The Effects of Fatigue on Lower Extremity Kinetics and Kinematics in Subjects with Known Ankle Instability

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The Effects of Fatigue on Lower Extremity Kinetics and Kinematics in Subjects with Known Ankle Instability

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

By

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List of Abbreviations

SEBT = star excursion balance test
CAIT = Cumberland Ankle Instability Tool
GRF = ground reaction force
GRFv = the vertical (or z) component of the ground reaction force
LE = lower extremity
CAI = Chronic Ankle Instability
FAI = Functional Ankle Instability
MAI = Mechanical Ankle Instability
FLA = fatiguing landing activity
ATFL = anterior talofibular ligament
PTFL = posterior talofibular ligament
CFL = calcaneofibular ligament
COP = center of pressure
EMG = electromyogram
EM = electromagnetic
COM = center of mass
ICC = intraclass correlation coefficient
LMM = linear mixed model
GEE = generalized estimating equations
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Abstract

THE EFFECTS OF FATIGUE ON LOWER EXTREMITY KINETICS AND KINEMATICS IN SUBJECTS WITH KNOWN ANKLE INSTABILITY

By Lindsay E. Clayton, M.S.

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

Virginia Commonwealth University, 2015

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The goal of this study was to evaluate biomechanical differences between healthy subjects and those with ankle instability during the gradual onset of lower extremity fatigue from a landing activity. An understanding of these differences is needed in order to prevent future injury to or further debilitation in individuals with ankle instability. A functional fatiguing activity was designed to focus fatigue on the quadriceps muscles, as those are the muscles most frequently fatigued during sport. Measures were taken throughout the progression of fatigue with a force plate and a motion tracking system and included vertical ground reaction force and lower extremity kinetics, kinematics, and energetics. The time required to reach self-reported fatigue and a balance assessment, the Star Excursion Balance Test, before and after the onset of fatigue was also recorded. Significant differences were observed between groups in peak ground reaction force, ground reaction force impulse, and frontal plane ankle joint impulse.
Results indicated that subjects with ankle instability not only exhibited a different baseline for most measurements than normal subjects, but also managed the progression of fatigue differently. With this information and information from further studies, recommendations and/or training schemes could be made and implemented to help those with ankle instability avoid recurrent injuries.
Chapter 1 - Introduction

The ankle sprain is one of the most common musculoskeletal injuries, with over 20,000 sprains occurring per day in the United States.\textsuperscript{1,2} The American College of Sports Medicine (ACSM) says that this is especially the case during sporting activities, where about fifty percent of all injuries that take place are ankle sprains.\textsuperscript{1} Because of their frequent occurrence, these sprains are often seen as inconsequential. Hertel reports that more than half of people who suffer an ankle sprain will not seek professional medical attention.\textsuperscript{2} The ACSM estimates that 40 percent of those patients who do seek care do not fully recover either due to misdiagnosis or improper treatment.\textsuperscript{1} Previous research has shown that, when left untreated or inadequately treated, a number of deficits, including instability in the ankle joint, may arise.\textsuperscript{2-5} It is important to understand how these deficits may affect patients, particularly in the case of athletes returning to high levels of physical activity, in order to prevent future injury or further debilitation.

Some type of jumping and landing are often performed in a variety of sports. Epidemiological studies have shown that ankle sprains have the highest occurrence in indoor-court sports, like basketball and volleyball, where jumps and landings are important skills.\textsuperscript{3,15} Previous studies have examined the lower extremity (LE) mechanics of landing due to the higher likelihood of injury resulting from increased loading rates and magnitudes.\textsuperscript{6} Various influences, including landing techniques, joint stiffness, drop height, the type of sport participation, gender, and fatigue, have all been investigated to determine their effect on
potential injury mechanisms.\textsuperscript{7-11} Biomechanical differences in subjects with instability in the ankle from past ankle sprain or sprains have also been studied during landing activities.\textsuperscript{12-14}

To our knowledge, no studies have examined changes in biomechanical variables in subjects with ankle instability over the course of an LE fatiguing landing protocol. This study built upon previous works by expanding fatiguing studies to populations with ankle instability. The purpose of this study was to evaluate landing biomechanics differences between healthy subjects and those with ankle instability during the programmatic onset of lower extremity fatigue. Objective measures included ground reaction forces (GRF) as well as LE kinetics, kinematics, and energetics. The onset of fatigue was self-reported, however changes in the subjects’ balance before and after a fatiguing activity was also measured. The goals of this study were to determine if people with ankle instability differed from healthy subjects in the force or speed of their landing or in the flexion of LE joints as they accept the force from that landing. This study also attempted to reveal potential changes in internal mechanics of the LE joints of the two different populations, including internal forces acting on each joint and work performed by those joints, during the progression of fatigue that might predispose those with instability to future injuries.

**Ankle Instability**

Ankle sprains occur when the ligaments of the ankle are stretched or torn. They are graded on a one to three scale of increasing severity and are often accompanied by some combination of pain, swelling, or bruising.\textsuperscript{1} Damage to the lateral ligaments of the ankle is most common and comprises about 80 percent of all ankle sprains.\textsuperscript{16} At between 30 and 50 percent,
lateral sprains also have the highest rates of development of residual symptoms and chronic conditions, like instability.\textsuperscript{2,3,5,16,19}

Lateral ankle sprains are frequently called inversion or supination sprains, named after the mechanism of injury. Inversion describes a movement in the frontal plane where the plantar surface of the foot rotates to face medially. The true ankle joint, the talocrural, is a hinge joint at the conjunction of the distal ends of the tibia and fibula and the superior aspect of the talus. This joint, in addition to distal articulations, becomes capable of more movement beyond simple flexion and extension movements in the sagittal plane.\textsuperscript{17} The ankle is described as a “complex” with articulations stemming mainly from three different joints: the talocrural joint, subtalar joint, and distal tibiofibular syndesmosis. Hertel\textsuperscript{2} describes supination in the open chain as a combination of plantar flexion, inversion, and medial rotation of the foot. Planatar flexion is the movement of the plantar surface of the foot away from the shank. During this motion, the talus rotates clockwise, when viewed from the right, away from the anterior surface of the leg, gliding past the inferior aspects of the tibia and fibula. Inversion, the medial rotation of the plantar surface of the foot, causes a gliding or rotation of the calcaneus past the talus in the frontal plane. Adduction is the medial rotation of the foot in the transverse plane and occurs between the talus and calcaneus. The ligaments that would prevent excess movement in this fashion, the talofibular ligaments (posterior and anterior) and the calcaneofibular ligament on the lateral side, are those typically affected either alone or in combination.\textsuperscript{2,17,18} The malleoli of the tibia and fibula provide structural stability to the talocrural joint by locking the talus into place with its mortise-like structure. The axis of rotation for the joint extends from lateral malleolus to the medial malleolus. Because the
medial malleolus is superior to its lateral counterpart, the axis lies at eight degrees off the transverse plane. During loading, the ankle is vulnerable to inversion because the foot favors a perpendicular alignment with this axis of rotation. This alignment is what makes inversion sprains so common, no matter if injury is caused by poor landing, an uneven surface, or a cutting maneuver.

Most sprains are treated without surgery, no matter the grade, with the main concern being to alleviate the short-term symptoms of pain and swelling. A varying period of immobilization, depending on severity, is recommended in conjunction with the local application of ice and compression and with elevation of the limb. A study by Freeman examined the presence of a lasting instability in the ankle one year post incident in subjects that had been treated with elastic bracing and physical therapy, immobilization for six weeks in a below knee cast, or surgical ligament repair. The researchers used various stress radiographs to determine if those patients who complained of instability had observable mechanical deficits. In the stress radiographs, clinicians move the ankle into certain positions and place manual tension on the joint while taking an x-ray. By examining the gapping between bones created by the applied tension, the researcher could judge if the ankle joint was mechanically unstable (larger gaps observed) or if the separation between the bones was within normal limits. In about a third of their cases, the ankle appeared healthy clinically and in the radiographs but the patient was experiencing unexpected inversions during activity. Of the twelve treated by bracing, five experienced these inversions. Of the sixteen treated by immobilization, six experiences lasting instability. Of the fourteen treated by immobilization, nine still had an incident of unexpected inversion. In a more recent study by Konradsen et
648 subjects who had suffered a lateral ankle sprain were interviewed seven years post injury regarding residual symptoms. Thirty-two percent of patients reported chronic pain, swelling, or incidences of recurrent sprains. Three quarters of these patients felt that these conditions negatively affected their activity levels. Nineteen percent of all of the subjects had experienced three or more sprains a year since the inciting incident, or felt that they would have experienced three or more sprains without the use of external support. In attempts to better understand the potential pathologies behind ankle instability, researchers in the field have created two different classes of instability, mechanical and functional, both falling under the broader term of Chronic Ankle Instability (CAI).

**Chronic Ankle Instability**

Mechanical ankle instability (MAI) results from clinically observable changes to the anatomy of the ankle after a sprain. Hertel discusses several different changes that are seen in the mechanically unstable ankle alone or in combination in his instability paradigm. These changes include degenerative and synovial changes, arthrokinematic restrictions, and pathological laxity. Degenerative and synovial changes can present as lesions, hypertrophy, pain, and inflammation. Arthrokinematic impairments describe any kind of improper, small movement at the joint surface. This usually presents as a subluxation or decreased range of motion. Pathological laxity is due to damage to the structure of the ligaments, whether through stretching or tearing. As injury to the anterior talofibular ligament (ATFL) and the calcaneofibular ligament (CFL) are most common, the laxity can usually be found in the talocrural or subtalar joints, if any laxity exists.
There are several methods that can be used to test for mechanical deficits. For ATFL laxity, an anterior drawer test can be performed and a distal fibular positional change measurement can be taken.\textsuperscript{2,23} The anterior drawer test is performed by placing the ankle with a suspected laxity into slight plantar flexion (about 15 degrees) and pulling forward on the foot from the calcaneus while stabilizing the supine patient’s leg.\textsuperscript{24} Deficits in the ATFL and the CFL will present as abnormal anterior movement of the foot away from the leg.\textsuperscript{2} Parasher \textit{et al.}\textsuperscript{23} have suggested that the use of a plastic goniometer with a fixed arm to ensure identical plantar flexion degree produces excellent inter-tester reliability when the anterior displacement of the foot is quantified with a pair of calipers. Distal fibular position is measured as the difference in distances from the same posterior position to the most anterior edge of tibial and fibular malleoli. A laxity in the supporting structures of the talocrural joint can result in displacement of the distal fibula. Alternatively, a slight displacement or dislocation of the inferior fibula, classified as an arthrokinematic restriction, can create slack in the lateral ligaments and allow for excess and potentially harmful range of motion.\textsuperscript{2} Talar tilt and subtalar glide tests are used to assess the amount of ATFL, CFL, or PTFL laxity effect in the frontal plane.\textsuperscript{2} In the talar tilt test, the distal leg is stabilized and the tester holds the calcaneus, ensure the foot is in slight plantar flexion, and invert the talus. Medial translation or gapping can indicate laxity in the CFL. Repeating the test in full plantar flexion or full dorsiflexion will reflect injury in the AFTL or PTFL respectively. Subtalar glide uses a very similar grip to the talar tilt, but the patient is lying on their side contralateral to the ankle to be tested instead of in a supine position. The plantar surface of the foot is kept level as the calcaneus is moved medially and laterally instead of into inversion.\textsuperscript{24} The presence and location of pain during physical manipulations is also an
important indicator of which ligaments or joints may be impacted.\textsuperscript{2,24} It is recommended that all measures be confirmed with radiograph when available.\textsuperscript{2,23,25}

Freeman\textsuperscript{22} described functional ankle instability (FAI) as the sensation of the ankle giving way despite a normal clinical examination. He and his associates followed a group of patients for a year post-injury. Subjects were selected for the study if they had no history of ankle injury and if the ligament injured was on the lateral aspect of the ankle. Freeman’s\textsuperscript{22} research was very important because it was one of the first to definitely distinguish between mechanical and functional instability. He and his team used radiographs and a selection of the clinical tests for mechanical instability to determine that in a majority of the patients who complained of instability, said instability could not be attributed to muscle weakness in the calf, anterior-posterior instability of the talus, a weak spot in the ligament, or a separation in the distal tibiofibular syndesmosis. Subjects were physically more similar to healthy, uninjured patients than patients who had not suffered an injury but had congenital deficits.\textsuperscript{22}

Functional instability after a lateral ankle sprain is thought to result from neuromuscular deficits. In Hertel’s\textsuperscript{2} paradigm, deficits believed to be possible contributors to functional instability include weakness, diminished postural control, impaired neuromuscular control, and reduced proprioception. Postural control is any movement or compensation that acts to maintain, restore, or achieve an alignment with the line of gravity over the base of support.\textsuperscript{28} Postural control deficits have been observed both statically and dynamically.\textsuperscript{2,16,27} Reduced neuromuscular response time and nerve conduction velocity, both in the ankle and in other parts of the LE, are often found in FAI subjects. This seems to indicate that neuromuscular
deficiencies in the ankle may cause the central nervous system to adapt and manifest into irregularities in proximal LE joints.\(^2\) Reduced proprioception manifests as impaired senses of joint movement, position, and force production. The receptors, called proprioceptors, respond to stretch and are found in the skeletal muscle and connective tissue, including ligaments, tendons, and joint capsules.

Sekir et al.\(^{38}\) investigated most of these deficits and the effect of training as a means of improving function in participants with FAI. Subjects were given a number of tests before beginning an exercise regime. Strength measurements of the invertor and evertor muscles were taken using an isokinetic dynamometer. The following day, proprioception was measured on the same dynamometer. The researchers asked the participants to recreate certain degrees of inversion to gauge joint position sense. Participants also performed a one-legged stance to test balance control and several hopping tasks to provide information to the researchers on the level of potential impairment. For a period of six weeks following these tests, subjects took part in an exercise protocol consisting of inversion and eversion movements on the dynamometer. After all sessions had taken place, the above measurements were repeated and Sekir et al.\(^{38}\) found that all measures had improved.

Winter et al.\(^{39}\) also examined training on proprioceptive sense in a select group of athletes: speed skaters. Ankle injuries are common in speed skating, and loss of position sense and kinesthesia can severely affect performance in the sport. The authors used a proprioception measurement similar to that of Sekir et al.,\(^{38}\) but examined plantar and
dorsiflexion, pronation, and supination. Winter et al.\textsuperscript{39} found that kinesthesia during plantar flexion in individuals with FAI improved with isokinetic training.

Gribble and Robinson\textsuperscript{40} and Bullock-Saxon\textsuperscript{41} both observed changes in proximal joints that stemmed from ankle dysfunction. Gribble and Robinson\textsuperscript{40} took isokinetic measures of hip, knee, and ankle torques resulting from flexion and extension (dorsi- and plantar flexion in the ankle) in participants with instability and compared it to those same torques in healthy subjects. The authors found that there was reduced strength in ankle plantar flexion and in knee flexion and extension, but found no significant change at the hip.\textsuperscript{40} Bullock-Saxon\textsuperscript{41} looked at loss of sensation in the ankle and the consequences it might have had on hip function post sprain. The author found that, along with a loss of perception of external vibration in the ankle, there was a delay in the activation in the gluteus maximus muscle at the hip.\textsuperscript{41} Both studies indicated that neuromuscular control may be affected more than just locally, but cause and effect could not be specifically pinpointed. They suggest that proximal alterations may be protective in nature.\textsuperscript{40}

Konradsen\textsuperscript{42} studied the motor control necessary to prevent or recover from sudden inversion, a common sprain mode. The author discusses the importance of proprioception in this mechanism during normal gait. If the individual cannot sense joint position before heel strike, the ankle may be over-inverted and the person would not be able to correctly load that limb. When this happens, there is a peripheral muscle response and usually compensation in the trunk and upper limbs to alter the center of mass (COM). Electromyogram (EMG) measures of electrical stimulation in the muscles show that muscles on the lateral aspect of the leg react
first and are followed later by activation of muscles in the proximal thigh. Konradsen\textsuperscript{42} reports that previous studies have observed a faster activation time of the evertor muscles in people with FAI when compared to those without.

Dundas \textit{et al.}\textsuperscript{43} examined neuromuscular control in participants with ankle instability, in those who had suffered a sprain but did not experience instability, and in uninjured participants as they were stepping down off of a curb. The authors found that there was significantly more tibialis anterior muscle activation immediately following landing and less plantar flexion in subject who had previously experienced a sprain, but not in those who suffered from resultant instability. This suggests a protective mechanism acting through lateral leg muscles.\textsuperscript{43}

Hertel \textit{et al.}\textsuperscript{29} assessed changes in postural control in subjects after lateral ankle sprain with the hope of being able to better predict recurrent ankle sprain and the risk of developing FAI. Center of pressure (COP) is a useful, objective measure in postural control studies because it is the point on the support surface (between the foot and floor) at which the ground reaction forces act. As people employ postural control techniques, the COP is the theoretical location at which all forces applied to the ground, needed in order to establish their line of gravity within the base of support, are summed. COP excursions are the movements that occur as those forces change. Hertel \textit{et al.}\textsuperscript{29} examined several variables associated with COP excursions; including length, amplitude, and velocity; as the subject maintained quiet, one-legged stance. Though this is considered a static measure, the COP is constantly shifting as the subject makes minute changes to maintain erect posture. The authors defined length as the total distance the COP traveled in each plane (frontal and sagittal) throughout the collection. Hertel \textit{et al.}\textsuperscript{29} used
root mean square velocity as the velocity of the COP is always changing. They found that there were significant changes in all variables when the subjects maintained stance on the injured ankle versus the uninjured foot with the injured limb exhibiting a higher COP length and velocity in both planes. The authors also found that there was some improvement, but were unable to determine if these improvements were a result of a learning effect or reduced deficiency.²⁹

Ross et al.³⁰ used additional measures to predict whether individuals had FAI or not. The authors also employed the static stance procedure. They calculated the arithmetic mean of the COP velocity as opposed to the quadratic mean. They also added the additional measurement of COP area ellipse. For the COP area measure, an ellipse that encloses 95% of all COP positions throughout the collection is fit over the COP excursions. A larger area is indicative of increased instability. The authors found that COP area measures were most reliable when it came to predicting that a subject had FAI.³⁰

There are several different methods for calculating the COP area, apart from the ellipse method. Wollseifen³¹ compared this method with convex hull approximation and a mean circle method area. The author used principal component analysis to determine the size of the best fit ellipse. In this method, the eigenvalues are calculated and the eigenvectors form the axes of the ellipse. In the convex hull method, all the COP excursions are enclosed by the closest fitting polygon possible. The total area of this polygon is considered the COP area. In the mean circle method, all COP points are encircled with the radius of each circle being equal to the distance from the instantaneous COP location to the mean location of all data. The average of all the
radii is calculated and the COP area is the area of the mean circle centered over the mean COP location. Wollseifen\textsuperscript{31} found that ellipse method and the convex hull method were comparable.

\textit{Star Excursion Balance Test}

While static measures are a good way to assess postural control, they may not be the best method to determine differences between subjects with and without FAI, as they do not closely match conditions in which someone with FAI might be most affected by their limitations. The Star Excursion Balance Test (SEBT) is a dynamic measure that is often used to assess shortcomings associated with FAI since it requires the integration of both strength and sensorimotor activity.\textsuperscript{32} The SEBT is performed while maintaining one-legged stance on the limb to be evaluated while the other limb reaches out in a specified direction. The individual must maintain full, constant, un-shifting contact of the stance limb without transferring weight to the reaching limb. The maximum distance the subject can reach while maintaining the stance criteria is recorded as the excursion distance. The maximum excursion is recorded in eight different directions each 45 degrees apart: anterior, anteromedial, medial, posteromedial, posterior, posterolateral, lateral, and anterolateral. Together, the excursion directions form the ‘star’ which is the root of the test’s name. While it was originally developed as a method of therapy, Gribble \textit{et al.}\textsuperscript{32} performed a review of the use of the SEBT as a tool to assess impairments in the LE. The authors were able to combine the reports of many researchers and give recommendations based on those findings for considerations when using the SEBT as a clinical assessment tool. They discussed the presence of a learning effect, which allowed subjects to reach further with each subsequent trial. They reported that there appeared to be a
plateau in this effect and recommend that subjects be given sufficient practice sessions before trials. They also found that all reach distances were highly relatable. This lead to the idea that it may not be necessary to evaluate the excursions in all 8 directions, but that a full picture could be extrapolated from performing the reach in the anterior, posteromedial, and posterolateral directions only. The authors also suggested normalizing for leg length to improve the ability to compare data across subjects.32

Reimann and Schmitz33 discussed the relationship between the different static measures and the SEBT in healthy subjects. The authors compared different modes of single leg quiet stance, including on a stable or on a shifting surface, to two function tests: the SEBT and a single leg hop-stabilization. They used the main outcome measures for each test: the average COP velocity for the static tests and maximum excursion distance for the SEBT. They found that there was no significant correlation between the static tests and the SEBT. They concluded that the inclusion of functional tests was important in assessing postural control and that an individual’s deficits may not be fully evaluated without such tests.

Gribble et al.32 discussed the use of the SEBT to diagnose LE deficiencies, rather than solely quantifying known conditions. They report that previous researchers had determined that subjects with ankle instability were neither able to reach as far as those without instability nor as far as their own self when standing on their injured limb versus their uninjured limb. However, twenty-five percent of studies reviewed did not report any differences between the normal and instability groups. Gribble et al.32 attributed this to improper selection criteria for those studies’ instability groups.
There is no true universally accepted definition for CAI. All agree that MAI is a pathologic finding, however previous studies have used varying criteria and have reported conflicting findings regarding the incidence of almost all modes of impairment in FAI and the effectiveness of training. FAI criteria have ranged from perceived to actual instability or giving way, to weakness, to loss of voluntary control of range of motion, to chronic pain, and more. Hertel’s CAI paradigm says that a combination of mechanical and functional ankle sprain will equal a predisposition for recurrent ankle sprain. Hiller et al. have said that this does not allow for a full understanding and that recurrent sprain does not necessarily stem from a combination of MAI and FAI, but can exist on its own. The authors proposed a new CAI model that could provide up to seven combinations of impairment from the overlap of three classes of resultant instability: mechanical instability, perceived instability, and recurrent sprains. Hiller et al. used an anterior drawer test to classify subjects as having MAI, a questionnaire to determine any perceived instability, and a defined recurrent sprain as a history of three or more sprains in the same ankle. The authors examined differences in balance and recovery from perturbation between all seven new categories. They found that their model fit all examined cases, as opposed to the Hertel paradigm which would have left 47 cases uncategorized. A number of researchers and clinicians in the field have formed the International Ankle Consortium and have attempted to address the problem of varying criteria by issuing an official position statement on inclusion criteria for CAI based on the field of research. The authors agree that the first criteria is a history of at least one ankle sprain with first incidence being more than year prior to testing. They say that the ankle sprain should have affected the lateral ligaments, resulted in pain and swelling, and interrupted the
individual’s physical activity for at least one day. They recommend that the subject should be injury free for at least 3 months prior to testing. For the second criteria, the Consortium stated that the individual should have additionally experienced one or more of the following: recurrent ankle sprain, giving way, or feelings of instability. The authors defined recurrent ankle sprain as the incidence of two or more sprains in the same ankle. Giving way was described as, “the regular occurrence of uncontrolled and unpredictable episodes of excessive inversion of the rear foot (usually experienced during initial contact during walking or running), which do not result in ankle sprain.” The authors say that at least 2 incidences of giving way in the 6 months prior to testing should be perceived to meet these criteria. Feelings of instability was defined as, “the situation whereby during activities of daily living and sporting activities the participant feels that the ankle joint is unstable and is usually associated with the fear of sustaining an acute ligament sprain.” In addition, the Consortium recommend the use of a self-reported measure of instability or disfunction. The present study used these recommendations for the selection of the instability population.

*Cumberland Ankle Instability Tool*

The Cumberland Ankle Instability Tool (CAIT) is one such subjective measure. Developed by Hiller et al., this questionnaire seeks to rate the level of FAI an individual may experience. The CAIT started as a collection of questions commonly asked to people with FAI and was whittled down to nine questions (CAIT III) that gives a score on a scale of 0 (extreme debilitation) to 30 (no instability). The questions asked subjects to qualify during what activities and at which frequencies they had experienced pain or feelings of an unstable ankle. While this
questionnaire allows for measures of both ankles, it does not require comparisons to be made between them allowing it to be used in individuals with bilateral impairment. Hiller et al.,\textsuperscript{34} found that a cutoff score of 27.5 yielded the best reliability, with scores above this value representing normal or no instability. The authors determined that the CAIT questionnaire had a sensitivity of 82.9% and a selectivity of 74.7%.\textsuperscript{34} This implies that the CAIT is a simple, valid, and reliable tool to measure severity of functional ankle instability.

Donahue et al.\textsuperscript{35} performed a review of the CAIT and similar surveys to see how well each was at determining, without the researchers using study inclusion criteria, that a subject likely had FAI as characterized by history or sprain and giving way. The authors determined that the CAIT with a cutoff score of 27 was one of the only measures of the seven questionnaires given to subject that reliably predicted FAI. The authors suggested that lowering the cutoff score might produce an even better result given the subjective nature of the CAIT and the fact that definitions of each condition is not described on the questionnaire.\textsuperscript{35} In a follow up to this study, Simon et al.\textsuperscript{36} claim that the CAIT is still a reliable measure, but may not be as consistent as they once thought. They propose a new subjective measurement tool that they say has better reliability even than when the two highest rated measures from their previous evaluation of seven questionnaires were used in combination.\textsuperscript{36} However, its use has yet to be fully examined.

Wright et al.\textsuperscript{5} reported on the validity of the CAIT in their study that clinically examined differences between people who had developed FAI after ankle sprain and those who had not. The authors compared both the individual’s self-reported history of instability, or lack thereof,
and objective measures including laxity, range of motion, and presence of pain. Wright et al.\textsuperscript{5} used inclusion criteria similar to that recommended by the Consortium\textsuperscript{4} for sprain and giving way. Subjects also had to score a 27 or lower on the CAIT. Participants in the FAI group had a greater incidence of mechanical laxity and pain at the end of the range of motion, but range of motion over all did not appear to be significantly altered. The researchers found that individuals without a history of sprain scored an average of 28.78 (± 1.78) on the CAIT. Subjects with an ankle sprain but no resultant instability scored on average 27.72 (± 1.69). Those with resultant instability scored with the lowest score and highest variation at 20.52 (± 2.94).\textsuperscript{5} The CAIT scores related to the clinical findings, though the results also leaned towards lowering the cutoff score for an accurate representation.

Wright et al.\textsuperscript{37} used the same criteria as their previous work in a recalibration of the CAIT cutoff for participants with CAI. Using self-reported limitations (including pain, functional impairments, and instability) and clinical measures, the researchers found that the Youden index was highest when a cutoff of 25 was assigned. Wright et al.\textsuperscript{37} went on to perform a validation and found that that score afforded excellent sensitivity at 0.966 compared to Hiller et al.’s\textsuperscript{34} cutoff of 27 which Wright et al.\textsuperscript{37} determined to be 0.860.

**Biomechanics of Landing**

Examining certain variables during a landing activity allows for an even better functional measure of potential impairment then the SEBT. Key variables include the ground reaction forces and joint kinematics, kinetics, and energetics.
Ground Reaction Force

The GRF is an example of Newton’s Third Law: as a body is exerting a force on the ground, the ground is exerting an opposite force on that body. It is from the GRF that the body experiences internal forces during walking, running, or landing. If the LE is not able to properly handle these forces, the body may collapse. The neuromusculoskeletal system is responsible for mitigating the internal forces caused by the GRF in a way that decelerates the body and protects the LE from injury. David Winter explains that the GRF itself is not very telling in and of itself when it comes to determining how the neuromusculoskeletal system is able to attenuate the reaction force. Human movement is, in essence, a combination of moments (or torques) at the joint axes of rotation. These moments include internal moments due to muscle contraction and external moments resulting from GRF kinetic chain propagation. Internal moments are the product of muscle force and their respective moment arms. Muscle force is created when muscles contract after recruitment from either central or reflexive motor neuron activation. The moment arm for that force can be defined as the perpendicular distance from the axis of rotation to the line of muscle force application (defined by the origin and insertion of the muscle). Movement results from an imbalance in these internal and external moments. In addition to examining joint moments, joint movement must also be considered.

Joint Kinematics

Kinematics is the study of motion without reference to the forces that cause that motion. Variables involved in the study of kinematics are position, time, velocity, and acceleration. For the purposes of studying the biomechanics of a landing, kinematics includes
the range of motion of the joints; the velocities and accelerations of the bending in those joints; and the position, velocity, and acceleration of the body as a whole as it makes contact with the ground. Movement in the joints contributes to the body’s deceleration upon landing and in regaining upright posture. Joint kinematics alone cannot provide any information regarding internal forces, but, when combined with the GRF, enough information is given to determine the moments acting around the joints.

During a landing, a body’s momentum is reduced to zero mainly by action of the vertical component of the GRF (GFRv). The horizontal components are friction forces, and not large enough to cause a significant effect on deceleration. In order for deceleration to happen in a controlled manner, the LE must attenuate this force. This is mostly achieved by joint flexion. In the closed kinetic chain (after impact), this relates to extensor moments at the hip, knee, and ankle. Over time, the effects of this attenuation can be seen in changes in the vertical component of the ground reaction force (GRFv). Any change in force over time is equal to the impulse which, according to Newton’s Second Law, is related to the change in momentum. Momentum is calculated by multiplying the body’s mass by its velocity. In the case of a landing, this is the person’s mass times gravity over the amount of time they were in the air with gravity accelerating them towards the ground. The impulse required to fully decelerate the body will be equal to that momentum since it must be reduced to zero. Adding mass or landing from a higher vertical position will both increase momentum and, thus, required impulse. Impulse can be altered by alterations in the force or by changes in the amount of time required for the landing.
A study by Seegmiller and McCaw\textsuperscript{9} examined the changes in GRFv and GRF impulse when subjects jumped from three different heights. The authors were mainly interested in any differences that might be apparent in participants who were recreationally active versus those who were athletes in order to study why the athletes, specifically gymnasts, often experienced injuries post loading. During landing impact, two peaks are seen in the GRFv. The first is called the impact peak and occurs within moments of first contact. The second is called the active peak is the maximum reaction to the total landing. The authors saw that there were no differences in the two groups when landing from the lowest height, but that the gymnasts landed with greater peak forces than the recreational athletes. However, because the gymnasts reached the peaks faster, the GRF impulses were not different at any height. Seegmiller and McCaw\textsuperscript{9} deduced that differences in landing strategy probably had an effect on the increased impact forces.

Landing strategy is related to the order in which the foot makes contact with the surface during a landing and to the amount of resultant flexion in the joints in the LE. A study by Self and Paine\textsuperscript{47} examined the mechanics of four different landings: a self-selected, natural landing; a stiff landing with minimal knee flexion and self-selected, natural plantar flexor contraction; a softer landing with minimal knee flexion but with contracted plantar flexors allowing them to land toe-heel; and a flat footed landing. The authors observed that when the participants concentrated on keeping their calves flexed, they were able to land in a toe to heel fashion which resulted in the least amount of impact force. Landing with stiff knees and flat feet resulted in the highest GRFv.\textsuperscript{47}
Joint Kinetics and Energetics

More information can be gathered about the effects of landing if internal forces are taken into account. Kinetics is the study of the forces that are acting on a body or, in the case of the present study, on the joints of the LE themselves. These forces include the reaction forces at each joint and the moments and impulses that accompany those forces. A number of muscles act on the joints as they bend. These muscles attach at different points around the joint on which they act, causing a moment around the joint with the moment arm being the distance from the joint center to the attachment point. All the muscles acting on a joint, both agonist and antagonist, gives a net muscle or joint moment. David Winter explained that, in the LE, the combination of the ankle, knee, and hip moments allow us to maintain posture. He called this the support moment because it equated to the amount of net muscle action required to keep a person upright during stance. During landings, muscle activity required to decelerate the body is also of concern. Angular impulse can be found by taking the integral of the joint moment over the time required to decelerate the body. As was the case with the GRFv impulse, this impulse is equal to the change in momentum. Joint impulse is responsible for decelerating the body.

In order to calculate the internal forces in landing biomechanics, an inverse solution must be used. The researcher can observe the amount of movement at the joint and the amount of force translated at impact, but it is difficult and invasive to directly measure the internal forces. Inverse dynamics uses the impact force and joint movements along with
anthropometric data to calculate the internal reaction forces. Inverse dynamics methods will be discussed later.

By combining kinematics and kinetics, the energetics of the system can be examined and even more can be understood about how much work each joint is doing. Power is the rate at which work is done. For the landing activities, angular power and angular work are often calculated.\textsuperscript{7,10,11,48} Angular power is calculated by multiplying the joint moment by its angular velocity. If the joint moment and angular velocity are acting in the same direction (the direction of movement) the power is positive and muscle groups responsible for such a movement would be shortening. Negative power indicates an eccentric (or lengthening) contraction.\textsuperscript{44} Work occurs when force is applied over a distance. In this case, it is the joint moment and change in joint position. Angular work is calculated by finding the integral of the power. If work is negative, the muscles are absorbing energy. If work is positive, the muscles are releasing energy. In landing biomechanics, work is negative because the muscles are absorbing the impact energy by lengthening during that impact and allowing time for the body to decelerate in a controlled fashion.\textsuperscript{6}

It is this energy transfer that makes landing biomechanics a useful tool in studying potential injury mechanisms. High energy transfer often translates to injuries.\textsuperscript{7,10,11,48} Devita and Skelly\textsuperscript{48} discuss the importance of a soft versus stiff landing in lowering LE injury risk. The soft landing was characterized by greater flexion at impact and a greater range of motion throughout the landing. The stiff landings were performed with a more upright posture and less total flexion. The authors reported that the time characteristics of the ground reaction
force were similar in both landings, but that stiff landings resulted in a higher ground reaction force and, thus, a higher impulse. The larger linear impulse related to higher total angular impulses in the joints. Joint moments were predominantly extensor moments in the hip and knee and plantar flexor in the ankle. High peak moments were experienced with the stiff landings which lead to a plantar flexor angular impulse that was 25% higher than the soft landing. Joint work was negative, which indicated that the muscles were absorbing energy. Interestingly, work at the hip and knee were higher for the soft landing at 54% and 46% higher, respectively, but lower by 14% in the ankle. Devita and Skelly discuss that this means that the ankle becomes more and more responsible for absorbing most of the kinetic energy as stiffness of the landing increases but that hip and knee can contribute more during a soft landing. The muscular system is able to absorb more energy overall in the soft landing than the stiff, resulting in less kinetic energy left over to be absorbed by other tissues which is a potential source of injury.

The contribution of each joint was also studied by Podraza and White and Fong et al. as a means of determining anterior cruciate ligament injury risk. Podraza and White found that an increase in knee flexion caused the peak GRFv to decrease and extensor moments in the knee to increase. This is in agreement with Devita and Skelly’s findings. The authors suggest that the increased knee flexion puts the quadriceps at a mechanical advantage for absorbing the kinetic energy transferred from the landing, but also has the potential to overload them. Fong et al. observed that smaller amounts of ankle-dorsiflexion during landing were coupled with a smaller amount of knee flexion. The authors demonstrated that passive dorsiflexion range of motion was associated with a person’s ability to control their landing position. This
has important implications for ankle instability as dorsiflexion range of motion can be limited with FAI. Terada et al. further examined the link between anterior cruciate ligament injuries and ankle instability. They found that subjects with ankle instability exhibited less knee flexion during a landing activity.

*Landings and Ankle Instability*

Delahunt *et al.* examined changes in LE kinetics and kinematics in order to determine potential causes for ankle inversion injury. Subjects performed landings following a jump off a platform. The authors found that participants exhibited less dorsiflexion and more ankle inversion. They observed a larger ground reaction force and a faster time to peak GRFv. Delahunt *et al.* concluded that both of these pointed to a deficit in the neuromuscular control that would allow the subject to reach a stable ankle position after impact.

Brown *et al.* studied movement variability across subjects with varying modalities of ankle instability to those without. Subjects performed a stop jump maneuver while kinematic data and the GRF were collected. The authors found that subjects with MAI exhibited the least amount of dorsiflexion while those with FAI exhibited the least amount of ankle inversion, contrary to the findings of Delahunt *et al.* However, movement variability was not significantly different. They also observed that people with instability had less variation for hip flexion and abduction, but were fairly comparable to controls in terms of knee flexion. The authors discussed that these changes could be a sign of a deficit or of a protective mechanism. The limited variation in the hip and knee suggested that the participants were unable to properly anticipate the maneuver due to a shortcoming in a “feed-forward neuromuscular
control\textsuperscript{13} system. They also report that such a reduction in variation would be more likely to cause injuries to proximal joints, sparing the ankle.\textsuperscript{13}

Caulfield and Garrett\textsuperscript{14} compared subjects who had FAI to those who did not in a landing study. The authors found that people with ankle instability reached their peaks faster than those without and that the peaks were higher. Caulfield and Garrett\textsuperscript{14} stated that this indicated that the subjects are landing more stiffly. This study also highlighted an important theory regarding control of the landing. In a previous study, Caulfield and Garrett\textsuperscript{46} examined changes in knee and ankle flexion after a landing. They observed that subjects with FAI experienced more flexion overall in both their knees and ankles when compared to normal subjects.\textsuperscript{46} When linking this study to their later work, the authors observed that the knee and ankle flexions were also inappropriately timed and were thus unable to help reduce impact force.\textsuperscript{14} Because of the timing of when the differences were found in the force, the authors were able to tell that subjects’ differences were not due to some deficit in the reflex, but rather with some issue in the “pre-programming of motor control.”\textsuperscript{14} Participants with FAI were not landing in a way that would reduce impact force by controlling and lengthening impact deceleration.

\textbf{Fatigue}

Neuromuscular fatigue has many definitions based on how the researcher intends to measure it. For a sustained contraction, it can be said that fatigue is the inability to maintain the necessary force to hold that contraction. It was defined by Bigland –Ritchie (1983) to be any “reduction in the force-generating capacity of the neuromuscular system that occurs during
sustained activity. This allowed fatigue to be measured as decreased performance, as defined by reduced force production, as an activity was repeated over time. More recently, the definition was altered to include more than just the characteristics of force production. Enoka and Stuart (1992) wanted to be able to account for both performance and the effort put forth to maintain that performance.

Barry and Enoka discuss four different trends in fatigue research. The first, task dependency, states that the mechanism for fatigue varies just as the task causing the fatigue may vary. Differences in complexity, duration, intensity, and frequency are all accounted for under this category. Another category covers the connection between force and endurance. For example, the amount a time a person is able to hold a sustained contraction. Muscle wisdom, the third trend, is often observed during sustained contraction studies. It describes a decrease in the frequency of motor unit discharge, which relates to a decrease in functional strength. The last theme discussed is the perception of effort. If a person feels that the task is longer, more complex, or of higher intensity, they may fatigue more rapidly than if the task is unknown. Martin et al. investigated two potential mechanisms, central and peripheral, of fatigue in a treadmill protocol. Central fatigue is caused by the decrease in firing of motor neurons. Peripheral is characterized by the muscle’s inability to meet metabolic demand. The protocol called for low-intensity, high-endurance exercise and, thus, performance was most affected by central fatigue. This type of fatigue could be considered protective in that it doesn’t allow the body to move in a way that would cause the impacts often associated with injury.

Neuromuscular Fatigue and Ankle Instability
Steib et al.\textsuperscript{54,55} also performed a fatiguing treadmill study, but introduced subjects with ankle instability. The authors discuss the decline in sensorimotor control with the onset of fatigue and the implications with a population with pre-fatigue sensorimotor deficits. The authors used postural control to assess post-fatigue differences between CAI and control groups. Max excursion distances during a SEBT\textsuperscript{54} and COP velocity\textsuperscript{55} were used to measure potential changes in sensorimotor control. The authors found that participants with CAI, who already had lower reach distances in certain directions pre-fatigue, worsened significantly with fatigue in the anterior direction compared to controls.\textsuperscript{54} Their COP velocity also increased more than that of controls.\textsuperscript{55} The findings were significant because, not only are people with CAI starting with a deficit, but the same amount of activity causes a greater decrease in function and control.

Gribble et al.\textsuperscript{56} also used postural control to assess neuromuscular fatigue, but distinct fatiguing protocols were used to target different joints. This allowed the researchers to study neuromuscular fatigue contributions from each joint of the LE so that separate conclusions could be drawn regarding the role of fatigue and the effects from the injury history. The study revealed that participants with CAI relied more on proximal joints for control of stability, especially during fatigued conditions.\textsuperscript{56}

Neuromuscular Fatigue and Landings

Many studies have sought to classify the effects of fatigue on different types of landings as both are commonly associated with the performance of sporting activities.\textsuperscript{6} Brazen et al.\textsuperscript{59} examined changes in joint angles, GRFv, and the time to stabilization, a stability measure, in a
drop landing before and after a fatiguing protocol. The protocol was designed to mimic sporting activities and included agility drills, side to side hops, and vertical jumps. Data were not collected during the fatiguing protocol. The authors found that knee flexion and ankle plantar flexion at contact increased after fatigue. They also observed that GRFv increased with fatigue and discussed the increased risk for injury that this poses.59

This finding was in conflict with other studies that found that GRFv decreased with fatigue.8,11 Coventry et al.8 expected that a decrease in shock attenuation would accompany fatigued landings, however, the authors found that alterations in the joint kinetics and energetics, rather than overall decrease in attenuation, were the outcomes of fatigue. The reduction in GRFv seemed to accompany an increased knee flexion at impact.8 Madigan and Pidcoe11 had similar findings. Their study incorporated the fatiguing protocol into the landing protocol so that the effects of fatigue could be found as fatigue progressed. The authors found that the maximum flexions, not just the flexions at impact, increased with fatigue. In agreement with Devita and Skelly48, Madigan and Pidcoe11 found that increased extensor moments at the hip and knee and accompanied the joint position changes. The fact that knee impulse or negative work did not change significantly indicated that landings were controlled in a way that allowed the hip and ankle to compensate for quad fatigue.

Hypothesis

The hypothesis for the present study is based on this idea of compensation. Neuromuscular fatigue alters the performance of a muscle and causes a reduction in force production. When angular momentum is held fairly constant, changes in kinetics and
energetics at the knee due to fatigue cause changes at the hip and ankle in order to maintain
the angular impulse necessary to decelerate the body. Subjects with ankle instability however,
display a reduced ability to alter landing patterns to suit different activities. They often display
deficits in the control mechanisms necessary to efficiently attenuate forces that are significantly
worsened by fatigue. This study aims to determine how subjects with ankle instability
compensate for fatigue during a landing activity. **The hypothesis for this study is that subjects
with ankle instability will employ landing strategies that deflect landing forces away from the
ankle and toward proximal structures resulting in faster fatigue rates when compared to a
normal population.** This has the potential to increase the risk for injury in all LE structures,
not just those injured during sprains.
Chapter 2 – Methods

The purpose of this study was to determine the differences between healthy subjects and those with ankle instability in the effects caused by fatigue on the biomechanics of landing. The protocol was modeled after a previous study designed by Madigan and Pidcoe, but modified for the inclusion of subjects with ankle instability. Madigan and Pidcoe also were attempting to characterize changes in LE biomechanics during a fatiguing landing activity (FLA). The authors’ hypothesis was that a fatiguing of the quadriceps muscles would cause a change of the internal joint forces and energetics and the hip and ankle compensated for the fatigue in the knee. Madigan and Pidcoe surmised that there would be a change in the net moments at the knee and cause this redistribution in order to provide the support moment necessary, according to the principle by David Winter, to prevent LE collapse. Madigan and Pidcoe designed the FLA to functionally fatigue the knee extensors as a whole because it was more feasible than isolating the single extensor muscles. Madigan and Pidcoe designed the FLA meet four criteria:

1. To keep the landing movement simple by making it a predominantly vertical landing with primarily sagittal plane movement.

2. To utilize a single-leg landing movement since studies have indicated that bilateral landings can involve significant asymmetry.

3. To generate relatively high LE forces during the landing movement that are typically associated with activities that involve impact with ground.

4. To integrate a fatigue inducing activity with the landing movement to allow changes in landing biomechanics to be monitored as fatigue progressed.

The authors also used electromyogram (EMG) fatigue analysis to quantify that fatigue was indeed occurring in the knee extensors during the FLA and that fatigue had indeed been
reached at the conclusion of the trial. As their protocol did achieve the expected fatigue, EMG analysis was not used in the present investigation to reduce the invasive component of Madigan and Pidcoe’s study. A SEBT was added to the FLA protocol the goal of which was to qualify the fatigue experienced by the subjects. Kinetic and kinematic data were collected for both the FLA and the SEBT.

Twenty eight healthy subjects (male: 8, female: 20, mean age: 24.1 ± 3.98 years, mean weight 66.9 ± 14.50 kg, and height: 174.0 ± 12.92 cm) volunteered to participate in this investigation. Thirteen of those subjects (male: 4, female: 9, mean age: 23.9 ± 3.70 years, mean weight 60.9 ± 9.12 kg, and height: 171.5 ± 12.73 cm) were part of a control group. These subjects had had one or fewer sprains in their dominant ankle, no feelings of instability on or giving way of the ankle. Fifteen of the total subjects (male: 5, female: 10, mean age: 24.4 ± 4.23 years, mean weight 72.2 ± 16.20 kg, and height: 176.2 ± 12.79 cm) were classified as having CAI by the criteria established by the International Ankle Consortium. These subjects had a history of at least one sprain in their dominant ankle as well as one or more of the following: recurrent sprains, any feelings of instability in the ankle, two or more experiences of giving way of the ankle in the past six months. All subjects reported having had no acute LE injuries in the past six months. Prior to participation, all subjects gave informed consent in keeping with VCU Institutional Review Board regulations. Prior to any testing, the subjects were given a CAIT questionnaire per International Ankle Consortium recommendation. This questionnaire was used as a compliment to their reported history of instability and was chosen because of its proven reliability in predicting the presence of CAI. Subjects in the CAI group also were required to have a CAIT score of 25 or below based on Wright et al.’s recalibration. Subjects
identified their dominant leg by reporting which foot they would use to kick a ball. The participants’ dominant legs were tested. Participants were asked to participate in two separate days of testing, no more than 72 hours apart, so that reliability analysis could be performed. All subjects wore their own shoes.

**Instrumentation**

Data were collecting using The MotionMonitor™ by Innovated Sports Training. 3-dimensional kinematic data were collected using an electromagnetic (EM) motion analysis system (Ascension Flock-of-Birds™). Six wired sensors communicated X, Y, and Z positional data with a maximal error of 7.8mm per meter of distance between sensors and receiver. The system reports a static linear accuracy of 1.8mm RMS with a resolution of 0.5mm (at a 30.5cm distance from the transmitter) and a rotational accuracy of 0.5° RMS with a resolution of 0.1°. The full technical and physical details can be found in Appendix B. A PCIM-DAS1602/16 (Measurement Computing) was used for data acquisition. A non-conducting force plate by Bertec Corporation (Model 4060_NC) collected GRF data. The force plate was mounted into a wooden testing platform so that it would be at the same level of the testing. Both kinematic and kinetic data were sampled at 1000Hz. A Butterworth filter with a cutoff frequency of 20 Hz was used on these data.

**Procedures**

After taking the CAIT questionnaire and identifying their dominant leg, participants were given an introduction to the procedure. The researcher demonstrated each part of the protocol with the correct form, which will be explained in later paragraphs, so that it could be accurately
replicated. Subjects were then allowed sufficient practice of each part of the procedure. Care was taken in allowing the subject sufficient practice of the SEBT as this in particular has been shown to have a learning effect and plateau associated with its use.32

Warm up and Setup

After familiarizing themselves with the procedures, the participants were instructed to warm up on an LE ergometer by Monark (Model 818E) at a self-selected pace for 10 minutes. The ergometer was set at a low workload of 1.5 kp or 75 Watts. The duration and workload was selected for consistency with the previous study.6 While the subjects were warming up, the researcher set up The MotionMonitorTM by assigning predetermined points marked on the floor to be the world axes locations. The force platform was also calibrated and its location relative to the world axes was defined. The EM motion tracking sensors were laid out so that they were convenient to quickly attach to the participant after they finished their warmup.

EM sensors were attached to the subject using elastic Velcro bands. Pre-wrap was used under the bands to keep them from slipping during the activity. Distal wire tethers were tucked under the proximal bands to keep them wrapped tight to the participant to reduce the risk of tripping. Subjects were equipped with six sensors, attached in a distal to proximal order. The first two were inserted into each of the subject’s shoes between the tongue and the laces. Subjects were allowed to re-lace their shoes to a comfortable tightness after these sensors were inserted. The third was attached to the subject’s dominant leg, about 12 cm superior to the ankle. The non-dominant leg did not receive a sensor. The next two sensors were attached to each thigh, about 20 cm superior to the knee. The sixth sensor was attached over the
participant’s sacrum. All tethers were secured to the ceiling to prevent tripping. A analysis to assess possible sensor slippage after performance of the protocol will be discussed later in this paper.

The subject was calibrated to The MotionMonitor™ by collecting their weight during a quiet stance on the force platform and their height with a movable sensor. The moveable sensor was also used to identify bony landmarks that corresponded to the joint centers and the segment endpoints for the link-segment model. Hip joint centers were identified using the Bell Method which used the location of the left and right anterior superior iliac spine and the sacral sensor to position the pelvis. Knee joint centers were identified by the lateral epicondyles. Ankle joint centers were identified by the lateral malleoli. Distal segment endpoint for the foot was defined as the anterior aspect of the second phalanx. This end point was identified with participant confirmation as participants were shod. The left side of Figure 1 shows the sensor position on the body. The sacral sensor, indicated in red, was placed on the dorsal aspect of the body. The right side of Figure 1 indicates the bony landmarks that were identified on individual subjects by palpation.

Star Excursion Balance Test

When subject setup was completed, the participant was asked to perform the SEBT. The participant was told to align their dominant foot at the back right corner of the force plate, close to the origin. Their bodies faced the positive y axis. The SEBT directions were mapped out in tape and labeled. The subjects were instructed to maintain a one-footed stance and
perform a maximal excursion in each of the 8 directions shown in Figure 2. All subjects began with the anterior reach direction and performed all subsequent reaches in the appropriate order, depending on their test limb, ending with the anterolateral direction. If the testing leg was the right leg, subjects performed the excursions counter clockwise. If the left limb was dominant, the SEBT was performed clockwise. Participants were allowed to tap their toe at the end of their excursion, but were instructed not to transfer weight to the non-support limb. The participants were told that they could compensate with their upper body as needed to maintain balance but to keep hands planted on their hips. The SEBT was performed on the force plate to allow for the collection of COP information. The SEBT was performed both before and once after the FLA. Data from the positions of the feet were collected as well as the COP and the vertical force. During analysis, weight transfer was measured by monitoring off-loaded weight.
from the stance limb located on the force plate. If the subject transferred more than twenty percent of their weight to the non-support foot, the collection was discarded.

**Fatiguing Landing Activity**

The FLA was performed in cycles of two landings and three one-legged squats. The participants stood with one foot aligned on the front and inside edge of a 19.5 cm box. Keeping hands on hips, the participants were instructed to switch to a one-legged stance with their dominant foot on the box and their contralateral limb suspended in the air. They were instructed to skip off the box by swinging the contralateral limb forward and performed a one-legged landing on the dominant foot. Figure 3 shows this method. Subjects were instructed to skip from the box rather than jump so that they would not change the vertical height of the jump by a significant amount. This was done to control and standardize the vertical translational energy carried into the landing. They were also told to aim for the center of the
force plate just off of the front of the box. Horizontal translational effects were kept to a minimum by instructing subjects to jump forward enough to clear the box, and not more. For the portion of the activity designed to accelerate fatigue, shown in Figure 4, the subjects started with the same one-legged suspended stance with which they began the landing activity. Participants were instructed to reach the non-support limb towards the ground, leading with the heel, producing a one-legged squat without making floor contact. A light gate was used to ensure that the subjects reached the same squat depth throughout the FLA. Subjects were instructed to perform the activities at an even pace throughout, as quickly as they could while still achieving controlled landings. The researcher provided verbal encouragement to ensure that the subjects kept pace. Subjects performed the FLA in cycles until fatigue was reached. Participants were told that fatigue was defined as the inability to maintain support without collapse and that they should continue until they felt like their next jump would result in a fall. Again, subjects were provided with verbal encouragement to continue jumping until they reached fatigue.

The MotionMonitor™ collected data via a circular buffer and was triggered by a signal from the force plate upon impact. The software collected kinematic data the half-second before landing and the kinematic and kinetic information from the landings for two seconds after impact. Hip flexion, hip abduction, knee flexion, ankle dorsiflexion, and foot inversion angles were exported for off-line processing. From the force plate, the GRFv data during landing were collected. MotionMonitor™ also provided LE net joint moments based on the link-segment model for impulse, power, and work calculations.
Figure 3: FLA Landing Portion. This picture illustrates the one-legged take off and one-legged landing of the landing portion of the FLA.

Figure 4: FLA Fatiguing Portion. This picture illustrates the one-legged squat procedure for the fatiguing portion of the FLA.
Data Analysis and Statistics

Inverse Dynamics

A link segment model described by Winter\textsuperscript{44} was used to calculate the LE internal forces and moments. Rules for the link segment model are as follows:

1. Each segment has a fixed mass located as a point mass at its COM (which will be the center of gravity in the vertical direction).
2. The location of each segment’s COM remains fixed during the movement.
3. The joints are considered to be hinge (or ball-and-socket) joints.
4. The mass moment of inertia of each segment about its mass center (or about either proximal or distal joints) is constant during the movement.
5. The length of each segment remains constant during the movement (e.g., the distance between hinge or ball-and-socket joints remains constant).\textsuperscript{44}

These rules allow solving for the LE internal joint forces to be broken down into three 2-dimensional problems. Once the LE is broken into segments, calculations are made using each segment separately from the bottom up (or distal-to-proximal in an ‘inverse dynamic’ approach). Figure 5 shows the free body diagram that is created based on the link segment model. Joint moments are denoted by M and internal joint forces are labeled as R for reaction force. Note that the moments and reaction forces on the distal ends of proximal segments are equal and opposite to the forces and moments on the proximal ends of the distal segments. This is what makes it necessary to start with the foot segment. When the LE makes contact with the ground, the ground reaction force acts upwards. For the present research, ground reaction forces and moments were included in addition to those see in Figure 5.

By the convention established for this work, the vertical (Z) axis is positive pointing down into the floor and, for the sagittal plane, the horizontal (Y) axis is positive in the direction
the person is facing. Moments are positive when acting in a counterclockwise fashion. Vertical and horizontal forces and moments in the sagittal plane were summed using the equations of motion:

\[ \sum F_z = m \times a_z \]  
\[ \sum F_y = m \times a_y \]  
\[ \sum M = I_0 \times \alpha \]

where \( m \) is equal to the segment mass, \( a_z \) and \( a_y \) are equal to the linear acceleration of the COM in the vertical and horizontal directions respectively, \( I_0 \) is equal to the segment of the mass moment of inertia, and \( \alpha \) is the angular acceleration of the segment. An example calculation can be seen in Appendix C. Anthropometric data for segment lengths (reported as a proportion of total body height) were from Drillis and Contini (1966) and the segment mass (reported as a proportion of total body mass), location of segment COM, and segment radii of gyration were reported by Dempster.44

**MATLAB**

Custom MATLAB (R2014a, The MathWorks) scripts were created to analyze the data. All programs can be viewed in Appendix D. All data collected for each jump were exported as text files by MotionMonitor™. Exported data included the GRFv; joint angles for hip flexion and abduction, knee flexion, and ankle dorsiflexion and inversion; and net joint moments for those
movements. Data from each landing were compiled into 3D matrices comprising of all collected landing data for that trial. For each landing, the impact phase was selected. The impact phase was defined as the first 200ms after impact and was selected because it contains some of the most relevant information regarding joint kinetics and energetics of landing. After finding the impact phase, the impact and active peaks of the GRFv were identified and the total impulse of the impact phase of the GRFv was calculated. Figure 6 shows an example of impact phase identification and impact and active peak selection. The crosshatched area represents the GRFv impulse. Certain characteristics of the joint movements were identified including the peaks, the positions at impact, and the range of motion between the impact and active peaks. The joint torques were computed for the impact phase and joint impulses and angular

Figure 5: Example Link-Segment Model and Free-Body Diagram. David Winter's link segment model and free body diagram for inverse dynamics calculations. Note that in the present study, ground reaction forces and moments are also included in the free-body diagram.
velocities over the total impact phase were found. Joint power and work were calculated. Data from successive landings (the pair of landings performed together during the FLA) were averaged and used to represent the entire cycle. This data collapse reduced jump variability. Variables of interest including GRFv peaks and impulses, max joint positions, joint impulses, and joint works were normalized for the total number of jumps by dividing all the data into 10% groups based on the number of cycles that were performed. The individual features were plotted over time to show how fatigue progressed over the course of the entire trial. Madigan demonstrated that these data fit a second order polynomial and the present study confirmed
that the data appeared to follow that trend. A second order polynomial was fit to the data and outliers were filtered based on this fit.

Custom programs were also made for the SEBT. A similar program imported all files for one trial’s worth of excursions, COP data, and moments around the force plate. The root mean squared COP velocities for both the anterior-posterior directions and the medial-lateral directions were calculated. The area enclosing the total movement of COP was estimated using convex hull approximation. The processed data were moved into Excel (Microsoft, 2010) and SPSS (IBM SPSS for Statistics 22) for comparison.

Since sensor slippage was a potential source of error, additional analyses were performed through custom programming to measure sensor movement from the beginning to end of the FLA. Sensor position was only noted if the participant was in quiet stance. As this was not in the original instructions, quiet stance was only observed before and after the FLA in 15 trials (out of a total 56). Half of a second of quiet stance was captured just before the start of each SEBT activity (pre and post). A person was determined to be in quiet stance if the position of the sensors were relatively unchanging (maximum change in range=1.2 cm, mean range=.24 ± 0.28 cm). To compare the pre and post values, the person had to be standing the same way as they were preparing to begin the SEBT. Stance position/posture was confirmed using the GRFv and center of pressure values. Prior to SEBT trials, the subject is standing in a natural two footed stance with one foot placed on the back corner of the force plate. The force plate characteristics can tell us how the person is standing. If the individual has centered their weight differently, it could result in changes in their posture, which would be evident in
differences in GRF values. Posture was confirmed if pre and post-activity stance GRF vectors were within 20% or each other. Following confirmation, position data from the all sensors in the x, y, and z dimensions were exported from both pre and post-trial quiet stance epochs. Two methods of comparison were performed, absolute height and relative sensor difference. For the first method, the absolute height of each sensor (foot, shank, thigh, and sacrum) was computed and pre and post height (z) values were compared. The second method of determining the magnitude of sensor slippage compared relative distances between sensor pairs. The magnitude of the vector connecting each sensor pair was computed. These comparisons included the distance between the sacrum and thigh, sacrum and shank, sacrum and foot, thigh and shank, thigh and foot, and shank and foot for both the pre and post-activity stances. Pre and post-activity vector lengths were then compared, resulting in difference errors.

Statistics

For both the FLA and the SEBT, an intraclass correlation coefficient (ICC) was used to test for reliability between the first day of collection and the second, identical collection less than 72 hours after trial one. For the SEBT, a generalized estimating equation (GEE) was used to model population averages. This model incorporated the test population group (unstable vs. control), the fatigue state, direction of reach, test day, gender, and interactions between as factors. An exchangeable correlation matrix structure and a linear response scale were used. The exchangeable correlation structure assumes that responses within subjects are equally correlated. The same model was used for each of the different dependent variables. For the
FLA, linear mixed models were used as repeated measures analysis to test for effects due to fatigue, instability, or the combination of both for each biomechanical variable of interest as the FLA progressed. Linear mixed models were used for this analysis because the repeated measures were identically spaced. A Toeplitz repeated covariance type was used. As needed, the participant’s gender and the number of cycles of the FLA that they were able to perform were also factored in to this model to improve normalcy. A quartile-quartile plot of the residuals was used for this purpose and the strength of the model was given by the Akaike Information Criteria (AIC). The best model, based on the AIC, was chosen by selecting the most complex model without sacrificing normalcy. For the slip model, histograms were used to display the frequency of slip within a range of 0-5 mm, 5 mm to 1 cm, 1 cm to 2 cm, 2 cm to 3 cm, and over 3 cm. Plots and tables were created using Microsoft Excel.
Chapter 3 – Results

The mean CAI score for all subjects was 23.1 ± 4.65. The mean score for the control group was 26.8 ± 2.51. The high score for the control group was 30 and the low score was 21. The mean score for the CAI group was 19.7 ± 3.36 with all subjects scoring at or below a 25 in accordance with Wright et al.’s recommendation. The low score for the CAI group was 12.

Star Excursion Balance Test

Means and standard deviations for all variables of interest in the SEBT are reported in Table 1. ICC values, reported in Appendix A.3, revealed that not all measures displayed day to day reliability. It was therefore decided to include test day as a within subjects factor for GEE analysis.

The P values from that analysis are reported in Table 2. Gender is a significant factor for almost every variable of interest. Only the vertical, twisting moments about the force plate were not significant for gender. Between test populations, significant differences were found for the normalized reach distance (P=0.038) and the X maximum (P=0.002) and mean moments (P=0.005) around the force plate. Controls were able to reach farther than those with instability with an average, normalized reach distance of 0.66 leg lengths compared to 0.58 m for those with CAI. Moments around the X-axis of the force plate relate to anterior-posterior postural control techniques. Controls had a mean moment of -84.52 Nm and a maximum (absolute value) moment of -95.36 Nm. Participants with CAI had a mean moment of -112.15 Nm and a maximum (absolute value) moment of -127.54 Nm. Significant differences were
Table 1: SEBT Results. Means and standard deviations for each variable separated by group, fatigue state, and reach direction (continued on next page).

<table>
<thead>
<tr>
<th>Test Population</th>
<th>Test Day</th>
<th>Fatigue State</th>
<th>Reach Direction</th>
<th>Normalized Reach Distance (ratio of reach to leg length)</th>
<th>Medial-Lateral COPvel (m/s)</th>
<th>Anterior-Posterior COPvel (m/s)</th>
<th>Area of COP Trajectory (m^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>1</td>
<td>pre fatigue</td>
<td>anterior</td>
<td>0.54 (± 0.12)</td>
<td>1.02 (± 0.59)</td>
<td>1.19 (± 0.72)</td>
<td>2.18E-4 (± 1.93E-4)</td>
</tr>
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<td></td>
<td></td>
<td></td>
<td>post-medial</td>
<td>0.72 (± 0.12)</td>
<td>1.05 (± 0.58)</td>
<td>1.9 (± 1.16)</td>
<td>2.18E-4 (± 1.26E-4)</td>
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<td>post-lateral</td>
<td>0.65 (± 0.12)</td>
<td>1.13 (± 0.69)</td>
<td>1.96 (± 1.22)</td>
<td>3.02E-4 (± 2.67E-4)</td>
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<td></td>
<td></td>
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<td>anterior</td>
<td>0.57 (± 0.1)</td>
<td>0.92 (± 0.46)</td>
<td>0.99 (± 0.53)</td>
<td>2.75E-4 (± 2.05E-4)</td>
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<td></td>
<td></td>
<td>post-medial</td>
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<td>0.98 (± 0.52)</td>
<td>1.73 (± 1.08)</td>
<td>2.40E-4 (± 1.34E-4)</td>
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<td></td>
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<td>post-lateral</td>
<td>0.66 (± 0.11)</td>
<td>1.01 (± 0.62)</td>
<td>1.83 (± 1.2)</td>
<td>3.14E-4 (± 2.22E-4)</td>
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<td>anterior</td>
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<td>1.16 (± 0.48)</td>
<td>1.17 (± 0.54)</td>
<td>3.00E-4 (± 1.51E-4)</td>
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<td></td>
<td></td>
<td>post-medial</td>
<td>0.76 (± 0.15)</td>
<td>1.06 (± 0.42)</td>
<td>2.17 (± 1.12)</td>
<td>3.58E-4 (± 1.87E-4)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>post-lateral</td>
<td>0.68 (± 0.17)</td>
<td>1.2 (± 0.61)</td>
<td>2.36 (± 1.28)</td>
<td>3.69E-4 (± 2.49E-4)</td>
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<td>1.1 (± 0.51)</td>
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<td>2.43E-4 (± 2.01E-4)</td>
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<td>2.23 (± 1.28)</td>
<td>3.45E-4 (± 3.31E-4)</td>
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<td>2.27 (± 1.31)</td>
<td>4.77E-4 (± 3.31E-4)</td>
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<td>2</td>
<td>pre fatigue</td>
<td>anterior</td>
<td>0.5 (± 0.09)</td>
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<td>1.28 (± 0.99)</td>
<td>1.83E-4 (± 1.17E-4)</td>
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<td>post-medial</td>
<td>0.64 (± 0.14)</td>
<td>0.87 (± 0.42)</td>
<td>1.98 (± 1.36)</td>
<td>1.68E-4 (± 1.00E-4)</td>
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<td>1.99 (± 1.32)</td>
<td>2.5E-4 (± 3.25E-4)</td>
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<td>anterior</td>
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<td>0.79 (± 0.34)</td>
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<td>2.22E-4 (± 1.46E-4)</td>
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<td>1.94E-4 (± 9.00E-5)</td>
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<td>anterior</td>
<td>0.54 (± 0.11)</td>
<td>1.16 (± 1.11)</td>
<td>1.38 (± 1.18)</td>
<td>3.15E-4 (± 3.34E-4)</td>
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<tr>
<td></td>
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<td>post-medial</td>
<td>0.69 (± 0.12)</td>
<td>0.88 (± 0.52)</td>
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<td>2.65E-4 (± 1.71E-4)</td>
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<td>post-lateral</td>
<td>0.59 (± 0.14)</td>
<td>0.98 (± 0.55)</td>
<td>2.08 (± 1.26)</td>
<td>2.76E-4 (± 1.70E-4)</td>
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<td>anterior</td>
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<td>1.07 (± 0.77)</td>
<td>1.36 (± 1.19)</td>
<td>3.05E-4 (± 3.86E-4)</td>
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<td>0.67 (± 0.1)</td>
<td>0.82 (± 0.43)</td>
<td>2.01 (± 1.51)</td>
<td>2.05E-4 (± 9.30E-5)</td>
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<td>post-lateral</td>
<td>0.58 (± 0.13)</td>
<td>1.01 (± 0.62)</td>
<td>2.15 (± 1.62)</td>
<td>2.42E-4 (± 1.27E-4)</td>
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<tr>
<td>Test Population</td>
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<td>Fatigue State</td>
<td>Reach Direction</td>
<td>X Max Moment (Nm)</td>
<td>X Mean Moment (Nm)</td>
<td>Y Max Moment (Nm)</td>
<td>Y Mean Moment (Nm)</td>
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<td>-----------------</td>
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<td>post-lateral</td>
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<td>91.56 (± 22.56)</td>
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<td>86.64 (± 14.43)</td>
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<td>-115.32 (± 34.66)</td>
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<td>86.62 (± 16.12)</td>
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<td>96.9 (± 19.58)</td>
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<td>anterior</td>
<td>-52.72 (± 12.95)</td>
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<td>98.32 (± 21.14)</td>
<td>90.38 (± 19.34)</td>
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<td>post-medial</td>
<td>-113.76 (± 35.06)</td>
<td>-102.43 (± 32.62)</td>
<td>100.35 (± 24.62)</td>
<td>90.21 (± 21.85)</td>
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<td>post-lateral</td>
<td>-113.92 (± 26.83)</td>
<td>-101.62 (± 27.49)</td>
<td>100.31 (± 28.13)</td>
<td>90.12 (± 24.63)</td>
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<td>-90.62 (± 28.01)</td>
<td>101.29 (± 31.18)</td>
<td>91.38 (± 30.01)</td>
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<td>pre fatigue</td>
<td>anterior</td>
<td>-91.77 (± 31.38)</td>
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<td>108.92 (± 45.08)</td>
<td>99.37 (± 43.88)</td>
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<td>post-medial</td>
<td>-148.73 (± 42.89)</td>
<td>-133.6 (± 32.88)</td>
<td>101.48 (± 58.61)</td>
<td>91.95 (± 56.08)</td>
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<td>post-lateral</td>
<td>-149.07 (± 38.58)</td>
<td>-133.61 (± 38.57)</td>
<td>109.51 (± 52.77)</td>
<td>99.77 (± 51.1)</td>
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<td>2</td>
<td>pre fatigue</td>
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<td>-94.04 (± 24.43)</td>
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<td>111.56 (± 42.1)</td>
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<td>95.58 (± 47.73)</td>
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<td>anterior</td>
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<td>-69.3 (± 31.13)</td>
<td>114.34 (± 39.55)</td>
<td>102.04 (± 40.54)</td>
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<td>-144.88 (± 55.77)</td>
<td>-128.73 (± 55.3)</td>
<td>103.32 (± 30.28)</td>
<td>92.1 (± 25.78)</td>
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<td>108.61 (± 35.47)</td>
<td>98.34 (± 35.6)</td>
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<td>104.04 (± 34.85)</td>
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<td>110.13 (± 36.62)</td>
<td>100.74 (± 37.86)</td>
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</table>
**Table 2: SEBT P Values.** P values for factors and interactions from Generalized Estimating Equations analysis. Significant values (P≤0.05) are highlighted in yellow.

<table>
<thead>
<tr>
<th>Factor or Interaction</th>
<th>Norm. Reach Distance</th>
<th>Medial-Lateral COPvel</th>
<th>Anterior-Posterior COPvel</th>
<th>Area of COP trajectory</th>
<th>X Max Moment</th>
<th>X Mean Moment</th>
<th>Y Max Moment</th>
<th>Y Mean Moment</th>
<th>Z Max Moment</th>
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<td>Gender</td>
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<td>0.000</td>
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<td>0.782</td>
<td>0.164</td>
<td>0.002</td>
<td>0.005</td>
<td>0.236</td>
<td>0.335</td>
<td>0.234</td>
<td>0.409</td>
</tr>
<tr>
<td>Test Day</td>
<td>0.025</td>
<td>0.093</td>
<td>0.261</td>
<td>0.022</td>
<td>0.426</td>
<td>0.310</td>
<td>0.775</td>
<td>0.860</td>
<td>0.466</td>
<td>0.532</td>
</tr>
<tr>
<td>Fatigue</td>
<td>0.839</td>
<td>0.003</td>
<td>0.236</td>
<td>0.668</td>
<td>0.613</td>
<td>0.504</td>
<td>0.339</td>
<td>0.204</td>
<td>0.733</td>
<td>0.156</td>
</tr>
<tr>
<td>Reach Direction</td>
<td>0.000</td>
<td>0.005</td>
<td>0.000</td>
<td>0.028</td>
<td>0.000</td>
<td>0.000</td>
<td>0.216</td>
<td>0.166</td>
<td>0.596</td>
<td>0.315</td>
</tr>
<tr>
<td>Test Population * Test Day</td>
<td>0.906</td>
<td>0.998</td>
<td>0.662</td>
<td>0.660</td>
<td>0.692</td>
<td>0.672</td>
<td>0.809</td>
<td>0.804</td>
<td>0.725</td>
<td>0.691</td>
</tr>
<tr>
<td>Test Population * Fatigue</td>
<td>0.793</td>
<td>0.997</td>
<td>0.314</td>
<td>0.508</td>
<td>0.522</td>
<td>0.450</td>
<td>0.202</td>
<td>0.287</td>
<td>0.905</td>
<td>0.719</td>
</tr>
<tr>
<td>Test Day * Fatigue</td>
<td>0.084</td>
<td>0.306</td>
<td>0.063</td>
<td>0.328</td>
<td>0.539</td>
<td>0.345</td>
<td>0.285</td>
<td>0.251</td>
<td>0.100</td>
<td>0.112</td>
</tr>
</tbody>
</table>
observed between test days for normalized reach distance (P=0.025) and area enclosing the COP trajectory (P=0.022). Participants reached an average of 0.03 leg lengths further on day two versus day one. The area needed to enclose the COP trace increased on the second day as well from an average of $2.34 \times 10^{-4}$ m$^2$ on the first day to $3.04 \times 10^{-4}$ m$^2$ on the second day. Only medial-lateral RMS COP velocity was significantly differently from the pre fatigued state to the post with a P value of 0.003. The velocity of medial-lateral COP excursion before fatigue was 1.01 m/s and after fatigue averaged 0.97 m/s. Figure 7 shows the average excursions in all eight excursion directions as an aerial representation of the differences between test groups.

**Figure 7: Mean Reach Distances.** A representation of mean reach distances for both groups before and after fatiguing protocol in all eight of the performed excursion directions.
and fatigue state. Reach direction was a significant factor for normalized reach distance (P=0.000), medial-lateral RMS COP excursion velocity (P=0.005), anterior-posterior RMS COP excursion velocity (P=0.000), area enclosing COP trajectory (P=0.028), and the maximum and mean moments around the X-axis of the force plate (P=0.000 and P=0.000). Subjects were able to reach the furthest in the posteromedial direction with an average of 0.70 leg lengths. Posterolateral reaches averaged a distance of 0.61 leg lengths while anterior reaches averaged 0.55 leg lengths. Participants exhibited the highest medial-lateral RMS COP velocities in the posterolateral direction (mean=1.03 m/s) followed by the anterior direction (mean=1.00 m/s) and then the posteromedial direction (mean=0.93 m/s). The anterior-posterior COP velocity was slightly different with the highest velocity in the posterolateral direction (mean=2.06 m/s) followed by the posteromedial direction (mean=2.00 m/s) and then the anterior direction (mean=1.24 m/s). The COP trajectory was enclosed by average areas of 2.59e-4 m² for the anterior reach direction, 2.46e-4 m² for the posteromedial reach direction, and 3.05e-4 m² for the posterolateral reach direction. The averages by reach direction for mean and maximum (absolute value) moments around the X-axis of the force plate were -63.04 Nm and -75.22 Nm for anterior reaches, -120.39 Nm and -134.30 Nm for posteromedial reaches, and -116.18Nm and -130.19 Nm for posterolateral reaches.

Fatigue Landing Activity

The average number of cycles of the FLA performed by the control group was 19.7 (±17.9). CAI subjects performed less jumps with an average of 14.3 (±7.0) cycles of the FLA. The ICC measures for GRFv and kinematic data for all subjects can be seen in Appendix A.4. Data from both days of a subject’s testing were analyzed as repeated measures when the ICC
was at or above 0.8, indicating strong test retest reliability. It was reasoned that, if the data did not show at least good retest reliability, there could be lingering fatigue effects from the subject’s first day of performing the FLA. The researchers were concerned that it would not be an accurate representation of the subject’s original, unfatigued state. The joint torques, impulses, power, and work all displayed an ICC below 0.6 for all classes and fatigue states. Upon further investigation, it was discovered that certain collections possessed artifacts at the point of impact in the ankle flexion and foot inversion. This was most likely due to sensor slippage. The artifacts were not large enough to affect peak flexions or peak to peak range of motions. However, when moments were calculated up the kinetic chain, effects were noticeable. It was decided that subjects with significant artifacts would not be considered in the joint kinetic and energetics analysis. Six subjects were eliminated from the instability group, leaving nine, and seven subjects were eliminated from the control group, with six subjects remaining in this group. The mean values for all subjects for kinematic and GRFv variables of interest can be seen in Table 3. One way ANOVA revealed no significant differences between the pre-fatigue states of the CAI participants and the control participants. Significant differences were observed between the groups in post-fatigue GRFv active peak and GRFv impulse. Controls reached an average GRFv peak of 4.02 (±0.7) times their body weight. CAI participants landed with smaller GRFv 3.6 (±0.52) times their body weight. For the control group, significant differences were observed between the pre- and post- fatigue states for maximum knee flexion and peak to peak knee flexion. In the fatigued state, a higher max flexion was reached, but there was a smaller change between peaks. For CAI participants, GRFv, maximum knee flexion, and maximum ankle dorsiflexion changed significantly between
### Table 3: FLA Kinetic and Kinematic Results

Means and standard deviations for data separated by test class and fatigue state.

<table>
<thead>
<tr>
<th></th>
<th>CAI</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>pre-fatigue</td>
<td>post-fatigue</td>
</tr>
<tr>
<td>GRFv Impact Peak (%BW)</td>
<td>1.23 (± 0.33)</td>
<td>1.12 (± 0.38)</td>
</tr>
<tr>
<td>GRFv Active Peak (%BW)</td>
<td>3.72 (± 0.34)</td>
<td>3.6 (± 0.52)</td>
</tr>
<tr>
<td>Time to max GRFv (s)</td>
<td>63.62 (± 13.5)</td>
<td>59.82 (± 10.28)</td>
</tr>
<tr>
<td>GRFv Load Rate (kN/sec)</td>
<td>153.49 (± 54.01)</td>
<td>177.91 (± 64.08)</td>
</tr>
<tr>
<td>GRFv impulse (BW*s)</td>
<td>0.39 (± 0.03)</td>
<td>0.37 (± 0.04)</td>
</tr>
<tr>
<td>Hip Flexion Maximum (deg)</td>
<td>36.38 (± 11.44)</td>
<td>40.99 (± 13.64)</td>
</tr>
<tr>
<td>Knee Flexion Maximum (deg)</td>
<td>50.17 (± 7.07)</td>
<td>58.34 (± 9.92)</td>
</tr>
<tr>
<td>Ankle Dorsiflexion Maximum (deg)</td>
<td>16.01 (± 4.82)</td>
<td>20.09 (± 6.63)</td>
</tr>
<tr>
<td>Hip Abduction Maximum (deg)</td>
<td>9.53 (± 7.11)</td>
<td>8.48 (± 7.1)</td>
</tr>
<tr>
<td>Ankle Inversion Maximum (deg)</td>
<td>5.82 (± 5.76)</td>
<td>5.94 (± 7.38)</td>
</tr>
<tr>
<td>Hip Flexion Peak2Peak change (deg)</td>
<td>-0.7 (± 4.18)</td>
<td>0.66 (± 3.79)</td>
</tr>
<tr>
<td>Knee Flexion Peak2Peak change (deg)</td>
<td>12.55 (± 7.7)</td>
<td>11.65 (± 6.94)</td>
</tr>
<tr>
<td>Ankle Flexion Peak2Peak change (deg)</td>
<td>33.76 (± 15.3)</td>
<td>31.86 (± 15.52)</td>
</tr>
<tr>
<td>Hip Abduction Peak2Peak change (deg)</td>
<td>-6.17 (± 4.37)</td>
<td>-5.32 (± 4.9)</td>
</tr>
<tr>
<td>Ankle Inversion Peak2Peak change (deg)</td>
<td>-8.56 (± 7.45)</td>
<td>-9.51 (± 7.69)</td>
</tr>
</tbody>
</table>

### Table 4: FLA Energetics Results

Means and standard deviations for data separated by test class and fatigue state.

<table>
<thead>
<tr>
<th></th>
<th>CAI</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>pre-fatigue</td>
<td>post-fatigue</td>
</tr>
<tr>
<td>Joint impulse Hip Extension (Nm<em>sec)/(kg</em>m)</td>
<td>0.211 (± 0.159)</td>
<td>0.245 (± 0.131)</td>
</tr>
<tr>
<td>Joint Impulse Knee Extension (Nm<em>sec)/(kg</em>m)</td>
<td>0.056 (± 0.051)</td>
<td>0.058 (± 0.044)</td>
</tr>
<tr>
<td>Joint Impulse Ankle Plantar Flexion (Nm<em>sec)/(kg</em>m)</td>
<td>0.048 (± 0.024)</td>
<td>0.047 (± 0.021)</td>
</tr>
<tr>
<td>Joint impulse Hip Adduction (Nm<em>sec)/(kg</em>m)</td>
<td>0.093 (± 0.108)</td>
<td>0.079 (± 0.095)</td>
</tr>
<tr>
<td>Joint impulse Ankle Eversion (Nm<em>sec)/(kg</em>m)</td>
<td>-0.007 (± 0.085)</td>
<td>-0.008 (± 0.087)</td>
</tr>
<tr>
<td>Joint Negative Work Hip Sagittal (W)/(kg*m)</td>
<td>0.091 (± 0.282)</td>
<td>0.176 (± 0.307)</td>
</tr>
<tr>
<td>Joint Negative Work Knee Sagittal (W)/(kg*m)</td>
<td>0.171 (± 0.18)</td>
<td>0.238 (± 0.188)</td>
</tr>
<tr>
<td>Joint Negative Work Ankle Sagittal (W)/(kg*m)</td>
<td>0.225 (± 0.153)</td>
<td>0.197 (± 0.107)</td>
</tr>
<tr>
<td>Joint Negative Work Hip Frontal (W)/(kg*m)</td>
<td>-0.053 (± 0.319)</td>
<td>-0.135 (± 0.263)</td>
</tr>
<tr>
<td>Joint Negative Work Ankle Frontal (W)/(kg*m)</td>
<td>-0.033 (± 0.23)</td>
<td>-0.008 (± 0.236)</td>
</tr>
</tbody>
</table>
the pre- and post-fatigued states. Impulse was reduced from .39 (± 0.03) %BW*s to .37 (± 0.04). Both knee flexion and ankle dorsiflexion maximums increased with fatigue. Typical joint articulations seen over the impact phase at the beginning and at the end of the FLA are displayed in Figure 8a for the control group and Figure 8b for the CAI group. The change in sagittal and frontal plane moments from pre-fatigued state to fatigued state can be seen in Figure 9. Typical net joint moment values over the impact phase for the unfatigued and the fatigued state are displayed in Figure 9a for the control group and Figure 9b for the CAI group. The means for joint impulse and negative work can be seen in Table 4. The only significant difference between classes and fatigue states revealed by the one way AVOVA tests was in the pre-fatigue sagittal joint work in the hip. The controls had predominantly extensor work while the CAI group absorbed more energy through their hip flexors.

Figure 8a: Typical Joint Articulations for Controls. Sagittal and frontal plane joint articulations for a normal subject from the first cycle (green) and the last (red).
Figure 9b: Typical Joint Articulations for CAI Subjects. Sagittal and frontal plane joint articulations for a normal subject from the first cycle (green) and the last (red).

Figure 9a: Typical Joint Moments for Controls. Sagittal and frontal plane moments for a normal subject from the first cycle (green) and the last (red). Units are Nm*sec normalized for body weight and height.
Estimated marginal means from the FLA were determined from linear mixed models and plotted for all biomechanical variables of interest. Table 5 shows the P values for the linear mixed models analysis. All parts of Figure 10 show the estimated marginal means and standard error of the mean produced by the LMM changing as time progresses for each group: control and CAI. Figure 10a displays the estimated marginal means for the impact peaks of the GRFv. The linear mixed model indicated that the impact peak would decrease about 5% for the control group and about 9% for the CAI group. The model determined that the presence, or lack thereof, of instability (p<.025) was a significant between subjects factor. Other significant effects (p<.05) in impact peak were due to gender and athletic condition, as estimated by the number of cycles of the FLA performed. Figure 10b displays the active peaks in the GRFv. The normal group was modeled at a 6% increase with fatigue while the CAI group was shown to
Table 4: P values for FLA Linear Mixed Models Analysis. Significant values (P≤0.05) are highlighted in yellow. Values that approached significance (P≤0.01) are highlighted in red.

<table>
<thead>
<tr>
<th></th>
<th>impact peak</th>
<th>active peaks</th>
<th>time to max GRFv</th>
<th>max load rate</th>
<th>GRFv Impulse</th>
<th>Max Hip Flexion</th>
<th>Max Knee Flexion</th>
<th>Max Ankle Flexion</th>
<th>Max Hip Abduction</th>
<th>Max Ankle Inversion</th>
</tr>
</thead>
<tbody>
<tr>
<td>instability</td>
<td>0.025</td>
<td>0.001</td>
<td>0.938</td>
<td>0.398</td>
<td>0.055</td>
<td>0.359</td>
<td>0.800</td>
<td>0.670</td>
<td>0.191</td>
<td>0.098</td>
</tr>
<tr>
<td>percent FLA</td>
<td>0.068</td>
<td>0.611</td>
<td>0.000</td>
<td>0.000</td>
<td>0.004</td>
<td>0.006</td>
<td>0.000</td>
<td>0.000</td>
<td>0.212</td>
<td>0.466</td>
</tr>
<tr>
<td>gender</td>
<td>0.021</td>
<td>0.136</td>
<td>0.093</td>
<td>0.000</td>
<td>0.005</td>
<td>0.257</td>
<td>0.526</td>
<td>0.625</td>
<td>0.014</td>
<td>0.017</td>
</tr>
<tr>
<td>cycles</td>
<td>0.000</td>
<td>0.000</td>
<td>0.005</td>
<td>0.000</td>
<td>0.001</td>
<td>0.004</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>0.001</td>
</tr>
<tr>
<td>instability*percent FLA</td>
<td>0.750</td>
<td>0.005</td>
<td>0.092</td>
<td>0.231</td>
<td>0.003</td>
<td>0.911</td>
<td>0.129</td>
<td>0.116</td>
<td>0.025</td>
<td>0.430</td>
</tr>
</tbody>
</table>

Decrease by 4%. The model found that instability (p<.001) and the combined interaction of instability and the percent completion of the FLA (p<.005) were both significant. Athletic condition also appeared to be a factor (p<.001). Figure 10c shows the modeled decrease in time from impact to peak GRFv to be 10% for the controls and 7% for those with instability. Fatigue (p<.001) was a significant effect in the change in estimated means. Athletic condition also caused differences (p<.001). Figure 10d displays the marginal means for the maximum GRFv loading rate as increasing with fatigue 13% for the controls and 16% for the CAI group. The progression of fatigue (p<.001) was a significant factor in the change of the loading rates. Gender and athletic condition also had effects (p<.001). The model means for impulse of the impact phase GRFv are displayed in Figure 10e. The progression of fatigue (p<.004) and the interaction of fatigue and instability (p<.003) were significant. The effect due to presence of
instability approached significance ($p=.055$). Gender ($p<.005$) and athletic condition ($p<.001$) were also significant. The model shows that the CAI group experienced a 5% drop in GRFv impulse while the normal group would experience a parabolic effect: increasing about 3% halfway through the FLA and then dropping back to approximately the original value. Figure 10f shows an 8% increase in maximum hip flexion of the normal group and an 11% increase for those with instability. Progression of fatigue ($p<.006$) was significant as was athletic condition ($p<.004$). Maximum knee flexion is seen in Figure 10g. Normal subjects were estimated to experience a 14% increase in flexion over the course of fatigue while the CAI group had a 17% increase. Again, fatigue ($p<.001$) and the number of cycles performed caused significant differences ($p<.001$). Figure 10h shows maximum ankle dorsiflexion. The model estimated a 23% increase for the controls and a 27% increase for those with CAI. Progress of fatigue and number of cycles were both significant ($p<0.001$). Figure 10i displays modeled hip abduction. The model estimated that abduction would increase by 22% for controls and decrease by 15% for CAI group. The interaction of fatigue and instability was significant ($p<0.025$). Gender also had an effect ($p<0.04$). Figure 10j illustrates a 27% and 3% mean increases in ankle inversion for the control and CAI groups respectively. There were no significant effects from instability or fatigue, however, gender ($p<0.017$) and athletic condition ($p<0.001$) had effects. Figure 10k shows that there was an increase of 8% and 18% in hip extension impulse in the control and CAI groups respectively. Estimated marginal means for knee extension impulse can be seen in Figure 10l. There was a slight increase in impulse of 10% for the controls and 7% for the CAI group. Ankle plantar flexion impulse is displayed in Figure 10m. Controls demonstrated a 12% decrease in impulse while the CAI group averaged an estimated 2% decrease. Figure 10n shows
an approximately 8% decrease in hip adduction impulse for the CAI group. The normal group shows a 2% decrease, however, the means increase by about 8% for the first 40% of the FLA. Frontal plane ankle impulses are seen in Figure 10o. CAI group demonstrated slight, eversion impulses while the normal subjects experienced inversion impulse throughout the FLA. The magnitude of the impulse decreased with fatigue for both groups: normal by 17% and CAI by 40%. While none of the models for joint impulse revealed any significant factors, the effect that instability group had on frontal plane ankle impulse approached significance at p < .063. Figure 10p shows the sagittal plane hip negative work. This value increased by 40% for the normal group and by 93% for the CAI group. Knee negative work increased as well, as seen in Figure 10q. The normal group experienced a 36% increase while the amount of energy absorption at the knee of those with instability increased by 44%. The progression of fatigue (p < .003) was a significant factor in this model. As illustrated in Figure 10r, ankle work in the sagittal plane changed very little. The controls showed a 1% decrease and the CAI group showed an 11% decrease. Frontal plane hip negative work was abductive, as seen in Figure 10s. A 27.3% increase was seen for those without impairment while those with instability experienced an 11% increase. Frontal plane ankle negative work, seen in Figure 10t, was predominantly in the evertors and varied considerably for both groups. Overall, with fatigue, about 135% more energy was absorbed within the collection frame for the control group while negative work increased in the CAI group by 400%.
**Figure 10a: GRFv Impact Peak.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

**Figure 10b: GRFv Active Peak.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
**Figure 10c: Time to GRFv Maximum.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

**Figure 10d: Maximum GRFv Loading Rate.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10e: GRFv Impulse. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10f: Maximum Hip Flexion. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10g: Maximum Knee Flexion. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10h: Maximum Ankle Dorsiflexion. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10i: Maximum Hip Abduction. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10j: Maximum Ankle Inversion. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10k: Joint Impulse, Hip, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10l: Joint Impulse, Knee, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10m: Joint Impulse, Ankle, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10n: Joint Impulse, Hip, Frontal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
Figure 10o: Joint Impulse, Ankle, Frontal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

Figure 10p: Joint Negative Work, Hip, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
**Figure 10q**: Joint Negative Work, Knee, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

**Figure 10r**: Joint Negative Work, Ankle, Sagittal Plane. This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
**Figure 10s : Joint Negative Work, Hip, Frontal Plane.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.

**Figure 10t : Joint Negative Work, Ankle, Frontal Plane.** This figure displays the estimated marginal means produced by the linear mixed model to represent the population means for each group as fatigue progresses. Error bars give the standard error of each estimated mean.
**Slip Model**

Sensor slip was assessed on 15 trails that had quiet stance at both the beginning and end of data collection. Differences in distance from each sensor to the ground from pre to post fatiguing activity are displayed in Figure 11 and Figure 12. Figure 11 shows both the magnitude of the error and the direction (positive for superior/proximal slip and negative for inferior/distal slip) for each of the examined trials. Figure 12 shows the absolute value of the errors in histogram format. Figure 13 illustrates the distribution of these errors in a histogram format. The average error from the foot to the sacrum was 16mm, to the thigh was 3.0mm, and to the shank 3.7mm. Note that the largest error is associated with movement of the sacral sensor. In fact, this sensor was seen to slip in both a superior and inferior direction. The thigh and shank sensors only slipped inferiorly. The slip model shows that the sensor on the sacrum shifts the most with 46.7% of subjects showing a total of 1-2 cm of sensor migration. The thigh sensor shows 60% of subjects with 0 to 5 mm of movement. The shank sensor shows 53.3% of subjects with 0 to 5 mm of movement.
Figure 11: Sensor to Ground Slip Error. This bar graph shows the slip direction and magnitude of slip error.
Figure 12: Histogram of Absolute Error. Note that Thigh, shank, and foot sensor errors are predominantly less than 5mm.
Figure 13: Histogram of Relative Error between Select Sensor Pairs. Note that the majority of Thigh and shank errors are less than 5mm.
Chapter 4 – Discussion

This study aimed to track differences in the progression of fatigue between individuals with ankle instability and those without. Changes in the biomechanics of landing, including GRFs, LE kinematics, kinetics, and energetics, during the development of fatigue were studied in the hopes of discovering differences between the study populations that might lead to the recurrent sprains and instability observed in people with CAI.

Slip Model

The first set of histograms show the absolute value of the error from pre to post-activity sensor to ground differences (Figure 12). It can be seen that the sacral sensor slipped the most. The sensor was attached to the body via a Velcro belt. On recommendation from Madigan (2001), the wires for inferior sensors were threaded through the belt to reduce tripping hazard during the experiment. It is likely that this caused movement in the belt and migration of the sacral sensor during the repeated jumps. The sacral sensor was also worn on the outside of clothing, shifts of which could also cause slippage of the sensor. The sensor on the dorsal surface of the foot was laced into the shoe of the participant and, being thus secured, had the least amount of slippage. It was therefore chosen to compare all other sensors to the foot as a relative measure of slippage.

The pre to post-activity three-dimensional distance differences between sensors seen in Figure 13 gives the most complete picture of relative sensor movement. These data demonstrate that most of the shank and thigh sensors movement is less than 5mm, with an
average movement of 3.7mm and 3.0mm respectively. The sacral sensor had the largest average movement of 16.0mm.

Sensor slippage is a source of error because of the way the motion tracking system computes joint location and movement information. During subject setup, data are collected while the subject is standing in anatomic neutral. The sensor positions are noted relative to segment endpoints (bony landmarks) as identified by a movable sensor. The system then creates a rigid body model using the calculated segment lengths and sensor locations. Joint angles are calculated by computing the intersection angle of adjacent segments. When slippage occurs, it affects this angular assessment. This creates potential errors in both joint angle and derivatives of that angle.

It was determined that, in this study, the changes in the sacral sensor would produce negligible errors in joint position data since that slippage was primarily translational. Segment length and orientation would be preserved. As a result, derivatives of these data would also be minimally impacted.

*Star Excursion Balance Test*

The data presented in Figure 7 show that, on average, those with CAI are unable to reach the same excursion distances as those without instability in the ankle. It is unclear from this data, however, the reason for this deficit. Neither passive nor closed-chain range of motion tests were performed for the present study, but it has been observed that movement patterns can have significant effects on reach distance. In the diagonal excursion directions, this is harder to quantify given the multiplane nature of the excursion. Terada *et al.* report that
weight-bearing dorsiflexion range of motion is almost correlated (p=0.051) with excursion distance in the posteromedial direction. In the present study, the deficit became more exaggerated with subsequent reaches, especially in those extending posterior and lateral. Recall that for a lateral reach, the participant is crossing over their supporting limb with their reaching limb. Participants were allowed to compensate for this shift by leaning with their torso, yet it could be observed during testing that this might have cause participants to invert slightly, which would place stress on the lateral structures of the ankle. Gabriner et al.\textsuperscript{62} found that greater eversion strength, as measured by a dynamometer, positively correlated to a larger excursion distance. It is also possible that subjects may have made the decision not to perform a maximum excursion. Given the mode of ankle sprain injuries, it is not surprising that subjects with CAI may fall short of those without instability given their predisposition for giving way or potential fear of reinjury.

Pre- and post-fatigue excursion distances for both groups are also seen in Figure 7. Contrary to some previous literature,\textsuperscript{32} these values were generally unaffected by fatigue. The majority of changes that were observed, though slight, seemed to indicate that subjects in both groups may improve with fatigue. Steib et al.\textsuperscript{54,55} found that fatigue most often caused a decrease in maximum excursion and many other studies have similar findings.\textsuperscript{32,56} The authors did not test excursions in diagonal directions and used a different protocol for the fatiguing activity. They also recorded three trials in each direction. The present study allowed sufficient practice before any recording took place, but only allowed for one trial before and after fatigue. Gribble et al.\textsuperscript{32} do report a learning effect that can be present with the SEBT. It is possible that the time elapse between SEBT practice and recorded trials caused a similar effect. In future,
multiple trials should be recorded as was done by Steib et al.\textsuperscript{54,55} It is also possible that subjects did not reach fatigue or were able to recover. Subjects were allowed up to two minutes of rest between the FLA and the post-fatigue SEBT. We must assume that these are unlikely possibilities, given their successful use in previous research protocols.\textsuperscript{6,32,54}

This study took measurements relating to the COP that are most commonly reserved for static stance: COP velocity and COP trajectory area. In no cases was the COP area found to be a reliable test re-test measure during the SEBT activity. Because it was not a repeatable measure based on the ICC, it was not tested for between or within subject effects. The COP velocity was more repeatable; however, no significant differences were seen between or within either group. It can therefore be determined that these measures were inconclusive as applied to the present study. The measurements were not sensitive to the differences between the two groups. This is most likely due to the dynamic nature of the test. The COP will move in response to the shift in weight as the person extends their leg. This measurement could not distinguish between the instable or normal population nor could it differentiate between fatigue states.

The most repeatable and most significant measures were those of the moments created by the application of the GRFv to the COP around the x-axis of the force plate. These moments were produced by anterior-posterior control strategies. Both the maximum and mean moments were greater in those with CAI than in those without instability. To produce greater moments, either the applied force or the moment arm must increase. Since the participants were maintaining a single leg base of support throughout the activity, it follows that those with CAI
made longer COP excursions, increasing the moment arm, during the same activity. In static activity, this would relate to an increase in COP area and indicate a balance deficit. Flexions in the sagittal plane would be most responsible for changes in the anterior-posterior direction. One possible reason for exacerbated COP excursion with CAI is that a reduced range of dorsiflexion might result in a proximal shift of control to larger muscle groups. Another possible cause could be the reduced force sense at the ankle joint in participants with CAI. Some fine tuning capability may be lost in either of these cases resulting in further COP excursions and increased moments.

The intended purpose of the SEBT in the present study was to use the well-established information on the effects of fatigue on SEBT performance as validation that fatigue had occurred. Gribble et al. discuss that fatigue of the proximal muscles may result in more of an ankle-strategy for postural control. The authors go on to say that neuromuscular fatigue of the ankle may result in few compensatory contractions in normal subjects. This would limit COP excursions. As stated above, in the absence of changes in magnitude of force, smaller COP excursions could be observed as smaller GRFv moments. In the present study, a reduction in moments was generally observed for the normal subjects before and after fatigue. By the reasoning Gribble et al., this could be a strong indicator that proximal muscle fatigue has occurred in the normal subjects.

Fatigued Landing Activity

Madigan used EMG analysis to demonstrate that fatigue of the quadriceps did indeed occur in the participants of that protocol. The most important assumption that is made moving
forward is that, by replicating that protocol, participants experienced a similar fatigue from the activity. The reliability of data obtained for this study has been previously discussed. Generally, however, the ICC values for test re-test consistency were comparable to those obtained by Madigan. Some variability may have been introduced by using a 24-72 hour window for subject completion of the second day of trials, but we hope that any learning or residual fatigue effects that may have been introduced were decreased by careful sampling, as discussed previously. It is with these assumptions that we hope to demonstrate that deviations from the results for the aforementioned study, and studies similar to it, are due to the altered study population and characteristics therein.

As seen in Figure 14, the GRFv impact peak decreased in the study from 1.23 times BW at baseline to 1.12 BW when fatigued for the CAI group and decreased from 1.33 to 1.28 BW for the controls. Landing took place from a 19.5 cm box. Decker et al. had subjects to land from 40 cm and reported GRFv impact peaks between 1.51 BW (males) and 1.58 (females).
However, participants in that study landed with two feet: one foot on the force plate and the other on the ground level to the force plate. Seegmiller and McCaw\textsuperscript{9} also conducted trials in this way with one foot on and one foot off the force plate. The authors reported an impact peak between 0.89 and 0.96 BW for a landing from 30 cm for recreational athletes and college athletes, respectively. Increasing the height to 60 cm, the researchers found that the impact peaks increased to 1.53 and 2.22 BW.\textsuperscript{9} By this relationship, we might expect that their recreationally active athletes would have landed with an impact peak of about .67 BW from a height of 20 cm. This would put the present study’s impact peaks within an appropriate range for single footed landing.

We accept that the impact peak decreases with fatigue because GRFv active peak has decreased with fatigue in previous literature.\textsuperscript{8,11} Coventry \textit{et al.}\textsuperscript{8} reported active peak values to fall from 3.9 BW to 3.63 BW before and after fatigue when landing from a height of about 35 cm. Pidcoe and Madigan\textsuperscript{11} reported a 12% decrease from 3.69 BW unfatigued to 3.24 fatigued from a height of 25 cm. CAI participants for the present study averaged a 3% decrease from 3.72 to 3.60 BW. These values are higher than Pidcoe and Madigan’s\textsuperscript{11}, even with the shorter box height. However, a higher maximum GRFv was reported in ankle instability participants versus control in a landing study by Caulfield and Garrett.\textsuperscript{14} So, we might assume a smaller decrease with fatigue could be significant.

The present study did not find this expected trend for the control group. As seen in Figure 10b, this group experienced an increase in GRFv with fatigue. One possible reason for this could have been due to deviation from the instructions. Subjects were instructed to skip
from the box so as not to increase vertical height before landing. An increase in vertical height would explain an increase in GRFv. However, as seen in Figure 15, while some amount of “jumping up” rather than “jumping out” did occur, it did not seem to be correlated to the progression of fatigue. Different protocols, including dropping from a bar, could have been used to ensure a more constant drop height. However, this type of drop would produce additional upper extremity movement which could negatively impact the consistency of the LE performance. Additional correction or guidance regarding jump form might have improved the consistency of the drop height for the present study. Subjects had been instructed to swing their non-support leg forward to gain the momentum necessary to horizontally translate from the box. Some subjects may have performed this swing in addition to bending their knees and rapidly extending to generate the force to takeoff. Additional constraints for physical ability may also have reduced this variability. All subjects claimed to recreationally active, however, using a more specific definition of recreationally active might have resulted in a more homogeneous sample population.

Devita and Skelly\textsuperscript{48} report several factors that might cause the GRFv to change in landing scenarios where starting height has not changed. The primary cause is the stiffness or softness of the landing. The present study observed a 27% decrease in knee flexion range of motion from landing to GRFv peak, meaning that subjects were stiffer with fatigue and would exhibit a higher peak GRFv. Another reason discussed was the movement variability before impact resulting in different flexions at the moment of impact. If the subject is already flexed when landing, they may be limited in how much more flexion they can produce. Brazen \textit{et al.}\textsuperscript{59}
actually report an increase in GRFv for a one legged fatiguing study. However, the fatiguing protocol and experimental set up were different.

A slight decrease was seen in time to peak GRFv for both the controls and the CAI group, even with the apparent difference in soft versus stiff landing strategy. This change was reflected in the maximum loading rate. The GRFv loading rate for CAI participants increased with faster landing time. In the control group, this faster landing time is accompanied by an increase in GRFv maximum, accounting for the difference in load rate seen in Figure 10d.

Figure 15: GRFv versus Sacral Height. Two representative samples from the control group illustrating the change in GRFv maximum with respect to sacral sensor height as fatigue progresses in ten percent intervals.

A slight decrease was seen in time to peak GRFv for both the controls and the CAI group, even with the apparent difference in soft versus stiff landing strategy. This change was reflected in the maximum loading rate. The GRFv loading rate for CAI participants increased with faster landing time. In the control group, this faster landing time is accompanied by an increase in GRFv maximum, accounting for the difference in load rate seen in Figure 10d.
Temporal pre-fatigue values are comparable to previous studies for both groups.\textsuperscript{8,11,14,48} Caulfield and Garrett\textsuperscript{14} report that the timing of peak GRFv will be slightly faster for individuals with FAI than that of controls. They report that this change was accompanied by a difference in the medial-lateral and anterior-posterior GRF, which was not reported in the present study. Delahunt \textit{et al}.\textsuperscript{12} also found a faster loading for CAI participants. They explain that the significance of this was that the ankle may not yet have been in its closest packed position when it had to attenuate the highest load, increasing the risk for injury.\textsuperscript{12}

With a decrease in GRFv, a decreased GRFv impulse over the 200 ms impact phase was expected. As seen in Figure 10e, this decrease in impulse with fatigue is observed in the present study’s CAI group with values trending from 0.39 BW*s to about 0.37 BW*s. This pattern was not observed in the control group whose pre- and post-fatigue values appear the same. The impulse was not constant throughout the FLA. Values increased until about 40-50\% through the FLA and then begin to decrease. It is possible that there was too much time between the warm-up activity and the FLA. The control participants may not have begun to fatigue until part way through the activity. The increase seen at the beginning of the time trend could be a warming up effect.

Increases in sagittal plane flexions are responsible for many of the changes seen in the GRFv. With an increase in flexion, the participant is able to slow their landing. This does not change the amount of momentum that they have built up from their drop, but it does cause the participant to extend their deceleration beyond the impact phase and causes the drop in impulse that is observed.\textsuperscript{6,8,48} The control group pre-and post-fatigue maximum flexions were
very consistent with those from similar protocols in previous literature.\textsuperscript{6,8} An interesting difference can be seen the comparing the progression of control group knee flexion in the present study to the same from Pidcoe and Madigan’s\textsuperscript{11} findings. Pidcoe and Madigan\textsuperscript{11} report that there is a steady increase in maximum knee flexion. Figure 10g demonstrates that the control group exhibits a more or less constant knee flexion until 40%-50% of the FLA has been completed followed by a more pronounced increase. Examining Figure 15 more closely, we can see that there is a trend in increasing sacral sensor height and increasing GRFv maximum for the first half of the representative data. This supports the differences seen in the GRFv impulse. Since GRFv appears to be increasing and flexion appears to remain the same, there would be no mechanism for extending the deceleration beyond the impact phase. Pre-fatigue flexions values for the CAI group are in agreement with previous studies both in maximum flexion and range of motion.\textsuperscript{12,13,51} Given the similarity in the GRFv characteristics, the trend of increasing maximum flexion with fatigue are not unexpected and no significant differences were noted. The plots of the estimated marginal means, Figures 10f and 10h, do reveal some visually interesting differences. The first is an almost parallel mean shift in Figure 10f. Maximum dorsiflexion seen in Figure 10h begin and ends in similar places, however, ankle dorsiflexion maximum values increase faster with fatigue and seem to reach a plateau. Both of these differences allude to potential differences in how each joint will handle the dissipation of the impact forces.

Joint articulation in the frontal plane was less pronounced, as would be expected for this type of activity. Pre-fatigue values for the controls in the present study are in agreement with the values found by Kernozak \textit{et al.}\textsuperscript{64} Peak to peak hip abduction values vary little, as can be
seen in Table 3. We can therefore say that our finding that the progression of fatigue had little effect on the controls was in agreement with the results obtained by Madigan. Maximum hip abduction values for the CAI participants differed significantly with fatigue from that of the controls by decreasing as the FLA progressed. This would suggest that the CAI subjects exhibited more adduction as well. Maximum ankle inversions were higher than those reported by Powell et al. and by Delahunt et al. However, this could be because the protocols varied slightly from the present study. Delahunt et al. observed that those with ankle instability exhibited less ankle eversion with landing and the present study had similar findings.

In the sagittal plane, joint impulses, as seen in Table 4, were found to be predominantly in the hip, followed by the knee, and then the ankle. This was true for both groups and throughout the duration of the FLA. Energy absorption was found to take place predominantly in the ankle, then knee, then hip for pre-fatigue CAI participants. However, this switched to a predominately knee LE energy absorption, followed by ankle, and then hip after completion of the FLA. For the controls, energy absorption was found to take place mostly in the knee, followed by ankle and hip respectively. Overall, as displayed in Table 6, this lead to increased hip and knee extensor impulses and energy absorption and decreases in the ankle. This seems to indicate that the quadriceps were not the primary target of the functional fatiguing protocol as was established by Pidcoe and Madigan. For the control group, this might be attributed to the late onset of fatigue as was demonstrated in the GRFv impulse trend shift after 40-50%. In the ankle, this could point to a significant difference in the landing strategy of individuals with
Table 6: Sagittal Plane Changes. Summary of sagittal plane joint kinetic and energetic changes over the progression of fatigue.

<table>
<thead>
<tr>
<th>Sagittal Plane</th>
<th>CAI Group</th>
<th>Control Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip:</td>
<td>Extensor Impulse: ↑18%</td>
<td>Extensor Impulse: ↑8%</td>
</tr>
<tr>
<td></td>
<td>Energy absorption: ↑93%</td>
<td>Energy absorption: ↑40%</td>
</tr>
<tr>
<td>Knee:</td>
<td>Extensor Impulse: ↑7%</td>
<td>Extensor Impulse: ↑10%</td>
</tr>
<tr>
<td></td>
<td>Energy absorption: ↑44%</td>
<td>Energy absorption: ↑36%</td>
</tr>
<tr>
<td>Ankle:</td>
<td>Plantar Flexion Impulse: ↓2%</td>
<td>Plantar Flexion Impulse: ↓12%</td>
</tr>
<tr>
<td></td>
<td>Energy absorption: ↓11%</td>
<td>Energy absorption: ↓1%</td>
</tr>
</tbody>
</table>

Table 7: Frontal Plane Changes. Summary of frontal plane joint kinetic and energetic changes over the progression of fatigue.

<table>
<thead>
<tr>
<th>Frontal Plane</th>
<th>CAI Group</th>
<th>Control Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip:</td>
<td>\textit{Adduction} Impulse: ↓8%</td>
<td>\textit{Adduction} Impulse: ↓2%</td>
</tr>
<tr>
<td></td>
<td>\textit{Abductor} Impulse:</td>
<td>\textit{Abductor} Impulse:</td>
</tr>
<tr>
<td></td>
<td>Energy absorption: ↑11%</td>
<td>Energy absorption: ↑27%</td>
</tr>
<tr>
<td>Ankle:</td>
<td>\textit{Eversion} Impulse:  ↓40%</td>
<td>\textit{Inversion} Impulse: ↓17%</td>
</tr>
<tr>
<td></td>
<td>\textit{Evertor} Impulse:</td>
<td>\textit{Evertor} Impulse:</td>
</tr>
<tr>
<td></td>
<td>Energy absorption: ↑400%</td>
<td>Energy absorption: ↑135%</td>
</tr>
</tbody>
</table>

CAI. Pre-fatigue net negative work values were most similar to those found in Coventry et al.\textsuperscript{8} The magnitude of the values were smaller because the drop height was decreased, but the relative contributions were the same. Coventry et al.\textsuperscript{8} acknowledge this difference by stating that their participants experienced greater trunk flexion than had previously been seen by Pidcoe and Madigan\textsuperscript{11} and by DeVita and Skelly\textsuperscript{48}. Trunk flexion was not measured in the present study, however, we can assume that this may be a factor here due to Coventry's et al.\textsuperscript{8} report that the rest of the LE kinematics were similar to the values in the present study and those found by Pidcoe and Madigan\textsuperscript{11} and by DeVita and Skelly\textsuperscript{48}. Additionally, when
questioned after completion of the FLA, many CAI participants reported that they felt more fatigue in their calves than in their quadriceps. This may also suggest an altered strategy.

This finding can also be demonstrated in the frontal plane. Dickin et al.\textsuperscript{66} observed an increase in both ankle frontal power and moment with fatigue. This supports the present study’s finding of an increase in energy absorption. The difference between the controls and the CAI subjects could be a result of the altered landing strategy. As the participants have landed, they have experienced dorsiflexion and eversion. In the closed kinetic chain, the triplanar nature of the ankle joint complex would cause an internal rotation of the shank with dorsiflexion and eversion.\textsuperscript{17} Figure 16 demonstrates that subjects with CAI may experience less internal rotation. Figure 16a shows that there is a relatively fixed amount of internal rotation after the peak GRFv while Figure 16b shows they subject entering recovery with external rotation. This may indicate that the CAI participant is producing a stronger contraction in the gastrocnemius in order to introduce more stability or stiffness to the joint by using this biarticulate muscle.

\textbf{Figure 16: Internal Tibial Rotation.} Two representative examples of the internal tibial rotation of A) CAI participant and b) control participant displayed alongside GRFv for impact point reference. Each line indicates a separate landing collection.
The fact that CAI participants are unable to fine tune this contraction, as a normal subject would, reflects a classic symptom of CAI according to Hertel’s paradigm: impaired neuro muscular control. A co-contraction of the gastrocnemius and an anterior leg muscle, like the tibialis anterior, would also work to introduce added joint stiffness. Future studies would be needed to examination the activation of muscles in this area.

In summary, significant differences were observed between groups in peak ground reaction force, ground reaction force impulse, and frontal plane ankle joint impulse. Results indicated that subjects with ankle instability started at a disadvantaged state for most measurements and also managed the progression of fatigue differently. It is suggested that future studies use a more homogeneous population in terms of physical ability or activity level. Prospective researchers will want to carefully manage the set up protocol to ensure subjects are adequately warmed up before beginning the FLA. Future work in this area should include EMG analysis of the muscles of the LE, especially of the gastrocnemius. Information gathered in this study may lead to a better understanding of the failure modes that make the CAI population more susceptible to recurrent sprain and other impairments. It is hoped that further research of these failure modes could allow for the development of appropriate prophylactics, training devices, or more accurate risk evaluation tools that would reduce the occurrence of injuries of this kind.
References


Appendix A: Supplemental Figures and Tables

1) Overview of LE joint articulations

2) Detail of ankle articulations


3) ICC values for SEBT variables of interest

<table>
<thead>
<tr>
<th>Test Population</th>
<th>Control</th>
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<tbody>
<tr>
<td>Fatigue State</td>
<td>Prefatigue</td>
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<td>Reach Direction</td>
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<td>A</td>
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<tr>
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<table>
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<td>Reach Direction</td>
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<td>RMS vel COPx</td>
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<td>RMS vel COPy</td>
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<td>Mean AP moment</td>
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## 4) ICC values for FLA variables of interest

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<th>Test Population</th>
<th>Control</th>
<th>Instability</th>
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<td></td>
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<td>Postfatigue</td>
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<td>Impact Peak ICC</td>
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<td>GRFv Loading Rate ICC</td>
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Appendix B: Supplemental Materials

1) Cumberland Ankle Instability Tool (III) given to participants

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<th>RIGHT</th>
</tr>
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<tbody>
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<td>Never</td>
<td></td>
<td></td>
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<tr>
<td>During sport</td>
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<tr>
<td>Running on uneven surfaces</td>
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<tr>
<td>Running on level surfaces</td>
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<tr>
<td>Walking on uneven surfaces</td>
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<tr>
<td>Walking on level surfaces</td>
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</table>

<table>
<thead>
<tr>
<th>2. My ankle feels UNSTABLE</th>
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</tr>
</thead>
<tbody>
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<td>Never</td>
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<td></td>
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<tr>
<td>Sometimes during sport (not every time)</td>
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<tr>
<td>Frequently during sport (every time)</td>
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<td></td>
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<tr>
<td>Sometimes during daily activity</td>
<td></td>
<td></td>
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<tr>
<td>Frequently during daily activity</td>
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<table>
<thead>
<tr>
<th>3. When I make SHARP turns, my ankle feels UNSTABLE</th>
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<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sometimes when running</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Often when running</td>
<td></td>
<td></td>
</tr>
<tr>
<td>When walking</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>4. When going down the stairs, my ankle feels UNSTABLE</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>If I go fast</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Occasionally</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Always</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>5. My ankle feels UNSTABLE when standing on ONE leg</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>On the ball of my foot</td>
<td></td>
<td></td>
</tr>
<tr>
<td>With my foot flat</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>6. My ankle feels UNSTABLE when</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I hop from side to side</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I hop on the spot</td>
<td></td>
<td></td>
</tr>
<tr>
<td>When I jump</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>7. My ankle feels UNSTABLE when</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I run on uneven surfaces</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I jog on uneven surfaces</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I walk on uneven surfaces</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I walk on a flat surface</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>8. TYPICALLY, when I start to roll over (or twist) on my ankle, I can stop it</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Immediately</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Often</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sometimes</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Never</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I have never rolled over on my ankle</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>9. After a TYPICAL incident of my ankle rolling over, my ankle returns to normal</th>
<th>LEFT</th>
<th>RIGHT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Almost immediately</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Less than one day</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1-2 days</td>
<td></td>
<td></td>
</tr>
<tr>
<td>More than 2 days</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I have never rolled over on my ankle</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
2) Most recent informed consent form

RESEARCH PARTICIPANT INFORMATION AND CONSENT FORM

TITLE: The Effects of Functional Fatiguuing of Dominant Leg in Subjects with Ankle Instability

VCU IRB PROTOCOL NUMBER: HM20001280

INVESTIGATOR: Peter Pidcoe, PhD, DPT, PT

If any information contained in this consent form is not clear, please ask the study doctor or the study staff to explain any information that you do not fully understand. You may take home an unsigned copy of this consent form to think about or discuss with family or friends before making your decision.

In this consent form, “you” always refers to the research participant. If you are a legally authorized representative, please remember that “you” refers to the study participant.

PURPOSE OF THE STUDY

The purpose of this study is to determine if there is a change in which leg muscle groups have the greatest role in recovering from special exercise testing in subjects with ankle instability.

DESCRIPTION OF THE STUDY

To begin this study you will be asked to fill out a Cumberland Ankle Instability Tool (CAIT) questionnaire. This questionnaire is a self-reported qualification of your ankle instability. The research will use your CAIT result to assign you to a testing group and to determine which ankle will be tested. If one ankle is determined to be unstable according to the CAIT scale, that ankle will be tested. If both ankles are unstable, your dominant leg will be tested.

The activity will begin with a warm up on an exercise bike for ten minutes at a comfortable pace and at minimal resistance. After your warm up, the researcher will attach 6 small, wired sensors to certain landmarks on your lower body with elastic belts and will use a 7th sensor to identify the locations of the joints in your legs to the computer. You will then perform a Star Excursion Balance Test (SEBT). Standing on one leg (the one to be tested), you will slide your other leg along a piece of tape until you feel unstable. You will repeat this action with the leg extended in front of your body, to the side, and to the back.

Version 1.9, March 6, 2015

Page 1 of 6

Approved by the VCU IRB on 3/10/2015
During the main data collection portion of the study, you will be asked to perform a series of one-legged drop jumps and one-legged squats repeatedly until you feel unsteady. Force plates underneath you will collect center of pressure information from your foot as you land. The electromagnetic 3D motion tracking system will collect information on the position of your body as you jump, land, and recover.

After you report being tired, you will be asked to complete the SEBT again within 3 minutes of completing the drop jump protocol. The sensors will then be removed.

Your participation in this study will last up to 120 minutes. Approximately 15-20 individuals will participate in this study.

PROCEDURES

If you decide to be in this research study, you will be asked to sign this consent form after you have had all your questions answered. At your first study visit (Visit 1), background information will be obtained to determine the extent of ankle instability, if present. You will be asked to complete the Cumberland Ankle Instability Test (CAIT) to determine your level of disability. Please answer as honestly as possible as this test will determine if your level of instability should exclude you from this study for safety reasons. Your height and weight will also be collected by the researcher. During your first visit you will also be allowed to practice the protocol with and without the motion tracking equipment on your body. Once you are ready to begin, you will be asked to perform the study as described above. You will perform the SEBT while the researcher marks the length of your excursion in each direction. The researcher will record the movement of your center of pressure on the force plate as you recover your balance during each extension of your leg. During the main collection, the researcher will collect information about the ground reaction forces after a land and the position of your body during landing, paying specific attention to the angles of your hip, knee, and ankle joints.

Your second visit (Visit 2), which should be scheduled within 48 hours of Visit 1, you will perform the warm-up, first SEBT, landing protocol, and post fatigue SEBT on the same leg for comparison purposes. Visit 2 should be scheduled within 48 hours of Visit 1. Both visits should be scheduled for approximately the same time of day and you should follow your normal routine throughout the testing period.

RISKS AND DISCOMFORTS

Participants will experience slight tiredness in their leg muscles during the study, but the risk is minimal. When the participant feels fatigued, he or she will stop performing the drop jump protocol. The participant may also feel some general fatigue or soreness. This risk for this is also minimal. The participant will be given ample time to rest and recover after completing the protocol. As with all studies where personnel data is collected, there is minimal risk of breach of confidentiality. All analyzed data will be de-identified and
identifiable information will be stored in a locked file cabinet. Only researchers will have access to that data.

USE AND DISCLOSURE OF PROTECTED HEALTH INFORMATION

Your privacy is important to us. During this study, we will ask you to share identifiable health information with us. This health information is Protected Health Information, so it will be protected like your other medical records are protected. We are asking you to authorize the release of your research information in the specific situations described below:

Types of Personal Health Information That May Be Collected by This Study
The following types of information may be used to conduct this research study:
- Complete health record
- Diagnosis & treatment
- Discharge summary codes
- History and physical exam
- Consultation reports
- Laboratory test results
- X-ray reports
- Photographs, videotapes
- Complete billing record
- X-ray films / images
- Itemized bill
- Information about drug or alcohol abuse
- Information about Hepatitis B or C tests
- Information about psychiatric care
- Information about sexually transmitted diseases
- Other (specify): names, email addresses and phone numbers

Authority to Request or Release Protected Health Information
The following people and/or groups may request my Protected Health Information and the Principal Investigator may release my information to them:

- Health Care Providers at the VCUHS
- Research Collaborators and Study Staff
- Data Safety Monitoring Boards
- Data Coordinators
- Study Sponsor
- Institutional Review Boards
- Government/Health Agencies
- Others as Required by Law

Once your health information has been disclosed to anyone outside of this study, the information may no longer be protected under this authorization.

Right to Revoke Authorization and Re-disclosure
You may change your mind and revoke (take back) the right to use your Protected Health Information at any time. Even if you revoke this Authorization, the researchers may still use or disclose health information they have already collected about you for this study. If you revoke this Authorization you may no longer be allowed to participate in the research study. To revoke this Authorization, you must write to the Principal Investigator.

BENEFITS TO YOU AND OTHERS

Version 1.0, March 6, 2015
The information gathered during the study may lead to a better understanding of the mechanisms behind certain injuries experienced in people with ankle instability and the probability that an injury may occur.

COSTS

There are no charges for the study visits. You may receive a parking voucher to pay for parking during your visits.

ALTERNATIVE TREATMENT

Your alternative is not to participate in this study.

CONFIDENTIALITY

Potentially identifiable information about you will include your reported, related medical history. Data is being collected only for research purposes. Your data will be identified by ID numbers, not names, and stored separately from other records in a locked research area. All personal identifying information will be kept in password protected files and these files will be deleted five (5) years after the completion of the study. Other physical records will be kept in a locked file cabinet for five (5) years after the study ends and will be destroyed at that time. Access to all data will be limited to study personnel. A data and safety monitoring plan is established.

You should know that research data or (medical information if applicable) about you may be reviewed or copied by the sponsor of the research or by Virginia Commonwealth University. Personal information about you might be shared with or copied by authorized officials of the Federal Food and Drug Administration, or the Department of Health and Human Services (if applicable).

Although results of this research may be presented at meetings or in publications, identifiable personal information pertaining to participants will not be disclosed.

VOLUNTARY PARTICIPATION AND WITHDRAWAL

Your participation in this study is voluntary. You may decide to not participate in this study. Your decision not to take part will involve no penalty or loss of benefits to which you are otherwise entitled. If you do participate, you may freely withdraw from the study at any time. Your decision to withdraw will involve no penalty or loss of benefits to which you are otherwise entitled.

Your participation in this study may be stopped at any time by the study doctor or the sponsor without your consent. The reasons might include:

• the study doctor thinks it necessary for your health or safety;

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• you have not followed study instructions;
• the sponsor has stopped the study; or
• administrative reasons require your withdrawal.

QUESTIONS
If you have any questions, complaints, or concerns about your participation in this research, contact:

Peter Pidcoe, 804-628-3655, pepidcoe@vcu.edu
or
Lindsay Clayton, 410-218-3287, claytonle@vcu.edu

The researcher/study staff named above is the best person(s) to call for questions about your participation in this study.

If you have general questions about your rights as a participant in this or any other research, you may contact:

Office of Research
Virginia Commonwealth University
800 East Leigh Street, Suite 3000
P.O. Box 980568
Richmond, VA 23298
Telephone: (804) 827-2157

Contact this number for general questions, concerns, or complaints about research. You may also call this number if you cannot reach the research team or if you wish to talk to someone else. General information about participation in research studies can also be found at http://www.research.vcu.edu/irb_volunteers.htm.

Do not sign this consent form unless you have had a chance to ask questions and have received satisfactory answers to all of your questions.

CONSENT
I have been provided with an opportunity to read this consent form carefully. All of the questions that I wish to raise concerning this study have been answered.

By signing this consent form, I have not waived any of the legal rights or benefits, to which I otherwise would be entitled. My signature indicates that I freely consent to participate in this research study. I will receive a copy of the consent form once I have agreed to participate.
3) Technical details for Ascension Flock of Birds™ system

### Technical

**Tracking Range:**
- Mid-Range Transmitter: ±30" (75cm) for specified accuracy, ±36" (90cm) for slightly reduced accuracy.
- Extended-Range Transmitter: ±8'-10" (2.4-3.05m) depending on environmental conditions.

**Angular Range:**
- ±180° Azimuth & Roll, ±90° Elevation

**Static Accuracy:**
- Position: 0.07" (1.8mm) RMS
- Orientation: 0.5° RMS

**Static Resolution:**
- Position: 0.02' (0.5mm) @ 12" (30.5cm)
- Orientation: 0.1° @ 12" (30.5cm)

**Update Rate:**
- Up to 144 measurements/second

**Outputs:**
- X, Y, Z positional coordinates and orientation angles, or rotation matrix

**Interface:**
- RS-232 with selectable baud rates to 115,200

**Format:**
- Binary

**Modes:**
- Point or Stream

### Physical

**Transmitters:**
- Mid-Range Transmitter 3.75" (9.6cm) cube with 10' (3.05m) cable; or
- Extended-Range Transmitter: 12" (30.5cm) cube with 20' (6.1m) cable

**Sensor:**
- 1.0" x 1.0" x 0.8" (25.4mm x 25.4mm x 20.3mm) cube or in optional 3D Pointer ("Wanda") with 10' (3.05m) or 35' (10.7m) cable

**Enclosure:**
- 9.5" x 11.5" x 2.6" (24cm x 29cm x 6.6cm)

**Power:**
- User provided or optional external plug-in, US/European version

**Operating Temperature:**
- 10°C to 40°C (50°F to 104°F)

**Operating Humidity:**
- 10% to 90% non-condensing

* Accuracy verified over range from 20.3cm to 76.2cm at constant orientation with Mid-Range Transmitter.

www.ascension-tech.com
Appendix C: Inverse Dynamics Calculations

Inverse Dynamics

1) Thigh
2) Shank
3) Foot

Ankle

\[ \Sigma F_y = m a_y \]
\[ F_y + R_y = m a_y \Rightarrow R_y = m a_y - F_y \]
\[ \Sigma F_z = m a_z \]
\[ -F_z + R_z + m g = m a_z \Rightarrow R_z = m a_z - m g + F_z \]
\[ \Sigma M = I_0 \alpha \]
\[ M_A - m g (d \cdot \cos \theta) + F_y (l \cdot \sin \theta) + F_z (l \cdot \cos \theta) + M_{GRF} = I_0 \alpha \]
\[ M_A = I_0 \alpha + m g d \cdot \cos \theta - F_y l \cdot \sin \theta - F_z l \cdot \cos \theta - M_{GRF} \]

Shank

\[ \Sigma F_y = m a_y \]
\[ -R_{y_{Ankle}} + R_{y_{Knee}} = m a_y \Rightarrow R_{y_{Knee}} = m a_y + R_{y_{Ankle}} \]
\[ \Sigma F_z = m a_z \]
\[ -R_{z_{Ankle}} + R_{z_{Knee}} + m g = m a_z \Rightarrow R_{z_{Knee}} = m a_z - m g + R_{z_{Ankle}} \]
\[ \Sigma M = I_0 \alpha \]
\[ I_0 \alpha = M_K - M_A + m g (d \cdot \sin \theta) - R_{z_{Ankle}} (l \cdot \sin \theta) - R_{y_{Ankle}} (l \cdot \cos \theta) \]
\[ M_K = I_0 \alpha + M_A - m g (d \cdot \sin \theta) - R_{z_{Ankle}} (l \cdot \sin \theta) - R_{y_{Ankle}} (l \cdot \cos \theta) \]
Inverse Dynamics

\[ \Sigma F_y = m a_y \]
\[ -R_{\text{y, hip}} + R_{\text{y, knee}} = m a_y \]
\[ R_{\text{y, hip}} = m a_y + R_{\text{y, knee}} \]

\[ \Sigma F_z = m a_z \]
\[ R_{\text{z, hip}} + mg - R_{\text{z, knee}} = m a_z \]
\[ R_{\text{z, hip}} = m a_z - mg + R_{\text{z, knee}} \]

\[ \Sigma M = I_0 \alpha \]
\[ M_H - M_h - mg(l \sin \theta) + R_{\text{z, knee}} (l \sin \theta) - R_{\text{y, knee}} (l \cos \theta) = I_0 \alpha \]
\[ M_H = I_0 \alpha + M_h + mgd(\sin \theta) - R_{\text{z, knee}} (l \sin \theta) + R_{\text{z, knee}} (l \cos \theta) \]

*Ground reaction forces and moments, segment flexion/extension angles, and segment linear accelerations are taken from the motion tracking system.

l = segment length = (subject's height) (ratio of segment length to total body length)**
m = segment mass = (subject's mass) (ratio of segment mass to total body mass)**
d = distance to COM = (segment length) (ratio of COM distance to segment length)**

**these ratios can be found in related literature

I_o = (segment mass) (radius of gyration)**
Appendix D: MATLAB™ Code

%lindsay_sebtmatrix.m
%compiles Star Excursion Balance Test files into matrices separated by
%trial day and fatigue state.
%includes values from the 0.25 seconds before and after the max excursion

root_name=input('enter the root of the file names you wish to use:','s');
%code=input('subject number:','s');

disp ' ;disp '...building the matrices...' ;disp ' ;

%%% pre jump sebt
A_pre = zeros(501,13,8);   %pre-allocating for speed

pre_f_name=strcat(root_name,'_sebtpre');

for jj=1:1:8

    f_name=strcat(pre_f_name,sprintf('%.4d',jj));
%f_name=strcat(pre_f_name,sprintf('%.d',jj));
    f_name=strcat(f_name,'.exp');
    Y=load(f_name);

    %find the max distance between planted foot and excursion foot and then
    %build a matrix out of the 250ms before and 250ms after.
    x_dist=Y(:,2)-Y(:,5);
    y_dist=Y(:,3)-Y(:,6);
    distance=sqrt(x_dist.^2+y_dist.^2);
    [max_dist,point]=max(distance,[],1);
    plot(distance)
    hold on
    linemax1=[point,point];
    linemax2=[min(distance), max(distance)];
    plot(linemax1,linemax2,'r')
    hold off
    pause(1)
    close
    if point<251
        disp 'point<251'
        verify=input('Do you wish to graphically reset the max? Y or N :','s');
        if strcmp(verify,'y')
            plot(distance);hold on;
            [x,~] = ginput(2);
            pre=round(x(1));
            post=round(x(2));
            max_dist=max(distance(pre:post),[],1);
            point = find(distance==max_dist,1);
            plot(distance)
            hold on
            linemax1=[max_dist,max_dist];
            linemax2=[min(distance), max(distance)];
            plot(linemax2,linemax1,'r')
        end
    end

end

elseif point>(length(distance) - 250)
    disp 'point to close to end'
    verify = input('Do you wish to graphically reset the max? Y or N :','s');
    if strcmp(verify,'y')
        plot(distance); hold on;
        [x,~] = ginput(2);
        pre = round(x(1));
        post = round(x(2));
        max_dist = max(distance(pre:post),[],1);
        point = find(distance == max_dist,1);
        plot(distance)
        hold on
        linemax1 = [max_dist, max_dist];
        linemax2 = [min(distance), max(distance)];
        plot(linemax2,linemax1,'r')
        hold off
        pause(1)
        close
    end
else
    disp 'point within limits'
end

Z = Y(point-250:point+250,:);
A_pre(:,:,jj) = Z;

end
%pre_f_name = strcat(code,'_sebtpre');
S.(pre_f_name) = A_pre;
    save('sebt_matrices.mat', '-struct','S','-append')
    disp 'pre test data saved in sebt_matrices.mat';

%% post jump sebt
A_post = zeros(501,13,8); %pre-allocating for speed
post_f_name = strcat(root_name,'_sebtpost');

for jj=1:1:8
    f_name = strcat(post_f_name,sprintf('%.4d',jj));
    %f_name = strcat(post_f_name,sprintf('%d',jj));
    f_name = strcat(f_name, '.exp');
    Y = load(f_name);
%find the max distance between planted foot and excursion foot and then
%build a matrix out of the 250ms before and 250ms after.
 x_dist=Y(:,2)-Y(:,5);
y_dist=Y(:,3)-Y(:,6);
distance=sqrt(x_dist.^2+y_dist.^2);
[max_dist,point]=max(distance,[],1);
plot(distance)
hold on
linemax1=[point,point];
linemax2=[min(distance), max(distance)];
plot(linemax1,linemax2,'r')
hold off
pause(1)
close
if point<251
 disp 'point<251'
 verify=input('Do you wish to graphically reset the max? Y or N :','s');
 if strcmp(verify,'y')
 plot(distance);hold on;
 [x,-] = ginput(2);
 pre=round(x(1));
 post=round(x(2));
 max_dist=max(distance(pre:post),[],1);
 point = find(distance==max_dist,1);
 plot(distance)
 hold on
 linemax1=[max_dist,max_dist];
 linemax2=[min(distance), max(distance)];
 plot(linemax2,linemax1,'r')
 hold off
 pause(1)
 close
 end
elseif point>(length(distance)-250)
 disp 'point to close to end'
 verify=input('Do you wish to graphically reset the max? Y or N :','s');
 if strcmp(verify,'y')
 plot(distance);hold on;
 [x,-] = ginput(2);
 pre=round(x(1));
 post=round(x(2));
 max_dist=max(distance(pre:post),[],1);
 point = find(distance==max_dist,1);
 plot(distance)
 hold on
 linemax1=[max_dist,max_dist];
 linemax2=[min(distance), max(distance)];
 plot(linemax2,linemax1,'r')
 hold off
 pause(1)
 close
 end
else
 disp 'point within limits'

end

Z=Y(point-250:point+250,:);
A_post(:,:,jj) =Z;

end

%post_f_name=strcat(code,'_sebtpost');
S.(post_f_name) = A_post;
    save('sebt_matrices.mat','-struct','S','-append')
    disp 'post test data saved in sebt_matrices.mat';

disp ' ';disp 'finished!';disp ' ';

% end code
% this program finds the excursion distance in each direction and tracks the
% center of pressure before and after the max distance is reached. Moments
% around the x,y,and z axes of the force plate are also calculated. by
% convention, X moments correspond to anterior-posterior changes,
% Y moments correspond to medial-lateral changes, and Z moments relate to
% LE longitudinal twisting.

% find max excursion
% evaluate change in center of pressure

clear all
clc
sampling_rate = 1000; % set to 1000Hz
T = 1 / sampling_rate; % period
load sebt_matrices

subject=input('Please enter subject ID_date: ','s');
prebit=strcat(subject,'_sebtpre');
postbit=strcat(subject,'_sebtpost');
pre_name=eval(prebit);
post_name=eval(postbit);

% assign initial variables
pre_xpos_left=pre_name(:,2,:);
pre_ypos_left=pre_name(:,3,:);
pre_zpos_left=pre_name(:,4,:);
pre_xpos_right=pre_name(:,5,:);
pre_ypos_right=pre_name(:,6,:);
pre_zpos_right=pre_name(:,7,:);
pre_vert_force=pre_name(:,8,:);
pre_COPy=pre_name(:,9,:);
pre_COPx=pre_name(:,10,:);
pre_momentx=pre_name(:,11,:);
pre_momenty=pre_name(:,12,:);
pre_momentz=pre_name(:,13,:);

post_xpos_left=post_name(:,2,:);
post_ypos_left=post_name(:,3,:);
post_zpos_left=post_name(:,4,:);
post_xpos_right=post_name(:,5,:);
post_ypos_right=post_name(:,6,:);
post_zpos_right=post_name(:,7,:);
post_vert_force=post_name(:,8,:);
post_COPy=post_name(:,9,:);
post_COPx=post_name(:,10,:);
post_momentx=post_name(:,11,:);
post_momenty=post_name(:,12,:);
post_momentz=post_name(:,13,:);

% make each var 2D
pre_xpos_left=squeeze(pre_xpos_left);
pre_ypos_left=squeeze(pre_ypos_left);
pre_zpos_left=squeeze(pre_zpos_left);
pre_xpos_right=squeeze(pre_xpos_right);
pre_ypos_right=squeeze(pre_ypos_right);
pre_zpos_right=squeeze(pre_zpos_right);
pre_vert_force=squeeze(pre_vert_force);
pre_COPy=squeeze(pre_COPy);
pre_COPx=squeeze(pre_COPx);
pre_momentx=squeeze(pre_momentx);
pre_momenty=squeeze(pre_momenty);
pre_momentz=squeeze(pre_momentz);

post_xpos_left=squeeze(post_xpos_left);
post_ypos_left=squeeze(post_ypos_left);
post_zpos_left=squeeze(post_zpos_left);
post_xpos_right=squeeze(post_xpos_right);
post_ypos_right=squeeze(post_ypos_right);
post_zpos_right=squeeze(post_zpos_right);
post_vert_force=squeeze(post_vert_force);
post_COPy=squeeze(post_COPy);
post_COPx=squeeze(post_COPx);
post_momentx=squeeze(post_momentx);
post_momenty=squeeze(post_momenty);
post_momentz=squeeze(post_momentz);

%% max excursion
%[len wid]=size(pre_xpos_left);

for dir=1:8
    pre_x_dist(dir)=max(pre_xpos_left(:,dir))-max(pre_xpos_right(:,dir));
    pre_y_dist(dir)=max(pre_ypos_left(:,dir))-max(pre_ypos_right(:,dir));
    max_excursion_pre(dir)=sqrt(pre_x_dist(dir)^2+pre_y_dist(dir)^2);
end

for dir=1:8
    post_x_dist(dir)=max(post_xpos_left(:,dir))-max(post_xpos_right(:,dir));
    post_y_dist(dir)=max(post_ypos_left(:,dir))-max(post_ypos_right(:,dir));
    max_excursion_post(dir)=sqrt(post_x_dist(dir)^2+post_y_dist(dir)^2);
end

%% Max excursion display

U_pre=[0 sqrt(max_excursion_pre(2)^2/2) max_excursion_pre(3)
sqrt(max_excursion_pre(4)^2/2) ... 
   0 -sqrt(max_excursion_pre(6)^2/2) -max_excursion_pre(7) -
   sqrt(max_excursion_pre(8)^2/2)];
V_pre=[max_excursion_pre(1) sqrt(max_excursion_pre(2)^2/2) 0 -
   sqrt(max_excursion_pre(4)^2/2) ... 
   -max_excursion_pre(5) -sqrt(max_excursion_pre(6)^2/2) 0
   sqrt(max_excursion_pre(8)^2/2)];
U_post=[0 sqrt(max_excursion_post(2)^2/2) max_excursion_post(3)
sqrt(max_excursion_post(4)^2/2) ... 
   0 -sqrt(max_excursion_post(6)^2/2) -max_excursion_post(7) -
   sqrt(max_excursion_post(8)^2/2)];
V_post=[max_excursion_post(1) sqrt(max_excursion_post(2)^2/2) 0 -
   sqrt(max_excursion_post(4)^2/2) ... 
   -max_excursion_post(5) -sqrt(max_excursion_post(6)^2/2) 0
   sqrt(max_excursion_post(8)^2/2)];

%compass plot later at figure 2
%% root mean square velocity of center of pressure excursions
for ii=1:8
    for jj=1:500
        AA(jj,ii)=((pre_COPx(jj+1,ii)-pre_COPx(jj,ii))/T)^2;
        BB(jj,ii)=((pre_COPy(jj+1,ii)-pre_COPy(jj,ii))/T)^2;
        CC(jj,ii)=((post_COPx(jj+1,ii)-post_COPx(jj,ii))/T)^2;
        DD(jj,ii)=((post_COPy(jj+1,ii)-post_COPy(jj,ii))/T)^2;
    end
end
AA=sum(AA);
BB=sum(BB);
CC=sum(CC);
DD=sum(DD);
for ii=1:8
    VEL_preCOPx(ii)=sqrt(AA(ii)/500);
    VEL_preCOPy(ii)=sqrt(BB(ii)/500);
    VEL_postCOPx(ii)=sqrt(CC(ii)/500);
    VEL_postCOPy(ii)=sqrt(DD(ii)/500);
end

%% COP sway area
%using convex hull approximation, a polygonal area is fit over the COP trajectories to give an estimated sway area.
%center the COP over the average
[length depth]=size(pre_COPx);
for i=1:depth
    M1X(i)=mean(pre_COPx(:,i));
    M1Y(i)=mean(pre_COPy(:,i));
    M2X(i)=mean(post_COPx(:,i));
    M2Y(i)=mean(post_COPy(:,i));
end
for i=1:depth
    for j=1:length;
        pre_COPx(j,i)=pre_COPx(j,i)-M1X(i);
        pre_COPy(j,i)=pre_COPy(j,i)-M1Y(i);
        post_COPx(j,i)=post_COPx(j,i)-M2X(i);
        post_COPy(j,i)=post_COPy(j,i)-M2Y(i);
    end
end

% find the sway area
for pos=1:8
    k_pre = convhull(pre_COPx(:,pos),pre_COPy(:,pos));
    pre_sway_A(pos)=polyarea(pre_COPx(k_pre,pos),pre_COPy(k_pre,pos));
end

for pos=1:8
    k_post= convhull(post_COPx(:,pos),post_COPy(:,pos));
    post_sway_A(pos)=polyarea(post_COPx(k_post,pos),post_COPy(k_post,pos));
end
%% tables and plots
% table of the max excursions and max velocities pre and post in each
direction
settitle=strcat(subject,'_sebt');

aa=[max_excursion_pre(1) max_excursion_post(1) VEL_preCOPx(1) ...
    VEL_postCOPx(1) VEL_preCOPy(1) VEL_postCOPy(1) pre_sway_A(1)
post_sway_A(1)];
bb=[max_excursion_pre(2) max_excursion_post(2) VEL_preCOPx(2) ...
    VEL_postCOPx(2) VEL_preCOPy(2) VEL_postCOPy(2) pre_sway_A(2)
post_sway_A(2)];
cc=[max_excursion_pre(3) max_excursion_post(3) VEL_preCOPx(3) ...
    VEL_postCOPx(3) VEL_preCOPy(3) VEL_postCOPy(3) pre_sway_A(3)
post_sway_A(3)];
dd=[max_excursion_pre(4) max_excursion_post(4) VEL_preCOPx(4) ...
    VEL_postCOPx(4) VEL_preCOPy(4) VEL_postCOPy(4) pre_sway_A(4)
post_sway_A(4)];
e=[max_excursion_pre(5) max_excursion_post(5) VEL_preCOPx(5) ...
    VEL_postCOPx(5) VEL_preCOPy(5) VEL_postCOPy(5) pre_sway_A(5)
post_sway_A(5)];
f=[max_excursion_pre(6) max_excursion_post(6) VEL_preCOPx(6) ...
    VEL_postCOPx(6) VEL_preCOPy(6) VEL_postCOPy(6) pre_sway_A(6)
post_sway_A(6)];
gg=[max_excursion_pre(7) max_excursion_post(7) VEL_preCOPx(7) ...
    VEL_postCOPx(7) VEL_preCOPy(7) VEL_postCOPy(7) pre_sway_A(7)
post_sway_A(7)];
hh=[max_excursion_pre(8) max_excursion_post(8) VEL_preCOPx(8) ...
    VEL_postCOPx(8) VEL_preCOPy(8) VEL_postCOPy(8) pre_sway_A(8)
post_sway_A(8)];

dat=[aa;bb;cc;dd;ee;ff;gg;hh];
cnames={'Max Excursion Pre';'RMS COPx Veli'
'Pre';...'
    'RMS COPy Veli Post';'RMS COPx Veli'
'Post';...'
    'Sway Area Pre';'Sway Area Post'};
columnformat = {'bank', 'bank', 'bank', 'bank', 'bank', 'short',
    'short'};
\textbf{rnames}=\{'Straight Front', 'Diagonal Front Left', 'Straight Left', 'Diagonal
Back Left',...'
    'Straight Back', 'Diagonal Back Right', 'Straight Right',...'
    'Diagonal Front Right'};
\textbf{parent} = figure('Position',[120 200 1150 250]);
figure(1)
uicontrol('String',settitle,'Position',[20 225 200 30]);
\textbf{TABLE} =uitable('Parent',parent,'Data',dat,'\textbf{ColumnName}',cnames, ...'
    'RowName',rnames,'\textbf{Position}',[20 20 1100 200],'\textbf{ColumnFormat}',
columnformat);

figure(2)
h1=compass(U_pre,V_pre,'r');
hold on
h2=compass(U_post,V_post,'--g');
title(strvcat('Max Excursions in Each Direction:','red=pre fatigue, green=post fatigue'))

hold off

figure(3)
plot(pre_COPx(:,1), pre_COPy(:,1), 'r.' )
hold on
plot(post_COPx(:,1), post_COPy(:,1), 'g.' )
title('forward')
xlabel('COPx'), ylabel('COPy')
legend('pre', 'post')
k_pre = convhull(pre_COPx(:,1),pre_COPy(:,1));
k_post= convhull(post_COPx(:,1),post_COPy(:,1));
plot(pre_COPx(k_pre,1),pre_COPy(k_pre,1), 'r- ', post_COPx(k_post,1),post_COPy(k_post,1), 'g- ')
hold off

figure(4)
plot(pre_COPx(:,4), pre_COPy(:,4), 'r.' )
hold on
plot(post_COPx(:,4), post_COPy(:,4), 'g.' )
title('diagonal back left')
xlabel('COPx'), ylabel('COPy')
legend('pre', 'post')
k_pre = convhull(pre_COPx(:,4),pre_COPy(:,4));
k_post= convhull(post_COPx(:,4),post_COPy(:,4));
plot(pre_COPx(k_pre,4),pre_COPy(k_pre,4), 'r- ', post_COPx(k_post,4),post_COPy(k_post,4), 'g- ')
hold off

figure(5)
plot(pre_COPx(:,6), pre_COPy(:,6), 'r.' )
hold on
plot(post_COPx(:,6), post_COPy(:,6), 'g.' )
title('diagonal back right')
xlabel('COPx'), ylabel('COPy')
legend('pre', 'post')
k_pre = convhull(pre_COPx(:,6),pre_COPy(:,6));
k_post= convhull(post_COPx(:,6),post_COPy(:,6));
plot(pre_COPx(k_pre,6),pre_COPy(k_pre,6), 'r- ', post_COPx(k_post,6),post_COPy(k_post,6), 'g- ')
hold off

%% moment analysis
% bodyweight=input('bodyweight in kg:');
% height=input('height in cm:');
for pos=1:8
    max_pre_momentx(pos)=max(pre_momentx(:,pos));
    max_pre_momenty(pos)=max(pre_momenty(:,pos));
    max_pre_momentz(pos)=max(pre_momentz(:,pos));
    max_post_momentx(pos)=max(post_momentx(:,pos));
    max_post_momenty(pos)=max(post_momenty(:,pos));
    max_post_momentz(pos)=max(post_momentz(:,pos));

    max_pre_momentx(pos)=max_pre_momentx(pos)./(bodyweight);
    max_pre_momenty(pos)=max_pre_momenty(pos)./(bodyweight);

end
% max_pre_momentz(pos)=max_pre_momentz(pos)./(bodyweight);
% max_post_momentx(pos)=max_post_momentx(pos)./(bodyweight);
% max_post_momenty(pos)=max_post_momenty(pos)./(bodyweight);
% max_post_momentz(pos)=max_post_momentz(pos)./(bodyweight);
end

for pos=1:8
    avg_pre_momentx(pos)=mean(pre_momentx(:,pos));
    avg_pre_momenty(pos)=mean(pre_momenty(:,pos));
    avg_pre_momentz(pos)=mean(pre_momentz(:,pos));
    avg_post_momentx(pos)=mean(post_momentx(:,pos));
    avg_post_momenty(pos)=mean(post_momenty(:,pos));
    avg_post_momentz(pos)=mean(post_momentz(:,pos));
end

%% pdf generator
name=settitle;
psname = strcat(name,'.ps');
print ('-dpsc2', psname, '-append', '-f1')
print ('-dpsc2', psname, '-append', '-f2')
print ('-dpsc2', psname, '-append', '-f3')
print ('-dpsc2', psname, '-append', '-f4')
print ('-dpsc2', psname, '-append', '-f5')
pdfname= strcat(name,'.pdf');
ps2pdf('psfile', psname, 'pdffile', pdfname, 'gspapersize', 'a4',
       'deletepsfile', 1)

%% output for excel
% these arrays hold the information that will be exported into excel for % statistical analysis.

%results= [(max excursion pre) (pre RMS COPx vel) (pre RMS COPy vel) (pre sway area); (max excursion post) (postRMS COPx vel) (postRMS COPy vel) (post sway area)]
result_1= [max_excursion_pre(1) VEL_preCOPx(1) VEL_preCOPy(1) pre_sway_A(1)]
            max_pre_momentx(1) avg_pre_momentx(1) max_pre_momenty(1) ...
            avg_pre_momenty(1) max_pre_momentz(1) avg_pre_momentz(1); ...
            max_excursion_post(1) VEL_postCOPx(1) VEL_postCOPy(1) post_sway_A(1); ...
            max_post_momentx(1) avg_post_momentx(1) max_post_momenty(1) ...
            avg_post_momenty(1) max_post_momentz(1) avg_post_momentz(1);
result_4= [max_excursion_pre(4) VEL_preCOPx(4) VEL_preCOPy(4) pre_sway_A(4)]
            max_pre_momentx(4) avg_pre_momentx(4) max_pre_momenty(4) ...
            avg_pre_momenty(4) max_pre_momentz(4) avg_pre_momentz(4); ...
max_excursion_post(4) VEL_postCOPx(4) VEL_postCOPy(4) post_sway_A(4) ...
max_post_momentx(4) avg_post_momentx(4) max_post_momenty(4) ...
avg_post_momenty(4) max_post_momentz(4) avg_post_momentz(4)];

result_6= [max_excursion_pre(6) VEL_preCOPx(6) VEL_preCOPy(6)
pre_sway_A(6) ...
max_pre_momentx(6) avg_pre_momentx(6) max_pre_momenty(6) ...
avg_pre_momenty(6) max_pre_momentz(6) avg_pre_momentz(6);...
max_excursion_post(6) VEL_postCOPx(6) VEL_postCOPy(6) post_sway_A(6) ...
max_post_momentx(6) avg_post_momentx(6) max_post_momenty(6) ...
avg_post_momenty(6) max_post_momentz(6) avg_post_momentz(6)];

varname=settitle;
filename=strcat('sebt_results_anterior');
S.(varname) = result_1;
save(filename, '-struct', 'S','-append');

varname=settitle;
filename=strcat('sebt_results_postmedial');
S.(varname) = result_4;
save(filename, '-struct', 'S','-append');

varname=settitle;
filename=strcat('sebt_results_postlateral');
S.(varname) = result_6;
save(filename, '-struct', 'S','-append');
close all

%%end code
%% lindsay_datamatrix.m
%This script will compile all jumps from one trial into a 3D matrix
%data were exported as .txt but can be form excel with few modifications
%each jump was recorded as a separate file. this portion grabs each file to
%compile it
root_name=input('enter the root of the file names you wish to use:', 's');
last_jump=input('enter the number on the end of the last file:');
disp ' ';disp '...building the matrix...';disp ' ';
for jj=0:1:last_jump
    f_name=strcat(root_name, sprintf('_%.4d', jj));
    f_name=strcat(f_name, '.exp');
    Y=load(f_name);
    if jj==0
        Z=Y(1:2000,:); use this if excel
        matrix_depth=last_jump+1;
        A = zeros(length(Z),16,matrix_depth); %pre-allocating- change depending on specific collection
    end
    %Y=xlsread(f_name); use this if excel
    %Z=Y(7:2000,:); use this if excel
    Z=Y(1:2000,:);
    A(:, :, jj+1) =Z;
end
disp ' ';disp 'finished!';disp ' ';

%% save the compiled data
%this section will save the 3D matrix that was just filled to the file
"data_matrices.mat". It will assign
%it a new variable name to match the root name.
ans=input('Do you want to save the compiled matrix? Y or N:', 's');
if ans=='y' || ans=='Y'
    S.(root_name) = A;
    save('data_matrices.mat', '-struct', 'S', '-append')
    disp 'saved in data_matrices.mat';
else
    disp 'data not saved';
end

%% end code
%% lindsay_main.m

% This is the main calculation script.
% lindsay_datamatrix.m must be run before this script.
% This script will find the following:
% Ground reaction force
% The start and stop point of the impact phase (first 200ms after impact)
% A plot verification of impact phase for each jump
% Max GRFv
% The time taken to reach the max GRFv
% The impulse of the impact phase of the GRFv
% The GRFv max loading rate
% Kinematics
% The maximum degree of hip, knee, and ankle flexion (sagital plane) and
% hip abduction and ankle inversion (frontal plane) during impact phase
% The angle of the joints at the moment of impact
% Inverse dynamics - calls a separate function to find angular
% accelerations and net joint moment
% The impulse of the NJM for the impact phase
% The joint power for the impact phase
% The work of the NJM for the impact phase

clear all
close all
clc

load data_matrices  % this variable file contains all compiled matrices

sampling_rate = 1000;  % set to 1000Hz
T = 1 / sampling_rate;  % period

% select the trial: ex. s12_jun11
root_name2=input('Please enter the root name of the trial you wish to access:
', 's');
root_name=eval(root_name2);

% Column names from raw data
% these should be changed if export from motion monitor is changed
reference_column=1;
hipflex_column=2;
hipabduc_column=3;
kneeflex_column=4;
ankleflex_column=5;
ankleinv_column=6;
% hipmomemtx_column=7;
hipmomemty_column=7;
hipmomemtz_column=8;
hipmomemty_column=9;
% kneemomemtx_column=10;
kneemomety_column=11;
kneemometz_column=12;
% ankelmomemtx_column=13;
anklemomety_column=14;
anklemometz_column=15;
GRFv_column=16;
%% Impact Phase Selection
%This portion will select the start and stop point for impact phase and
%allow the user to used graphic input to manually reset an impact phase if
%needed. Because of how the data collection was windowed, there is usually
%some times where the force is equal to zero. This loop selects the frame
%after the last zero as the start time.

[i,j,k]=size(root_name);

for jump=1:1:k
    GRFv_total(:,jump)=-1*root_name(:,GRFv_column,jump);
    [peak GRFv,time_to_peak_GRFv]=max(GRFv_total(:,jump),[],1);
    A=GRFv_total(1:time_to_peak_GRFv,jump);
    ind=find(A==0,1,'last');
    start(:,jump)=ind+1;
    stop(:,jump)=ind+201;
end

%% graphical verification of impact phase selection
%This part "plays back" what the computer selected for impact phase
verify=input('Do you wish to graphically verify the phase limits? Y or N:','s');
if strcmp(verify,'y')
    for which_one=1:1:k
        B=-1*root_name(:,GRFv_column,which_one);
        time=0:1:length(B)-1;
        plot(time,B,'b'); hold on;
        x1=[start(which_one),start(which_one)];
        x2=[stop(which_one),stop(which_one)];
        y=[min(B), max(B)];
        plot(x1,y,'r',x2,y,'r');
        title(strcat('Impact Phase Verification. Jump:',num2str(which_one)))
        xlabel('time in ms');
        ylabel('Ground Reaction Force / Z');
        hold off;
        pause(2)
        close all
    end
end

%% Manual impact phase selection
%this portion allows the user to choose if they would like to reset the
%start/stop values. new start value is selected graphically
verify=input('manually set a specific jumps phase limits? Y or N:','s');
while verify=='y'||verify=='Y'
    which_one=input('Which Jump: ');
    B=-1*root_name(:,GRFv_column,which_one);
    time=0:1:length(B)-1;
    plot(time,B,'b'); hold on;
    axis([350 600 0 500]);
    [start(which_one),~] = ginput(1);
    start(which_one)=floor(start(which_one));
    stop(which_one)=start(which_one)+200;
    x1=[start(which_one),start(which_one)];
    x2=[stop(which_one),stop(which_one)];
y=[min(B), max(B)];
plot(x1,y,'r',x2,y,'r')
title(strcat('Impact Phase Verification. Jump:',num2str(which_one)))
ylabel('Ground Reaction Force / Z');
xlabel('time in ms');
hold off;
pause(2)
close all
verify=input('Another? y or n: ','s');

%% GRFv
%% This portion calculates the
%% max GRFv, the time taken to reach the max GRFv,
%% the GRFv impulse, and max loading rate
for jump=1:1:k
    GRFv(:,jump)=GRFv_total(start(jump):stop(jump),jump);
end

%active peak = max GRFv or second peak
for jump=1:1:k
    [max_GRFv,time_to_max_GRFv]=max(GRFv,[],1);
end

%impact peak= first peak in the GRFv
for jump=1:1:k;
    %temporary filter is used to reduce noise and allow clearer active peak
    %selection
    [b,a] = butter(30,150/(1000/2),'low');
    CC=filtfilt(b,a,GRFv(:,jump));
    [pks,locs]=findpeaks(CC);
    if numel(pks)==0
        impact_peak(jump)=0;
        time_to_impact(jump)=1;
        disp(strcat('no impact peak detected for jump: ',num2str(jump)));
    elseif abs(time_to_max_GRFv(jump)-locs(1))<=25;
        impact_peak(jump)=0;
        time_to_impact(jump)=1;
        disp(strcat('no impact peak detected for jump: ',num2str(jump)));
    elseif locs(1)<=3;
        [impact_peak(jump),time_to_impact(jump)]=max(GRFv(locs(2)-
            3:locs(2)+3,jump));
    else
        [impact_peak(jump),time_to_impact(jump)]=max(GRFv(locs(1)-
            3:locs(1)+3,jump));
    end
end

%allows the user to "play back" the impact and active peak selections
verify=input('Do you wish to graphically verify the impact and active peaks?
Y or N: ','s');
if strcmp(verify,'y')
    for jump=1:1:k
        C=GRFv(:,jump);
end


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plot(C,'k'); hold on;
tempmax=[max_GRFv(jump), max_GRFv(jump)];
tempimpact=[impact_peak(jump), impact_peak(jump)];
y=[0, 250];
plot(y,tempmax,'r-',y,tempimpact,'g-')
axis([0 250 0 3500])
title(strcat('MAX Verification. Jump:',num2str(jump)))
ylabel('Ground Reaction Force / Z');
xlabel('time in ms');
hold off;
pause(2)
close all
end
end
%% allows the user to manually reset the impact selections
% using graphic input, the user visually identifies the impact peak and
% clicks once before and once after it. program finds the max in that
% window
manual_set=input('Do you want to manually set any of the impact peaks? Y or N :','s');
while strcmp(manual_set,'y')
    which_one=input('Please enter the jump# you wish to reset: ');
    D=GRFv(:,which_one);
    plot(D,'k');hold on;
    [x,~] = ginput(2);
    pre=round(x(1));
    post=round(x(2));

    [impact_peak(which_one),time_to_impact(which_one)]=max(GRFv(pre:post,which_one));
    tempmax=[max_GRFv(which_one), max_GRFv(which_one)];
    tempimpact=[impact_peak(which_one), impact_peak(which_one)];
    y=[0, 250];
    plot(y,tempmax,'r-',y,tempimpact,'g-')
    axis([0 250 0 3000])
    title(strcat('MAX Verification. Jump:',num2str(which_one)))
    ylabel('Ground Reaction Force / Z');
    xlabel('time in ms');
    hold off;
    pause(2)
close all

    manual_set=input('Manually set another? Y or N:','s');
end

%GRFv impulse and loading rate
for jump=1:1:k
    GRFv_impulse(jump)=trapz(GRFv(:,jump));
end

for jump=1:1:k
    for step=1:1:(length(GRFv)-1)
        GRFv_load_rate(step,jump)= (GRFv(step+1,jump)-GRFv(step,jump))/T;
    end
end
GRFv_load_rate=transpose(GRFv_load_rate);  
max_GRFv_load_rate=max(GRFv_load_rate,[],2);  
%% The maximum degree of hip, knee, and ankle flexion and hip abduction  
% during impact phase  
% note that by convention a positive ankle flexion indicates dorsiflexion  
and negative  
% indicates plantar flexion

for jump=1:1:k

   hip_flexion(:,jump)=root_name(start(jump):stop(jump),hipflex_column,jump);
   knee_flexion(:,jump)=root_name(start(jump):stop(jump),kneeflex_column,jump);
   ankle_flexion(:,jump)=root_name(start(jump):stop(jump),ankleflex_column,jump);
   hip_abduction(:,jump)=root_name(start(jump):stop(jump),hipabduc_column,jump);
   ankle_inversion(:,jump)=root_name(start(jump):stop(jump),ankleinv_column,jump);

end  
% correction values taken from the neutral stance views
hip_flexion_correction=input('please enter the neutral stance hip flexion value: ');  
knee_flexion_correction=input('please enter the neutral stance knee flexion value: ');  
ankle_flexion_correction=input('please enter the neutral stance ankle flexion value: ');  
hip_abd_correction=input('please enter the neutral stance hip ABduction value: ');  
ankle_inv_correction=input('please enter the neutral stance ankle inversion value: ');  

   hip_flexion_correction=0-hip_flexion_correction;
   knee_flexion_correction=0-knee_flexion_correction;
   ankle_flexion_correction=0-ankle_flexion_correction;
   hip_abd_correction=0-hip_abd_correction;
   ankle_inv_correction=0-ankle_inv_correction;

   hip_flexion=hip_flexion+hip_flexion_correction;
   knee_flexion=knee_flexion+knee_flexion_correction;
   ankle_flexion=ankle_flexion+ankle_flexion_correction;
   hip_abdduction=hip_abdduction+hip_abd_correction;
   ankle_inversion=ankle_inversion+ankle_inv_correction;

max_hip_flexion=max(hip_flexion,[],1);
max_knee_flexion=max(knee_flexion,[],1);
max_ankle_flexion=max(ankle_flexion,[],1);
max_hip_abduction=max(hip_abduction,[],1);
max_ankle_inversion=max(ankle_inversion,[],1);

% hip_extension=-hip_flexion;
% knee_extension=-knee_flexion;
% ankle_extension=-ankle_flexion;
% hip_Abduction=-hip_abduction;
%% The angle of the joints at the moment of impact peak
for jump=1:1:k
    hip_flexion_impact(jump)=hip_flexion(time_to_impact(jump),jump);
    knee_flexion_impact(jump)=knee_flexion(time_to_impact(jump),jump);
    ankle_flexion_impact(jump)=ankle_flexion(time_to_impact(jump),jump);
    hip_abduction_impact(jump)=hip_abduction(time_to_impact(jump),jump);
    ankle_inversion_impact(jump)=ankle_inversion(time_to_impact(jump),jump);
end

%% The angle of the joints at the moment of active peak
for jump=1:1:k
    hip_flexion_active(jump)=hip_flexion(time_to_max_GRFv(jump),jump);
    knee_flexion_active(jump)=knee_flexion(time_to_max_GRFv(jump),jump);
    ankle_flexion_active(jump)=ankle_flexion(time_to_max_GRFv(jump),jump);
    hip_abduction_active(jump)=hip_abduction(time_to_max_GRFv(jump),jump);
    ankle_inversion_active(jump)=ankle_inversion(time_to_max_GRFv(jump),jump);
end

%% The impact phase joint impulse (area under the moment curve)
BW=input('please enter body weight in kilograms: ');
BWn=BW*9.80665002864; %kilos to newtons
LL=input('please enter height in centimeters: ');
LL=LL/100; %height in meters
norm=BW*LL;

%if using the moments output by MotionMonitor, the following line is not
%needed and the moments in the following for loop should be commented
%back
[ ankle_moment_x,knee_moment_x,hip_moment_x ] =
inverse_dynamics(root_name2,start,stop,BW,LL,T,hip_flexion,knee_flexion,ankle_flexion);

for jump=1:1:k
    hip_moment_x(:,jump)=root_name(start(jump):stop(jump),hipmomemtx_column,jump);
    hip_moment_y(:,jump)=root_name(start(jump):stop(jump),hipmomemty_column,jump);
    hip_moment_z(:,jump)=root_name(start(jump):stop(jump),hipmomemtz_column,jump);
end
% knee_moment_x(:, jump) = root_name(start(jump):stop(jump), kneemomemtx_column, jump);
% knee_moment_y(:, jump) = root_name(start(jump):stop(jump), kneemomemty_column, jump);
% knee_moment_z(:, jump) = root_name(start(jump):stop(jump), kneemomemtz_column, jump);
% ankle_moment_x(:, jump) = root_name(start(jump):stop(jump), anklemomemtx_column, jump);
% ankle_moment_y(:, jump) = root_name(start(jump):stop(jump), anklemomemty_column, jump);
% ankle_moment_z(:, jump) = root_name(start(jump):stop(jump), anklemomemtz_column, jump);
end

hip_moment_x = hip_moment_x ./ norm;
hip_moment_y = hip_moment_y ./ norm;
knee_moment_x = knee_moment_x ./ norm;
ankle_moment_x = ankle_moment_x ./ norm;
ankle_moment_y = ankle_moment_y ./ norm;

subplot(5,1,1)
plot(hip_moment_x)
title('hip moment sagittal')
subplot(5,1,2)
plot(knee_moment_x)
title('knee moment sagittal')
subplot(5,1,3)
plot(ankle_moment_x)
title('ankle moment sagittal')
subplot(5,1,4)
plot(hip_moment_y)
title('hip moment frontal')
subplot(5,1,5)
plot(ankle_moment_y)
title('ankle moment frontal')

momcheck = input('make sure moments are as expected. Continue? Y or N:','s');
if strcmp(momcheck,'n')
    break
end

for jump=1:1:k
    joint_impulse_hipx(jump)=trapz(hip_moment_x(:,jump));
    joint_impulse_hipy(jump)=trapz(hip_moment_y(:,jump));
    joint_impulse_kneex(jump)=trapz(knee_moment_x(:,jump));
    joint_impulse_anklex(jump)=trapz(ankle_moment_x(:,jump));
    joint_impulse_ankley(jump)=trapz(ankle_moment_y(:,jump));
end

joint_impulse_hipx=joint_impulse_hipx.*T;
% The impact phase joint work
% power is the moment times the angular velocity
% work is the area under the power curve (for the impact phase)

%angular velocities calculated here

for jump=1:1:k
    for step=1:1:200
        ang_vel_hip_flex(step,jump) = (hip_flexion(step+1,jump) - hip_flexion(step,jump))/T*pi/180;
    end
    for step=1:1:200
        ang_vel_hip_abduc(step,jump) = (hip_abduction(step+1,jump) - hip_abduction(step,jump))/T*pi/180;
    end
    for step=1:1:200
        ang_vel_knee_flex(step,jump) = (knee_flexion(step+1,jump) - knee_flexion(step,jump))/T*pi/180;
    end
    for step=1:1:200
        ang_vel_ankle_flex(step,jump) = (ankle_flexion(step+1,jump) - ankle_flexion(step,jump))/T*pi/180;
    end
    for step=1:1:200
        ang_vel_ankle_inversion(step,jump) = (ankle_inversion(step+1,jump) - ankle_inversion(step,jump))/T*pi/180;
    end
end

%joint power for sagittal and frontal(hip and ankle)

for jump=1:1:k
    for step=1:1:199
        joint_power_hipx(step,jump) = hip_moment_x(step,jump) .* ang_vel_hip_flex(step,jump);
    end
    for step=1:1:199
        joint_power_hipy(step,jump) = hip_moment_y(step,jump) .* ang_vel_hip_abduc(step,jump);
    end
    for step=1:1:199
        joint_power_kneex(step,jump) = knee_moment_x(step,jump) .* ang_vel_knee_flex(step,jump);
    end
    for step=1:1:199
        joint_power_anklex(step,jump) = ankle_moment_x(step,jump) .* ang_vel_ankle_flex(step,jump);
    end
    for step=1:1:199
        joint_power_ankley(step,jump) = ankle_moment_y(step,jump) .* ang_vel_ankle_inversion(step,jump);
    end
end
end
end

% joint work for sagital and frontal

for jump=1:1:k
    joint_work_hipx(jump)=trapz(joint_power_hipx(:,jump));
    joint_work_hipy(jump)=trapz(joint_power_hipy(:,jump));
    joint_work_kneex(jump)=trapz(joint_power_kneex(:,jump));
    joint_work_anklex(jump)=trapz(joint_power_anklex(:,jump));
    joint_work_ankley(jump)=trapz(joint_power_ankley(:,jump));
end

joint_work_hipx=joint_work_hipx.*T;
joint_work_hipy=joint_work_hipy.*T;
joint_work_kneex=joint_work_kneex.*T;
joint_work_anklex=joint_work_anklex.*T;
joint_work_ankley=joint_work_ankley.*T;
break

%%
% clears the irrelevant variables loaded with data_matrices.mat and saves
% all calculated variables of interest for further analysis

clear s7_sept10 s9_nov18 s10_nov5 s11_oct27 s13_oct13 s14_jan31 s15_feb3 s16_feb4...
    s17_feb9 s18_feb9 s20_mar3 s21_mar3 s22_mar4 s23_mar10 s24_mar10 s25_mar11...
    s26_mar11 s27_mar12 s28_mar12 s29_mar17 s30_mar17 s31_mar20 s32_mar23 s33_mar25...
    s34_mar25 s35_mar27 s36_mar7 s37_mar7 s7_sept11 s9_nov20 s10_nov7 s11_oct29...
    s13_oct27 s14_feb3 s15_feb5 s16_feb6 s17_feb12 s18_feb11 s20_mar4 s21_mar5...
    s22_mar5 s23_mar12 s24_mar11 s25_mar13 s26_mar12 s27_mar13 s28_mar13 s29_mar23...
    s30_mar21 s31_mar24 s32_mar24 s33_mar27 s34_mar27 s35_mar2 s36_mar8 s37_mar9

disp 'Finished! Saving';

% save(root_name2)
clear
clc

% end code
function [ A ] = invdyn_matrix( root_name, last_jump )
% builds 3d matrices from the segment and forceplate data needed
% for inverse dynamics calculations,
% rows= data, columns=variables, depth=all jumps
for jj=0:1:last_jump
    f_name=strcat(root_name,sprintf('_%.4d',jj));
    f_name=strcat(f_name, '.exp');

    %Y=xlsread(f_name); use this if excel
    Y=load(f_name);

    if jj==0
        %Z=Y(7:2000,:); use this if excel
        Z=Y(1:2000,:);
        matrix_depth=last_jump+1;
        A = zeros(length(Z),11,matrix_depth); %pre-allocating- change depending on specific collection
    end

    %Y(7:2000,:);
    Z=Y(1:2000,:);
    A(:,:,jj+1) =Z;
end

% ay_thigh(:,nn)=root_name(start:stop,4,nn); %linear accelerations at the CoM
% az_thigh(:,nn)=root_name(start:stop,5,nn);
% ay_shank(:,nn)=root_name(start:stop,6,nn);
% az_shank(:,nn)=root_name(start:stop,7,nn);
% ay_foot(:,nn)=root_name(start:stop,8,nn);
% az_foot(:,nn)=root_name(start:stop,9,nn);
% FY(:,nn)=root_name(start:stop,10,nn); %ground reaction forces
% FZ(:,nn)=root_name(start:stop,11,nn);
% end code
function [ NJM_ankle,NJM_knee,NJM_hip ] = inverse_dynamics(root_name2,start,stop,BW,LL,T,hip_flexion,knee_flexion,ankle_flexion)
%calculates net joint moments in the sagittal plane
load invdyn_matrices
root_name=eval(root_name2);
[~,~,k]=size(root_name);

for nn=1:k
    ay_thigh(:,nn)=root_name(start:stop,4,nn); %linear accelerations at the CoM
    az_thigh(:,nn)=root_name(start:stop,5,nn);
    ay_shank(:,nn)=root_name(start:stop,6,nn);
    az_shank(:,nn)=root_name(start:stop,7,nn);
    ay_foot(:,nn)=root_name(start:stop,8,nn);
    az_foot(:,nn)=root_name(start:stop,9,nn);
    FY(:,nn)=root_name(start:stop,10,nn); %ground reaction forces
    FZ(:,nn)=root_name(start:stop,11,nn);
    M_grf(:,nn)=root_name(start:stop,12,nn);
end

theta_hip=hip_flexion*pi/180; %convert to radians
theta_knee=knee_flexion*pi/180;
theta_ankle=ankle_flexion*pi/180;

% angular accelerations
%BOXCAR function used below smooths the angular accelerations
ang_vel_hip= diff(theta_hip)./T;
ang_vel_knee= diff(theta_knee)./T;
ang_vel_ankle= diff(theta_ankle)./T;

[ang_vel_hip]= BOXCAR2( ang_vel_hip,10);
[ang_vel_knee]= BOXCAR2( ang_vel_knee,10);
[ang_vel_ankle]= BOXCAR2( ang_vel_ankle,10);

ang_accel_hip= diff(ang_vel_hip)./T;
ang_accel_knee= diff(ang_vel_knee)./T;
ang_accel_ankle= diff(ang_vel_ankle)./T;

[ang_accel_hip]= BOXCAR2( ang_accel_hip,10);
[ang_accel_knee]= BOXCAR2( ang_accel_knee,10);
[ang_accel_ankle]= BOXCAR2( ang_accel_ankle,10);

% anthropometrics
mass=BW; %in kilograms
height=LL; %in meters
gravity=9.80665002864;

%segment length from Drillis and Contini(1966)
SegmentLength_thigh=.245*height;
SegmentLength_shank=.246*height;
SegmentLength_foot=.152*height;
% segment weight, CoM distance, and segment mass moment of inertia from
Dempster
Dist_CoM2thigh=0.433*SegmentLength_thigh;
Dist_CoM2shank=0.433*SegmentLength_shank;
Dist_CoM2foot=0.5*SegmentLength_foot;

SegmentMass_thigh=0.1*mass;
SegmentMass_shank=0.0465*mass;
SegmentMass_foot=0.0145*mass;

I_thigh=SegmentMass_thigh*(SegmentLength_thigh*0.323)^2;
I_shank=SegmentMass_shank*(SegmentLength_shank*0.302)^2;
I_foot=SegmentMass_foot*(SegmentLength_foot*0.475)^2;

SegmentWeight_hip=0.1*mass*gravity;
SegmentWeight_knee=0.0465*mass*gravity;
SegmentWeight_ankle=0.0145*mass*gravity;

%% ankle formulas
for jump=1:k
    for aa=1:length(ang_accel_ankle)
        Ry_ankle(aa,jump)=SegmentMass_foot*ay_foot(aa,jump)-FY(aa,jump);
        Rz_ankle(aa,jump)=SegmentMass_foot*az_foot(aa,jump)+FZ(aa,jump)-SegmentWeight_ankle;

        NJM_ankle(aa,jump)=I_foot*ang_accel_ankle(aa,jump)+SegmentWeight_ankle*(Dist_CoM2foot*cos(theta_ankle(aa,jump))... -FY(aa,jump)*(SegmentLength_foot*sin(theta_ankle(aa,jump)))-FZ(aa,jump)*(SegmentLength_foot*cos(theta_ankle(aa,jump)))-M_grf;
    end
end

%% knee formulas
for jump=1:k
    for kk=1:length(ang_accel_knee)
        Ry_knee(kk,jump)=SegmentMass_shank.*ay_shank(kk,jump)+Ry_ankle(kk,jump);
        Rz_knee(kk,jump)=SegmentMass_shank.*az_shank(kk,jump)+Rz_ankle(kk,jump)-SegmentWeight_knee;

        NJM_knee(kk,jump)=I_shank*ang_accel_knee(kk,jump)+NJM_ankle(kk,jump)-SegmentWeight_knee*(Dist_CoM2shank*sin(theta_knee(kk,jump))... +Ry_ankle(kk,jump)*(SegmentLength_shank*cos(theta_knee(kk,jump)))+Rz_ankle(kk,jump)*(SegmentLength_shank*sin(theta_knee(kk,jump)));
    end
end

%% hip formulas
for jump=1:k
    for hh=1:length(ang_accel_hip)
        Ry_hip(hh,jump)=SegmentMass_thigh*ay_thigh(hh,jump)+Ry_knee(hh,jump);
Rz_hip(hh,jump)=SegmentMass_thigh*az_thigh(hh,jump)+Rz_knee(hh,jump)-SegmentWeight_hip;

NJM_hip(hh,jump)=I_thigh*ang_accel_hip(hh,jump)+NJM_knee(hh,jump)+SegmentWeight_hip*(Dist_CoMthigh*sin(\theta_hip(hh,jump)))...+
Ry_knee(hh,jump)*(SegmentLength_thigh*cos(\theta_hip(hh,jump)))-Rz_knee(hh,jump)*(SegmentLength_thigh*sin(\theta_hip(hh,jump)));

end

end

% M(:,:,1)=NJM_ankle;
% M(:,:,2)=NJM_knee;
% M(:,:,3)=NJM_hip;
%

end

% end code
function [ adj_metric, std_frame ] = BOXCAR2( metric, frame_size)

%BOXCAR2.m moving average of raw data
% frame_size selects how many of the surrounding data points are averaged
% metric is any data where data is in columns and each column is new jump

[l,d]=size(metric);
int=frame_size-1;
for jump=1:d
    for n=1:1:int;
        adj_metric(n,jump)=mean(metric(n:n+int,jump));
        std_frame(n,jump)=std(metric(n:n+int,jump));
    end
end
end

% end code
%lindsay_split2.m
%loads all the raw data that was calculated by lindsay_main
%divides it into 10% intervals and takes the average of those intervals
%using the SEGMENTER.m function.
%finds a second order polynomial fit using the FIT2.m function
%plots the data

clear all
close all

settitle=input('Please enter the name of the file you wish to load: ', 's');
load(settitle)
settitle=strcat(settitle, ' with 10percent 2nd');

%average the jump values to get cycle values (not used until the end)
cycles=floor(length(max_GRFv)/2);
% additional calculations
% peak to peak time
for jj=1:length(max_GRFv)
    p2p_time(jj)=time_to_max_GRFv(jj)-time_to_impact(jj);
end

p2p_ankeflex=ankle_flexion_active-ankle_flexion_impact;
p2p_kneeflex=knee_flexion_active-knee_flexion_impact;
p2p_histflex=hip_flexion_active-hip_flexion_impact;
p2p_hipabdup=hip_abduction_active-hip_abduction_impact;
p2p_ankleinv=ankle_inversion_active-ankle_inversion_impact;
% average the jumps into 10% intervals
%if the person jumped less than 10 times, no averaging occurs.
jumps=length(max_GRFv);
if jumps>10;
    [ max_GRFv,mg_std ] = SEGMENTER( max_GRFv);
end

jumps=length(impact_peak);
if jumps>10;
    [ impact_peak,ip_std ] = SEGMENTER( impact_peak);
end

jumps=length(time_to_max_GRFv);
if jumps>10;
    [ time_to_max_GRFv,ttmg_std ] = SEGMENTER( time_to_max_GRFv);
else
    time_to_max_GRFv=transpose(time_to_max_GRFv);
end

jumps=length(GRFv_impulse);
if jumps>10;
    [ GRFv_impulse,gi_jwa_std ] = SEGMENTER( GRFv_impulse);
end

jumps=length(max_GRFv_load_rate);
if jumps>10;
    [ max_GRFv_load_rate,mglr_std ] = SEGMENTER( max_GRFv_load_rate);
else if jumps<=10
max_GRFv_load_rate=transpose(max_GRFv_load_rate);
end

%% Degrees flexion: max and peak2peak

jumps=length(p2p_ankleflex);
if jumps>10;
    [p2p_ankleflex,p2p_ankle_std] = SEGMENTER( p2p_ankleflex);
end

jumps=length(p2p_ankleinv);
if jumps>10;
    [p2p_ankleinv,p2p_ankleinv_std] = SEGMENTER( p2p_ankleinv);
end

jumps=length(p2p_kneeflex);
if jumps>10;
    [p2p_kneeflex,p2p_knee_std] = SEGMENTER( p2p_kneeflex);
end

jumps=length(p2p_hipflex);
if jumps>10;
    [p2p_hipflex,p2p_hip_std] = SEGMENTER( p2p_hipflex);
end

jumps=length(p2p_hipabd);
if jumps>10;
    [p2p_hipabd,p2p_hipabd_std] = SEGMENTER( p2p_hipabd);
end

jumps=length(max_hip_flexion);
if jumps>10;
    [max_hip_flexion,hf_std] = SEGMENTER( max_hip_flexion);
end

jumps=length(max_hip_abduction);
if jumps>10;
    [max_hip_abduction,hab_std] = SEGMENTER( max_hip_abduction);
end

jumps=length(max_knee_flexion);
if jumps>10;
    [max_knee_flexion,kf_std] = SEGMENTER( max_knee_flexion);
end

jumps=length(max_ankle_flexion);
if jumps>10;
    [max_ankle_flexion,af_std] = SEGMENTER( max_ankle_flexion);
end

jumps=length(max_ankle_inversion);
if jumps>10;
    [max_ankle_inversion,ai_std] = SEGMENTER( max_ankle_inversion);
end

%%

jumps=length(joint_impulse_hipx);
if jumps>10;
    [ joint_impulse_hipx,jih_std ] = SEGMENTER( joint_impulse_hipx);
end

jumps=length(joint_impulse_kneex);
if jumps>10;
    [ joint_impulse_kneex,jik_std ] = SEGMENTER( joint_impulse_kneex);
end

jumps=length(joint_impulse_anklex);
if jumps>10;
    [ joint_impulse_anklex,jia_std ] = SEGMENTER( joint_impulse_anklex);
end

jumps=length(joint_impulse_ankley);
if jumps>10;
    [ joint_impulse_ankley,jiay_std ] = SEGMENTER( joint_impulse_ankley);
end

jumps=length(joint_impulse_hipy);
if jumps>10;
    [ joint_impulse_hipy,jihy_std ] = SEGMENTER( joint_impulse_hipy);
end

jumps=length(joint_work_hipx);
if jumps>10;
    [ joint_work_hipx,jha_std ] = SEGMENTER( joint_work_hipx);
end

jumps=length(joint_work_kneex);
if jumps>10;
    [ joint_work_kneex,jwk_std ] = SEGMENTER( joint_work_kneex);
end

jumps=length(joint_work_anklex);
if jumps>10;
    [ joint_work_anklex,jwa_std ] = SEGMENTER( joint_work_anklex);
end

jumps=length(joint_work_ankley);
if jumps>10;
    [ joint_work_ankley,jway_std ] = SEGMENTER( joint_work_ankley);
end

jumps=length(joint_work_hipy);
if jumps>10;
    [ joint_work_hipy,jwhy_std ] = SEGMENTER( joint_work_hipy);
end

% find the unfatigued and fatigues values and display a table

aa=[max_GRFv(1) max_GRFv(length(max_GRFv)) max_GRFv(length(max_GRFv))
    max_GRFv(1)];
aa_1=[impact_peak(1) impact_peak(length(impact_peak))
    impact_peak(length(impact_peak))
    impact_peak(1)];
bb=[time_to_max_GRFv(1) time_to_max_GRFv(length(time_to_max_GRFv)) ...
time_to_max_GRFv(length(time_to_max_GRFv)) - time_to_max_GRFv(1);
cc=[GRFv_impulse(1) GRFv_impulse(length(GRFv_impulse)) ... 
   GRFv_impulse(length(GRFv_impulse)) - GRFv_impulse(1)];
dd=[max_GRFv_load_rate(1) max_GRFv_load_rate(length(max_GRFv_load_rate)) ... 
   max_GRFv_load_rate(length(max_GRFv_load_rate)) - max_GRFv_load_rate(1)];
ee=[max_hip_flexion(1) max_hip_flexion(length(max_hip_flexion)) ... 
   max_hip_flexion(length(max_hip_flexion)) - max_hip_flexion(1)];
ff=[max_hip_abduction(1) max_hip_abduction(length(max_hip_abduction)) ... 
   max_hip_abduction(length(max_hip_abduction)) - max_hip_abduction(1)];
gg=[max_knee_flexion(1) max_knee_flexion(length(max_knee_flexion)) ... 
   max_knee_flexion(length(max_knee_flexion)) - max_knee_flexion(1)];
hh=[max_ankle_flexion(1) max_ankle_flexion(length(max_ankle_flexion)) ... 
   max_ankle_flexion(length(max_ankle_flexion)) - max_ankle_flexion(1)];
ii=[joint_impulse_hipx(1) joint_impulse_hipx(length(joint_impulse_hipx)) ... 
   joint_impulse_hipx(length(joint_impulse_hipx)) - joint_impulse_hipx(1)];
jj=[joint_impulse_hipy(1) joint_impulse_hipy(length(joint_impulse_hipy)) ... 
   joint_impulse_hipy(length(joint_impulse_hipy)) - joint_impulse_hipy(1)];
kk=[joint_impulse_kneex(1) joint_impulse_kneex(length(joint_impulse_kneex)) ... 
   joint_impulse_kneex(length(joint_impulse_kneex)) - joint_impulse_kneex(1)];
ll=[joint_impulse_anklex(1) joint_impulse_anklex(length(joint_impulse_anklex)) ... 
   joint_impulse_anklex(length(joint_impulse_anklex)) - joint_impulse_anklex(1)];
mn=[joint_work_hipx(1) joint_work_hipx(length(joint_work_hipx)) ... 
   joint_work_hipx(length(joint_work_hipx)) - joint_work_hipx(1)];
nn=[joint_work_hipy(1) joint_work_hipy(length(joint_work_hipy)) ... 
   joint_work_hipy(length(joint_work_hipy)) - joint_work_hipy(1)];
no=[joint_work_kneex(1) joint_work_kneex(length(joint_work_kneex)) ... 
   joint_work_kneex(length(joint_work_kneex)) - joint_work_kneex(1)];
pp=[joint_work_anklex(1) joint_work_anklex(length(joint_work_anklex)) ... 
   joint_work_anklex(length(joint_work_anklex)) - joint_work_anklex(1)];
dat=[aa; aa_1; bb; cc; dd; ee; ff; gg; hh; ii; jj; kk; ll; mm; nn; oo; pp];
cnames={'Unfatigued'; 'Fatigued'; 'Change'};
columnformat = {'bank', 'bank', '+'};
rnames={'GRFv max (N)', 'impact peak (N)', 'GRFv max time (msec)', 'GRFv impulse', ... 
   'GRFv max loading rate', 'max hip flexion', 'max hip abduction', ... 
   'max knee flexion', 'max ankle dorsiflexion', 'hip flexion impulse', ... 
   'hip abduction impulse', 'knee flexion impulse', 'ankle flexion impulse' ... 
   ', hip ext negative work', 'hip abduction negative work', 'knee ext negative work', 'ankle pf negative work'};
parent = figure('Position',[200 200 625 400]);
figure(1)
uc = uicontrol('String', settitle, 'Position', [20 350 200 30]);
TABLE =uitable('Parent', parent, 'Data', dat, 'ColumnName', cnames, ... 
   'RowName', rnames, 'Position', [20 20 500 325], 'ColumnFormat', columnformat);
%% GRFv max (active peak) and impact peak
%line of best FIT and remove outliers using the FIT2.m function

[FIT2_impact_peak, adj_impact_peak, ~, r_squared_1] = FIT2(impact_peak);
[FIT2_max_GRFv, adj_max_GRFv, ~, r_squared_2] = FIT2(max_GRFv);

%plot
figure(2)
jumps = 1:1:length(max_GRFv);
plot(adj_impact_peak, 'ko')
title('Impact peak and active peak as fatigue progresses')
hold on
plot(jumps, FIT2_impact_peak, 'g')
plot(adj_max_GRFv, 'bo'), xlabel('jump'), ylabel('GRFv in Newtons')
plot(jumps, FIT2_max_GRFv, 'r')
hold off
legend('impact peak', strcat('trendline with R^2=', num2str(r_squared_1)), ...
       'active peak', strcat('trendline with R^2=', num2str(r_squared_2)),'Location', 'NorthEastOutside')

%% time to max
[FIT2_time_to_max_GRFv, adj_time_to_max_GRFv, ~, r_squared_3] = FIT2(time_to_max_GRFv);

%% GRFv loading rate
[FIT2_max_GRFv_load_rate, adj_max_GRFv_load_rate, ~, r_squared_4] = FIT2(max_GRFv_load_rate);
figure(3)
plot(adj_max_GRFv_load_rate, 'o'), xlabel('jump'), ylabel('Loading Rate in Newtons/s')
title('Max GRFv Loading Rate as fatigue progresses')
hold on
plot(jumps, FIT2_max_GRFv_load_rate, 'r')
legend('GRFv loading rate', strcat('trendline with R^2=', num2str(r_squared_4)),'Location', 'NorthEastOutside')

%% GRFv impulse
[FIT2_GRFv_impulse, adj_GRFv_impulse, ~, r_squared_5] = FIT2(GRFv_impulse);
figure(4)
plot(adj_GRFv_impulse, 'o'), xlabel('jump'), ylabel('Impulse in Newton seconds')
title('GRFv impulse as fatigue progresses')
hold on
plot(jumps, FIT2_GRFv_impulse, 'r')
legend('GRFv impulse', strcat('trendline with R^2=', num2str(r_squared_5)),'Location', 'NorthEastOutside')

%% Maximum Flexions
% hip flex
[FIT2_max_hip_flexion, adj_max_hip_flexion, ~, r_squared_6] = FIT2(max_hip_flexion);
[FIT2_max_hip_abduction, adj_max_hip_abduction, slope_7, r_squared_7] = FIT2(max_hip_abduction);
figure(5)
plot(adj_max_hip_flexion, 'bo')
hold on
plot(FIT2_max_hip_flexion,'b--')
plot(adj_max_hip_abduction,'ro')
plot(FIT2_max_hip_abduction,'r--')

% knee flex
[FIT2 max knee flexion,adj_max_knee_flexion,~,r_squared_8] =
FIT2(max_knee_flexion);
plot(adj_max_knee_flexion,'ko')
plot(FIT2_max_knee_flexion,'k--')
% ankle flex
[FIT2 max ankle flexion,adj_max_ankle_flexion,~,r_squared_9] =
FIT2(max_ankle_flexion);
[FIT2 max ankle inversion,adj_max_ankle_inversion,~,r_squared_9_inversion] =
FIT2(max_ankle_inversion);
plot(adj_max_ankle_flexion,'go')
plot(FIT2_max_ankle_flexion,'g--')
plot(adj_max_ankle_inversion,'co')
plot(FIT2_max_ankle_inversion,'c--')

xlabel('jump'),ylabel('flexion in degrees')
title('Max Flexions as Fatigue Progresses')
legend('max hip flexion','trendline with
R^2=',num2str(r_squared_6)),...
'max hip abduction','trendline with
R^2=',num2str(r_squared_7)),...
'max knee flexion','trendline with R^2=',num2str(r_squared_8)),...
'max ankle dorsiflexion','trendline with
R^2=',num2str(r_squared_9)),...
'max ankle inversion','trendline with
R^2=',num2str(r_squared_9_inversion)),...
'Location','NorthEastOutside')
hold off

[~,adj_p2p_hipflex,~,r_squared_6_2] = FIT2(p2p_hipflex);
[~,adj_p2p_kneeflex,~,r_squared_8_2] = FIT2(p2p_kneeflex);
[~,adj_p2p_ankleflex,~,r_squared_9_2] = FIT2(p2p_ankleflex);
[~,adj_p2p_hipabd,~,r_squared_6_2abd] = FIT2(p2p_hipabd);
[~,adj_p2p_ankleinv,~,r_squared_9_2inv] = FIT2(p2p_ankleinv);

%% Joint Impulse
% hip impulse
[FIT2 joint impulse hipx,adj_joint_impulse_hipx,~,r_squared_10] =
FIT2(joint_impulse_hipx);

figure(6)
plot(adj_joint_impulse_hipx,'bo')
hold on
plot(jumps, FIT2_joint_impulse_hipx,'b--')

% knee impulse
[FIT2 joint impulse kneex,adj_joint_impulse_kneex,~,r_squared_12] =
FIT2(joint_impulse_kneex);
plot(adj_joint_impulse_kneex,'ko')
plot(FIT2_joint_impulse_kneex,'k--')

% ankle impulse
[FIT2_joint_impulse_anklex,adj_joint_impulse_anklex,~,r_squared_13] = 
FIT2(joint_impulse_anklex);

plot(adj_joint_impulse_anklex,'go')
plot(jumps, FIT2_joint_impulse_anklex,'g--')
xlabel('jump'),ylabel('Joint Impulse in in N*s')
title('Joint Impulse as Fatigue Progresses')
legend('hip flexion impulse',strcat('trendline with
R^2=',num2str(r_squared_10)),'...
'knee flexion impulse',strcat('trendline with
R^2=',num2str(r_squared_12)),'...
'ankle flexion impulse',strcat('trendline with
R^2=',num2str(r_squared_13)),'...
'Location','NorthEastOutside')
hold off

%% Joint impulse frontal plane
%hip abduction
[FIT2_joint_impulse_hipy,adj_joint_impulse_hipy,~,r_squared_11] = 
FIT2(joint_impulse_hipy);
%ankle inversion
[FIT2_joint_impulse_ankley,adj_joint_impulse_ankley,~,r_squared_13y] = 
FIT2(joint_impulse_ankley);

figure(7)
plot(adj_joint_impulse_hipy,'bo')
hold on
plot(jumps, FIT2_joint_impulse_hipy,'b--')
plot(adj_joint_impulse_ankley,'go')
plot(jumps, FIT2_joint_impulse_ankley,'g--')
legend('hip abduction impulse',strcat('trendline with
R^2=',num2str(r_squared_11)),'...
'ankle inversion impulse',strcat('trendline with
R^2=',num2str(r_squared_13y)),'...
'Location','NorthEastOutside')
hold off

%% Joint Negative Works
% hip negative work
[FIT2_joint_work_hipx,adj_joint_work_hipx,~,r_squared_14] = 
FIT2(joint_work_hipx);

figure(8)
jumps=1:1:length(joint_work_hipx);
plot(adj_joint_work_hipx,'bo')
hold on
plot(jumps,FIT2_joint_work_hipx,'b--')

% knee negative work
[FIT2_joint_work_kneex,adj_joint_work_kneex,~,r_squared_16] = 
FIT2(joint_work_kneex);
plot(adj_joint_work_kneex,'ko')
plot(jumps, FIT2_joint_work_kneex,'k--')

% ankle negative work
[FIT2_joint_work_anklex,adj_joint_work_anklex,~,r_squared_17] = FIT2(joint_work_anklex);

plot(adj_joint_work_anklex,'go')
plot(jumps, FIT2_joint_work_anklex,'g--')

xlabel('jump'),ylabel('Negative Work')
title('Joint Negative work as Fatigue Progresses')
legend('hip ext negative work',strcat('trendline with R^2=',num2str(r_squared_14)),...
    'knee ext negative work',strcat('trendline with R^2=',num2str(r_squared_16)),...
    'ankle pf negative work',strcat('trendline with R^2=',num2str(r_squared_17)),...
    'Location','NorthEastOutside')
hold off

% Joint work frontal plane

% hip abduction
[FIT2_joint_work_hipy,adj_joint_work_hipy,~,r_squared_15] = FIT2(joint_work_hipy);

% ankle inversion
[FIT2_joint_work_ankley,adj_joint_work_ankley,~,r_squared_17y] = FIT2(joint_work_ankley);

figure(9)
plot(adj_joint_work_hipy,'bo')
hold on
plot(jumps, FIT2_joint_work_hipy,'b--')
plot(adj_joint_work_ankley,'go')
plot(jumps, FIT2_joint_work_ankley,'g--')
legend('hip adduction work',strcat('trendline with R^2=',num2str(r_squared_15)),...
    'ankle eversion work',strcat('trendline with R^2=',num2str(r_squared_17y)),...
    'Location','NorthEastOutside')
hold off

% peak joint torque

abs_hip_moment_x = abs(hip_moment_x);
abs_knee_moment_x = abs(knee_moment_x);
abs_ankle_moment_x = abs(ankle_moment_x);
abs_hip_moment_y = abs(hip_moment_y);
abs_ankle_moment_y = abs(ankle_moment_y);

%dont wanna just do max in case it’s a negative torque
for cc=1:k
    [~,when1(cc)] = max(abs_hip_moment_x(:,cc));
    [~,when2(cc)] = max(abs_knee_moment_x(:,cc));
    [~,when3(cc)] = max(abs_ankle_moment_x(:,cc));
    [~,when4(cc)] = max(abs_hip_moment_y(:,cc));
    [~,when5(cc)] = max(abs_ankle_moment_y(:,cc));
peak_hip_moment_x(cc) = hip_moment_x(when1(cc),cc);
peak_knee_moment_x(cc) = knee_moment_x(when2(cc),cc);
peak_ankle_moment_x(cc) = ankle_moment_x(when3(cc),cc);
peak_hip_moment_y(cc) = hip_moment_y(when4(cc),cc);
peak_ankle_moment_y(cc) = ankle_moment_y(when5(cc),cc);
end

jumps=length(hip_moment_x);
if jumps>10;
   [ joint_torque_hipx, JThx_std ] = SEGMENTER(peak_hip_moment_x);
end

jumps=length(knee_moment_x);
if jumps>10;
   [ joint_torque_kneex, JTlx_std ] = SEGMENTER(peak_knee_moment_x);
end

jumps=length(ankle_moment_x);
if jumps>10;
   [ joint_torque_anklex, JTax_std ] = SEGMENTER(peak_ankle_moment_x);
end

jumps=length(hip_moment_y);
if jumps>10;
   [ joint_torque_hipy, JThy_std ] = SEGMENTER(peak_hip_moment_y);
end

jumps=length(ankle_moment_y);
if jumps>10;
   [ joint_torque_ankley, JTay_std ] = SEGMENTER(peak_ankle_moment_y);
end

% hip flex
[FIT2_joint_torque_hipx, adj_joint_torque_hipx, ~, r_squared_18] = FIT2(joint_torque_hipx);
% knee flex
[FIT2_joint_torque_kneex, adj_joint_torque_kneex, ~, r_squared_19] = FIT2(joint_torque_kneex);
% ankle flex
[FIT2_joint_torque_anklex, adj_joint_torque_anklex, ~, r_squared_20] = FIT2(joint_torque_anklex);
% hip abduction
[FIT2_joint_torque_hipy, adj_joint_torque_hipy, ~, r_squared_21] = FIT2(joint_torque_hipy);
% ankle inversion
[FIT2_joint_torque_ankley, adj_joint_torque_ankley, ~, r_squared_22] = FIT2(joint_torque_ankley);
% joint torque at impact peak
for cc=1:k
   impact_hip_moment_x(cc) = hip_moment_x(time_to_impact(cc),cc);
   impact_knee_moment_x(cc) = knee_moment_x(time_to_impact(cc),cc);
   impact_ankle_moment_x(cc) = ankle_moment_x(time_to_impact(cc),cc);
   impact_hip_moment_y(cc) = hip_moment_y(time_to_impact(cc),cc);
end
impact_ankle_moment_y(cc) = ankle_moment_y(time_to_impact(cc),cc);
end

jumps=length(hip_moment_x);
if jumps>10;
    [ impact_joint_torque_hipx,JThx_std_impact ] = SEGMENTER(impact_hip_moment_x);
end

jumps=length(knee_moment_x);
if jumps>10;
    [ impact_joint_torque_kneex,JTkx_std_impact ] = SEGMENTER(impact_knee_moment_x);
end

jumps=length(ankle_moment_x);
if jumps>10;
    [ impact_joint_torque_anklex,JTax_std_impact ] = SEGMENTER(impact_ankle_moment_x);
end

jumps=length(hip_moment_y);
if jumps>10;
    [ impact_joint_torque_hipy,JThy_std_impact ] = SEGMENTER(impact_hip_moment_y);
end

jumps=length(ankle_moment_y);
if jumps>10;
    [ impact_joint_torque_ankley,JTay_std_impact ] = SEGMENTER(impact_ankle_moment_y);
end

% hip flex
 [~,adj_impact_joint_torque_hipx,~,r_squared_18_impact] = FIT2(impact_joint_torque_hipx);
% knee flex
 [~,adj_impact_joint_torque_kneex,~,r_squared_19_impact] = FIT2(impact_joint_torque_kneex);
% ankle flex
 [~,adj_impact_joint_torque_anklex,~,r_squared_20_impact] = FIT2(impact_joint_torque_anklex);
% hip abduction
 [~,adj_impact_joint_torque_hipy,~,r_squared_21_impact] = FIT2(impact_joint_torque_hipy);
% ankle inversion
 [~,adj_impact_joint_torque_ankley,~,r_squared_22_impact] = FIT2(impact_joint_torque_ankley);

% joint torque at active peak
for cc=1:k
    active_hip_moment_x(cc) = hip_moment_x(time_to_max_GRFv(cc),cc);
    active_knee_moment_x(cc) = knee_moment_x(time_to_max_GRFv(cc),cc);
    active_ankle_moment_x(cc) = ankle_moment_x(time_to_max_GRFv(cc),cc);
    active_hip_moment_y(cc) = hip_moment_y(time_to_max_GRFv(cc),cc);
end
active_ankle_moment_y(cc) = ankle_moment_y(time_to_max_GRFv(cc),cc);
end

jumps=length(hip_moment_x);
if jumps>10;
    [ active_joint_torque_hipx,JThx_std_active ] = SEGMENTER(active_hip_moment_x);
end

jumps=length(knee_moment_x);
if jumps>10;
    [ active_joint_torque_kneex,JTkx_std_active ] = SEGMENTER(active_knee_moment_x);
end

jumps=length(ankle_moment_x);
if jumps>10;
    [ active_joint_torque_anklex,JTax_std_active ] = SEGMENTER(active_ankle_moment_x);
end

jumps=length(hip_moment_y);
if jumps>10;
    [ active_joint_torque_hipy,JThy_std_active ] = SEGMENTER(active_hip_moment_y);
end

jumps=length(ankle_moment_y);
if jumps>10;
    [ active_joint_torque_ankley,JTay_std_active ] = SEGMENTER(active_ankle_moment_y);
end

% hip flex
 [~,adj_active_joint_torque_hipx,~,r_squared_18_active] = FIT2(active_joint_torque_hipx);
% knee flex
 [~,adj_active_joint_torque_kneex,~,r_squared_19_active] = FIT2(active_joint_torque_kneex);
% ankle flex
 [~,adj_active_joint_torque_anklex,~,r_squared_20_active] = FIT2(active_joint_torque_anklex);
% hip abduction
 [~,adj_active_joint_torque_hipy,~,r_squared_21_active] = FIT2(active_joint_torque_hipy);
% ankle inversion
 [~,adj_active_joint_torque_ankley,~,r_squared_22_active] = FIT2(active_joint_torque_ankley);

%% First order fit for comparison
% finds slope of the first three segments and the last three segments to
% get the trend that the second order fit finds but first order for easier
% comparison. No additional outlier removal.
% have not added the p2p and the frontal plane stuff here yet.
[~,fatigued_slope_12,fatigued_r_squared_12] = FIT_WO_outlierremoval(adj_joint_impulse_kneex(length(adj_joint_impulse_kneex)-2:length(adj_joint_impulse_kneex)),1);

[~,unfatigued_slope_13,unfatigued_r_squared_13] = FIT_WO_outlierremoval(adj_joint_impulse_anklex(1:3),1);
[~,fatigued_slope_13,fatigued_r_squared_13] = FIT_WO_outlierremoval(adj_joint_impulse_anklex(length(adj_joint_impulse_anklex)-2:length(adj_joint_impulse_anklex)),1);

[~,unfatigued_slope_14,unfatigued_r_squared_14] = FIT_WO_outlierremoval(adj_joint_work_hipx(1:3),1);
[~,fatigued_slope_14,fatigued_r_squared_14] = FIT_WO_outlierremoval(adj_joint_work_hipx(length(adj_joint_work_hipx)-2:length(adj_joint_work_hipx)),1);

[~,unfatigued_slope_16,unfatigued_r_squared_16] = FIT_WO_outlierremoval(adj_joint_work_kneex(1:3),1);
[~,fatigued_slope_16,fatigued_r_squared_16] = FIT_WO_outlierremoval(adj_joint_work_kneex(length(adj_joint_work_kneex)-2:length(adj_joint_work_kneex)),1);

[~,unfatigued_slope_17,unfatigued_r_squared_17] = FIT_WO_outlierremoval(adj_joint_work_anklex(1:3),1);
[~,fatigued_slope_17,fatigued_r_squared_17] = FIT_WO_outlierremoval(adj_joint_work_anklex(length(adj_joint_work_anklex)-2:length(adj_joint_work_anklex)),1);

unfatigued_r_squared=[unfatigued_r_squared_1 unfatigued_r_squared_2 unfatigued_r_squared_3 unfatigued_r_squared_4 unfatigued_r_squared_5 unfatigued_r_squared_6 unfatigued_r_squared_8 unfatigued_r_squared_9 unfatigued_r_squared_10 unfatigued_r_squared_12 unfatigued_r_squared_13 unfatigued_r_squared_14 unfatigued_r_squared_16 unfatigued_r_squared_17];
fatigued_r_squared=[fatigued_r_squared_1 fatigued_r_squared_2 fatigued_r_squared_3 fatigued_r_squared_4 fatigued_r_squared_5 fatigued_r_squared_6 fatigued_r_squared_8 fatigued_r_squared_9 fatigued_r_squared_10 fatigued_r_squared_12 fatigued_r_squared_13 fatigued_r_squared_14 fatigued_r_squared_16 fatigued_r_squared_17];

% exports an array that can be compiled with other subjects' results for comparison.

% results will a compiled matrix in the following format:
% number of cycles,fatigued,slope of line of best FIT for first few unfatigued values, slope for last few fatigued values, R^2
% impact peak
% active peak
% time to max
% loading rate
% GRFv impulse
% hip flex max
% knee flex max

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% ankle flex max
% hip abduction max
% ankle inversion max
% p2p hip
% p2p knee
% p2p ankle
% p2p hip abduction
% p2p ankle inversion
% hip flexion impulse
% knee flexion impulse
% ankle flexion impulse
% hip abduction impulse
% ankle inversion impulse
% hip negative work
% knee negative work
% ankle negative work
% hip abduction work
% ankle inversion work

exp_impact_peak=adj_impact_peak/BWn; %normalized for body weight
exp_max_GRFv=adj_max_GRFv/BWn; %normalized for body weight
exp_time_to_max_GRFv=adj_time_to_max_GRFv; %left in msec
exp_max_GRFv_load_rate=adj_max_GRFv_load_rate/1000; %kilonewtons per second
exp_GRFv_impulse=adj_GRFv_Impulse/BWn/1000; %normalized for body weight.
units BW's
exp_max_hip_flexion=adj_max_hip_flexion; %in degrees
exp_max_knee_flexion=adj_max_knee_flexion; %in degrees
exp_max_ankle_flexion=adj_max_ankle_flexion; %in degrees
exp_hip_abduction_max=adj_max_hip_abduction; %in degrees
exp_ankle_inversion_max=adj_max_ankle_inversion; %in degrees
exp_p2p_hipflex=adj_p2p_hipflex; %in degrees
exp_p2p_kneeflex=adj_p2p_kneeflex; %in degrees
exp_p2p_ankleflex=adj_p2p_ankleflex; %in degrees
exp_p2p_hipabd=adj_p2p_hipabd; %in degrees
exp_p2p_ankleinv=adj_p2p_ankleinv; %in degrees
exp_joint_impulse_hipx=adj_joint_impulse_hipx; %units are (Nm*sec)/(kg*m)
%normalized for body weight and height
exp_joint_impulse_kneex=adj_joint_impulse_kneex; %units are (Nm*sec)/(kg*m)
%normalized for body weight and height
exp_joint_impulse_anklex=adj_joint_impulse_anklex; %units are (Nm*sec)/(kg*m)
%normalized for body weight and height
exp_joint_impulse_hipy=adj_joint_impulse_hipy; %units are (Nm*sec)/(kg*m)
%normalized for body weight and height
exp_joint_impulse_ankley=adj_joint_impulse_ankley; %units are (Nm*sec)/(kg*m)
%normalized for body weight and height
exp_joint_work_hipx=adj_joint_work_hipx; %units are W/(kg*m) %normalized for body weight and height
exp_joint_work_kneex=adj_joint_work_kneex; %units are W/(kg*m) %normalized for body weight and height
exp_joint_work_anklex=adj_joint_work_anklex; %units are W/(kg*m) %normalized for body weight and height
exp_joint_work_hipy=adj_joint_work_hipy; %units are W/(kg*m) %normalized for body weight and height
exp_joint_work_ankley=adj_joint_work_ankley; %units are W/(kg*m) %normalized for body weight and height
exp_joint_torque_hipx=adj_joint_torque_hipx; %units are (Nm/(kg*m)) %normalized for body weight and height
if length(exp_joint_work_hipx)<10
    aa=zeros(1,10);
    exp_impact_peak(n numel(aa))=0;
    exp_max_GRFv(n numel(aa))=0;
    exp_max_to_max_GRFv(n numel(aa))=0;
    exp_max_GRFv_load_rate(n numel(aa))=0;
    exp_GRFv_impulse(n numel(aa))=0;
    exp_max_hip_flexion(n numel(aa))=0;
    exp_max_knee_flexion(n numel(aa))=0;
    exp_max_ankle_flexion(n numel(aa))=0;
    exp_p2p_hipflex(n numel(aa))=0;
    exp_p2p_kneeflex(n numel(aa))=0;
    exp_p2p_ankleflex(n numel(aa))=0;
    exp_joint_impulse_hipx(n numel(aa))=0;
    exp_joint_impulse_kneex(n numel(aa))=0;
    exp_joint_impulse_anklex(n numel(aa))=0;
    exp_joint_work_hipx(n numel(aa))=0;
    exp_joint_work_kneex(n numel(aa))=0;
    exp_joint_work_anklex(n numel(aa))=0;
    exp_hip_abduction_max(n numel(aa))=0;
    exp_ankle_inversion_max(n numel(aa))=0;
    exp_p2p_hipabd(n numel(aa))=0;
    exp_p2p_ankleinv(n numel(aa))=0;
    exp_joint_impulse_hipy(n numel(aa))=0;
    exp_joint_impulse_ankley(n numel(aa))=0;
exp_joint_work_hipy(numel(aa))=0;
ex_p_joint_work_ankley(numel(aa))=0;
ex_p_joint_torque_hipx(numel(aa))=0;
ex_p_joint_torque_kneex(numel(aa))=0;
ex_p_joint_torque_anklex(numel(aa))=0;
ex_p_joint_torque_hipy(numel(aa))=0;
ex_p_joint_torque_ankley(numel(aa))=0;
ex_impact_joint_torque_hipx(numel(aa))=0;
ex_impact_joint_torque_kneex(numel(aa))=0;
ex_impact_joint_torque_anklex(numel(aa))=0;
ex_impact_joint_torque_hipy(numel(aa))=0;
ex_impact_joint_torque_ankley(numel(aa))=0;
ex_active_joint_torque_hipx(numel(aa))=0;
ex_active_joint_torque_kneex(numel(aa))=0;
ex_active_joint_torque_anklex(numel(aa))=0;
ex_active_joint_torque_hipy(numel(aa))=0;
ex_active_joint_torque_ankley(numel(aa))=0;
end

results_10=[cycles exp_impact_peak_r_squared_1;... cycles exp_max_GRFv_r_squared_2;... cycles exp_time_to_max_GRFv_r_squared_3;... cycles exp_max_GRFv_load_rate_r_squared_4;... cycles exp_GRFv_impulse_r_squared_5;... cycles exp_max_hip_flexion_r_squared_6;... cycles exp_max_knee_flexion_r_squared_8;... cycles exp_max_ankle_flexion_r_squared_9;... cycles exp_hip_abduction_max_r_squared_7;... cycles exp_ankle_inversion_max_r_squared_9_inversion;... cycles exp_p2p_hipflex_r_squared_6_2;... cycles exp_p2p_kneeflex_r_squared_8_2;... cycles exp_p2p_ankleflex_r_squared_9_2;... cycles exp_p2p_hipab_r_squared_6_2abd;... cycles exp_p2p_ankleinv_r_squared_9_2inv;... cycles exp_joint_impulse_hipx_r_squared_10;... cycles exp_joint_impulse_kneex_r_squared_12;... cycles exp_joint_impulse_anklex_r_squared_13;... cycles exp_joint_impulse_hipy_r_squared_11;... cycles exp_joint_impulse_ankley_r_squared_13y;... cycles exp_joint_work_hipx_r_squared_14;... cycles exp_joint_work_kneex_r_squared_16;... cycles exp_joint_work_anklex_r_squared_17;... cycles exp_joint_work_hipy_r_squared_15;... cycles exp_joint_work_ankley_r_squared_17y;... cycles exp_joint_torque_hipx_r_squared_18;... cycles exp_joint_torque_kneex_r_squared_19;... cycles exp_joint_torque_anklex_r_squared_20;... cycles exp_joint_torque_hipy_r_squared_21;... cycles exp_joint_torque_ankley_r_squared_22;... cycles exp_impact_joint_torque_hipx_r_squared_18_impact;... cycles exp_impact_joint_torque_kneex_r_squared_19_impact;... cycles exp_impact_joint_torque_anklex_r_squared_20_impact;... cycles exp_impact_joint_torque_hipy_r_squared_21_impact;... cycles exp_impact_joint_torque_ankley_r_squared_22_impact;... cycles exp_active_joint_torque_hipx_r_squared_18_active;... cycles exp_active_joint_torque_kneex_r_squared_19_active;... cycles exp_active_joint_torque_anklex_r_squared_20_active;...
cycles exp_active_joint_torque_hipy r_squared_21_active;...
cycles exp_active_joint_torque_ankley r_squared_22_active;

%% pdf generator
psname = strcat(settitle, '.ps');
print ('-dpsc2', psname, '-append', '-f1')
print ('-dpsc2', psname, '-append', '-f2')
print ('-dpsc2', psname, '-append', '-f3')
print ('-dpsc2', psname, '-append', '-f4')
print ('-dpsc2', psname, '-append', '-f5')
print ('-dpsc2', psname, '-append', '-f6')
print ('-dpsc2', psname, '-append', '-f7')
print ('-dpsc2', psname, '-append', '-f8')
print ('-dpsc2', psname, '-append', '-f9')
pdfname= strcat(settitle,'.pdf');
ps2pdf('psfile', psname, 'pdffile', pdfname, 'gspapersize', 'a4',
'deletepsfile', 1)
close all

%%
varname=settitle(1:3);
day=input('Test day 1 or 2? ','s');
filename=strcat('results_generator_day',day);
S.(varname) = results_10;
save(filename, '-struct', 'S','-append');

%% end code
function [ newmetric,newmetric_std ] = SEGMENTER( metric)
%This program breaks the data into 10% intervals.
% The program finds the number of landings that need to be in each trial
% by dividing the total number of landings by 10 and then then rounding
% down to the nearest integer. It then finds the remainder and
% distributes those landings throughout the middle of FLA. The mean and
% standard deviation for each 10% interval is found.
% For example, if a person jumps 25 times, their 10% intervals will be two
% jumps per interval with a remainder of 5. The first two intervals and the
% last three intervals will be
% the average of two jumps, but the middle will be the average of three.
% Ex: newmetric=[mean(1:2) mean(3:4) mean(5:7) mean(8:10) mean(11:13)...
% mean(14:16) mean(17:19) mean(20:21) mean(22:23) mean(24:25)];

jumps=length(metric);
group=floor(jumps/10);
n=rem(jumps,10);

switch n
    case 0
        newmetric=[mean(metric(1:group)) mean(metric(1+group:2*group))...
            mean(metric(1+2*group:3*group))
            mean(metric(3*group+1:4*group))...
            mean(metric(4*group+1:5*group))
            mean(metric(5*group+1:6*group))...
            mean(metric(6*group+1:7*group))
            mean(metric(7*group+1:8*group))...
            mean(metric(8*group+1:9*group))
            mean(metric(9*group+1:10*group))];
    case 1
        newmetric=[mean(metric(1:group)) mean(metric(1+group:2*group))...
            mean(metric(1+2*group:3*group))
            mean(metric(3*group+1:4*group))...
            mean(metric(4*group+1:5*group+1))
            mean(metric(5*group+2:6*group+1))...
            mean(metric(6*group+2:7*group+1))
            mean(metric(7*group+2:8*group+1))...
            mean(metric(8*group+2:9*group+1))
            mean(metric(9*group+2:10*group+1))];
    case 2
        newmetric=[mean(metric(1:group)) mean(metric(1+group:2*group))...
            mean(metric(1+2*group:3*group))
            mean(metric(3*group+1:4*group))...
            mean(metric(4*group+1:5*group+1))
            mean(metric(5*group+2:6*group+2))...
            mean(metric(6*group+3:7*group+2))
            mean(metric(7*group+3:8*group+2))...
            mean(metric(8*group+3:9*group+2))
            mean(metric(9*group+3:10*group+2))];
    case 3
        newmetric=[mean(metric(1:group)) mean(metric(1+group:2*group))...
            mean(metric(1+2*group:3*group))
            mean(metric(3*group+1:4*group+1))...
            mean(metric(4*group+2:5*group+2))
            mean(metric(5*group+3:6*group+3))...
            mean(metric(6*group+4:7*group+3))
            mean(metric(7*group+4:8*group+3))...
\[
\text{mean}(\text{metric}(8*\text{group}+4:9*\text{group}+3))
\]
\[
\text{mean}(\text{metric}(9*\text{group}+4:10*\text{group}+3))
\]
\[
\text{case 4}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1:\text{group})) \ \text{mean}(\text{metric}(1+\text{group}:2*\text{group})) \ldots \\
\text{mean}(\text{metric}(1+2*\text{group}:3*\text{group}+1))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+1:4*\text{group}+1))
\]
\[
\text{mean}(\text{metric}(4*\text{group}+2:5*\text{group}+2))
\]
\[
\text{mean}(\text{metric}(5*\text{group}+3:6*\text{group}+3))
\]
\[
\text{mean}(\text{metric}(6*\text{group}+4:7*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(7*\text{group}+5:8*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(8*\text{group}+5:9*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(9*\text{group}+5:10*\text{group}+4))
\]
\[
\text{case 5}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1:\text{group})) \ \text{mean}(\text{metric}(1+\text{group}:2*\text{group})) \ldots \\
\text{mean}(\text{metric}(1+2*\text{group}:3*\text{group}+1))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+2:4*\text{group}+2))
\]
\[
\text{mean}(\text{metric}(4*\text{group}+3:5*\text{group}+3))
\]
\[
\text{mean}(\text{metric}(5*\text{group}+4:6*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(6*\text{group}+5:7*\text{group}+5))
\]
\[
\text{mean}(\text{metric}(7*\text{group}+6:8*\text{group}+5))
\]
\[
\text{mean}(\text{metric}(8*\text{group}+6:9*\text{group}+5))
\]
\[
\text{mean}(\text{metric}(9*\text{group}+6:10*\text{group}+5))
\]
\[
\text{case 6}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1:\text{group})) \ \text{mean}(\text{metric}(1+\text{group}:2*\text{group})) \ldots \\
\text{mean}(\text{metric}(1+2*\text{group}:3*\text{group}+1))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+3:4*\text{group}+3))
\]
\[
\text{mean}(\text{metric}(4*\text{group}+4:5*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(5*\text{group}+5:6*\text{group}+5))
\]
\[
\text{mean}(\text{metric}(6*\text{group}+6:7*\text{group}+6))
\]
\[
\text{mean}(\text{metric}(7*\text{group}+7:8*\text{group}+7))
\]
\[
\text{mean}(\text{metric}(8*\text{group}+8:9*\text{group}+7))
\]
\[
\text{mean}(\text{metric}(9*\text{group}+8:10*\text{group}+7))
\]
\[
\text{case 7}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1:\text{group})) \ \text{mean}(\text{metric}(1+\text{group}:2*\text{group})) \ldots \\
\text{mean}(\text{metric}(2+2*\text{group}:3*\text{group}+2))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+4:4*\text{group}+4))
\]
\[
\text{mean}(\text{metric}(3*\text{group}+5:4*\text{group}+5))
\]
\[
\text{mean}(\text{metric}(5*\text{group}+6:6*\text{group}+6))
\]
\[
\text{mean}(\text{metric}(6*\text{group}+7:8*\text{group}+8))
\]
\[
\text{mean}(\text{metric}(8*\text{group}+9:10*\text{group}+8))
\]
\[
\text{case 8}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1:\text{group})) \ \text{mean}(\text{metric}(1+\text{group}:2*\text{group})) \ldots \\
\text{mean}(\text{metric}(2+2*\text{group}:3*\text{group}+2))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+6:4*\text{group}+6))
\]
\[
\text{mean}(\text{metric}(4*\text{group}+7:4*\text{group}+7))
\]
\[
\text{mean}(\text{metric}(5*\text{group}+8:6*\text{group}+8))
\]
\[
\text{mean}(\text{metric}(6*\text{group}+9:8*\text{group}+9))
\]
\[
\text{mean}(\text{metric}(9*\text{group}+10:10*\text{group}+10))
\]
\[
\text{case 9}
\]
\[
\text{newmetric}=[\text{mean}(\text{metric}(1+\text{group})) \ \text{mean}(\text{metric}(2+\text{group}:2*\text{group}+2)) \ldots \\
\text{mean}(\text{metric}(3+2*\text{group}:3*\text{group}+3))]
\]
\[
\text{mean}(\text{metric}(3*\text{group}+8:4*\text{group}+8))
\]
mean(metric(4*group+5:5*group+5))
mean(metric(5*group+6:6*group+6))...
mean(metric(6*group+7:7*group+7))
mean(metric(7*group+8:8*group+8))...
mean(metric(8*group+9:9*group+9))
mean(metric(9*group+10:10*group+9));
otherwise
disp('error in switch')
end

switch n
case 0
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))... std(metric(1+2*group:3*group)) std(metric(3*group+1:4*group))...
                std(metric(4*group+1:5*group)) std(metric(5*group+1:6*group))...
                std(metric(6*group+1:7*group)) std(metric(7*group+1:8*group))...
                std(metric(8*group+1:9*group)) std(metric(9*group+1:10*group))];
case 1
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))... std(metric(1+2*group:3*group)) std(metric(3*group+1:4*group))...
                std(metric(4*group+1:5*group+1))...
                std(metric(5*group+2:6*group+1))...
                std(metric(6*group+2:7*group+1))...
                std(metric(7*group+2:8*group+1))...
                std(metric(8*group+2:9*group+1))]
    case 2
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))... std(metric(1+2*group:3*group)) std(metric(3*group+1:4*group))...
                std(metric(4*group+1:5*group+1))...
                std(metric(5*group+2:6*group+2))...
                std(metric(6*group+3:7*group+2))...
                std(metric(7*group+3:8*group+2))...
                std(metric(8*group+3:9*group+2))]
    case 3
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))... std(metric(1+2*group:3*group)) std(metric(3*group+1:4*group))...
                std(metric(4*group+1:5*group+1))...
                std(metric(5*group+2:6*group+3))...
                std(metric(6*group+3:7*group+3))...
                std(metric(7*group+4:8*group+3))...
                std(metric(8*group+4:9*group+3))]
    case 4
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))... std(metric(1+2*group:3*group)) std(metric(3*group+1:4*group))...
                std(metric(4*group+1:5*group+1))...
                std(metric(5*group+2:6*group+4))...
                std(metric(6*group+4:7*group+4))...
                std(metric(7*group+5:8*group+4))...
                std(metric(8*group+5:9*group+4))]
    case 5
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))...
std(metric(1+2*group:3*group+1))
std(metric(3*group+2:4*group+2))
... 
std(metric(4*group+3:5*group+3))
std(metric(5*group+4:6*group+4))
... 
std(metric(6*group+5:7*group+5))
std(metric(7*group+6:8*group+5))
... 
std(metric(8*group+6:9*group+5))
std(metric(9*group+6:10*group+5)));
case 6
    newmetric_std=[std(metric(1:group)) std(metric(1+group:2*group))
    ...
    std(metric(1+2*group:3*group+1))
std(metric(3*group+2:4*group+2))
... 
std(metric(4*group+3:5*group+3))
std(metric(5*group+4:6*group+4))
... 
std(metric(6*group+5:7*group+5))
std(metric(7*group+6:8*group+6))
... 
std(metric(8*group+6:9*group+5))
std(metric(9*group+6:10*group+5))]
    newmetric_std=[std(metric(1:group+1)) std(metric(2+group:2*group+2))
    ...
    std(metric(2+2*group:3*group+2))
std(metric(3*group+3:4*group+3))
... 
std(metric(4*group+4:5*group+4))
std(metric(5*group+5:6*group+5))
... 
std(metric(6*group+6:7*group+5))
std(metric(7*group+7:8*group+6))
... 
std(metric(8*group+6:9*group+5))
std(metric(9*group+6:10*group+5))]
    newmetric_std=[std(metric(1:group+1)) std(metric(2+group:2*group+2))
    ...
    std(metric(2+2*group:3*group+2))
std(metric(3*group+3:4*group+3))
... 
std(metric(4*group+4:5*group+4))
std(metric(5*group+5:6*group+5))
... 
std(metric(6*group+6:7*group+5))
std(metric(7*group+7:8*group+6))
... 
std(metric(8*group+6:9*group+5))
std(metric(9*group+6:10*group+5))]
    newmetric_std=[std(metric(1:group+1)) std(metric(2+group:2*group+2))
    ...
    std(metric(2+2*group:3*group+2))
std(metric(3*group+3:4*group+3))
... 
std(metric(4*group+4:5*group+4))
std(metric(5*group+5:6*group+5))
... 
std(metric(6*group+6:7*group+5))
std(metric(7*group+7:8*group+6))
... 
std(metric(8*group+6:9*group+5))
std(metric(9*group+6:10*group+5))]
    otherwise
        disp('error in switch')
end
end

%% end code
function [new_dat,adj_metric,slope,r_squared] = FIT2(metric)
%FIT2 adjusts data by removing outliers and calculates line of best fit with
%R^2 value
% same as FIT but does a second order
% find the line of best fit using MATLAB functions polyfit and polyval
cycle=1:length(metric);
p = polyfit(cycle,metric,2);
dat=polyval(p,cycle);

%find the residuals (difference between the actual data and the fit
for jj=cycle
    residuals(jj)=abs(dat(jj)-metric(jj));
end

%find the mean and standard deviation of the residuals
mean_res=mean(residuals);
std_res=std(residuals);
for kk=1:1:length(metric)
    if residuals(kk) > (2*std_res)
        adj_metric(kk)=dat(kk);
    else
        adj_metric(kk)=metric(kk);
    end
end

%find new line of best fit with adjusted data and calculate R^2
new_p = polyfit(cycle,adj_metric,2);
new_dat=polyval(new_p,cycle);

n=length(cycle);
x=cycle;
y=adj_metric;
\[ r = \frac{n\sum(x \cdot y) - (\sum(x) \cdot \sum(y))}{\sqrt{n\sum(x^2) - (\sum(x))^2} \cdot \sqrt{n\sum(y^2) - (\sum(y))^2}} \]
\[ r^2 \]
\[ y_{resid} = y - new\_dat \]
\[ SS_{resid} = \sum(y_{resid}^2) \]
\[ SS_{tot} = (n-1) \cdot \text{var}(y) \]
\[ r^2 = 1 - \frac{SS_{resid}}{SS_{tot}} \]
slope=new_p(1);
end

%% end code
% This code quantifies what slip might have occurred in the sensors

load slip_data

% from the raw_slipdata.mat file, column references are as follows:

% line 1 :   Frame #
% line 2 :   Sensor #10 / Position / X / Time
% line 3 :   Sensor #10 / Position / Y / Time
% line 4 :   Sensor #10 / Position / Z / Time
% line 5 :   Sensor #10 / Euler Angles (Z Y'X") / Z / Time
% line 6 :   Sensor #10 / Euler Angles (Z Y'X") / Y' / Time
% line 7 :   Sensor #10 / Euler Angles (Z Y'X") / X" / Time
% line 8 :   Sensor #2 / Position / X / Time
% line 9 :   Sensor #2 / Position / Y / Time
% line 10 :  Sensor #2 / Position / Z / Time
% line 11 :  Sensor #2 / Euler Angles (Z Y'X") / Z / Time
% line 12 :  Sensor #2 / Euler Angles (Z Y'X") / Y' / Time
% line 13 :  Sensor #2 / Euler Angles (Z Y'X") / X" / Time
% line 14 :  Sensor #3 / Position / X / Time
% line 15 :  Sensor #3 / Position / Y / Time
% line 16 :  Sensor #3 / Position / Z / Time
% line 17 :  Sensor #3 / Euler Angles (Z Y'X") / Z / Time
% line 18 :  Sensor #3 / Euler Angles (Z Y'X") / Y' / Time
% line 19 :  Sensor #3 / Euler Angles (Z Y'X") / X" / Time
% line 20 :  Sensor #4 / Position / X / Time
% line 21 :  Sensor #4 / Position / Y / Time
% line 22 :  Sensor #4 / Position / Z / Time
% line 23 :  Sensor #4 / Euler Angles (Z Y'X") / Z / Time
% line 24 :  Sensor #4 / Euler Angles (Z Y'X") / Y' / Time
% line 25 :  Sensor #4 / Euler Angles (Z Y'X") / X" / Time
% line 26 :  Forceplate #0 / World Ref. Frame / Force / X / Time
% line 27 :  Forceplate #0 / World Ref. Frame / Force / Y / Time
% line 28 :  Forceplate #0 / World Ref. Frame / Force / Z / Time
% line 29 :  Forceplate #0 / World Ref. Frame / Center of Pressure / X / Time
% line 30 :  Forceplate #0 / World Ref. Frame / Center of Pressure / Y / Time
% line 31 :  Forceplate #0 / World Ref. Frame / Center of Pressure / Z / Time

%sensor 10 is sacrum
%sensor 4 is thigh
%sensor 3 is shank
%sensor 2 is foot

%% part one: check stance
% examines force plate characteristics to determine if similar stance was
% obtained pre and post

for jj=1:18;
    ii=jj*2;
    post_x=mean(dat(:,26,ii-1));
    pre_x=mean(dat(:,26,ii));
    post_y=mean(dat(:,27,ii-1));
pre_y=mean(dat(:,27,ii));
post_z=mean(dat(:,28,ii-1));
pre_z=mean(dat(:,28,ii));

post_forcemag=sqrt((post_x)^2+(post_y)^2+(post_y)^2);
pre_forcemag=sqrt((pre_x)^2+(pre_y)^2+(pre_y)^2);

percentdiff_forcemag(jj)=(post_forcemag-pre_forcemag)/abs(pre_forcemag)*100;
if abs(percentdiff_forcemag(jj))>30
    disp(strcat('forceplate dissimilarity at pair: ',num2str(jj)))
end

end

%% part 2: relative difference between sensors
% examines distances between sensors to determine if slippage occurred

for jj=1:18;
    ii=jj*2;
    sen10_post_x=mean(dat(:,2,ii-1));
    sen10_pre_x=mean(dat(:,2,ii));
    sen10_post_y=mean(dat(:,3,ii-1));
    sen10_pre_y=mean(dat(:,3,ii));
    sen10_post_z=mean(dat(:,4,ii-1));
    sen10_pre_z=mean(dat(:,4,ii));
    sen10=[sen10_pre_x,sen10_pre_y,sen10_pre_z;
           sen10_post_x,sen10_post_y,sen10_post_z];

    sen4_post_x=mean(dat(:,20,ii-1));
    sen4_pre_x=mean(dat(:,20,ii));
    sen4_post_y=mean(dat(:,21,ii-1));
    sen4_pre_y=mean(dat(:,21,ii));
    sen4_post_z=mean(dat(:,22,ii-1));
    sen4_pre_z=mean(dat(:,22,ii));
    sen4=[sen4_pre_x,sen4_pre_y,sen4_pre_z;
           sen4_post_x,sen4_post_y,sen4_post_z];

    sen3_post_x=mean(dat(:,14,ii-1));
    sen3_pre_x=mean(dat(:,14,ii));
    sen3_post_y=mean(dat(:,15,ii-1));
    sen3_pre_y=mean(dat(:,15,ii));
    sen3_post_z=mean(dat(:,16,ii-1));
    sen3_pre_z=mean(dat(:,16,ii));
    sen3=[sen3_pre_x,sen3_pre_y,sen3_pre_z;
           sen3_post_x,sen3_post_y,sen3_post_z];

    sen2_post_x=mean(dat(:,8,ii-1));
    sen2_pre_x=mean(dat(:,8,ii));
    sen2_post_y=mean(dat(:,9,ii-1));
    sen2_pre_y=mean(dat(:,9,ii));
    sen2_post_z=mean(dat(:,10,ii-1));
    sen2_pre_z=mean(dat(:,10,ii));
    sen2=[sen2_pre_x,sen2_pre_y,sen2_pre_z;
           sen2_post_x,sen2_post_y,sen2_post_z];
end
sen2_pre_y = mean(dat(:,9,ii));
sen2_post_z = mean(dat(:,10,ii-1));
sen2_pre_z = mean(dat(:,10,ii));

sen2 = [sen2_pre_x, sen2_pre_y, sen2_pre_z;  
        sen2_post_x, sen2_post_y, sen2_post_z];

% sacrum to thigh differences
[diff_pre_post, percentchange] = relative_difference(sen10, sen4);
sacrum_thigh(jj,:) = [diff_pre_post, percentchange];

% sacrum to shank differences
[diff_pre_post, percentchange] = relative_difference(sen10, sen3);
sacrum_shank(jj,:) = [diff_pre_post, percentchange];

% sacrum to foot differences
[diff_pre_post, percentchange] = relative_difference(sen10, sen2);
sacrum_foot(jj,:) = [diff_pre_post, percentchange];

% thigh to shank differences
[diff_pre_post, percentchange] = relative_difference(sen4, sen3);
thigh_shank(jj,:) = [diff_pre_post, percentchange];

% thigh to foot differences
[diff_pre_post, percentchange] = relative_difference(sen4, sen2);
thigh_foot(jj,:) = [diff_pre_post, percentchange];

% shank to foot differences
[diff_pre_post, percentchange] = relative_difference(sen3, sen2);
shank_foot(jj,:) = [diff_pre_post, percentchange];

end

%% part 3: distance to ground
% examines the distance of each sensor from the ground

for jj = 1:18;
    ii = jj*2;
    sen10_post_z = abs(mean(dat(:,4,ii-1)));  
    sen10_pre_z = abs(mean(dat(:,4,ii)));  
    percentdiff_z_10_floor(jj) = (sen10_pre_z - sen10_post_z)/abs(sen10_post_z)*100;
    if abs(percentdiff_z_10_floor(jj)) > 10
        disp(strcat('dissimilarity in z 10-floor at pair: ', num2str(jj)))
    end

    sen4_post_z = abs(mean(dat(:,22,ii-1)));  
    sen4_pre_z = abs(mean(dat(:,22,ii)));  
    percentdiff_z_4_floor(jj) = (sen4_pre_z - sen4_post_z)/abs(sen4_post_z)*100;
    if abs(percentdiff_z_4_floor(jj)) > 10
        disp(strcat('dissimilarity in z 4-floor at pair: ', num2str(jj)))
    end

    sen3_post_z = abs(mean(dat(:,16,ii-1)));
sen3_pre_z=abs(mean(dat(:,16,ii)));% Compute the absolute mean of the 16th sensor pre- and post-values.
percentdiff_z_3_floor(jj)=(sen3_pre_z-sen3_post_z)/abs(sen3_post_z)*100;% Compute the percentage difference for the 3rd floor.
if abs(percentdiff_z_3_floor(jj))>10
    disp(strcat('dissimilarity in z 3-floor at pair: ',num2str(jj)))% Display if the dissimilarity is above 10%.
end% End of if statement.

sen2_post_z=abs(mean(dat(:,10,ii-1)));% Compute the absolute mean of the 10th sensor post-values.
sen2_pre_z=abs(mean(dat(:,10,ii)));% Compute the absolute mean of the 10th sensor pre-values.
percentdiff_z_2_floor(jj)=(sen2_pre_z-sen2_post_z)/abs(sen2_post_z)*100;% Compute the percentage difference for the 2nd floor.
if abs(percentdiff_z_2_floor(jj))>10
    disp(strcat('dissimilarity in z 2-floor at pair: ',num2str(jj)))% Display if the dissimilarity is above 10%.
end% End of if statement.

absdiff_z_sacrum(jj)=sen10_post_z-sen10_pre_z;% Compute the absolute difference for the sacrum.
absdiff_z_thigh(jj)=sen4_post_z-sen4_pre_z;% Compute the absolute difference for the thigh.
absdiff_z_shank(jj)=sen3_post_z-sen3_pre_z;% Compute the absolute difference for the shank.
absdiff_z_foot(jj)=sen2_post_z-sen2_pre_z;% Compute the absolute difference for the foot.
end% End of the function.

avgdiff_z_sacrum=mean(percentdiff_z_10_floor);% Compute the average percentage difference for the sacrum.
avgdiff_z_thigh=mean(percentdiff_z_4_floor);% Compute the average percentage difference for the thigh.
avgdiff_z_shank=mean(percentdiff_z_3_floor);% Compute the average percentage difference for the shank.
avgdiff_z_ankle=mean(percentdiff_z_2_floor);% Compute the average percentage difference for the ankle.

%% end code
function [ diff_pre_post,percentchange ] = relative_difference( prox,dist)
%two sensors are inputed and the program calculates the length of the vector
%between between the two sensors in their pre and post locations.
%the distance vectors are compared and an actual length and percent
%difference is given.
pré_x=prox(1,1)-dist(1,1);
pré_y=prox(1,2)-dist(1,2);
pré_z=prox(1,3)-dist(1,3);
post_x=prox(2,1)-dist(2,1);
post_y=prox(2,2)-dist(2,2);
post_z=prox(2,3)-dist(2,3);

distance_pre=sqrt(pre_x^2+pre_y^2+pre_z^2);
distance_post=sqrt(post_x^2+post_y^2+post_z^2);

diff_pre_post=distance_pre-distance_post;
percentchange=(distance_post-distance_pre)/abs(distance_pre)*100;

end

%% end code
Vita

Lindsay Ellen Clayton was born on December 22, 1988 in Fairfax County, Virginia, but was raised in Anne Arundel County, Maryland. She graduated from Severn School, Severna Park, Maryland in 2007. She received her Bachelor of Science degree in Engineering Science and Mechanics from Virginia Tech, Blacksburg, Virginia in 2011. She subsequently worked for half a year as an engineering consultant in the Biotech-in-a-Box science outreach laboratory in Blacksburg, Virginia before pursuing her Master of Science.