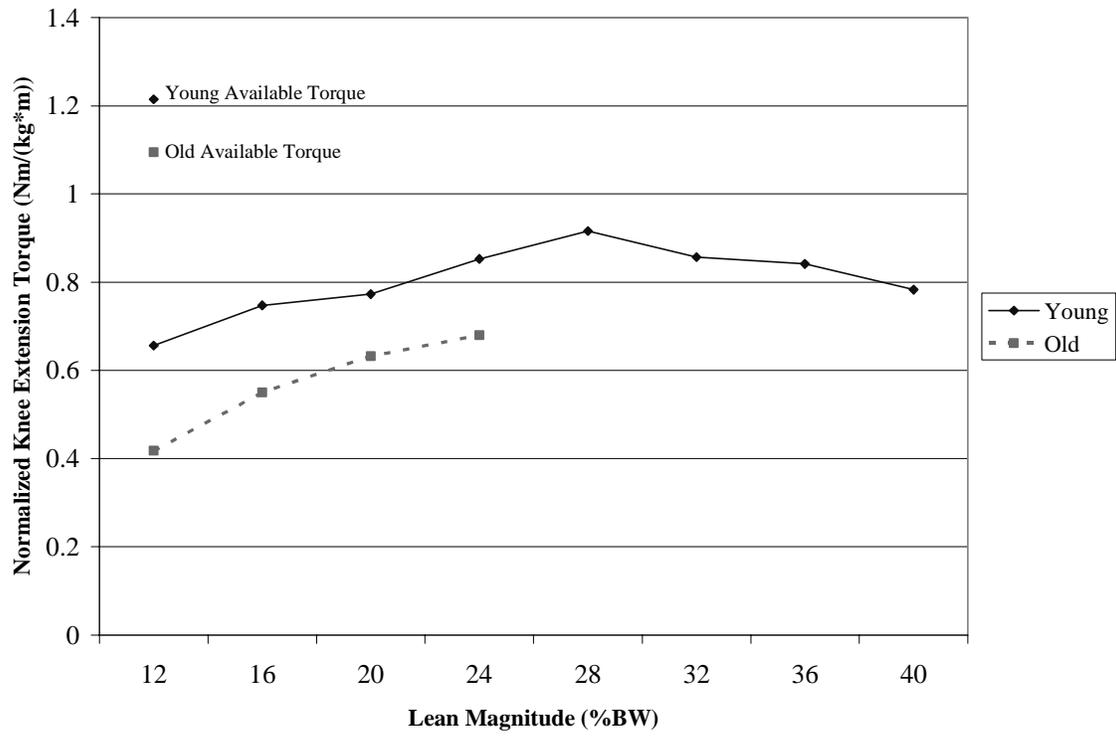
Figure 3-4. Peak knee extensor torques after stepping.
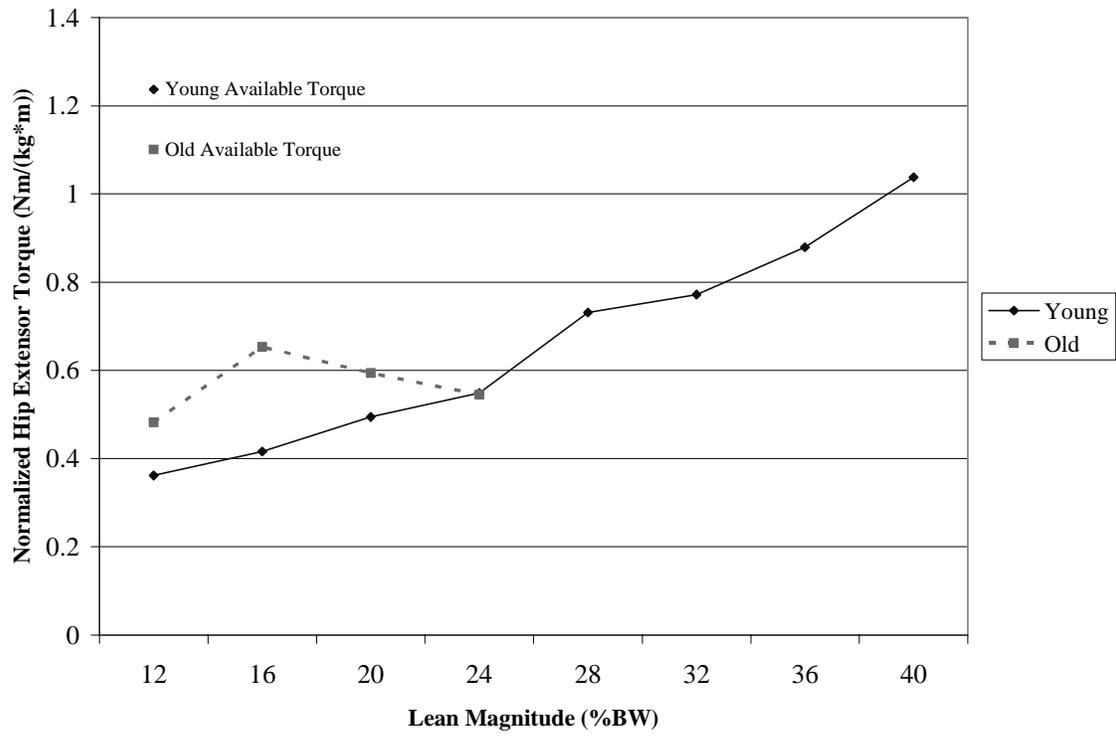
Figure 3-5. Peak hip extensor torques after stepping.

Appendix A: Data Collection Information

Details of equipment position and settings, data collected, and the preliminary screening of older subjects are included here.

A1. Laboratory Equipment Positioning
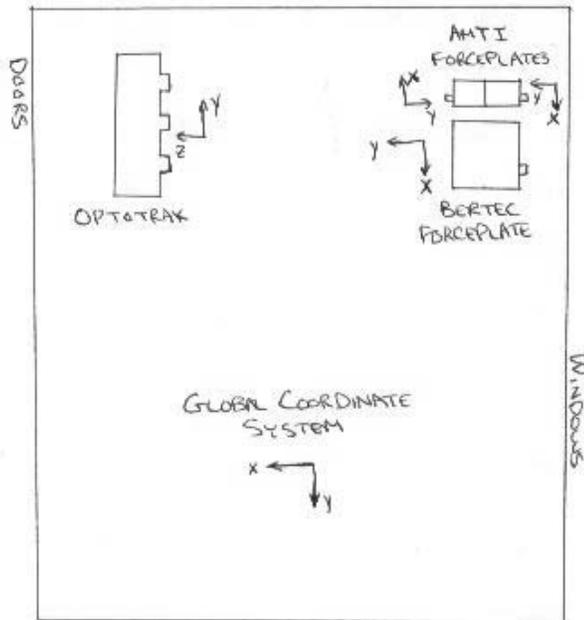
Figure A1-1. Equipment positions for testing the right lower limb.
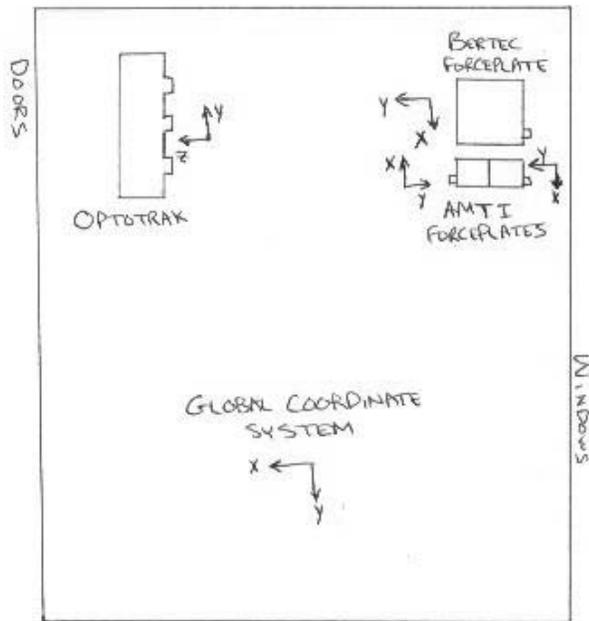
Figure A1-2. Equipment positions for testing the left lower limb.

A2. Data Collection Form

BALANCE SYMMETRY STUDY

Setup Checklist

1. Turn on:
     - control unit
     - position sensor
     - ODAU
     - force platform amplifiers
     - load cell amplifiers
     - power supply to solenoid
2. Create computer directory with subject initials and date (i.e. *mm021002*)
3. Run COLLECT from this directory
4. Setup data acquisition parameters in COLLECT

ODAU file collection parameters
length of collection: 5
data file trial #: 1
data file extension: .sym

OPTOTRAK collection parameters
frame rate: 200
port 1: 6 markers
port 2: 4 markers

ODAU collection parameters
Frame rate: 1000
Num analog input chan: 16
Gain: 1
Input mode: disabled
Input mode: single
Scan freq: 20000

OPTOTRAK file collection parameters
length of data collection: 5 sec
data file trial #: 1
data file extension: .sym

OPTOTRAK system parameters
marker frequency: 2100
duty cycle: 70
strober voltage: 8
dynamic duty cycle: on

5. Setup video camera
6. Get out supplies
     - double-sided adhesives
     - athletic tape
     - ACE bandages
     - razor
     - lean control belt
     - harness
     - measuring tape
     - IR diodes
7. Check placement of IR diodes on force platforms
8. Sweep off force platforms

BALANCE SYMMETRY STUDY

Data Collection Sheet

Subject name:_____ Date:_____Start time:_____

   1. Make sure subject is wearing proper clothing (tank-top shirt, bicycle shorts, low-top shoes, socks)
   2. Give summary of protocol and demonstrate balance recovery movement
   3. Obtain informed consent
   4. Turn on music
   5. Collect anthropometric measurements: age: _____

               handed (verbal):_____   gender: _____

               footed (verbal):_____   weight: _____lbs

                     stature: _____cm   sitting height: _____cm

           R. ankle circum: _____cm   R. sphyrion height: _____cm

           R. knee circum: _____cm   R. tibiale height: _____cm

           R. thigh circum: _____cm   R. trochanteric height:_____cm

            R.foot length: _____cm   L. trochanteric height:_____cm

   6. Ask subject to empty pockets
   7. Ask subject if they need to use the restroom
   8. Mark toe line on Donald and Willy, and mark 30% body length line on Bertha
   9. Calculate lean magnitude voltages using Excel spreadsheet
   10. Don lean control cable belt and harness
   11. Ask subject to practice balance recovery and record foot used: (R / L)
   12. Select lean control cable height # _____
   13. Measure height of lean control cable at subject's back: _____cm
   14. Apply IR diodes
        1   5th metatarsal head
        2   heel
        3   lateral malleolus
        4   lateral femoral epicondyle
        5   greater trochanter
        6   acromion
       use tape, ACE wraps, and twisty ties to secure all leads

   15. Apply masking tape to dark shoes
   16. Check ROM and adjust leads if necessary
   17. Zero bertha

18. Close window blinds
19. Turn on 'real-time' data collection to check marker signals
20. Start video recording   Tape # _____   Counter # _____
21. Collect file of subject in anatomical position while standing on Bertha
22. Perform balance recovery trials using a 12% initial lean magnitude and 4% lean increment until subject fails twice at a single lean magnitude

Record which leg is being used to step: (R / L)

File number	Task	Comments
	Anatomical position	

23. Deactivate markers
24. Stop video recording
25. Offer subject a chair to rest
26. Remove weights from lean cable structure
27. Disconnect markers 1 and 2 from force platform strober
28. Move Bertha to opposite side of Willy and Donald and place spacer between them
29. Re-connect markers 1 and 2 from force platform strober
30. Move lean cable structure
31. Replace weights on lean cable structure
32. Remove and reapply IR diodes

    1   5th metatarsal head
    2   heel
    3   lateral malleolus
    4   lateral femoral epicondyle
    5   greater trochanter
    6   acromion

use tape, ACE wraps, and twisty ties to secure all leads

33. Check ROM and adjust leads if necessary
34. Re-apply tape to mark toe line on Donald and Willy, and mark 30% body length on Bertha
35. Turn on 'real-time' data collection to check marker signals
36. Check viewing area of camcorder
37. Start video recording
38. Collect file of subject in anatomical position while standing on Bertha
39. Perform balance recovery trials using a 12% initial lean magnitude and 4% lean increment until subject fails twice at a single lean magnitude

Record which leg is being used to step: (R / L)

File number	Task	Comments
	Anatomical position	

40. Deactivate markers
41. Stop video recording
42. Record end time :_____
43. Remove all markers and equipment from subject. Thank them, and see them out the door.
44. Measure distance from floor to bottom of lanyard.   _____ cm
45. Convert data files
     Convert 'R' data files to 'C' data files.
     Convert 'O' data files to 'V' data files
     Exit COLLECT and enter DISPLAY
     Convert 'C' files to 'KIN' files
     Convert 'V' files to 'GRF' files
     Exit DISPLAY and enter Windows Explorer
     Create subdirectories 'rawdata' and 'textdata'
     Move 'R' and 'O' files into 'rawdata' directory
     Move 'KIN' and 'GRF' files into 'textdata' directory
     Delete 'C' and 'V' files
46. Turn off system:
     - control unit
     - position sensor
     - ODAU
     - force platform amplifiers

- load cell amplifiers
- power supply to solenoid

A3. Preliminary Health Screening Form for Older Adults

SUBJECT SCREENING FORM

Name: _____

Phone: _____

Email: _____

Sex: Male   Female

Age: _____

Height: _____

Weight: _____

Subject screening questions

(Y / N) 1.   Do you have any medical condition that interferes significantly with your daily activities?
(Y / N) 2.   Do you fall down a lot?
(Y / N) 3.   Have you had any type of joint replacement or joint fusion?
(Y / N) 4.   Within the last 6 months, have you had a severe head injury, concussion, or been "knocked out"?
(Y / N) 5.   Have you ever had problems with dizziness or loss of balance that seemed to occur frequently or lasted for an extended period of time?
(Y / N) 6.   Within the last 6 months, have you had an ear infection or drainage from the ear?
(Y / N) 7.   Do you have any trouble with your vision not correctable by glasses?
(Y / N) 8.   Do you have any problems caused by arthritis, muscle problems, or broken bones, etc. that limit your ability to walk or bend your joints?
(Y / N) 9.   Within the last 6 months, have you had any arm or leg injuries?
(Y / N) 10.   Have you been diagnosed with or are being treated for osteoporosis?
(Y / N) 11.   Are you being treated with corticosteriods for any condition?
(Y / N) 12.   Are both of your legs the same length?

13.   How many days a week do you participate in physical activities?
       0-1 days / 2-4 days / 5-7 days. (Circle answer)

Appendix B: Matlab Code for Study #1

Matlab code was used to convert the text data into matlab format (B1), find the maximum harness force to separate successful trials from Harness Failures (B2 and B3), find the kinematic times of interest (B4), and find the step length and velocity (B5). Some modifications of the routine in Appendix B4 were made for specific files to aid the program in finding the correct piece of force plate data and therefore the correct time values. The numerical output from the code in Appendices B4 and B5 were verified with the graphs produced by these programs.

B1. Optoreader

Program created by Dr. Madigan to convert text data to Matlab data for easy manipulation in Matlab.

```
% optoreader.m
% This program is designed to convert files exported from Optotrak
% DISPLAY software into MatLab Files

basefilename=input('Input Base File Name: ','s');
extfilename=input('Input File Extension: .','s');
ftrial=input('Input First File #: ');
ltrial=input('Input Last File #: ');

for cfn=ftrial:ltrial,
   clear rawdata header col_header hdr hdrline;
   if cfn<10,
      scfn=['00' num2str(cfn)];
   elseif cfn>9 & cfn<100,
      scfn=['0' num2str(cfn)];
   end;
   current_file_name = [basefilename scfn '.' extfilename];


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%
         % read in header information

   fid=fopen(current_file_name);
   if fid ==-1, error('File could not be read.');end;
   for i = 1:20,
      hdrline=fgetl(fid);
      hdr(i,:)=cellstr(hdrline);
   end;
   header=char(hdr);
   clear hdr hdrline;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%
         % parse the header

  k= findstr(',',header(20,:)); %determine # of columns in comma delimited file
         n_col=length(k)+1;
         for i=1:n_col,
     if i==1;
        X=char(header(20,1:k(i)-1));
```

```
        elseif i==n_col;
            X=char(header(20,k(i-1)+1:length(header)));
        else
            X=char(header(20,k(i-1)+1:k(i)-1));
        end;
        col_header(i)=cellstr(X);
    end;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%
            % read in data

    fprintf('Reading in data from file %2.0f...',cfn);
    f_error=0;
            i=1;
            while f_error==0,
    eval('rawdata(:,i)=fscanf(fid,"%f,");','f_error=1;');
    i=i+1;
            end;
    rawdata=rawdata';
    fclose(fid);


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%
    % interpolates to remove missing data points

    fprintf('  Interpolating missing data points...\n');
    for col=1:n_col-1,
        for row = 2:length(rawdata),
            if abs(rawdata(row,col))>10000,
                rawdata(row,col)=rawdata(row-1,col);
            end;
        end;
    end;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%
    % save data

    save_file_name = [basefilename scfn '.mat'];
    save(save_file_name,'rawdata','header','col_header'); %write data to file

end; % end of cfn loop that loops through data files

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%
% clear extraneous variables

clear X basefilename cfn col current_file_name extfilename f_error fid
clear ftrial i k ltrial n_col row save_file_name scfn
```

B2. Maximum Harness Force

Program to find the maximum force a subject put on the harness for each trial. Output from this program was used to determine if a subject passed or failed a trial that was noted as passing (no Multistep Failure) during data collection.

```
% HarnessMaxForce.m
% Program to find the maximum force (lb) from the load cell that the harness attaches to
% Input a data file (harness data is column 14 of data, column 16 with frame numbers) and subject weight
% Output if the subject load cell held over 30% of the subject's weight
% Started 5/13/02, Emily Lloyd
% Last Revised: 7/2/02, Emily Lloyd

% Input

subject_weight = input ('Input subject weight in pounds: ');


basefilename=input('Input Subject''s Initials: ','s');
ftrial=input('Input First File #: ');
ltrial=input('Input Last File #: ');

for cfn=ftrial:ltrial,                                        % from optoreader program
   clear rawdata header col_header hdr hdrline;
   if cfn<10,
      scfn=['00' num2str(cfn)];
   elseif cfn>9 & cfn<100,
      scfn=['0' num2str(cfn)];
   end;
   current_file_name = [basefilename 'grf$#' scfn '.mat'];

load(current_file_name, 'rawdata');

% get files

harness_voltage = rawdata(:,16);                 % harness_voltage (channel 14 (0-15) of analog data)
start_data = rawdata(:,17);                       % solenoid signal (channel 15 (0-15) of analog data)
frame = rawdata(:,1);                             % frame number (put in front of analog data)

% Calculations to find start time

time = frame.*0.001;

start_frame = 0;
started = 0;

counter=1;

while started==0
  if start_data(counter)>1
    started=1;
    start_frame = counter;
  elseif counter==5000
    fprintf('Solenoid was not released - no start time.\n')
    start_frame = 10000;
    started=1;
  end
  counter = counter + 1;
end
```

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```
start_time = start_frame*0.001;

% Get rid of occasions when raw harness_voltage data = 10

for counter=1:5000
  if harness_voltage(counter)>9.99                          % ==10?
    harness_voltage(counter)=(harness_voltage(counter-1)+harness_voltage(counter+1))/2;
  end
end

% Filter data

harness_voltage = zerophaseshift_filter_10Hz(harness_voltage);



% Calculations for 30% method

harness_force = 14.86*harness_voltage - 1.25;
max_acceptable_force = 0.3*subject_weight;

% Set beginning of data equal to 0 lbs

total=0;
                            % Find average force in second 0.25 sec
for counter=250:500
  total=total + harness_force(counter);
end
harness_average = total/250;

harness_force = harness_force - harness_average;  % Set harness_force to zero initially


max_force = 0;

for counter=50:4950                      % Exclude the beginning and end, where the force changes because of
filtering
  if harness_force(counter) > max_force
    max_force = harness_force(counter);
  end
end

max_percentage = max_force./subject_weight * 100;



% Output file name

fprintf(current_file_name)

% Output for 30% method

if max_force > max_acceptable_force
  fprintf(' - FAIL - The maximum harness force, %5.2f lb, is greater than the maximum allowable force %5.2f
lb.\n',max_force,max_acceptable_force)
  fprintf('The maximum percentage of harness force to body weight is %5.2f \n',max_percentage)
end
```

B3. Low-pass zero-phase-shift Butterworth Filter with a 10 Hz cutoff Frequency

This function was called by the previous program HarnessMaxForce.m(Appendix B2) to filter the harness data. An initial filtering function written by Dr. Madigan was revised to fit various cutoff frequencies throughout this research.

```
function filtered_data = zerophaseshift_filter_10Hz(data_to_be_filtered);

% this function lowpass filters kinematic data using a 4th order
% zero-phase-shift filter.  This involves using a 2nd order filter
% forward then reverse through the data.
% see Winter DA. Biomechanics and Motor Control of Human Movement. Pg 36-43.

sampling_rate=1000;
fc=10;
corrected_fc = fc/.802;  %corrects the fc since filtering data twice affects original fc
[filternum,filterdenom] = butter(2,corrected_fc/(sampling_rate/2));
reversed_data_to_be_filtered=data_to_be_filtered(length(data_to_be_filtered):-1:1);
f1 = filter(filternum,filterdenom,data_to_be_filtered);
f2 = f1(length(f1):-1:1);
f3 = filter(filternum,filterdenom,f2);
filtered_data = f3(length(f3):-1:1);
```

B4. Kinematic Time Values

This program uses force plate and solenoid data to compute the different kinematic times examined in this study. Start points of each event (release, reaction, lift off, and landing) (defined by Thelen et al. 1997) were also output. Figure 2-2 illustrates an example of the graphical output from this program used to verify the numbers computed.

```
% TripTimes.m
% Program to find the reaction time, weight transfer time, and step time
% Input a data file (grf)
% Output the times
% Started 5/22/02, Emily Lloyd
% Last Revised: 6/21/02, Emily Lloyd

% Input

basefilename=input('Input Subject''s Initials: ','s');
cfn=input('Input File #: ');


% from optoreader program
   clear rawdata header col_header hdr hdrline;
   if cfn<10,
      scfn=['00' num2str(cfn)];
   elseif cfn>9 & cfn<100,
      scfn=['0' num2str(cfn)];
   end;
   current_file_name = [basefilename 'grf$#' scfn '.mat'];


load(current_file_name, 'rawdata');

% Choose Units

Units=input('Press 1 for SI units, 2 for English Units: ');

if Units==2
   fprintf('Please note that this option is for viewing the plot only.  Times noted here are only estimates.\n')
end

% get files

DSFx = rawdata(:,8);                    % Donald's SFx (Signal in the Fx direction) (channel 6 (0-15) of
analog data)
DSFy = rawdata(:,9);                    % Donald's SFy (Signal in the Fy direction) (channel 7 (0-15) of
analog data)
DSFz = rawdata(:,10);                   % Donald's SFz (Signal in the Fz direction) (channel 8 (0-15) of
analog data)
DSMx = rawdata(:,11);                   % Donald's SMx (Signal in the Mx direction) (channel 9 (0-15) of
analog data)
DSMy = rawdata(:,12);                   % Donald's SMy (Signal in the My direction) (channel 10 (0-15) of
analog data)
DSMz = rawdata(:,13);                   % Donald's SMz (Signal in the Mz direction) (channel 11 (0-15) of
analog data)

BSFx = rawdata(:,2);                    % Bertha's SFx (Signal in the Fx direction) (channel 0 (0-15) of
analog data)
BSFy = rawdata(:,3);                    % Bertha's SFy (Signal in the Fy direction) (channel 1 (0-15) of
analog data)
```

```
BSFz = rawdata(:,4);                              % Bertha's SFz (Signal in the Fz direction) (channel 2 (0-15) of
analog data)
BSMx = rawdata(:,5);                              % Bertha's SMx (Signal in the Mx direction) (channel 3 (0-15) of
analog data)
BSMy = rawdata(:,6);                              % Bertha's SMy (Signal in the My direction) (channel 4 (0-15) of
analog data)
BSMz = rawdata(:,7);                              % Bertha's SMz (Signal in the Mz direction) (channel 5 (0-15) of
analog data)

%harness_voltage = rawdata(:,16);                % harness_voltage (channel 14 (0-15) of analog data)
solenoid = rawdata(:,17);                         % solenoid signal (channel 15 (0-15) of analog data)
frame = rawdata(:,1);                             % frame number (put in front of analog data)

% Account for gain settings

bertha_gains = [10 10 2 5 5 5];

BSFx=BSFx./bertha_gains(1);
BSFy=BSFy./bertha_gains(2);
BSFz=BSFz./bertha_gains(3);
BSMx=BSMx./bertha_gains(4);
BSMy=BSMy./bertha_gains(5);
BSMz=BSMz./bertha_gains(6);

donald_gains = [1996.6 1996.6 1996.6 1996.4 1997 4000]; % for Donald, from Amplifier Calibration Sheet

Vexcite = 4.991;                                  % Volts, nominal value of 5 V, from Donald's Amplifier Calibration
Sheet

CF = donald_gains*Vexcite*10^-6;                 % Conversion Factor, OR6-5 Manual, page 3

% Get forces and moments for Donald and Bertha using Calibration Matricies

DonaldC = [1.512 0.005 0.016 -0.002 -0.003 0.006;
   0.006 1.516 0.023 -0.002 -0.011 0.000;
   0.024 0.025 5.893 0.000 0.007 -0.001;
   0.000 0.000 0.000 0.604 -0.005 -0.003;
   0.000 0.000 0.000 0.001 0.605 -0.002;
   0.000 0.002 0.000 -0.001 -0.004 0.306];

BerthaC = [863.0 -11.2 -5.5 0.1 2.5 -3.3;
   22.6 855.5 -13.4 -7.5 6.0 -3.7;
   -37.3 -0.5 1805.6 -1.6 -11.2 -17.1;
   -3.4 -72.9 -2.6 908.3 -2.7 2.8;
   73.8 -3.3 0.1 -1.5 908.9 6.5;
   1.8 -5.0 1.4 3.8 3.3 302.1];

for counter=1:5000
   DonaldS = [DSFx(counter); DSFy(counter); DSFz(counter); DSMx(counter); DSMy(counter); DSMz(counter)];
   Donald = DonaldC * DonaldS;
   DFx(counter) = Donald(1)./CF(1);
   DFy(counter) = Donald(2)./CF(2);
   DFz(counter) = Donald(3)./CF(3);
   DMx(counter) = Donald(4)./CF(4);
   DMy(counter) = Donald(5)./CF(5);
   DMz(counter) = Donald(6)./CF(6);
end

for counter=1:5000
   BerthaS = [BSFx(counter); BSFy(counter); BSFz(counter); BSMx(counter); BSMy(counter); BSMz(counter)];
   Bertha = BerthaC * BerthaS;
   BFz(counter) = Bertha(3);
```

```
end

if Units==2
        BFz = BFz./4.44822;                                          % Changes Newtons to lbs
        DFz = DFz./4.44822;                                          % Changes Newtons to lbs
end

% Filter forces and moments

DFz = zerophaseshift_filter_100Hz(DFz);

BFz = zerophaseshift_filter_100Hz(BFz);

% Calculations to find start time

time = frame.*0.001;

start_index = 0;
started = 0;

counter=1;

while started==0
  if solenoid(counter)>1
    started=1;
    start_index = counter;
  elseif counter==5000
    fprintf('Solenoid was not released - no start time.\n')
    start_index = 10000;
    started=1;
  end
  counter = counter + 1;
end

start_time = frame(start_index)*0.001;

% Calculations to find Reaction Time

DFzBaseline = DFz(250);
epsilon = 30;
past_point=0;
minimum = DFzBaseline;
minimum_index = 0;
for counter=start_index:1:(start_index+200)
  if (DFz(counter)<minimum) & (past_point==0)
    minimum = DFz(counter);
    minimum_index = counter;
  end
  if DFz(counter)>(DFzBaseline+epsilon) & (past_point==0)
    past_point = 1;
  end
end

Reaction = frame(minimum_index)*0.001;

Reaction_Time = Reaction - start_time;
% plot(time, DFz,'b',time, BFz,'g',time, solenoid*10,'r')
% Calculations to find Weight Transfer Time

DFz_final = 0;
DFz_total = 0;
```

```
for i=4500:4749                % CHANGED TO 3500:3749 WHEN SUBJECT MOVES BACK TO DONALD
BEFORE THE END OF THE 5 SECONDS
   DFz_total = DFz(i) + DFz_total;
end
DFz_final = DFz_total/250;

minimum = DFz_final;
minimum_counter2 = 0;
lift_off_found = 0;
for counter=minimum_index:(minimum_index+400)
  if (DFz(counter)<=minimum) & (lift_off_found==0)
    minimum = DFz(counter);
    minimum_index2 = counter;
    lift_off_found = 1;
  end
end

Lift_off = frame(minimum_index2)*0.001;

Weight_Transfer_Time = Lift_off - Reaction;

% Calculations to find Step Time

BFz_initial = 0;
BFz_total = 0;
for i=250:499
   BFz_total = BFz(i) + BFz_total;
end
BFz_initial = BFz_total/250;

counter = 1;
found=0;
while found==0
  if BFz(counter)>(BFz_initial+10)
    found=1;
    landing_index=counter;
  elseif counter == 5000
    fprintf('No force on Bertha found.\n')
    found = 1;
  else counter = counter+1;
  end
end

Landing = frame(landing_index) * 0.001;

Step_Time = Landing - Lift_off;

Landing_Time = Landing - start_time;

% Output

start_time
Reaction
Reaction_Time
Lift_off
Weight_Transfer_Time
Landing
Step_Time
Landing_Time
if Units==1
  plot(time, DFz,'b',time, BFz,'g',time, solenoid*10,'r')
  legend('Donald','Bertha','Solenoid')
```

```
    ylabel('Force (N)')
    xlabel('Time (s)')
    title('Force versus Time')
end
if Units==2
    plot(time, DFz,'b',time, BFz,'g',time, solenoid*10,'r')
    legend('Donald','Bertha','Solenoid')
    ylabel('Force (lb)')
    xlabel('Time (s)')
    title('Force versus Time')
end
```

B5. Step Length and Step Velocity Calculations

Marker data from the fifth metatarsal head were used to find step length. Step time was input by the user so that step velocity could be calculated. Note that because a segment of time was visually chosen by the operator while viewing the fifth metatarsal head location data, no splining or filtering were employed.

```
% Step_length_velocity.m
% Program to find step length
% Input a kin data file
% Output step length from 5th metatarsal head
% Started 6/19/02, Emily Lloyd
% Last Revised: 6/20/02, Emily Lloyd

% Input

basefilename=input('Input Subject''s Initials: ','s');
cfn=input('Input File #: ');

                                                              % from optoreader
program
   clear rawdata header col_header hdr hdrline;
   if cfn<10,
      scfn=['00' num2str(cfn)];
   elseif cfn>9 & cfn<100,
      scfn=['0' num2str(cfn)];
   end;

current_file_name = [basefilename 'kin$#' scfn '.mat']; %extfilename];
clear rawdata;
load(current_file_name, 'rawdata');
kin_data = TransferkinGCS(rawdata);

% Get relevant columns

metay = kin_data(:,2)./1000;          % Column 2 in kin_data is the y direction of the 5th metatarsal head, in m
time = kin_data(:,31);                % Column 31 in kin_data is time, in seconds

% Plot data
figure(1)
plot(time,metay)
xlabel('Time (seconds)')
ylabel('Length (meters)')
title('Y Position of the Fifth Metatarsal')

%fprintf('Please view plot, then press any key to continue.\n')
%break

subject_height=input('Input subject height in m: ');
ti0 = input('Input beginning of first 0.25 second period (smallest interval is 0.005 s): ');
tf0 = input('Input beginning of second 0.25 second period (smallest interval is 0.005 s): ');
step_time = input('Input step time (seconds): ');

% Find relevant frame numbers

counter = 1;

while time(counter)<ti0
   counter = counter + 1;
end
```

```matlab
counter1 = counter;
counter2 = counter1 + 24;

while time(counter)<tf0
    counter = counter + 1;
end

counter3 = counter;
counter4 = counter3 + 24;

% Find averages

totaly = 0;

for counter=counter1:counter2
    totaly = metay(counter)+totaly;
end

initialy = totaly/25;

totaly = 0;

for counter=counter3:counter4
    totaly = metay(counter)+totaly;
end

finaly = totaly/25;

step_length = abs(finaly-initialy);              % meters
step_length_bh = step_length/subject_height * 100;     % % body length

step_velocity = step_length/step_time;              % meters/second
step_velocity_bh = step_length_bh/step_time /100;       % body lengths/second

fprintf('Step length is %4.3f meters.\n',step_length)
fprintf('Step length is %4.3f %% body lengths.\n',step_length_bh)
fprintf('Step velocity is %4.3f meters/second.\n',step_velocity)
fprintf('Step velocity is %4.3f body lengths/second.\n',step_velocity_bh)
```

Appendix C: Matlab Code for Study #2

       Matlab code was used to convert the text data into matlab format (C1). Afterwards, a splining program (C2) was used, with a special version for files missing the first and/or last 5 data points for any marker data (C3). Subrountines for the splining program converted the grand reaction force signal into forces and moments (C4), transferred the grand reaction force (C5) and marker data (C6) into the global coordinate system, converted the marker location data into units of meters (C7), set the origin for data analysis (C8), and ultimately allowed the used to spline the marker data (C9) for the initial spline program (C2) or splined missing data and replaced missing beginning and end values with the closest existing data (C10) for the modified splining program (C3). An addition subroutine during the splining programs plotted all of the marker data after the initial splining data to allow the user to confirm that the appropriate data points had been splined (C11). The Hanavan model (1964, see Chapter 3 for citation) was used to create anthropometric files for all subjects (C12). The joint torques program (C13) used inverse dynamics analysis, the matlab files output from the splining programs, and the anthropometric files to output graphs of the joint torques and allow the user to select the range of the graph they were interested in so that numerical values returned were from the appropriate part of the trip recovery. Three different functions filtered the marker data (C14), grand reaction force data (C15), and harness load cell data (C16). The initial offset in grand reaction force data was adjusted to zero (C17), and the grand reaction force was downsampled to match the frequency of the marker data (C18). The center of pressure on the large Bertec forceplate used for landing was calculated (C19). Leg angles (C20), center of mass locations for the body segments (C21), and center of mass linear and angular accelerations (C22) were computed and used by another subroutine to calculate joint torques using inverse dynamics (C23). An additional routine allowed the user to specify the domain of the torque plots of interest to the study and returned the maximum and minimum torque values in that domain (C24).

In the code for the Joint Torques program, ankle dorsiflexion, knee extension, and hip flexion are defined to be positive and ankle plantarflexion, knee flexion, and hip extension are defined to be negative.

C1. Optoreader2

Original program by Dr. Madigan slightly modified to convert text data to Matlab data for easy manipulation in Matlab. The modification removed a loop to replace missing data, instead leaving all signals as collected so that missing data could be splined at a later time.

```
% optoreader2.m
% This program is designed to convert files exported from Optotrak
% DISPLAY software into MatLab Files
% Revised from optoreader.m on 5/16/02 by Emily Lloyd

basefilename=input('Input Base File Name: ','s');
extfilename=input('Input File Extension: .','s');
ftrial=input('Input First File #: ');
ltrial=input('Input Last File #: ');
```

```matlab
subjectname=input('Input Subject Name: ','s');

for cfn=ftrial:ltrial,
   clear rawdata header col_header hdr hdrline;
   if cfn<10,
      scfn=['00' num2str(cfn)];
   elseif cfn>9 & cfn<100,
      scfn=['0' num2str(cfn)];
   end;
   current_file_name = [basefilename scfn '.' extfilename];


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%
           % read in header information

   fid=fopen(current_file_name);
   if fid ==-1, error('File could not be read.');end;
   for i = 1:20,
      hdrline=fgetl(fid);
      hdr(i,:)=cellstr(hdrline);
   end;
   header=char(hdr);
   clear hdr hdrline;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%
           % parse the header

   k= findstr(',',header(20,:)); %determine # of columns in comma delimited file
           n_col=length(k)+1;
           for i=1:n_col,
      if i==1;
         X=char(header(20,1:k(i)-1));
      elseif i==n_col;
         X=char(header(20,k(i-1)+1:length(header)));
      else
         X=char(header(20,k(i-1)+1:k(i)-1));
      end;
      col_header(i)=cellstr(X);
   end;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%
           % read in data

   fprintf('Reading in data from file %2.0f...',cfn);
   f_error=0;
           i=1;
           while f_error==0,
      eval('rawdata(:,i)=fscanf(fid,"%f,");','f_error=1;');
      i=i+1;
           end;
   rawdata=rawdata';
   fclose(fid);


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%
```

```
    % save data

    save_file_name = [subjectname basefilename scfn '.mat'];
    save(save_file_name,'rawdata','header','col_header'); %write data to file

end; % end of cfn loop that loops through data files

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%
% clear extraneous variables

clear X basefilename cfn col current_file_name extfilename f_error fid
clear ftrial i k ltrial n_col row save_file_name scfn subjectname
```

C2. Splining Program

Program to convert grand reaction signal into forces and moments, convert all data to the global coordinate system, and allow splining of the kinematic data by the user.

```
% SpliningProgram.m
% Program to call other programs to get the desired results
% Input a data file
% Output calculations
% Started 6/17/02, Emily Lloyd
% Last Revised: 9/18/02, Emily Lloyd

% Input file information ********************************************

basefilename=input('Input Subject''s Initials: ','s');
cfn=input('Input File #: ');

        % from optoreader program
  clear rawdata header col_header hdr hdrline;
  if cfn<10,
    scfn=['00' num2str(cfn)];
  elseif cfn>9 & cfn<100,
    scfn=['0' num2str(cfn)];
  end;

% Input grf data, make data into forces and moments, put into GCS *******
current_file_name = [basefilename 'grf$#' scfn '.mat'];
load(current_file_name, 'rawdata');
grf_data = GetGRFOutput(rawdata);
grf_data = TransfergrfGCS(grf_data);

% Input kin data, put into GCS ********************************************
current_file_name = [basefilename 'kin$#' scfn '.mat'];
clear rawdata;
load(current_file_name, 'rawdata');
kin_data = TransferkinGCS(rawdata);
kin_data = mm_to_m(kin_data);
Set_kin_origin;

% Splining loop (for kin data) ********************************************

splining_needed = 1;

while splining_needed==1
  kin_data = SplinekinIndex(kin_data);

  plot_all_kin_data_index;

  fprintf('\n');
  splining_needed = input('Is further splining needed?  Input 1 for yes, 2 for no: ');
end

% Save splined kin data and grf data ********************************

  save_file_name = [basefilename 'kin' scfn '.mat'];
  save(save_file_name,'kin_data'); %write data to file

  save_file_name = [basefilename 'grf' scfn '.mat'];
  save(save_file_name,'grf_data'); %write data to file
```

C3. Splining Program for Missing Beginning or Ending Kinematic Data

Program to convert grand reaction signal into forces and moments, convert all data to the global coordinate system, and allow splining of the kinematic data by the user. This program is an altered version of the program in Appendix C2 that allows the user to replace missing kinematic data at the beginning or end of the file with the nearest data value so that matrix size can be maintained (data at the extreme ends of collection were not necessary for joint torque analysis at landing).

```
% SpliningProgram2.m
% Program to call other programs to get the desired results
% Input a data file
% Output calculations
% Started 1/22/03, Emily Lloyd
% Last Revised: 1/22/03, Emily Lloyd

% Input file information **********************************************

basefilename=input('Input Subject''s Initials: ','s');
cfn=input('Input File #: ');

            % from optoreader program
    clear rawdata header col_header hdr hdrline;
    if cfn<10,
        scfn=['00' num2str(cfn)];
    elseif cfn>9 & cfn<100,
        scfn=['0' num2str(cfn)];
    end;

% Input grf data, make data into forces and moments, put into GCS *******
current_file_name = [basefilename 'grf$#' scfn '.mat'];
load(current_file_name, 'rawdata');
grf_data = GetGRFOutput(rawdata);
grf_data = TransfergrfGCS(grf_data);

% Input kin data, put into GCS ****************************************
current_file_name = [basefilename 'kin$#' scfn '.mat'];
clear rawdata;
load(current_file_name, 'rawdata');
kin_data = TransferkinGCS(rawdata);
kin_data = mm_to_m(kin_data);
Set_kin_origin;

% Splining loop (for kin data) ****************************************

splining_needed = 1;

while splining_needed==1
    kin_data = SplinekinIndexBegEnd(kin_data);

    plot_all_kin_data_index;

    fprintf('\n');
    splining_needed = input('Is further splining needed?  Input 1 for yes, 2 for no: ');
end

% Save splined kin data and grf data *********************************

    save_file_name = [basefilename 'kin' scfn '.mat'];
```

```
save(save_file_name,'kin_data'); %write data to file

save_file_name = [basefilename 'grf' scfn '.mat'];
save(save_file_name,'grf_data'); %write data to file
```

C4. Convert the Grand Reaction Force Signal into Forces and Moments

This function converts grand reaction force signal into forces and moments on the large force place (named Bertha) and the AMTI forceplate nearest the door (Donald).

```
function [grf] = GetGRFOutput(rawdata)
% Function to put grf data into correct units
% From TripTimes.m
% Input a grf data file
% Output correct grf values
% Started 7/22/02, Emily Lloyd
% Last Revised: 7/22/02, Emily Lloyd

% get files

DSFx = rawdata(:,8);          % Donald's SFx (Signal in the Fx direction) (channel 6 (0-15) of
analog data)
DSFy = rawdata(:,9);          % Donald's SFy (Signal in the Fy direction) (channel 7 (0-15) of
analog data)
DSFz = rawdata(:,10);         % Donald's SFz (Signal in the Fz direction) (channel 8 (0-15) of
analog data)
DSMx = rawdata(:,11);         % Donald's SMx (Signal in the Mx direction) (channel 9 (0-15) of
analog data)
DSMy = rawdata(:,12);         % Donald's SMy (Signal in the My direction) (channel 10 (0-15) of
analog data)
DSMz = rawdata(:,13);         % Donald's SMz (Signal in the Mz direction) (channel 11 (0-15) of
analog data)

BSFx = rawdata(:,2);          % Bertha's SFx (Signal in the Fx direction) (channel 0 (0-15) of
analog data)
BSFy = rawdata(:,3);          % Bertha's SFy (Signal in the Fy direction) (channel 1 (0-15) of
analog data)
BSFz = rawdata(:,4);          % Bertha's SFz (Signal in the Fz direction) (channel 2 (0-15) of
analog data)
BSMx = rawdata(:,5);          % Bertha's SMx (Signal in the Mx direction) (channel 3 (0-15) of
analog data)
BSMy = rawdata(:,6);          % Bertha's SMy (Signal in the My direction) (channel 4 (0-15) of
analog data)
BSMz = rawdata(:,7);          % Bertha's SMz (Signal in the Mz direction) (channel 5 (0-15) of
analog data)

% Account for gain settings

bertha_gains = [10 10 2 5 5 5];

BSFx=BSFx./bertha_gains(1);
BSFy=BSFy./bertha_gains(2);
BSFz=BSFz./bertha_gains(3);
BSMx=BSMx./bertha_gains(4);
BSMy=BSMy./bertha_gains(5);
BSMz=BSMz./bertha_gains(6);

donald_gains = [1996.6 1996.6 1996.6 1996.4 1997 4000]; % for Donald, from Amplifier Calibration Sheet

Vexcite = 4.991;              % Volts, nominal value of 5 V, from Donald's Amplifier Calibration
Sheet

CF = donald_gains*Vexcite*10^-6;   % Conversion Factor, OR6-5 Manual, page 3

% Get forces and moments for Donald and Bertha using Calibration Matricies
```

```
DonaldC = [1.512 0.005 0.016 -0.002 -0.003 0.006;
  0.006 1.516 0.023 -0.002 -0.011 0.000;
  0.024 0.025 5.893 0.000 0.007 -0.001;
  0.000 0.000 0.000 0.604 -0.005 -0.003;
  0.000 0.000 0.000 0.001 0.605 -0.002;
  0.000 0.002 0.000 -0.001 -0.004 0.306];

BerthaC = [863.0 -11.2 -5.5 0.1 2.5 -3.3;
  22.6 855.5 -13.4 -7.5 6.0 -3.7;
  -37.3 -0.5 1805.6 -1.6 -11.2 -17.1;
  -3.4 -72.9 -2.6 908.3 -2.7 2.8;
  73.8 -3.3 0.1 -1.5 908.9 6.5;
  1.8 -5.0 1.4 3.8 3.3 302.1];

for counter=1:5000
  DonaldS = [DSFx(counter); DSFy(counter); DSFz(counter); DSMx(counter); DSMy(counter); DSMz(counter)];
  Donald = DonaldC * DonaldS;
  DFx(counter) = Donald(1)./CF(1);
  DFy(counter) = Donald(2)./CF(2);
  DFz(counter) = Donald(3)./CF(3);
  DMx(counter) = Donald(4)./CF(4);
  DMy(counter) = Donald(5)./CF(5);
  DMz(counter) = Donald(6)./CF(6);
end

for counter=1:5000
  BerthaS = [BSFx(counter); BSFy(counter); BSFz(counter); BSMx(counter); BSMy(counter); BSMz(counter)];
  Bertha = BerthaC * BerthaS;
  BFx(counter) = Bertha(1);
  BFy(counter) = Bertha(2);
  BFz(counter) = Bertha(3);
  BMx(counter) = Bertha(4);
  BMy(counter) = Bertha(5);
  BMz(counter) = Bertha(6);
end

for index = 1:5000

grf(index,2) = BFx(index);
grf(index,3) = BFy(index);
grf(index,4) = BFz(index);
grf(index,5) = BMx(index);
grf(index,6) = BMy(index);
grf(index,7) = BMz(index);

grf(index,8) = DFx(index);
grf(index,9) = DFy(index);
grf(index,10) = DFz(index);
grf(index,11) = DMx(index);
grf(index,12) = DMy(index);
grf(index,13) = DMz(index);

grf(index,1)=rawdata(index,1);
grf(index,14)=rawdata(index,14);
grf(index,15)=rawdata(index,15);
grf(index,16)=rawdata(index,16);
grf(index,17)=rawdata(index,17);

end        % ends for statement
```

C5. Transfer Grand Reaction Force Data to the Global Coordinate System

Function to put grand reaction force data into the global coordinate system.

```
function [grf] = TransfergrfGCS(rawdata)
% Program to put grf.mat data into the Global Coordinate System (GCS)
% Input a grf data file
% Output grf data in GCS
% Started 6/17/02, Emily Lloyd
% Last Revised: 9/25/02, Emily Lloyd

% Read in file rawdata

grf(:,1) = rawdata(:,1);                        % Frame numbers are the
same
grf(:,16) = rawdata(:,16);                       % Harness load cell is the
same
grf(:,17) = rawdata(:,17);                       % Solenoid pulse is the same

% *********************NOTE THAT ALL VALUES ARE MULTIPLIED BY -1 TO REPRESENT THE
REACTION FORCES (FORCES
% ON THE SUBJECT) INSTEAD OF THE FORCES PUT ON THE FORCE
PLATE*******************************************

% Bertha
grf(:,2) = -1*rawdata(:,3);                      % GCS Bertha Fx(2) = original Bertha Fy(3)
grf(:,3) = -1*rawdata(:,2);                      % GCS Bertha Fy(3) = original Bertha Fx(2)
grf(:,4) = rawdata(:,4);                         % GCS Bertha Fz(4) = original Bertha Fz(4) *-1

grf(:,5) = -1*rawdata(:,6);                      % GCS Bertha Mx(5) = original Bertha My(6)
grf(:,6) = -1*rawdata(:,5);                      % GCS Bertha My(6) = original Bertha Mx(5)
grf(:,7) = rawdata(:,7);                         % GCS Bertha Mz(7) = original Bertha Mz(7) *-
1

% Donald
grf(:,8) = rawdata(:,9);                         % GCS Donald Fx(8) = original Donald Fy(9)
times -1
grf(:,9) = rawdata(:,8);                         % GCS Donald Fy(9) = original Donald Fx(8)
times -1
grf(:,10) = rawdata(:,10);                       % GCS Donald Fz(10) = original Donald Fz(10)
times -1

grf(:,11) = rawdata(:,12);                       % GCS Donald Mx(11) = original Donald
My(12) * -1
grf(:,12) = rawdata(:,13);                       % GCS Donald My(12) = original Donald
Mx(11) * -1
grf(:,13) = rawdata(:,13);                       % GCS Donald Mz(13) = original Donald
Mz(13) times -1

% Willy
grf(:,14) = -1*rawdata(:,14);                    % GCS Willy Fy(14) = original Willy Fx(14)
% NOTE THIS IS NOW WILLY IN THE Y DIRECTION OF THE GCS
grf(:,15) = rawdata(:,15);                       % GCS Willy Fz(15) = original Willy Fz(15) *-1

% OUTPUT grf
```

C6. Transfer Marker Data to the Global Coordinate System

Function to put marker data into the global coordinate system.

```
function [kin] = TransferkinGCS(rawdata)
% Program to put kin.mat data into the Global Coordinate System (GCS)
% Input a kin data file
% Output kin data in GCS
% Started 6/17/02, Emily Lloyd
% Last Revised: 9/19/02, Emily Lloyd

% Read in file rawdata

% Marker 1
kin(:,1) = rawdata(:,4);                    % GCS x1(1) = original z1(4)
kin(:,2) = -1*rawdata(:,3);                 % GCS y1(2) = original y1(3) * -1
kin(:,3) = rawdata(:,2);                    % GCS z1(3) = original x1(2)

% Marker 2
kin(:,4) = rawdata(:,7);                    % GCS x2(4) = original z2(7)
kin(:,5) = -1*rawdata(:,6);                 % GCS y2(5) = original y2(6) * -1
kin(:,6) = rawdata(:,5);                    % GCS z2(6) = original z2(5)

% Marker 3
kin(:,7) = rawdata(:,10);                   % GCS x3(7) = original z3(10)
kin(:,8) = -1*rawdata(:,9);                 % GCS y3(8) = original y3(9) * -1
kin(:,9) = rawdata(:,8);                    % GCS z3(9) = original x3(8)

% Marker 4
kin(:,10) = rawdata(:,13);                  % GCS x4(10) = original z4(13)
kin(:,11) = -1*rawdata(:,12);               % GCS y4(11) = original y4(12) * -1
kin(:,12) = rawdata(:,11);                  % GCS z4(12) = original x4(11)

% Marker 5
kin(:,13) = rawdata(:,16);                  % GCS x5(13) = original z5(16)
kin(:,14) = -1*rawdata(:,15);               % GCS y5(14) = original y5(15) * -1
kin(:,15) = rawdata(:,14);                  % GCS z5(15) = original x5(14)

% Marker 6
kin(:,16) = rawdata(:,19);                  % GCS x6(16) = original z6(19)
kin(:,17) = -1*rawdata(:,18);               % GCS y6(17) = original y6(18) * -1
kin(:,18) = rawdata(:,17);                  % GCS z6(18) = original x6(17)

% Marker 7
kin(:,19) = rawdata(:,22);                  % GCS x7(19) = original z7(22)
kin(:,20) = -1*rawdata(:,21);               % GCS y7(20) = original y7(21) * -1
kin(:,21) = rawdata(:,20);                  % GCS z7(21) = original x7(20)

% Marker 8
kin(:,22) = rawdata(:,25);                  % GCS x8(22) = original z8(25)
kin(:,23) = -1*rawdata(:,24);               % GCS y8(23) = original y8(24) * -1
kin(:,24) = rawdata(:,23);                  % GCS z8(24) = original x8(23)

% Marker 9
kin(:,25) = rawdata(:,28);                  % GCS x9(25) = original z9(28)
kin(:,26) = -1*rawdata(:,27);               % GCS y9(26) = original y9(27) * -1
kin(:,27) = rawdata(:,26);                  % GCS z9(27) = original x9(26)

% Marker 10
kin(:,28) = rawdata(:,31);                  % GCS x10(28) = original z10(31)
kin(:,29) = -1*rawdata(:,30);               % GCS y10(29) = original y10(30) * -1
```

```
kin(:,30) = rawdata(:,29);                          % GCS z10(30) = original x10(29)

% Time/frame
kin(:,31) = rawdata(:,1) / 200;                     % Time = frame / 200

% OUTPUT kin
```

C7. Convert Marker Data from Millimeters to Meters

Function to convert marker data from units of millimeters to meters.

```
function [kin] = mm_to_m(kin_data)
% Function to convert mm kin data to m
% Input a kin data file in mm
% Output same file in m
% Started 9/12/02, Emily Lloyd
% Last Revised: 9/19/02, Emily Lloyd

for column=1:30
   for counter=1:1000
      kin(counter,column)=kin_data(counter,column)/1000;
   end
end

kin(:,31) = kin_data(:,31);
```

C8. Set Marker Origin

Program to set the origin of the global coordinate system to the center of the Bertec force plate.

```
% Set_kin_origin.m
% Program to center all kinematic data at Bertha's origin
% Started 9/2/02, Emily Lloyd
% Last Revised: 9/19/02, Emily Lloyd

marker_9_y = 0;
marker_9_z = 0;
marker_10_y = 0;
marker_10_z = 0;
for counter=1:10
   marker_9_y = marker_9_y + kin_data(counter,26);
   marker_9_z = marker_9_z + kin_data(counter,27);
   marker_10_y = marker_10_y + kin_data(counter,29);
   marker_10_z = marker_10_z + kin_data(counter,30);
end

marker_9_y = marker_9_y/10;
marker_9_z = marker_9_z/10;
marker_10_y = marker_10_y/10;
marker_10_z = marker_10_z/10;

yoffset = (marker_9_y + marker_10_y)/2;
zoffset = (marker_9_z + marker_10_z)/2;

zoffset = zoffset + 0.01;      % for measurements in m

for counter = 1:10
   for index = 1:1000
      kin_data(index,(3*counter - 1)) = kin_data(index,(3*counter - 1)) - yoffset;    % sets all y data
      kin_data(index,(3*counter)) = kin_data(index,(3*counter)) - zoffset;          % sets all z data
   end
end
```

C9. Splining Function

Function that allows user to select the marker data that need to be replaced and replaces them with a cubic spline using data from the five closest points on both sides to the splined data.

```
function [kin] = SplinekinIndex(kin)
% Program to use spline to replace missing data points
% Input a kin data file
% Output splined kin data
% Started 6/25/02, Emily Lloyd
% Last Revised: 1/16/03, Emily Lloyd

indexvector = 1:1000;
for column = 1:30

fprintf ('\n')
fprintf ('You are looking at column %1.f of the kinematic data \n',column)

clear marker;
marker = kin(:,column);

% Take out excessive marker values

for counter = 1:1000,
   if abs(marker(counter)>10000)
      marker(counter)=5000;
   end
end

% when missing points are found, find their x values, and the x values of the 5 points previous and the 5 points after

% Plot data

figure(column)
grid on
plot(indexvector,marker,'.-')
ylim([-5000 5000]);
xlabel('Index Number')
ylabel('Position (mm)')
title('Position of the Marker')

% Need splining?

continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');

see_graph_again = 2;
if continue_splining ==2
   fprintf('Press 1 to see the graph again with automatic y limits, \n');
   see_graph_again = input('2 if you do not want to see the graph again: ');
end

if see_graph_again == 1
   figure(column)
   grid on
   plot(indexvector,marker,'.-')
   xlabel('Index Number')
   ylabel('Position (mm)')
   title('Position of the Marker')
   continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');
```

```
end

while continue_splining==1

% identify missingxpoints

xbegin = input('Enter index of first missing x point: ');
xend = input('Enter index of last missing x point: ');

% make matrices - missingxpoints, previousx, afterx, previousy, aftery

missingxpoints = indexvector(xbegin:xend);

previousx = [(xbegin - 5) (xbegin - 4) (xbegin - 3) (xbegin - 2) (xbegin - 1)];
afterx = [(xend + 1) (xend + 2) (xend + 3) (xend + 4) (xend + 5)];

previousy = [marker(xbegin - 5) marker(xbegin - 4) marker(xbegin - 3) marker(xbegin - 2) marker(xbegin - 1)];
aftery = [marker(xend + 1) marker(xend + 2) marker(xend + 3) marker(xend + 4) marker(xend + 5)];

x = [previousx afterx];
y = [previousy aftery];
xx = [missingxpoints];
yy = spline(x,y,xx)
%yy = interp1(x,y,xx);

clear x y xx missingxpoints previousx afterx previousy aftery;

% replace y of xx with yy (replace missing points with splined points)

missingindex = 1;                    % variable for index of yy

for index=xbegin:xend
    marker(index)=yy(missingindex);
    missingindex = missingindex + 1;
end

clear yy

% Plot data again

figure(column)
grid on
plot(indexvector,marker,'.-')
xlabel('Index Number')
ylabel('Position (m)')
title('Position of the Marker')

% Still need splining?

continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');

end          % ends while statement that allows user to continuously pick points that need to be splined

kin(:,column) = marker;

end                       % ends for statement
```

C10. Revised Splining Function

Function that allows user to select the marker data that need to be replaced. Replaces beginning or end data with the nearest value and replaces other data with a cubic spline using data from the five closest points on both sides to the splined data.

```
function [kin] = SplinekinIndexBegEnd(kin)
% Program to use spline to replace missing data points
% Input a kin data file
% Output splined kin data
% Started 1/16/03, Emily Lloyd
% Last Revised: 1/22/03, Emily Lloyd

indexvector = 1:1000;
for column = 1:30

fprintf ('\n')
fprintf ('You are looking at column %1.f of the kinematic data \n',column)

clear marker;
marker = kin(:,column);

% Take out excessive marker values

for counter = 1:1000,
   if abs(marker(counter)>10000)
      marker(counter)=5000;
   end
end

% when missing points are found, find their x values, and the x values of the 5 points previous and the 5 points after

% Plot data

figure(column)
grid on
plot(indexvector,marker,'.-')
ylim([-5000 5000]);
xlabel('Index Number')
ylabel('Position (mm)')
title('Position of the Marker')

% Need splining?

continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');

see_graph_again = 2;
if continue_splining ==2
   fprintf('Press 1 to see the graph again with automatic y limits, \n');
   see_graph_again = input('2 if you do not want to see the graph again: ');
end

if see_graph_again == 1
   figure(column)
   grid on
   plot(indexvector,marker,'.-')
   xlabel('Index Number')
   ylabel('Position (mm)')
   title('Position of the Marker')
   continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');
```

```
end

while continue_splining==1

% determine normal splining or missing beginning or end points

special = input('Press 1 for standard splining, 2 if beginning or end points are missing: ');

% for missing beginning or end points
if special==2

    beg_or_end = input('Press 1 for missing start points, 2 for missing end points: ');

    if beg_or_end==1               % missing beginning points
        beg_pt = input('Enter last missing data point: ');
        for index=1:beg_pt
            marker(index)=marker(beg_pt+1);
        end
    end

    if beg_or_end==2               % missing end points
        end_pt = input('Enter first missing data point: ');
        for index=end_pt:1000
            marker(index)=marker(end_pt-1);
        end
    end

end % ends special splining

if special==1   % normal splining

% identify missingxpoints

xbegin = input('Enter index of first missing x point: ');
xend = input('Enter index of last missing x point: ');

% make matrices - missingxpoints, previousx, afterx, previousy, aftery

missingxpoints = indexvector(xbegin:xend);

previousx = [(xbegin - 5) (xbegin - 4) (xbegin - 3) (xbegin - 2) (xbegin - 1)];
afterx = [(xend + 1) (xend + 2) (xend + 3) (xend + 4) (xend + 5)];

previousy = [marker(xbegin - 5) marker(xbegin - 4) marker(xbegin - 3) marker(xbegin - 2) marker(xbegin - 1)];
aftery = [marker(xend + 1) marker(xend + 2) marker(xend + 3) marker(xend + 4) marker(xend + 5)];

x = [previousx afterx];
y = [previousy aftery];
xx = [missingxpoints];
yy = spline(x,y,xx)
%yy = interp1(x,y,xx);

clear x y xx missingxpoints previousx afterx previousy aftery;

% replace y of xx with yy (replace missing points with splined points)

missingindex = 1;                        % variable for index of yy

for index=xbegin:xend
    marker(index)=yy(missingindex);
    missingindex = missingindex + 1;
end
```

```
clear yy

end % ends normal splining

% Plot data again

figure(column)
grid on
plot(indexvector,marker,'.-')
xlabel('Index Number')
ylabel('Position (m)')
title('Position of the Marker')

% Still need splining?

continue_splining = input('Press 1 if splining is needed, 2 if splining is complete: ');

end          % ends while statement that allows user to continuously pick points that need to be splined

kin(:,column) = marker;

end                        % ends for statement
```

C11. Plot Marker Data to Confirm Proper Splining

Program that plots all three directions (x, y, and z) of all 10 markers. Allows user to see the data in a normal plot and assess if all bad data have been splined.

```
% Plot_all_kin_data_index.m
% Program to plot all kinematic data, columns 1 - 30, each on their own plot
% Input a kin data file
% Output plots
% Started 7/25/02, Emily Lloyd
% Last Revised: 7/25/02, Emily Lloyd

index = 1:1000

for column = 1:10
   figure(column)
   plot(index,kin_data(:,column))
   xlabel('Index')
   ylabel('Position (mm)')
   title('Position of the Marker')
end

fprintf('Press any key to continue');
pause;

for column = 11:20
   figure(column)
   plot(index,kin_data(:,column))
   xlabel('Index')
   ylabel('Position (mm)')
   title('Position of the Marker')
end

fprintf('\n');
fprintf('Press any key to continue');
pause

for column = 21:30
   figure(column)
   plot(index,kin_data(:,column))
   xlabel('Index')
   ylabel('Position (mm)')
   title('Position of the Marker')
end
```

C12. Hanavan Model Anthropometric Estimations

Program written by Dr. Madigan that estimates the mass, center of mass location, and moment of inertia for the foot, shank, and thigh. Uses the Hanavan Model (1964, see citation in Chapter 3) to estimate the mass, center of mass position, and moment of inertia of the foot, shank, and thigh. Outputs values into a MatLab file that is referenced by the program to calculate joint torques (Appendix C13).

```
% Hanavan anthropometric model
%
% This model can be used to determine each segment's mass,
% COM location, and moment of inertia.  This script is run
% as a subroutine to invdyn.m
%
% Body segments are numbered as follows within this script:
% 1 head
% 2 upper torso
% 3 lower torso
% 4 hand
% 5 arm
% 6 forearm
% 7 thigh
% 8 shank
% 9 foot
%
% At the end of this script, the body segments specific to the
% invdyn.m are renumbered as follows:
% 1 foot
% 2 shank
% 3 thigh
%
% segment masses are in variable 'sm(segment #)'.
% segment COM location are along longitudinal axis as a distance
%    'com(segment #)' from the proximal end.
% segment moment of inertia are for rotation in the sagittal plane
%    and are in variable 'i(segment #)'.  Segment moment of inertia
%    is only determined for thighs, shanks, and feet
%
% input data-------------------------------------------------------


clear;
datacollection = input('Enter output file name without extension: ','s');

fprintf('Enter measurements in cm.\n');
ankc=input('Ankle circumference: '); ankc=ankc/2.54;
%axilc=input('Axillary arm circumference: ');
%buttd=input('Buttock depth: ');
%chesb=input('Chest breadth: ');
%chesd=input('Chest depth: ');
%elbc=input('Elbow circumference: ');
%fistc=input('Fist circumference: ');
%foarl=input('Forearm length: ');
footl=input('Foot length: '); footl=footl/2.54;
gknec=input('Knee circumference: '); gknec=gknec/2.54;
%headc=input('Head circumference: ');
%hipb=input('Hip breadth: ');
%shldh=input('Shoulder height: ');
%sith=input('Sitting height: ');sith=sith/2.54;
```

```
sphyh=input('Sphyrion height: ');sphyh=sphyh/2.54;
height=input('Height: ');height=height/2.54;
%subh=input('Substernale height: ');
thihc=input('Thigh Circumference: ');thihc=thihc/2.54;
tibh=input('Tibiale height: ');tibh=tibh/2.54;
troch=input('Trochanteric height: ');troch=troch/2.54;
%uparl=input('Upper arm length: ');
weight=input('Weight (lbs): ');
%waisb=input('Waist breadth: ');
%waisd=input('Waist depth: ');
%wrisc=input('Wrist circumference: ');

% determine segment weights-----------------------------------
hnt=(.47*weight)+12;
bua=(.08*weight)-2.9;
bfo=(.04*weight)-0.5;
bh=(.01*weight)+.7;
bul=(.18*weight)+3.2;
bll=(.11*weight)-1.9;
bf=(0.02*weight)+1.5;
calculated_w=hnt+bua+bfo+bh+bul+bll+bf;
diff=weight-calculated_w;

hnt=hnt+(diff/13);
bua=bua+(2*diff/13);
bfo=bfo+(2*diff/13);
bh=bh+(2*diff/13);
bul=bul+(2*diff/13);
bll=bll+(2*diff/13);
bf=bf+(2*diff/13);

% head segment (1)-------------------------------------------
%sw(1)=.079*weight;
%sm(1)=sw(1)/32.2;
%sl(1)=height-shldh;
%com(1)=.5;

% upper torso (2)--------------------------------------------
%r(2)=.5*chesb;
%rr(2)=.25*(chesd+waisd);
%sl(2)=shldh-subh;
%v2(2)=pi*r(2)*rr(2)*sl(2);
%v3(2)=pi*(.5*hipb)*(.25*(waisd+buttd))*(sith-(height-subh));
%delta(2)=(hnt-sw(1))/(v2(2)+((1.01/.92)*v3(2)));
%sw(2)=delta(2)*v2(2);
%sm(2)=sw(2)/32.2;
%com(2)=.5;

% lower torso (3)--------------------------------------------
%r(3)=.5*hipb;
%rr(3)=.25*(waisd+buttd);
%sl(3)=sith-(height-subh);
%com(3)=.5;
%sw(3)=hnt-sw(1)-sw(2);
%sm(3)=sw(3)/32.2;

% hand (4) --------------------------------------------------
%r(4)=fistc/(2*pi);
%rr(4)=r(4);
%sl(4)=2*r(4);
%com(4)=.5;
%sw(4)=.5*bh;
```

```
%sm(4)=sw(4)/32.2;

% arm (5) ---------------------------------------------------
%r(5)=axilc/(2*pi);
%rr(5)=elbc/(2*pi);
%sl(5)=uparl;
%sw(5)=.5*bua;
%sm(5)=sw(5)/32.2;

%seg=5;
%delta(seg)=(3*sw(seg))/(sl(seg)*((r(seg)^2)+(r(seg)*rr(seg))+(rr(seg)^2))*pi);
%mu(seg)=rr(seg)/r(seg);
%sigma(seg)=1+mu(seg)+mu(seg)^2;
%com(seg)=(1+(2*mu(seg))+(3*mu(seg)^2))/(4*sigma(seg));
%aa(seg)=(9/(20*pi))*((1+(mu(seg)^2)+(mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
%bb(seg)=(3/80)*((1+(4*mu(seg))+(10*mu(seg)^2)+(4*mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
%i(seg)=(aa(seg)*(sm(seg)^2)/(delta(seg)*sl(seg)))+(bb(seg)*sm(seg)*sl(seg)^2);

% forearm (6) ------------------------------------------------------
%sl(6)=foarl;
%r(6)=elbc/(2*pi);
%rr(6)=wrisc/(2*pi);
%sw(6)=.5*bfo;
%sm(6)=sw(6)/32.2;

%seg=6;
%delta(seg)=(3*sw(seg))/(sl(seg)*((r(seg)^2)+(r(seg)*rr(seg))+(rr(seg)^2))*pi);
%mu(seg)=rr(seg)/r(seg);
%sigma(seg)=1+mu(seg)+mu(seg)^2;
%com(seg)=(1+(2*mu(seg))+(3*mu(seg)^2))/(4*sigma(seg));
%aa(seg)=(9/(20*pi))*((1+(mu(seg)^2)+(mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
%bb(seg)=(3/80)*((1+(4*mu(seg))+(10*mu(seg)^2)+(4*mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
%i(seg)=(aa(seg)*(sm(seg)^2)/(delta(seg)*sl(seg)))+(bb(seg)*sm(seg)*sl(seg)^2);

% thigh (7) ---------------------------------------------------
r(7)=thihc/(2*pi);
rr(7)=gknec/(2*pi);
%sl(7)=height-sith-tibh; % actual hanavan code underestimates thigh length
sl(7)=troch-tibh;  % my modified thigh length
%delsh(7)=sith-(height-troch);
sw(7)=.5*bul;
sm(7)=sw(7)/32.2;

seg=7;
delta(seg)=(3*sw(seg))/(sl(seg)*((r(seg)^2)+(r(seg)*rr(seg))+(rr(seg)^2))*pi);
mu(seg)=rr(seg)/r(seg);
sigma(seg)=1+mu(seg)+mu(seg)^2;
com(seg)=(1+(2*mu(seg))+(3*mu(seg)^2))/(4*sigma(seg));
aa(seg)=(9/(20*pi))*((1+(mu(seg)^2)+(mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
bb(seg)=(3/80)*((1+(4*mu(seg))+(10*mu(seg)^2)+(4*mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
i(seg)=(aa(seg)*((sm(seg))^2)/(delta(seg)*sl(seg)))+(bb(seg)*sm(seg)*(sl(seg))^2);

% shank (8) ---------------------------------------------------
r(8)=gknec/(2*pi);
rr(8)=ankc/(2*pi);
sl(8)=tibh-sphyh;
sw(8)=.5*bll;
sm(8)=sw(8)/32.2;

seg=8;
delta(seg)=(3*sw(seg))/(sl(seg)*((r(seg)^2)+(r(seg)*rr(seg))+(rr(seg)^2))*pi);
mu(seg)=rr(seg)/r(seg);
```

```
sigma(seg)=1+mu(seg)+mu(seg)^2;
com(seg)=(1+(2*mu(seg))+(3*mu(seg)^2))/(4*sigma(seg));
aa(seg)=(9/(20*pi))*((1+(mu(seg)^2)+(mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
bb(seg)=(3/80)*((1+(4*mu(seg))+(10*mu(seg)^2)+(4*mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
i(seg)=(aa(seg)*(sm(seg)^2)/(delta(seg)*sl(seg)))+(bb(seg)*sm(seg)*sl(seg)^2);

% foot (9) ------------------------------------------------------
r(9)=.5*sphyh;
rr(9)=.6443*r(9);
sl(9)=footl;
sw(9)=.5*bf;
sm(9)=sw(9)/32.2;

seg=9;
delta(seg)=(3*sw(seg))/(sl(seg)*((r(seg)^2)+(r(seg)*rr(seg))+(rr(seg)^2))*pi);
mu(seg)=rr(seg)/r(seg);
sigma(seg)=1+mu(seg)+mu(seg)^2;
com(seg)=(1+(2*mu(seg))+(3*mu(seg)^2))/(4*sigma(seg));
aa(seg)=(9/(20*pi))*((1+(mu(seg)^2)+(mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
bb(seg)=(3/80)*((1+(4*mu(seg))+(10*mu(seg)^2)+(4*mu(seg)^3)+(mu(seg)^4))/sigma(seg)^2);
i(seg)=(aa(seg)*(sm(seg)^2)/(delta(seg)*sl(seg)))+(bb(seg)*sm(seg)*sl(seg)^2);

% convert data to SI units -----------------------------------

for loop = 1:length(sm),
   sm(loop)=sm(loop)*14.59; % converts slugs to kilograms
   i(loop)=i(loop)*14.59*.0254*.0254; % converts slug*in*in to kg*m*m
end;

% output data --------------------------------------------------
%fid=fopen('out.txt','w');
fprintf('\n\n\nSegment masses\n');
fprintf('   thigh: %5.2f kg\n',sm(7));
fprintf('   shank: %5.2f kg\n',sm(8));
fprintf('   foot: %5.2f kg\n\n',sm(9));
fprintf('Segment COM location (percent of total length from proximal joint)\n');
fprintf('   thigh: %4.3f\n',com(7));
fprintf('   shank: %4.3f\n',com(8));
fprintf('   foot: %4.3f\n\n',com(9));
fprintf('Segment moment of inertia in sagittal plane\n');
fprintf('   thigh: %6.3f kg*m*m\n',i(7));
fprintf('   shank: %6.3f kg*m*m\n',i(8));
fprintf('   foot: %6.3f kg*m*m\n\n',i(9));
%fclose(fid);

% renumber body segments and save data -----------------------------
smtemp=sm;
momentofinertiatemp=i;
comtemp=com;
clear sm i com;
% foot (segment 1)
sm(1)=smtemp(9);
momentofinertia(1)=momentofinertiatemp(9);
com(1)=comtemp(9);
% shank (segment 2)
sm(2)=smtemp(8);
momentofinertia(2)=momentofinertiatemp(8);
com(2)=comtemp(8);
% thigh (segment 3)
sm(3)=smtemp(7);
momentofinertia(3)=momentofinertiatemp(7);
com(3)=comtemp(7);
```

```
clear smtemp momentofinertiatemp comtemp;
weight=weight*.4536;
height=height*.0254;

% save data -------------------------------------------------------
sf=[ datacollection '_anthro.mat'];
eval(['save ',sf,' sm momentofinertia com weight height']);
fprintf('Anthropometric data saved in file %s\n',sf);

clear bul calculated_w delta diff filename footl gknec hnt loop mu r;
clear rr seg sf sigma sl sphyh sw thihc tibh troch aa ankc ans bb bf;
clear bfo bh bll bua ;
```

C13. Joint Torques Program

Main program to calculate joint torques of the stepping lower limb using inverse dynamics.

```
% JointTorques.m
% Program to call other programs to get the desired results
% Input a data file
% Output calculations
% Started 7/22/02, Emily Lloyd
% Last Revised: 5/23/03, Emily Lloyd

% Input

basefilename=input('Input Subject''s Initials: ','s');
cfn=input('Input File #: ');

        % from optoreader program
  clear rawdata header col_header hdr hdrline;
  if cfn<10,
    scfn=['00' num2str(cfn)];
  elseif cfn>9 & cfn<100,
    scfn=['0' num2str(cfn)];
  end;

clear kin_data grf_data

% Load grf data
current_file_name = [basefilename 'grf' scfn '.mat'];
load(current_file_name, 'grf_data');

% load kin data
current_file_name = [basefilename 'kin' scfn '.mat'];
load(current_file_name, 'kin_data');

% load hanavan data
current_file_name = [basefilename '_anthro.mat'];
load(current_file_name)

% Filter data

for column=1:30
  kin_data(:,column) = zerophaseshift_filter_7Hz2(kin_data(:,column));          % Filter kin_data at 7Hz
end

for column=2:15
  grf_data(:,column) = zerophaseshift_filter_100Hz2(grf_data(:,column));        % Filter forces and moments at 100Hz
end

grf_data(:,16) = zerophaseshift_filter_10Hz2(grf_data(:,16));          % Filter harness data at 10Hz

fixgrfoffset; % zeroes Bertha's initial forces and moments

grf_data = down_sample_grf(grf_data);

Find_COP_Bertha;

Find_leg_angles2; % do this for t = 0.5 - 4.5 to take out filtering affects????
Find_COM_locations3;
Find_COM_acc;
```

Find_torques4;

Find_maxes;

C14. 7 Hz Zero Phase Shift Butterworth Filter

Zero phase shift Butterworth filter with a cutoff frequency of 7 Hz, applied to the splined marker data.

```
function filtered_data = zerophaseshift_filter_7Hz(data_to_be_filtered);

% this function lowpass filters kinematic data using a 4th order
% zero-phase-shift filter.  This involves using a 2nd order filter
% forward then reverse through the data.
% see Winter DA. Biomechanics and Motor Control of Human Movement. Pg 36-43.

% Uses filtfilt instead of filter function
% Revised by Emily Lloyd, 5/21/03

sampling_rate=200;
fc=7;
corrected_fc = fc/.802;  %corrects the fc since filtering data twice affects original fc
[filternum,filterdenom] = butter(2,corrected_fc/(sampling_rate/2));
reversed_data_to_be_filtered=data_to_be_filtered(length(data_to_be_filtered):-1:1);
f1 = filtfilt(filternum,filterdenom,data_to_be_filtered);
f2 = f1(length(f1):-1:1);
f3 = filtfilt(filternum,filterdenom,f2);
filtered_data = f3(length(f3):-1:1);
```

C15. 100 Hz Zero Phase Shift Butterworth Filter

Zero phase shift Butterworth filter with a cutoff frequency of 100 Hz, applied to the grand reaction force data.

```
function filtered_data = zerophaseshift_filter_100Hz2(data_to_be_filtered);

% this function lowpass filters kinematic data using a 4th order
% zero-phase-shift filter.  This involves using a 2nd order filter
% forward then reverse through the data.
% see Winter DA. Biomechanics and Motor Control of Human Movement. Pg 36-43.

% Uses filtfilt instead of filter function
% Revised by Emily Lloyd, 5/21/03

sampling_rate=1000;
fc=100;
corrected_fc = fc/.802;  %corrects the fc since filtering data twice affects original fc
[filternum,filterdenom] = butter(2,corrected_fc/(sampling_rate/2));
reversed_data_to_be_filtered=data_to_be_filtered(length(data_to_be_filtered):-1:1);
f1 = filtfilt(filternum,filterdenom,data_to_be_filtered);
f2 = f1(length(f1):-1:1);
f3 = filtfilt(filternum,filterdenom,f2);
filtered_data = f3(length(f3):-1:1);
```

C16. 10 Hz Zero Phase Shift Butterworth Filter

Zero phase shift Butterworth filter with a cutoff frequency of 100 Hz, applied to the harness load cell data.

```
function filtered_data = zerophaseshift_filter_10Hz(data_to_be_filtered);

% this function lowpass filters kinematic data using a 4th order
% zero-phase-shift filter.  This involves using a 2nd order filter
% forward then reverse through the data.
% see Winter DA. Biomechanics and Motor Control of Human Movement. Pg 36-43.

% Uses filtfilt instead of filter function
% Revised by Emily Lloyd, 5/21/03

sampling_rate=1000;
fc=10;
corrected_fc = fc/.802;  %corrects the fc since filtering data twice affects original fc
[filternum,filterdenom] = butter(2,corrected_fc/(sampling_rate/2));
reversed_data_to_be_filtered=data_to_be_filtered(length(data_to_be_filtered):-1:1);
f1 = filtfilt(filternum,filterdenom,data_to_be_filtered);
f2 = f1(length(f1):-1:1);
f3 = filtfilt(filternum,filterdenom,f2);
filtered_data = f3(length(f3):-1:1);
```

C17. Zero Grand Reaction Forces

Program that zeroes the Bertec force plate's initial forces and moments using the first second of each trial's data, before any forces or moments were applied to the force plate.

```
% fixgrfoffset.m
% Input grf_data
% Output grf data with a zero reading in the beginning
% Started 3/5/03, Emily Lloyd
% Last Revised: 3/5/03, Emily Lloyd

ave(1) = mean(grf_data(26:50,2));
ave(2) = mean(grf_data(26:50,3));
ave(3) = mean(grf_data(26:50,4));
ave(4) = mean(grf_data(26:50,5));
ave(5) = mean(grf_data(26:50,6));
ave(6) = mean(grf_data(26:50,7));

for counter = 1:1000
    grf_data(counter,2) = grf_data(counter,2) - ave(1);
    grf_data(counter,3) = grf_data(counter,3) - ave(2);
    grf_data(counter,4) = grf_data(counter,4) - ave(3);
    grf_data(counter,5) = grf_data(counter,5) - ave(4);
    grf_data(counter,6) = grf_data(counter,6) - ave(5);
    grf_data(counter,7) = grf_data(counter,7) - ave(6);
end
```

C18. Down Sample Grand Reaction Force Data

Function to down sample grand reaction force data so that it matches the frequency and data collection times of the marker data.

```
function [final_data] = down_sample_grf(given_data)
% Program to bring frequency of grf (1000 Hz) down to frequency of kin (200 Hz)
% Input a grf data file
% Output grf data with 1/5 of original points
% Started 9/12/02, Emily Lloyd
% Last Revised: 9/19/02, Emily Lloyd

for counter = 0:999
   for column = 1:17
      final_data(counter + 1,column) = given_data(5*counter + 1,column);
   end
end
```

C19. Calculate Center of Pressure on the Bertec Force Plate

Program to find the center of pressure location necessary for the inverse dynamics analysis.

```
% Find_COP_Bertha.m
% Program to find center of pressure on Bertha
% Input a data file
% Output calculations
% Started 7/24/02, Emily Lloyd
% Last Revised: 3/6/03, Emily Lloyd

for index = 1:1000
    if grf_data(index,4)>(44.5)          % if Fz on Bertha is greater than 44.5 Newtons, the COP exists
       COPexists(index) = 1;
    else COPexists(index) = 0;
    end
end

% set xCOP and yCOP to 0 initially
xCOP = zeros(1000,1);
yCOP = zeros(1000,1);

% Find COP
for index = 1:1000
    if COPexists(index)==1    % MULTIPLIED BY -1 TO TAKE INTO ACCOUNT THAT GCS FORCES AND
MOMENTS ARE REACTIONARY
       xCOP(index) = grf_data(index,6)./grf_data(index,4);          % x = My / -Fz for GCS
       yCOP(index) = grf_data(index,5)./grf_data(index,4);          % y = Mx / Fz for GCS
    end
end
```

C20. Calculate Leg Angles

Program to calculate segment (foot = theta12, shank = theta34, thigh = theta45, HAT(Head, Arms, and Torso) = theta56) and joint angles.

```
% Find_leg_angles2.m
% Calculates ankle, knee, and hip angles
% Started 9/3/02, Emily Lloyd
% Last Revised: 5/22/03, Emily Lloyd

% Calculate angles made with horizontal

for counter = 1:1000
    theta56(counter) = atan2((kin_data(counter,18) - kin_data(counter,15)),(kin_data(counter,17) -
kin_data(counter,14)));
    theta45(counter) = atan2((kin_data(counter,15) - kin_data(counter,12)),(kin_data(counter,14) -
kin_data(counter,11)));
    theta34(counter) = atan2((kin_data(counter,12) - kin_data(counter,9)),(kin_data(counter,11) - kin_data(counter,8)));
    theta12(counter) = atan((kin_data(counter,6) - kin_data(counter,3))/(kin_data(counter,5) - kin_data(counter,2)));
    hip_angle(counter) = -1 * (theta56(counter) - theta45(counter));
    knee_angle(counter) = theta45(counter) - theta34(counter);
    ankle_angle(counter) = -1 * (theta34(counter) - theta12(counter) + pi/2);
end
```

C21. Calculate Center of Mass Positions

Program to calculate the position of the center of mass of the foot, shank, thigh, and HAT.

```
% Find_COM_locations3.m
% Calculates COM of foot, shank, and thigh
% Started 3/4/02, Emily Lloyd
% Last Revised: 3/4/03, Emily Lloyd

for counter = 1:1000

   % thigh y location
   COM_thigh(counter,1) = kin_data(counter,14) + com(3)*(kin_data(counter,11) - kin_data(counter,14));

   % thigh z location
   COM_thigh(counter,2) = kin_data(counter,15) + com(3)*(kin_data(counter,12) - kin_data(counter,15));

   % shank y location
   COM_shank(counter,1) = kin_data(counter,11) + com(2)*(kin_data(counter,8) - kin_data(counter,11));

   % shank z location
   COM_shank(counter,2) = kin_data(counter,12) + com(2)*(kin_data(counter,9) - kin_data(counter,12));

   % foot y location
   COM_foot(counter,1) = kin_data(counter,8) + com(1)*(kin_data(counter,2) - kin_data(counter,8));

   % foot z location
   COM_foot(counter,2) = kin_data(counter,9) + com(1)*(kin_data(counter,3) - kin_data(counter,9));

end
```

C22. Calculate Linear and Rotational Accelerations

Program to calculate the linear accelerations of the segment center of masses and the rotational accelerations of the segments.

```
% Find_COM_acc.m
% Calculates linear and angular COM of foot, shank, and thigh
% Started 9/5/02, Emily Lloyd
% Last Revised: 3/4/03, Emily Lloyd

% Linear acceleration
for counter = 2:999
    a_thigh(counter,1) = (COM_thigh(counter + 1,1) - 2*COM_thigh(counter,1) + COM_thigh(counter -
1,1))/(0.005^2); % y
    a_thigh(counter,2) = (COM_thigh(counter + 1,2) - 2*COM_thigh(counter,2) + COM_thigh(counter -
1,2))/(0.005^2); % z
    a_shank(counter,1) = (COM_shank(counter + 1,1) - 2*COM_shank(counter,1) + COM_shank(counter -
1,1))/(0.005^2); % y
    a_shank(counter,2) = (COM_shank(counter + 1,2) - 2*COM_shank(counter,2) + COM_shank(counter -
1,2))/(0.005^2); % z
    a_foot(counter,1) = (COM_foot(counter + 1,1) - 2*COM_foot(counter,1) + COM_foot(counter - 1,1))/(0.005^2); %
y
    a_foot(counter,2) = (COM_foot(counter + 1,2) - 2*COM_foot(counter,2) + COM_foot(counter - 1,2))/(0.005^2); %
z
end

% Angular acceleration
for counter = 2:999
    alpha_hip(counter) = (hip_angle(counter + 1) - 2*hip_angle(counter) + hip_angle(counter - 1))/(0.005^2);
    alpha_knee(counter) = (knee_angle(counter + 1) - 2*knee_angle(counter) + knee_angle(counter - 1))/(0.005^2);
    alpha_ankle(counter) = (ankle_angle(counter + 1) - 2*ankle_angle(counter) + ankle_angle(counter - 1))/(0.005^2);
    alpha_12(counter) = (theta12(counter + 1) - 2*theta12(counter) + theta12(counter - 1))/(0.005^2);
    alpha_34(counter) = (theta34(counter + 1) - 2*theta34(counter) + theta34(counter - 1))/(0.005^2);
    alpha_45(counter) = (theta45(counter + 1) - 2*theta45(counter) + theta45(counter - 1))/(0.005^2);
    alpha_56(counter) = (theta56(counter + 1) - 2*theta56(counter) + theta56(counter - 1))/(0.005^2);
end

% Account for counter = 1 and counter = 1000
counter = 1;
a_thigh(counter,1) = 0; % y
a_thigh(counter,2) = 0; % z
a_shank(counter,1) = 0; % y
a_shank(counter,2) = 0; % z
a_foot(counter,1) = 0; % y
a_foot(counter,2) = 0; % z
alpha_hip(counter) = 0;
alpha_knee(counter) = 0;
alpha_ankle(counter) = 0;
alpha_12(counter) = 0;
alpha_34(counter) = 0;
alpha_45(counter) = 0;
alpha_56(counter) = 0;

counter = 1000;
a_thigh(counter,1) = 0; % y
a_thigh(counter,2) = 0; % z
a_shank(counter,1) = 0; % y
a_shank(counter,2) = 0; % z
a_foot(counter,1) = 0; % y
a_foot(counter,2) = 0; % z
```

```
alpha_hip(counter) = 0;
alpha_knee(counter) = 0;
alpha_ankle(counter) = 0;
alpha_12(counter) = 0;
alpha_34(counter) = 0;
alpha_45(counter) = 0;
alpha_56(counter) = 0;


for counter = 1:1000
    time(counter,1) = (counter)*0.005; % check this for accurate timing
end
```

C23. Calculate Joint Torques

Program that calculates joint torques of the ankle, knee, and hip of the stepping leg using inverse dynamics. This program outputs plots of the torque normalized to the product of height and mass and the absolute torque.

```
% Find_torques2.m
% Input masses, accelerations
% Calculate joint torques
% Started 9/12/02, Emily Lloyd
% Last Revised: 5/7/03, Emily Lloyd

% Programmed for stepping with right foot, but should work for both

g = 9.81;   % N/ms^2
COPz = 0;   % COP on Bertha is at surface, where z = 0

% masses, COM, and I for body segments
m_foot = sm(1);
I_foot = momentofinertia(1);

m_shank = sm(2);
I_shank = momentofinertia(2);

m_thigh = sm(3);
I_thigh = momentofinertia(3);

% Find ankle torques
% Use FBD of foot          % zCOP = 0

for index = 1:1000
   if COPexists(index) == 0
      Ry_ankle(index) = m_foot * a_foot(index,1);
      Rz_ankle(index) = m_foot * (a_foot(index,2) + g); % R = ma + mg
      d1 = COM_foot(index,1) - kin_data(index,8);
      d2 = kin_data(index,9) - COM_foot(index,2);
      M_ankle(index) = I_foot * alpha_12(index) + Rz_ankle(index) * d1 + Ry_ankle(index) * d2;
   else
      Ry_ankle(index) = m_foot * a_foot(index,1) - grf_data(index,3);
      Rz_ankle(index) = m_foot * (a_foot(index,2) + g) - grf_data(index,4); % R = ma + mg - Rzbertha
      d1 = COM_foot(index,1) - kin_data(index,8);
      d2 = kin_data(index,9) - COM_foot(index,2);
      d3 = yCOP(index) - COM_foot(index,1);
      d4 = COM_foot(index,2) - COPz;
      M_ankle(index) = I_foot * alpha_12(index) + Rz_ankle(index) * d1 + Ry_ankle(index) * d2 - ...
         grf_data(index,4) * d3 - grf_data(index,3) * d4;
   end
end

% Find knee torques
% Use FBD of shank

for index = 1:1000
   Ry_knee(index) = m_shank * a_shank(index,1) + Ry_ankle(index);
   Rz_knee(index) = m_shank * (a_shank(index,2) + g) + Rz_ankle(index);   % R = ma + mg + Rzankle
   M_knee(index) = I_shank * alpha_34(index) - Rz_knee(index) * (kin_data(index,11) - COM_shank(index,1)) + ...
      Ry_knee(index) * (kin_data(index,12) - COM_shank(index,2)) + M_ankle(index) - ...
      Rz_ankle(index) * (COM_shank(index,1) - kin_data(index,8)) + Ry_ankle(index) * (COM_shank(index,2) -
kin_data(index,9));
end
```

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```
% Find hip torques
% Use FBD of thigh

for index = 1:1000
    Ry_hip(index) = m_thigh * a_thigh(index,1) + Ry_knee(index);
    Rz_hip(index) = m_thigh * (a_thigh(index,2) + g) + Rz_knee(index);      % R = ma + mg + Rzknee
    M_hip(index) = I_thigh * alpha_45(index) + Rz_hip(index) * (COM_thigh(index,1) - kin_data(index,14)) + ...
        Ry_hip(index) * (kin_data(index,15) - COM_thigh(index,2)) + M_knee(index) + ...
        Rz_knee(index) * (kin_data(index,11) - COM_thigh(index,1)) + Ry_knee(index) * (COM_thigh(index,2) -
kin_data(index,12));
end


for index = 1:1000
    M_ankle_norm(index) = M_ankle(index)/(height*weight);
    M_knee_norm(index) = M_knee(index)/(height*weight);
    M_hip_norm(index) = M_hip(index)/(height*weight);
end

max_z = max(grf_data(:,4));
for index = 1:1000
    grf_data_outline(index) = grf_data(index,4)/max_z;
end

figure(1)
plot(1:1000,M_ankle_norm,1:1000,COPexists,1:1000,grf_data_outline)
ylim([-2 2]);
title('Ankle Normalized Torque')
legend('Torque','COP Exists','GRF z data normalized')

figure(2)
plot(1:1000,M_ankle,1:1000,COPexists,1:1000,grf_data_outline)
title('Ankle Torque (Nm)')
legend('Torque','COP Exists','GRF z data normalized')

figure(3)
plot(1:1000,M_knee_norm,1:1000,COPexists,1:1000,grf_data_outline)
ylim([-2 2]);
title('Knee Normalized Torque')
legend('Torque','COP Exists','GRF z data normalized')

figure(4)
plot(1:1000,M_knee,1:1000,COPexists,1:1000,grf_data_outline)
title('Ankle Torque (Nm)')
legend('Torque','COP Exists','GRF z data normalized')

figure(5)
plot(1:1000,M_hip_norm,1:1000,COPexists,1:1000,grf_data_outline)
ylim([-2 2]);
title('Hip Normalized Torque')
legend('Torque','COP Exists','GRF z data normalized')

figure(6)
plot(1:1000,M_hip,1:1000,COPexists,1:1000,grf_data_outline)
title('Hip Torque (Nm)')
legend('Torque','COP Exists','GRF z data normalized')
```

C24. Find Peak Torque Values

Program to find the maximum and minimum normalized (to the product of height and mass) joint torque, absolute torque, and index of the data at which this peak torque occurs given a start and end point by the user.

```
% Find_maxes.m (M_ankle_norm,M_ankle,M_knee_norm,M_knee,M_hip_norm,M_hip)
% Program to find maximum torque given a range containing the maximum torque
% Input joint moments
% Output max joint moments
% Started 5/21/02, Emily Lloyd
% Last Revised: 5/22/03, Emily Lloyd

%%% ANKLE

% Identify range for ankle moment

xbegin_ank = input('Enter index of first x point in ankle range: ');
xend_ank = input('Enter index of last x point in ankle range: ');

% Find maximum normalized ankle moment, that index, then ankle moment

max_ank_norm = M_ankle_norm(xbegin_ank);
min_ank_norm = M_ankle_norm(xbegin_ank);
max_ank_index = xbegin_ank;
min_ank_index = xbegin_ank;
for index_ank = xbegin_ank:xend_ank
   if M_ankle_norm(index_ank) > max_ank_norm
      max_ank_norm = M_ankle_norm(index_ank);
      max_ank_index = index_ank;
   end
   if M_ankle_norm(index_ank) < min_ank_norm
      min_ank_norm = M_ankle_norm(index_ank);
      min_ank_index = index_ank;
   end
end

max_ank = M_ankle(max_ank_index);
min_ank = M_ankle(min_ank_index);

% Output

max_ank_norm
max_ank
max_ank_index
min_ank_norm
min_ank
min_ank_index

%%% KNEE

% Identify range for knee moment

xbegin_k = input('Enter index of first x point in knee range: ');
xend_k = input('Enter index of last x point in knee range: ');

% Find maximum normalized knee moment, that index, then knee moment

max_k_norm = M_knee_norm(xbegin_k);
min_k_norm = M_knee_norm(xbegin_k);
```

```
max_k_index = xbegin_k;
min_k_index = xbegin_k;
for index_k = xbegin_k:xend_k
   if M_knee_norm(index_k) > max_k_norm
     max_k_norm = M_knee_norm(index_k);
     max_k_index = index_k;
   end
   if M_knee_norm(index_k) < min_k_norm
     min_k_norm = M_knee_norm(index_k);
     min_k_index = index_k;
   end
end

max_k = M_knee(max_k_index);
min_k = M_knee(min_k_index);

% Output

max_k_norm
max_k
max_k_index
min_k_norm
min_k
min_k_index

%%% HIP

% Identify range for hip moment

xbegin_h = input('Enter index of first x point in hip range: ');
xend_h = input('Enter index of last x point in hip range: ');

% Find maximum normalized hip moment, that index, then hip moment

max_h_norm = M_hip_norm(xbegin_h);
min_h_norm = M_hip_norm(xbegin_h);
max_h_index = xbegin_h;
min_h_index = xbegin_h;
for index_h = xbegin_h:xend_h
   if M_hip_norm(index_h) > max_h_norm
     max_h_norm = M_hip_norm(index_h);
     max_h_index = index_h;
   end
   if M_hip_norm(index_h) < min_h_norm
     min_h_norm = M_hip_norm(index_h);
     min_h_index = index_h;
   end
end

max_h = M_hip(max_h_index);
min_h = M_hip(min_h_index);

% Output

max_h_norm
max_h
max_h_index
min_h_norm
min_h
min_h_index
```

Appendix D: Data for Study #1

Maximum lean results and modes of failure are presented here.

Table D-1. Record of identification of dominance, failure mode, and dominance from trip recovery results.

	Dominant		Maximum Lean (%BW)				
Subject	Hand	Foot	Right	Mode of Failure	Left	Mode of Failure	Dominance
1	R	R	36	Harness	24	Harness	Right
2	R	R	28	Harness	32	Multistep	Left
3	R	R	32	Harness	36	Multistep	Left
4	R	R	32	Harness	36	Harness	Left
5	R	R	40	Harness	32	Harness	Right
6	R	R	44	Harness	40	Harness	Right
7	R	R	36	Multistep	28	Harness	Right
8	R	R	28	Harness	32	Harness	Left
9	R	R	36	Harness	40	Harness	Left
10	R	R	32	Harness	24	Harness	Right

Older Subjects

	Dominant		Maximum Lean (%BW)				
Subject	Hand	Foot	Right	Mode of Failure	Left	Mode of Failure	Dominance
11	R	R	24	Multistep	24	Multistep	None
12	R	R	16	Multistep	16	Multistep	None
13	R	R	12	Multistep	20	Multistep	Left
14	R	R	36	Multistep	32	Multistep	Right
15	R	R	20	Harness	24	Multistep	Left
16	R	R	20	Multistep	20	Multistep	None
17	R	R	16	Multistep	24	Multistep	Left
18	R	R	20	Multistep	12	Multistep	Right
19	R	R	16	Multistep	20	Multistep	Left
20	R	R	<12	Multistep	<12	Multistep	Unknown

Appendix E: Data for Study #2

Peak extensor torques values are presented here. Only 19 subjects have reported joint torques as one older subject did not recover from the smallest lean magnitude.

Table E-1. Peak ankle plantarflexor torques

%BW	Subject	12	16	20	24	28	32	36	40	44
Young Subjects	6	0.5183	0.6598	0.5073	0.8518	1.0251	0.9945	0.6077	0.7322	**
	5	0.0616	0.4640	0.4116	0.3861	0.6445	0.7238	0.6848	0.7822	
	7	0.5789	0.6256	*	0.7958	0.6270	0.7768	0.7451		
	1	0.2449	0.4277	0.7138	0.5677	0.4503	0.6692	0.6298		
	9	0.3201	0.5057	**	**	0.6474	**	0.5333		
	3	*	0.1500	0.5591	0.3137	0.6040	0.4982			
	4	**	**	**	**	0.7076	0.6423			
	10	*	*	*	*	*	*			
	2	*	0.5713	0.7874	0.7096	0.8623				
	8	0.4294	0.1861	0.3566	0.2772	0.4705				
Average		0.3589	0.4488	0.5560	0.5574	0.6710	0.7175	0.6401	0.7572	***
Standard Deviation		0.1740	0.1778	0.1535	0.2184	0.1699	0.1508	0.0715	0.0250	***
Older Subjects	14	0.4552	0.3704	0.4575	0.5177	0.6233	0.7599	0.7819		
	11	0.4624	0.2131	0.4499	0.4943					
	15	0.5158	0.4709	0.7967						
	16	0.6453	0.5378	0.6529						
	12	0.4949	0.5917							
	17	0.3853	0.6221							
	18	0.4582	0.6025							
	19	0.7429	0.7775							
	13	0.2963								
Average		0.4951	0.5233	0.5893	0.5060	0.6233	0.7599	0.7819		
Standard Deviation		0.1323	0.1723	0.1672	0.0165	***	***	***		
* Data corresponding to a trial with hip flexion										
** Successful trial but torque calculations not possible due to incomplete data										
*** Calculations impossible due to insignificant sample size										

Table E-2. Peak knee extensor torques.

%BW	Subject	12	16	20	24	28	32	36	40	44
Young Subjects	6	0.7613	0.6310	0.8778	0.8937	1.0057	0.9463	1.0439	0.9596	**
	5	0.4695	0.5339	0.6957	0.7933	0.7520	0.4831	0.6128	0.6064	
	7	0.8669	0.9209	*	0.8371	1.1253	1.0676	1.0557		
	1	0.4210	0.6639	0.6934	0.6824	0.9786	0.8037	0.7930		
	9	0.5076	0.3425	**	**	0.6327	**	0.7006		
	3	*	1.1460	1.0581	1.0491	0.9925	1.0290			
	4	**	**	**	**	0.6379	0.8106			
	10	*	*	*	*	*	*			
	2	*	0.9226	0.5561	0.8511	0.8976				
	8	0.9100	0.8170	0.7573	0.8616	1.2217				
Average		2.5044	2.6451	3.9802	4.1581	3.8914	5.3058	6.7010	13.8553	***
Standard Deviation		3.9745	4.7992	6.7316	7.6571	8.1277	10.8994	13.1039	18.4876	***
Older Subjects	14	0.5641	0.9340	0.7562	0.6745	0.7938	0.6447	0.6493		
	11	0.2668	0.5185	0.6081	0.6851					
	15	0.6341	0.6863	0.7758						
	16	0.3807	0.4234	0.3888						
	12	0.2828	0.3765							
	17	0.5687	0.6192							
	18	0.3091	0.5216							
	19	0.2921	0.3195							
	13	0.4607								
Average		0.8081	1.0220	1.8521	3.0056	4.4607	5.7720	6.8766		
Standard Deviation		1.1958	1.4284	2.7321	4.0284	***	***	***		

* Data corresponding to a trial with hip flexion

** Successful trial but torque calculations not possible due to incomplete data

*** Calculations impossible due to insignificant sample size

Table E-3. Peak hip extensor torques.

%BW	Subject	12	16	20	24	28	32	36	40	44
Young Subjects	6	0.3387	0.3258	0.3524	0.7168	0.9224	0.7243	1.0028	0.9516	**
	5	0.2123	0.2375	0.2448	0.2886	0.5512	0.8726	0.8520	1.1245	
	7	0.4469	0.2555	*	0.3329	0.3972	0.5057	0.4724		
	1	0.2682	0.5483	0.8664	0.7521	0.7264	0.9203	0.7578		
	9	0.4573	0.6342	**	**	1.0975	**	1.3108		
	3	*	0.1494	0.2181	0.3833	0.4978	0.8782			
	4	**	**	**	**	0.7594	0.7314			
	10	*	*	*	*	*	*			
	2	*	0.7915	0.6003	0.7909	0.8425				
	8	0.4468	0.3840	0.6835	0.5758	0.7889				
Average		0.3617	0.4158	0.4943	0.5486	0.7315	0.7721	0.8792	1.0381	***
Standard Deviation		0.1051	0.2217	0.2624	0.2124	0.2196	0.1538	0.3092	0.1223	***
Older Subjects	14	0.3835	0.4858	0.5064	0.5317	0.9808	1.1472	1.0862		
	11	0.4537	0.3790	0.5423	0.5579					
	15	0.3449	0.4248	0.7140						
	16	0.4753	0.4783	0.6121						
	12	0.7551	0.9100							
	17	0.5085	0.7297							
	18	0.6140	0.8957							
	19	0.6350	0.9222							
	13	0.1720								
Average		0.4824	0.6532	0.5937	0.5448	0.9808	1.1472	1.0862		
Standard Deviation		0.1738	0.2357	0.0914	0.0185	***	***	***		

* Data corresponding to a trial with hip flexion

** Successful trial but torque calculations not possible due to incomplete data

*** Calculations impossible due to insignificant sample size

Vita

Emily Marie Lloyd

Emily Marie Lloyd was raised in Alexandria, Virginia. She attended Saint Clement Day School for first and second grade and subsequently Alexandria City Public Schools until her high school graduation in June 1998. While in junior high and high school she served on the Board of Directors of a local children's charity. As an undergraduate at Virginia Tech, she founded a chapter of the American Indian Science and Engineering Society (AISES). She interned at Ford Motor Company in the summers of 2000 and 2001. In May 2002 she graduated from Virginia Tech with a bachelor's degree in Mechanical Engineering. She was certified as a Basic Emergency Medical Technician in August of 2002. She became an Engineer in Training in June of 2003.